J Neurophysiol 118: 363–373, 2017.
First published April 5, 2017; doi:10.1152/jn.00813.2016.

RESEARCH ARTICLE | *Control of Movement*

Increased neuromuscular consistency in gait and balance after partnered, dance-based rehabilitation in Parkinson's disease

Jessica L. Allen,[1] J. Lucas McKay,[1] Andrew Sawers,[2] Madeleine E. Hackney,[3,4] and Lena H. Ting[1,5]

[1]*Wallace H. Coulter Department of Biomedical Engineering, Emory University and Georgia Institute of Technology, Atlanta, Georgia;* [2]*Department of Kinesiology, University of Illinois at Chicago, Chicago, Illinois;* [3]*Atlanta Department of Veterans Affairs Center of Excellence for Visual and Neurocognitive Rehabilitation, Atlanta, Georgia;* [4]*Division of General Medicine and Geriatrics, Department of Medicine, Emory University School of Medicine, Atlanta, Georgia; and* [5]*Division of Physical Therapy, Department of Rehabilitation Medicine, Emory University School of Medicine, Atlanta, Georgia*

Submitted 14 October 2016; accepted in final form 5 April 2017

Allen JL, McKay JL, Sawers A, Hackney ME, Ting LH. Increased neuromuscular consistency in gait and balance after partnered, dance-based rehabilitation in Parkinson's disease. *J Neurophysiol* 118: 363–373, 2017. First published April 5, 2017; doi: 10.1152/jn.00813.2016.—Here we examined changes in muscle coordination associated with improved motor performance after partnered, dance-based rehabilitation in individuals with mild to moderate idiopathic Parkinson's disease. Using motor module (a.k.a. muscle synergy) analysis, we identified changes in the modular control of overground walking and standing reactive balance that accompanied clinically meaningful improvements in behavioral measures of balance, gait, and disease symptoms after 3 wk of daily Adapted Tango classes. In contrast to previous studies that revealed a positive association between motor module number and motor performance, none of the six participants in this pilot study increased motor module number despite improvements in behavioral measures of balance and gait performance. Instead, motor modules were more consistently recruited and distinctly organized immediately after rehabilitation, suggesting more reliable motor output. Furthermore, the pool of motor modules shared between walking and reactive balance increased after rehabilitation, suggesting greater generalizability of motor module function across tasks. Our work is the first to show that motor module distinctness, consistency, and generalizability are more sensitive to improvements in gait and balance function after short-term rehabilitation than motor module number. Moreover, as similar differences in motor module distinctness, consistency, and generalizability have been demonstrated previously in healthy young adults with and without long-term motor training, our work suggests commonalities in the structure of muscle coordination associated with differences in motor performance across the spectrum from motor impairment to expertise.

NEW & NOTEWORTHY We demonstrate changes in neuromuscular control of gait and balance in individuals with Parkinson's disease after short-term, dance-based rehabilitation. Our work is the first to show that motor module distinctness, consistency, and generalizability across gait and balance are more sensitive than motor module number to improvements in motor performance following short-term rehabilitation. Our results indicate commonalities in muscle coordination improvements associated with motor skill reacquisition due to rehabilitation and motor skill acquisition in healthy individuals.

Address for reprint requests and other correspondence: L. H. Ting, 1760 Haygood Dr., Suite W200, Atlanta, GA 30322 (e-mail: lting@emory.edu).

muscle coordination; muscle synergy; electromyography; dance; exercise

FEATURES OF MUSCLE COORDINATION associated with differences in gait and balance performance may provide important insight into neural mechanisms of motor performance, particularly in neurological disorders. Motor module (a.k.a. muscle synergy) analysis has been used to provide such insight and has identified differences in neuromuscular control across levels of motor performance in both healthy and impaired populations (for reviews see Bizzi and Cheung 2013; Ivanenko et al. 2013; Ting et al. 2015). Motor modules are defined as groups of coactive muscles with a fixed spatial structure that are flexibly recruited over time to transform movement goals into biomechanical outputs (Allen and Neptune 2012; Berniker et al. 2009; Chvatal et al. 2011; d'Avella and Bizzi 2005; Ting and Macpherson 2005). In an effort to advance the analysis of muscle coordination, we recently developed more refined motor module-based metrics of neuromuscular control. Specifically, we identified differences in the distinctness and consistency of motor modules as a function of motor skill in healthy, young adults (Sawers et al. 2015). However, it remains unclear whether similar changes accompany improvements in motor performance following rehabilitation. Understanding general principles of neuromuscular control that underlie improvements in motor performance with rehabilitation may help improve patient screening for rehabilitation prescription and guide the development of new interventions to enhance the reacquisition of movement skills lost through injury or disease.

The number of motor modules recruited to perform a motor task is frequently used as a measure of neuromuscular complexity, with higher complexity (i.e., more motor modules) associated with better motor performance. Increased neuromuscular complexity is observed as motor development progresses (Dominici et al. 2011) and because of long-term motor training (i.e., ballet dancers vs. nondancers; Sawers et al. 2015). Conversely, reduced neuromuscular complexity has been identified in various populations that exhibit impaired motor performance such as individuals after stroke (Cheung et

al. 2012; Clark et al. 2010) and those with spinal cord injury (Fox et al. 2013; Hayes et al. 2014; Pérez-Nombela et al. 2017), cerebral palsy (Steele et al. 2015; Tang et al. 2015), and Parkinson's disease (PD) (Rodriguez et al. 2013). A single prior study has demonstrated changes in neuromuscular complexity within the same individuals due to rehabilitation where increased neuromuscular complexity, i.e., more motor modules, was associated with improved motor performance following rehabilitation (e.g., increased walking speed after stroke; Routson et al. 2013).

However, the number of motor modules alone may be insufficiently sensitive to distinguish important and clinically relevant impairments in motor performance and, subsequently, any improvements with rehabilitation. Individuals with neurological motor impairments who recruit the same number of motor modules can exhibit widely varying levels of motor performance [e.g., stroke (Clark et al. 2010), spinal cord injury (Hayes et al. 2014), etc.]. Among stroke survivors, rehabilitation that is successful in improving motor performance does not always result in increased motor module number (Routson et al. 2013). Thus a given number of motor modules does not directly translate to a specific level of motor performance. In the case of PD, movement may be substantially impaired although the number of motor modules observed during gait is comparable to that of neurotypical control subjects (Rodriguez et al. 2013). Furthermore, although treatment with L-DOPA has beneficial effects on gait (Smulders et al. 2016), it does not alter the number of recruited motor modules (Roemmich et al. 2014).

Whereas the number of motor modules identifies consistent features of muscle coordination underlying multiple movement observations, variation in muscle coordination within those same observations may also reflect differences in motor performance. Generating consistent and well-coordinated movements requires recruitment of motor modules that are consistently and distinctly organized around producing required motor output. However, increased variability in muscle recruitment (e.g., Miller et al. 1996; Robichaud et al. 2009), increased coactivation (e.g., Dietz et al. 1995; Lamontagne et al. 2000; Lünenburger et al. 2006), and less distinct motor module organization (e.g., Clark et al. 2010; Fox et al. 2013; Hayes et al. 2014) have previously been identified in individuals with motor impairment. We recently observed greater consistency and distinctness of motor modules for walking and balance among expert professional ballet dancers compared with novice nondancers (Sawers et al. 2015). These differences may reflect greater stability of motor output across repetitions of a task (consistency) that is organized around producing more well-defined biomechanical output (distinctness), leading to superior motor performance. Whether short-term, intensive rehabilitation in motor-impaired populations results in similar improvements in motor module consistency and distinctness remains unknown.

Generalization of motor modules, i.e., the ability to use the same motor modules across different motor behaviors, may also be an important feature of muscle coordination relevant to understanding the effects of rehabilitation. Animal studies suggest that shared motor modules across a range of hindlimb motor tasks may share common neural substrates (Cheung et al. 2005; d'Avella et al. 2003; d'Avella and Bizzi 2005; Hart and Giszter 2004). Similarly in humans, shared motor modules have been identified across a range of lower limb motor tasks, such as across gait and balance tasks (Chvatal and Ting 2012, 2013; Oliveira et al. 2012, 2013a). However, because gait and balance performance can be differently affected by aging and PD (Horak et al. 2016; Park et al. 2016), the same motor modules may no longer be recruited across these two motor tasks. Sharing of motor modules across motor tasks may be critical for practice of tasks during rehabilitation to generalize to other activities often performed in daily life. We previously found that long-term training over many years in professional ballet dancers leads to better motor performance on an untrained beam-walking task, which was associated with recruiting more common motor modules across motor tasks, compared with nondancers (Sawers et al. 2015). Whether increased generalization of motor modules underlies improved motor performance after rehabilitation is unknown.

Here we hypothesized that changes in neuromuscular control similar to those associated with motor skill acquisition also underlie motor skill reacquisition through rehabilitation. To test this hypothesis, we examined changes in neuromuscular control of gait and balance induced by an exercise-based Adapted Tango (AT) dance program. AT has previously been shown to improve clinical measures of both gait and balance performance in individuals with PD (Hackney and Earhart 2010; McKay et al. 2016; McKee and Hackney 2013). While we recently demonstrated in a small cohort of individuals with mild to moderate PD that improvements in clinical tests of gait and balance after AT were accompanied by changes in ankle muscle coactivity during automatic postural responses to anterior/posterior balance perturbations (McKay et al. 2016), we do not know how muscle activity was changed across both gait and balance. Therefore, in the present study we analyzed electromyography (EMG) data from muscles across the leg and trunk during overground walking and multidirectional postural perturbations to examine whether changes in multimuscle coordination (i.e., motor modules) would be associated with observed motor improvements in both gait and balance. We predicted that after AT rehabilitation these individuals would *1*) recruit more consistent and distinct motor modules and *2*) increase the proportion of motor modules shared between walking and reactive balance, suggesting that generalizability of neuromuscular control across motor tasks was improved after AT.

METHODS

Study Population and Data Sources

We performed motor module analysis on EMG data collected as a secondary outcome measure of a small pilot cohort study (McKay et al. 2016). Briefly, participants with a diagnosis of "definite" idiopathic PD (Racette et al. 1999) participated in a short-duration, high-volume AT rehabilitation intervention. Each participant completed fifteen 1.5-h AT lessons taught by an experienced professional ballroom dance instructor over the course of 3 wk. In addition to the primary clinical outcome measures (below), a convenience sample ($n = 9$) of the entire cohort ($n = 22$) was allocated to additional balance and gait testing with EMG before and after the intervention. Of these, complete EMG data suitable for motor module analysis were available for only six participants (Table 1) because of equipment failure at posttest for the remaining three. All participants provided written informed consent before participating according to protocols approved by the institutional review boards at both Emory University and the Georgia

Table 1. *Participant demographics*

	Age, yr	Sex	Height, m	Mass, kg	PD Duration, yr	UPDRS-III	H&Y	CBF	PD Phenotype	Medications
PR1	68	M	1.8	80.6	5	26	2	24	PIGD (0.14/1.00)	C/L, Ent, Rop
PR2	79	M	1.68	68.0	3	40	2	19	PIGD (0.57/1.00)	C/L, Ama
PR3	64	M	1.75	79.3	11	25	2.5	20	PIGD (0.00/0.50)	C/L, Ent
PR7	36	M	1.83	74.7	6	29	2	24	TD (1.71/0.00)	C/L
PR8	81	F	1.65	48.9	14	31	3	22	PIGD (0.00/0/50	C/L, Rop
PR9	56	M	1.85	82.9	3	28	2	22	Indet (0.71/0.50)	C/L

PD, Parkinson's disease; UPDRS-III, Unified Parkinson's Disease Rating Motor Subscale III; H&Y, Hoehn and Yahr scale; PIGD, postural instability/gait disability dominant; TD, tremor dominant; Indet, indeterminate C/L, carbidopa/levodopa; Ent, entacapone; Rop, ropinerole; Ama, amantadine. Physical function reported with composite physical function (Rikli and Jones 1999). PD phenotype presented as the ratio of average scores on UPDRS-III for posture and gait items/tremor items. Participant codes are as in McKay et al. (2016).

Institute of Technology. All participants were prescribed and taking antiparkinsonian medications throughout the study. All assessments occurred at a self-determined, optimal time consistent between pre- and posttests. While we did not explicitly control for medication wear off during the experiment, the amount of wearing off should be consistent within a participant at pretest and posttest since participants were tested at the same time of day corresponding to their self-determined optimal ON state. In addition, we did not observe any deterioration in movement quality during any session. Participants were classified as tremor dominant (TD), postural instability/gait disability dominant (PIGD), or indeterminate based on Unified Parkinson Disease Rating Scale Motor Subscale III (UPDRS-III) scores, following the methodology of Stebbins and colleagues (Table 1; Stebbins et al. 2013). Briefly, average scores for UPDRS-III items related to tremor and UPDRS-III items related to posture and gait were calculated for each participant. The ratio between these averages was used to classify participants as TD ($\geq$1.5), PIGD ($\leq$1), or indeterminate otherwise.

Clinical Outcomes

Clinical outcomes included motor examination of PD symptoms (UPDRS-III; Goetz et al. 2008) and behavioral measures of balance and gait [Berg Balance Scale (BBS; Berg et al. 1995); Fullerton Advanced Balance scale (FAB; Klein et al. 2011), Dynamic Gait Index (DGI; Shumway-Cook and Woollacott 1995), preferred gait speed, fast gait speed, and 6-min walk test (6MWT; Enright 2003)].

Muscle Activity Assessments

During walking assessments, each participant walked overground at self-selected walking speed for ~7.5 m. Participants were instructed to walk as they would normally while maintaining their head level. At least three trials were collected per participant.

During reactive balance assessments, we recorded postural responses to ramp-and-hold translations of the support surface during standing while participants stood on an instrumented platform that translated in 12 equally spaced directions in the horizontal plane (see Fig. 1*B*). Participants were instructed to maintain balance without stepping. Three trials in each of the twelve directions were collected in random order. The perturbation level was adjusted for each participant such that they could perform the set of perturbations without stepping. This level was determined at pretest by delivering three to six initial perturbations to select the highest perturbation level among six predetermined levels at which a participant could reliably maintain balance without stepping. The same perturbation level from the pretest assessment was used in the posttest, even if the participant could withstand a higher perturbation level at posttest. All participants used *level 3* (displacement 7.5 cm, velocity 15 cm/s, acceleration 0.1 g) except *participants PR7* and *PR9*, who used *level 4* (10 cm, 20 cm/s, 0.2 g). Stance width was self-selected by each participant at the beginning of the pretest and enforced through all trials during pre- and posttests. In one participant (*PR1*), self-selected stance width was not correctly enforced, and this participant used a 9.5-cm wider stance width at posttest.

Surface EMG activity was recorded at 1,080 Hz from 13 muscles of the right side leg and lower back. Muscles recorded from included rectus abdominus (REAB), external oblique (EXOB), erector spinae (ERSP), gluteus medius (GMED), tensor fascia lata (TFL), biceps femoris long head (BFLH), rectus femoris (RF), vastus medialis (VMED), medial gastrocnemius (MGAS), lateral gastrocnemius (LGAS), soleus (SOL), peroneus longus (PERO), and tibialis anterior (TA). Three-dimensional kinematics were also measured with an eight-camera Vicon motion analysis system at 120 Hz and a custom 25-marker set that included head-arms-trunk, thigh, shank, and foot segments.

EMG Data Processing

All EMG data were high-pass filtered at 35 Hz, demeaned, rectified, and low-pass filtered at 40 Hz with custom MATLAB routines. To extract motor modules, we first generated subject-specific EMG data matrices for each condition [4 conditions = 2 tasks (walking and reactive balance) $\times$ 2 time points (pretest and posttest)] as follows. To fully capture the underlying variability, the EMG data matrices included the whole data set of EMG rather than averaged data (e.g., over trials for reactive balance or gait cycles for walking). Across both behaviors the EMG data matrices were normalized to the maximum activation observed during walking.

For walking, at least five total gait cycles per walking condition were included in the analyses. EMG data were averaged over 75-ms bins, and data from the first and last two steps as identified from kinematic markers on the heels were removed in order to avoid gait initiation and termination, as in a previous study (Chvatal and Ting 2013). Trials were concatenated end to end to form an $m \times t$ data matrix, where m is the number of muscles (13) and t the number of conditions (trials $\times$ time bins). The number of data points in the walking data matrix varied across subjects, with a minimum size of 115 points.

For reactive balance, EMG data were analyzed during four different time bins: one before the perturbation and three during the automatic postural response (APR; Fig. 1*B*), as in a previous study (Chvatal et al. 2011). Specifically, mean muscle activity was calculated during a 120-ms background period that ended 170 ms before the perturbation and during each of three 75-ms bins beginning either 120 or 150 ms after perturbation onset depending on the level of the applied perturbation. Latencies of 150 and 120 ms were used for the *level 3* (participants *PR1*, *PR2*, *PR3*, *PR8*) and *level 4* (participants *PR7*, *PR9*) perturbations, respectively. These onsets are based on the earliest observed onset of muscle activity across all muscles and perturbation directions previously observed in healthy, young adults during identical levels of applied perturbations. Mean muscle activity values for each muscle during each bin during each trial were assembled to form an $m \times t$ data matrix, where m is the

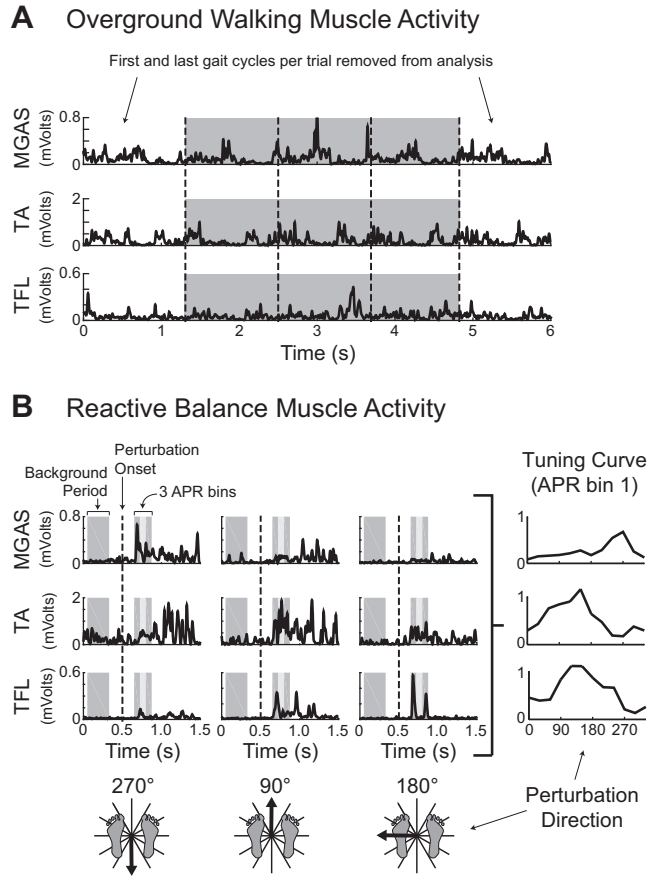

Fig. 1. Example processed EMG from select muscles during overground walking (*A*) and reactive balance (*B*). *A*: muscle activity for walking was recorded while participants walked overground at their self-selected speed for at least 3 trials of 7.5 m each. For each trial, the first and last gait cycles were removed to avoid gait initiation and termination. Dashed lines represent right heel strikes, and the shaded region represents the gait cycles analyzed for 1 trial. Data from all trials for a subject were concatenated before motor module extraction to form an $m \times t$ data matrix, where m is the number of muscles and t the number of time points across all trials. *B*: muscle activity for reactive balance was assessed through ramp-and-hold perturbations in 12 evenly spaced directions. *Left*: responses to backward, forward, and leftward perturbations are illustrated. EMG responses occurred ~120–150 ms after perturbation onset (denoted by vertical dashed lines). Mean EMG activity was calculated during a background period before the perturbation and during three 75-ms time bins during the automatic postural response (APR, shaded regions). *Right:* tuning curves of mean muscle activity from perturbation responses as a function of perturbation directions for the first APR bin. Before motor module extraction, the tuning curves were assembled to form an $m \times t$ data matrix, where m is the number of muscles and t the number of data points (3 trials × 12 directions × 4 time bins = 144).

number of muscles (13) and t the number of data points (3 trials × 12 directions × 4 time bins = 144).

Motor Module Extraction

Motor modules for each subject at each observation time point (pretest, posttest) were extracted separately from the EMG data matrices derived from walking and from reactive balance with non-negative matrix factorization (Lee and Seung 1999) such that EMG = W × C, where W is an $m \times n$ matrix with n modules and C is the $n \times t$ matrix of motor module activation coefficients. Each column of W represents the weights of each muscle in a module, and each row of C represents how much the corresponding module was activated over all data points. To ensure equal weighting of each muscle during the extraction process, each row in the EMG data matrices (i.e., muscle vector) was scaled to unit variance before motor module extraction and rescaled to original units afterwards (Torres-Oviedo and Ting 2007).

The number of motor modules, n, per condition was chosen as follows. From each EMG data matrix 1–13 motor modules (W) were extracted and the goodness of fit between actual and reconstructed EMG was evaluated with variability accounted for (VAF), defined as 100 × squared uncentered Pearson's correlation coefficient (Zar 1999). The number of motor modules was chosen such that the lower bound of the 95% confidence interval (CI) on VAF exceeded 90% (Cheung et al. 2009; Hayes et al. 2014). The 95% CI was found by implementing a bootstrapping procedure in which the EMG data matrix was resampled 500 times with replacement. The VAF of the reconstructed EMG was recalculated for each resampling, and 95% CIs were constructed from these bootstrapped VAF values at each module number (Fig. 2).

Data Analysis

Nine metrics were used to examine motor module changes with rehabilitation.

Motor module number (n_{walk}, $n_{balance}$). Motor module number was defined as the number of motor modules independently extracted for each task.

Motor module coactivity ($W_{mus,walk}$, $W_{mus,balance}$). Motor module coactivity was defined as the number of significantly active muscles per module, which reflects the sparsity of motor module composition. Greater motor module sparsity has been hypothesized to reflect more

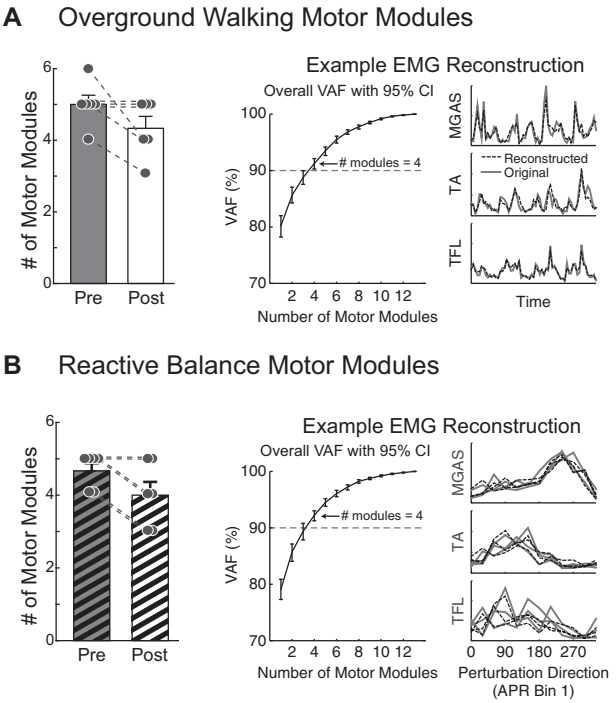

Fig. 2. Number of motor modules and goodness of fit in overground walking (*A*) and reactive balance (*B*). *Left*: the number of motor modules (mean ± SD) recruited during overground walking and reactive balance either decreased or remained the same after Adapted Tango (AT) rehabilitation. Connected circles denote the numbers of motor modules for each subject before and after AT rehabilitation. *Center*: the number of motor modules selected accounted for ≥90% of the overall variability accounted for (VAF) as depicted by plots from an example subject. *Right*: EMG signals were well reconstructed with the extracted motor modules in both walking and reactive balance as depicted in the example original vs. reconstructed EMG plots from a representative subject (light solid lines, original EMG; dark dashed lines, reconstructed EMG).

efficient neuromuscular control (Hayes et al. 2014; Sawers et al. 2015). Significantly active muscles were computed by establishing 95% CIs for the contribution, i.e., the values of the elements W_{ij} to each muscle i in each module j extracted from the previously bootstrapped version of the EMG data sets. Significantly active muscles were considered those whose 95% CI did not include 0.

Motor module generalizability (%shared). Motor module generalizability was defined as the percentage of motor modules recruited across both walking and reactive balance. First, the number of similar motor modules across walking and reactive balance ($n_{similar}$) was identified with Pearson's correlation coefficients (r), as in a previous study (Chvatal and Ting 2013). A pair of motor modules were considered "similar" if $r > 0.684$, which corresponds to the critical value of r^2 for 13 muscles at $P = 0.01$. The amount of motor module similarity was expressed as a percentage to account for the fact that each participant recruited a different number of motor modules. The percentage of similar motor modules was calculated as $100 \times [n_{similar}/(n_{walk} + n_{balance} - n_{similar})]$.

Motor module variability ($R95_{walk}$, $R95_{balance}$). Motor module variability was defined as the variability of motor module structure across different movement observations. This analysis quantifies the variability of motor module spatial structure (W) across different subsets of the EMG data set with a multistep process (Sawers et al. 2015). First, each EMG matrix was resampled 100 times in which 80% of the data was randomly sampled without replacement. From each resampled matrix a new set of motor modules was extracted, where the number of motor modules, n, was identical to the number previously identified from the entire data set. Then, Sammon mapping was used to map and plot each subject's set of resampled motor modules in a two-dimensional (2D) space (De Marchis et al. 2013). This procedure generated a new set of 2D vectors from the set of 13-dimensional vectors (i.e., 13 muscles) while conserving the structure (point-to-point Euclidean distance) of the original data set by minimizing differences in the distance between points from the two data sets (Sammon 1969). To allow comparison of the 2D maps across all conditions, Sammon mapping was applied to a matrix that contained all of the resampled motor modules (i.e., all motor modules from both walking and reactive balance across all participants at both pre- and posttest). Each data point in the resulting map is a 2D representation of one of the resampled motor modules. Finally, the resulting 2D motor module vectors for each participant and task were organized into clusters with k-means clustering, where the number of clusters was set equal to the number of motor modules, n, previously identified for that task. The variability of each motor module was quantified as the radius of a circle that encompassed all of the cluster points in that module to 95% confidence (R95; see Fig. 4) and was then averaged across all modules within a task.

Motor module distinctness (d_{walk}, $d_{balance}$). Motor module distinctness was defined as the mean distance between the R95 circles of each module (d; see Fig. 4), where the more distinct the motor modules are for a task the greater the distance.

Statistical Analyses

For preliminary analysis, changes in the number of motor modules (n_{walk}, $n_{balance}$) from pre- to posttest were compared to the null value 0 with signed-rank tests. Because of the small sample size, we considered further analyses of individual motor module outcomes unlikely to be informative. Therefore to examine changes in motor module metrics with rehabilitation, we tested whether a composite outcome measure of all motor module outcomes described above would exhibit consistent changes across all participants from pre- to posttest. We defined a "direction of expected change" for each outcome measure separately based on observed and hypothesized changes (see Table 3; see RESULTS for description). We modeled the number of the nine separate motor module outcomes that changed in the expected direction from pre- to posttest for each participant as a binomial random variable with nine independent Bernoulli trials with probability of success 0.5 [$X \sim B(n = 9, p_0 = 0.50)$]. That is, we compared the observed proportion of outcome measures that changed in the expected direction $\hat{p}$ to that which would be expected under the null hypothesis that each participant tossed nine independent, but fair coins. We compared the averaged observed proportion, $\hat{p}$, to the null value of $p_0 = 0.5$ with a Wald test.

Secondary analyses were applied to each outcome to calculate the effect size of the change induced with AT rehabilitation. Effect sizes were calculated with Cohen's d, calculated as differences in means between posttest and pretest divided by standard deviation at pretest.

RESULTS

Performance on clinical outcomes in the present study is summarized in Table 2. With effect size cutoff points suggested by Cohen (1992), at posttest medium improvements were observed in PD symptoms (UPDRS-III, d = 0.55), medium to large improvements were observed in clinical balance measures (BBS, d = 1.17; FAB, d = 0.83; DGI, d = 0.87), small to medium effects were observed on overground gait [Timed Up and Go test (TUG), d = 0.46; 6MWT, d = 0.79), and negligible effects were observed on gait speed (preferred,

Table 2. *Clinical measures of balance and gait before and after 3-wk high-volume Adapted Tango rehabilitation intervention*

	Participants					
Clinical Outcome	*PR1*	*PR2*	*PR3*	*PR7*	*PR8*	*PR9*
UPDRS-III						
Pretest	26	40	25	29	31	28
Posttest	26	33	26	19	30	27
Change	0	−7	+1	−10	−1	−1
BBS						
Pretest	51	52	54	56	51	54
Posttest	55	53	56	56	56	56
Change	+4	+1	+2	0	+5	+2
FAB						
Pretest	29	23	30	36	26	34
Posttest	37	28	34	35	30	38
Change	+8	+5	_4	−1	+4	+4
DGI						
Pretest	18	19	20	23	17	24
Posttest		22	21	24	23	23
Change		+3	+1	+1	+6	−1
TUG						
Pretest	10.16	8.03	6.96	5.65	7.72	7.03
Posttest	7.87	8.66	6.13	5.65	7.34	6.5
Change	−2.29	+0.63	−0.84	0	−0.38	−0.53
6MWT, m						
Pretest	371.9	350.5	478.5	403.2	477.0	451.1
Posttest	433.4	387.1	452.6	442.0	528.0	548.6
Change	+61.5	+36.6	−25.9	38.8	51	97.5
Gait speed, m/s (preferred)						
Pretest	1.11	0.95	1.32	1.19	1.66	1.36
Posttest	1.07	0.85	1.41	1.36	1.36	1.66
Change	−0.04	−0.10	+0.09	+0.18	−0.29	+0.30
Gait speed, m/s (fast)						
Pretest	1.59	1.60	1.95	1.87	1.98	2.07
Posttest	1.57	1.31	2.19	1.95	1.93	2.24
Change	−0.02	−0.29	+0.24	+0.08	−0.04	+0.17

UPDRS-III, Unified Parkinson's Disease Rating Scale, Part III: Motor Exam; BBS, Berg Balance Scale; FAB, Fullerton Advanced Balance Scale; DGI, Dynamic Gait Index; TUG, Timed Up and Go test; 6MWT, 6-min walk test. Participant codes are as in McKay et al. (2016).

d = 0.08; fast, d = 0.11)]. Where effects were observed, they were consistently larger than effect sizes reported previously for the entire cohort from which these participants were sampled (cf. McKay et al. 2016; UPDRS-III, d = 0.47; BBS, d = 0.59; FAB, d = 0.56; DGI, d = 0.53; TUG, d = 0.31; 6MWT, d = 0.37).

In contrast with previous studies that have demonstrated that improvements in motor performance are associated with an increase in the number of recruited motor modules, no one in our study cohort increased the number of motor modules recruited in either walking or reactive balance (Fig. 2). Median (±interquartile range) changes in motor module number were -0.5 ± 1 and -1 ± 1 for n_{walk} and $n_{balance}$ with effect sizes of -0.82 and -1.29, respectively. Results of signed-rank tests indicated that neither of these changes could be discriminated from the null value of 0 ($S = -3$, $P = 0.25$ and $S = -5$, $P = 0.125$, for n_{walk} and $n_{balance}$, respectively). To evaluate whether this observation is robust across different criteria to determine motor module number, we performed a post hoc analysis in which we calculated the change in motor module number with four additional criteria: *1*) overall VAF > 85%, *2*) overall VAF > 90%, *3*) overall VAF > 95%, and *4*) lower bound of the 95% CI on VAF > 85%. Across all criteria, we observed no increase in the number of motor modules after rehabilitation in both walking and reactive balance in any participant.

Similarly, in contrast with our previous study that demonstrated that motor module coactivity (W_{mus}) is lower in individuals with superior balance performance (Sawers et al. 2015), at least half of the participants studied here increased motor module coactivity at posttest (Fig. 3*C*). Three and four of six participants increased module coactivity for walking and reactive balance, respectively. Across all participants, motor module coactivity changed from 6.18 ± 1.03 to 7.61 ± 1.77 (effect size = 1.39) for walking and from 6.23 ± 0.96 to 8.32 ± 2.03 (effect size = 2.17) for reactive balance. Post hoc correlation analyses revealed a significant relationship between a decrease in motor module number and an increase in motor module coactivity across both walking and reactive balance ($r = -0.8523$, $P < 0.01$; see Fig. 5).

Consistent with our prediction that motor modules would become more consistent and distinct after AT, most participants decreased motor module variability and increased motor module distinctness in both walking and reactive balance (Fig. 4). Five and three participants decreased motor module variability at posttest in walking and reactive balance, respectively.

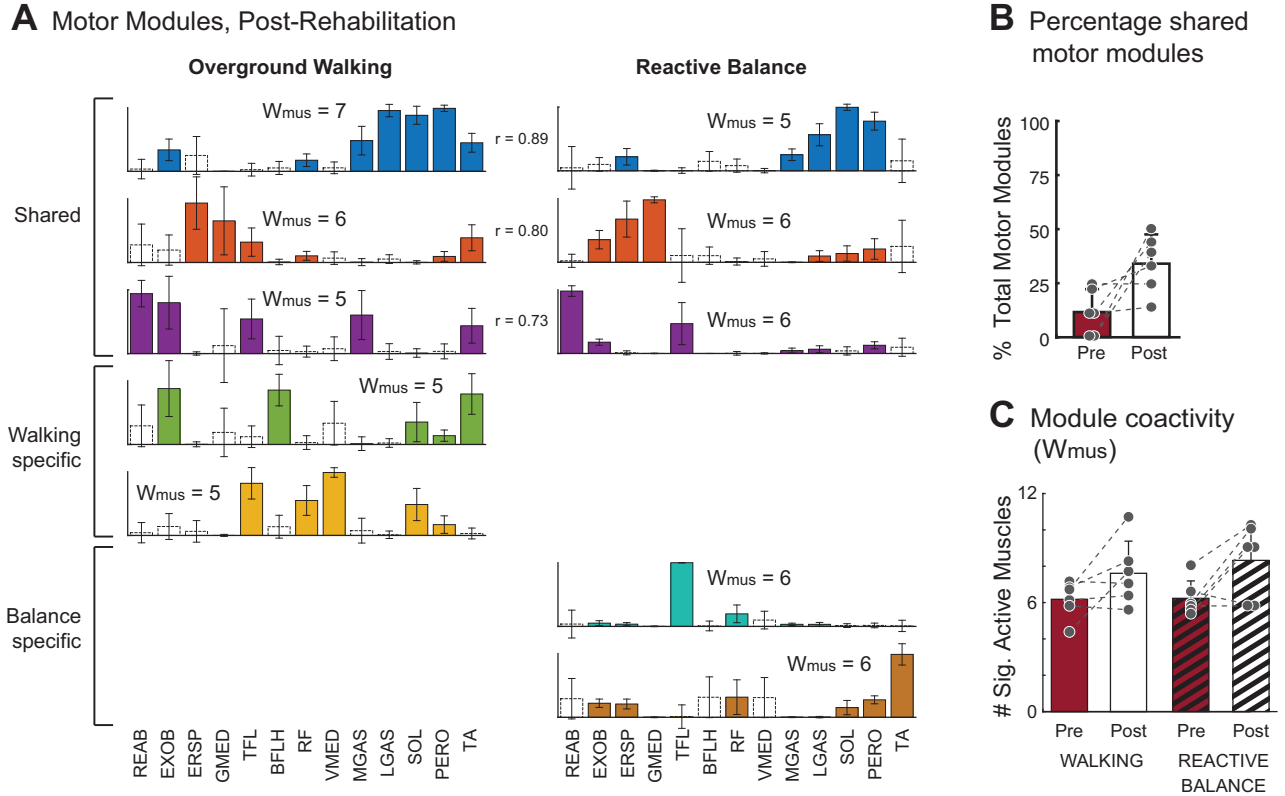

Fig. 3. Motor module sharing and coactivity in overground walking and reactive balance. *A*: representative motor modules during walking (*left*) and reactive balance (*right*). Motor modules were extracted from each behavior independently. Motor modules that were identified as similar between tasks are represented with the same color across tasks. *B*: % of motor modules shared between walking and reactive balance increased from before (Pre) to after (Post) AT rehabilitation in 5 of the 6 participants. Connected circles denote the value for each participant. Shared motor modules are those pairs of motor modules across behaviors in which $r \geq 0.684$. Amount of sharing was quantified as % of total number of unique motor modules (i.e., 42.9% of the motor modules, or 3 of 7, were shared across behaviors in the representative subject in *A*). *C*: motor module coactivity increase from before (Pre) to after (Post) AT rehabilitation in both walking and reactive balance in most participants. Motor module coactivity was quantified as the average number of significantly active muscles per module (W_{mus}). Significantly active muscles represent those whose activation was consistently >0, despite variations over movement repetitions, and muscles were classified as significantly active if their 95% CI did not include 0, whereas nonsignificantly active muscle had 95% CIs that included 0 (i.e., filled bars with solid borders vs. open bars with dashed borders in the representative motor modules in *A*).

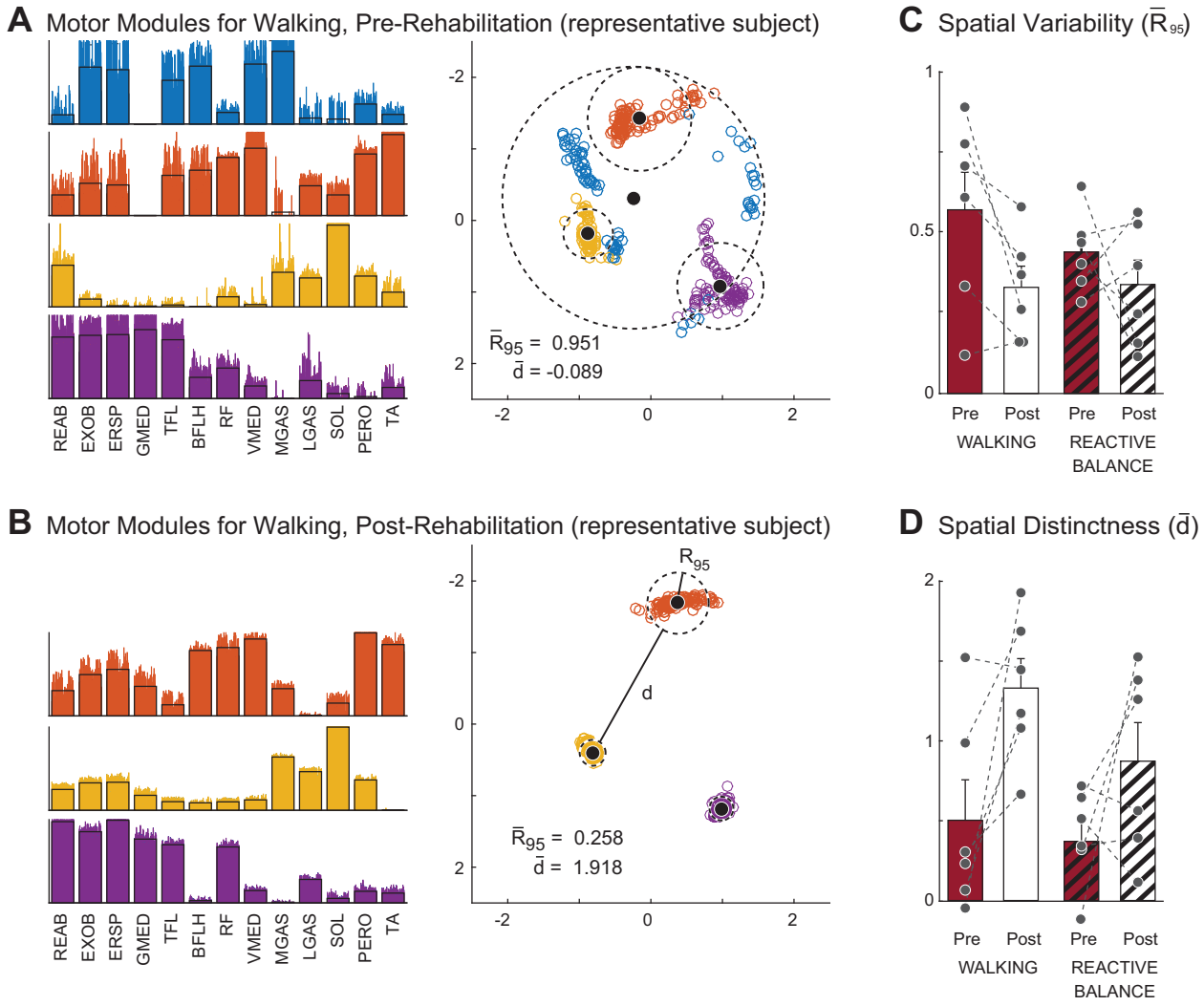

Fig. 4. Spatial motor module variability and distinctness. Example motor modules and cluster plots for walking before rehabilitation (*A*) and after rehabilitation (*B*) depicting motor module consistency (R95) and distinctness (*d*). *Left:* colored bars for each muscle weighting represent the contribution of a muscle within a module over each of the 100 different resampled module extractions. Black bars indicate the mean across all resampled extractions. *Right*: each point in a cluster is a 2-dimensional representation of 1 of the 100 resampled motor modules as depicted on *left*. *C*: motor module variability decreased from before (Pre) to after (Post) AT rehabilitation in most participants for both walking and reactive balance. *D*: motor module distinctness increased in most participants after AT rehabilitation in both walking and reactive balance. Connected circles denote the value for each participant.

Five and four increased motor module distinctness in walking and reactive balance, respectively. Across all participants, motor module variability decreased from 0.57 ± 0.29 to 0.33 ± 0.16 (effect size $= -0.84$) for walking and from 0.44 ± 0.13 to 0.34 ± 0.18 (effect size $= -0.37$) for reactive balance. Motor module distinctness increased from 0.50 ± 0.62 to 1.33 ± 0.45 (effect size $= 1.17$) for walking and from 0.35 ± 0.36 to 0.83 ± 0.56 (effect size $= 0.59$) for reactive balance.

Consistent with our prediction that motor module generalization across walking and balance would increase after AT, five of six participants increased the percentage of motor modules shared between walking and reactive balance at posttest, with the remaining participant having no change (Fig. 3*B*; $11.6 \pm 10.6\%$ to $34.0 \pm 13.5\%$; effect size $= 2.11$). To examine whether this increased generalization was due to motor modules for walking becoming more like those for reactive balance, or vice versa, post hoc analysis was performed with Pearson's correlation coefficients to examine how many of the motor modules at pretest were similar to those recruited at posttest for each motor task. This analysis revealed a greater change in the motor modules recruited for walking than those recruited for reactive balance, with only $25.5 \pm 25.0\%$ of the motor modules recruited for walking in the pretest also recruited in the posttest, compared with $46.7 \pm 21.0\%$ for reactive balance.

Overall, we found that the proportion of participants who exhibited changes in our motor module metrics in the expected direction at posttest were higher than what would be expected by chance. The directions of expected change for each motor module metric for the overall statistical test (Table 3) were chosen as follows. For motor module number, direction of expected change was defined as lack of an increase in motor module number (i.e., reduction or no change in number), which was chosen because of the observation that all participants improved motor performance after rehabilitation without an

Table 3. *Frequency of outcome measures that did vs. did not change in expected direction*

Outcome Measure	Direction of Expected Change	No. of Participants	
		Expected change	Nonexpected change
n_{walk}	− or =	6	0
$n_{balance}$	− or =	6	0
$\%_{shared}$	+	5	1
d_{walk}	+	5	1
$R95_{walk}$	−	5	1
$W_{mus,walk}$	+	4	2
$d_{balance}$	+	4	2
$R95_{balance}$	−	3	3
$W_{mus,balance}$	+	4	2

n, Motor module number (i.e., complexity); $\%_{shared}$, proportion of motor module shared across walking and reactive balance (i.e., generalizability); d, motor module distinctness, R95, motor module variability; W_{mus}, motor module coactivity.

increase in motor module number. Similarly, because decrease in motor module number was associated with an increase in W_{mus}, we defined the direction of expected change for W_{mus} as an increase in value. Finally, the directions of expected change for motor module variability (decrease), distinctness (increase), and generalizability (increase) were defined based on our hypothesized changes. With these definitions, the average proportion of outcomes that changed in the expected direction from pretest to posttest across all participants (our composite outcome measure) was 0.78 ± 0.32 (Table 4), which is significantly higher than the proportion 0.50 that would be expected by chance ($Z_W = 2.00$, $P = 0.02$). As an alternative approach, we also compared the average number of outcomes that changed in the expected direction for each participant (Table 4) to the value that would be expected under the null hypothesis (4.5) with a t-test that yielded test statistic $t = 2.11$, $P = 0.09$.

DISCUSSION

Here we show that efficacious gait and balance rehabilitation in individuals with PD is associated with changes in neuromuscular control during walking and reactive balance responses. Our work is the first to show that motor module distinctness and consistency may act as markers of improved motor performance after rehabilitation. Furthermore, we demonstrate that increased generalization of motor modules across gait and balance tasks that are controlled by different neural substrates may also indicate improved motor function after rehabilitation. As prior work demonstrates only a modest reduction in motor module number in PD compared with age-matched control subjects, the metrics of motor module consistency, distinctness, and generalizability may be more sensitive to changes in neuromuscular control underlying motor improvements with rehabilitation. Moreover, as similar differences in the distinctness, consistency, and generalization of motor modules have been demonstrated between young adults with and without long-term specialty motor training, there may be commonalities in the structure of muscle coordination associated with differences in motor performance across the spectrum ranging from impairment to expertise.

Our work demonstrates that the number of motor modules recruited for a motor task may not always be the most appropriate metric to identify changes in neuromuscular control that contribute to improvements in motor performance with rehabilitation, particularly in individuals with PD. While the number of recruited motor modules is often associated with motor performance (Cheung et al. 2012; Clark et al. 2010; Fox et al. 2013; Hayes et al. 2014; Pérez-Nombela et al. 2017; Tang et al. 2015), a prior study demonstrated that many individuals with PD have reduced motor performance without exhibiting differences in motor module number (Rodriguez et al. 2013). Moreover, in PD motor module number is not affected by dopaminergic medications that improve motor function (Roemmich et al. 2014), suggesting that aspects of neuromuscular control not captured by motor module number can be affected by PD. Consistent with these prior findings, no increases in motor module number were observed in any of the participants studied here despite clinically meaningful improvements on behavioral measures of balance, gait, and disease symptoms. Interestingly, some participants in our study actually decreased motor module number.

Our novel motor module analyses reveal how consistently and distinctly the structure of each motor module, and therefore its corresponding motor output, is maintained over repeated movements. In contrast to standard motor module analysis based on analysis of the entire data set, we performed multiple analyses on subsets of the data for each participant to identify variations in the structure of motor modules (Sawers et al. 2015). Each analysis identifies slightly different muscle contributions to each motor module. Consistency reflects within-module difference in motor module structure, which we showed decreased after rehabilitation. Our consistency analysis revealed that some motor modules at pretest were highly inconsistent and may not have represented stable neural solutions (Fig. 4*A*); in some cases these were eliminated after rehabilitation (Fig. 4*B*). Distinctness reflects between-module differences in motor module structure, which we showed increased after rehabilitation. Recruiting motor modules that are more distinct in structure may result in motor modules that are organized around producing more well-defined biomechanical output, leading to better motor performance.

As a proxy for the efficiency of movement, our measure of motor module coactivation quantifies the sparsity of muscle representation within a module; the more significantly active muscles within a module, the less sparse that module. Surprisingly, we found that most participants increased motor module coactivity after short-term rehabilitation, whereas healthy individuals who receive long-term motor training (>10 yr) exhibit less muscle coactivation within their motor modules

Table 4. *Frequency of outcome measures that did vs. did not change in expected direction for each participant*

Participant	No. of Outcome Measures	
	Expected change	Nonexpected change
PR1	8	1
PR2	9	0
PR3	9	0
PR7	2	7
PR8	9	0
PR9	5	4

Participant codes are as in McKay et al. (2016).

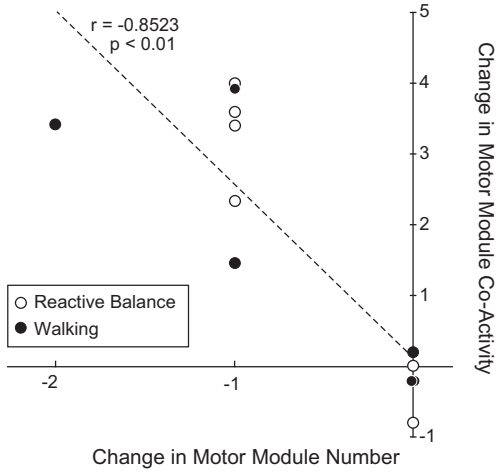

Fig. 5. Increased motor module coactivity was associated with a reduction in motor module number after AT rehabilitation. Motor module coactivity increased in those participants who decreased motor module number, whereas those participants who had no change in motor module number had only minor changes in motor module coactivity. Values for each participant for walking are represented by circles.

(Sawers et al. 2015). Specifically, it was those individuals who decreased motor module number who exhibited increased muscle coactivation within each module (Fig. 5). One possible interpretation is that participants prioritized the ability to reliably generate specific biomechanical output through the consistent recruitment of a module over being more energetically efficient in their movements. It may be that once participants establish appropriate motor modules, continued rehabilitation would reduce the amount of muscle coactivation within each module, similar to what is seen after long-term training. Note that a prior analysis on the same cohort showed a decrease in the coactivation between two antagonistic ankle muscles (McKay et al. 2016); here the increased coactivation within motor modules represents differences in the structure of multimuscle coordination across multiple joints. Increased motor module coactivation was primarily a result of a return toward more appropriate simultaneous activity of anatomically similar muscles (e.g., ankle plantarflexors) and/or coactivation of muscles crossing different joints.

Finally, we found motor module generalizability across tasks to be lower in individuals with PD than reported previously in healthy, young adults and to increase in association with improved motor performance after AT rehabilitation. Prior studies of motor-impaired populations have quantified motor modules within a single motor task (e.g., locomotor tasks in Clark et al. 2010; Rodriguez et al. 2013; Steele et al. 2015). However, studies in unimpaired humans and in animals show that motor modules are typically shared across multiple behaviors because of common neural substrates (e.g., Cheung et al. 2005; Chvatal and Ting 2012, 2013; d'Avella et al. 2003; d'Avella and Bizzi 2005; Hart and Giszter 2004; Oliveira et al. 2012, 2013a). For example, we previously demonstrated in healthy young adults that a common set of motor modules are used across walking and reactive balance (Chvatal and Ting 2012, 2013), which are mediated by different spinal and brain stem circuits. In contrast, the individuals with PD tested here initially exhibited little sharing of motor modules across walking and reactive balance.

Taken together with results from prior studies, the changes in module distinctness, consistency, and generalization observed after AT rehabilitation in parkinsonian patients are consistent with improved basal ganglia function. Prior studies have demonstrated that exercise can improve the trial-by-trial variability of fractionated EMG burst patterns observed during reaching tasks in individuals with moderate PD (David et al. 2016; Robichaud et al. 2009). Similar changes in EMG are observed with antiparkinsonian medications or stimulation of the subthalamic nucleus (Vaillancourt et al. 2004). Additionally, reduced gait variability has been reported after pallidotomy in PD patients (Siegel and Metman 2000). Thus increased motor module consistency and distinctness could reflect changes within dopaminergic systems in the basal ganglia or their targets, perhaps by increasing the efficiency of striatal dopamine transmission through use-dependent plasticity (Petzinger et al. 2007). Furthermore, the loss of automatic movements in favor of conscious control is a hallmark of PD (Kelly et al. 2012; Petzinger et al. 2013), and reduced cortical contributions to gait have been demonstrated in animal models of PD after exercise-based training (Petzinger et al. 2010). Successful partner dance involves concurrent performance of attention, navigation, memory, and gait tasks (McKee and Hackney 2013). We speculate that increased generalization of motor modules across walking and reactive balance could indicate improved automatic control of gait—including dynamic balance during gait—after AT. Our prior work demonstrates that reactive balance modules during standing are also used in balance responses during walking (Chvatal and Ting 2012, 2013). Increased gait automaticity is further supported by our observation that walking motor modules after AT became more similar to the reactive balance motor modules that are likely mediated by brain stem balance centers (Stapley and Drew 2009).

In this pilot study we provide evidence that the motor module metrics of consistency, distinctness, and generalizability may be related to clinically meaningful improvements in motor performance after rehabilitation that cannot be explained by increases in motor module number. However, there are several limitations that must be addressed to identify the

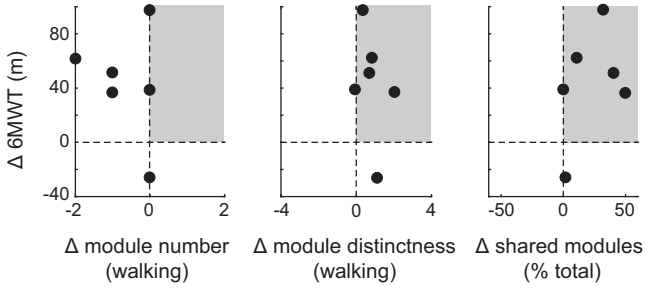

Fig. 6. Examples of associations between clinical scores vs. changes in motor module metrics. The change in walking endurance (6MWT, *y*-axis) is illustrated for each subject vs. the change in motor module number for walking (*left*), motor module distinctness for walking (*center*), and % of motor modules shared between walking and reactive balance (*right*). In general, changes in motor module number did not appear related to improvements in motor performance (e.g., 6MWT). In contrast, other motor module metrics (e.g., distinctness and % shared across tasks) demonstrated trends such that increases in these metrics were in general accompanied by increases in motor performance. Values for each participant are represented by closed circles. Shaded regions denote where there is an increase in both the clinical score and the motor module metric.

relationship between these metrics and motor performance. Because of our small sample size ($n = 6$), we were unable to associate changes in our motor module metrics with overall improvement (or lack thereof) at the level of individual participants or with improvements in specific clinical gait and balance measures, although the trends in these relationships are promising (e.g., Fig. 6). Furthermore, for these metrics to be clinically relevant they must be stable across days (i.e., demonstrate no change) in individuals who do not participate in rehabilitation and have no motor performance improvements. While we did not include a control group in the present study, some support for the stability of our motor module metrics can be seen in the highest-functioning participant (*PR7*), who experienced little change in the clinical domain (as measured with our subset of clinical tests) and was also unchanged in the motor module domain. Nonetheless, future studies incorporating a larger cohort of individuals with appropriate control groups will be necessary to examine the repeatability/robustness of these motor module metrics. In addition, larger cohorts will be necessary to identify the specific relationship of motor module consistency, distinctness, and generalizability to clinical measures of motor performance and whether there are particular improvements that are induced by AT compared with standard of care in PD.

GRANTS

This work was supported in part by National Institutes of Health (NIH) Grants R21 HD-075612 and R01 HD-46922, National Science Foundation EFRI 1137229, Tango Under the Tent, Inc., and the Emory Udall Center. J. L. Allen was supported by NIH Grants T32 NS-007480 and F32 NS-087775. J. L. McKay was supported by the Atlanta Clinical and Translational Science Institute KL2-Mentored Clinical and Translational Research Program (NIH RR-025008, UL1 TR-000454, and KL2 TR-000455). M. E. Hackney was supported by Department of Veterans Affairs R&D Service Career Development Awards E7108M and N0870W.

DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the authors.

AUTHOR CONTRIBUTIONS

J.L.A., J.L.M., A.S., M.E.H., and L.H.T. conceived and designed research; J.L.A. and J.L.M. analyzed data; J.L.A., J.L.M., A.S., and L.H.T. interpreted results of experiments; J.L.A. prepared figures; J.L.A. and L.H.T. drafted manuscript; J.L.A., J.L.M., A.S., M.E.H., and L.H.T. edited and revised manuscript; J.L.A., J.L.M., A.S., M.E.H., and L.H.T. approved final version of manuscript; J.L.M. and M.E.H. performed experiments.

REFERENCES

Allen JL, Neptune RR. Three-dimensional modular control of human walking. *J Biomech* 45: 2157–2163, 2012. doi:10.1016/j.jbiomech.2012.05.037.

Berg K, Wood-Dauphinee S, Williams JI. The Balance Scale: reliability assessment with elderly residents and patients with an acute stroke. *Scand J Rehabil Med* 27: 27–36, 1995.

Berniker M, Jarc A, Bizzi E, Tresch MC. Simplified and effective motor control based on muscle synergies to exploit musculoskeletal dynamics. *Proc Natl Acad Sci USA* 106: 7601–7606, 2009. doi:10.1073/pnas.0901512106.

Bizzi E, Cheung VC. The neural origin of muscle synergies. *Front Comput Neurosci* 7: 51, 2013. doi:10.3389/fncom.2013.00051.

Cheung VC, d'Avella A, Bizzi E. Adjustments of motor pattern for load compensation via modulated activations of muscle synergies during natural behaviors. *J Neurophysiol* 101: 1235–1257, 2009. doi:10.1152/jn.01387.2007.

Cheung VC, d'Avella A, Tresch MC, Bizzi E. Central and sensory contributions to the activation and organization of muscle synergies during natural motor behaviors. *J Neurosci* 25: 6419–6434, 2005. doi:10.1523/JNEUROSCI.4904-04.2005.

Cheung VC, Turolla A, Agostini M, Silvoni S, Bennis C, Kasi P, Paganoni S, Bonato P, Bizzi E. Muscle synergy patterns as physiological markers of motor cortical damage. *Proc Natl Acad Sci USA* 109: 14652–14656, 2012. doi:10.1073/pnas.1212056109.

Chvatal SA, Ting LH. Voluntary and reactive recruitment of locomotor muscle synergies during perturbed walking. *J Neurosci* 32: 12237–12250, 2012. doi:10.1523/JNEUROSCI.6344-11.2012.

Chvatal SA, Ting LH. Common muscle synergies for balance and walking. *Front Comput Neurosci* 7: 48, 2013. doi:10.3389/fncom.2013.00048.

Chvatal SA, Torres-Oviedo G, Safavynia SA, Ting LH. Common muscle synergies for control of center of mass and force in nonstepping and stepping postural behaviors. *J Neurophysiol* 106: 999–1015, 2011. doi:10.1152/jn.00549.2010.

Clark DJ, Ting LH, Zajac FE, Neptune RR, Kautz SA. Merging of healthy motor modules predicts reduced locomotor performance and muscle coordination complexity post-stroke. *J Neurophysiol* 103: 844–857, 2010. doi:10.1152/jn.00825.2009.

Cohen J. A power primer. *Psychol Bull* 112: 155–159, 1992.

d'Avella A, Bizzi E. Shared and specific muscle synergies in natural motor behaviors. *Proc Natl Acad Sci USA* 102: 3076–3081, 2005. doi:10.1073/pnas.0500199102.

d'Avella A, Saltiel P, Bizzi E. Combinations of muscle synergies in the construction of a natural motor behavior. *Nat Neurosci* 6: 300–308, 2003. doi:10.1038/nn1010.

David FJ, Robichaud JA, Vaillancourt DE, Poon C, Kohrt WM, Comella CL, Corcos DM. Progressive resistance exercise restores some properties of the triphasic EMG pattern and improves bradykinesia: the PRET-PD randomized clinical trial. *J Neurophysiol* 116: 2298–2311, 2016. doi:10.1152/jn.01067.2015.

De Marchis C, Schmid M, Bibbo D, Castronovo AM, D'Alessio T, Conforto S. Feedback of mechanical effectiveness induces adaptations in motor modules during cycling. *Front Comput Neurosci* 7: 35, 2013. doi:10.3389/fncom.2013.00035.

Dietz V, Zijlstra W, Prokop T, Berger W. Leg muscle activation during gait in Parkinson's disease: adaptation and interlimb coordination. *Electroencephalogr Clin Neurophysiol* 97: 408–415, 1995. doi:10.1016/0924-980X(95)00109-X.

Dominici N, Ivanenko YP, Cappellini G, d'Avella A, Mondì V, Cicchese M, Fabiano A, Silei T, Di Paolo A, Giannini C, Poppele RE, Lacquaniti F. Locomotor primitives in newborn babies and their development. *Science* 334: 997–999, 2011. doi:10.1126/science.1210617.

Enright PL. The six-minute walk test. *Respir Care* 48: 783–785, 2003.

Fox EJ, Tester NJ, Kautz SA, Howland DR, Clark DJ, Garvan C, Behrman AL. Modular control of varied locomotor tasks in children with incomplete spinal cord injuries. *J Neurophysiol* 110: 1415–1425, 2013. doi:10.1152/jn.00676.2012.

Goetz CG, Tilley BC, Shaftman SR, Stebbins GT, Fahn S, Martinez-Martin P, Poewe W, Sampaio C, Stern MB, Dodel R, Dubois B, Holloway R, Jankovic J, Kulisevsky J, Lang AE, Lees A, Leurgans S, LeWitt PA, Nyenhuis D, Olanow CW, Rascol O, Schrag A, Teresi JA, van Hilten JJ, LaPelle N; Movement Disorder Society UPDRS Revision Task Force. Movement Disorder Society-sponsored revision of the Unified Parkinson's Disease Rating Scale (MDS-UPDRS): scale presentation and clinimetric testing results. *Mov Disord* 23: 2129–2170, 2008. doi:10.1002/mds.22340.

Hackney ME, Earhart GM. Effects of dance on gait and balance in Parkinson's disease: a comparison of partnered and nonpartnered dance movement. *Neurorehabil Neural Repair* 24: 384–392, 2010. doi:10.1177/1545968309353329.

Hart CB, Giszter SF. Modular premotor drives and unit bursts as primitives for frog motor behaviors. *J Neurosci* 24: 5269–5282, 2004. doi:10.1523/JNEUROSCI.5626-03.2004.

Hayes HB, Chvatal SA, French MA, Ting LH, Trumbower RD. Neuromuscular constraints on muscle coordination during overground walking in persons with chronic incomplete spinal cord injury. *Clin Neurophysiol* 125: 2024–2035, 2014. doi:10.1016/j.clinph.2014.02.001.

Horak FB, Mancini M, Carlson-Kuhta P, Nutt JG, Salarian A. Balance and gait represent independent domains of mobility in Parkinson disease. *Phys Ther* 96: 1364–1371, 2016. doi:10.2522/ptj.20150580.

Ivanenko YP, Cappellini G, Solopova IA, Grishin AA, Maclellan MJ, Poppele RE, Lacquaniti F. Plasticity and modular control of locomotor patterns in neurological disorders with motor deficits. *Front Comput Neurosci* 7: 123, 2013. doi:10.3389/fncom.2013.00123.

Kelly VE, Eusterbrock AJ, Shumway-Cook A. A review of dual-task walking deficits in people with Parkinson's disease: motor and cognitive contributions, mechanisms, and clinical implications. *Parkinsons Dis* 2012: 918719, 2012. doi:10.1155/2012/918719.

Klein PJ, Fiedler RC, Rose DJ. Rasch Analysis of the Fullerton Advanced Balance (FAB) Scale. *Physiother Can* 63: 115–125, 2011. doi:10.3138/ptc.2009-51.

Lamontagne A, Richards CL, Malouin F. Coactivation during gait as an adaptive behavior after stroke. *J Electromyogr Kinesiol* 10: 407–415, 2000. doi:10.1016/S1050-6411(00)00028-6.

Lee DD, Seung HS. Learning the parts of objects by non-negative matrix factorization. *Nature* 401: 788–791, 1999. doi:10.1038/44565.

Lünenburger L, Bolliger M, Czell D, Müller R, Dietz V. Modulation of locomotor activity in complete spinal cord injury. *Exp Brain Res* 174: 638–646, 2006. doi:10.1007/s00221-006-0509-4.

McKay JL, Ting LH, Hackney ME. Balance, body motion, and muscle activity after high-volume short-term dance-based rehabilitation in persons with Parkinson disease: a pilot study. *J Neurol Phys Ther* 40: 257–268, 2016. doi:10.1097/NPT.0000000000000150.

McKee KE, Hackney ME. The effects of adapted tango on spatial cognition and disease severity in Parkinson's disease. *J Mot Behav* 45: 519–529, 2013. doi:10.1080/00222895.2013.834288.

Miller RA, Thaut MH, McIntosh GC, Rice RR. Components of EMG symmetry and variability in parkinsonian and healthy elderly gait. *Electroencephalogr Clin Neurophysiol* 101: 1–7, 1996. doi:10.1016/0013-4694(95)00209-X.

Oliveira AS, Gizzi L, Kersting UG, Farina D. Modular organization of balance control following perturbations during walking. *J Neurophysiol* 108: 1895–1906, 2012. doi:10.1152/jn.00217.2012.

Oliveira AS, Silva PB, Lund ME, Gizzi L, Farina D, Kersting UG. Effects of perturbations to balance on neuromechanics of fast changes in direction during locomotion. *PLoS One* 8: e59029, 2013a. doi:10.1371/journal.pone.0059029.

Park JH, Mancini M, Carlson-Kuhta P, Nutt JG, Horak FB. Quantifying effects of age on balance and gait with inertial sensors in community-dwelling healthy adults. *Exp Gerontol* 85: 48–58, 2016. doi:10.1016/j.exger.2016.09.018.

Pérez-Nombela S, Barroso F, Torricelli D, de Los Reyes-Guzmán A, Del-Ama AJ, Gómez-Soriano J, Pons JL, Gil-Agudo Á. Modular control of gait after incomplete spinal cord injury: differences between sides. *Spinal Cord* 55: 79–86, 2017. doi:10.1038/sc.2016.99.

Petzinger GM, Fisher BE, McEwen S, Beeler JA, Walsh JP, Jakowec MW. Exercise-enhanced neuroplasticity targeting motor and cognitive circuitry in Parkinson's disease. *Lancet Neurol* 12: 716–726, 2013. doi:10.1016/S1474-4422(13)70123-6.

Petzinger GM, Fisher BE, Van Leeuwen JE, Vukovic M, Akopian G, Meshul CK, Holschneider DP, Nacca A, Walsh JP, Jakowec MW. Enhancing neuroplasticity in the basal ganglia: the role of exercise in Parkinson's disease. *Mov Disord* 25, *Suppl* 1: S141–S145, 2010. doi:10.1002/mds.22782.

Petzinger GM, Walsh JP, Akopian G, Hogg E, Abernathy A, Arevalo P, Turnquist P, Vucković M, Fisher BE, Togasaki DM, Jakowec MW. Effects of treadmill exercise on dopaminergic transmission in the 1-methyl-4-phenyl-1,2,3,6-tetrahydropyridine-lesioned mouse model of basal ganglia injury. *J Neurosci* 27: 5291–5300, 2007. doi:10.1523/JNEUROSCI.1069-07.2007.

Racette BA, Rundle M, Parsian A, Perlmutter JS. Evaluation of a screening questionnaire for genetic studies of Parkinson's disease. *Am J Med Genet* 88: 539–543, 1999. doi:10.1002/(SICI)1096-8628(19991015)88:5<539::AID-AJMG19>3.0.CO;2-S.

Rikli RA, Jones CJ. Development and validation of a functional fitness test for community-residing older adults. *J Aging Phys Act* 7: 129–161, 1999. doi:10.1123/japa.7.2.129.

Robichaud JA, Pfann KD, Leurgans S, Vaillancourt DE, Comella CL, Corcos DM. Variability of EMG patterns: a potential neurophysiological marker of Parkinson's disease? *Clin Neurophysiol* 120: 390–397, 2009. doi:10.1016/j.clinph.2008.10.015.

Rodriguez KL, Roemmich RT, Cam B, Fregly BJ, Hass CJ. Persons with Parkinson's disease exhibit decreased neuromuscular complexity during gait. *Clin Neurophysiol* 124: 1390–1397, 2013. doi:10.1016/j.clinph.2013.02.006.

Roemmich RT, Fregly BJ, Hass CJ. Neuromuscular complexity during gait is not responsive to medication in persons with Parkinson's disease. *Ann Biomed Eng* 42: 1901–1912, 2014. doi:10.1007/s10439-014-1036-2.

Routson RL, Clark DJ, Bowden MG, Kautz SA, Neptune RR. The influence of locomotor rehabilitation on module quality and post-stroke hemiparetic walking performance. *Gait Posture* 38: 511–517, 2013. doi:10.1016/j.gaitpost.2013.01.020.

Sammon JW Jr. A nonlinear mapping for data structure analysis. *IEEE Trans Comput* C-18: 401–409, 1969. doi:10.1109/T-C.1969.222678.

Sawers A, Allen JL, Ting LH. Long-term training modifies the modular structure and organization of walking balance control. *J Neurophysiol* 114: 3359–3373, 2015. doi:10.1152/jn.00758.2015.

Shumway-Cook A, Woollacott MH. *Motor Control: Theory and Practical Applications*. Baltimore, MD: Lippincott, Williams, & Wilkins, 1995.

Siegel KL, Metman LV. Effects of bilateral posteroventral pallidotomy on gait of subjects with Parkinson disease. *Arch Neurol* 57: 198–204, 2000. doi:10.1001/archneur.57.2.198.

Smulders K, Dale ML, Carlson-Kuhta P, Nutt JG, Horak FB. Pharmacological treatment in Parkinson's disease: effects on gait. *Parkinsonism Relat Disord* 31: 3–13, 2016. doi:10.1016/j.parkreldis.2016.07.006.

Stapley PJ, Drew T. The pontomedullary reticular formation contributes to the compensatory postural responses observed following removal of the support surface in the standing cat. *J Neurophysiol* 101: 1334–1350, 2009. doi:10.1152/jn.91013.2008.

Stebbins GT, Goetz CG, Burn DJ, Jankovic J, Khoo TK, Tilley BC. How to identify tremor dominant and postural instability/gait difficulty groups with the movement disorder society unified Parkinson's disease rating scale: comparison with the unified Parkinson's disease rating scale. *Mov Disord* 28: 668–670, 2013. doi:10.1002/mds.25383.

Steele KM, Rozumalski A, Schwartz MH. Muscle synergies and complexity of neuromuscular control during gait in cerebral palsy. *Dev Med Child Neurol* 57: 1176–1182, 2015. doi:10.1111/dmcn.12826.

Tang L, Li F, Cao S, Zhang X, Wu D, Chen X. Muscle synergy analysis in children with cerebral palsy. *J Neural Eng* 12: 046017, 2015. doi:10.1088/1741-2560/12/4/046017.

Ting LH, Chiel HJ, Trumbower RD, Allen JL, McKay JL, Hackney ME, Kesar TM. Neuromechanical principles underlying movement modularity and their implications for rehabilitation. *Neuron* 86: 38–54, 2015. doi:10.1016/j.neuron.2015.02.042.

Ting LH, Macpherson JM. A limited set of muscle synergies for force control during a postural task. *J Neurophysiol* 93: 609–613, 2005. doi:10.1152/jn.00681.2004.

Torres-Oviedo G, Ting LH. Muscle synergies characterizing human postural responses. *J Neurophysiol* 98: 2144–2156, 2007. doi:10.1152/jn.01360.2006.

Vaillancourt DE, Prodoehl J, Verhagen Metman L, Bakay RA, Corcos DM. Effects of deep brain stimulation and medication on bradykinesia and muscle activation in Parkinson's disease. *Brain* 127: 491–504, 2004. doi:10.1093/brain/awh057.

Zar JH. *Biostatistical Analysis*. Upper Saddle River, NJ: Prentice-Hall, 1999, p. 663.