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Full Length Article

Task specificity and neural adaptations after balance learning in young adults

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ABSTRACT

Background: Only 30 min of balance skill training can significantly improve behavioral and neuromuscular outcomes. However, it is unclear if such a rapidly acquired skill is also retained and transferred to other untrained balance tasks.
Research question: What are the effects of a single balance training session on balance skill acquisition, retention, and transferability and on measures of neural plasticity examined by transcranial magnetic brain stimulation (TMS) and inter-muscular coherence?
Methods: Healthy younger adults ($n = 36$, age 20.9, 18 M) were randomly assigned to: Balance training (BT); Active control (cycling training, CT) or non-active control (NC) and received a 20-min intervention. Before, immediately and ~ 7 days after the interventions, we assessed performance in the trained wobble board task, untrained static standing tasks and dynamic beam walking balance tasks. Underlying neural plasticity was assessed by tibialis anterior motor evoked potential, intracortical facilitation, short-interval intracortical inhibition and long-interval intracortical inhibition using TMS and by inter-muscular coherence.
Results: BT, but not CT (18%, $d = 0.32$) or NC (−1%, $d = -0.02$), improved balance performance in the trained, wobble board task by 207% (effect size $d = 2.12$). BT retained the acquired skill after a 1-week no-training period (136%, $d = 1.57$). No changes occurred in 4 measures of balance beam walking, in 8 measures of static balance, in 8 measures of intermuscular coherence, and in 4 TMS measures of supra-spinal plasticity (all $p > 0.05$).
Significance: Healthy young adults can learn a specific balance skill very rapidly but one should be aware that while such improvements were retained, the magnitude of transfer (32%, $d = 0.94$) to other balancing skills was statistically not significant. Additional studies are needed to determine the underlying neural mechanisms of rapid balance skill acquisition, retention, and transfer.

1. Introduction

Several weeks of balance training (BT) consisting of postural stabilization tasks on wobble boards, spinning tops, soft mats, and

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cushions improves standing and dynamic balance and reduces the risks for falls and lower extremity injuries (Beck et al., 2007; Gruber et al., 2007; Schubert et al., 2008; Taube et al., 2007; Taube, Gruber, & Gollhofer, 2008). In manual motor skill learning, substantial skill acquisition already occurs after a single training session (Berghuis et al., 2019).While BT usually comprises multiple training sessions over several weeks, akin to manual motor skill learning, a single, 30-min BT on an unstable device successfully improved performance in the trained balance skill (van Dieën, van Leeuwen, & Faber, 2015). However, it is unknown if such a rapidly acquired balance skill is also retained, which would indicate that the skill was not only acquired but also consolidated into motor memory. In addition, training specificity, transferability, and the neural mechanisms underlying adaptations and retention following a single BT session remain unexplored.

It is possible that not only balance exercises, but also exercises producing muscle activation not specific to balance outcomes such as stationary cycling, could improve balance performance. Two, 5-week-long interventions at the same cardiovascular load consisting of either BT or a non-balance specific stationary cycling training, both improved balance performance in Parkinsonian patients and healthy older adults (Tollar, Nagy, & Hortobágyi, 2019). Yet in other studies balance-specific intervention improved the balance-specific outcomes more effectively (Schubert et al., 2008; Taube, Gollhofer, & Lauber, 2020; Tollár et al., 2019). Previous balance training studies in young adults for the most part lacked active control groups and did not test the possibility that non-specific muscle contraction could improve balance performance (Beck et al., 2007; Gruber et al., 2007; Schubert et al., 2008; Taube et al., 2007, 2008). Therefore, our first hypothesis was that 20 min of balance specific exercise but not exercise producing muscle activity not specific to balance performance would improve task-specific balance which would be retained after ~7 days of rest.

In addition to training specificity, transferability of a trained balance skill to other balance tasks also remains unclear. Balance training studies in older adults and Parkinson's patients suggest that the skill to maintain one's balance is a general ability and when trained, improvements occur independent of the type and nature of a test (Agmon, Perry, Phelan, Demiris, & Nguyen, 2011; Tollar et al., 2019; Tollár et al., 2019). These studies reported improvements in standing sway and walking speed, tasks that have a different spatial and temporal structure than the training task and tend to support the idea that contractile activity affords a physical conditioning effect which acts as a mediator for the non-balance specific adaptations to balance skill training. However, in young adults, the effects of balance training appear to be task-specific: Four weeks of slackline training improved balance performance on the rope but not sway in quiescent standing (Keller, Pfusterschmied, Buchecker, Müller, & Taube, 2012). In addition, improvements in a trained balance task as well as in an untrained task following BT revealed highly task-specific effects with no transfer even to balance tasks on the same device as the training device or tasks involving the trained perturbation direction (Giboin, Gruber, & Kramer, 2015). Therefore, the present study aimed to examine transfer not only to static standing balance but also to a highly challenging dynamic balance task. Because this highly challenging dynamic task is more in line with the spatial and temporal structure of the trained task compared to the less challenging static balance task, our second hypothesis was that the level of transfer would increase when tested with a task that is increasingly more difficult and similar to the training task.

Coherence between EMG signals from different muscles, i.e., intermuscular coherence, indicates the presence of common synaptic inputs to the motor neuron pools of two or more muscles (Farina, Merletti, & Enoka, 2014). The central nervous system (CNS) drives muscles to discharge action potentials at a number of frequencies. The frequency at which coherence emerges can provide information about the contribution of distinct circuits to the presynaptic input (Grosse, Cassidy, & Brown, 2002). Data from spinal cord injury patients suggest that coherence in the 2–14 Hz frequency range originates from spinal circuits (Aguiar, Baker, Gant, Bohorquez, & Thomas, 2018; Norton & Gorassini, 2006) while intermuscular coherence within the 15–30 Hz frequency band has been shown to originate from the cortex (Grosse et al., 2002). During standing, intermuscular coherence was shown in the 0–5 Hz frequency band and reflects a co-modulation of muscle activity which is suggested to originate in the subcortical system (Boonstra et al., 2008; De Luca & Erim, 2002). Intermuscular coherence in the 15-30 Hz band was also reported in standing and since corticomuscular coherence has been reported mainly in this band, it has been suggested to reflect corticospinal contributions (Nandi, Hortobágyi, van Keeken, Salem, & Lamoth, 2019; Walker, Piitulainen, Manlangit, Avela, & Baker, 2020; Watanabe, Saito, Ishida, Tanabe, & Nojima, 2018). Intermuscular coherence analyses in these frequency bands can therefore provide insights into the neural mechanisms underlying rapid motor skill acquisition (De Luca & Erim, 2002; Grosse et al., 2002). Indeed, increasing the difficulty of a postural task increases intermuscular coherence in alpha or beta frequency bands for agonist or antagonist muscle pairs, respectively (Nandi et al., 2019). Acquisition of a visuo-motor skill involving the ankle joint affected the central drive to the motor neuron pool and increased intramuscular coherence within the tibialis anterior muscle at 15-35 Hz (Perez, Lundbye-Jensen, & Nielsen, 2006). Therefore, our third hypothesis was that short-term balance learning would increase intermuscular coherence between agonist muscle pairs but reduce intermuscular coherence between antagonist pairs mostly in the 15–30 Hz frequency band.

Transcranial magnetic brain stimulation (TMS) can aid the interpretation and further specify the neural mechanisms of adaptations to BT. On the corticospinal level, acquisition of a motor skill in a single session has been consistently associated with reductions in short interval intracortical inhibition (SICI) and with increases in corticospinal excitability at rest (Perez, Lungholt, Nyborg, & Nielsen, 2004). However, the pattern of adaptation is task-dependent. While strength training tends to reduce SICI (Weier, Pearce, & Kidgell, 2012), practice of tasks that requires complex movement sequences tend to increase SICI (Dai et al., 2016). This increase in SICI may help focus control and suppress unwanted movements and increase differences in the excitability states between active and resting muscles. Switching from a simple to more challenging postural tasks consistently leads to increased corticospinal excitability and reduced intracortical inhibition (Papegaaij, Baudry, Négyesi, Taube, & Hortobágyi, 2016). Several weeks of BT alters the balance between inhibition and excitation by reducing the corticospinal and cortical excitability (Beck et al., 2007; Penzer, Duchateau, & Baudry, 2015; Schubert et al., 2008). Two weeks of BT on an unstable platform also brought about correlated improvements in balance performance and increases in SICI (Mouthon & Taube, 2019). The time course of such changes is unknown but based on TMS data of manual skill acquisition (Berghuis et al., 2019), it is conceivable that such changes are already present after the very first BT session.

Taking the neurophysiological data together, our fourth hypothesis is that BT would lead to task-specific adaptations in cortical and corticospinal excitability during standing but not after cycling and not during sitting. Overall, the aim of the present study was to examine the effects of 20 min of unstable balance board training on balance skill acquisition, retention, and transfer and to probe the underlying neurophysiological mechanisms by intermuscular coherence and TMS.

2. Methods

2.1. Participants

Thirty-six healthy young adults participated in this study (18 M, age: 20.9 ± 1.9y, height: 1.8 m ±0.1, mass: 69.5 kg ±8.2). Exclusion criteria were: prior wobble board training, a history of neurological or orthopedic disorders, seizures, head trauma, suspicion of pregnancy, metal implants, pacemaker, or blood relatives with a history of seizures. Participants signed an informed consent document which complied with the Declaration of Helsinki and was approved by the University Medical Centre Ethics Committee.

2.2. Interventions

Participants were randomly assigned to a balance-specific training (BT, $n = 12$), a non-balance specific training: cycling training (CT, n = 12), or to a non-active control (NC, n = 12) group. BT consisted of bipedal standing on a 0.4-m diameter wooden wobble board (maximal 3D tilting angle: 20°; Miniboard, Sensamove®, Groessen, NL). Feet were placed on two marked lines positioned 0.2-m apart to ensure consistent foot placement across measurements. In the BT group, participants performed 3 blocks (5 min apart) of 8, 45-s-long bipedal standing trials on the wobble board with 30-s of rest in between. Instruction was: 'Keep the board horizontal, with the arms folded at the chest'. For safety, participants wore a harness suspended from the ceiling, that allowed free movement. The board's deviation from horizontal and a 5° circular deviation target were displayed on a screen in front of the participant at eye level. The CT group cycled on a seated ergometer at a resistance set to a rate of perceived exertion (RPE) of 12–13 and completed 4, 5-min-long bouts with 2 min of rest in between. The NC group sat at a table in the lab, read newspapers or watched TV for 20 min.

2.3. Data acquisition and analysis

At baseline, the international physical activity questionnaire (IPAQ) and the Pittsburgh sleep quality index (PSQI) were used to determine general physical activity in daily life and the quality of sleep. Subjects also performed the short physical performance battery (SPPB) including standing balance, walking speed and chair stand test to assess lower extremity function. Leg preference was determined by asking which foot was used when kicking a ball, pushing an object with the foot and stamping on the floor. Before (pretest), immediately (posttest) and ~ 7 days after the interventions (retention-test), we assessed performance of the trained wobble board task and of untrained static and dynamic balance tasks and probed underlying neural plasticity by TMS and by analyzing intermuscular coherence of recorded muscle activity.

The assessment of the wobble board task, a task similar to the BT intervention task, consisted of standing on the board and keeping the unstable board 'in balance' during 2, 30-s trials with 20-s of rest in between. Before the wobble board test, participants were familiarized with the wobble board during two 15-s trials. Using Sensamove software, the board angle was determined based on the acceleration sensors embedded in the board. Balance performance was defined as the percent of total time the board angle was within 5° relative to horizontal in anterior-posterior (AP) and medio-lateral (ML) direction) and averaged for the two trials.

In the dynamic balance task, participants walked twice on 3- and 6 cm-wide, 3 m-long wooden beams with arms folded across the chest with or without a calculation task (subtraction-by-7 between 300 and 900). Participants were allowed to become familiar with the beam walking task. Distance walked without stepping off the beam or unfolding the arms, was recorded by measuring the distance between the start of the beam and the position of the heel of the foot that was on the beam when balance was lost and averaged for the two trials under each condition.

In the static balance task, participants stood as still as possible, arms folded across the chest, during 8 different standing conditions, 20s per condition. Participants were asked to stand with feet in a wide or narrow position on a rigid or foam surface with eyes open or closed in a random order. Using two force plates (Bertec 4060–08, Columbus, OH, USA), force and moment data were sampled at 200 Hz. The Centre of Pressure (COP) was calculated (LabVIEW, v2015, National Instruments, Austin, TX, USA) and filtered using a low-pass Butterworth filter with a cut off frequency of 5 Hz. From the ML and AP COP signals, the resultant COP vector was calculated. The distances between two consecutive data points of the resultant COP signal were summed and then divided by the sum of the time between these data points to calculate the mean resultant COP velocity for each of the 8 standing balance task conditions.

Muscle activity of the Tibialis Anterior (TA), Peroneus Longus (PL), Soleus (SOL), Lateral Gastrocnemius (LG), Rectus Femoris (RF) and Biceps Femoris (BF) of the dominant leg was recorded using surface wireless pre-amplified EMG sensors (37x25x15mm, Trigno™ Wireless System, Delsys, Natick, MA, USA) placed over the bulk of the muscle belly of each individual muscle, located by palpating during voluntary muscle contraction following SENIAM guidelines. The EMG signal was amplified 1000×, and sampled at 5 kHz (1401 A/D Power Board, Signal software v5.11, Cambridge Electronic Design Ltd., Cambridge UK). EMG data was bandpass filtered using a 4th order dual pass Butterworth filter with 10 Hz and 1000 Hz high and low cutoffs. The signal was rectified using the Hilbert-transform and intermuscular coherence (EMG-EMG coherence) was computed between three synergistic muscle pairs: SOL-LG, SOL-PL, LG-PL, and three antagonistic muscle pairs: TA-SOL, TA-LG and RF-BF. To calculate intermuscular coherence, we determined the auto-spectrum (f_{xx} and f_{yy}) of each muscle as well as the cross spectrum (f_{xy}) of each muscle pair using Welch's periodogram method. Intermuscular coherence was estimated for pairs of EMG signals by normalizing f_{xy} squared by the product of f_{xx} and f_{yy} at each frequency (λ) (Halliday et al., 1995).

$$|R_{xy}(\lambda)|^2 = \frac{|f_{xy}(\lambda)|^2}{f_{xx}(\lambda)f_{yy}(\lambda)}$$

Intermuscular coherence estimates were considered statistically significant when they exceeded the confidence limit (at $\alpha = 0.05$) for the number of disjoint segments (L) used to estimate the spectrum (Rosenberg, Amjad, Breeze, Brillinger, & Halliday, 1989), which was determined by:

$$1-(1-\alpha)^{\frac{1}{L-1}}$$

To estimate the pooled coherence between agonist-agonist (AG-AG) muscle pairs the synergistic muscle pairs were combined. The antagonistic muscle pairs were combined to estimate pooled coherence between agonist-antagonist (AG-ANT) muscle pairs. The pooled coherence analysis were performed using the following equation (Amjad, Halliday, Rosenberg, & Conway, 1997):

$$\frac{\left|\Sigma_{i=1}^{k} f_{xy}(\lambda)L_i\right|^2}{\left(\Sigma_{i=1}^{k} f_{xx}(\lambda)L_i\right)\left(\Sigma_{i=1}^{k} f_{yy}(\lambda)L_i\right)}$$

Where k is the number of muscle pairs pooled together (3 for AG-AG and 3 for AG-ANT) and L_i is the total number of segments used to estimate the spectrum. In order to compare coherence estimations across participants and during different experimental conditions, all estimates were z-transformed by computing the Fisher transformation of the estimates. Thereafter, the frequency distribution was assessed by integrating the single pair and pooled coherence in two different frequency bands: 0-12 Hz (alpha) and 13-30 Hz (beta). These frequency bands were chosen based on the frequency intervals reported in previous standing studies, and probable neural origin of coherent signals (Grosse et al., 2002; Nandi et al., 2019). Single pair and pooled coherences values were only included in the ANOVA if the level of coherence exceeded the confidence limit.

Two single-pulse magnetic stimulators (Magstim Model 2002, The Magstim Co., Whitland, UK), a Bistim module and a double cone coil (inner loop diameter 110 mm) were used to magnetically stimulate the primary motor cortex contralateral to the preferred leg. The coil was moved in 1 cm increments to determine the hot-spot location for the TA. The hot-spot location was defined as the location where the largest and most consistent motor evoked potential (MEP) was obtained. To ensure consistent positioning of the coil, participants wore a cloth cap on which the hot-spot could be marked. The coil was positioned so that the current in the coil flowed in an anterior-to-posterior direction. The resting motor threshold (rMT) and the active motor threshold (aMT) were determined in sitting and standing respectively by systematically varying the stimulation output to find the lowest level of stimulator output at which three out of five MEPs had a peak-to-peak amplitude of at least 50 μV. During sitting and standing, participants received in a random order 10-paired pulses each for the short interval intracortical inhibition (SICI), long interval intracortical inhibition (LICI) and intracortical facilitation (ICF) protocols and 10 single pulses. For eliciting SICI, (GABAa mediated) and ICF, the conditioning and test pulse were set at 70% and 110% MT, respectively. For LICI (GABAb mediated), the conditioning and test pulse were set at 120% and 110% MT respectively. The intensity for the single pulses was set at 110% MT. An inter-stimulus interval (ISI) of 3, 13 and 100-ms was used for SICI, ICF and LICI, respectively, parameters were chosen based on an extensive piloting in a previous study (Nandi et al., 2018).

The filtered EMG was rectified and the peak-to-peak amplitude was determined for each of the 10 trials and then averaged for the MEP, SICI, LICI and ICF condition in both sitting and standing. SICI, LICI and ICF conditions were then divided by the MEP value and expressed as a percentage of the MEP amplitude. Values below 100% indicate inhibition while values above 100% represent facilitation.

2.4. Statistical analyses

Statistical analyses were performed in SPSS (version 25, IBM Corp, Armonk, NY, USA). All variables were checked for normal distribution using the Shapiro-Wilk test prior to analysis. Outcomes that were not distributed normally were log transformed. We computed posttest minus pretest and retention minus pretest delta scores for the wobble board task, static balance task, dynamic balance task, TMS outcome measures and intermuscular coherence values. We compared normally distributed outcome measure between BT, CT, and NC at baseline using a one-way ANOVA. One-way ANOVAs on the normally distributed delta scores compared BT, CT, and NC. A significant effect, characterized by partial η^2 effect size (ES), was interpreted as a group by time interaction and was followed by a Tukey's posthoc. Within group changes were further characterized by Cohen's ESs (small:0.20; moderate:0.50; large:0.80) (Cohen, 1988). Data that was non-normally distributed, even after transformation, was analyzed using an independent samples Kruskal-Wallis test. The Holm method was used to correct for familywise error. Pearson's correlation was computed between pre-to-post delta scores for the time in balance on the board and the MEP, SICI, LICI and ICF values assessed during standing. The level of significance was set at $p < 0.05$.

3. Results

Table 1 shows that the three groups did not differ at pretest. All participants were right-leg dominant, except for one in BT and one in CT.

3.1. Behavioral outcomes

There was a group by time effect for time in balance on the balance board (H(2) = 17.86, $p < 0.00$ Table 2), so that time in balance pre-to-posttest improved by 207% (d = 2.12) in BT, which exceeded the +18% (d = 0.32) and − 1% (d = −0.02) adaptations in CT and NC (Fig. 1). These improvements were retained after ~7 days ($F_{2,33} = 8.72$, $p < 0.00$, Table 3), so that time in balance pre-to-retention test improvements in BT (136%, d = 1.57) exceeded improvements in CT (33%, d = 0.48) and in NC (6%, d = 0.10, Fig. 1).

In the static balance transfer tasks, neither intervention affected COP velocity in the 8 conditions between pre-to-posttest (all $p >$ 0.05, Table 2) or between pre-to-retention test (all $p > 0.05$, Table 3).

Neither intervention did significantly improve the distance walked on the wide or the narrow beam during single and dual task at posttest (Table 2) and retention (Table 3). To highlight, beam walking distance increased after BT during single task at retention by 32% (d = 0.94) on the narrow and by 15% (d = 1.06) on the wide beam with minimal changes in the other two groups (−10% to +0.2%). However, this interaction did not survive Holm's correction for familywise error.

3.2. Intermuscular coherence and TMS outcomes

There were no significant intervention effects on coherence in any muscle pairs or in any tasks (all $p > 0.05$). Fig. 2 shows representative pooled coherence data during the rigid surface, wide stance, eyes open task (RWEO) before and after the training intervention in the balance training group. In addition, neither intervention affected TMS outcomes between pre-to-post (Table 2, all p > 0.05) and between pre-to-retention test (all $p > 0.05$, Table 3).

Table 1

Participant characteristics and pretest scores for balance behavior and TMS measures in balance training group (BT, $n = 12$, 6 males), cycling training active control group (CT, n = 12, 6 males) and no-intervention control group (NC, n = 12, 6 males).

	BT	CT	NC	*p*-value
Age (years)	20.67 ± 1.07	20.58 ± 1.93	21.58 ± 2.50	$0.37^{\#}$
Height (m)	1.79 ± 0.10	1.79 ± 0.09	1.77 ± 0.08	$0.88^{\#}$
Weight (kg)	70.08 ± 10.26	68.58 ± 8.04	69.75 ± 6.18	0.90
SPPB score	11.92 ± 0.29	12.00 ± 0.00	11.83 ± 0.58	$0.56^{\#}$
IPAQ score	3730.92 ± 1617.33	4656.17 ± 2290.39	2650.33 ± 1085.58	0.06^{T}
PSQI score	4.00 ± 1.41	4.00 ± 1.41	4.50 ± 1.68	$0.53^{\#}$
Balance Board – time in balance (%)	16.13 ± 9.24	18.96 ± 10.41	29.91 ± 18.57	0.10^{T}
Balance Beam wide – distance (m)				
ST	2.52 ± 0.48	2.86 ± 0.28	2.93 ± 0.24	$0.05^{\#}$
DT	2.58 ± 0.47	2.92 ± 0.23	2.75 ± 0.32	$0.06^{\#}$
Balance Beam narrow– distance (m)				
ST	1.59 ± 0.69	1.89 ± 0.76	2.38 ± 0.48	0.02◆
DT	1.65 ± 0.50	1.89 ± 0.77	1.92 ± 0.75	0.71^{T}
Standing Balance – COP velocity (cm/s)				
RWEO	0.45 ± 0.10	0.47 ± 0.17	0.50 ± 0.17	0.65
RNEO	0.60 ± 0.22	0.56 ± 0.18	0.54 ± 0.14	0.70
RWEC	0.80 ± 0.26	0.70 ± 0.37	0.66 ± 0.20	0.46^{T}
RNEC	0.96 ± 0.35	0.96 ± 0.32	0.97 ± 0.34	1.00^{T}
FWEO	0.86 ± 0.61	0.79 ± 0.17	0.82 ± 0.13	$0.39^{\#}$
FNEO	0.81 ± 0.25	0.87 ± 0.26	0.94 ± 0.19	0.36^{T}
FWEC	1.74 ± 0.61	2.07 ± 0.96	1.82 ± 0.55	$0.74^{\#}$
FNEC	2.87 ± 1.30	2.64 ± 1.27	2.58 ± 0.61	0.79
TMS sitting				
rMT	60.33 ± 10.31	57.09 ± 8.63	59.50 ± 10.19	0.70
MEP (mV)	0.18 ± 0.08	0.19 ± 0.14	0.21 ± 0.17	$0.95^{\#}$
SICI (% of MEP sitting)	73.85 ± 43.85	68.51 ± 33.47	72.93 ± 41.32	0.94
LICI (% of MEP sitting)	23.39 ± 26.12	23.69 ± 15.55	18.26 ± 14.95	0.57^{T}
ICF (% of MEP sitting)	156.15 ± 53.45	188.44 ± 94.48	142.39 ± 41.80	0.28^{T}
TMS standing				
aMT	58.80 ± 11.34	56.91 ± 11.16	54.75 ± 10.68	0.71
MEP (mV)	0.26 ± 0.17	0.18 ± 0.14	0.18 ± 0.17	$0.22^{\#}$
SICI (% of MEP standing)	78.23 ± 38.40	56.37 ± 27.24	94.54 ± 41.53	0.05◆
LICI (% of MEP standing)	33.14 ± 36.54	31.56 ± 39.45	77.47 ± 110.80	0.45^{T}
ICF (% of MEP standing)	210.81 ± 103.55	253.74 ± 135.72	184.45 ± 58.75	0.41^{T}

Values are presented as means ± SD; Bold p-value indicate significant between group differences ($p < 0.05$); #: Kruskal-Wallis test; T: one-way ANOVA was performed on transformed data; ◆: Did not survive Holm correction for family wise error (Balance Beam narrow ST: $p = 0.017$ cut off, TMS standing SICI: $p = 0.010$ cut off); SPPB: Short Physical Performance Battery (max. Score of 12); IPAQ: International Physical Activity Questionnaire – short version (score in MET hours); PSQI: Pittsburgh Sleep Quality Index (score > 5 indicates poor sleep); ST: single task; DT: dual task; R: rigid surface; F: foam surface; EO: eyes open; EC: eyes closed, W: wide stance, N: narrow stance; rMT: resting Motor Threshold; aMT: active motor threshold; MEP: motor evoked potential; SICI: short-interval intracortical inhibition, LICI: long-interval intracortical inhibition; ICF: intracortical facilitation.

Table 2

Pretest vs. posttest difference scores for all balance behavior and TMS measures in balance training group (BT, n = 12), cycling training active control group (CT, *n* = 12) and no-intervention control group (NC, n = 12) as well as the results of the Analysis of Variance (ANOVA) or Kruskal-Wallis test.

	Δ Score (post-pretest)			ANOVA		
	BT	CT	NC	F(2,33) / H(2)	P-value	Partial η^2
Balance board - time in balance (%)	+33.44 ± 19.12	+3.50 ± 8.99	−0.32 ± 10.24	17.86$^{\#}$	**<0.00**^{a}	0.51
Balance beam wide – distance (m)						
ST	+0.39 ± 0.43	+0.07 ± 0.42	−0.01 ± 0.34	6.05$^{\#}$	0.05◆	0.17
DT	+0.35 ± 0.41	−0.06 ± 0.42	+0.13 ± 0.29	4.45$^{\#}$	0.11	0.13
Balance beam narrow– distance (m)						
ST	+0.53 ± 0.82	+0.34 ± 0.77	+0.09 ± 0.50	1.19	0.32	0.07
DT	+0.09 ± 0.58	+0.03 ± 0.94	−0.02 ± 0.69	0.07	0.94	0.00
Standing balance – COP velocity (m/s)						
RWEO	+0.08 ± 0.11	+0.03 ± 0.14	+0.01 ± 0.23	0.69	0.51	0.04
RNEO	+0.07 ± 0.42	+0.09 ± 0.15	+0.06 ± 0.16	0.23$^{\#}$	0.89	0.01
RWEC	−0.06 ± 0.24	−0.05 ± 0.23	+0.03 ± 0.20	0.52	0.60	0.03
RNEC	+0.06 ± 0.23	+0.00 ± 0.30	−0.05 ± 0.33	0.42	0.66	0.02
FWEO	−0.01 ± 0.58	−0.08 ± 0.28	−0.05 ± 0.16	4.11$^{\#}$	0.13	0.12
FNEO	+0.20 ± 0.72	+0.05 ± 0.33	−0.09 ± 0.20	2.99$^{\#}$	0.22	0.09
FWEC	+0.16 ± 0.86	−0.31 ± 0.92	−0.25 ± 0.61	1.21	0.31	0.07
FNEC	−0.35 ± 0.89	+0.03 ± 0.96	−0.07 ± 1.03	0.52	0.60	0.03
TMS sitting						
rMT	+2.92 ± 4.85	+3.27 ± 6.86	−2.83 ± 6.75	5.43$^{\#}$	0.07	0.16
MEP (mV)	−0.04 ± 0.06	−0.02 ± 0.17	−0.06 ± 0.20	2.66$^{\#}$	0.26	0.08
SICI (% of MEP sitting)	+2.62 ± 32.37	+6.70 ± 31.81	+6.77 ± 36.80	0.26$^{\#}$	0.88	0.01
LICI (% of MEP sitting)	+17.45 ± 57.80	+12.89 ± 54.03	+1.92 ± 20.95	0.33$^{\#}$	0.85	0.01
ICF (% of MEP sitting)	+76.06 ± 74.92	+43.06 ± 109.62	+85.60 ± 103.94	0.63	0.54	0.04
TMS standing						
aMT	+3.50 ± 5.93	+1.64 ± 5.01	−0.50 ± 4.17	1.86	0.17	0.10
MEP (mV)	−0.06 ± 0.11	+0.04 ± 0.12	+0.04 ± 0.13	2.72	0.08	0.14
SICI (% of MEP standing)	+12.77 ± 74.92	−4.99 ± 18.20	−15.72 ± 44.71	0.44$^{\#}$	0.80	0.01
LICI (% of MEP standing)	−22.03 ± 31.48	−2.46 ± 33.25	−44.19 ± 115.48	0.73$^{\#}$	0.69	0.02
ICF (% of MEP standing)	+1.34 ± 96.99	−31.22 ± 85.26	+5.24 ± 75.36	0.65	0.53	0.04

Values are presented as means ± SD. Bold *p*-values indicate a significant group x time effect ($p < 0.05$); #: Kruskal-Wallis test; ◆: Did not survive Holm correction for family wise error ($p = 0.013$ cut off); a: Pairwise comparisons indicates that increases in balance board performance in BT exceeded ($p < 0.05$) the changes in CT and NC; ST: single task; DT: dual task; R: rigid surface; F: foam surface; EO: eyes open; EC: eyes closed, W: wide stance, N: narrow stance; rMT: resting Motor Threshold; aMT: active motor threshold; MEP: motor evoked potential; SICI: short-interval intracortical inhibition, LICI: long-interval intracortical inhibition; ICF: intracortical facilitation.

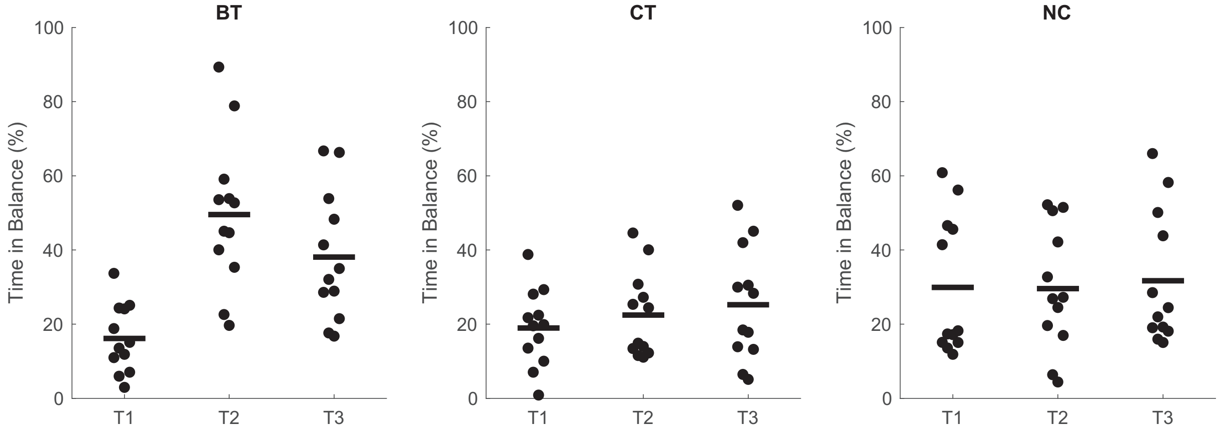

Fig. 1. Time in balance (%) on the unstable balance board at pretest (T1), posttest (T2) and retention (T3) for the balance training (BT), cycling training (CT), and the no-intervention control (NC) group. Horizontal short lines denote the mean value, each dot represents one participant at one of the measurement moments.

3.3. Correlation analyses

Increases in time in balance on the board did not correlate with the MEP during standing at pre-test ($r < 0.19$; $p > 0.293$). Improvements in time in balance on the board did not correlate with changes in the MEP, SICI, LICI and ICF, assessed during standing, at

Table 3
Pretest vs. retention difference scores for all balance behavioral and TMS measures in the balance training group (BT, n = 12), cycling training active control group (CT, $n = 12$) and no-intervention control group (NC, $n = 12$) as well as the results of the Analysis of Variance (ANOVA) or Kruskal-Wallis test.

	Δ Score (retention-pretest)			ANOVA		
	BT	CT	NC	F(2,33) / H(2)	P-value	Partial η^2
Balance Board – time in balance (%)	+21.95 ± 14.44	+6.28 ± 10.09	+1.80 ± 12.34	8.72	**<0.00**^{a}	0.35
Balance Beam wide – distance (m)						
ST	+0.39 ± 0.54	+0.07 ± 0.42	−0.05 ± 0.36	6.04$^{\#}$	0.05◆	0.17
DT	+0.26 ± 0.57	−0.10 ± 0.29	+0.13 ± 0.44	4.03$^{\#}$	0.13	0.12
Balance Beam narrow – distance (m)						
ST	+0.48 ± 0.86	−0.03 ± 0.80	+0.14 ± 0.79	1.21	0.31	0.07
DT	+0.29 ± 0.66	−0.05 ± 0.64	+0.23 ± 0.67	0.88	0.42	0.05
Standing Balance – COP velocity (m/s)						
RWEO	+0.00 ± 0.12	+0.10 ± 0.19	−0.05 ± 0.15	3.10	0.06	0.16
RNEO	−0.04 ± 0.20	+0.06 ± 0.17	+0.04 ± 0.26	0.74	0.48	0.04
RWEC	−0.11 ± 0.18	−0.02 ± 0.14	+0.00 ± 0.20	1.29	0.29	0.07
RNEC	−0.04 ± 0.51	−0.08 ± 0.26	−0.12 ± 0.26	0.16	0.86	0.01
FWEO	−0.03 ± 0.71	+0.07 ± 0.35	−0.15 ± 0.17	2.63$^{\#}$	0.27	0.08
FNEO	+0.06 ± 0.17	+0.04 ± 0.27	−0.04 ± 0.20	0.66	0.53	0.04
FWEC	−0.17 ± 0.51	−0.45 ± 0.78	−0.14 ± 0.61	0.86	0.43	0.05
FNEC	−0.50 ± 0.75	−0.19 ± 1.03	−0.36 ± 0.75	0.40	0.68	0.02
TMS sitting						
rMT	+2.08 ± 9.51	+4.00 ± 7.40	+5.67 ± 7.55	0.57	0.57	0.03
MEP (mV)	−0.01 ± 0.10	−0.05 ± 0.15	+0.02 ± 0.29	0.78$^{\#}$	0.68	0.02
SICI (% of MEP sitting)	−13.78 ± 25.32	+5.77 ± 68.63	+8.97 ± 37.73	0.81	0.46	0.05
LICI (% of MEP sitting)	−4.11 ± 16.25	+0.52 ± 27.04	+48.03 ± 123.11	5.07$^{\#}$	0.08	0.15
ICF (% of MEP sitting)	+59.47 ± 58.65	+40.41 ± 155.91	+130.10 ± 228.33	1.18$^{\#}$	0.56	0.03
TMS standing						
aMT	+0.92 ± 8.62	+0.91 ± 5.70	+3.17 ± 4.26	1.86$^{\#}$	0.40	0.05
MEP (mV)	+0.04 ± 0.16	−0.02 ± 0.19	+0.08 ± 0.16	0.98	0.39	0.06
SICI (% of MEP standing)	−17.95 ± 45.17	−13.82 ± 21.43	−20.06 ± 51.74	0.07	0.93	0.00
LICI (% of MEP standing)	−17.96 ± 33.05	−12.48 ± 32.72	−24.58 ± 151.56	0.01$^{\#}$	1.00	0.00
ICF (% of MEP standing)	−16.87 ± 145.95	−30.73 ± 138.17	+51.85 ± 122.36	1.27	0.29	0.07

Values are presented as means ± SD. Bold p-values indicate a significant group x time effect ($p < 0.05$); #: Kruskal-Wallis test; ◆: Did not survive Holm correction for family wise error ($p = 0.008$ cut off); a: Tukey post hoc tests indicates that increase in balance board performance in BT exceeded ($p < 0.05$) the changes in CT and NC; ST: single task; DT: dual task; R: rigid surface; F: foam surface; EO: eyes open; EC: eyes closed, W: wide stance, N: narrow stance; rMT: resting Motor Threshold; aMT: active motor threshold; MEP: motor evoked potential; SICI: short-interval intracortical inhibition, LICI: long-interval intracortical inhibition; ICF: intracortical facilitation

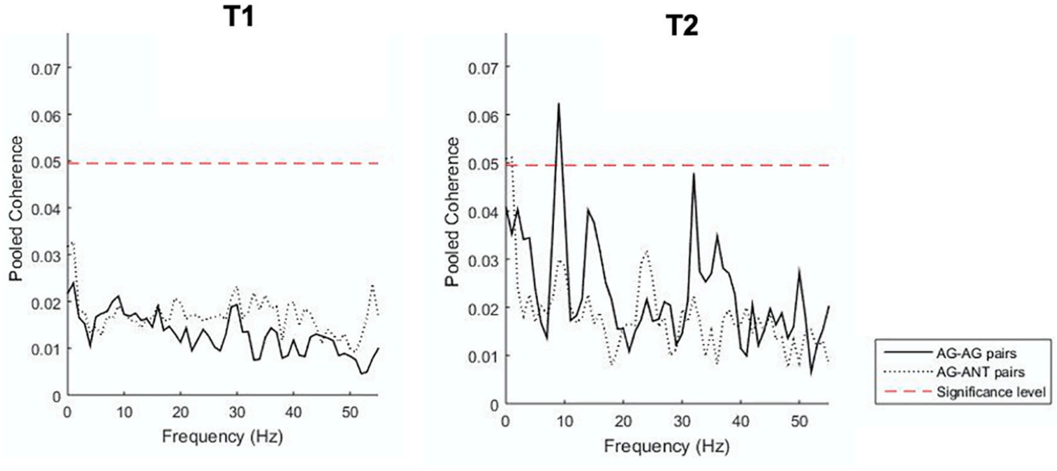

Fig. 2. Pooled intermuscular coherence of synergistic (AG-AG) and antagonistic (AG-ANT) muscle pairs of the balance training group (BT) during solid surface, wide stance, eyes open standing at pretest (T1) and posttest (T2). Solid black lines depict AG-AG muscles, broken black line depict AG-ANT muscles and the red line depicts the significance level. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

post-test (all $r < 0.23$; $p > 0.180$) or retention-test (all $r < 0.16$; $p > 0.340$) in BT.

4. Discussion

We examined the effects of one session of wobble board training on balance skill acquisition, retention, training specificity, and transferability and probed the underlying neural plasticity with intermuscular coherence and TMS in healthy younger adults. The 207% immediate increase and the retention of a 136% increase of the trained balance skill after one week exceeded the increases in the active and passive control groups. Contrary to our hypothesis, improvements in the trained balance skill did not transfer to the untrained standing balance tasks or the untrained beam walking tasks. In addition, the balance improvements failed to induce neural adaptations during sitting and standing.

The substantial increase and retention in the trained balance task after a single session of BT on an unstable board confirms previous data that acquiring a new balance skill is possible during a single, 30-min-long BT session on an unstable device (van Dieën et al., 2015). Similar to manual skill learning, the rapidly improved balance performance was retained, indicating that the skill was not just acquired but consolidated into motor memory. The present results are not unexpected, as the bulk of improvements in balancing skill can occur after only six, 16-s-long balancing trials on an unstable board (van Dieën et al., 2015). Combining the current short-term BT data with those reported previously, we might need to look at the changes reported by long-term BT studies from another perspective. Based on the rapidity and magnitude of manual skill acquisition data we suspected that similar rapid initial learning also dominates balance improvements after long term BT. Even the largest session-normalized improvement of 38% following several weeks of BT was of a smaller magnitude (Giboin et al., 2015; Penzer et al., 2015; Ruffieux, Mouthon, Keller, Walchli, & Taube, 2017; Taube et al., 2007; Taubert et al., 2010; Taubert, Mehnert, Pleger, & Villringer, 2016) than the improvements observed here. While BT and manual motor skill learning appear to share a similar time course, BT tends to produce greater initial gains than the practice of manual motor skills probably because the number of degrees of freedom in a whole body balance task affords a greater potential for re-organizing and optimizing the spatiotemporal structure of the skill compared with manual skills comprising 1–2 degrees of freedom (Berghuis, Semmler, Opie, Post, & Hortobágyi, 2017).

We examined the specificity of training according to intervention type and the current data confirm the principle of training specificity by showing that balance improvements certainly require BT in healthy younger adults. Unlike BT, short-term contractile activity while cycling does not train muscle activation patterns that control and keep the center of gravity within the base of support. Thus, cycling does not seem to afford the rapid, large, and highly task-specific gains in balancing skills and is insufficient to improve balance performance in healthy younger adults. This is not in line with recent interventions in mobility-limited older adults and Parkinson patients showing that seated cycling at the same perceived and cardiovascular load as balance-focused exergaming similarly improved balance and non-balance outcomes (Tollar et al., 2019; Tollár et al., 2019). These studies support the idea that contractile activity affording a physical conditioning effect acts as a mediator of the non-balance specific adaptations to balance skill training (Agmon et al., 2011; Tollar et al., 2019; Tollár et al., 2019). The training-specific effects may be related to participants' age, health, and mobility status, with non-balance specific effects occurring in mobility-limited older adults (Agmon et al., 2011; Tollar et al., 2019; Tollár et al., 2019) but not in healthy younger adults after a single BT session.

We also examined specificity according to transferability of the acquired balance skill. We observed no immediate transfer of the learned balance skill to the static balance or beam walking tasks immediately after BT and CYC (Table 2). At retention, narrow and wide beam walking distance increased by 32% ($d = 0.94$) and 15% ($d = 1.06$) however the *p* value did not survive the Holm's correction. Contrary to our hypothesis we found a lack of transfer to all static balance and beam walking tasks. This is in line with previous studies reporting no transfer even to balance tasks on the same device as the training device or tasks involving the trained perturbation direction (Giboin et al., 2015). Our findings are not in line with the body of literature reporting transferability of a newly learned balanced skill to a non-practiced balance skill (Agmon et al., 2011; Tollar et al., 2019) but this could be confirmed in a study with a sample size greater than the one we used in the present study. However it also possible that the poor transferability of a newly learned balance skill is related to the low correlations between balance outcomes in different balance tasks (Kiss, Schedler, & Muehlbauer, 2018), suggesting low mechanistic links between these tasks. Training on a wobble board, as in the present study, improved the ability to control counter-rotation mechanisms for balance maintenance on the tilting, unstable board (de Silva, Mrachacz-Kersting, Oliveira, & Kersting, 2018). However, such mechanisms may be needed to a much smaller extent while walking on a narrow beam and not needed at all during standing even with sensory perturbations and a reduced base of support. Our data add and confirm evidence that if younger adults train on an unstable surface the test-task must be similar in spatiotemporal structure to the trained task, a finding that has implications for designing and testing training outcomes in athletes who perform BT (Giboin et al., 2015). The current data also adds to the ongoing debate concerning the suitability of standing balance sway to quantify BT effects and, in a broader sense, to assess fall risk (Hortobágyi et al., 2020).

Despite the substantial increase and retention in the trained balance task, we found no adaptations in intermuscular coherence or TMS measures of neural plasticity, resulting in a dissociation between behavioral and neural adaptation. These data agree with a number of manual skill learning (Berghuis et al., 2019) and also BT studies reporting neural changes only during postural perturbation (Beck et al., 2007; Schubert et al., 2008) or in a task similar to the trained task (Ruffieux et al., 2017). However, our data do not agree with the imaging data, showing localized increases in motor cortical thickness after one session of unstable balance board training (Taubert et al., 2016). Several studies reported correlations between adaptations in cortical/corticospinal excitability and changes in balance control (Taube et al., 2007; Taubert et al., 2010; Taubert, Lohmann, Margulies, Villringer, & Ragert, 2011). In addition, a 2-week-long BT intervention reduced motor thresholds during balancing and strengthened the association between improved postural control and increases in SICI (Mouthon & Taube, 2019). Reasons for the dissociation we observed could include differences in the time course of adaptations between behavior and plasticity, low difficulty and different spatiotemporal structure of the test task (sitting, standing) compared with the trained skill, and muscles other (i.e., soleus, peroneus) than the one we tested (tibialis anterior) using TMS showing more prominent adaptations. Additionally, it is possible that neural adaptations in other parts of the brain, like the cerebellum, underlie the observed performance improvements.

One limitation of this study is that the difficulty of the transfer tasks, both the widely used static balance and the dynamic beam walking balance task, was low (ceiling effect), which made the tasks insensitive to detect training-specific adaptations in healthy younger adults. Also, due to the already extensive testing battery, we probed only one muscle by TMS, potentially missing adaptations in other, more responsive muscles. We elicited the TMS stimulus during quiet standing but we did not control for muscle activity. Albeit minimal, sway in standing may affect the amplitude of the evoked potentials because muscle activation increases the amplitude of the MEP elicited by TMS, and the targeted TA muscle is especially active during anterior posterior sway. Not controlling for muscle activity might have therefore increased the variability between trials. We also did not assess intermuscular coherence or administer TMS during the trained balance task. Therefore, we have no neurophysiological data to characterize the specific adaptations to the balance exercise task (Mouthon & Taube, 2019; Taube et al., 2007). While corrected for, the high number of statistical comparisons increased the likelihood of a type I error and the small sample sizes increased the risk of a type II error.

In conclusion, one session of BT on a wobble board but not seated cycling or rest improved this specific balance skill in healthy younger adults by margins observed in long-term studies with no transfer to other balancing skills and changes in neural plasticity as assessed by intermuscular coherence and TMS. Future studies will examine in more detail and by other methods the neural mechanisms and time course underlying the remarkable level of specificity of balance skill learning.

Author statement

Conception and design of study: LB, CL, TH;
Acquisition of data: LB;
Analysis and/or interpretation of data: LB, TN, CL, TH;
Drafting the manuscript: LB, CL, TH;
Revising the manuscript critically for important intellectual content: LB, TN, CL, TH;
Approval of the version of the manuscript to be published: L. B. M. Bakker, T. Nandi, C. J. C. Lamoth, T Hortobágyi;

Trial registration

The study was registered as a clinical trial (NL6691)

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Declaration of Competing Interest

None

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