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Short communication

Characterizing the balance-dexterity task as a concurrent bipedal task to investigate trunk control during dynamic balance

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ABSTRACT

The purpose of the study was to characterize the Balance-Dexterity Task as a means to investigate a concurrent bipedal lower-extremity task and trunk control during dynamic balance. The task combines aspects of single-limb balance and the lower-extremity dexterity test by asking participants to stand on one limb while compressing an unstable spring with the contralateral limb to an individualized target force. Nineteen non-disabled participants completed the study, and performance measures for the demands of each limb – balance and dexterous force control – as well as kinematic and electromyographic measures of trunk control were collected. Given five practice trials, participants achieved compression forces ranging from 100 to 139 N (mean 121.2 ± 12.3 N), representing 14.4–23.0% of body weight (mean 18.7 ± 2.4%), which were then presented as target forces during test trials. Dexterous force control coefficient of variation and average magnitude of the center of pressure (COP) resultant velocity were associated such that greater variability in force control was accompanied by greater COP velocity (R = 0.598, p = 0.007). Trunk coupling, quantified as the coefficient of determination (R^2) of a frontal plane thorax and pelvis angle-angle plot, varied independently of any measure of balance or dexterous force control. The Balance-Dexterity Task is a continuous, dynamic balance task where bipedal coordination and trunk coupling can be concurrently observed and studied.

1. Introduction

The purpose of this study was to characterize the Balance-Dexterity Task and to evaluate its use in investigating a concurrent bipedal lower-extremity task and trunk control during dynamic balance (Fig. 1). The Balance-Dexterity Task was developed by combining single-limb balance (Schneiders et al., 2010) with the lower-extremity dexterity test (LED-test) (Lyle et al., 2013a). The traditional LED-test involves compression of an unstable spring while semi-seated on a bicycle seat with arms resting on a support surface and quantifies lower-limb dexterity since the compression force achieved is associated with performance on the cross-agility test (R^2 = 0.63) but not hip extensor strength (R^2 = 0.04), knee extensor strength (R^2 < 0.01), or knee flexor strength (R^2 = 0.02) (Lyle et al., 2013a, 2013b). In investigating athletic performance measures with a principal component analysis approach, dexterous force control and balance were found to quantify distinctly different aspects of performance (Lawrence et al., 2015). Adding this dexterous force control demand to the balance demands of single-limb stance can be viewed as a concurrent lower-extremity bipedal task and allows us to study motor control processes involved in successful task execution.

The characterization framework started by quantifying and evaluating performance measures for the demands of each limb – balance and dexterous force control – then continued by examining relationships between these measures. Next, trunk coordination was quantified and associations between trunk coordination and task performance measures were tested. Finally, factors potentially contributing to trunk coordination were explored including muscle activation data.

2. Methods

2.1. Participants and instrumentation

Nineteen non-disabled participants with no back or lower-extremity injury or pain in the last year and no conditions which would affect balance were recruited for the study with Institutional Review Board approval and informed consent (12 females, 7 males; 23.9 ± 3.3yrs; 169.1 ± 10.4 cm; 67.1 ± 10.8 kg; BMI 23.3

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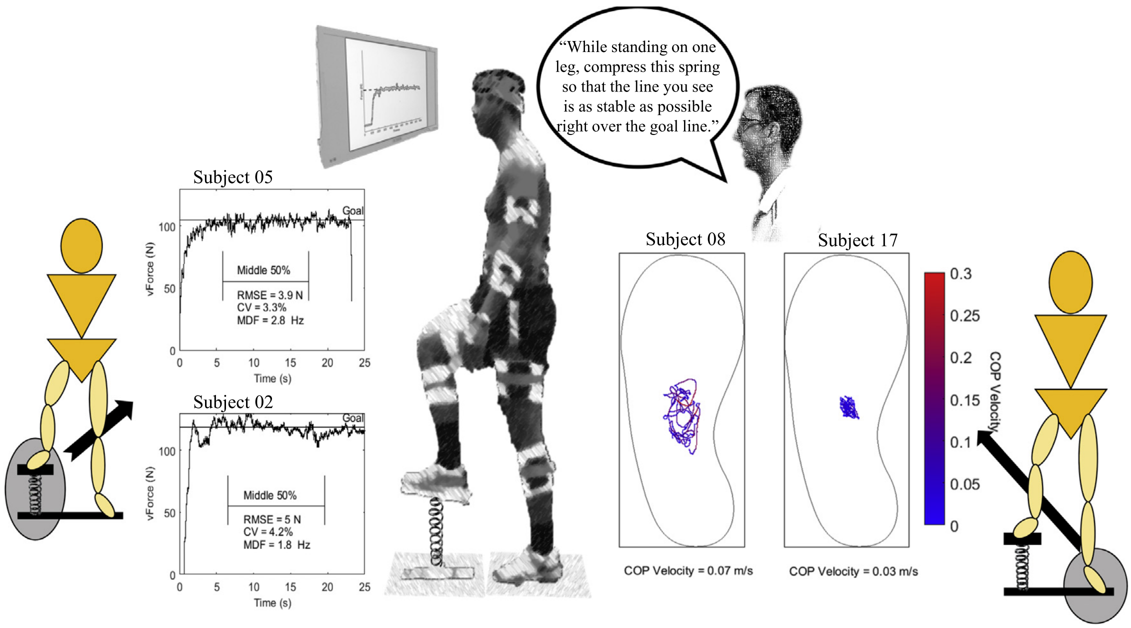

Fig. 1. The Balance-Dexterity Task with representative data showing examples of highest and lowest balance outcome measures including center of pressure (COP) measures (right) and dexterous vertical force (vForce) control outcome measures including root-mean-squared error (RMSE), coefficient of variation (CV) and median frequency (MDF) (left).

± 1.8). Participants were instrumented with a full-body retroreflective marker set as well as surface electromyography (EMG) of the external oblique (EO), rectus abdominis (RA), and gluteus maximus (GMax) and medius (GMed) and fine-wire EMG of the internal oblique (IO), lumbar multifidus (MF), and erector spinae (ES) at the level of L4 (Noraxon Wireless EMG; Scottsdale, AZ; 3000 Hz). Surface EMG were collected with bipolar silver/silver chloride electrodes with an interelectrode distance of 22 mm placed per guidelines from SENIAM (Hermens et al., 2006), and fine-wire EMG were collected with a pair of 50 µm nickel-chromium alloy wires insulated with nylon with distal 2 mm exposed and loaded into a 25-gauge hypodermic needle and sterilized. Insertions were done under ultrasound guidance, and protocols were adapted from Perotto et al. (2011). All muscles were instrumented on the side contralateral to the participant's preferred kicking limb, hereafter referred to as the stance side. Motion data were captured with an 11-camera Qualisys Oqus System (Gothenburg, Sweden; 250 Hz), and kinetic data were captured with Advanced Medical Technology Inc. force plates (Watertown, MA; 3000 Hz).

2.2. Procedures

Participants completed a 30 s trial of double-limb standing (preferred stance width) and three 30 s trials of single-limb standing on the stance side. Participants were introduced to the Balance-Dexterity Task, which used a custom device made by mounting polyvinyl chloride (PVC) adaptors to boards with a spring between them (spring characteristics: outside diameter 1.750 in [4.445 cm], inside diameter 1.336in [3.393 cm], free length 12.0in [30.48 cm], rate 28.0 lbs/in [49.0 N/cm], wire diameter 0.207 in [0.526 cm], and total coils 27.5; Compression Spring #805, Century Spring Corp., Commerce, CA). A similar instrumented device is available from Neuromuscular Dynamics, LLC (La Crescenta, CA). Participants were shown real-time feedback of the vertical force under the spring, and instructed: "While standing on one leg, compress this spring so that the line is first as high, then as stable as possible" (Fig. 1). Each trial lasted 20–25 s. After one familiarization trial and five practice trials, the mean of the middle 50% of the last three practice trials were used to calculate an individual's individualized target compression force. This value is different from the compression force achieved during the traditional LED-test because (1) the goal of the Balance-Dexterity Task, in contrast to the LED-test's goal of measuring maximum dexterous control ability, is to use dexterous force control to perturb balance; (2) for the LED-test, at least 20–25 attempts are required to produce a stable maximum indicating the compression described here is not maximal (Lyle et al., 2013a, 2013b); (3) in pilot testing it was found that, without the seat and arm rests, giving subjects more than five practice trials to achieve a stable maximum led to creative but confounding strategies sometimes including a deep squat with the stance leg or wedging the spring into a contorted shape. Note that the compression forces achieved cannot be directly compared to the Lyle et al. series of studies because spring stiffness parameters were different – 36.8 N/cm (Lyle et al., 2013a) and 49.0 N/cm in the current study. In addition, a direct comparison is not warranted because of the methodological differences between the LED-test where lower-extremity dexterity capability is quantified and the Balance-Dexterity Task where dexterous force control (not necessarily one's maximum capacity) is used to perturb standing balance. After practice, participants used a visual analog scale (VAS) to report how difficult the task was (0 Anchor: "Not difficult at all" and 10 Anchor: "Extremely difficult"), how confident they were they could complete the task successfully (0 Anchor: "Not confident at all" and 10 Anchor: "Extremely confident"), and how much attention the task required (0 Anchor: "No attention at all" and 10 Anchor: "All my attention").

Participants then completed five trials where a dotted line on the computer screen indicating their individualized target was shown with the instructions: "While standing on one leg, compress this spring so that the line is as stable as possible directly over the dotted goal line." Three trials were interspersed where the spring was replaced with a stable block of the same height, and the same target instructions were given. Five participants were brought in on a separate day for re-testing to assess test-retest reliability of outcomes measures. Results are reported as two-way random effects model ICC(2,5)s for absolute agreement, standard error of measurement (SEM) and minimal detectable change (MDC) (de Vet et al., 2006).

2.3. Data analysis

All trials were trimmed so the middle 50% of the task was analyzed. Kinematic and force plate data were low-pass filtered with cutoff frequencies of 12 Hz and 50 Hz, respectively. Surface and fine-wire EMG data were band-pass filtered between 20 and 500 Hz and 20 and 1000 Hz, respectively, all using a dual-pass 4th order Butterworth filter. EMG data were rectified and smoothed with a moving weighted average window of 500 ms. Signals were normalized to maximum voluntary isometric contractions (MVICs). Note one subject was excluded when reporting EMG data because of corrupted MVIC data.

Balance demands were quantified with center of pressure (COP) measures from the stance limb including the average magnitude of the COP resultant velocity, also characterized as average speed, and COP area measured with a 95% confidence ellipse, as well as the average magnitude of the center of mass (COM) resultant velocity, also characterized as average speed (Fig. 1, Eqs. (A) and (B)). COM was estimated with Visual3D software (C-Motion, Inc., Germantown, MD) Model Center of Gravity function using parameters from Dempster (1955) and Hanavan (1964). One participant was excluded from COM analyses due to an occluded marker on the arm that prevented full-body COM from being calculated. Dexterous force control was measured using the vertical force produced under the spring and quantified as root-mean-squared error (RMSE) from the reproducible, submaximal compression goal line; coefficient of variation (CV); and median frequency (MDF) of the detrended force (Fig. 1). Muscle activation data were averaged to acquire a mean activation amplitude for each trial. Muscle activation ratios were calculated in a frame-by-frame manner including deep-to-superficial ratios for the paraspinals (MF:ES) and abdominals (IO:EO) and co-contraction ratios for the deep trunk muscles (MF, IO) and superficial trunk muscles (ES, EO) (Eq. (C)).

$$Average\ Magnitude\ of\ COP\ Resultant\ Velocity = \frac{\sum_{i=1}^{N-1}\sqrt{\left(\frac{COP_{x,i+1}-COP_{x,i}}{\Delta t}\right)^2+\left(\frac{COP_{y,i+1}-COP_{y,i}}{\Delta t}\right)^2}}{N-1} \quad (A)$$

$$Average\ Magnitude\ of\ COM\ Resultant\ Velocity = \frac{\sum_{i=1}^{N-1}\sqrt{\left(\frac{COM_{x,i+1}-COM_{x,i}}{\Delta t}\right)^2+\left(\frac{COM_{y,i+1}-COM_{y,i}}{\Delta t}\right)^2+\left(\frac{COM_{z,i+1}-COM_{z,i}}{\Delta t}\right)^2}}{N-1} \quad (B)$$

$$Muscle\ Activation\ Ratio = \frac{\sum_{i=1}^{N}\frac{Muscle1_i}{Muscle2_i}}{N} \quad (C)$$

Symbols in these equations include x: medial-lateral position, y: anterior-posterior position, z: vertical position, Δt: change in time, N: total data points in one trial, and for deep-to-superficial ratios, Muscle 1 was MF for paraspinals and IO for abdominals, and for co-contraction ratios, the muscle whose mean amplitude was lower was Muscle 1.

Trunk control was quantified by tracking thorax and pelvis motion and acquiring EMG signals from trunk musculature. The pelvis segment was defined in a static trial with retroreflective markers on the iliac crests and greater trochanters and tracked during dynamic trials with markers on anterior superior iliac spines and L5-S1. The thorax segment was defined in a static trial with markers on the acromion processes and on the iliac crests and tracked during dynamic trials with markers on acromion processes, sternal notch, and T1. Rotation of these segments was quantified relative to the global (lab) coordinate system using a Cardan X-Y-Z sequence to give sagittal, frontal, and transverse plane rotations of these segments in space. Using an angle-angle plot of thorax and pelvis frontal plane rotation, a coefficient of determination (R^2) was calculated where a high R^2 would indicate highly coupled thorax and pelvis motion and a low R^2 would indicate more dissociated or independent motion of the thorax and pelvis. This metric has been used to distinguish participants with and without low back pain through frontal and transverse plane trunk coupling during gait (Crosbie et al., 2013; van den Hoorn et al., 2012). Also from this angle-angle plot, instantaneous coupling angles were calculated and the percentages of time spent in in-phase coupling and in anti-phase coupling were extracted, all per equations presented in Needham (Needham et al., 2014). In addition, more traditional range-of-motion metrics were acquired including trunk (thorax relative to pelvis), thorax, and pelvis angular excursions. Measures of dexterous force control (RMSE, CV, MDF) were compared between stable block and Balance-Dexterity Task conditions using paired t-tests for normally distributed data and Wilcoxon ranked sum tests for nonparametric data. Balance measures (COP, COM) were compared between double-limb stance, single-limb stance, the stable block condition, and the Balance-Dexterity Task using repeated measures analysis of variance (ANOVA) with Bonferroni-corrected post-hoc tests in cases of a significant main effect. Given the characterization nature of this study, relationships among trunk control variables and task performance measures were explored using descriptive statistics and bivariate Pearson and Spearman correlations. For all tests, α=0.05 (PASW Statistics, IBM Corp., Armonk, NY).

3. Results

3.1. Task performance

Given five practice trials, participants achieved dexterous control of the spring with compression forces ranging from 100 to 139 N (mean 121.2 ± 12.3 N), representing 14.4–23.0% of body weight (mean 18.7 ± 2.4%) (Fig. 2A). A handful of participants stumbled in the first trial before feeling how much the spring compresses and buckles, but after practice all participants were able to complete the task safely and successfully, defined as maintaining a positive spring compression force throughout the trial and without falling or movement of the stance foot on the force plate. There were no significant sex differences in any task performance measures. Reliability statistics for all primary outcome measures are summarized (Table 1).

Measures of performance of the two primary demands in the Balance-Dexterity Task – balance and dexterous force control – were evaluated. Dexterous force control was quantified using RMSE (6.54 ± 2.5 N), CV (3.60 ± 1.2%), and MDF (2.42 ± 0.5 Hz), and none of these measures were associated with each other ($0.346 \geq p \geq 0.129$) (Fig. 2B-D). Comparing these measures between the Balance-Dexterity Task and the stable block condition, where force control was required but without instability of a spring, revealed significant differences between conditions for all three measures – RMSE (Z = 170.0, p = 0.003), CV (t(18) = −2.884, p = 0.010), and MDF (t(18) = −9.328, p < 0.001).

Most measures of balance demands were associated (COM velocity and COP velocity: R = 0.457, p = 0.057; COM velocity and COP area: R = 0.505, p = 0.033; COP velocity and COP area: R = 0.476, p = 0.040). Using a repeated-measures ANOVA, a main effect of task was found for all three measures – COM velocity (F(3,15) = 24.5, p < 0.001), COP velocity (F(3,16) = 116.6, p < 0.001), and COP

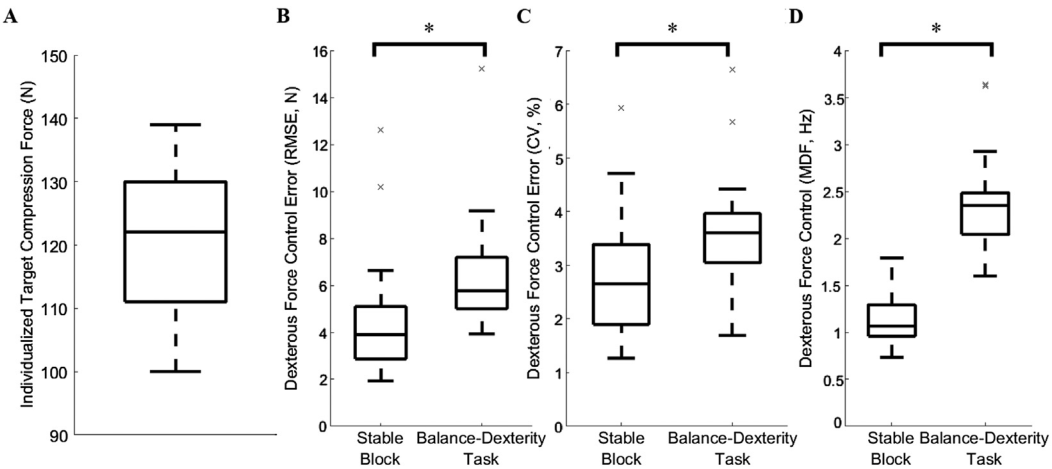

Fig. 2. (A) Submaximal, reproducible compression forces acquired from five practice trials and presented as a target during stable block and Balance-Dexterity Task trials. Measures of dexterous force control in the stable block and Balance-Dexterity Task conditions including root-mean-squared-error (B), coefficient of variation (C), and median frequency (D). *p < 0.05. Medians and quartiles are shown, with outliers defined as >1.5-times the interquartile range above the upper quartile marked as "x".

Table 1
Reliability statistics for primary outcome measures from a subset of five participants who underwent re-testing. ICC: intraclass correlation coefficient, SEM: standard error of measurement; MDC: minimal detectable change, BW: body weight, RMSE: root-mean-squared error, CV: coefficient of variation, COP: center of pressure.

Measure	ICC (2,5)	SEM	MDC
Compression Force (N)	0.875	5.19	14.39
Compression Force (%BW)	0.964	0.63	1.75
Dexterous Force RMSE (N)	0.916	0.59	1.64
Dexterous Force CV (%)	0.972	0.27	0.75
COP Velocity (cm/s)	0.999	0.05	0.14
Trunk Coupling (R^2)	0.684	0.04	0.11
In-Phase Coupling (%)	0.947	0.64	1.77

area ($F(3,16) = 33.4$, $p < 0.001$). Only for COP velocity, however, were all conditions significantly different from one another (Fig. 3).

The relationships between these two demands were investigated by testing associations between performance variables. Greater dexterous force CV was associated with greater COP velocity ($R = 0.598$, $p = 0.007$) (Fig. 4A). In addition, those who reported greater perceived task difficulty rated on a VAS also exhibited greater RMSE relative to the target compression force ($\rho = 0.640$, $p = 0.003$) (Fig. 4B).

3.2. Trunk coordination

The primary outcome measure for trunk kinematic coordination – trunk coupling R^2 (Fig. 5) – was compared to traditional segment excursion measures and measures derived from thorax and pelvis coupling angle. Trunk coupling R^2 was not associated with thorax ($R = 0.230$, $p = 0.354$), pelvis ($R = 0.330$, $p = 0.174$), or trunk ($R = 0.190$, $p = 0.440$) excursions (Fig. 6A-C). This measure was correlated with the percent of time spent in-phase coupling ($R = 0.850$, $p < 0.001$) and was negatively associated with the percent of time spent anti-phase coupling ($R = -0.800$, $p < 0.001$) (Fig. 6D). Trunk coupling R^2 was not significantly associated with any of the measures of task demands described above ($0.872 \geq p \geq 0.279$). Taken together, trunk coupling R^2 captures something different than simple measures of excursion, but similar to a phase analysis of segment coupling angle. Also, given the lack of significant associations between trunk coupling and any of the task performance measures, trunk coupling seems to be modulated independently of task performance.

Muscle activation levels during the stable block and Balance-Dexterity Task conditions normalized to MVICs are summarized (Fig. 7). The mean activation levels for all muscles were greater in the Balance-Dexterity Task compared to the stable block

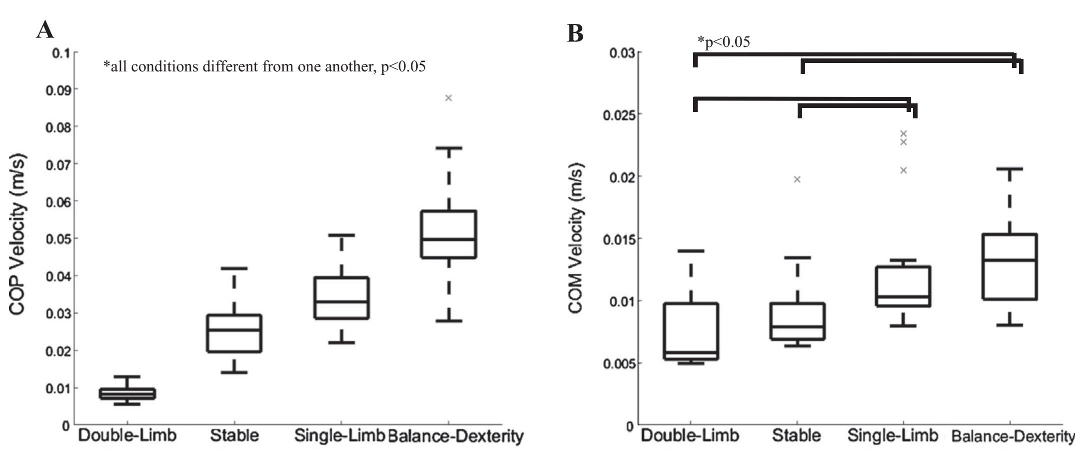

Fig. 3. (A) Center of pressure (COP) and (B) center of mass (COM) average resultant velocity in four conditions. Medians and quartiles are shown, with outliers defined as >1.5-times the interquartile range above the upper quartile marked as "x".

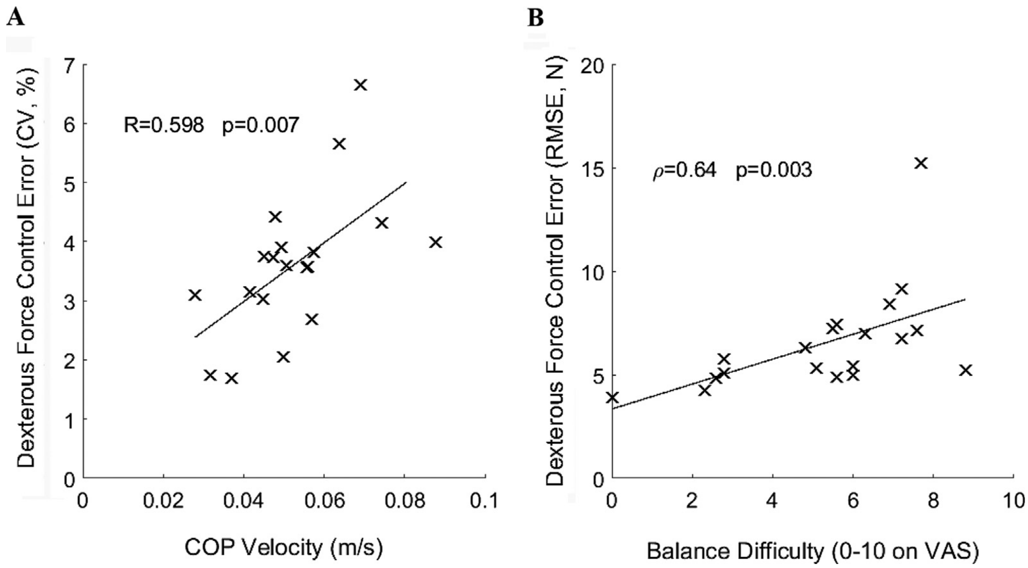

Fig. 4. (A) Association between coefficient of variation (CV) of dexterous force control center of pressure (COP) average resultant velocity. (B) Association between participants' self-reported assessment of task difficulty on a visual analog scale (VAS) and root-mean-squared-error (RMSE) of dexterous force relative to the reproducible, submaximal compression goal line.

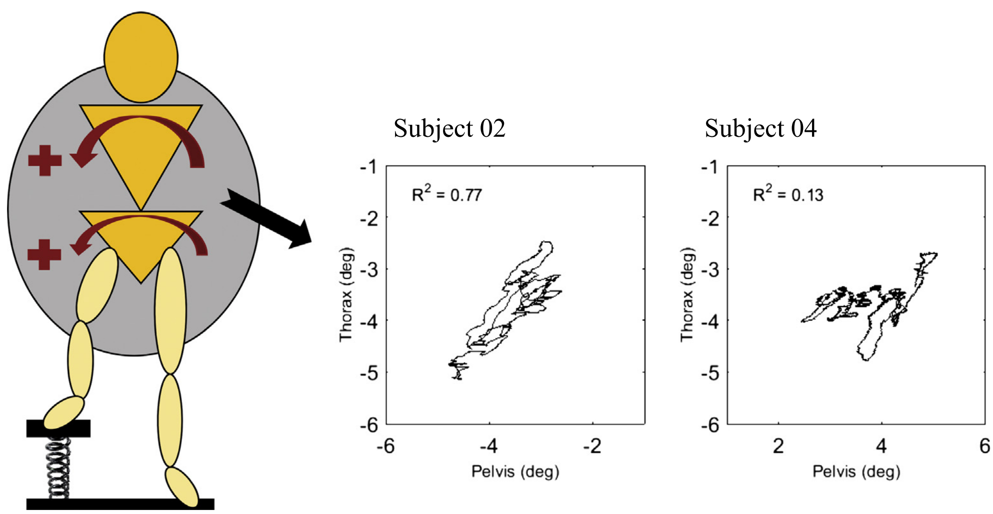

Fig. 5. Representative examples of high (left) and low (right) frontal plane trunk coupling quantified with a coefficient of determination for the thorax and pelvis angle-angle plot (R^2).

condition, but this only reached significance for the abdominals – IO ($p = 0.006$), EO ($p < 0.001$), RA ($p < 0.001$) – and GMed ($p < 0.001$). Trunk coupling R^2 was positively associated with MF: ES ratio ($R = 0.517$, $p = 0.028$) and negatively associated with GMed activation ($R = -0.503$, $p = 0.033$). The association with MF:ES ratio, however, was contingent on one outlier without which significance was not detected ($R = 0.428$, $p = 0.086$). When all EMG variables were tested in a stepwise multiple linear regression to predict trunk coupling R^2, without the MF:ES outlier, none entered the model.

4. Discussion

The Balance-Dexterity Task invokes larger COP velocities than double- and single-limb stance with eyes open and more variable dexterous force control compared to a stable block condition. The large COP velocity utilized in the task was likely necessary to control COM motion, (Winter, 1995) as we saw that COP velocity was higher in the Balance-Dexterity Task than in single-limb standing, but COM velocity was not different. Variability in dexterous force control was associated with variability in balance control, indicated by a positive correlation between dexterous force CV and stance limb COP velocity. In addition, dexterous force RMSE was associated with participants' perceived task difficulty. RMSE is most directly related to the visual error signal provided to the participants as they view vertical force in reference to their individualized target line, which could explain its association with a self-report measure of task difficulty.

Frontal plane trunk coupling was altered independently of task performance. Frontal plane motion was analyzed for two reasons – there was minimal sagittal and transverse plane motion during the task, and frontal plane motion is frequently the target of investigation and intervention in single-limb balance tasks (Asai et al., 2013; Winter, 1995; Winter et al., 1996). Trunk coupling R^2 was not associated with excursion measures but was correlated with

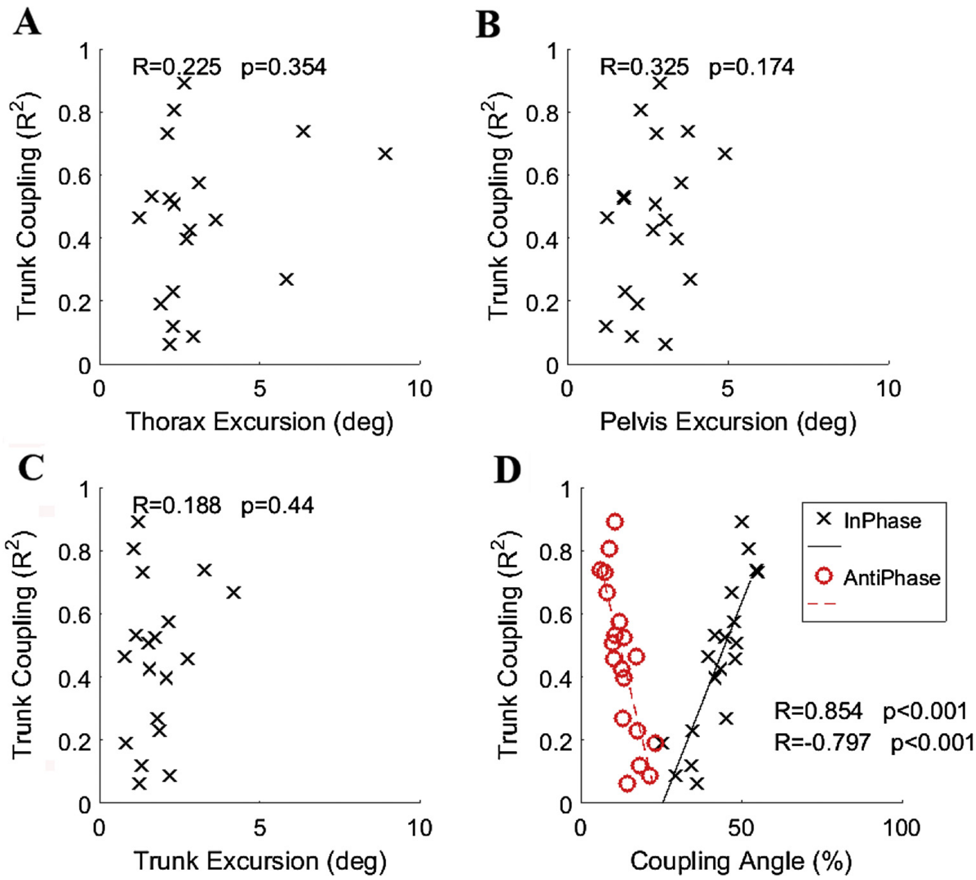

Fig. 6. Associations between trunk coupling quantified as the coefficient of determination of a thorax-pelvis frontal plane angle-angle plot (R^2) and (A–C) frontal plane segment excursions and (D) the percent of time spent in-phase and anti-phase coupling.

the percent of time in-phase and oppositely correlated with the percent of time anti-phase coupling, justifying use of this simpler measure to capture trunk coordination. Trunk coupling was not associated with any of the balance or dexterous force control demand measures. Trunk coordination, therefore, may prove to be a useful target in this task for observing or modifying movement patterns in various patient, aging, or athletic populations.

Muscle activation levels mostly ranged between 1%MVIC and 5%MVIC, and abdominal and GMed activations were greater in the Balance-Dexterity Task compared to the stable block condition. While muscle activation levels were low, they correspond well to those predicted in a seminal modeling study by Cholewicki and McGill reporting that at least 1–3%MVC activation of MF and ES was required to achieve sufficient lumbar spine stability in a neutral, standing posture (Cholewicki and McGill, 1996). Our findings support this model's predictions and suggest that the relative activation between these muscle groups may be important. In addition, the significant association between GMed activation and trunk coupling, though not maintained in the multiple regression model, highlight this muscle's important role in trunk control during conditions of single-limb stance (Penney et al., 2014). The paucity of associations in the multiple regression indicates that these non-disabled persons have redundant motor control processes available to control balance, dexterous force control, and trunk coupling. It is hypothesized that stronger associations will be present in patient populations with dysfunctional systems or in non-disabled controls when one or more sensory inputs are knocked out or perturbed. Future work should more rigorously test how MF:ES and GMed activation governs trunk coupling through a wider variety of tasks and analysis techniques.

In addition, future work utilizing the Balance-Dexterity Task can address limitations of the current study. This proof-of-concept and task characterization study was conducted on a convenience sample of nineteen healthy young adults. Future studies should utilize larger sample sizes to present more robust normative values as well as study more diverse populations including athletes, older adults, and patient groups. The focus on trunk control in the present study was motivated by our lab's research into trunk control impairments and persons suffering from low back pain. The small ranges of motion observed at the trunk (<5°) may be subject to measurement error, but this would be minimal given published marker accuracy of <2 mm and angle errors of <1° at most reference angles using a similar motion capture system (Vander Linden et al., 1992). In addition, the trunk coupling outcomes quantify relative motion of the thorax and pelvis, not absolute angle measures of these segments. An additional value of the task is it was not pain provoking at the lower-extremities or the trunk, yet provided insight into trunk control around an unstable neutral posture. The Balance-Dexterity Task is also well-suited for an investigation into hip- and ankle-strategy approaches to balance among others. Adding EMG signals from muscles around these joints and utilizing more intensive inverse dynamics analyses will answer questions about which strategies are used in different conditions and populations.

In conclusion, the Balance-Dexterity Task is a continuous dynamic balance task from which many fruitful measures of movement and motor control processes may be observed. These include measures of balance and dexterous force control, between which variability was positively associated, and trunk coupling, which was modulated independent of task performance.

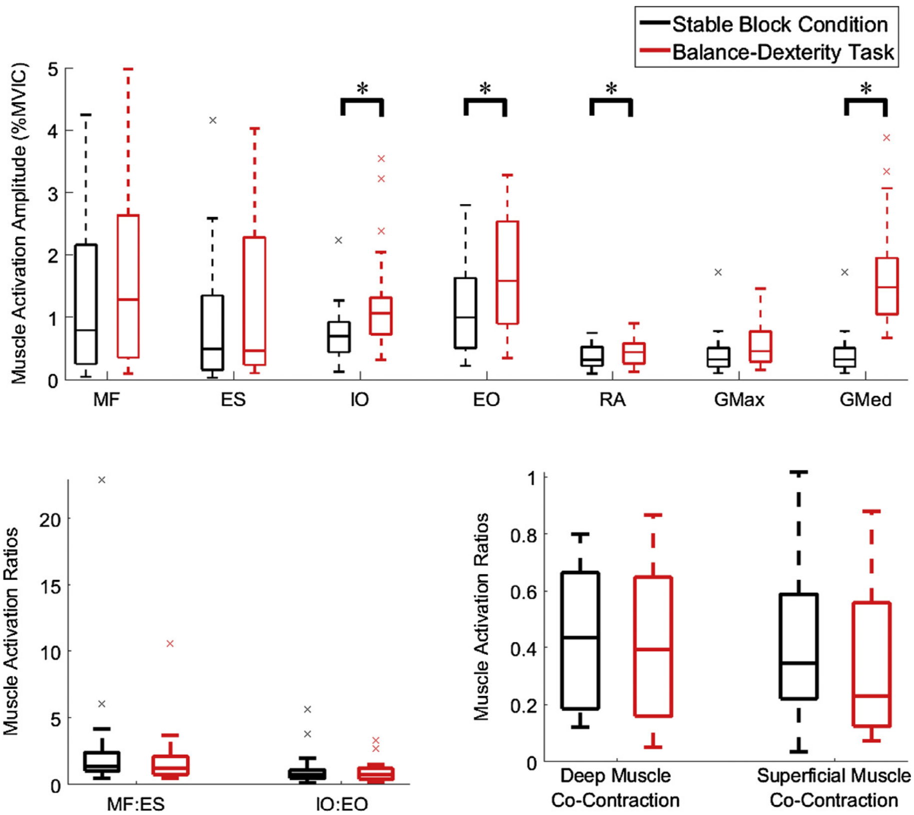

Fig. 7. (A) Muscle activation levels during the stable block and Balance-Dexterity Task conditions normalized to maximum voluntary isometric contractions (MVICs). MF: multifidus, ES: erector spinae, IO: internal oblique, EO: external oblique, RA: rectus abdominis, GMax: gluteus maximus, and GMed: gluteus medius. (B) Deep-to-superficial muscle activation ratios. (C) Co-contraction ratios for deep muscles (MF and IO) and superficial muscles (ES and EO). *$p < 0.05$ between conditions. Medians and quartiles are shown, with outliers defined as >1.5-times the interquartile range above the upper quartile marked as "x". One outlier with MF and ES values of 21%MVIC and 10%MVIC during the Balance-Dexterity Task, respectively, is not shown to preserve axis scale.

5. Conflict of interest statement

Authors have no financial or personal conflicts of interest to disclose.

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References

Asai, T., Doi, T., Hirata, S., Ando, H., 2013. Dual tasking affects lateral trunk control in healthy younger and older adults. Gait Posture 38, 830–836. https://doi.org/10.1016/j.gaitpost.2013.04.005.

Cholewicki, J., McGill, S.M., 1996. Mechanical stability of the in vivo lumbar spine. Clin. Biomech. 11, 1–15.

Crosbie, J., de Faria Negrão Filho, R., Nascimento, D.P., Ferreira, P., 2013. Coordination of spinal motion in the transverse and frontal planes during walking in people with and without recurrent low back pain. Spine (Phila. Pa. 1976) 38, E286–E292. https://doi.org/10.1097/BRS.0b013e318281de28.

de Vet, H.C., Terwee, C.B., Ostelo, R.W., Beckerman, H., Knol, D.L., Bouter, L.M., 2006. Minimal changes in health status questionnaires: distinction between minimally detectable change and minimally important change. Health Qual. Life Outcomes 4, 3–7. https://doi.org/10.1186/1477-7525-4-54.

Dempster, W.T., 1955. The anthropometry of body action. Ann. N.Y. Acad. Sci. 63, 559–585. https://doi.org/10.1111/j.1749-6632.1955.tb32112.x.

Hanavan, E.P., 1964. A mathematical model of the human body. Aerosp. Med. Res. Lab., 1–149

Hermens, H., Freriks, B., Merletti, R., Rau, G., Disselhorst-Klug, C., Stegeman, D., Hagg, G., 2006. The Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles Project (SENIAM) [WWW Document]. URL seniam.org.

Lawrence, E.L., Cesar, G.M., Bromfield, M.R., Peterson, R., Valero-Cuevas, F.J., Sigward, S.M., 2015. Strength, multijoint coordination, and sensorimotor processing are independent contributors to overall balance ability. Biomed. Res. Int. 2015. https://doi.org/10.1155/2015/561243.

Lyle, M.A., Valero-Cuevas, F.J., Gregor, R.J., Powers, C.M., 2013b. Lower extremity dexterity is associated with agility in adolescent soccer athletes. Scand. J. Med. Sci. Sports 25, 81–88. https://doi.org/10.1111/sms.12162.

Lyle, M.A., Valero-Cuevas, F.J., Gregor, R.J., Powers, C.M., 2013a. The lower extremity dexterity test as a measure of lower extremity dynamical capability. J. Biomech. 46, 998–1002. https://doi.org/10.1016/j.jbiomech.2012.11.058.

Needham, R., Naemi, R., Chockalingam, N., 2014. Quantifying lumbar-pelvis coordination during gait using a modified vector coding technique. J. Biomech. 47, 1020–1026. https://doi.org/10.1016/j.jbiomech.2013.12.032.

Penney, T., Ploughman, M., Austin, M.W., Behm, D.G., Byrne, J.M., 2014. Determining the activation of gluteus medius and the validity of the single leg stance test in chronic, nonspecific low back pain. Arch. Phys. Med. Rehabil. 95, 1969–1976. https://doi.org/10.1016/j.apmr.2014.06.009.

Perotto, A.O., Delagi, E.F., Iazzetti, J., Morrison, D., 2011. Anatomical Guide for the Electromyographer. Charles C Thomas Publisher, LTD., Springfield, Illinois.

Schneiders, A.G., Sullivan, S.J., Gray, A.R., Hammond-Tooke, G.D., McCrory, P.R., 2010. Normative values for three clinical measures of motor performance used in the neurological assessment of sports concussion. J. Sci. Med. Sport 13, 196–201. https://doi.org/10.1016/j.jsams.2009.05.004.

van den Hoorn, W., Bruijn, S.M., Meijer, O.G., Hodges, P.W., van Dieen, J.H., 2012. Mechanical coupling between transverse plane pelvis and thorax rotations during gait is higher in people with low back pain. J. Biomech. 45, 342–347. https://doi.org/10.1016/j.jbiomech.2011.10.024.

Vander Linden, D.W., Carlson, S., Robin, L., 1992. Reproducibility and accuracy of angle measurements obtained under static and dynamic conditions with the motion analysis video system. J. Am. Phys. Ther. Assoc. 72, 300–305.

Winter, D.A., 1995. Human balance and posture control during standing and walking. Gait Posture 3, 193–214.

Winter, D.A., Prince, F., Frank, J.S., Powell, C., Zabjek, K.F., 1996. Unified theory regarding A/P and M/L balance in quiet stance. J. Neurophysiol. 75, 2334–2343.