2018

Self-Powered Robots to Reduce Motor Slacking During Upper-Extremity Rehabilitation: A Proof of Concept Study

E. P. Washabaugh

Emma Treadway
etreadwa@trinity.edu

R. Brent Gillespie

C. D. Remy

C. Krishnan

Repository Citation

Washabaugh, E. P., Treadway, E., Gillespie, R. B., Remy, C. D., & Krishnan, C. (2018). Self-powered robots to reduce motor slacking during upper-extremity rehabilitation: a proof of concept study. *Restorative Neurology and Neuroscience, 36*(6), 693-708. doi: 10.3233/RNN-180830

Self-powered robots to reduce motor slacking during upper-extremity rehabilitation: a proof of concept study

Edward P. Washabaugh, Emma Treadway, [...], and Chandramouli Krishnan

Abstract

Background:

Robotic rehabilitation is a highly promising approach to recover lost functions after stroke or other neurological disorders. Unfortunately, robotic rehabilitation currently suffers from “motor slacking”, a phenomenon in which the human motor system reduces muscle activation levels and movement excursions, ostensibly to minimize metabolic- and movement-related costs. Consequently, the patient remains passive and is not fully engaged during therapy. To overcome this limitation, we envision a new class of body-powered robots and hypothesize that motor slacking could be reduced if individuals must provide the power to move their impaired limbs via their own body (i.e., through the motion of a healthy limb).

Objective:

To test whether a body-powered exoskeleton (i.e. robot) could reduce motor slacking during robotic training.

Methods:

We developed a body-powered robot that mechanically coupled the motions of the user’s elbow joints. We tested this passive robot in two groups of subjects (stroke and able-bodied) during four exercise conditions in which we controlled whether the robotic device was powered by the subject or by the experimenter, and whether the subject’s driven arm was engaged or at rest. Motor slacking was quantified by computing the muscle activation changes of the elbow flexor and extensor muscles using surface electromyography.

Results:

Subjects had higher levels of muscle activation in their driven arm during self-powered conditions compared to externally-powered conditions. Most notably, subjects unintentionally activated their driven arm even when explicitly told to relax when the device was self-powered. This behavior was persistent throughout the trial and did not wane after the initiation of the trial.

Conclusions:

Our findings provide novel evidence indicating that motor slacking can be reduced by self-powered robots; thus demonstrating promise for rehabilitation of impaired subjects using this new class of wearable system. The results also serve as a foundation to develop more sophisticated body-powered robots (e.g., with controllable transmissions) for rehabilitation purposes.

Keywords: Rehabilitation robotics, hemiparesis, active engagement, EMG, metabolic cost, wearable, cable robot

Introduction

Robotic rehabilitation is a fast emerging and promising approach to recover lost functions after stroke or other neurological disorders (Caleo, 2015; Loureiro et al., 2011). Rehabilitation robots are gaining popularity because they are able to partially automate, standardize, and potentially increase the dosage of the therapy that patients can receive following injury (Huang and Krakauer, 2009). For the most part, robots that are used in therapy are assistive, where motors on the device power the subject by either unweighing their limb or fully aiding their movement (Marchal-Crespo and Reinkensmeyer, 2009). Typically, the motors used in these devices are expensive and heavy when worn by the subject and pose an inherent safety issue, because if the control system is not properly validated, errors could potentially lead to injury. Additionally, during this type of training, subjects are susceptible to motor slacking (Marchal-Crespo and Reinkensmeyer, 2009; Reinkensmeyer et al., 2009) – a phenomenon where the motor system reduces muscle activation levels and movement excursions, which may minimize metabolic- and movement-related costs. Hence, the patient remains passive and is not fully engaged in the training, which is not optimal for the recovery process (Duret et al., 2015; Israel et al., 2006; Lotze et al., 2003). Although roboticists are working on new control algorithms to better engage these patients (i.e., assist as needed, error reduction, error augmentation, etc.), research on these paradigms have found only limited benefits (Amirabdollahian et al., 2007; Marchal-Crespo and Reinkensmeyer, 2009; Patton et al., 2006a; Stein et al., 2007; Wirz et al., 2005). Thus, there is a critical need for unique robotic devices that can minimize slacking (i.e., letting the robot do all the work) and increase engagement in training.

Body-power is an alternative to external power, and can be harvested in order to drive robotic devices. In its most simple form, a rehabilitation robot can operate using body-power by having the subject's own limbs drive, or self-power the movement of the device using a direct transmission system. This is common in prosthetics, where movement of the residual limb is used to power the opening and closing of a hooked end effector. Indeed, many patients prefer body-powered robotic prostheses as opposed to modern myoelectric controlled models (Biddiss and Chau, 2007; Millstein et al., 1986), which may be in part due to the proprioceptive position and force feedback provided through the cable to the residual limb (Simpson, 1974). We believe that this same benefit would apply to body-powered rehabilitation robots. However, in the proposed body-powered robotic system, the subject would be driving their impaired limb for therapy, rather than a prosthesis. There are several practical advantages to training with body-powered rehabilitation robots. In addition to providing proprioceptive feedback through the robot's transmission system, a body-powered rehabilitation robot would improve safety as well as overall comfort since the user is in direct control of the system. These devices would also be cost effective and lightweight since they use transmissions to route power between joints rather than expensive and heavy motors. This could greatly increase their utility both within and outside the clinic, as patients could afford to continue training in their own homes. More importantly, body-powered rehabilitation robots could reduce motor slacking during robotic-assisted rehabilitation, thereby facilitating active engagement during therapy.

How could a self-powered robotic device reduce motor slacking during training? In part, slacking during robot-assisted exercise is a consequence of a subject's motor system exploiting assistive forces from the robot to minimize effort associated with movement (Casadio and Sanguineti, 2012; Casadio et al., 2013; Emken et al., 2007b; Reinkensmeyer et al., 2009). That is, unless subjects are either provided external motivation or are self-motivated to exert additional power during training, they will naturally tend to reduce their effort when performing a motor task (Casadio and Sanguineti, 2012; Emken et al., 2007b; Hidler et al., 2008). This is problematic in externally-powered (i.e., active motor driven) rehabilitation robots, as the user tends to extract as much power out of the robot as possible; thus, they reduce their own power output and slack. Alternatively, if the robot does not produce net power, this adverse incentive could be rendered ineffective, which is the basis of operation for body-powered rehabilitation robots. In this new paradigm, the robot does not inject additional power to the system during training; rather, the mechanical power is transferred between the driving (powering) and driven (powered) joints of the subject. We note that although the joints are intended to either drive or be driven in therapy, the flow of power may not always be in the direction from the driving to the driven joints. However, in theory, since no net power is introduced into the system, the user will remain engaged and will actively participate in training while using a body-powered rehabilitation robot. Potentially, body-powered robots could even increase active participation by the user and reduce slacking compared to an externally-powered robot, as reducing effort in the driven joint would necessitate additional effort from the driving joint. This finding could be of great importance for patients who are impaired due to neurological injury (e.g., stroke, spinal cord injury, and multiple sclerosis), as these populations are prone to compensations and motor slacking (due to learned disuse), which is potentially detrimental to their recovery (Israel et al., 2006; Secoli et al., 2011; Taub et al., 2006).

Thus, this study was performed to determine whether a device that harnesses the user's own body-power (i.e., self-powered) reduced slacking as opposed to when the device was externally-powered. To test this, we first designed and fabricated a low-cost and lightweight body-powered device that mechanically coupled the user's elbow joints. We then designed an experiment to test able-bodied (neurologically intact) and stroke subjects while using the device during various self-powered and externally-powered exercise conditions. Throughout the experiment, we measured muscle activation as a means to determine the subject's effort and the level of motor slacking. We hypothesized that motor slacking can be effectively reduced if individuals provide the power to move the limb via their own body power (i.e., from their opposite limb). We found that the use of body power to drive a rehabilitation robot can be an effective means to actively engage the driven limb during robotic-assisted training.

Materials and Methods

Device Design

We developed a wearable body-powered exoskeleton device that mechanically coupled the motions of the user's elbow joints (Fig. 1 and Video, Supplemental Digital Content 1). Bowden cables were used to realize a fixed 1:1 transmission ratio between the joint positions of the exoskeleton device. Each elbow was spanned using two 3D-printed ABS plastic exoskeleton segments: one for the upper arm, and one for the forearm. The two segments were connected on either side of the elbow joint by a hinged ball bearing connection. A rotary optical encoder (US Digital E2-1024-250-NE-D-T-B) was used to measure the elbow angle of each arm.

Fig 1.

Passive body-powered robot that mechanically couples elbow joints.

Two Bowden cables were employed to allow transmission of power during both extension and flexion of the elbow. The outer housing of the Bowden cable was anchored to the upper arm exoskeleton segment via typical mounting brackets on one arm; on the other arm, the cable and housing ran through a hole in a custom L-shaped bracket, allowing for tensioning of the cable by spinning a nut along the housing winding. The ends of the inner cable were wound around and were mounted to a capstan-style pulley incorporated in the lower arm segment design. By virtue of this pulley arrangement, a constant transmission ratio was achieved regardless of the elbow angle.

Custom molded cuffs were employed to secure each exoskeleton to the arm via hook and loop fastening elastic straps. The cuffs were custom fabricated by the University of Michigan Orthotics and Prosthetics Center to fit an arm shape. A variety of cuff sizes were made in order to fit the device to typical small, medium, or large arms. The straps had some elasticity in order to account for expansion along the arm's diameter as muscles contract, thereby ensuring that the device did not restrict the user's range of motion. These cuffs were bolted to the supporting 3D-printed exoskeleton structure, which passed below each arm.

Human Subjects Experiment

Experiments were performed to evaluate the merits of self-powered training using our body-powered device (Video, Supplemental Digital Content 1, which provides a description of the device and an overview of the experiment). Seven neurologically intact able-bodied adults (N = 7 [2 females and 5 males]; age: 24.0±7.0 years; height: 176.8±11.1 cm; weight: 69.3±14.0 kg) and six stroke survivors (N = 6 [4 females and 2 males]; age: 54.5±15.4 years; height: 167.2±11.4 cm; weight: 74.5±19.1 kg; upper-extremity Fugl-Meyer: 52.7±15.6; time since stroke onset: 3.4±2.9 years; Modified Ashworth Scale: 0.5 (0–2) [Median (Range)]) participated in the study. All able-bodied subjects were right hand dominant based on their own self-report, which is highly correlated with handedness questionnaires (Coren, 1993). Stroke subjects were included in this study if they 1) were between 18 and 75 years of age, 2) had a unilateral stroke at least 6 months prior to participation, and 3) had a cortical or sub-cortical lesion that was documented by radiological (CT or MRI) or clinical findings. Exclusion criteria included: 1) uncontrolled diabetes or hypertension, 2) Mini-Mental State Exam score < 22, 3) severe aphasia or inability to understand instructions, and 4) severe elbow spasticity (Modified Ashworth > 2), contractures, or limitations in joint range of motion. All procedures were carried out in accordance with the University of Michigan Human Subjects Institutional Review Board, and before participating in the study, all subjects provided written informed consent that also permitted the use of picture media in journals and other publications.

In this study, each subject wore the Bowden cable device described above during four conditions in which we controlled whether the device was powered by the subject (self-powered) or by the experimenter (externally-powered), and if the subject's driven arm was engaged (active) or at rest (resting) (Fig. 2).

Fig 2.

Schematic of the experimental protocol.

1. Active Self-powered (C1): The subject wore the device on both arms and was instructed to use both arms to perform a target-matching task.
2. Resting Self-powered (C2): The subject wore the device on both arms but was instructed to rest (i.e., relax) their non-dominant/affected arm and perform the task with their dominant/unaffected arm.
3. Active Externally-powered (C3): The device was externally-powered by the experimenter who wore the driving side of the device and performed the target-matching task, while the subject was instructed to contract in coordination with the externally delivered power using their non-dominant/affected, driven arm.
4. Resting Externally-powered (C4): The device was externally-powered by the experimenter who wore the driving side of the device and performed the target-matching task, while the subject was instructed to rest their driven arm.

During testing, the subject was positioned with their forearm secured into the cuff of the lower arm of the device and the elbow joint center aligned with the rotational axis of the device, and was asked to keep their wrist approximately in a neutral position (i.e., wrist and forearm at 0°). The proximal part of the device was then fixed to a riser located on a table, so as to allow for full extension of the elbow without interference. In this configuration, the subject performed the self-powered conditions. For the externally powered conditions, the device was placed onto the experimenter's arm in the same fashion. The same experimenter powered the driving arm for each subject group in the study. Further, the subjects were given clear instructions for each condition and the same experimenter gave those instructions for each group of patients. Instructions were given at the beginning of each trial, and the experimenter obtained verbal confirmation that the subject understood the instructions. Steps were also taken to make sure that the subjects were not influenced by the knowledge of the experiment (e.g., subjects were given no indication about the hypothesis, subjects were not coached during the experiment, and subjects were not given knowledge of the results [i.e., muscle activation during the experiment]). The target-matching task was performed for ten minutes in total (two-five minute blocks with a one minute rest in between each block to minimize fatigue and subject discomfort) during each condition.

Target-Matching Task The target-matching task required the device user to follow a 0.3 Hz sinusoidal trajectory shown on a 42 inch LCD TV monitor (placed at about 6 feet in front of the subject) by flexing and extending their elbow (Fig. 3A). The template wave was shifted and scaled so that the minimum corresponded to the subject's fully extended position, while the maximum amplitude corresponded to approximately 70% of the subject's range of motion (maximum flexion) while wearing the device. While matching the target trajectory, the elbow joint angles of the driven limb were recorded in real-time using the encoder on the device and displayed as feedback. Communication with the encoder was carried out using a quadrature to USB adapter (QSB, US Digital, Vancouver, WA, USA) and LabVIEW 2011 software (National Instruments Corp., Austin, TX, USA), which sampled the data at 30 Hz. While the experimenter was matching the target (i.e., conditions 3 and 4), subjects were not given visual feedback of the sinusoidal trajectory. This was done so that the subject would reliably contract in phase with the experimenter rather than oppose them using their corresponding antagonist muscles, as we observed during pilot testing.

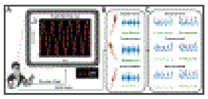

Fig 3.

Schematic of the training task, experimental setup, and data processing.

Electromyography Prior to the experiment, surface electromyography (EMG) electrodes (Trigno, Delsys, Natick, MA) were placed over the muscle bellies of the elbow flexors (biceps brachii and brachioradialis) and extensors (triceps brachii) of both the driven (the arm being powered) and driving arm (the arm supplying the power) of the study participants (Fig. 3B). Electrodes were placed according to established guidelines (www.seniam.org) (Krishnan et al., 2013b; Ranganathan and Krishnan, 2012). For fixation, the EMG electrodes were tightly secured to the skin using self-adhesive tapes and cotton elastic bandages. The quality of the EMG signals was visually inspected to ensure that the electrodes were appropriately placed. The raw EMG data of both the subject and experimenter were collected during their respective conditions using custom software written in LabVIEW 2011 (National Instruments Corp., Austin, TX, USA), and were sampled at 1000 Hz.

Data Analysis

The subject's muscular effort while using the device was evaluated using the EMG data collected during the four testing conditions. For processing, the raw EMG data were band-pass filtered (20–500 Hz), rectified, and smoothed using a zero phase-lag low-pass Butterworth digital filter (8th order, 6 Hz Cut-off) (Krishnan et al., 2013b; Washabaugh et al., 2016). The resulting EMG profiles of the four conditions were then segmented at each period of the target-matching task (i.e., cycle-by-cycle). Periods of the sinusoidal trajectory used in the matching task were identified using accelerometer data collected from the Trigno EMG sensors. We then calculated the cycle-by-cycle maximum by finding the maximum EMG amplitude over each period of the sinusoidal trajectory (Fig. 3C). The cycle-by-cycle maximum values were then used to compute two outcome variables that quantified motor slacking and its time course (i.e., whether the behavior was consistent during training): (1) average muscle activation, and (2) slope of the cycle-by-cycle maximums. The average muscle activation values were calculated by computing the mean of the cycle-by-cycle maximums and were normalized as follows: in the driven arm, the values for each condition were expressed as a percentage of values obtained during condition 1 (C1/C1, C2/C1, C3/C1, and C4/C1). For the driving arm, the subject's EMG values were expressed as a percentage of condition 1 (C1/C1, C2/C1), while the experimenter's EMG values (while driving the device) were expressed as a percentage of condition 3 (C3/C3, C4/C3). The slope was calculated by performing a linear regression over the cycle-by-cycle maximums. Here, lower *average muscle activation* would indicate motor slacking and a difference in *slope* (i.e., differences in slope across conditions) would indicate that the time-course of slacking behavior was inconsistent between conditions.

Statistical Analysis

All statistical analyses were performed separately for both groups using SPSS for Windows version 22.0 (SPSS Inc., Chicago, IL, USA). First, descriptive statistics were computed for each variable. Then, to examine the effects of the device on motor slacking, we performed three-way repeated measures analysis of variance (ANOVA) on the *average muscle activity* and *slope* with group as a between-subjects factor and condition and muscle as within-subjects factors. For the driven arm, analyses were performed to compare the self-powered and externally-powered conditions when the driven arm was either active (i.e., condition 1 vs 3) or resting (i.e., condition 2 vs 4). For the driving arm, analyses were performed to compare the active and resting conditions when the device was either self-powered by the subject (i.e., condition 1 vs 2) or externally powered by the experimenter (i.e., condition 3 vs 4). Driving arm analyses were performed to demonstrate the increased effort during resting conditions. A significant three-way interaction (condition × muscle × group) was followed up by two-factor repeated measures ANOVA for each group. Significant main effects or interaction effects were followed up by paired t-tests with Bonferroni correction. A significance level of $\alpha = 0.05$ was used for all statistical analyses and all tests were two-sided.

Results

Driven arm

Active Self-Powered and Active Externally-Powered (conditions 1 and 3) Three-factor repeated measures ANOVA revealed a significant main effect of condition on average muscle activation [$F(1,11)=5.055$, $p=0.046$]. Muscle activation was higher when the subject self-powered their driven arm compared with the condition in which their arm was externally-powered (Fig. 4). There were no significant group or interaction effects for average muscle activation (see supplemental appendix 1A for details). There were also no significant main effects or interactions for the slope of the muscle activation (see supplemental appendix 1B for details). The time-course of muscle activation can be seen in Fig. 5.

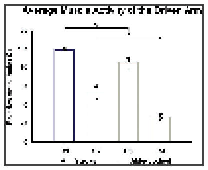

Fig 4.

Average muscle activity of the driven arm.

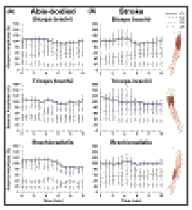

Fig 5.

Driven arm cycle-by-cycle muscle activation.

Resting Self-Powered and Resting Externally-Powered (conditions 2 and 4) There was a significant condition × muscle × group interaction on average muscle activation [$F(2,22)=4.280$, $p=0.027$]. Again, there was no significant effect for group (see supplemental appendix 1C for details). Two-factor repeated measures ANOVA for each group revealed a significant main effect of condition for both the stroke [$F(1,5) = 9.943$, $p = 0.025$] and able-bodied subjects [$F(1,6) = 18.934$, $p = 0.005$]. Muscle activation was higher when the subject self-powered their driven arm compared with the condition in which their arm was externally-powered (Fig. 4).

In able-bodied subjects, there was also a significant condition × muscle interaction effect on average muscle activation [$F(2,12) = 3.947$, $p=0.048$]. Post hoc analysis showed that this occurred because the biceps and brachioradialis muscle activation decreased to a larger extent than the triceps muscle during externally-powered condition (a decrease in average muscle activation of 34.2 ± 10.1%, 20.6 ± 4.6%, and 13.6 ± 4.0%, respectively), although all muscles showed significant differences in muscle activation between the two conditions ($p = 0.004$ to 0.015). There were no other significant main effects or interactions for average muscle activation within the groups (see supplemental appendix 1D for details).

Three-way repeated measures ANOVA indicated a significant muscle × group interaction effect [$F(2,22)=6.198$, $p=0.007$] on the slope of muscle activation, where post hoc testing showed that the biceps had a larger (i.e., more positive) slope than the triceps and brachioradialis in stroke survivors ($p= 0.015$ to 0.016). There were no other main or interaction effects (see supplemental appendix 1E for details).

Driving arm muscle activity

Active Self-Powered and Resting Self-Powered (Conditions 1 and 2) There was a significant main effect of condition on average muscle activation [$F(1,11)=12.933$. $p=0.004$]. As expected, muscle activation was higher in the subject's driving arm when the subject was resting their driven arm compared with when the subject was actively using their driven arm during the task (Fig. 6). There were no other significant group or interaction effects seen in the subjects' driving arm for either average muscle activation or slope of the muscle activation (see supplemental appendix 1F for details). The time-course of muscle activation for the driving arm can be seen in Fig. 7.

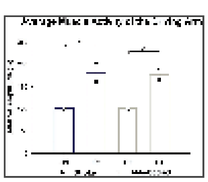

Fig 6.

Average muscle activity of the driving arm.

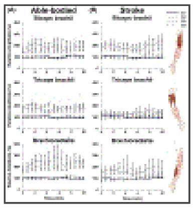

Fig 7.

Driving arm cycle-by-cycle muscle activation.

Active Externally-Powered and Resting Externally-Powered (conditions 3 and 4) There was a significant main effect of condition on average muscle activation [$F(1,11)=31.249$, $p<0.001$]. Muscle activation was higher in the experimenter's driving arm when the subject was resting their driven arm compared with when the subject was actively using their driven arm during the task (Fig. 6). This is analogous to the extra power that a motor would exert if the subject was slacking during training. There were no other significant group or interaction effects seen in the experimenter's driving arm for average muscle activation or slope of the muscle activation (see supplemental appendix 1G for details).

Discussion

This study tested a novel hypothesis that motor slacking during rehabilitation can be reduced by using self-powered rehabilitation robots. To test this hypothesis, we first designed and fabricated a device that mechanically linked a subject's elbows via a cable transmission. This device was then utilized in a human subject experiment in which the subject operated the device in various self-powered or externally-powered exercise conditions. The results of this study indicate that the device reduced slacking when it was self-powered by the subject compared to when the device was externally-powered by the experimenter. Most notably, when the device was self-powered, subjects unintentionally activated their driven arm even when explicitly told to slack by relaxing their arm. This finding is important, as it provides the basis for further testing of body-powered robots during rehabilitation of individuals with neurological impairments.

The field of rehabilitation robotics has seen significant growth in recent years, particularly due to the ability of robots to provide high-intensity task-oriented therapy while reducing the physical labor exerted by therapists. However, emulating the many functions of a therapist is a major challenge for a robot. Therapists provide interventions gauged to a patient's needs, provide gentle, compliant manual assistance when needed, and occasionally provide extra resistance to encourage the patient to exert more effort. Producing a robot that emulates a therapist presents two major technical challenges: 1) an interface is required that ensures safety, supports patient-centered adaptation, and promotes patient engagement, and 2) compact and portable actuation is required to source sufficient power in a manner that is backdriveable and compliant. Both of these challenges are augmented if the robotic device is to be taken out of the clinic and into the home. Accordingly, a number of robotic devices have been built to improve the robotic interface and promote active involvement of the patient during the therapy (Brochard et al., 2010; Chang et al., 2018; Hogan and Krebs, 2011; Scott and Dukelow, 2011; Washabaugh et al., 2018; Washabaugh et al., 2016; Washabaugh and Krishnan, 2018). A variety of controllers have also been developed for these devices, including techniques referred to as assist-as-needed, error augmentation, error reduction, and constraint-induced control strategies (Johnson et al., 2005; Marchal-Crespo and Reinkensmeyer, 2009; Reinkensmeyer and Patton, 2009; Simon et al., 2007). Experience with such devices and controllers suggests that patients remain more actively engaged when the robot requires them to initiate motion, and show greater improvements in motor recovery (Abdollahi et al., 2014; Kantak et al., 2013; Krishnan et al., 2013a; Krishnan et al., 2013b; Krishnan et al., 2012; Patton et al., 2006b; Rong et al., 2017).

However, such robots are often large, bulky, and expensive, thereby rendering them impractical to be engineered into a wearable device that could be taken home. Further, assist-as-needed control algorithms that require patient's neurophysiological signals (e.g., EMG or EEG) to infer their intention to perform a particular movement (Brauchle et al., 2015; Rosen et al., 2001; Stein, 2009) are sensitive to electrode placement and skin properties, and often lack fidelity due to crosstalk from neighboring muscles or brain regions (Marchal-Crespo and Reinkensmeyer, 2009). More importantly, researchers have shown that incorporating sophisticated control algorithms (e.g., anti-slacking watchdog algorithms) could actually discourage motor exploration and learning over time (Sans-Muntadas et al., 2014). However, appropriately designed body-powered robots could potentially address many of these issues at a relatively low cost without compromising on safety.

A self-powered device provides a training environment that is somewhat similar to constraint-induced control strategies for rehabilitation robots—derived from Constraint-Induced Movement Therapy techniques (Shaw et al., 2005; Taub et al., 1999; Uswatte and Taub, 2013)—where the patient is encouraged to use their impaired limb through a forced use feedback and reinforcement scheme. Indeed, our results demonstrated that the device encouraged use of the driven or impaired limb. The average muscle activation of the driven arm was higher during the self-powered conditions as opposed to the externally-powered conditions. This was the case irrespective of whether the subject was told to actively engage or rest their muscles. These findings were consistent between the different muscle groups, albeit to a greater extent in the antigravity muscles, such as biceps brachii and brachioradialis, than the gravity-assisted triceps brachii muscle. Regression slopes derived from the cycle-by-cycle muscle activation levels also indicated that this trend of greater muscle activation persisted throughout the trial and did not wane after the initiation of the trial. This finding is particularly encouraging considering that patients often slack after the initiation of the trial while using some of the existing assist-as-needed control algorithms: these algorithms typically necessitate active involvement of the patient only to trigger the robotic device, after which the robot takes control of the movement (Marchal-Crespo and Reinkensmeyer, 2009).

Training with an elbow-elbow coupling device also resembles bimanual therapy. Although typical bimanual training does not provide mechanical coupling of the limbs, it has long been a proposed strategy to improve function in individuals with stroke (Mudie and Matyas, 2000; Stinear and Byblow, 2004; Whitall et al., 2000). Bimanual training is seen as potentially beneficial because it improves bimanual coordination and facilitates reorganization of ipsilesional and contralesional networks (Bertolucci et al., 2018; Grefkes et al., 2008; Kantak et al., 2017; Luft et al., 2004; Stinear and Byblow, 2004; Swinnen, 2002). Studies testing the effects of bimanual training have also found improved movement performance (e.g., kinematics, velocity, movement smoothness) and motor recovery of the paretic limb following training (Cunningham et al., 2002; Harris-Love et al., 2005; Lin et al., 2010). However, it is unclear to what extent the reorganization of cortical networks (typically evaluated using fMRI or transcranial magnetic stimulation) due to bimanual training are directly related to functional recovery; although it appears that improvements in functional recovery are more evident in individuals who show evidence of cortical reorganization (Luft et al., 2004). Bimanual training has also been criticized as there is a concern that use of the non-paretic limb may promote compensation and impede recovery of the paretic limb; although there is currently little evidence to support this concern (Wolf, 2007). Thus, future studies should evaluate the neuroplastic effects of our proposed training, as it may have meaningful clinical implications for training with a body-powered device.

The device we created for this study (featuring a fixed 1:1 transmission coupling the elbows using a Bowden cable) was the most basic implementation of a body-powered rehabilitation robot, and was intended only to provide proof of concept for the use of self-powered robotic devices for training. In its current form and under the testing conditions in this study, the device operates very similarly to how a patient may directly couple their limbs following neurological injury (i.e., use their sound arm to grasp the wrist of their impaired arm and move it to a desired position). This method is widely used by stroke survivors with upper extremity impairments, as it is a useful means to provide self-range of motion exercise to the impaired limb or set the limb in a desired posture. However, bimanual limb–limb coupling is just a subset of how body-powered robots could operate as a whole.

Indeed, many improvements can be made to the device to further implement the body-powered concept. More sophisticated designs for body-powered rehabilitation robots would use controllable transmissions (e.g., based on digital hydraulics) (Gan et al., 2015; Treadway et al., 2018), where responsiveness of the driven arm could be diversified by altering the transmission ratio provided between the arms (and even allow for negative transmission ratios). This would change the proprioceptive feedback and perceived effort of the subject when driving their impaired limb, which could be used to shift the cost function for using the impaired arm and increase engagement in training. For example, subjects could be penalized for relying on their unimpaired limb by reducing the transmission ratio, which would increase the effort required to drive the impaired limb. Additionally, a similar effect could be achieved if the device is deliberately and programmatically made dissipative. When the amount of dissipated energy scales in proportion to the power transmitted through the device, slacking would be further penalized, as it would require more net effort to drive the dissipative transmission with a single limb than under conditions in which both joints provided their own power.

These modifications would permit the device to remain self-powered and passive, because although external power is routed to the switches and valves of the device, there is no flow of external power from the device to the user. Alternate extensions of the device could diversify the type of training that is provided using these devices. For example, by coupling the proximal and distal joints within a limb (e.g., shoulder – elbow or elbow – wrist), we could re-train normal synergies by targeting the abnormal synergies that are commonly observed after stroke (e.g., arm flexor synergies) (Ellis et al., 2005; Krishnan and Dhaher, 2012). Similarly, lower extremity joints could be coupled to fabricate a body-powered gait rehabilitation robot. However, using coupling constraints to coordinate complex motions would require variable transmissions to produce appropriate kinematic behaviors of the driven joint (Gan et al., 2015; Treadway et al., 2018).

The findings of this study have meaningful implications for designing rehabilitation robots, as motor slacking is commonly observed when rehabilitating individuals with neurological impairments using externally driven robots. This slacking has been theorized to be detrimental to recovery and has been posited as a key reason for the limited effectiveness of robotic-assisted therapy in comparison to conventional therapies (Casadio and Sanguineti, 2012; Emken et al., 2007a; Secoli et al., 2011). Our findings suggest that the use of body-powered robots can reduce slacking and encourage active participation during rehabilitation. For example, stroke often results in learned disuse (Taub et al., 2006), where the individual reduces the use of their impaired limb (i.e., slacks) and compensates with their ipsilateral or less impaired limb. By linking the impaired and unimpaired arms of a stroke survivor using a self-powered robot, we can potentially harness the body power generated by their unimpaired limb during compensation as a means to enlist effort from the impaired arm. Indeed, such coupling of motions has been shown to be an excellent brain priming tool to facilitate motor cortical excitability and accelerate motor recovery after stroke (Byblow et al., 2012; Raghavan et al., 2017; Stinear et al., 2014). Thus, prolonged use of a self-powered robotic device could prevent the formation of learned disuse in acute cases, or improve function for those that are already affected, albeit this hypothesis has to be verified in clinical trials.

A key limitation of this study is that we only tested the device on a small cohort of subjects. This was done because this study was only intended to provide proof of concept for the use of body-power in rehabilitation robots and was not meant to show the clinical potential of the device. We note that increased muscle activation during self-powered conditions was observed in a majority of the subjects tested (12/13 in C2 vs. C4 and 8/13 in C1 vs.

C3), indicating that the results would not be expected to change with a larger sample size. We also note that we had an inherent assumption that motor slacking is detrimental to recovery based on principles of motor learning and neuroplasticity (Kleim and Jones, 2008). However, after neurological injury, subjects often have additional complications (e.g., co-contractions and spasticity) that can alter the control of routine volitional movements (Levin and Dimov, 1997; Levin et al., 2000; Musampa et al., 2007) and even interfere with motor adaptation (Subramanian et al., 2018). Hence, it is unclear if these complications would interfere with self-powered training, and more work is necessary to determine if increasing EMG (or reducing motor slacking) can lead to better functional outcomes. An additional limitation is that the order of the testing conditions was not randomized between subjects to avoid removal and reapplication of EMG electrodes and to improve efficiency in data collection time. However, data from another ongoing study in our lab, in which we randomized the order of testing, indicate that the order of testing has negligible effects on the findings of the study.

In summary, this study provides proof of concept for the use of self-powered robotic devices for use in therapy. We designed and fabricated a device that mechanically linked the user's elbows via a cable transmission, and ran a human subjects experiment to determine the device's ability to influence motor slacking. Our results indicate that the device increased muscle activation and reduced slacking when it was driven with the subjects' own body power, especially when the subject was instructed to slack by resting their arm during the training task. Although further research is necessary to understand the effects of such training, our results lay the foundation to develop more sophisticated body-powered robots (e.g., with controllable transmissions) for rehabilitation purposes.

Supplementary Material

Supplemental Digital Content 1

Video Abstract. Video that provides a description of the device and an overview of the experiment and results. m4v

Click here to view.(47M, m4v)

2

Click here to view.(18K, pdf)

Acknowledgements

The research reported in this publication was supported by the National Institutes of Health (Grant# R01 EB019834). Edward Washabaugh and Emma Treadway were supported by the National Science Foundation Graduate Research Fellowship Program under Grant No. DGE #1256260. Any opinions, findings, and conclusions or recommendations expressed in this material are those of the author(s) and do not necessarily reflect the views of the funding sources. Additionally, we thank Robert Saner, M.S.; Katelyn Fry, M.S.E.; and the University of Michigan Orthotics and Prosthetics Center for their assistance in data collection and device design and fabrication, respectively. Anatomical images were modified with permission of Pearson Education, Inc. (Copyright © 2007 Pearson Education, Inc., publishing as Benjamin Cummings). The authors have no financial relationships that may pose a conflict of interest.

Article information

Restor Neurol Neurosci. Author manuscript; available in PMC 2019 Oct 28.

Published in final edited form as:

Restor Neurol Neurosci. 2018; 36(6): 693–708.

doi: 10.3233/RNN-180830

PMCID: PMC6817341

NIHMSID: NIHMS1055952

PMID: 30400120

Edward P. Washabaugh,[1,2] Emma Treadway,[3] R. Brent Gillespie,[3,4] C. David Remy,[3,4] and Chandramouli Krishnan[1,2,3,4,*]

[1]NeuRRo Lab, Department of Physical Medicine and Rehabilitation, Michigan Medicine, Ann Arbor, MI, USA.

[2]Department of Biomedical Engineering, University of Michigan, Ann Arbor, MI, USA.

[3]Department of Mechanical Engineering, University of Michigan, Ann Arbor, MI, USA.

[4]Michigan Robotics Institute, University of Michigan, Ann Arbor, MI, USA.

[*]**Corresponding author**: Chandramouli Krishnan, Director, Neuromuscular & Rehabilitation Robotics Laboratory (NeuRRo Lab), Department of Physical Medicine and Rehabilitation, University of Michigan, 325 E Eisenhower Parkway (Suite 3013), Ann Arbor, MI - 48108, Phone: (319) 321-0117, Fax: (734-615-1770); mouli@umich.edu

The publisher's final edited version of this article is available at Restor Neurol Neurosci

See other articles in PMC that cite the published article.

References

1. Abdollahi F, Case Lazarro ED, Listenberger M, Kenyon RV, Kovic M, Bogey RA, Hedeker D, Jovanovic BD and Patton JL (2014). Error augmentation enhancing arm recovery in individuals with chronic stroke: a randomized crossover design. Neurorehabilitation and Neural Repair, 28(2), 120–128. [PubMed] [Google Scholar]
2. Amirabdollahian F, Loureiro R, Gradwell E, Collin C, Harwin W and Johnson G (2007). Multivariate analysis of the Fugl-Meyer outcome measures assessing the effectiveness of GENTLE/S robot-mediated stroke therapy. Journal of Neuroengineering and Rehabilitation, 44. [PMC free article] [PubMed] [Google Scholar]
3. Bertolucci F, Chisari C and Fregni F (2018). The potential dual role of transcallosal inhibition in post-stroke motor recovery. Restorative Neurology and Neuroscience, 36(1), 83–97. [PubMed] [Google Scholar]
4. Biddiss EA and Chau TT (2007). Upper limb prosthesis use and abandonment: a survey of the last 25 years. Prosthetics and Orthotics International, 31(3), 236–257. [PubMed] [Google Scholar]
5. Brauchle D, Vukelic M, Bauer R and Gharabaghi A (2015). Brain state-dependent robotic reaching movement with a multi-joint arm exoskeleton: combining brain-machine interfacing and robotic rehabilitation. Frontiers in Human Neuroscience, 9564. [PMC free article] [PubMed] [Google Scholar]
6. Brochard S, Robertson J, Medee B and Remy-Neris O (2010). What's new in new technologies for upper extremity rehabilitation? Current Opinion in Neurology, 23(6), 683–687. [PubMed] [Google Scholar]
7. Byblow WD, Stinear CM, Smith MC, Bjerre L, Flaskager BK and McCambridge AB (2012). Mirror symmetric bimanual movement priming can increase corticomotor excitability and enhance motor learning. PloS One, 7(3), e33882. [PMC free article] [PubMed] [Google Scholar]
8. Caleo M (2015). Rehabilitation and plasticity following stroke: Insights from rodent models. Neuroscience, 311180–194. [PubMed] [Google Scholar]
9. Casadio M and Sanguineti V (2012). Learning, retention, and slacking: a model of the dynamics of recovery in robot therapy. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 20(3), 286–296. [PubMed] [Google Scholar]
10. Casadio M, Tamagnone I, Summa S and Sanguineti V (2013). Neuromotor recovery from stroke: computational models at central, functional, and muscle synergy level. Frontiers in Computational Neuroscience, 797. [PMC free article] [PubMed] [Google Scholar]
11. Chang CK, Washabaugh EP, Gwozdziowski A, Remy CD and Krishnan C (2018). A Semi-passive Planar Manipulandum for Upper-Extremity Rehabilitation. Annals of Biomedical Engineering, 46(7), 1047–1065. [PMC free article] [PubMed] [Google Scholar]
12. Coren S (1993). Measurement of handedness via self-report: the relationship between brief and extended inventories. Perceptual and Motor Skills, 76(3 Pt 1), 1035–1042. [PubMed] [Google Scholar]
13. Cunningham CL, Stoykov ME and Walter CB (2002). Bilateral facilitation of motor control in chronic hemiplegia. Acta Psychologica, 110(2–3), 321–337. [PubMed] [Google Scholar]
14. Duret C, Hutin E, Lehenaff L and Gracies JM (2015). Do all sub acute stroke patients benefit from robot-assisted therapy? A retrospective study. Restorative Neurology and Neuroscience, 33(1), 57–65. [PubMed] [Google Scholar]
15. Ellis MD, Holubar BG, Acosta AM, Beer RF and Dewald JP (2005). Modifiability of abnormal isometric elbow and shoulder joint torque coupling after stroke. Muscle and Nerve, 32(2), 170–178. [PMC free article] [PubMed] [Google Scholar]
16. Emken JL, Benitez R and Reinkensmeyer DJ (2007a). Human-robot cooperative movement training: learning a novel sensory motor transformation during walking with robotic assistance-as-needed. Journal of Neuroengineering and Rehabilitation, 48. [PMC free article] [PubMed] [Google Scholar]
17. Emken JL, Benitez R, Sideris A, Bobrow JE and Reinkensmeyer DJ (2007b). Motor adaptation as a greedy optimization of error and effort. Journal of Neurophysiology, 97(6), 3997–4006. [PubMed] [Google Scholar]
18. Gan Z, Fry K, Gillespie RB and Remy CD (2015). A novel variable transmission with digital hydraulics, Intelligent Robots and Systems (IROS), 2015 IEEE/RSJ International Conference on. IEEE, pp. 5838–5843. [Google Scholar]
19. Grefkes C, Eickhoff SB, Nowak DA, Dafotakis M and Fink GR (2008). Dynamic intra- and interhemispheric interactions during unilateral and bilateral hand movements assessed with fMRI and DCM. Neuroimage, 41(4), 1382–1394. [PubMed] [Google Scholar]
20. Harris-Love ML, McCombe Waller S and Whitall J (2005). Exploiting interlimb coupling to improve paretic arm reaching performance in people with chronic stroke. Archives of Physical Medicine and Rehabilitation, 86(11), 2131–2137. [PubMed] [Google Scholar]

21. Hidler J, Hamm LF, Lichy A and Groah SL (2008). Automating activity-based interventions: the role of robotics. Journal of Rehabilitation Research and Development, 45(2), 337–344. [PubMed] [Google Scholar]
22. Hogan N and Krebs HI (2011). Physically interactive robotic technology for neuromotor rehabilitation. Progress in Brain Research, 19259–68. [PubMed] [Google Scholar]
23. Huang VS and Krakauer JW (2009). Robotic neurorehabilitation: a computational motor learning perspective. Journal of Neuroengineering and Rehabilitation, 65. [PMC free article] [PubMed] [Google Scholar]
24. Israel JF, Campbell DD, Kahn JH and Hornby TG (2006). Metabolic costs and muscle activity patterns during robotic- and therapist-assisted treadmill walking in individuals with incomplete spinal cord injury. Physical Therapy, 86(11), 1466–1478. [PubMed] [Google Scholar]
25. Johnson MJ, Van der Loos HF, Burgar CG, Shor P and Leifer LJ (2005). Experimental results using force-feedback cueing in robot-assisted stroke therapy. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 13(3), 335–348. [PubMed] [Google Scholar]
26. Kantak S, Jax S and Wittenberg G (2017). Bimanual coordination: A missing piece of arm rehabilitation after stroke. Restorative Neurology and Neuroscience, 35(4), 347–364. [PubMed] [Google Scholar]
27. Kantak SS, Jones-Lush LM, Narayanan P, Judkins TN and Wittenberg GF (2013). Rapid plasticity of motor corticospinal system with robotic reach training. Neuroscience, 24755–64. [PMC free article] [PubMed] [Google Scholar]
28. Kleim JA and Jones TA (2008). Principles of experience-dependent neural plasticity: implications for rehabilitation after brain damage. Journal of Speech, Language, and Hearing Research, 51(1), S225–239. [PubMed] [Google Scholar]
29. Krishnan C and Dhaher Y (2012). Corticospinal responses of quadriceps are abnormally coupled with hip adductors in chronic stroke survivors. Experimental Neurology, 233(1), 400–407. [PMC free article] [PubMed] [Google Scholar]
30. Krishnan C, Kotsapouikis D, Dhaher YY and Rymer WZ (2013a). Reducing robotic guidance during robot-assisted gait training improves gait function: a case report on a stroke survivor. Archives of Physical Medicine and Rehabilitation, 94(6), 1202–1206. [PubMed] [Google Scholar]
31. Krishnan C, Ranganathan R, Dhaher YY and Rymer WZ (2013b). A pilot study on the feasibility of robot-aided leg motor training to facilitate active participation. PloS One, 8(10), e77370. [PMC free article] [PubMed] [Google Scholar]
32. Krishnan C, Ranganathan R, Kantak SS, Dhaher YY and Rymer WZ (2012). Active robotic training improves locomotor function in a stroke survivor. Journal of Neuroengineering and Rehabilitation, 957. [PMC free article] [PubMed] [Google Scholar]
33. Levin MF and Dimov M (1997). Spatial zones for muscle coactivation and the control of postural stability. Brain Research, 757(1), 43–59. [PubMed] [Google Scholar]
34. Levin MF, Selles RW, Verheul MH and Meijer OG (2000). Deficits in the coordination of agonist and antagonist muscles in stroke patients: implications for normal motor control. Brain Research, 853(2), 352–369. [PubMed] [Google Scholar]
35. Lin KC, Chen YA, Chen CL, Wu CY and Chang YF (2010). The effects of bilateral arm training on motor control and functional performance in chronic stroke: a randomized controlled study. Neurorehabilitation and Neural Repair, 24(1), 42–51. [PubMed] [Google Scholar]
36. Lotze M, Braun C, Birbaumer N, Anders S and Cohen LG (2003). Motor learning elicited by voluntary drive. Brain, 126(Pt 4), 866–872. [PubMed] [Google Scholar]
37. Loureiro RC, Harwin WS, Nagai K and Johnson M (2011). Advances in upper limb stroke rehabilitation: a technology push. Medical and Biological Engineering and Computing, 49(10), 1103–1118. [PubMed] [Google Scholar]
38. Luft AR, McCombe-Waller S, Whitall J, Forrester LW, Macko R, Sorkin JD, Schulz JB, Goldberg AP and Hanley DF (2004). Repetitive bilateral arm training and motor cortex activation in chronic stroke: a randomized controlled trial. JAMA: Journal of the American Medical Association, 292(15), 1853–1861. [PMC free article] [PubMed] [Google Scholar]
39. Marchal-Crespo L and Reinkensmeyer DJ (2009). Review of control strategies for robotic movement training after neurologic injury. Journal of Neuroengineering and Rehabilitation, 620. [PMC free article] [PubMed] [Google Scholar]
40. Millstein SG, Heger H and Hunter GA (1986). Prosthetic use in adult upper limb amputees: a comparison of the body powered and electrically powered prostheses. Prosthetics and Orthotics International, 10(1), 27–34. [PubMed] [Google Scholar]
41. Mudie MH and Matyas TA (2000). Can simultaneous bilateral movement involve the undamaged hemisphere in reconstruction of neural networks damaged by stroke? Disability and Rehabilitation, 22(1–2), 23–37. [PubMed] [Google Scholar]
42. Musampa NK, Mathieu PA and Levin MF (2007). Relationship between stretch reflex thresholds and voluntary arm muscle activation in patients with spasticity. Experimental Brain Research, 181(4), 579–593. [PubMed] [Google Scholar]
43. Patton JL, Kovic M and Mussa-Ivaldi FA (2006a). Custom-designed haptic training for restoring reaching ability to individuals with poststroke hemiparesis. Journal of Rehabilitation Research and Development, 43(5), 643–656. [PubMed] [Google Scholar]
44. Patton JL, Stoykov ME, Kovic M and Mussa-Ivaldi FA (2006b). Evaluation of robotic training forces that either enhance or reduce error in chronic hemiparetic stroke survivors. Experimental Brain Research, 168(3), 368–383. [PubMed] [Google Scholar]
45. Raghavan P, Aluru V, Milani S, Thai P, Geller D, Bilaloglu S, Lu Y and Weisz DJ (2017). Coupled bimanual training using a non-powered device for individuals with severe hemiparesis: a pilot study. International Journal of Physical Medicine and Rehabilitation, 5(3). [PMC free article] [PubMed] [Google Scholar]
46. Ranganathan R and Krishnan C (2012). Extracting synergies in gait: using EMG variability to evaluate control strategies. Journal of Neurophysiology, 108(5), 1537–1544. [PMC free article] [PubMed] [Google Scholar]

47. Reinkensmeyer DJ, Akoner O, Ferris DP and Gordon KE (2009). Slacking by the human motor system: computational models and implications for robotic orthoses. Conference Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, 20092129–2132. [PubMed] [Google Scholar]
48. Reinkensmeyer DJ and Patton JL (2009). Can robots help the learning of skilled actions? Exercise and Sport Sciences Reviews, 37(1), 43–51. [PMC free article] [PubMed] [Google Scholar]
49. Rong W, Li W, Pang M, Hu J, Wei X, Yang B, Wai H, Zheng X and Hu X (2017). A Neuromuscular Electrical Stimulation (NMES) and robot hybrid system for multi-joint coordinated upper limb rehabilitation after stroke. Journal of Neuroengineering and Rehabilitation, 14(1), 34. [PMC free article] [PubMed] [Google Scholar]
50. Rosen J, Brand M, Fuchs MB and Arcan M (2001). A myosignal-based powered exoskeleton system. IEEE Transactions on systems, Man, and Cybernetics-part A: Systems and humans, 31(3), 210–222. [Google Scholar]
51. Sans-Muntadas A, Duarte JE and Reinkensmeyer DJ (2014). Robot-assisted motor training: assistance decreases exploration during reinforcement learning. Conference Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, 20143516–3520. [PubMed] [Google Scholar]
52. Scott SH and Dukelow SP (2011). Potential of robots as next-generation technology for clinical assessment of neurological disorders and upper-limb therapy. Journal of Rehabilitation Research and Development, 48(4), 335–353. [PubMed] [Google Scholar]
53. Secoli R, Milot MH, Rosati G and Reinkensmeyer DJ (2011). Effect of visual distraction and auditory feedback on patient effort during robot-assisted movement training after stroke. Journal of Neuroengineering and Rehabilitation, 821. [PMC free article] [PubMed] [Google Scholar]
54. Shaw SE, Morris DM, Uswatte G, McKay S, Meythaler JM and Taub E (2005). Constraint-induced movement therapy for recovery of upper-limb function following traumatic brain injury. Journal of Rehabilitation Research and Development, 42(6), 769–778. [PubMed] [Google Scholar]
55. Simon AM, Brent Gillespie R and Ferris DP (2007). Symmetry-based resistance as a novel means of lower limb rehabilitation. Journal of Biomechanics, 40(6), 1286–1292. [PubMed] [Google Scholar]
56. Simpson D (1974). The choice of control system for the multimovement prosthesis: extended physiological proprioception (epp), in: Herberts P (Ed.), The control of upper-extremity prostheses and orthoses. Thomas, pp. 146–150. [Google Scholar]
57. Stein J (2009). e100 NeuroRobotic system. Expert Review of Medical Devices, 6(1), 15–19. [PubMed] [Google Scholar]
58. Stein J, Narendran K, McBean J, Krebs K and Hughes R (2007). Electromyography-controlled exoskeletal upper-limb-powered orthosis for exercise training after stroke. American Journal of Physical Medicine and Rehabilitation, 86(4), 255–261. [PubMed] [Google Scholar]
59. Stinear CM, Petoe MA, Anwar S, Barber PA and Byblow WD (2014). Bilateral priming accelerates recovery of upper limb function after stroke: a randomized controlled trial. Stroke, 45(1), 205–210. [PubMed] [Google Scholar]
60. Stinear JW and Byblow WD (2004). Rhythmic bilateral movement training modulates corticomotor excitability and enhances upper limb motricity poststroke: a pilot study. Journal of Clinical Neurophysiology, 21(2), 124–131. [PubMed] [Google Scholar]
61. Subramanian SK, Feldman AG and Levin MF (2018). Spasticity may obscure motor learning ability after stroke. Journal of Neurophysiology, 119(1), 5–20. [PMC free article] [PubMed] [Google Scholar]
62. Swinnen SP (2002). Intermanual coordination: from behavioural principles to neural-network interactions. Nature Reviews: Neuroscience, 3(5), 348–359. [PubMed] [Google Scholar]
63. Taub E, Uswatte G, Mark VW and Morris DM (2006). The learned nonuse phenomenon: implications for rehabilitation. Europa Medicophysica, 42(3), 241–256. [PubMed] [Google Scholar]
64. Taub E, Uswatte G and Pidikiti R (1999). Constraint-Induced Movement Therapy: a new family of techniques with broad application to physical rehabilitation--a clinical review. Journal of Rehabilitation Research and Development, 36(3), 237–251. [PubMed] [Google Scholar]
65. Treadway E, Gan Z, Remy CD and Gillespie B (2018). Toward Controllable Hydraulic Coupling of Joints in a Wearable Robot. IEEE Transactions on Robotics, PP(99), 1–16. [PMC free article] [PubMed] [Google Scholar]
66. Uswatte G and Taub E (2013). Constraint-induced movement therapy: a method for harnessing neuroplasticity to treat motor disorders, Progress in Brain Research. Elsevier, pp. 379–401. [PubMed] [Google Scholar]
67. Washabaugh E, Guo J, Chang CK, Remy D and Krishnan C (2018). A Portable Passive Rehabilitation Robot for Upper-Extremity Functional Resistance Training. IEEE Transactions on Biomedical Engineering. [PMC free article] [PubMed] [Google Scholar]
68. Washabaugh EP, Claflin ES, Gillespie RB and Krishnan C (2016). A Novel Application of Eddy Current Braking for Functional Strength Training During Gait. Annals of Biomedical Engineering, 44(9), 2760–2773. [PMC free article] [PubMed] [Google Scholar]
69. Washabaugh EP and Krishnan C (2018). A wearable resistive robot facilitates locomotor adaptations during gait. Restorative Neurology and Neuroscience, 36(2), 215–223. [PMC free article] [PubMed] [Google Scholar]
70. Whitall J, McCombe Waller S, Silver KH and Macko RF (2000). Repetitive bilateral arm training with rhythmic auditory cueing improves motor function in chronic hemiparetic stroke. Stroke, 31(10), 2390–2395. [PubMed] [Google Scholar]
71. Wirz M, Zemon DH, Rupp R, Scheel A, Colombo G, Dietz V and Hornby TG (2005). Effectiveness of automated locomotor training in patients with chronic incomplete spinal cord injury: a multicenter trial. Archives of Physical Medicine and Rehabilitation, 86(4), 672–680. [PubMed] [Google Scholar]

72. Wolf SL (2007). Revisiting constraint-induced movement therapy: are we too smitten with the mitten? Is all nonuse "learned"? and other quandaries. Physical Therapy, 87(9), 1212–1223. [PubMed] [Google Scholar]