Quantifying Human Autonomy Recovery During Ankle Robot-Assisted Reversal of Foot Drop After Stroke

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***Abstract*—In this paper, we present a measure to quantify the profile of human autonomy recovery during ankle robot-assisted locomotor training to reverse foot drop after hemiparetic stroke. Underlying this kinetic-based measure is the estimation of the human ankle torque contribution during dynamic (step-by-step) patient-robot interaction, by measuring the device peak torques that meet desired design criteria across gait cycles. As with our previous clinical trials with the ankle robot, we employ an adaptive control approach for a six-week treadmill intervention in chronic stroke patients with well-defined ankle deficits. Here, we provide the first evidence of ankle robot-mediated changes in human autonomy index (HAI) and assess the worthiness of the HAI to predict changes in key volitional ankle neuromotor and whole-body functional gait outcomes. We also propose ways to incorporate the HAI into practice, including a model to customize the duration of robotic therapy for each patient based on minimally acceptable improvements in outcome measures.**

I. Introduction

Stroke is the most prevalent CNS etiology for physical disability, with about half of all patients experiencing persistent mobility impairments [1]. This problem is magnified in our aging population, projected to double its clinical stroke rates from 800,000 to 1.6 million clinical strokes per year by 2040 [1]. In the lower extremity (LE), the ankle is often adversely impacted to reduce gait speed, safety, postural balance, and lower limb motor control, rendering mobility recovery slow, unstable, and energetically costly [2-5]. Even after cessation of conventional therapy, ankle deficits afflict a large proportion (~30%) of chronic stroke survivors, leaving them no recourse beyond living with assistive devices (ADs) that provide safety for ambulation but also magnify abnormal compensatory behaviors, doing nothing to reverse ankle impairments.

In stark contrast to long-held beliefs that the chronic stroke phase (>6 months post-stroke) is impervious to functional improvement, numerous studies have shown that different gait training modalities can pierce that supposed recovery window ceiling. For example, partial body-weight supported treadmill and unsupported aerobic treadmill training (TM) improve gait velocity and fitness, even years after stroke [6,7]. However, meta-analyses show that gait interventions, whether plain exercise or robot-aided exercise, produce only modest incremental gains in gait velocity in those with mild-to-moderate gait deficits [8,9]. Simply put, we currently cannot re-engineer neuromotor recovery of activities of daily life (ADL) mobility in a major, life-changing manner using current locomotor therapies. Thus, according to the present state of neurorehabilitation science and practice, locomotor training can increase fitness and enable stroke patients to limp faster, but does not robustly improve underlying gait biomechanics or reduce dependence on ADs, with lingering controversy about whether task-specificity remains important during recovery. We address this gap using deficit severity adjusted adaptive control ankle robotics aimed at the hemiparetic ankle, and now seek to define the motor learning metrics that define its efficacy.

We have developed an impedance-controlled, highly backdrivable, 2 degree-of-freedom (DOF) actuated ankle robot ("Anklebot") to improve walking and balance functions after stroke [10], and has been the subject of numerous clinical trials [11-13]. The rationale to focus on the ankle was due to the critical role it plays in the biomechanics of gait and balance—the ankle musculature provides propulsion during mid-to-late stance, ground clearance during swing, and "shock absorption" during the landing phase of gait. Following a stroke, some (or all) of these ecological aspects of gait are disrupted. For example, "foot drop " is a common impairment caused by a weakness in the dorsiflexor muscles that lift the foot resulting in the slapping of the foot upon contact ("foot slap"), or even more dangerous, initial contact at the toes or lateral foot (as opposed to heel contact) that increases fall-risk and injuries. Another major deficit is reduced impulse at the hemiparetic ankle, which reduces gait forward propulsion, impair gait efficiency, and limits righting and reaction responses during balance perturbations to increase fall and injury risk. In our most recent study [13], this was exemplified by negative baseline impulses, indicating that as a group, the hemiparetic ankle-foot complex was serving as a brake, rather than for propulsion. The Anklebot can mediate motor learning for all of these ankle movement deficits, since it can be actuated in 2-DOFs [10], and is able to independently target multiple deficits (characterized by gait sub-tasks) within the stance and swing phases, by shaping the assistance profile across the gait cycle to accommodate the heterogeneity of initial hemiparetic stroke deficits and the profile of dynamic recovery.

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Recently, we first conducted the first randomized study to test the efficacy of Anklebot with an adaptive control architecture integrated into treadmill training versus impairment-focused seated Anklebot training (TMR). This study investigated the hypothesis that 6 weeks of TMR is more effective than matched dose impairment focused seated robotic training (SRT) across the paretic ankle to durably improve unassisted overground gait function and safety [13]. The primary finding of that study was that task-specific locomotor training (TMR), but not isolated massed practice across the affected joint (SRT), durably improves gait biomechanics and paretic ankle function during independent walking in chronic stroke survivors [13]. TMR progressively increased unassisted paretic swing to normal levels with retained improvements 6 weeks after cessation of training such that a majority of TMR graduates self-discarded their ankle braces. The significant increase in propulsive impulse with TMR to near-normal levels at retention, versus no change in SRT, contributes to ongoing increases in gait velocity after training ended. To our knowledge, this is the first therapy, robotic or otherwise, to therapeutically improve functional dorsiflexion and restore impaired push-off during independent walking in chronic stroke enabling individuals to self-reduce reliance on their assistive devices. These results support the *overarching* hypothesis that in the chronic phase post-stroke, locomotor task-integrated Anklebot training is superior to impairment-focused massed practice across the paretic joint for improving gait function, and raises the possibility that adaptive control ankle robotics could be applied to other central and peripheral neuromuscular etiologies for foot drop, beyond just stroke.

Our findings that 18 sessions of TMR can reverse foot drop across 6 weeks provides the first evidence that robot-mediated motor learning is occurring at the ankle. However, no prior studies, including ours, have determined a metric to quantify the temporal profile of human vs. robotic contribution across this motor learning phase, nor have any mathematical predictors of lower extremity motor learning rates been described to more efficiently plan the therapeutic trajectory required to produce autonomous functional recovery. An implicit assumption is that individuals with hemiparetic stroke must develop complete (i.e. 100%) autonomous volitional control in order to reverse their foot drop. However, it is known for individuals with motor incomplete spinal cord injury that as few as 15% of cortico-spinal axons are needed to produce some degree of ambulatory capacity. Although baseline vs. post training provide proof of ankle motor learning, the degree to which persons with hemiparetic stroke must recover autonomous volitional control in the setting of human-robot interaction to achieve such functional gains, and how to measure this, has not previously been examined. In this paper, we present a novel metric to quantity the human autonomy during robot-assisted gait therapy. We also present a model to predict the optimal number of robot training sessions to achieve desired effects on human autonomy, enabling: a) customization of robotic therapies for stroke patients; b) predicting profiles of recovery; and c) utilization of the index in comparative efficacy studies against other therapies and in other populations with ankle motor deficits secondary to central and peripheral neuromuscular, and orthopedic conditions.

II. Gaps In Robotics To Improve Gait Post-Stroke

Despite a renewed research thrust into locomotor-based robotics in neurological disease, conceptual approaches and technologies are widely divergent. One approach emphasizing whole body exoskeleton gait patterning has been shown to modestly improve fitness and gait in non-controlled studies. However, in randomized studies such patterned locomotor training has not shown advantage over standard PT, and in fact may be inferior [14,15]. This is partly due to the "stiff" nature of these robots, primarily focusing on repetitive, multi-joint patterning of gait cycles, while constraining natural limb dynamics, especially at the affected ankle. From a motor learning standpoint, current robotic gait trainers do not address potential over reliance on assistance from the robot to leverage volitional motor learning, and do not actuate (power) the ankle at all, but rather use it as a stable base to maintain balance. Hence, the ankle is the ignored joint in neuro-rehabilitation robotics. With impedance control [16-18] and other mechatronics advances, robotics now offers a flexible platform for re-engineering mobility recovery by leveraging motor learning. In stroke, impedance-controlled *arm* robotics that facilitate motor learning with a safe and spring-like programmable patient-robot interface have proven effective [19] and less expensive [20]; altering U.S. national care standards [21,22]. Conversely, LE robotics remain controversial, with consensus that current approaches are inferior to usual care, or possibly deleterious [21,22]. Although the technologies are sophisticated, the operating principles are not aligned with motor learning. Recognizing this gap and the key role of the ankle in gait and balance, we developed the Anklebot, designed to increase paretic ankle contributions to gait and balance functions in persons with hemiparesis [10]. The Anklebot accommodates the spectrum of motor deficit severities (Fig. 1); back-drivable hardware makes the robot "transparent" to "get out of the way" when the human performs the whole movement task. We first tested a seated, visually evoked/guided Anklebot protocol which demonstrated ankle motor learning in both sub-acute and chronic stroke [11,12], with carryover to improve gait speed and patterning (spatial-temporal inter-limb symmetry).

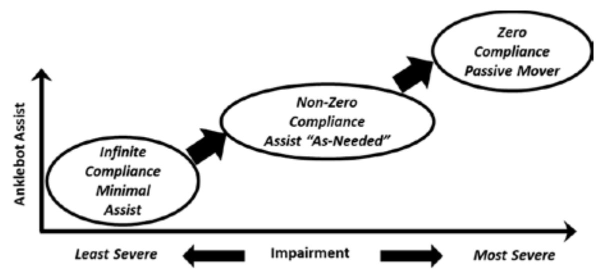

Figure 1. Impedance control principle: robotic assist vs. impairment.

III. Deficit-Adjusted Adaptive Ankle Robotics

To meet the requirements for ankle robotic locomotor therapy for stroke and other neurologic diseases, we developed adaptive control to precisely time robotic activation to sub-events and ankle directionality across the gait cycle [23,24]. This customizes robotic therapy to individual gait deficits (e.g., foot drop, weak push-off) and precisely timed impedance control support, in turn affording

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safety and maximum autonomy to effectively integrate the Anklebot into the context of locomotor learning (Fig. 2). A novel feature of the adaptive controller is the use of robust systems control that tolerates perturbations due to step-to-step variability, thereby increasing operational stability, and accommodates heterogeneous levels of mobility across the spectrum of stroke recovery [23]. The underlying concept is to precisely time robotic assistance, if needed, to gait sub-events derived from real-time signals via bilateral micro-switch insoles. This breakthrough allows, for the first time, the ability to: (1) Differentially target deficits in both stance and swing phases of gait, enabling tailoring of robotic assistance to individual deficits (e.g., foot drop, weak push-off, improper landing) via gait sub-event triggered actuation, (2) Tolerate step-to-step variability to prevent destabilization and ensure patient safety, (3) Progress and dynamically modulate robotic outputs, both in the immediate time frame (step-to-step) and over the course of therapy (inter-visit), as subject performance adapts. This novel *co-robotics* application defines a cooperative learning process between the subject and robot, which over time dynamically shapes robotic outputs to promote and elicit greater volitional effort toward durable locomotor learning. To our knowledge, TMR constitutes the only therapy to reverse foot drop, restore paretic leg propulsion, and correct heel-first landing in persons with chronic stroke. Hence, adaptive control co-robotics holds promise to shift practice paradigms for the care of hemiparetic stroke, providing a bioengineering solution to a previously immutable neurological deficit.

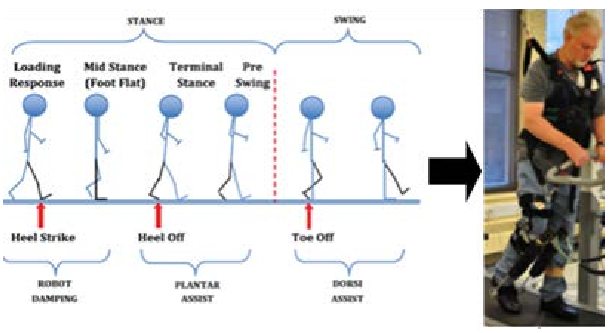

Figure 2. Adaptive timing control approach: gait sub-events are detected by foot switches to inform the ankle robot to precisely time robotic assistance to key deficits characterized by those sub-events.

IV. Human Autonomy Index

The human autonomy index is a way to quantify kinetic-based interaction between the patient and the robotic device during robot-assisted locomotor training. The human autonomy index (N_{HA}) is defined as ratio of cumulative steps for which the peak robot torque in a cycle is upper-bounded to the total number of steps (expressed as %) i.e.

$$\%N_{HA}/N = 100 * \{\textstyle\sum_{i=1}^{N} N_i^*\}/N, \tag{1}$$

where N_i^* is the i^{th} gait cycle for which

$$\tau_i^{max,R}(t_{i,SW}) \leq \delta,\ t_{SW} \in \left[t_{i,TO}, t_{i,FS}\right], \tag{2}$$

where $\tau_i^{max,R}$ is the peak robot torque in the i^{th} gait cycle, $\delta > 0$ is a design constant, and $t_{i,TO}$ and $t_{i,FS}$ are the instants of toe-off and foot strike in the i^{th} gait cycle, respectively. In this study, we adopt $\delta = 2.82$ Nm, which is the minimum single value device stiction accounting for the uncertainty in torque due to stiction [10]. The net ankle torque ($\vec{\tau}_{net}$) is the vector sum of the robot ($\vec{\tau}_R$) and human ($\vec{\tau}_H$) torques i.e.

$$\vec{\tau}_{net} = \vec{\tau}_H + \vec{\tau}_R. \tag{3}$$

Since the ankle robot is controlled by a simple impedance controller, the net robot torque may be expressed as:

$$\vec{\tau}_{net} = K\theta + B\dot{\theta}, \tag{4}$$

where K and B are the programmable controller stiffness and programmable controller damping, respectively. Without a force transducer, we can estimate human torque ($\widehat{\tau_H}$) as:

$$\widehat{\tau_H} = K\theta + B\dot{\theta} - \vec{\tau}_R, \tag{5}$$

In terms of peak torque, Eq. (5) reduces to:

$$\widehat{\tau_H}_i^{max,H}(t_{i,SW}) = K\theta_i^{max}(t_{i,SW}) + B\dot{\theta}_i^{max}(t_{i,SW}) - \tau_i^{max,R}(t_{i,SW}), \tag{6}$$

subject to the constraint $\tau_i^{max,R}(t_{i,SW}) \leq \alpha$, $t_{SW} \in [t_{i,TO}, t_{i,FS}]$, where θ is estimated online using a linearized model of the ankle and linear encoder outputs [10] and $\dot{\theta}$ is computed online using a Butterworth filter. Note that for a given assisted trial on any visit, K and B are held constant. Per Eq. (6), a lower value of $\tau_i^{max,R}(t_{i,SW})$ results in a higher value of $\widehat{\tau_H}_i^{max,H}(t_{i,SW})$, and vice versa, as expected. Compensating for adaptations due to device mass and walking with an actuated exoskeleton, we compute $\%N_{HA}/N$ across all sessions, starting with the second session and ending with the final session (e.g. 18th in our current study).

V. Results

The University of Maryland, Baltimore Institutional Review Board and Veterans Affairs Research & Development approved this study (HP-00046304); written informed consent was obtained. Eligibility included adults with mild-moderate severity chronic (>6 months) hemiparetic gait, paretic ankle dorsi-flexor manual muscle test score ≥ 2 (full ROM gravity eliminated) and ≤ 4 (full ROM against gravity, moderate resistance) in dorsiflexion and/or plantarflexion, and capacity to treadmill walk ≥ 0.12 m/sec for 3 min with handrail support. Exclusion criteria included conditions precluding exercise, concurrent physical therapy, and non-stroke mobility disability conditions. Clinical evaluations included screening for dementia, depression, medical and neurological exams, and treadmill exercise stress test [9]. Performance assessments included preferred speed overground walks over an 8-meter instrumented walkway (GaitRite, CIR Systems, Clifton, NJ) and over force plates (Bertec, Columbus, OH), clinical goniometry and robot-derived ankle kinematics and kinetics during unassisted and assisted preferred speed TM walking, respectively [13]. A

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total of N=14 subjects with hemiparetic gait deficits (foot drop and/or impaired push-off) who met eligibility criteria participated and completed this study.

A. Anklebot Treadmill Training Protocol

Ankle robot integrated treadmill training (TMR) was initiated by matching task difficulty to baseline ankle deficits, and progressed on performance over 18 sessions (3x weekly; 6 weeks) [13]. Each 1-hour session of TMR aimed for two 15–20-min trials, or as tolerated with rests, at preferred speed (increased from 0.34 to 0.45 m/s and duration from 16 to 37 min), to accumulate a mean number of 889 paretic steps/session, with robotic assistance provided to actuate swing dorsi-flexion or stance plantar-flexion, according to individual gait deficits (i.e. deficit-adjusted).

On each visit, subjects walk for 1 minute on the TM with the robot in a "record only" mode to ensure accurate capture of gait sub-events and assure precise timing of robotic actuation during subsequent assisted trials. The unassisted trials are also used to accommodate appreciable changes in gait (e.g. cycle time) to refine robotic trajectory parameters on a session-by-session basis. This is followed by two trials of assisted walking during which the robot continually records ankle kinematics. Level of robotic assistance in early sessions was adjusted to promote foot clearance and push-off, with a tapering of support in the latter sessions to promote autonomy. Robotic assistance was precisely timed to the gait sub-events of interest using insole micro-switches [23]. Primary outcomes obtained at entry, after 6-weeks training, and 6 weeks post-completion included preferred overground walking speed, paretic limb single support durations, and paretic anterior-posterior propulsive impulses. Secondary measures included paretic foot center of pressure (CoP) length and CoP symmetry (paretic-to-nonparetic) during stance [13]. Locomotor learning profile in the TMR group was measured by paretic peak swing angle and heel-first strikes (% footfalls) obtained from robot- and footswitch-measured data during unassisted 1-min treadmill walking respectively, before each session [13]. We recorded self-reported changes in utilization of ankle brace and/or assistive device. These results are reported in [13] and briefly summarized below—here, we report on the temporal profiles and baseline vs. post-completion changes in HAI (see Section IV). Changes in HAI are compared using t-tests across the baseline (second session) and completion (last session) time points, correlations are performed between subject-wise changes in HAI versus unassisted ankle kinematics (e.g. peak swing angle) and kinetic (e.g. peak mechanical DF power) outcomes, and linear regressions computed to derive the optimal intervention duration model for responders (as defined in Section IV). Two-tail significance was set at 0.05.

B. Clinical Findings

Clinical findings utilizing this approach are described elsewhere [13]. Briefly, in chronic stroke (N=14), we found that 6 weeks TMR (18 sessions) increases self-selected walking speed and that participants continued to progress for 6 weeks after cessation of the formal Anklebot training stimulus. Additionally, TMR produced significant increases in paretic leg impulse at post-training, functionally shifting from braking to propulsion and further improving this metric toward normal values across the retention period (~80% restoration toward normal, Fig. 3). The latter finding suggests that robotically assisting the paretic ankle into a more advantageous rocker position through the early stance phase biomechanically facilitates complete gait cycle engagement, thereby contributing to ongoing improvement in gait velocity. These gains at the level of the whole body (walking velocity) and at the level of the limb (paretic leg kinetics) are also reflected at the level of the paretic ankle. Specifically, by 6-weeks TMR significantly increases volitional paretic peak swing angle (P-PSW), a surrogate for ground clearance with gains sustained at retention (Fig. 3). Analysis of robot-derived kinematics from each unassisted TMR session revealed a power-law motor learning profile for P-PSW in all subjects, demonstrating emergence of autonomous swing clearance. Over a surprisingly brief period of time, subjects significantly improved their aerobic intensity indexed by self-selected training speed (33%) and duration (155%), even though cardiopulmonary fitness was not a training target. Collectively, these benefits in gait and balance function translated into an 85% reduction in AFO and/or AD use among stroke survivors with a mean latency greater than 3 years (post-stroke, 41.9 ± 20 months). This self-reported discarding of AFOs or switching to a less-supportive device during retention, suggests profoundly greater functional independence as well as ambulatory confidence in community environments. Notably, in contrast, current approaches to gait assistance/improvement (i.e. PT, AFO's, FES, TM without Anklebot) do not improve paretic ankle motor control and unassisted gait biomechanics.

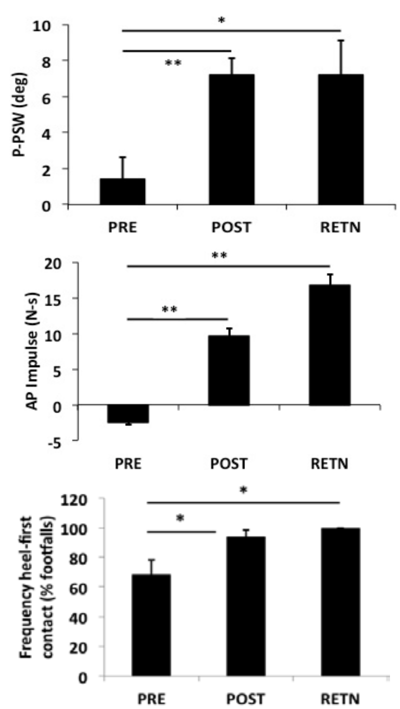

Figure 3. Effect of 6-week TMR on: paretic peak swing angle, anterior-posterior propulsion, and heel-first foot strikes in N=14 stroke patients.

C. TMR Mediated Increase in Human Autonomy Index

The primary finding is that on average HAI changed from 7.2±1.49% at baseline (visit 2) to 12.1±1.40% at post-completion (visit 18) (Fig. 4). Changes across subjects normalized to individual indices was 156±49%. This is a robust change given the relatively short intervention duration (18 one-hour sessions) and the small sample size (N=14), and concomitant with the finding that 85% TMR subjects self-discarded or reduced reliance on their ADs, which they had previously been prescribed for a mean of 41 months.

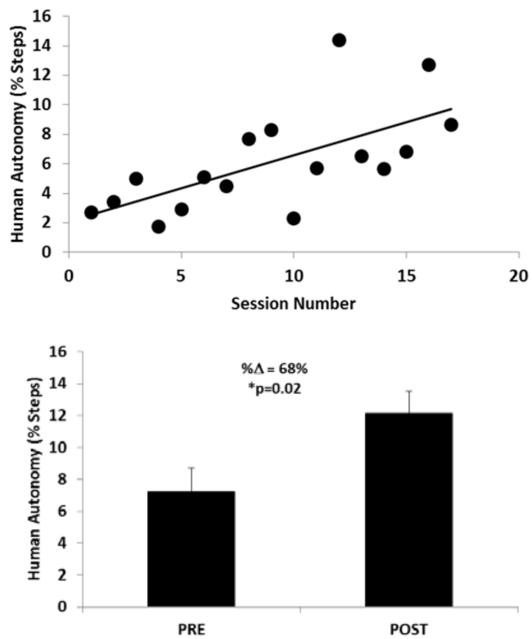

Figure 4. Effect of 6-week TMR on HAI for ankle dorsiflexion toward reversing foot drop in N=14 stroke patients. (Top) Exemplar HAI temporal profile; (Bottom) HAI group average across visit 2 (“PRE”)

D. HAI Predicts Independent Ankle Neuro-Motor Function

Correlations were performed to determine if changes in the HAI predicted changes in select ankle neuromotor and whole-body functional measures (Table I). The purpose was to test the hypothesis that improvements in HAI will strongly and positively correlate with key neuromotor measures of ankle function that are directly implicated in the *directionality* of the task targeted during robot-assisted therapy (DF/PF) e.g. swing clearance in the DF direction and anterior-posterior (AP) impulse in the PF direction. As shown in Table I, this hypothesis was found to be correct—the strongest correlations for improvements in HAI occur with improvements in unassisted swing clearance (indexed by peak swing angle), AP impulse, and volitional mechanical power in dorsiflexion. Additionally, the Dynamic Gait Index (DGI), which subsumes multiple locomotor activities (0-24), was also strongly correlated with gains in HAI. This suggests that changes in HAI for responders (i.e. patients who show ongoing improvements in HAI) can predict improvements in these ankle neuromotor and whole-body functional measures.

TABLE I. CHANGES IN HAI VS. OTHER MEASURES

ΔHAI (vs.)	Correlation Coefficient
ΔPeak Swing Angle (deg)	0.38
ΔAP Impulse (Ns)	0.37
ΔDynamic Gait Index Score	0.38
ΔMechanical Power in DF	0.42

VI. CLINICAL IMPLICATIONS AND FUTURE DIRECTIONS

How best to utilize this novel measure in practice is a pertinent question. We propose two applications: one, a model to predict the optimal duration for robot-assisted locomotor therapy; and two, an algorithm to probe ongoing HAI changes to select a direction-based (DF/PF/both) robotic prescription.

A. Model to Customize Robotic Intervention Duration

Locomotor learning may be quantified using robot measured HAI on each session. By curve-fitting the HAIs with multiple functions (linear, power etc.), we found that LSQ linear regression was the best fit, with the learning rate (α) defined as its slope (Fig. 5). For learning in HAI to have occurred, $\alpha > 0$ with a higher value corresponding to more rapid learning, and vice versa. Hence, in “responders” (i.e. $\alpha > 0$), the relative change in HAI is given by:

$$\frac{\Delta HAI}{HAI_0} = \alpha \frac{\Delta n}{n_0}, \tag{7}$$

where HAI_0 is the initial value at the initial session, n_0, and α is the learning rate. Our data show that 6-week TMR results in HAI learning at an average rate of 0.4 (range 0.31-0.66) (Fig. 5). Hence, to obtain a desired relative change in HAI i.e. $\left(\frac{\Delta HAI}{HAI_0}\right)^*$, we calculate the minimum Δn^* required to

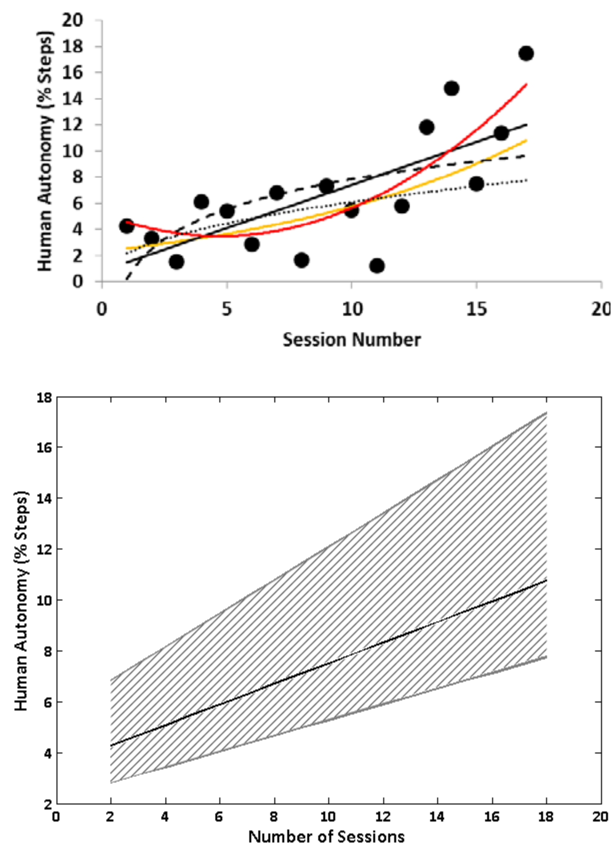

Figure 5. (Top) Curve-fitted HAI data for one stroke patient across sessions (visits 2-18) (-- Log, — Exponential, — Linear, — 2nd Order Polynomial, ... Power); (Bottom) Temporal profile of the linear fitted HAI averaged across N=14 subjects, upper- and lower-bounded by the range of learning rates (slopes) and initial HAIs (intercepts).

bring about this change i.e.

$$\Delta n^* = \frac{n_0}{\alpha}\left(\frac{\Delta HAI}{HAI_0}\right)^*$$

For example, to obtain a 90% gain in HAI i.e. $\left(\frac{\Delta HAI}{HAI_0}\right)^* = 0.9$, on average 5 visits (range: 3-6 visits) are required, which corresponds to ~1.5-2 weeks depending on the frequency of training (3xweekly or 2xweekly, respectively). Note that this model assumes that a patient will be a responder (i.e. $\alpha > 0$). In practice, this may be assessed by two "pre-intervention screening" visits separated by 48 hours to determine potential responders, characterized by "pre-intervention" learning rate $\alpha_0 > 0$). Then, the training should last until visit n^*, where

$$n^* = n_0\left[1 + \frac{1}{\alpha_0}\left(\frac{\Delta HAI}{HAI_0}\right)^*\right].$$

In practice, n^* will consist of uncertainty bounds given by:

$$n_-^* = n_0\left[1 + \frac{1}{\alpha_-^{exp}}\left(\frac{\Delta HAI}{HAI_0}\right)^*\right],$$

$$n_+^* = n_0\left[1 + \frac{1}{\alpha_+^{exp}}\left(\frac{\Delta HAI}{HAI_0}\right)^*\right].$$

where α_-^{exp} and α_+^{exp} are the minimum and maximum HAI learning rates as determined in N=14 stroke patients in this study (Fig. 6). Hence, the model may be expressed as:

$$min\left\{n_-^*, n_0\left[1 + \frac{1}{\alpha_0}\left(\frac{\Delta HAI}{HAI_0}\right)^*\right]\right\} \le n^* \le max\left\{n_+^*, n_0\left[1 + \frac{1}{\alpha_0}\left(\frac{\Delta HAI}{HAI_0}\right)^*\right]\right\}$$

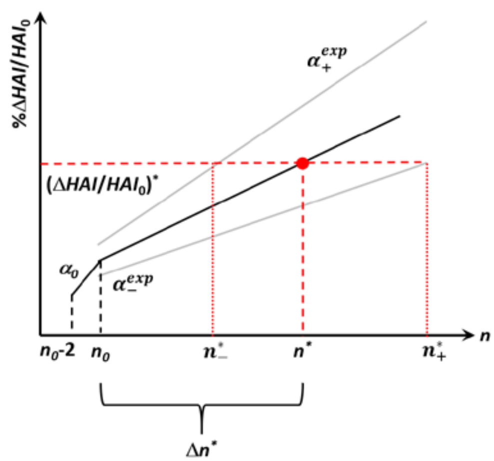

Figure 6. Optimal training duration model based on desired change in HAI. Two "pre-intervention" visits are conducted to determine potential responders ($\alpha_0 > 0$). The robotic intervention should last until $\min\{n^*, n_-^*\} \le n^* \le \max\{n^*, n_+^*\}$ (-- line), where the bounds (... lines) are computed by experimental HAI learning rates obtained in this study.

B. Utilizing the Human Autonomy Index

Future TMR interventions can be directionally bimodal by initially incorporating PF stance and DF swing assist ("prescription" R_{x1}) as-needed using our novel adaptive control system (Period 1: weeks 1-2, 3xweekly). As in the current TMR model, swing assist will focus on properly timed dorsiflexor control to assure ground clearance and orient the foot for heel-first landings, and stance assist will target plantar-flexor contributions during late stance to enhance paretic leg push-off. Initial targeting of both directional ranges will give a more precise indication of the locus of ankle deficits while walking. An advantage of this strategy is the blending of motor learning gains unique to stance and swing phases to culminate in progression towards more normal gait biomechanics over the entire gait cycle. Over the course of training, both the magnitude and type of support need to change (Fig. 7). Thus, intermittent (e.g. weekly) HAI probes in each direction will define subsequent focus in Period 2 (e.g. weeks 3-4) and Period 3 (e.g. weeks 5-6) on either DF (prescription R_{x2}) or PF (prescription R_{x3}), or continuing with both (prescription R_{x1}).

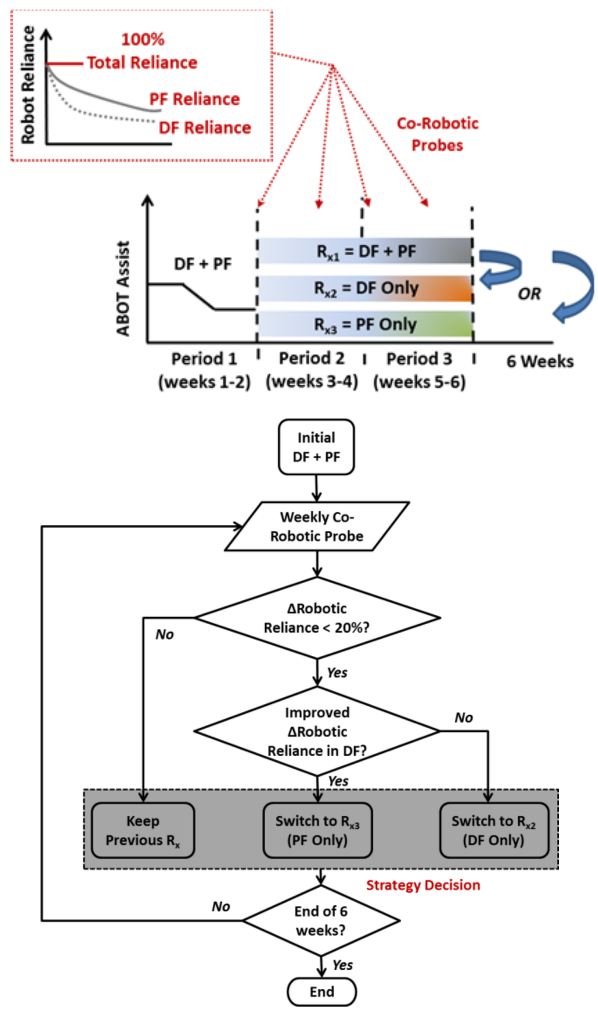

Figure 7. A suggested application of HAI: (Top) Blending initial bimodal DF-PF with subsequent unimodal (DF/PF) training based on ongoing changes in HAI in DF and PF directions; (Bottom) Algorithm to determine the robotic strategy (i.e. *prescription*: directionality of assist) based on intermittent HAI probes.

Our unpublished data reveals a 20% reduction in robotic reliance on DF swing assist over 6 weeks (2xweekly, 12 sessions). Hence we set a 20% change in robotic reliance as a threshold to switch prescriptions, as warranted (Fig. 7). For example, $\Delta HAI_{\mathrm{PF}} \geq 20\%$ reflecting more autonomy during stance, will lead to a switch in the robotic prescription from DF+PF (R_{x1}) to DF-only (R_{x2}). Similarly, $\Delta HAI_{\mathrm{DF}} \geq 20\%$ reflecting more autonomy during swing, will render a switch in the prescription from DF+PF (R_{x1}) to PF-only (R_{x3}). This enables the robot to target deficit(s) differentially or in combination, for each individual.

VII. Conclusion

We have developed a novel Human Autonomy Index as a metric quantifying user torque contribution during robot assisted locomotor training. The HAI estimates human effort and is shown to be correlated with key measures of volitional ankle neuromotor and whole-body functions in persons with chronic stroke, thereby providing a means to predicting patient recovery in those measures. Also, by using the HAI as an intermittent "probe" over the course of an intervention, we propose a new training model in which the directionality of assist targeted (dorsi/plantar-flexion, or both) for each patient is determined by the ongoing HAI temporal profile. As a benchmark of autonomy, the HAI can be used to predict the optimal duration of robotic therapy, rendering it valuable more generally for human-robot interaction control for rehabilitation and potentially other co-robotic applications.

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