Postural Control Strategies in Standing With Handrail Support and Active Assistance From Robotic Upright Stand Trainer (RobUST)

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***Abstract*— In people with severe neuromotor deficits of trunk and lower extremities, regaining balance in standing is often performed in rehabilitation with manual assistance, rigid body supports or by the use of handrails. To investigate and further expand postural control training in standing, we developed a Robotic Upright Stand Trainer (RobUST). In this study, we used RobUST to deliver trunk perturbations while simultaneously providing postural assistive forces on the pelvis in 10 able-bodied adults. Posture control responses with 'pelvic support' was then compared to 'no support' and 'hand supported' standing, with and without assistance from RobUST. We characterize postural imbalance with kinematic displacements and center of pressure (COP) outcomes, such as amplitude and root mean square of the excursions of COP. Surface electromyography (sEMG) was also applied to investigate muscle control. We additionally investigated ground reaction and handrail forces during standing to analyze how postural strategies and muscle mechanisms with 'pelvic support' via RobUST would differ from standing with 'no support' and with the 'handrail support'. Our results show that during perturbations, pelvic assistive support decreased kinematic and COP excursions compared to standing with no support. The pelvic assistance from RobUST showed similar level of COP changes as the use of handrail support but without reducing muscle activity or ground reaction forces. As expected, the maximum level of postural stability was observed when participants used the handrail and received pelvic assistive forces. In conclusion, RobUST demonstrates potential as a training device since it enhances postural balance without significantly removing muscular control mechanisms that are of interest in re-training postural control strategies in standing.**

***Index Terms*— Assistive robotic forces, center of pressure, EMG, handrail, kinematics, perturbations.**

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I. INTRODUCTION

POSTURAL responses are task and environment dependent, and hence require different neuromuscular control strategies appropriate to the task [1], [2]. Posture is central to effectively and efficiently perform activities of daily life [3]. External perturbations are frequent during daily life, from sudden bus stops to accidental crowd pushing. Unexpected perturbations disrupt the person's equilibrium and elicit muscle responses to restore balance in standing position. These are called *in-place* postural strategies [4]. When muscle responses cannot overcome the postural imbalance, a person needs to perform compensatory actions such as taking a step or reaching for an external support to avoid falling. These are termed as *change-in-support* postural strategies [3], [5]. This situation is frequent in individuals with neuromotor disorders who continuously step or reach for support to recover balance even under small perturbations [6]. The goal of postural rehabilitation in neuromotor disorders is to practice and relearn *in-place* and *change-in-support* postural strategies to adapt postural control to diverse perturbations present in everyday surroundings.

We have developed a novel cable-driven system, Robotic Upright Stand Trainer (RobUST), that can apply forces on the trunk and the pelvis [7]. Previously, we focused on the technical features of RobUST. However in this study, we investigate and compare postural control strategies that able-bodied participants use in standing with RobUST versus a traditional rehabilitation method, such as the use of a handrail.

In rehab settings, grasping a handrail is a recurrent *change-in-support* postural strategy adopted by individuals with impaired trunk and lower extremity control to prevent falls during unexpected perturbations [3], [5], [8]. The main functional goal of RobUST is to promote independent postural standing while encouraging users to elicit *in-place* postural strategies via assistive forces. The rationale is that RobUST would allow users to fully experience sensorimotor cues during trunk perturbations and balance recovery. In addition, RobUST can potentially train critical control strategies to maintain postural balance without the need to step or use an external support, such as a handrail. In this study, we predict that our cumulative experimental findings with RobUST will have a major impact on improving current training approaches in postural standing.

In our previous study with RobUST [7], we analyzed upper body displacements when perturbations were applied at the trunk and pelvis, with and without assistive forces. We observed that trunk perturbations were accompanied by large-amplitude postural imbalances, as opposed to perturbations when applied on the pelvis [7]. This previous study, however, did not consider i) a two level assistive force, ii) the neuromuscular mechanisms by which users control postural stability, iii) how the postural responses differed when using RobUST compared to traditional balance training methods, and iv) the potential additive assistance effects when combining assistive forces with a static support such as handrails. These scenarios are particularly interesting for individuals with profound lack of control of trunk and lower extremities.

For this study, RobUST provides randomized perturbations at the level of the trunk. We characterize postural stability with output variables from two force plates and body translation from motion capture cameras. Muscle control mechanisms are characterized with surface electromyography (sEMG). Specifically, we aim at answering the following questions: I. How does RobUST's assistive force field on the pelvis improve postural control and modulate muscle activity (sEMG) during direction-specific trunk perturbations? II. Can the RobUST's pelvic assistive force field improve postural stability to the same level as when participants support themselves with a handrail? III. Do we observe an additive effect in postural control when people stand holding a handrail while receiving pelvic assistive forces via RobUST?

We hypothesize that the use of RobUST will provide significant stabilizing effects, such as reducing postural excursions and variability. This increase in postural stabilization would be associated with higher muscle sEMG activity with RobUST compared to the traditional use of a handrail. This work is an extension of study [9].

II. System Design

A. System Hardware

RobUST is a cable-driven robotic system that can actuate belts placed on a participant's trunk and pelvis, as shown in Fig. 1. RobUST contains 14 motors (Maxon Motor, Switzerland) but only eight were used in this study, each instrumented with an encoder and a load-cell (LSB302 Futek, California) in series with the motor. Four motors are used to actuate the four cables attached to each of the two belts. A motion capture system (Vicon Vero 2.2, Denver) provides real-time information on the position and orientation of the two belts to the robotic controller, programmed in LabVIEW (National Instrument v2017). In this study, RobUST provides perturbative forces at the level of the trunk and assitive forces at the level of the pelvis. The system contains two force plates that participants stand on, (Bertec Force Plate V1, Ohio) and a

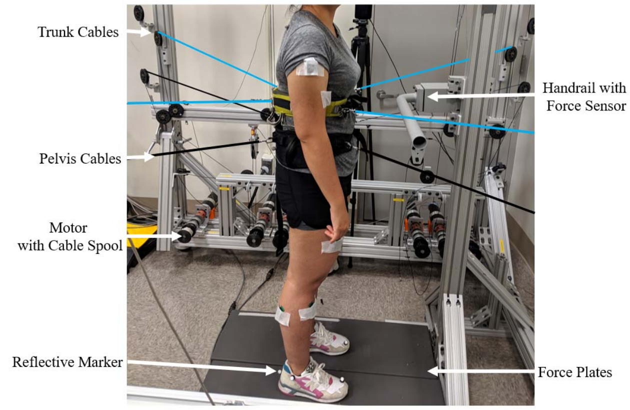

Fig. 1. A participant inside the RobUST system with labeled parts of the device. The trunk and pelvis cables are attached to the respective belts on the user.

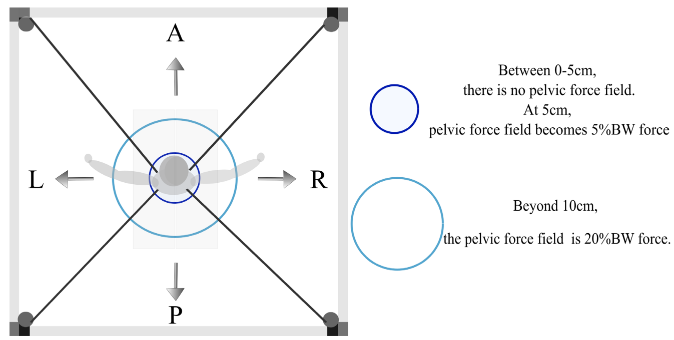

Fig. 2. Representation of the force field around a subject's pelvis and the directions of the trunk perturbations: Anterior (A), Right (R), Posterior (P), and Left (L).

height adjustable handrail (Bertec, Ohio) that measures forces applied onto the bar.

B. System Controller

A closed-loop PID controller modulates the tension to achieve the desired force. Further details on the PID controller are described in [7]. The force assistance in this study was configured differently from our previous research [7], [10]. However, the framework of how the tension planner distributes the force among the cables remains the same. The architecture of the assistive force controller resembles a virtual donut-shaped ring, Fig. 2, around the pelvis. In this study, the force controller has two main boundaries, the first is an inner radius of 5 cm and the second has an outer radius of 10 cm. Within the first boundary, the force controller creates a 'transparent' mode, i.e., nearly zero external forces are applied during motion. There is a force applied to keep the cables in tension. This transparent mode allows participants to move freely within their standing workspace. The motion capture system detects the Cartesian coordinates of the center of the belt and is used to determine whether the user is within the inner or outer boundary. Once the subject is outside the first boundary, an assistive force of 5% of body weight (BW) is applied. In this study, when the subject reaches the outer boundary, an assistive force of 20% BW is applied to keep the center

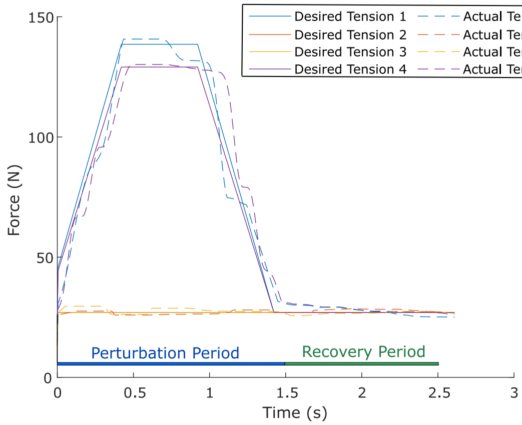

Fig. 3. A trunk perturbation force profile for a representative participant in the anterior direction. The perturbation duration is 1.5s followed by a recovery period of 1s.

of pelvis from going further. This 20%BW force was selected to ensure that participants would be safe and not fall. These forces can be adjusted to personalize RobUST to the motor and postural characteristics of the individual. The direction of the assistive force is towards the center, i.e., the neutral position of the subject when standing upright. This center position can be redefined through the graphical user interface of the RobUST system during the experiment, allowing re-centering of the geometrical center of the belt within force field boundaries if a new standing position is acquired. The RobUST force field had an average absolute error of 4.1N between the desired and actual forces in the x-direction and 4.0N in the y-direction.

III. Experiment Design

A. Procedure

The protocol had four experimental conditions and in each condition, RobUST delivered 8 randomized trunk perturbations. Two perturbations along each of the four directions were examined: anterior (A), posterior (P), Left (L), and Right (R). The trunk perturbation force was set to 20% of the participant's body weight. The force profile was trapezoid with 0.5 seconds of ramp up, 0.5 seconds of constant force equal to 20% body weight, and 0.5 seconds of ramp down to minimum tension, Fig. 3. The force magnitude and duration were chosen based on a previous study that showed kinematic displacements were adequate at these settings to study the response [7]. Participants were instructed to stand upright and recover balance without taking any step when RobUST delivered the trunk perturbation. However, the participants were advised to reach for the handlerail before taking a step, only if it was necessary. The position of the trunk belt was consistent across participants, placed on the lower ribs, Fig. 1. The four experimental conditions were as follows: No Support (NS), Handrail Support (HS), Pelvic Support (PS) from RobUST, and Handrail support along with Pelvic support from RobUST (HPS). In the NS, participants received trunk perturbations without handrail contact and without cables attached to the pelvic belt. In the HS condition, participants held onto a firm handrail placed at elbow's height but did not have cables attached to the pelvic belt. For the PS condition, RobUST provided assistive forces on the pelvis. In the HPS, participants held onto a handrail and received pelvic support from RobUST.

Kinematic position data of trunk, pelvis and feet were collected using nine Vicon cameras. Participants stood on two six-axis Bertec force plates, each foot on a different plate. Surface electromyography (sEMG) muscle activity was recorded by a 14 channel Delsys Trigno Wireless System (Delsys Incorporated, Massachusetts). Bilateral sEMG signals were registered from deltoids (DT), trapezius (TP), biceps (BC), erector spinae (ES), rectus femoris (RF), gastrocnemius (LG), and tibialis anterior (TA) and were recorded making up 14 channels. The abdominal muscles were not measured due to the trunk and pelvic belt locations.

B. Participants

Ten adults, 5 females and 5 males, with no musculoskeletal or neurological conditions participated in the experiment. The participants' characteristics were: height $170 \pm 12.3cm$, weight $70.7 \pm 18kg$, and age $26.8 \pm 4yrs$. Participants were informed about the research procedures and signed a written consent approved by the Institutional Review Board of Columbia University before participating.

C. Data Processing

Data was processed offline using MATLAB (Mathworks Inc.). Statistical analyses were carried with SPSS (version 27, IBM, 2020). Data recordings were synchronized with the perturbation onset; which was determined once cable tension increased above 5% from the desired onset. Data was segmented into two periods, Fig. 3, perturbation period, from onset to the end of the perturbation (1.5s), and the recovery period, which corresponded to a duration of 1s after the end of the perturbation.

Kinematic data was sampled at 100Hz. Reflective body markers were placed, three around each trunk and pelvic belt, one at each shoulder, (near the acromion), and three per foot (positioned about the toe, heel and ankle). Marker data was passed through a 4th order low-pass filter at 6Hz cut off frequency. The mean and amplitude (maximum-minimum) of the pelvis and upper trunk position was determined for each perturbation and recovery period. Local coordinate frames were calculated for each foot, pelvis and trunk based on ISB recommendations [11]. The feet and pelvic angles were calculated in reference to the initial ground frame at the start of perturbation, and the trunk angles are in reference to the local pelvic frame.

The force plate and handrail force data was recorded at 1000Hz and passed through a 4th order low-pass filter at 6Hz cut off frequency. We analyzed the root mean square (RMS) position of the ground COP, as defined in [12], [13], and the amplitude COP (maximum-minimum), as defined in [14]. The ground COP in the anterior-posterior (AP) and medio-lateral (ML) directions were exported from Vicon Nexus software (Vicon Vero 2.2). Center of pressure variables were normalized based on the participants BOS, ML COP by the width of BOS, and AP COP by the length of the BOS, similarly to

TABLE I
COP AND GRF VARIABLE MEANS AND STANDARD ERROR FOR EACH TEST CONDITION DURING TRUNK PERTURBATION PERIODS ($*p < 0.05$ IN EXP. CONDITION COMPARED TO NS, $*p < 0.05$ IN EXP. CONDITION COMPARED TO HS, $*p < 0.05$ IN EXP. CONDITION COMPARED TO PS, $*p < 0.05$ IN EXP. CONDITION COMPARED TO HPS)

Experimental Conditions	Variables	Anterior mean ± SE	Posterior mean ± SE	Dominant mean ± SE	Non-Dominant mean ± SE
No Support*	RMS COP	0.25 ± 0.02	0.12 ± 0.02	0.37 ± 0.08	0.28 ± 0.07
	COP Amplitude	0.46 ± 0.05	0.31 ± 0.03	0.70 ± 0.09	0.66 ± 0.07
	GRF (%bw)	1.02 ± 0.01	1.03 ± 0.01	1.03 ± 0.01	1.02 ± 0.01
Handrail Support*	RMS COP	0.13* ± 0.03	0.07** ± 0.018	0.34 ± 0.09	0.25 ± 0.09
	COP Amplitude	0.27* ± 0.06	0.17* ± 0.04	0.65* ± 0.10	0.66 ± 0.11
	GRF (%bw)	0.98**± 0.01	1.02 ± 0.01	1.00**± 0.01	1.00**± 0.01
Pelvic Support*	RMS COP	0.14** ± 0.03	0.07** ± 0.022	0.25* ± 0.08	0.20 ± 0.08
	COP Amplitude	0.26**± 0.05	0.22* ± 0.04	0.53* ± 0.07	0.52 ± 0.08
	GRF (%bw)	1.02** ± 0.01	1.03 ± 0.01	1.03** ± 0.01	1.02** ± 0.01
Handrail+Pelvic Support*	RMS COP	0.08** ± 0.02	0.01*** ± 0.009	0.19* ± 0.07	0.18 ± 0.06
	COP Amplitude	0.15** ± 0.03	0.08* ± 0.01	0.48 ± 0.06	0.48 ± 0.07
	GRF (%bw)	0.99**± 0.01	1.01 ± 0.01	1.00**± 0.01	0.99**± 0.01

Maki *et al.* [15]. Mean handrail forces and mean ground reaction forces were normalized by participant's weight. For the ML directions, the data was processed based on participants dominant and non-dominant hemibody side.

The average EMG value was removed from the entire signal. Then the EMG signal was band-pass filtered (60-500Hz), rectified, and low-pass filtered at 100Hz. The integrated EMG (iEMG) data was calculated and normalized by the integrated EMG activity obtained while participants stood still without postural disturbance (baseline EMG) [16]:

$$iEMG_{norm} = \frac{\int_0^t EMG - \int_0^t EMG_{baseline}}{\int_0^t EMG_{baseline}} \tag{1}$$

where t is dependent on the period, 1.5s for perturbation and 1s for recovery. For each participant, the baseline iEMG of each muscle group was registered during steady standing. The iEMG data was added for postural muscles (ES, RF, LG and TA) during anterior-posterior perturbations. Dominant and non-dominant muscles were examined during lateral perturbations.

D. Statistical Analysis

A total of 320 trials were examined. Data did not follow a normal distribution (Shapiro-Wilk test $= p < 0.05$) and was highly variable across trials, participants, and perturbation directions. We found that postural and muscle responses across conditions depended on both perturbation directions and experimental conditions. Generalized Estimating Equations (GEEs) account for within-subject correlation responses of many different distributions when data are clustered within subgroups [17]. Thus, GEEs were used to analyze events-in-trials following a repeated-measures procedure. In the analysis, participants and perturbation trials were used as clusters and experimental conditions and perturbation directions as within-subject variables. A linear model was selected. An exchangeable covariance structure was specified as correlation matrix based on the quasi-likelihood under independence criterion (QIC) goodness of fit coefficient, and because certain level of correlation between trails and within participants is expected. *Post-Hoc* testing with sequential Holm-Bonferroni method to correct multiple comparisons was applied if the statistical model was significant.

IV. RESULTS

A. Postural Stability

In the three external support conditions, the AP COP amplitude was significantly reduced in both anterior (Wald $\chi 2 = 44.79$, $p < 0.001$), and posterior perturbations (Wald $\chi 2 = 40.38$, $p < 0.001$), Fig. 4. The ML COP amplitude was also significantly different in the lateral directions, toward the dominant (Wald $\chi 2 = 15.357$, $p < 0.05$) and non-dominant hemibody (Wald $\chi 2 = 10.533$, $p < 0.05$). Table I summarizes means and standard errors for each support condition.

The use of a handrail, RobUST's PS, and the combination of both in HPS, assisted postural recovery by significantly decreasing the normalized RMS COP, Table I, during anterior (Wald $\chi 2 = 61.11$, $p < 0.001$) and posterior perturbations (Wald $\chi 2 = 31.42$, $p < 0.001$). In the ML directions, there was a significant effect towards the dominant hemibody (Wald $\chi 2 = 16.533$, $p < 0.001$) but not towards the non-dominant hemibody (Wald $\chi 2 = 6.87$, $p > 0.05$).

The pelvis amplitude displacement was significantly reduced in only the pelvic support conditions, PS and HPS, during anterior (Wald $\chi 2 = 12.63$, $p < 0.01$) and posterior perturbations (Wald $\chi 2 = 46.62$, $p < 0.001$), Fig. 5. When perturbations were towards the dominant hemibody (Wald $\chi 2 = p < 0.001$), ML pelvis amplitude decreased in PS and HPS, Fig. 6, and towards the non-dominant hemibody (Wald $\chi 2 = 16.53$, $p < 0.001$). The AP trunk amplitude displacement significantly decreased in the pelvic support conditions, PS and HPS, during anterior (Wald $\chi 2 = 30.27$, $p < 0.001$), and posterior perturbations (Wald $\chi 2 = 14.08$, $p < 0.005$). The ML trunk amplitude displacement significantly decreased in the PS condition, Fig. 6, during perturbations towards the dominant hemibody (Wald $\chi 2 = 17.82$, $p < 0.001$), and towards non-dominant hemibody (Wald $\chi 2 = 16.32$, $p < 0.001$).

B. Postural Control Mechanisms: Surface EMGs

The sEMG data of dominant postural muscles, trunk and lower extremities, show that participants executed

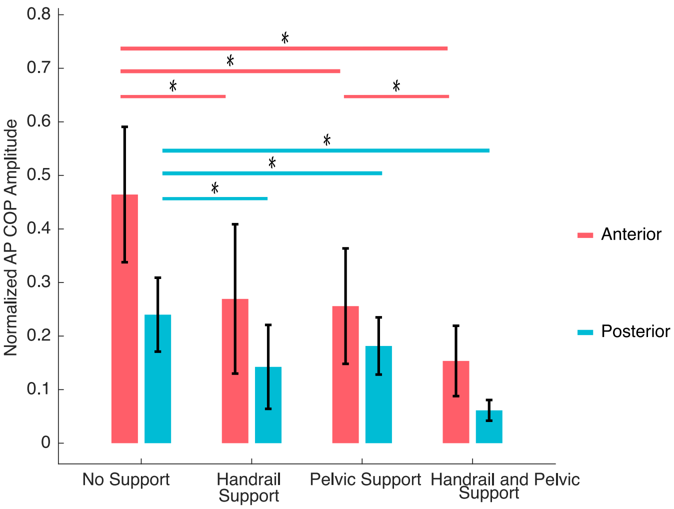

Fig. 4. Group averages of normalized COP amplitude in the AP direction across experimental conditions during anterior-posterior perturbations are shown. HPS offered the most stable postural control during anterior-posterior perturbations. $*p < 0.05$.

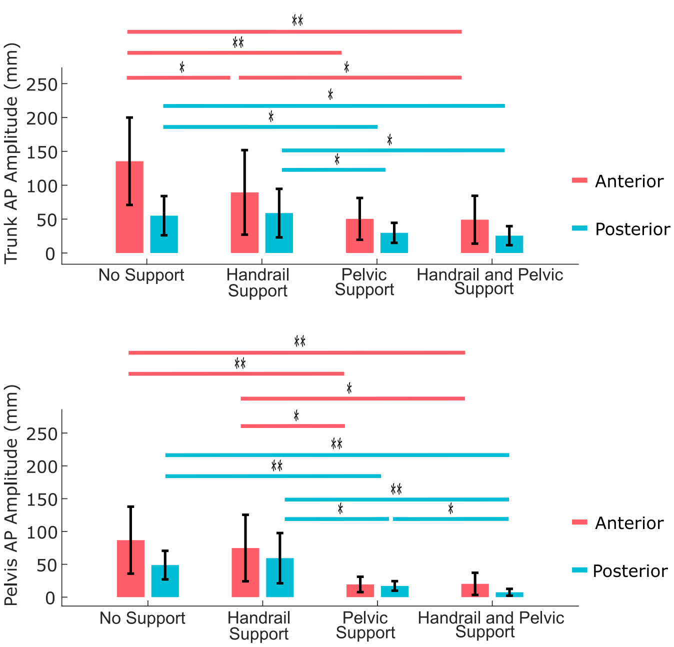

Fig. 5. Group averages of trunk and pelvis amplitude during anterior-posterior perturbations across the experimental conditions. PS and HPS provided the most decrease in amplitude. $*p < 0.05, **\,p < 0.005$.

specific-direction muscle responses during perturbations: anterior (ES: Wald $\chi 2$ z 19.39, $p < 0.001$; LG: Wald $\chi 2 = 34.93$, $p < 0.001$, RF: Wald $\chi 2 = 14.29$, $p < 0.005$, TA: Wald $\chi 2 = 10.28$, $p < 0.05$) and posterior (TA: Wald $\chi 2 = 29.63$, $p < 0.001$, RF: Wald $\chi 2 = 18.03$, $p < 0.001$). When RobUST delivered anterior perturbations, Fig. 7, participants showed significantly greater iEMG of ES muscles in NS (Mean $= 2.04 \pm 0.84$) than in HPS (Mean $= 0.15 \pm 0.17$, $p = 0.047$), HS (Mean $= -0.05 \pm 0.14$, $p = 0.014$). Participants also showed greater iEMG of ES muscles in PS(Mean $= 0.94 \pm 0.35$) than in HS ($p = 0.001$) and then in HPS ($p = 0.001$).

Similarly, the iEMG activity of LG was higher in NS (Mean $= 6.29 \pm 1.72$) than in HS (Mean $= 1.76 \pm 1.04$,

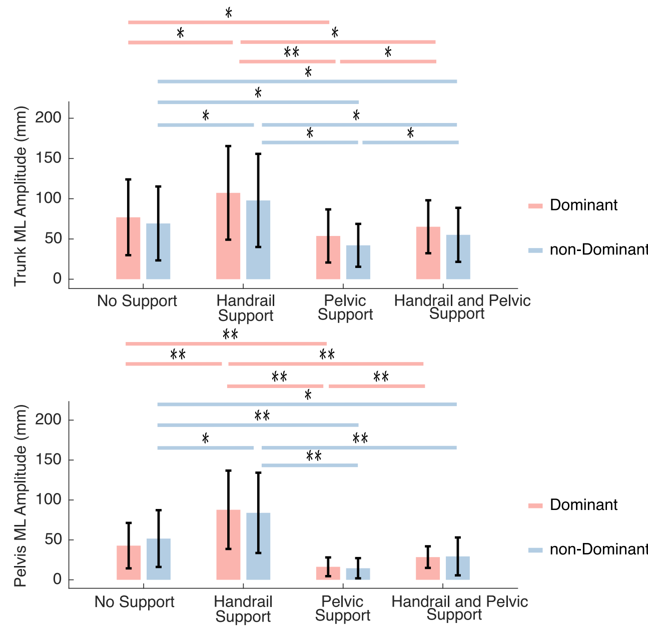

Fig. 6. Group averages of trunk and pelvis amplitude during medio-lateral perturbations across the experimental conditions. PS provided the most decrease in amplitude. HS significantly increased ML amplitude. $*p < 0.05, **\,p < 0.005$.

$p < 0.001$), PS (Mean $= 3.1 \pm 1.58$, $p < 0.001$) or HPS (Mean $= 0.41 \pm 0.28$, $p = 0.001$).

During posterior perturbations, the activation of TA had a significant role for participants to recover postural stability. Compared to NS (Mean $= 19.5 \pm 4.47$), the TA activity was significantly reduced when participants received PS (Mean $= 9.2 \pm 2.03$, $p < 0.005$) and HPS (Mean $= 0.91 \pm 0.67$, $p < 0.001$). The use of HS did not significantly reduce the muscle activity of TA compared to NS (Mean $= 6.42 \pm 4.4$, $p = 0.121$).

The EMG analysis of arm muscles showed high level of variability and did not reveal statistical differences among conditions in perturbation or recovery periods.

C. Force Output Responses

Mean ground reaction forces (GRF) were dependent on perturbation direction: anterior (Wald $\chi 2 = 95.84$, $p < 0.001$), posterior (TA: Wald $\chi 2 = 7.64$, p >0.05), non-dominant (Wald $\chi 2 = 33.83$, $p < 0.001$) and dominant (Wald $\chi 2 = 56.17$, $p < 0.001$). Both handrail conditions, HS and HPS, significantly reduced the GRF in all directions except posterior perturbation, Table I.

Mean handrail force magnitude was also dependent on perturbation direction, only during an anterior perturbation significance was found (Wald $\chi 2 = 42.57$, $p < 0.001$). The force magnitude in HPS was significantly less than HS Fig. 8. An interesting effect was identified in the recovery period from a posterior perturbation, the force magnitude in HPS was significantly less than HS, Fig. 8.

V. Discussion

In this study, we tested the dual action of RobUST to generate postural imbalance, via controlled perturbative forces

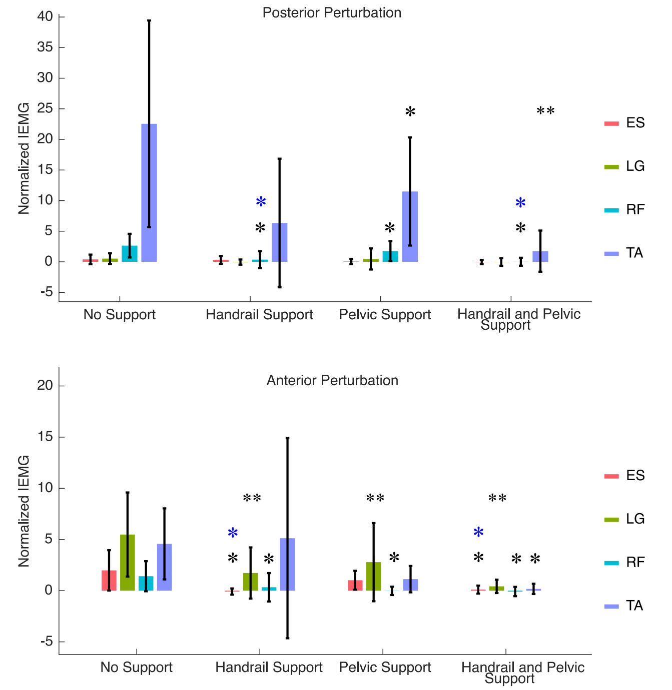

Fig. 7. The normalized iEMG averages across experimental conditions. ($*p < 0.05$, $** = p < 0.001$, in Exp. Condition Compared to NS, and $*p < 0.05$ in Exp. Condition Compared to PS). During posterior perturbations, the HS and HSP conditions significantly reduced RF compared to NS and PS. During anterior perturbations the HS and HPS significantly reduced ES compared to NS and PS. In PS the ES and TA were not significantly altered during the anterior perturbation.

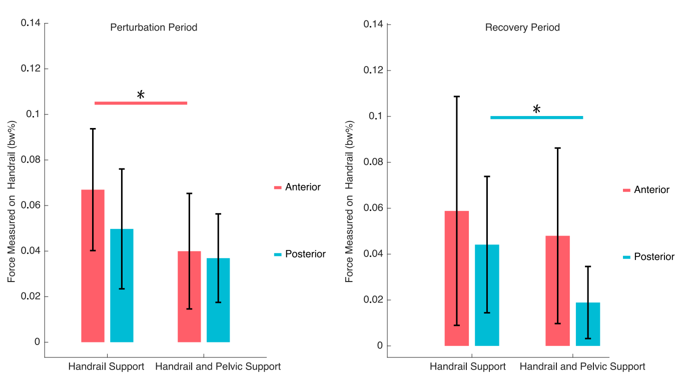

Fig. 8. Group averages of handrail force magnitude across perturbation and recovery period for anterior-posterior perturbations are shown. $*p < 0.05$.

on the trunk and stabilizing postural strategies through assistive forces on the pelvis. Then, we investigated how postural recovery with RobUST differed from standing with handrail support, which is a common postural paradigm and first step in rehab settings. Futhermore, we examined the potential combinatory effect of pelvic assistance from RobUST and handrail support to improve upright postrual balance.

The data analysis revealed that 1) the use of pelvic assistive force field via RobUST substantially improved postural stability in standing during trunk perturbations, mainly in the AP direction and towards the dominant hemibody; 2) RobUST provided similar level of COP postural stabilization as when participants supported their posture with the handrail; but interestingly, participants experienced less body displacements with pelvic assistive force support compared to postural support offered by a handrail 3) with pelvic support from RobUST, EMG activity of postural muscles resembled the muscles active during standing with no support 4) the combination of assistive force field and handrail support resulted in the greatest level of postural stability at the expense of substantial decrease in weight bearing forces between the feet and the ground.

A. The Stability Effects of Pelvic Support From RobUST

During postural imbalance, participants improved their stability in standing position while receiving assistive forces with RobUST compared to standing with no support. They experienced less degree of postural variability and decreased postural excursions with the assistive pelvic forces. Interestingly, participants showed similar COP stability outcomes while receiving pelvic support from RobUST as compared to when participants used a handrail for support. The use of the handrail decreased participant's natural weight bearing load, (i.e., ground reaction forces); whereas this effect was not observed with pelvic support mediated by RobUST. In standing, ground reaction forces have a relevant effect on postural control strategies. For instance, in slow low-amplitude horizontal ground displacements, healthy adults exert torques against the floor at the level of the ankles (i.e., ankle strategies) to recover sway [18]. From a rehabilitation standpoint, Olivetti *et al.* [19] has shown that weight bearing forces increase hip extensor strength in older people. Ground reaction forces also play an important role in some ankle orthotics to enhance lower limb alignment in the crouch position that children with cerebral palsy manifest [20]. Therefore, in future interventions, the pelvic assistive forces from RobUST would be beneficial in encouraging a natural ground reaction force profile for training standing postural control strategies in those with neuromotor disorders.

The postural muscles active with assistive forces at the pelvis were similar to the EMG of muscles active during standing without any support during anterior perturbations. However, this finding was not present for the other two handrail conditions, HS and HPS. The use of the handrail demanded other postural control adjustments that differed from *in-place* postural control strategies at the level of the ankle, knee, hip or a combination of these [21]. Hall *et al.* [22] showed the flexibility of the postural control system able bodied individuals have to adjust to different external rigid supports and body configurations. We observed direction specific postural adjustments, TA and RF in posterior perturbations, and a combination of LG, ES and TA in anterior perturbations. During pelvic support from RobUST these muscles were still active, but with reduced EMG activity. This data may justify the potential applicability of RobUST to train postural balance via systematic perturbations in individuals with standing control impairments without the need of adopting a *change-in-support* postural strategy. RobUST may be used in future training paradigms to encourage *in-place* postural strategies while performing reaching or hand-arm related tasks.

While stepping or reaching for support are essential to maintain balance and prevent falls, the postural strategy and neuromuscular demands to recover balance are different from those required for *in-place* postural strategies, which are highly impaired in individuals with neuromotor disorders [6].

Researchers have previously shown in individuals, with and without nervous system lesions, direction-specific postural adjustments during reactive postural control in standing via moving platforms [18], [23], [24]. In our study, the perturbative forces acting on the torso were associated with direction-specific muscle activity of trunk and lower limb muscles so that participants could be displaying a combination of hip and ankle muscle strategies [25]. This reactive postural muscle strategy is different from studies using surface perturbations. A study [26], demonstrated the presence of complex postural responses that combined muscle responses from the typical ankle strategy (disto-proximal recruitment of distal leg muscles) and hip strategy (proximo-distal recruitment of trunk and thigh muscles). Therefore, the origin of the perturbation, how the perturbation is delivered and the perturbative force profile, has an impact on the type of postural control responses. Although different perturbations can induce different muscle control responses, they could have similar secondary outcomes. Mansfield *et al.* [8], describe there is some postural transfer effect in platform perturbation training to cable perturbations about the center of mass, this effect was a decrease in handrail contact time. Further research is needed to determine other outcomes that can be transferred from body based perturbations to ground based perturbation training.

When pelvic support from RobUST was provided, body translations were reduced in the ML directions. Meanwhile, COP stability outcomes were dependent on the perturbation direction, and only significantly reduced when directed towards the dominant hemibody direction. A possible explanation may be the greater mediolateral BOS and that individuals have a higher force threshold to lateral perturbations than to anterior-posterior [27]. In our study, the force magnitude was set to 20%BW of the participant in each of the four directions. This force intensity may have not been strong enough to induce a significant level of instability in the ML directions. This interpretation is also supported by our EMG results. Dominant or non-dominant ES were not significantly reduced or augmented with the use of force field or handrail during lateral perturbations. This may also be partly explained because hip abductors, i.e. gluteus medius, are the primary muscles to control body COM within the frontal plane and EMG of such muscles were not registered in our study [28].

The effects in the different stability supports, between pelvic and handrail, are depicted by the responses in body translations as well. Assistive forces at the pelvis significantly reduced participant's trunk and pelvic displacements during the perturbations in all directions. However, in the handrail support, trunk and pelvis did not show a decrease in body translations in the ML perturbation directions. These results may indicate that trunk perturbation forces were high enough to displace the pelvic and trunk segments but low to cause destabilizing effect associated with a significant modulation of sEMG responses.

B. The Additive Support Effect of HPS

The application of a handrail and pelvic support reduced postural instability, i.e., COP and pelvic and trunk excursions. It also reduced the level of postural muscle activity required to control balance during anterior-posterior perturbations. Compared to standing without support, participants substantially reduced postural excursions and demonstrated a highly stable postural stance during perturbations.

Our data showed that handrail conditions, with or without the pelvic assistive support, were accompanied by a decrease in muscle activity and ground reaction forces. In other words, the excess of external fixed support can improve postural sway in standing at the expense of suppressing the active role of the neuromuscular system and weight bearing force distribution to control posture. While this strategy may not be the most efficient therapeutic strategy to retrain postural control in individuals with mild-to-moderate balance control disorders; the additional use of an external handrail in RobUST sheds light on its potential use to promote postural standing in individuals with severe loss of neuromotor postural control, such as in spinal cord injuries (SCI). However, as we observed in our analysis, the application of assistive pelvic support from RobUST may help modulate the amount of force exerted by the hands in postural training with external handrail assistance. We found that the combination of hand and pelvic support via RobUST significantly reduced the handrail force magnitude exerted by participants during anterior perturbations and during the recovery stage from posterior perturbations, Fig. 8. Patients with ambulatory SCI may acquire standing, however, they suffer from static and dynamic postural deficits that increase their risk to fall or transition from sitting to standing [29]. Hence, they would need to overcompensate with their arms and hands because of the severe lack of postural loss. The combination of the two support systems can alleviate the hand and weight bearing force distributions required to obtain stability to trunk perturbations. This provides a method to structure a postural rehabilitation paradigm based on the individual's assistance needs.

C. Study Limitations

A methodological limitation was the inability to register EMG activity of abdominal muscles due to the location of the belts and cables around the area. The data from these muscles could have expanded our understanding of trunk control during posterior perturbations. We must also acknowledge the possibility that RobUST in transparent mode may have provided a certain level of postural stability in stance to the able bodied participants. This effect would come from the minimum tensions required to prevent the cables from slacking. Similar to how light touch improves balance [30], the presence of the low tensioned cables may be enough to provide some stability.

Another study limitation was the application of 20%BW across all participants in the AP and ML directions. Other researchers like Komisar *et al.* [31] determine an individual's perturbation threshold per direction by delivering forces until the individual steps or falls. Performing the study at the threshold magnitude could have provided more insight into participant's reaction to trunk perturbation, especially in the ML directions.

VI. Conclusion

Pelvic assistive forces from RobUST allowed participants to have similar postural COP outcomes as holding a handrail, but without inhibiting as significantly the EMG activity of the postural muscles nor decreasing the ground reaction force distribution. The pelvic support via RobUST also decreased postural excursions for all perturbation directions. Additionally, the combination of the handrail and pelvic support provided the most postural stability and reduced the force magnitude exerted by the hands during anterior and posterior perturbations. RobUST could be systematically used in the training of individuals with neuromotor disorders to progressively build complex automatic postural reactions [32] in upright standing.

The findings of the present study show the promise of RobUST for future training paradigms that target specific muscle strategies for *in-place* and *change-in-support* postural strategies. RobUST may provide a new evaluation and training paradigm for postural balance training of neurologically impaired individuals who require external assistance and aids during postural stance.

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