Single limb cable driven wearable robotic device for upper extremity movement support after traumatic brain injury

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Abstract

Introduction: Recently, soft exosuits have been proposed for upper limb movement assistance, most supporting single joint movements. We describe the design of a portable wearable robotic device (WRD), "Armstrong," able to support three degrees-of-freedom of arm movements, and report on its feasibility for movement support of individuals with hemiparesis after traumatic brain injury (TBI).

Methods: We introduce Armstrong and report on a pilot evaluation with two male individuals post-TBI (T1 and T2) and two healthy individuals. Testing involved elbow flexion/extension with and without robotic-assisted shoulder stabilization; shoulder abduction with and without robotic-assisted elbow stabilization; and assisted shoulder abduction and flexion. Outcome measures included range of motion and root mean square trajectory and velocity errors.

Results: TBI subjects performed active, passive, hybrid and active assistive movements with Armstrong. Subjects showed improvements in movement trajectory and velocity. T1 benefited from hybrid, active, and assistive modes due to upper extremity weakness and muscle tone. T2 benefited from hybrid and assistive modes due to impaired coordination. Healthy subjects performed isolated movements of shoulder and elbow with minimal trajectory and velocity errors.

Conclusions: This study demonstrates the safety and feasibility of Armstrong for upper extremity movement assistance for individuals with TBI, with therapist supervision.

Introduction

Traumatic brain injury (TBI) affects 2.87 million Americans each year.[1] According to estimates from several states, 3.2 million to 5.3 million individuals are living with a TBI-related disability in the United States.[2–4] Overall, males account for approximately 59% of all reported TBI-related medical visits in the United States.[5] According to some estimates, TBI impacts lead to unemployment in 60.4% affected individuals between the ages of 16–60.[6] Another study indicates that approximately 60% of TBI survivors are males younger than 36 years of age with the ultimate goal of returning to prior level of function and employment.[7] A 5-year retrospective study indicated that only 41% of TBI patients return to pre-injury level for home management.[8] Results also indicated that 30% of these patients reported upper limb motor deficits five years after the injury. Furthermore, up to 35% of these individuals stopped activities such as cooking, shopping, yard care and childcare due to disability. Therefore it is reasonable to assume that improving upper limb

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function could result in improved independence and lower caregiver burden in this population.

Current rehabilitation research indicates that task-specific and intensive practice could significantly improve motor recovery and neuroplasticity after brain damage.[9,10] Shaw et al. recommended 6 h of daily task-specific training of the affected upper extremity to improve function and performance.[10] The greater effectiveness of intensive task-specific practice relative to standard therapy techniques suggests that repetitive motor practice is a crucial rehabilitation component, and presents a key opportunity for the introduction of robotics in rehabilitation. Upper-extremity robotic devices have been developed and used for training of stroke and spinal cord injury patients in order to increase the efficiency and data feedback of rehabilitation.[11–13]

Increasingly, exoskeleton-type robots have been proposed for rehabilitation, since the robot can more accurately align with the various joints of the upper limbs to support movements.[14] There has also been a growing trend in developing softer, more compliant and wearable versions of exoskeletons that might be suitable not just for rehabilitation, but also for providing movement assistance. Devices have been proposed for assisting hand function,[15–17] elbow flexion and extension,[18–20] and shoulder movements.[21,22] A few devices target multiple movements of the upper limb, spanning shoulder and elbow,[23] or shoulder, elbow, forearm, and wrist.[24]

These wearable robotic devices (WRDs) offer movement assistance, and also the potential advantages of bringing therapy to new venues including the home. This capability comes at an opportune time, given the $76.5 billion annual cost of TBI and the pressure on the healthcare systems of the United States and other nations. In the U.S., most healthcare plans limit the number and duration of rehabilitation visits (a typical rehabilitation visit is one hour or less), and the amount of therapy delivered per patient is declining due to decreases in length of stay and frequency of outpatient rehabilitation visits. While robotic devices are costly, they have the potential to increase cost-effectiveness of rehabilitation sessions by reducing physical demand on therapists (e.g. facilitating limb positioning) and allowing the collection of important performance measures. Recent evidence suggests greater effectiveness of robotic therapy compared with traditional care and intensive therapy for long term upper extremity rehabilitation (36 weeks) of highly involved patients with acquired brain injury.[11] In the aforementioned clinical study, additional cost analysis suggested comparable costs between robotic therapy, traditional care, and intensive therapy.[25] These findings, combined with current limitations in healthcare resources, suggest that robotic devices have the potential to be a cost-effective mode of upper extremity rehabilitation after brain injury.

Soft exosuits have been shown to increase range of motion in shoulder abduction/adduction, elbow flexion/extension, forearm pronation/supination, and wrist flexion extension following stroke.[24] In a small case study involving a portable upper-extremity WRD called CRUX, the authors demonstrated that the device could assist a stroke-impaired individual in the completion of elbow flexion-extension movements.[26] A number of other studies have reported beneficial effects of exosuits including reduced muscular effort and biological torque requirements to achieve movement, but these have only evaluated the hardware with able-bodied participants.[19,20]

In this paper, we present a wearable robotic device, "Armstrong," that assists with shoulder and elbow movements. This device is the first portable soft wearable robotic device to be tested on participants with TBI, and demonstrates the capability to adequately actuate the shoulder of an impaired individual in flexion and abduction - important for providing functional range of motion in TBI patients. Typical movements that capture functional workspace include hand at ear, over chest, and reaching to work surface. All of these include some level of proximal upper extremity movement including shoulder flexion and abduction and elbow flexion and extension. Regaining proximal shoulder and elbow movements are the biggest contributors to functional recovery.[27] Development of the device focused on accurate, repeatable shoulder and elbow manipulation by addressing core design challenges such as the body-machine interface, strategic actuator placement, and use of advanced materials to promote transparent feel and patient comfort and safety. A pilot study with both able-bodied and motor-impaired individuals is presented, with the aim of demonstrating the feasibility of using a soft, portable, wearable robotic device for assisting shoulder and elbow movement that could potentially translate to the rehabilitation domain.

Materials and methods

Wearable robotic device

The Armstrong wearable robotic device is fabricated of primarily soft materials with strategically placed rigid components. It can be powered by batteries, making it portable. The device manipulates the upper extremity allowing for three actuated degrees-of-freedom (DOF), namely shoulder abduction/adduction and flexion/extension as well as elbow flexion/extension. An included adjustable wearable jacket fits snugly around the

torso, and two custom cuffs encapsulate the upper and lower right arm (Figure 1). Actuation, power distribution, and electronics reside comfortably on the wearer's back, ensuring minimal distal mass. The mass of the system is less than 9 kg, and it can be powered from either a battery or external power supply. (The MIL-SPEC battery used for demonstrations adds 3 kg to the system mass, but this subsystem has not been a focus of development nor has it been optimized.) Synthetic tendons routed through conduit to the shoulder and elbow joints cross the joint and attach to the respective arm cuff distal to the joint. The limbs are actuated via pulling and releasing tendons in a coordinated fashion, similar to the approach demonstrated in other cable-actuated assistive exosuits and gloves.[15–17,22,23] Actuated range-of-motion of shoulder abduction/adduction, shoulder flexion/extension and elbow

flexion/extension are 0°–80°, –10°–80°, and 30°–130°, respectively (Figure 2). Maximum applied torques vary as a function of anthropometry and software limits placed on the synthetic tendons' tensions; typical values are in the 25–50 N-m range.

This device is focused on improving functional range of motion in TBI patients; three typical movements that capture the functional workspace were targeted. This workspace addresses the three common movement patterns previously mentioned: hand to ear, hand over chest, and reaching to a work surface. These motions are essential for completing self-care and activities of daily living such as meal preparation and eating, bathing and grooming, and dressing and undressing.

Hardware. The goal of creating a comfortable, lightweight, unobtrusive wearable device guided the iterative
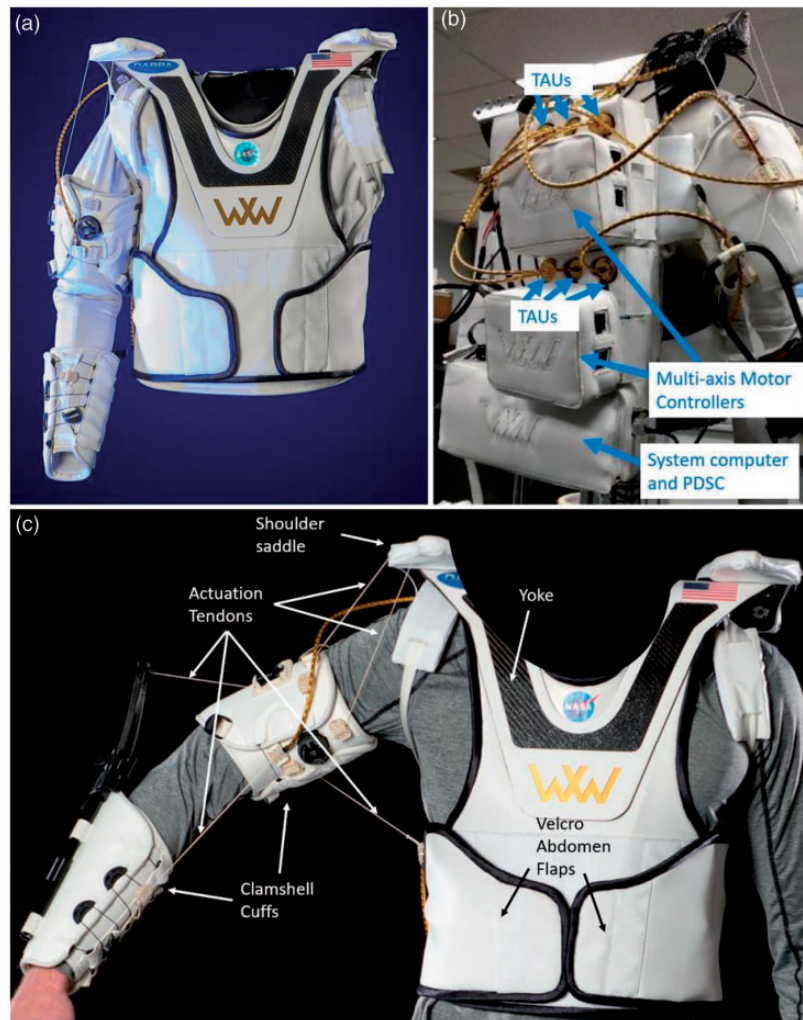

Figure 1. Armstrong—the soft, portable wearable robotic device utilized in this study—manipulates the upper extremity via pulling and releasing synthetic tendons in a coordinated fashion, allowing for three actuated degrees-of-freedom. It includes an adjustable wearable jacket fitting snugly around the torso, and two custom cuffs encapsulating the upper and lower right arm. Mass of the system is less than nine kilograms. (a) Front view. (b) Rear view. (c) Front view with subject's shoulder abducted.

development of Armstrong. The team strove to use soft, conformable materials where the device contacts the wearer, and only introduce rigid components where necessary for function. A large device volume was penalized to maintain a wearable, portable form factor. Distal weight was minimized by positioning heavier components on the trunk. Design trades contributed to expanding the shoulder extension and elbow extension bar away from the body, with the intent of creating sufficient torques. The rigidity of electrical components could not be avoided, but they were placed on the wearer's back to minimize interference. Other more structured components were integrated into the design in a manner to not impede the wearer's mobility.

Central to developing a WRD capable of comfortably, accurately, and safely positioning the limb is the design of the body-machine interface, or the physical interface between the user and the robot. If not properly addressed, the component of the tendon force going into the shoulder joint can be painful and cause joint wear. Large joint forces imparted due to the relatively large synthetic tendon forces necessary to manipulate the arm have been eliminated by utilizing a shoulder "saddle" with an extension and a carbon fiber "yoke" to distribute forces to the chest, back, and abdomen (Figure 1(c)). No load-bearing member is in contact with the shoulder.

Armstrong is capable of actuating the right arm. The WRD is donned by inserting the head through the yoke and wrapping Velcro flaps around the abdomen snugly. Comfortable clamshell cuffs on the upper arm and forearm contribute to the quick donning by the target population, without the need to insert the entire arm into a sleeve. The shape of the cuffs was chosen to minimize migration with respect to the user's skin or shirt, which is important for accurate control. Four tendons cross the shoulder joint, with proximal endpoints anchored on the shoulder saddle or torso and distal endpoints anchored to the upper arm cuff. Two tendons cross the elbow joint to control flexion/extension.

Six electromechanical actuators control the movement of the shoulder and elbow. These patented custom designed actuators,[28] dubbed Tendon Actuation Units (TAUs), are mounted on the back of the user, and are connected to various points on the body machine interface with a Bowden cable transmission that utilizes synthetic tendons. The TAUs work in a coordinated manner to move each degree of freedom at the shoulder and elbow. Each TAU comprises a motor with encoder, a pulley, an online tendon tension sensor (for control), a conduit force sensor (for redundancy/safety), conduit, and synthetic cable (tendon).

Electrical/software. The electrical subsystem (Figure 1(b)) is comprised of a backpack that houses electronics for a power distribution and safety controller (PDSC) and the system computer. Two custom multi-axis motor controllers reside above the lower backpack, each housing a Xilinx Spartan-6 field programmable gate array (FPGA) as its main processing unit. Three MicroBlaze processors are instantiated on each FPGA, for independent control of each actuator. Thus, each motor controller performs simultaneous real-time, closed loop control of three actuators (1 kHz loop rate), as well as I/O with load cell interface boards and communication with the system computer via EtherCAT. Each actuator's motor "tile" circuit
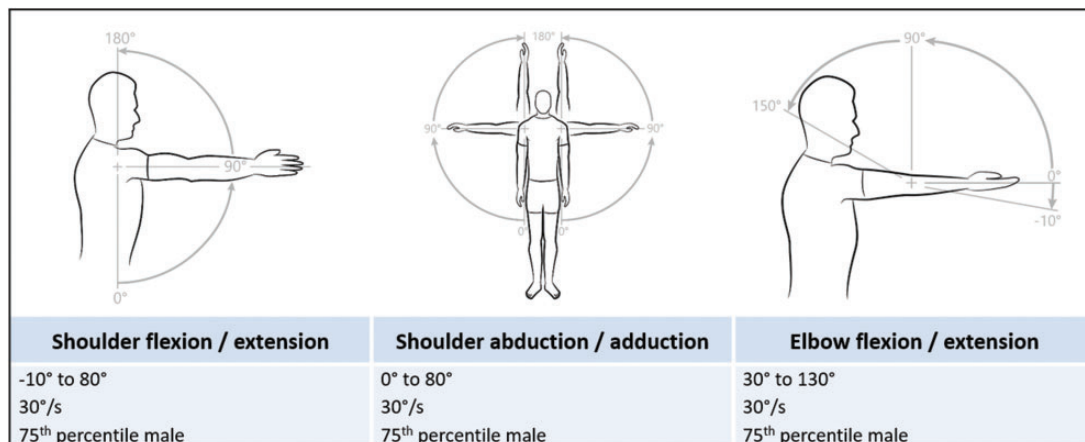

Shoulder flexion / extension	Shoulder abduction / adduction	Elbow flexion / extension
-10° to 80°	0° to 80°	30° to 130°
30°/s	30°/s	30°/s
75th percentile male	75th percentile male	75th percentile male

Figure 2. Armstrong was designed to achieve specified range of motion for shoulder flexion/extension, shoulder abduction/adduction, and elbow flexion/extension. These range of motion values were defined such that the functional workspace of Armstrong would support three fundamental movements common to activities of daily living (ADLs): hand to ear, hand over chest, and reaching to a work surface.

board hosts a Texas Instruments (TI) DRV8332 brushless DC motor driver, capable of sourcing electric current up to 8 A RMS; it also reads an incremental encoder and hall-effect sensors for motor commutation and closed loop control, and bus voltage and motor phase current measurements.

Armstrong utilizes the Robot Operating System (ROS) and Open Robot Control Software (OROCOS), a near real-time operating system based on ROS. In addition to the ROS message passing system, a custom library using shared memory to transfer data between processes is employed. This is the conduit for data to/from each motor controller and the PDSC; ROS messages are passed at 100 Hz.

Sensing/control. Sensing of the human arm position and joint torques is required for data recording, display, and control. As mentioned above, each TAU is comprised of a motor encoder and a tension sensor, allowing determination of length and proximal tension of each synthetic tendon. Shoulder control system development began on an idealized ball-and-socket testbed (Figure 3), where parameters such as three-dimensional positions of tendons with respect to joint center are known, tendon friction is negligible, and tendon connections to the "limb" are rigid. Assumptions break down, however, when translating from this apparatus to a compliant garment interfacing with the compliant flesh of a human. Therefore, additional position sensors have been employed by Armstrong to address this challenge. For the shoulder joint, which is treated in the controller as an ideal ball-and-socket joint, two Microstrain GX4-25 IMU sensors are utilized to determine the upper arm's orientation with respect to the fixed clavicle. One IMU is mounted on the shoulder saddle (proximal side of joint) and the other on the upper arm cuff (distal side of joint).

Several operational modes are available with Armstrong, selectable through a graphical user interface. In active assist mode, any or all of the controlled DOFs may be commanded simultaneously via coordinated control. In passive mode, the device provides minimal torque (sufficient to avoid slack in synthetic tendons) to the DOFs, allowing the wearer to freely move his or her limb. A hybrid mode allows one joint to be stabilized at a certain orientation while the other joint is free to be moved by the wearer, thereby aiding in weight offloading and joint isolation. Additionally, an assist-as-needed mode is available. This control mode uses the active assist mode with larger allowable position errors. In this mode, the device takes correcting action only if the error between a target and actual joint angle are above a configurable threshold. This grants the user maximum opportunity to rehabilitate with voluntary motion before the device overcomes the portions of the trajectory that are difficult or impossible for the subject to accomplish on his/her own. The intuitive user interface displays real-time data of commanded and actual joint positions.[29]

Each of the six TAUs uses a cascaded control scheme to ensure individual joint tracking while following safety and dynamic performance requests from the high level controller. The system is able to achieve tight performance tracking through several nested control loops that rely on sensor-fed feed-forward control terms and traditional PID controllers. To linearize
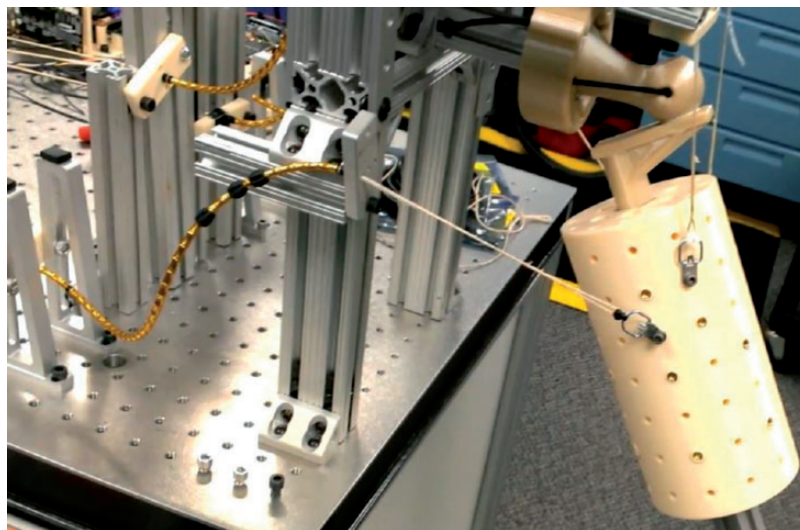

Figure 3. An idealized ball-and-socket testbed used in the development of Armstrong's control system. Responsive control was demonstrated with this testbed; when implementing on the wearable device, sensitivities to physical compliance and a moving center of shoulder joint rotation need to be accounted for.

and minimize contributions from the PID control loops, several feed-forward terms have been implemented into the nested loops to compensate for predicted electromechanical errors and non-linear physical characteristics of the TAUs.

In order to reduce sensitivities to physical compliance and compensate for idealized assumptions regarding the human-machine interface, such as the ability to control shoulder movement as in an idealized ball-and-socket joint, impedance control is performed at the actuator level. This provides a desired feel for each tendon, which can be affected by high-level determination of tendon length commands. A representation of this control scheme for the shoulder is shown in Figure 4.

Tendon length commands are determined from tendon endpoint geometry and desired joint angles using inverse kinematics. By incorporating the IMU-derived joint angles into the controller, joint angle error can be minimized. After calculating joint angle errors, these errors are fed into a length bias calculator, implemented essentially as an ID-controller, to ensure adequate speed of response, especially when changing direction. Gains were selected empirically, and remained unchanged for all users. The biased tendon

length command is acted on by the cascaded controller to achieve the desired joint angles and joint torques.

The use of inner and outer deadbands in the tendon length biaser allows for an assist-as-needed capability. When the joint angle error is outside the outer deadband, active tendon length biasing occurs; inside the inner deadband, no correction from the controller occurs. In between the two deadbands, the tendon length bias is latched; the controller will not attempt to further correct the joint angle error until journeying outside the outer deadband again. Deadband sizes can be configured to match a wearer's ability to follow a trajectory cue, and subsequently be used as a metric as a patient's ability improves.

In addition to the tendon length biaser, a force command limiter contributes to the spring-like feel of the controller. When the joint angle error is inside its respective outer joint angle error deadband, the tension command on a corresponding tendon is limited. This limits the tendon's effectiveness when joint angle errors shrink. When combined with the behavior of the tendon length biaser inside the deadbands, this effects a spring-like feel of the system. Conversely, a minimum tendon force of 1 lbf was levied on each tendon to
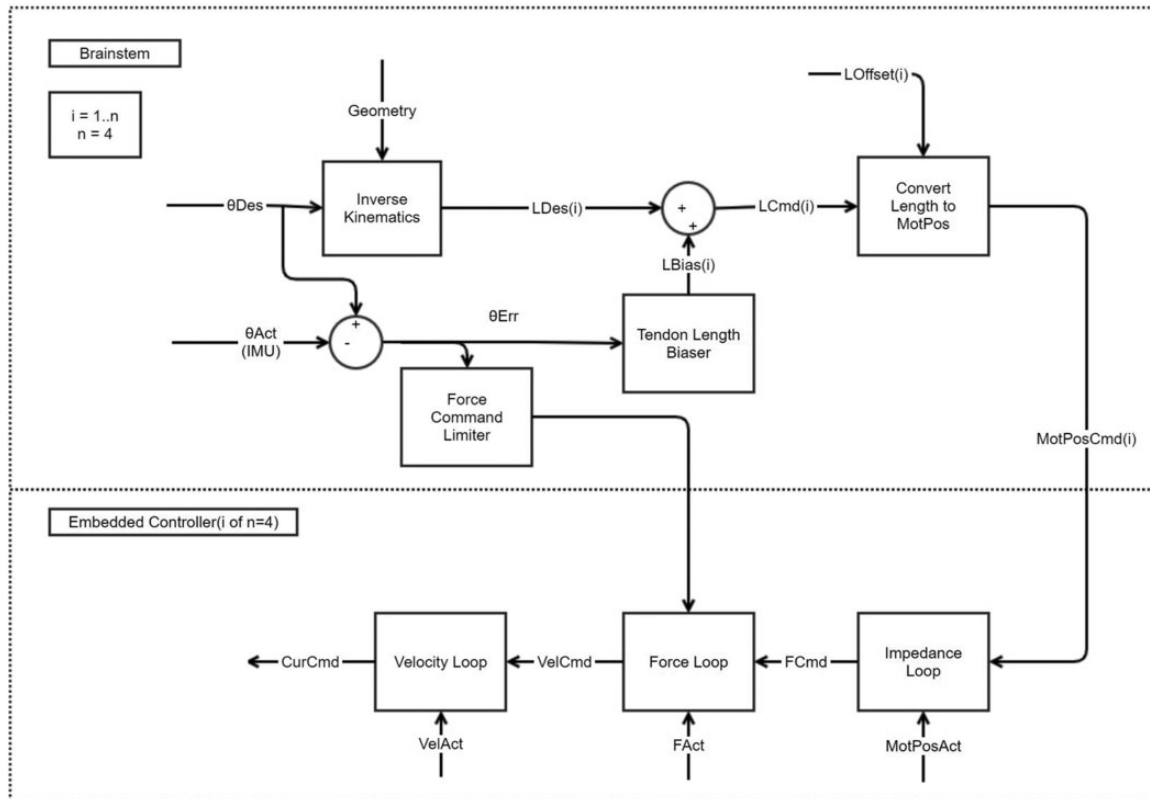

Figure 4. Control diagram for the shoulder joint. A tendon length bias calculator, based on IMU-derived joint angle data, is employed to make corrections to the model-based individual tendon length commands for each actuator. Impedance control is performed at the actuator level, providing a desired feel for each tendon.

ensure no tendon ever becomes slack, creating an uncontrollable situation.

Safety. Implemented safety measures include placement of tendons in such a way to minimize the risk of hyper-extension of shoulder and elbow joints. Joint compression and dislocation are also considered as potential risks. To control these risks each user is subjected to a "calibration" test. Data from force sensors in each tendon are then used to monitor and compare against the limits established during calibration. For force-sensing redundancy, data from the proximal conduit force sensor are compared to data from the on-line tension sensor; if the difference is outside a pre-set limit, a fault is emitted and motor power is removed. In addition to sensor-based hazard controls, system health- and communication-checking ensures that any component malfunction in the system will trigger a "safing" action, where all power to motors is removed.

Subjects

Two male subjects ("T1" and "T2") with diagnosis of Traumatic Brain Injury were recruited for the purpose of this study. Subject Demographics are included in Table 1. Subjects were included if they had the diagnosis of TBI confirmed by brain CT or MRI, were 18 years or older, had a TBI duration of greater than 3 months, and demonstrated right upper extremity hemiparesis. Subjects had no limitations in communicating or following instructions (Mini Mental score of > 24). Both subjects presented hemiparesis in right upper extremities with no significant limitation in the passive range of motion of shoulder and elbow. T1 was 29 years old and 14 months post injury while T2 was 32 years old and 53 months post injury. Therefore, the two subjects represented two different stages of chronicity after TBI. T1 had a more significant weakness and was unable to complete shoulder movements against gravity (active ROM shoulder flexion = 0°, shoulder abduction = 15°, elbow flexion = 125°). In addition, T1 presented with increased muscular tone in the right upper extremity as indicated by Modified Ashworth scale (MAS) = 1+ in the elbow flexor muscle group. T2 was able to actively achieve full range of elbow and shoulder motions against gravity with manual muscle testing. However, T2 had poor coordination as indicated by action tremor (during finger to nose testing). In order to ensure safety, subjects were included if they were independent with mobility at ambulatory (T1) and wheelchair (T2) level. Subjects were tested to ensure their right shoulder and elbow could be safely moved within the range achievable by the robotic device: shoulder flexion –10°–80°, shoulder abduction 0°–80°, and elbow flexion 30°–130°.

Two healthy subjects ("H1" and "H2") were recruited for comparison. Healthy subjects were 35 and 36 years of age with no history of neurological or orthopedic issues affecting movement of the right upper extremity. Healthy subjects replicated movements performed by the two TBI subjects.

Table 1. Demographic information for TBI subjects.

		TBI Subject 1	TBI Subject 2
Age		29 years	32 years
Gender		Male	Male
Hand dominance		Right	Right
Affected Upper extremity		Right	Right
Months since Injury		14	53
Level of education		College Degree	College Degree
Mini-mental status exam (MMSE)		30	28
Shoulder Flexion	Active-ROM	0°	135°
	Passive-ROM	130°	135°
Shoulder Abduction	Active-ROM	15°	150°
	Passive-ROM	110°	150°
Elbow Flexion	Active-ROM	125°	130°
	Passive-ROM	125°	130°
Modified Ashworth Rating Scale (MAS)		Elbow Flexors: 1+	Elbow Flexors: 0
		Elbow Extensors: 0	Elbow Extensors: 0
		Wrist Flex/Ext: 0	Wrist Flex/Ext: 0
		Shoulder Flex/Ex: 0	Shoulder Flex/Ex: 0
Resting tremor		Absent	Absent
Action tremor		Absent	Present

Note that MAS = 0 indicates no increase in muscle tone. MAS = 1 indicates slight increase in muscle tone, manifested by a catch and release or by minimal resistance at the end of the range of motion when the affected part(s) is moved in flexion or extension, MAS = 1+ indicates slight increase in muscle tone, manifested by a catch, followed by minimal resistance throughout the remainder (less than half) of the ROM. MMSE of > 24 is normal.

All subjects had access to an emergency stop button during the tests (placed in a reachable distance to the least impaired upper extremity for T1 and T2). Subjects were instructed to use the emergency button if they felt any discomfort or pain.

Testing protocol

Subjects participated in up to four sessions for engineering evaluation and garment fitting and one testing session for formal assessment. Each subject provided written informed consent under a protocol approved by the Institutional Review Board of Rice University (IRB-FY2016-234). Inclusion criteria were monitored by a physical therapist/neurologic clinical specialist. The physical therapist was present throughout all testing sessions. The Academy of Neurologic Physical Therapy of the American Physical Therapy Association (APTA) has developed the Evaluation Database to Guide Effectiveness (EDGE), in order to make recommendations for outcome measure utilization for therapy including a task force to evaluate outcome measures specifically for use in traumatic brain injury.[30] According to this task force, Modified Ashworth Scale, range of motion testing and coordination are considered essential for testing body function and structure. Therefore, we recorded data from Armstrong that would quantify range of motion and coordination with and without device assistance.

In this study, the physical therapist designed each testing session according to subjects' capabilities and limitations. Each subject performed different exercises depending on what was deemed appropriate based on therapist recommendation. For example, T1 performed exercises that would focus on reducing compensatory movements, while T2 performed exercises most appropriate to control action tremor and enhance movement accuracy. During each session, the subject was seated comfortably in an upright position. T1 was tested sitting on a stable surface with no back or arm support, while T2 was seated on his personal wheelchair with arm rests removed, as in Figure 5. The subjects were then assisted while donning the garment and were positioned facing a monitor that provided visual display of the movements. The duration of each session, including donning and doffing of the system, lasted between 1 and 2.5 h depending on the level of reported fatigue and subject availability. All subjects received safety precautions, such as potential for alterations in center of gravity, prior to testing.

During powered testing, subjects performed rehabilitation exercises selected from the list below and based on therapist discretion. Visual cues were given to the subject on a monitor, as shown in Figure 6, to provide a representation of the motion that was desired. All stabilization was accomplished by simply commanding a static trajectory for that degree of freedom.

Exercise A: Passive mode, constant velocity (13.3 deg/s for TBI subjects; 20.0 deg/s for healthy subjects) elbow flexion/extension cue between 30° and 90°. No shoulder or elbow stabilization.

Exercise B: Hybrid mode, constant velocity elbow flexion/extension cue between 30° and 90° with shoulder abduction and flexion stabilization.

Exercise C: Hybrid mode, constant velocity elbow flexion/extension cue between 30° and 90°, with additional elbow extension assistance torque and shoulder abduction and flexion stabilization.

Exercise D: Passive mode constant velocity (10 deg/s) shoulder abduction cue between 30° and 80°. No shoulder or elbow stabilization.

Exercise E: Hybrid mode, constant velocity shoulder abduction cue between 30° and 80° with elbow stabilization.

Exercise F: Assist-as-needed mode, constant velocity assist-as-needed shoulder abduction cue between 30° and 80°, with elbow stabilization.

Figure 5. Test Subjects T1 (left) and T2 (right) perform rehabilitation exercises to demonstrate Armstrong's capabilities. T1 was seated on a flat bench in an upright position; T2 remained in his personal wheelchair with arm supports removed.

Exercise G: Active Mode, WRD manipulates the passive limb through a defined range-of-motion. Shoulder and Elbow actuated.

Outcome measures

The actual joint angles and velocities were compared to their desired counterparts throughout each trajectory. A root-mean-squared (RMS) error (equation (1)) was calculated for each (Trajectory RMS = TRMS, Velocity RMS = VRMS), and these two numbers were used for comparisons among test cases. Root mean square error is commonly used in previous studies for quantifying movement trajectory.[31,32]

$$XRMS = \sqrt{\frac{1}{n}\left(X_1^2 + X_2^2 + \ldots + X_n^2\right)} \qquad (1)$$

where X_n is the nth sample of trajectory or (velocity) error.

The first few repetitions were excluded from each data set to ensure consistent performance by the subjects.

Results

TBI subject 1 ("T1")

During Exercise A, despite T1's ability to perform active elbow flexion and extension, presence of muscular tone impaired voluntary elbow extension. As a result, active elbow extension was achieved through compensatory movements of the shoulder during isolated elbow extension. The subject performed eight trials of elbow flexion and extension during Exercise A. Although the subject was able to follow the desired elbow trajectory (VRMS = 27.6 deg/s, TRMS = 20.9 deg), compensatory shoulder abduction up to 20 deg was observed during this trial. Figure 7 (left side) indicates the compensatory shoulder abduction observed in T1 in comparison with healthy subjects. T1 exhibited more trajectory and velocity error than healthy subjects (Table 2). Exercise B with stabilization of the shoulder (40° of abduction and 0° of flexion) was used to assist T1 in performing isolated elbow movement without shoulder compensation. As shown in Figure 7 (right side) this feature reduced compensatory shoulder abduction for T1 to <1 degree. Stabilization of the shoulder also improved movement quality as indicated by reduction in velocity and trajectory errors (VRMS = 23.9 deg/s, TRMS = 20.7 deg).

In order to further facilitate elbow isolation and overcome elbow flexion tone, additional assistance was applied to the elbow extension tendon in the subsequent trial (Exercise C). Elbow extension tendon force was set to 5 lbf (22 N) while the shoulder was stabilized as before. This elbow extension assistance further improved the movement by reducing elbow VRMS (22.2 deg/s) and TRMS (14.7 deg).
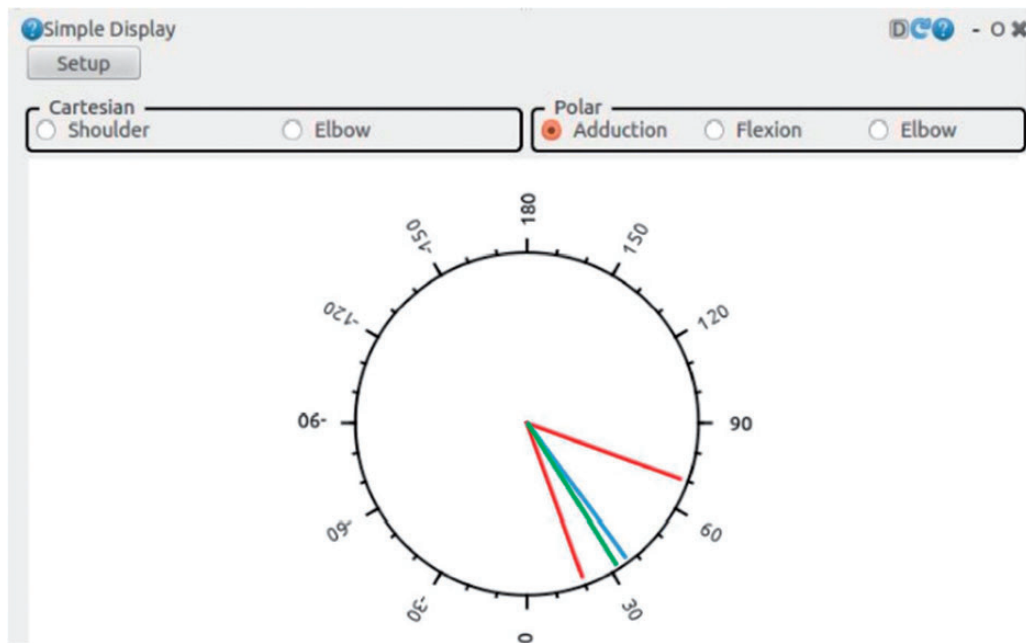

Figure 6. The simple trajectory cue displayed for the patient (blue), and the real-time angle (green). The red lines represent the range configured by the operator.
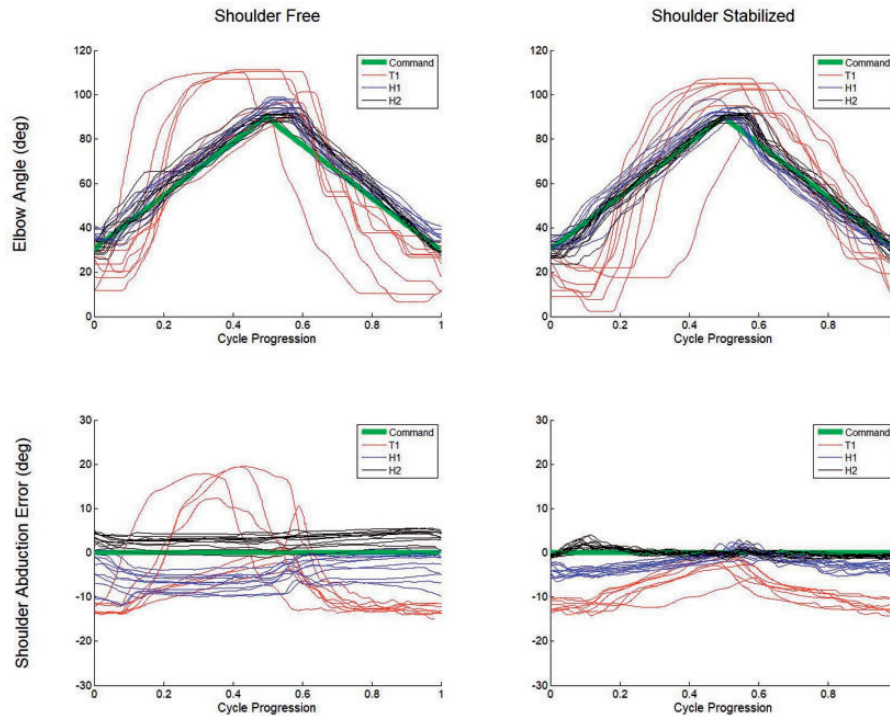
Figure 7. Elbow motion (top row) and shoulder abduction error (bottom row) for the free shoulder (left column) and stabilized (fixed) shoulder (right column) conditions, for T1, H1, and H2. When the shoulder was stabilized via actuation of the garment, there was no reduction in elbow range of motion, and shoulder abduction joint angle errors were noticeably reduced for all subjects. Reduced compensatory shoulder abduction for T1 was observed. Movement quality improved, as indicated by reduction in Velocity (from 27.6 to 23.9 deg/s) error.

Given T1's inability to perform much active shoulder flexion and abduction, these movements were carried out using the device's active assist mode with the robotic device moving the right upper extremity throughout the desired range of motion (Exercise G). T1 was able to achieve the commanded shoulder abduction and flexion ROM in this mode with no discomfort (Figure 8).

TBI subject 2 ("T2")

T2 presented with weakness in the right shoulder with normal elbow strength. T2 was able to move the right shoulder throughout the entire range, however indicated weakness against resistance as well as poor motor coordination as evident by clinical testing (positive finger to nose and supination/pronation test). The testing session for T2 emphasized robotic shoulder movements. Exercises D (passive mode) and E (hybrid mode) were performed, where the elbow was free and stabilized, respectively, to determine if reducing the demand on controlling degrees of freedom would improve movement quality, given the impairments in coordination noted above. Results are shown in Table 2 and Figure 9. Slight improvement in

Trajectory and Velocity errors in shoulder abduction were noted with elbow stabilization compared to free elbow. H2's Trajectory and Velocity errors similarly slightly improved when fixating the elbow; H1's Velocity error improved, but Trajectory error slightly increased.

Exercise F (Assist-as-needed mode) was used to improve movement trajectory in T2. The device's allowable shoulder abduction error was configured to six degrees; the device responded to correct any abduction error larger than this. This mode further improved Trajectory RMS (7.4 deg), but Velocity RMS error increased (16.1 deg/s). The assist-as-needed mode was not replicated in healthy individuals.

Discussion

The results demonstrate the feasibility of using the wearable robotic device, Armstrong, for upper extremity movement of patients with TBI. In this pilot evaluation, the Armstrong device was safely used to meet the individual needs of patients with TBI. The safety of the device was demonstrated as no adverse events or notable discomfort occurred during any subject session with

Table 2. Comparison of movement characteristics between TBI and healthy subjects.

	Subject					
	T1		**H1**		**H2**	
	Elbow VRMS Error (deg/s)	Elbow TRMS Error (deg)	Elbow VRMS Error (deg/s)	Elbow TRMS Error (deg)	Elbow VRMS Error (deg/s)	Elbow TRMS Error (deg)
Exercise A: Passive Mode	27.6	20.9	13.4	5.8	13.5	5.5
Exercise B: Hybrid: Shoulder Stabilized	23.9	20.7	13.8	5.3	14.1	4.0
Exercise C: Hybrid: Shoulder Stabilized; Elbow Extension assistance 5 lbf	22.2	14.7	14.4	4.7	16.0	4.0

|	Subject T2		**H1**		**H2**	
	Shoulder Abduction VRMS Error (deg/s)	Shoulder Abduction TRMS Error (deg)	Shoulder Abduction VRMS Error (deg/s)	Shoulder Abduction TRMS Error (deg)	Shoulder Abduction VRMS Error (deg/s)	Shoulder Abduction TRMS Error (deg)
Exercise D: Passive Mode	14.5	10.1	14.5	3.7	8.8	2.6
Exercise E: Hybrid: Elbow Stabilized	13.7	9.0	10.4	5.7	7.5	2.3
Exercise F: Assist-As-Needed; Stabilized Elbow	16.1	7.4	a	a	a	a

No data collected.
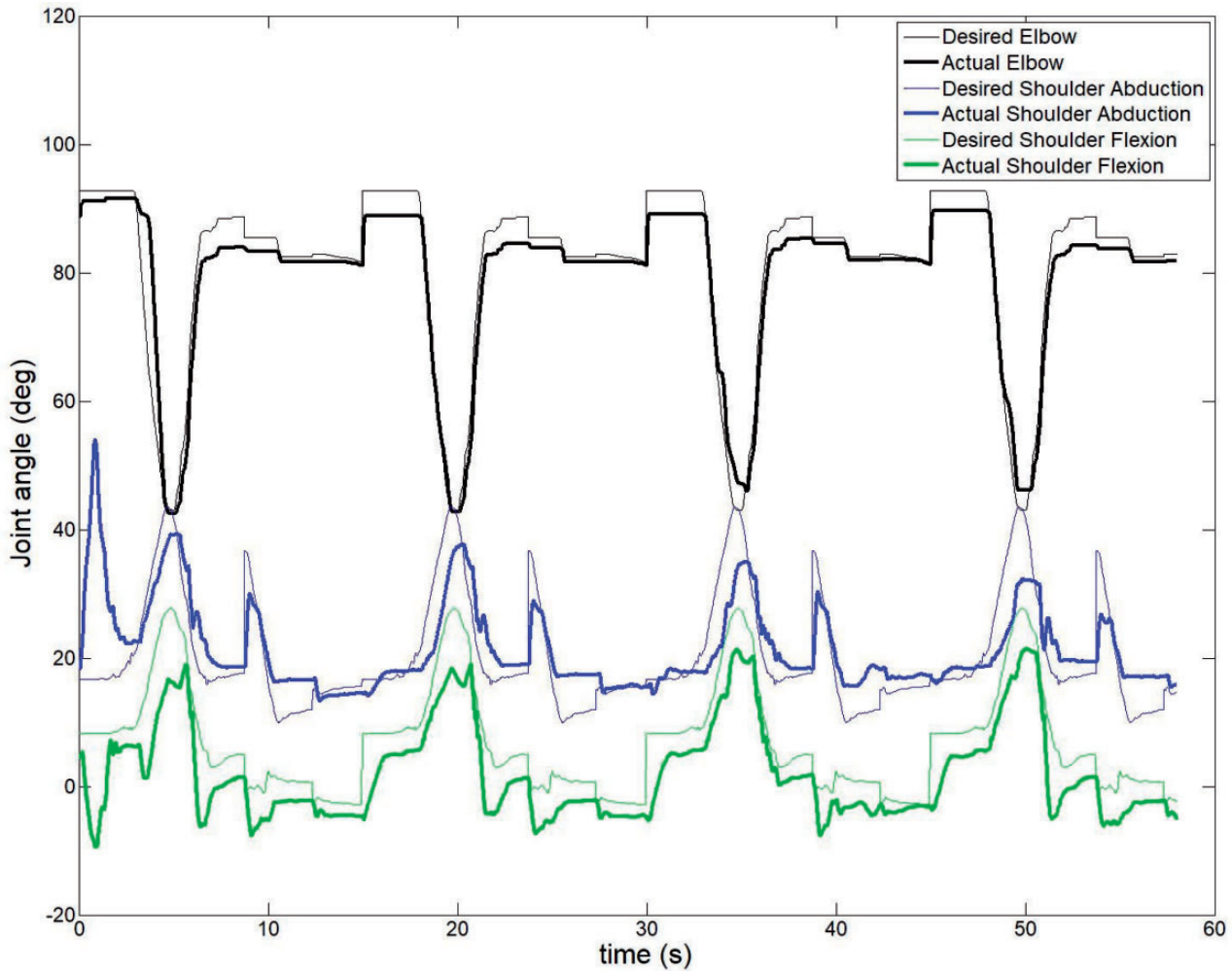
Figure 8. Desired versus actual joint angles during active assist mode for subject T1. Device accurately drove shoulder and elbow angles to commands for this subject, who was unable to perform much voluntary shoulder flexion or abduction.

the device operated in multiple modes and settings. Engineering and clinical staff were present for setup and supervision throughout the evaluation sessions. The key features evaluated included (1) robot-assisted stabilization of joints for better joint isolation, and (2) the optimized movement trajectory against abnormal tone in elbow extension by varying the level of elbow extension assistance.

Robot-assisted stabilization of joints was applicable and beneficial to both subjects despite their different levels of upper extremity involvement. For T1, the stabilization capability allowed for isolating the movement, overcoming the observed abnormal synergy, and reducing trajectory and velocity errors to values closer to that of healthy subjects. For T2, the stabilization feature slightly improved quality of shoulder movement. The reduction in trajectory errors is notable, since prior exosuits have had a negative effect on movement accuracy, though reduced muscular effort

was observed.[20] Future work should investigate whether these observed changes in movement characteristics are attributable to the increased range of motion, speed of motion, or motor control facilitated by the device.

A more advanced operational control mode (assist-as-needed) was used with T2 to achieve improved movement trajectories. This more advanced mode enables greater participation from the participant, and its appropriateness for use with this participant may be attributable to the fact that T2 was in a more advanced stage of recovery and was able to achieve full range of motion with no compensatory movements. This corresponds well with the Brunnström stages of recovery where in earlier stages synergy patterns are predominant (T1), and in later stages individual joint movements are possible and movement coordination approaches normal (T2).[33]

Given the low volume of subjects and data gathered in this pilot study, results do not show any statistically
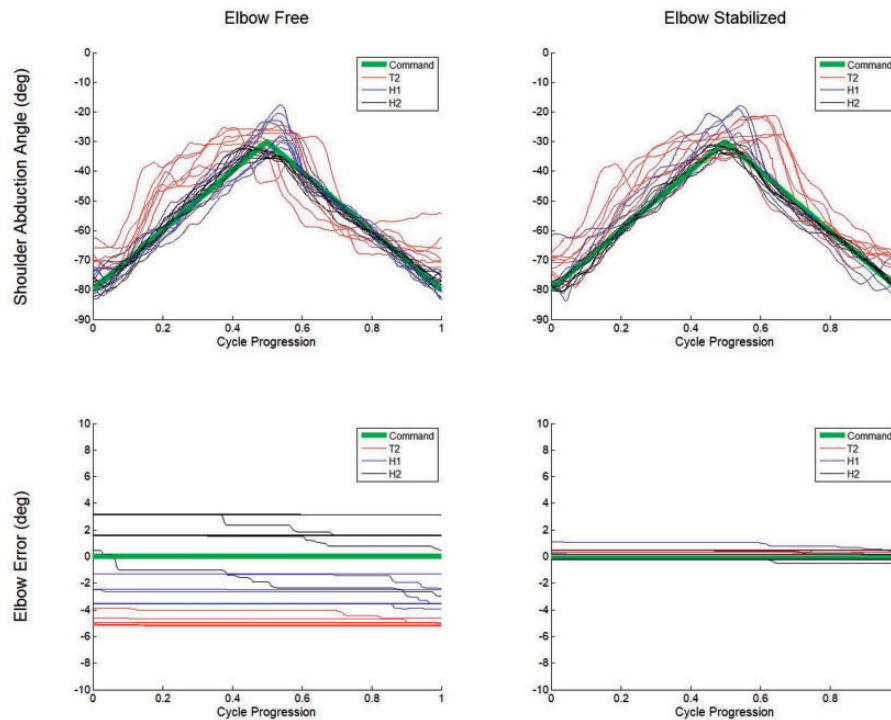
Figure 9. Shoulder abduction motion (top row) and elbow error (bottom row) for the free elbow (left column) and stabilized (fixed) elbow (right column) conditions, for T2, H1, and H2. When the elbow is stabilized via actuation of the garment, there is no reduction in shoulder abduction range of motion. Slight improvement in Trajectory (from 10.1 to 9.0 deg) and Velocity (from 14.5 to 13.7 deg/s) errors in shoulder abduction were observed for T2.

significant improvement of motor coordination with the suit, but demonstrate the proof of concept of using the Armstrong system for upper extremity movement support for both rehabilitation and assistance applications. The prototype system shows promise for future use as a rehabilitation tool that may be useful outside of a traditional clinical setting, pending future design updates and certifications. As described herein, the rehabilitation mode of the device operates under the assumption that the patient will be following a real-time visual cue. For a device to be used outside a clinical setting, this visual cue will need to be provided by a mobile device, or replaced by audio, tactile, or other cues.

Hardware updates will benefit the overall form factor cleanliness by addressing the unprotected tendon array and streamlining the elbow extension bar. System mass can be further reduced by optimizing design of major components, which has not yet been prioritized. The design attributes related to easy donning have been informed by experience with TBI subjects, including their verbal feedback. While the clamshell design of the arm cuffs removes the requirement for an impaired individual to thread his/her arm through a long sleeve, further improvements are warranted to allow one-handed cuff donning and adjustment.

These findings present the pilot results needed to support larger controlled clinical studies to further evaluate the safety, feasibility and efficacy of using portable wearable robotic devices to aid TBI patient rehabilitation.

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