Inducing Self-Selected Human Engagement in Robotic Locomotion Training

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***Abstract*—Stroke leads to severe mobility impairments for millions of individuals each year. Functional outcomes can be improved through manual treadmill therapy, but high costs limit patient exposure and, thereby, outcomes. Robotic gait training could increase the viable duration and frequency of training sessions, but robotic approaches employed thus far have been less effective than manual therapy. These shortcomings may relate to subconscious energy-minimizing drives, which might cause patients to engage less actively in therapy when provided with corrective robotic assistance. We have devised a new method for gait rehabilitation that harnesses, rather than fights, least-effort tendencies. Therapeutic goals, such as increased use of the paretic limb, are made easier than the patient's nominal gait through selective assistance from a robotic platform. We performed a pilot test on a healthy subject (N = 1) in which altered self-selected stride length was induced using a tethered robotic ankle-foot orthosis. The subject first walked on a treadmill while wearing the orthosis with and without assistance at unaltered and voluntarily altered stride length. Voluntarily increasing stride length by 5% increased metabolic energy cost by 4%. Robotic assistance decreased energy cost at both unaltered and voluntarily increased stride lengths, by 6% and 8% respectively. We then performed a test in which the robotic system continually monitored stride length and provided more assistance if the subject's stride length approached a target increase. This adaptive assistance protocol caused the subject to slowly adjust their gait patterns towards the target, leading to a 4% increase in stride length. Metabolic energy consumption was simultaneously reduced by 5%. These results suggest that selective-assistance protocols based on targets relevant to rehabilitation might lead patients to self-select desirable gait patterns during robotic gait training sessions, possibly facilitating better adherence and outcomes.**

I. Introduction

Mobility impairments caused by stroke can be mitigated by intensive manual treadmill therapy, but the costs of this treatment limit patient exposure and outcomes. Stroke affects 2.6% of adults in the United States each year [1], resulting in decreased walking speed, gait asymmetry, and increased energetic cost [2–4]. Manual gait training, in which therapists assist patient leg motions during treadmill walking, results in significant improvements, with outcomes tied to session duration and frequency [5–8]. Unfortunately, the demands of manual treadmill therapy restrict patient training time and, therefore, limit the resulting outcomes [9, 10].

Robotic rehabilitation offers several potential advantages over manual treadmill therapy. Robotic platforms can be more precise, repeatable, responsive, and powerful than humans; they can measure many signals that are inaccessible to a human therapist, such as electromyographic activity; they can endure longer, more intense, and more frequent training sessions; and the operating cost of a robotic platform are lower than that of equivalent throughput by teams of therapists [11]. In the long term, robotic gait training will therefore likely provide more effective rehabilitation than manual therapy.

Shortcomings in rehabilitation robotics technology have, as yet, prevented realization of these advantages. Robotic gait trainers are often structured around a trajectory control approach [12, 13], limiting their capabilities to the enforcement of predefined limb motions. While this approach captures some aspects of therapist-patient interaction, other potentially-crucial functions are neglected. For instance, therapists can provide verbal instructions and feedback, encouraging a patient to try harder and guiding coordination patterns. Interaction forces between the therapist's hands and the patient's legs might also encode important information, for instance related to the appropriateness of muscle activity [14]. This may partially explain why, despite some mixed initial results [15], multicenter randomized studies have shown robotic gait training to under-perform manual therapy [16, 17].

Robotic gait trainers with force control have sought to overcome these limitations and enable new rehabilitation methods. Inspired by upper-extremity haptic tools [e.g. 18], these machines use series elasticity to improve force control and backdrivability [19–22]. We have developed a tethered, torque-controlled, robotic ankle-foot orthosis with significantly higher torque bandwidth and lower mass than prior platforms [23, 24]. These properties are critical to effective intervention [25]; low bandwidth leads to undesired forces under rapidly-changing conditions [26, 27], while heavy orthoses interfere with limb motions and increase effort [28]. These improvements allow more precise control of human-robot interactions during treadmill therapy, and may enable new approaches to robotic gait training.

The tendency for humans to minimize energy consumption during locomotion might help to explain the shortcomings of conventional robotic therapy and to suggest new strategies for intervention. It has long been observed that, within task and physiological constraints, people tend to walk in ways that minimize energy use [29, 30], in terms of, e.g., walking speed [31], step length [32], step width [33], or even arm movement [34]. This 'least-effort' drive holds true for individuals with disabilities such as amputation [35], Spina

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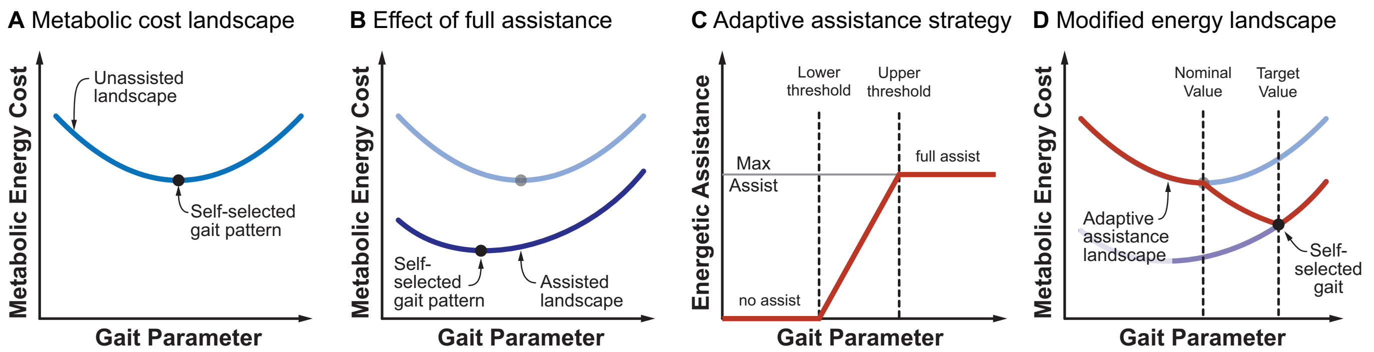

Fig. 1. Schematic representation of least-effort gait shaping methodology using hypothetical energy cost landscapes. **A.** We expect that patients will select to walk with gait parameters, such as step length or muscle activity level, that result in the least effort overall, indicated by metabolic energy consumption. **B.** Assistance from a robotic device could reduce effort and alter the landscape, but might not result in a desirable optimum. **C.** Using modulation of the amount of assistance, we can make reductions in effort a function of desirable gait parameter changes. **D.** This results in a new landscape, in which the least-effort coordination pattern corresponds to improvements in selected parameters, such as increased symmetry or paretic-limb muscle use.

bifida [36], and Down's syndrome [37]. Although the long-term energetic effects of improved mechanics on hemiparetic gait are clear [38], little is known about the immediate effects of, e.g., increased paretic-limb muscle activity. It may be that patients exhibiting hemiparetic gait are moving in ways that are (transiently) easiest, or feel easiest due to disruption of sensory pathways, and that conventional therapies must fight least-effort drives. During trajectory-controlled robotic gait rehabilitation, least-effort drives may act against therapeutic goals; with corrective control, less effort will result in greater robotic assistance, incentivizing reduced muscle use and not reinforcing weakened neural pathways. With improved control, perhaps robotic interventions could instead harness least-effort drives to induce engagement [39].

Here we describe an alternative approach to robotic rehabilitation which is intended to induce active engagement rather than enforce predefined trajectories. We present results from a pilot test that suggest that this approach could be used to shape patient behavior during training sessions, possibly leading to better outcomes of robotic gait rehabilitation.

II. Methods

We devised a technique for inducing self-selected changes in human gait, implemented it using a robotic ankle-foot orthosis, and performed a pilot test on a single healthy subject.

A. Energy-Cost Landscape Manipulation

We have formulated a method for systematically altering the relationship between gait parameters and metabolic energy cost. Humans tend to walk with gait parameters, such as speed or step length, that minimize energetic cost [e.g. 29, 30]. Enforcing altered gait tends to increase energy cost, creating bowl-like 'landscapes' of energy cost vs. gait parameter, with preferred gait at the lowest point (Fig. 1 A). We hypothesize that these landscapes could be manipulated by a rehabilitation robot, such that the minimum is moved to a more desirable location. During stroke rehabilitation, for instance, target gait parameters might encode greater symmetry or more appropriate paretic-limb muscle activity. We further hypothesize that under such conditions, humans will prefer, and self-select, to walk with the new optimal gait parameter value. In this way, patients could be subconsciously encouraged to engage more actively and suitably during robotic gait training sessions.

We propose a strategy of selective assistance to obtain desirable changes in the energy-cost landscape. Any mechanical intervention will alter the relationship between energy cost and gait parameters, but not necessarily in a desirable way. During continuous robotic assistance, for example, the optimal human coordination strategy might be to reduce all muscle activity somewhat (Fig. 1 B). Instead, we suggest providing energy-saving assistance only when desirable changes are observed, and in proportion to those changes (Fig. 1 C). This proportional assistance will move the optimal gait parameter toward the target value (Fig. 1 D). Proportional resistance could also be applied for changes in the opposite direction, increasing the energy cost of undesirable coordination patterns. Baseline assistance could be provided to allow nominal walking ability at early stages if necessary, and target parameters could be slowly adjusted as a patient's locomotor performance improved during the course of therapy. Dynamic coupling between selectively-applied assistance and the targeted gait parameter should also be considered, as this could augment, or interfere with [40], convergence to the new optimum.

B. Tethered Robotic Ankle-Foot Orthosis

Manipulating the energy-cost landscape requires a tool capable of selectively reducing energy cost. We previously developed a tethered robotic ankle-foot orthosis (Fig. 2) capable of precisely controlling and varying ankle joint torques, described in detail in [23, 41]. This platform provides a unique combination of low worn mass, high peak torque, and high torque bandwidth. A unilateral Bowden-cable tether allows zero ankle impedance when desired and provides very little interference with leg motions, verified in tests with a leg-like pendulum. A leaf spring provides series elasticity. Numerous hardware and software features protect human participants.

Assistance was provided through ankle push-off work done by the tethered orthosis. We have previously shown that

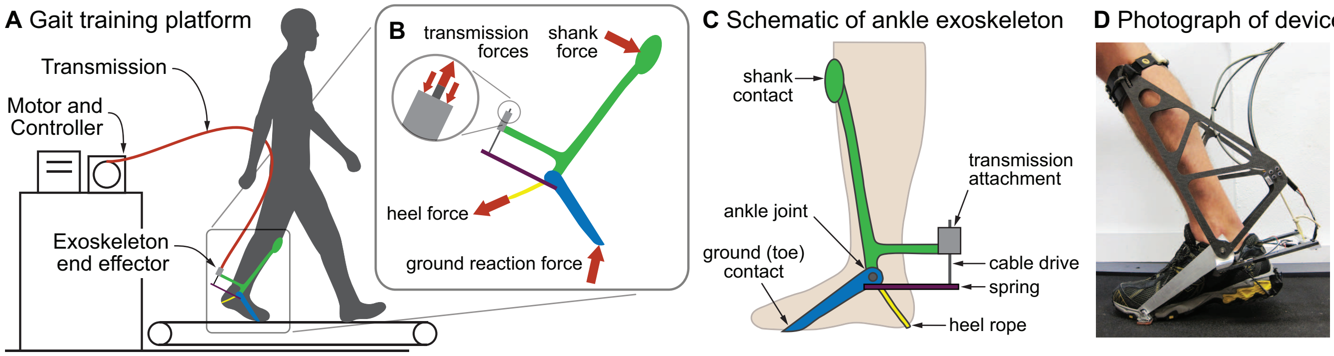

Fig. 2. Robotic gait trainer used in pilot study. **A.** The system comprises: (1) powerful off-board motor and control hardware, (2) a flexible tether transmitting mechanical power and sensor signals, and (3) a lightweight instrumented orthosis. **B.** Free-body diagram of the orthosis. An ankle plantarflexion torque is produced on the person's leg by pushing back on the proximal tibia, up on the heel, and down on the ground at the toe. **C.** Robotic orthosis schematic. The Bowden cable and fiberglass leaf spring provide an effect similar to the Achilles tendon. **D.** Photograph of the orthosis end-effector (mass = 0.53 kg).

increased prosthetic ankle push-off can reduce user effort [42]. Here, we augmented ankle push-off using an active orthosis. During stance, the motor rotated at a constant velocity, v_m, starting from a predefined initial angle, θ_0. The combination of motor and ankle displacements stretched the series spring, generating a varying ankle plantarflexion torque over the stance period. We tuned v_m and θ_0 until the natural dynamical interactions between human, spring, and motor during a typical step resulted in a peak orthosis torque of about half that observed for normal walking [similar to the tuning process in 43]. We always used positive values of v_m, resulting in net positive work provided to the user. During the swing phase, the orthosis provided zero impedance by maintaining slack in the transmission cable. Foot switches at the heel and toe were used to detect ground contact and switch between modes.

In selective the assistance mode, the level of assistance on each step was set by scaling motor control parameters as a function of the measured, desired, and nominal gait parameter values (Fig. 1 C). We first measured the nominal gait parameter (e.g. self-selected stride length) in a trial without the device. During selective assistance, we measured the gait parameter p on each step and compared it to the nominal, p_{nom}, and desired, p_{des} values. For p greater than p_{des} or less than p_{nom}, the scaling factor k was set to 1 or 0, respectively. Otherwise, k was set to $(p - p_{nom}) \cdot (p_{des} - p_{nom})^{-1}$. Motor control parameters were then set to $v'_m = k \cdot v_m$ and $\theta'_0 = k \cdot \theta_0$ for the ensuing stance period.

C. Experimental Methods

We performed pilot tests on a single healthy subject (N = 1, 72 kg, 0.90 m leg length, 22 yrs.) walking on a treadmill at 1.25 m$\cdot$s^{-1} while wearing the robotic orthosis on one ankle. Stride length was calculated from belt speed and stride period, measured using foot switches. Metabolic energy expenditure was calculated [44] using data from sampled-gas indirect respirometry, with quiet standing used as a baseline.

In the Landscape experiment, we tested whether the robotic orthosis could manipulate the energy cost landscape with respect to stride length. The subject first walked with self-selected gait and Nominal stride length was determined. The subject then walked while voluntarily maintaining a target stride length, with visual feedback provided on a monitor (Fig. 3, similar to [38, 45]). Four conditions were then applied in random order: (1) Nominal stride length, Unassisted, (2) Increased stride length, Unassisted, (3) Nominal stride length, Assisted, and (4) Increased stride length, Assisted. Increased stride length was defined as 5% greater than Nominal. During Unassisted conditions, the orthosis produced no torque. During Assisted conditions, maximal assistance was applied ($k = 1$). Each condition lasted 10 minutes to allow for subject adaptation. The subject was presented with all conditions on a single training day, two days prior to collection.

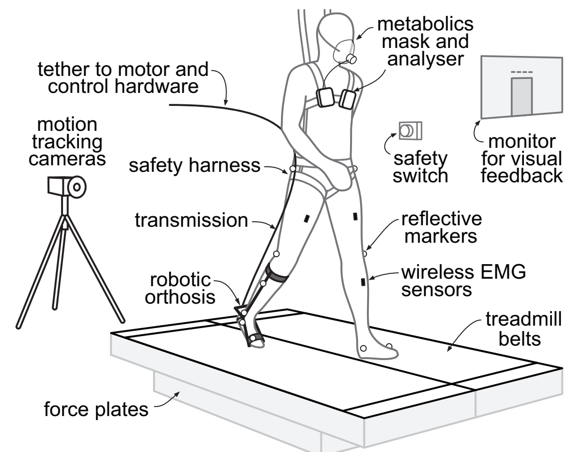

Fig. 3. Experimental setup. The pilot subject walked on a treadmill while wearing the tethered robotic ankle-foot orthosis on one leg. Metabolic energy expenditure was measured using a wearable indirect respirometry system. Stride length was calculated from stride time, as measured by foot switches.

In the Adaptive experiment, we tested whether selective assistance from the robotic orthosis would lead to self-selected changes in stride period. We used the nominal stride period determined in the first experiment, and set desired stride length to 5% above nominal. The subject then walked for 10 minutes while the orthosis provided assistance as a function of stride period as described above. The subject was naïve to this Adaptive controller. We compared stride length and metabolic rate from the 1^{st} and 10^{th} minutes of this condition.

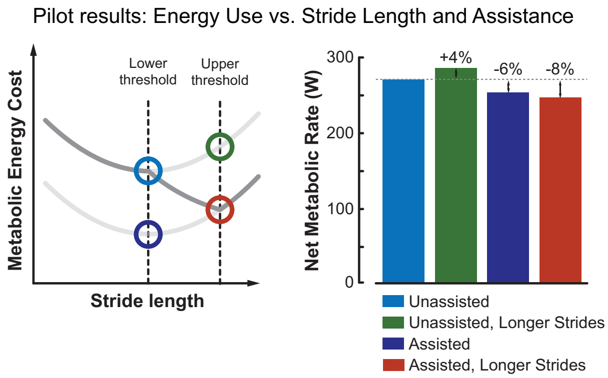

Fig. 4. Preliminary results from a test on a healthy subject, in which we sampled four points on the effort landscape: (Blue) Unassisted at Nominal stride length; (Green) Unassisted at Increased stride length; (Purple) Assisted at Nominal stride length; and (Red) Assisted at Increased stride length. During all conditions, subjects maintained a target stride length using visual feedback. During Assisted conditions, the robotic orthosis always provided maximal assistance. We found that voluntarily increasing stride length by 5% led to a 4% increase in energy cost. Applying robotic assistance reduced energy use by 6%. Doing both led to a net reduction of 8%.

III. Results

Mean stride length during self-selected Unassisted walking was 1.34 $\pm$ 0.02 m, with a corresponding net metabolic rate of 271 W (Fig. 4). Voluntarily increasing stride length by 5% using visual feedback resulted in a stride length of 1.40 $\pm$ 0.02 m, increasing net metabolic rate to 282 W.

With appropriate motor control parameters, the robotic ankle-foot orthosis provided significant torque and mechanical work on each step (Fig. 5 D). While voluntarily maintaining Nominal or Increased stride length, this assistance resulted in net metabolic rates of 254 W and 249 W, respectively (Fig. 4).

Adaptive assistance from the robotic orthosis caused the subject to slowly increase stride length towards the target value (Fig. 5 A). During the final minute of the test, stride length was 1.39 $\pm$ 0.02 m, a 4% increase over Nominal (Fig. 5 B). This corresponded to a net metabolic rate of 256 W, a 5% reduction (Fig. 5 C), and net work of 22.5 $\pm$ 1.1 J performed by the robotic orthosis each step (Fig. 5 D).

IV. Discussion

We devised a strategy for inducing desirable self-selected changes in human gait, implemented it using a tethered robotic ankle-foot orthosis, and performed pilot tests on a single subject in which stride length was targeted. We found that the robotic orthosis was capable of manipulating the energy-cost landscape, such that walking with target stride length and robotic assistance required less metabolic energy than walking with nominal stride length and no assistance. When this altered landscape was applied during freely-selected treadmill walking, the subject slowly adjusted their stride length towards the target value. These results are consistent with the idea that least-effort drives could be used to shape human behavior during robotic gait training.

The net energetic benefit of robotic assistance at increased step length observed in this study was likely due to a combination of reduced ipsilateral ankle plantarflexor force and reduced muscle work throughout the body. With assistance (Fig. 5 D) the orthosis contributed more than half the ankle torque expected for normal walking, and nearly four times the net joint work [46]. Proximate ankle muscles were likely less active, consuming less energy. Despite the fact that increased stride length was observed to increase energy cost, consistent with prior studies, robotic assistance more than compensated for mechanical disadvantages. Although more mechanical work may have been performed by the system as a whole, less of it seems to have been done by the human. This partitioning of effort resulted in a new optimum coordination pattern for the human. Although we report results from only one subject, we have performed a variety of pilot tests with similar protocols, including targeting changes in muscle activity, and consistently found that metabolic rate could be altered in this way.

Self-selected gait changes during the Adaptive experiment were consistent with the finding that humans tend to walk in ways that minimize energy cost, but additional dynamics may have played an important role. For example, humans initially walk with suboptimal speed when visual flow is manipulated, but then slowly converge to the optimal speed [47]. In experiments with dynamically-controlled treadmill speed, by contrast, humans do not discover the optimal cadence when it differs from their preferred speed-cadence relationship [40]. These differences may be related to the perception of device function as an externality, or to dynamic coupling between the device's input to the human and the human gait measurements used as input to the device controller. In the present study, increased ankle push-off provided by the exoskeleton could have caused a tendency toward longer strides, creating a positive feedback loop. Such dynamics could explain the observed trend in stride length over time, rather than energy cost minimization. For the purposes of inducing desirable patient activity during rehabilitation sessions either effect would have utility, while energy reductions might be more related to long term adherence. Additional experiments, for example in which an opposite change in stride length were targeted, would lend insight into the role of such coupling.

These results should be taken as promising initial findings, since the protocol was only performed on a single subject. More subjects must be tested, and appropriate statistical tools applied, before these trends are established. Other biomechanics measures must be considered, including joint torques and electromyographic activity, to understand the mechanisms underlying such trends. Nonetheless, our findings are consistent with feasibility of the proposed rehabilitation approach.

One might consider this approach analogous to positive verbal feedback in manual therapy, with social rewards for increased physical effort. It is different from existing robotic training, in which symmetric kinematics may be enforced, but not underlying neural and muscular activity. Although we have observed aftereffects in catch trials, our primary goal is improved activity *during* training. This differentiates the approach from split-belt treadmill training [e.g. 48], in which aftereffects demonstrate (temporary) desirable changes in gait parameters, but training activities are asymmetric.

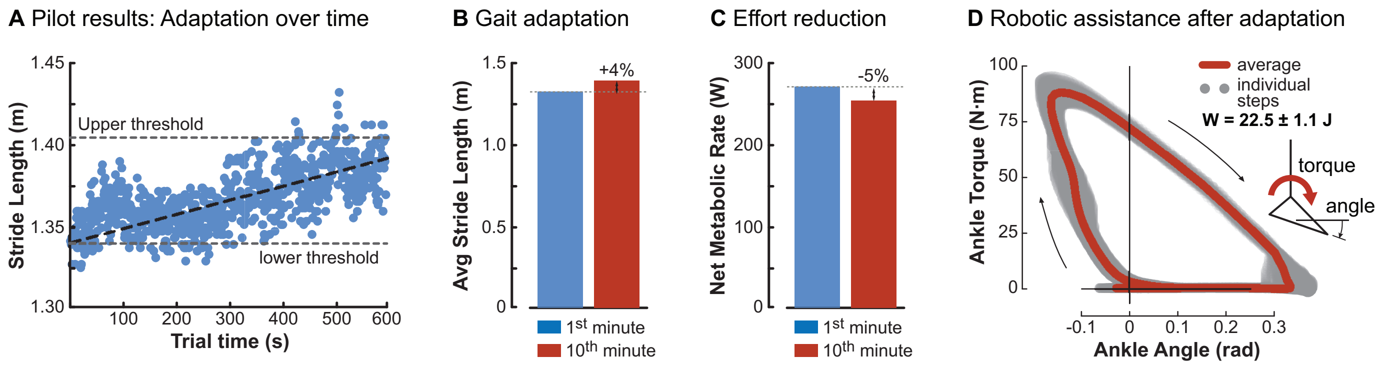

Fig. 5. Preliminary results show self-selected changes in gait parameters under Adaptive assistance. A healthy, naïve subject wore the robotic ankle orthosis as they walked on a treadmill at 1.25 m·s^{-1}. The target in this pilot test was to increase stride length. The subject was provided no direct feedback about stride length, the target, or robotic assistance. **A.** The subject slowly adjusted their gait pattern towards the rehabilitation target over the course of the trial. **B.** This resulted in increased self-selected stride length by the end of the trial. **C.** Energy use decreased during adaptation, driving the trend toward the desired behavior. Increasing stride length without the adaptive interventions would have, instead, required more effort and fought the pseudo-therapeutic goal of the test. **D.** In the final minute of the trial, the robotic orthosis provided significant plantarflexion torque and mechanical work to assist push-off during each step.

V. Conclusions and Future Work

Robotic gait training has the potential to provide increased therapeutic exposure to patients recovering from neurological injuries, if problems with active engagement can be overcome. We have described a method for inducing desirable changes in self-selected human gait suitable for use in robotic rehabilitation. Pilot data suggest that selective robotic assistance can alter the relationship between overall effort and gait parameters, and that subjects adopt the optimal gait pattern.

With refinement, this approach could be applied to individuals with mild impairment arising from stroke. Gait symmetry or appropriateness of electromyographic activity in the paretic limb could be used as target parameters, with robotic assistance applied to the contralateral limb. Proper gains might make increased symmetry or use of the paretic limb easier than the nominal coordination pattern. This could increase use of paretic-limb muscles and pathways during training sessions, possibly improving outcomes. Over the course of therapy, neural pathways and muscles could strengthen and the thresholds for assistance be adjusted. Eventually, wearable devices [e.g. 49] could apply adaptive assistance continually, providing greater exposure.

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