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# Effect of Motor-Assisted Elliptical Training Speed and Body Weight Support on Center of Pressure Movement Variability

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## Abstract

*Background:* A motor-assisted elliptical trainer is being used clinically to help individuals with physical disabilities regain and/or retain walking ability and cardiorespiratory fitness. Unknown is how the device's training parameters can be used to optimize movement variability and regularity. This study examined the effect of motor-assisted elliptical training speed as well as body weight support (BWS) on center of pressure (CoP) movement variability and regularity during training. *Methods:* CoP was recorded using in-shoe pressure insoles as participants motor-assisted elliptical trained at three speeds (20, 40, and 60 cycles per minute) each performed at four BWS levels (0%, 20%, 40%, and 60%). Separate two-way repeated measures ANOVAs (3 × 4) evaluated impact of training speed and BWS on linear variability (standard deviation) and nonlinear regularity (sample entropy) of CoP excursion (anterior-posterior, medial-lateral) for 10 dominant limb strides. *Findings:* Training speed and BWS did not significantly affect the linear variability of CoP in the anterior-posterior or medial-lateral directions. However, sample entropy in both directions revealed the main effect of training speed ($p < 0.0001$), and a main effect of BWS was observed in the medial-lateral direction ($p = 0.004$). Faster training speeds and greater levels of BWS resulted in more irregular CoP patterns. *Interpretation:* The finding that speed and BWS can be used to manipulate CoP movement variability when using a motor-assisted elliptical has significant clinical implications for promoting/restoring walking capacity. Further research is required to determine the impact of motor-assisted elliptical speed and BWS manipulations on functional recovery of walking in individuals who have experienced a neurologic injury or illness.

**Keywords:** gait, center of pressure, physical rehabilitation, variability, regularity

## 1. Introduction

Improving walking is a primary goal for many patients and their therapists following neurologic injuries and illnesses. Therapists employ a variety of approaches to address walking limitations, including practicing gait skills in real or simulated environments [1] as well as engaging in intensive repetition of stepping on treadmills and with robotic assistance [2].

Clinical research increasingly emphasizes the importance of integrating movement variability into gait training so patients explore and learn different movement strategies beneficial for navigating real world environments [3,4]. While movement variability is relatively easy to integrate into overground gait activities in naturalistic environments, it can be more challenging to assimilate during treadmill and robotic stepping given the activities' repetitive and constrained nature [5]. Purposeful modification of training parameters (e.g., speed, external assistance level) might provide a means for manipulating task complexity and promoting a wider repertoire of movement strategies so variability across repetitions is neither too constrained (i.e., lacking any variability across repetitions) nor too variable (i.e., overly disorganized and random) [6].

Various analytics characterize movement variability during walking. Linear variables (e.g., standard deviation of anterior-posterior or medial-lateral CoP displacement) provide insights into amplitude and dispersion of kinematic and kinetic data, yet fail to characterize a movement's temporally evolving nature (i.e., impact of preceding cycle's movements on subsequent cycles' movements) [4]. In contrast, non-linear measures reveal temporal variations in movement patterns across a series of movement strides/cycles, but fail to describe amplitude (e.g., dispersion from mean). In the field of non-linear dynamics, entropy is indicative of the system's regularity [6]. More regular systems display entropy values closer to zero [7]. Multiple non-linear tools exist for exploring data series regularity. For example, detrended fluctuation analysis and Lyapunov exponent require relatively long time series (i.e., many data points) to generate reliable estimates [8], while data length does not have as great an impact when calculating SampEn (SE) [9]. Combined, linear and non-linear measures offer a more comprehensive description of movement variability.

Neuromuscular control of gait has been characterized by analyzing center of pressure (CoP) progression across the plantar foot surface [10]. CoP advances sequentially from heel to forefoot across successive stance periods when nonpathological gait is performed at a comfortable speed [11]. During treadmill walking, linear and nonlinear measures of CoP progression have demonstrated sensitivity to both walking speed and amplitude of body weight support (BWS) [12–14], thus providing clinicians with two manipulatable training variables for altering movement variability during treadmill gait.

The ICARE, a motor-assisted elliptical (Fig. 1), is used to promote intensive locomotor practice in rehabilitation, medical fitness, and home settings [15]. Patients train at speeds


**Figure 1.** Individual using ICARE training system.

up to 65 cycles per minute (CPM) with a motor's assistance for pedal advancement [16]. Each ICARE cycle mimics a gait cycle's kinematic and muscle demands [17] and faster motor-assisted training speeds increase muscle demands in key muscle required for stability and shock absorption [18]. An integrated external BWS system helps individuals with weakness and/or balance deficits maintain upright posture while safely exploring stability limits. As strength and endurance improve, faster training speeds and lower levels of BWS are used to promote continued challenge. The ICARE uses an endpoint control strategy (i.e., user is only in physical contact with machine through pedals and handles) vs. other robotic devices which often use leg orthoses to provide more rigid limb movement control. ICARE's motor provides a guiding force to cycle each pedal in an ovaloid path while the user activates muscles in each limb to stabilize the body in sagittal (e.g., vastus lateralis) and frontal planes (e.g., gluteus medius) over the pedals [18]. Because physical guidance is provided only under the foot's plantar aspect, participants can vary kinematic and muscle activation patterns across strides. Plantar pressure recorded from a subject training at 40 CPM demonstrated posterior to anterior CoP progression with each cycle [19], however no other speed or BWS conditions were reported to help elucidate the influence of these parameters on movement variability. Understanding speed's and BWS's influence on anterior-posterior and medial-lateral COP motion could guide understanding of requisite muscle activation patterns and strategies the central nervous system uses to maintain stability over a dynamically shifting base of support.

This pilot study involved a secondary analysis of previously recorded data to help elucidate the influence of ICARE's speed and BWS training parameters on movement variability and to facilitate design of future prospective studies. We hypothesized that ICARE pedals' fixed ovaloid path would constrain *linear* measures of anterior-posterior and medial-lateral CoP variability (standard deviation of length of each trajectory across strides) across training speeds and levels of BWS. However, we hypothesized that similar to treadmill walking [12–14], faster motor-assisted ICARE speeds (which lead to an increased lower extremity muscle demands) as well as increased BWS would reduce participants' CoP control and result in greater *nonlinear* irregularity of anterior-posterior and medial-lateral CoP motion (SE). Understanding the effect of speed and BWS on movement variability is expected to provide clinicians with essential data to guide selection of training parameters to address not only cardiorespiratory training goals but also movement variability goals to promote adaptive walking skills.

## 2. Methods

### *2.1. Participants*

As this is a secondary analysis of a data set, the participants and methods have been reported previously [18]. In brief, five males and five females (26.8 ± 3.8 years; 174.2 ± 8.5 cm; 80.4 ± 13.2 kg) free from musculoskeletal, neurological, and cardiovascular disorders that might affect walking were recruited from staff at Madonna Rehabilitation Hospitals. After obtaining consent, demographic data were collected. All procedures were reviewed and approved by Madonna's institutional review board.

### *2.2. Apparatus and procedure*

The PedarX insole measuring system (Novel Electronics, St. Paul, Minnesota, USA) evaluated plantar pressures. Each insole, consisting of 99 capacitive sensors, was calibrated (Trublu® calibration device) according to the manufacturer's guidelines.

Testing was performed on a single ICARE (SportsArt Fitness, Woodinville, Washington, USA) [16,20]. A BWS harness (Maine Antigravity Systems, Inc., Portland, Maine), fitted to each participant and attached to an overhead BWS system (PnueWeight Unweighting System, Pnuemex, Sandpoint, Idaho, USA), provided predetermined external BWS.

The experimental protocol consisted of two sessions. Participants were encouraged to wear comfortable exercise clothing and their own self-selected athletic footwear. During session one, basic anthropometrics were recorded, and participants kicked a ball to determine lower extremity dominance. Participants tried ICARE, and the self-selected ICARE stride length was determined.

During session two, a pair of PedarX insoles (selected from an inventory of 14 sizes based on closest match to a shoe insole's length and width) was placed between the participant's shoe insoles and socked feet. Participants then lifted each foot sequentially for insole zeroing according to the manufacturer's guidelines. Next, each participant performed 12 motor-assisted ICARE conditions at their self-selected stride length (two minutes each). Conditions consisted of four BWS levels (0%, 20%, 40%, 60%), each performed at three motor-assisted elliptical training speeds (20, 40, 60 CPM). Participants were instructed to let the motor (that controlled speed of the elliptical pedals) guide their legs at each training speed [10]. Actual training speed and set training speed were monitored on the control console to ensure participants did not override the motor's assistance and cycle faster. To reduce the systematic influence of fatigue on performance across participants, activity order was randomized using a computer program (MATLAB version 7.2.0.232, The MathWorks, Inc., Natick, Massachusetts, USA). Two levels of randomization were applied. The first was for percentage of BWS. Then, within each BWS level, training speed was randomized. Between conditions, participants rested at least one minute.

### *2.3. Data analysis*

All participants were right lower extremity dominant based on kick test performance. Maximal vertical force delineated pedal strides. For consistency, ten consecutive dominant limb strides recorded with PedarX (100 Hz) during the final minute of each condition were trimmed and analyzed to calculate CoP variability and regularity (frontal and sagittal planes).

PedarX software recorded in-shoe CoP location as X-Y coordinates (origin defined as most medial and posterior insole points). Anterior-posterior and medial-lateral CoP data were directly exported from the PedarX software. Standard deviation and SE were calculated. Standard deviation, indicating the time series' linear variability, was evaluated by measuring how closely each anterior-posterior and medial-lateral CoP time series centered around the mean. SE, defined as the probability of future and previous CoP movement patterns being consistent, was quantified using methods described by Richman et al. [21]. SE of ten consecutive strides of CoP was quantified to determine the regularity of CoP across strides. After examining relative consistency with $r = 0.15, 0.2, 0.25, 0.3$ (r indicates similarity criterion), and the parameter values of $m = 2,3$ (m is the length of the data segment being compared), $m = 2$ and $r = 0.2$*standard deviation were selected for calculation of SE consistent with previous recommendations indicating appropriate parameters for CoP at 125 Hz are $m = 2–6$ and $r = 0.2$*STD [13]. SE of all participants was averaged and reported for each condition. SE values closer to zero indicate highly regular motion patterns, while greater values represent more irregularity. All calculations were performed using custom Matlab programs.

### *2.4. Statistics*

Data were inspected for spikes or outliers greater than three standard deviations. Following screening for normality, two-way repeated measure ANOVAs ($3 \times 4$) compared the effect of training speed (20, 40, 60 CPM) and BWS (0%, 20%, 40%, 60%) as well as their interaction on each dependent variable (anterior-posterior and medial-lateral standard deviations and anterior-posterior and medial-lateral CoP time series SEs) across conditions. Greenhouse-Geisser corrections were applied when sphericity was lacking. Statistical analyses were carried out using SPSS software version 24 (IBM Corp., Armonk, New York, USA). Statistical significance was set at alpha = 0.05. *A posteriori* power analyses of this pilot study's data were performed using G power to guide understanding of sample sizes required to sufficiently power (80%) future studies examining similar variables.

## 3. Results

Ten individuals participated. Two participants' data were removed from the repeated measure analysis because of incomplete data acquisition of one condition for each participant. All other data are reported from the eight participants. While ICARE stride length was adjustable, each participant maintained their same self-selected stride length across conditions as evidenced by data displayed on the console. The mean preferred stride length across participants was 1.16 m (SD = 0.18 m; range = 0.96–1.37). Figure 2 displays exemplar CoP excursion data from one participant training at the lowest and highest motor-assisted elliptical speeds and BWS levels.

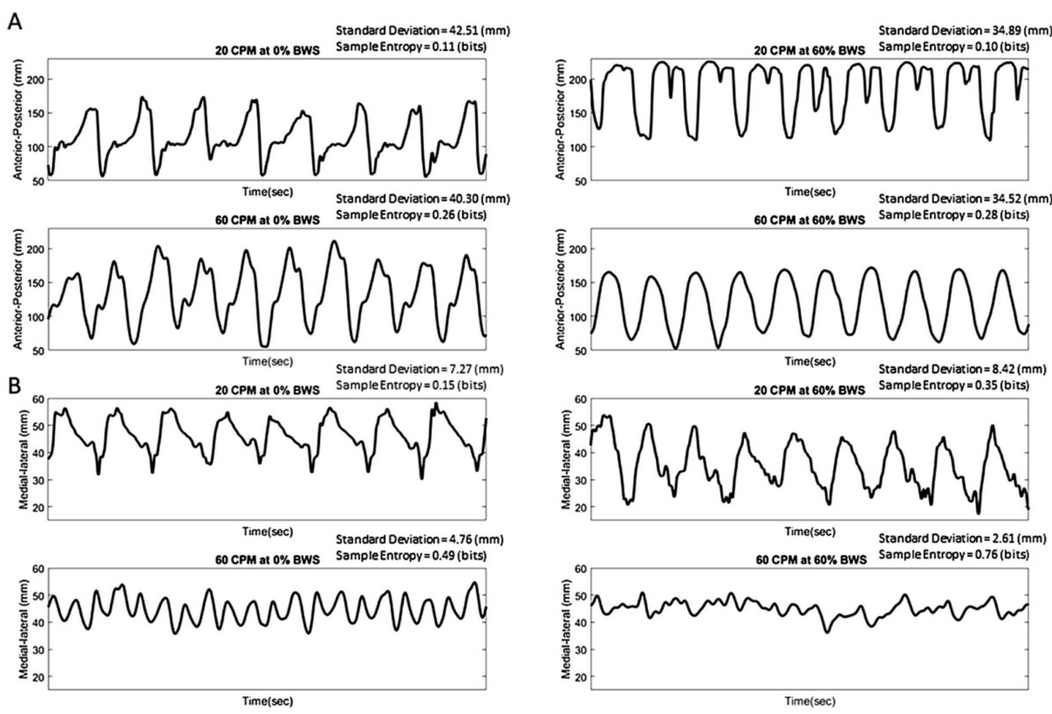


**Figure 2.** Exemplar Center of Pressure (CoP) trajectories across time (seconds) for differing body weight support (BWS) and speed (cycles per minute, CPM) conditions in (A) anterior-posterior and (B) medial-lateral directions. Calculated linear (standard deviation) and nonlinear (sample entropy) CoP variability values provided for each condition.

### *3.1. Anterior-posterior direction*

Training speed and BWS did not significantly affect anterior-posterior CoP linear variability within conditions ($F_{2,14} = 1.68$, $p = 0.22$) and ($F_{3,21} = 1.23$, $p = 0.32$) respectively (Fig. 3A). Anterior-posterior CoP SE demonstrated a main effect of training speed ($F_{2,14} = 27.59$, $p < 0.0001$; Fig. 4A) but not BWS ($F_{3,21} = 0.519$, $p = 0.67$) and no significant interaction ($F_{6,42} = 0.57$, $p = 0.75$). In particular, when averaged across BWS conditions, anterior-posterior CoP motion became more irregular as training speed increased (Fig. 4A).

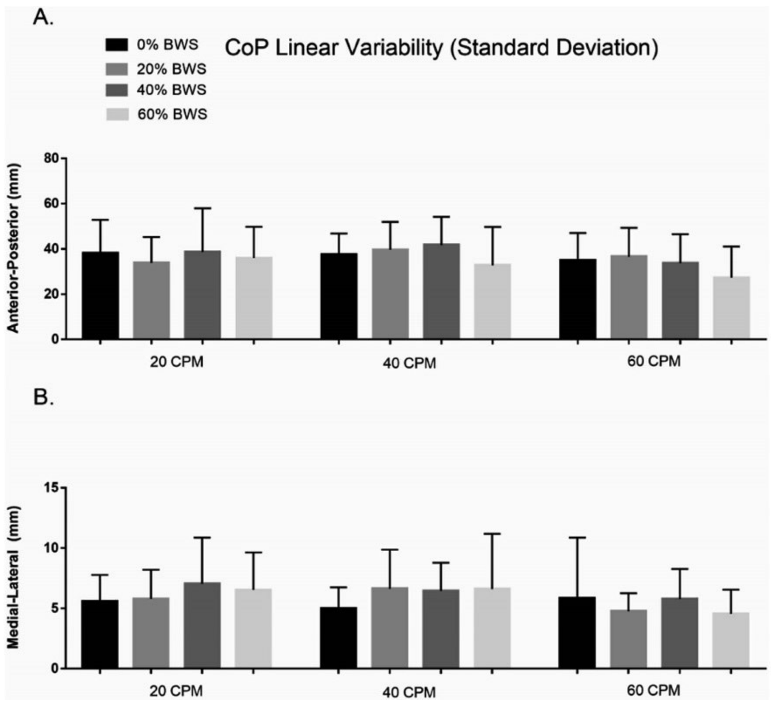


**Figure 3.** Linear variability (standard deviation) of dominant lower extremity CoP excursion ($n = 10$ strides) across three ICARE speeds (20, 40, and 60 cycles per minute, CPM) and four body weight support (BWS) conditions (0% BWS, 20% BWS, 40% BWS, and 60% BWS) in the (A) anterior-posterior and (B) medial-lateral directions. Training speed and BWS did not significantly impact the linear variability of CoP excursion in either the anterior-posterior or medial-lateral directions.

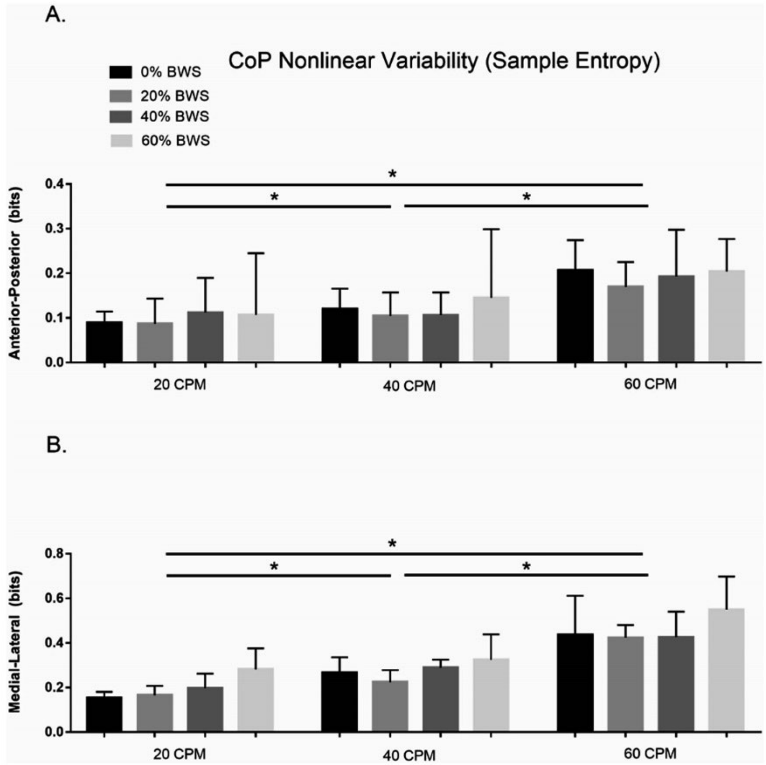


**Figure 4.** Nonlinear variability (sample entropy) of dominant lower extremity CoP excursion ($n = 10$ strides) across three ICARE speeds (20 CPM, 40 CPM, and 60 CPM) and four body weight support (BWS) conditions (0% BWS, 20% BWS, 40% BWS, and 60% BWS) in the (A) anterior-posterior and (B) medial-lateral directions. When averaged across BWS conditions, Anterior-Posterior CoP motion became more irregular as training speed increased ($p < 0.0001$). When averaged across BWS conditions, medial-lateral CoP motion also became more irregular as training speed increased ($p \leq 0.001$ for all pairwise comparisons). When averaged across speeds, medial-lateral CoP motion was more irregular at 60% BWS compared to 0% BWS ($p = 0.01$) and 20% BWS ($p = 0.006$). **Note:** Horizontal bars represent significant differences across training speeds ($p < 0.05$).

### *3.2. Medial-lateral direction*

Training speed and BWS did not significantly affect the medial-lateral CoP linear variability within conditions ($F_{2,14} = 1.34$, $p = 0.30$) and ($F_{3,21} = 0.56$, $p = 0.65$), respectively (Fig. 3B). Medial-lateral CoP SE revealed the main effect of training speed ($F_{1.09,7.68} = 53.31$, $p < 0.0001$) as well as BWS ($F_{3,21} = 6.05$ $p = 0.004$); however, there was no significant interaction ($F_{6,42} = 0.76$, $p = 0.60$) (Fig. 4B). In particular, when averaged across BWS conditions, medial-lateral CoP motion became more irregular as training speed increased ($p \leq 0.001$ for all pairwise comparisons). When averaged across speeds, medial-lateral CoP motion was more irregular at 60% BWS compared to 0% BWS ($p = 0.01$) and 20% BWS ($p = 0.006$).

## 4. Discussion

Successfully navigating homes and communities by foot requires that individuals be able to alter and adapt steps to varying speed, surface, and obstacle demands. Promoting variability in training is positively associated with better performance, adaptation, and quicker learning [22], as it helps individuals adapt their behaviors to environmental change. The relationship between variability and motor learning depends on the nature of intrinsic characteristics of the individual as well as task nature (environment) [23]. Given this need, gait rehabilitation should ideally integrate activities that promote variability in stepping to develop highly adaptable motor recovery [3,4]. However, as novel rehabilitation tools emerge, it is often not evident how a device's unique training parameters can be exploited to promote optimal movement variability. The current study combined linear and nonlinear analyses to explore variability of CoP progression as individuals trained on a motor-assisted elliptical to elucidate how training speed and BWS could be exploited to manipulate not only "how much" (linear) movement variability a user experienced but also "how organized" (nonlinear) the variability.

In support of our first hypothesis, anterior-posterior and medial-lateral linear measures of CoP movement variability (i.e., standard deviation of CoP progression) did not differ significantly across training speeds or BWS levels. We anticipated this finding because ICARE provides a fixed pedal path irrespective of training speed or level of BWS. Although users can vary kinematic patterns while using ICARE [18], and these variations would be expected to alter COP patterns, this was not observed in our study, possibly because of the relatively short recording period. Alternatively, the linear measure of variability we studied may not have been sufficiently sensitive to detect potential differences in peak anterior-posterior and medial-lateral COP excursions [17,24]. Given this is the first study to evaluate CoP excursion across ICARE speed and BWS conditions, we do not have other similar studies for comparison. However, previous research focused on CoP excursion during gait identified greater stride to stride variability during slow walking [25].

In support of our second hypothesis, greater irregularity of anterior-posterior and medial-lateral CoP motion (nonlinear SE) was observed with faster speeds and in the medial-lateral direction with greater BWS. Previous walking studies have also documented that entropy measures are speed dependent [26]. For example, greater nonlinear irregularity in anterior-posterior CoP excursion was documented with faster treadmill walking and was postulated to have resulted from demands of more rapidly loading the limb at faster speeds [27]. Possibly, increased gluteus maximus and medius demands at faster ICARE speeds [18] similarly contributed to the current study's observation of greater irregularity in anterior-posterior and medial-lateral CoP excursion.

Finally, we postulate that irregularity of medial-lateral CoP movement across strides was greater with higher BWS levels, in part, due to participant's ability to employ a wider range of strategies for shifting body weight onto their "stance" limb with increasing levels of external stabilization. We did not observe this pattern in the anterior-posterior direction, possibly due to the handhold's impact on guiding arm and trunk movement.

Collectively, this study's findings have several implications for clinical practice. First, we identified that ICARE has the ability to promote variability during training. This is important, as robotic gait trainers that promote overly prescriptive movement patterns may not encourage the highly adaptable movements that appear critical for successful navigation of complex community environments. Second, clinicians have the ability to manipulate variability using both speed and BWS. When combined with the ability to use similar ICARE variables to manipulate the demands placed on lower extremity muscles [18] and the cardiovascular system [28], clinicians have the capacity to customize training programs to address clients' unique needs. For example, if therapeutic goals target simultaneously promoting movement and cardiovascular fitness, clinicians might elect to increase training speed within/across sessions given that faster motor-assisted elliptical speeds increase nonlinear measures of movement variability and cardiovascular demands. Increasing ICARE training speed and decreasing motor assistance contribute to clinically relevant increases in heart rate, perceived exertion, and systolic blood pressure in young, nondisabled adults [28]. If these same goals also include promoting mass repetition of ICARE's gait-like movement pattern to promote lasting neuroplastic changes in a person recovering from a stroke, then clinicians might select to use a higher BWS level combined with faster training speed to allow movement variability, cardiovascular challenge, and sustained repetition. Additionally, while not explored in the current study, potential options for promoting greater variability in linear measures across strides might include purposefully altering ICARE step length or use of handholds during training. By integrating movement variability into therapeutic gait training, patients can explore and learn different movement strategies beneficial for navigating real world environments and overcoming challenges.

This study has limitations. First, since variability was not the original focus, we analyzed the same number of strides for each speed, leading to fewer data points at higher speeds, as data acquisition frequency was unchanged. It might be expected that the 100 Hz sampling rate influenced SE values and consistency of parameters selected for the SE algorithm [29,30]. Second, given our focus on understanding impact of different speeds and BWS levels on variability, movement variability during each individual's self-selected speed and BWS condition was not evaluated. Variability can be altered when conditions (e.g., speed) are imposed. Future research exploring how differing BWS levels within a self-selected speed or, vice versa, differing speeds within a self-selected BWS level, influence variability could help discern potential impacts of constraining these variables in individuals with and without disabilities. Finally, since this study was a secondary analysis of existing data, an *a priori* power analysis was not conducted. *A posteriori* power analysis identified the study was sufficiently powered to detect differences of medial-lateral SE across conditions. The effect of training speed on anterior-posterior SE was detected. However, future studies using similar procedures would require at least 23 participants to achieve 80% power to detect effect of BWS on anterior-posterior SE. This study was insufficiently powered to detect training speed's and BWS's main effects on linear variability. Our data suggest at least 10 participants would be required for detecting training speed's effect on anterior-posterior and medial-lateral linear variability. To detect BWS's effect on medial-lateral and anterior-posterior CoP linear variability, 22 and 11 participants, respectively, would be required.

In conclusion, this is the first study to evaluate variability and regularity of anterior-posterior and medial-lateral CoP motion with varying speed and BWS during motor-assisted elliptical training. Findings that variability can be manipulated through adjustments in speed and BWS have significant clinical implications for using ICARE to promote, restore, or preserve patient motor behavior. Exploration of functional outcomes for individuals with physical deficits before and following an ICARE intervention involving manipulation of speed and BWS is warranted.

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