

Effects of Random Oscillations on Balance Control in Healthy Young Adults

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EFFECTS OF RANDOM OSCILLATIONS ON BALANCE CONTROL IN HEALTHY YOUNG ADULTS

by

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ABSTRACT
EFFECTS OF RANDOM OSCILLATIONS ON BALANCE CONTROL IN HEALTHY YOUNG ADULTS

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Marquette University, 2017

In human walking, balance control is managed through proactive changes in spatio-temporal parameters of stepping [1]. It has been suggested that continuous disruptions to healthy young adult balance cause greater changes to overall variability of these parameters than a shift in the mean stepping parameters [2]. This suggests that walking may be occurring in a more *reactive* manner, modulating to maintain balance without increasing the mean significantly. Work using continuous oscillations to treadmill walking suggest there is an interplay between the predictability of a signal used to disrupt subject balance and the degree to which compensation occurs [3]. To determine how balance compensation occurs during continuous, unpredictable oscillations this work investigated the effects of unpredictable oscillations on human walking. A 6 Degree of Freedom Motion base was used to oscillate 12 subjects walking on a treadmill for seven different balance conditions: (1) Normal Walking (2) Pitch Amplitude Oscillations, (3) Pitch Frequency Oscillations, (4) Roll Amplitude Oscillations, (5) Roll Frequency Oscillations, (6) Medial-Lateral Amplitude Oscillations, and (7) Medial-Lateral Frequency Oscillations. Amplitude perturbations used a probabilistic multiplier to change the amplitude of an applied sine wave each period, maintaining timing, while frequency perturbations used the same multiplier to vary the timing of sine waves for each period.

Amplitude oscillations caused a greater degree of proactive control characterized by changes in temporal stepping parameters. Frequency oscillations resulted in a greater change in reactive control, demonstrating variability in stepping parameters immediately preceding and following peaks in accelerations peaks which exceed 0.5 m/s^2 . These observations suggest that healthy young adults shift to a reactive strategy of balance compensation when subject to more difficult, higher acceleration oscillations of support surface while maintaining a proactive rate of level walking at low accelerations.

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Chapter 1

Balance Control in Healthy Adult Walking: Exploration of Center of Mass Control in Predictable and Unpredictable Conditions

1.1 Physiology of Standing Balance and Reflex Compensation

Human walking is subject to multiple different types of balance disruption from environmental factors every day which may cause major injury or death due to falls; despite this fact, current literature demonstrates a poor understanding of the causes of balance disruption in human walking in healthy adults. Human beings primarily achieve locomotion through walking balance. Although seemingly simple, walking balance has been the subject of an immense body of literature [4]. Much of this literature focuses on disordered walking—the effects of diseases such as Multiple Sclerosis, Stroke, and Cerebral Palsy on walking kinematics, kinetics, and energy efficiency. This deficiency in understanding balance disruption during healthy gait puts researchers using new technologies, such as CAREN Motion Base systems, at a distinct disadvantage in both human motor control and rehabilitation studies.

1.1.1 Reflex Control of Center of Mass

Standing balance is achieved through manipulation of a bilateral three-joint system of the ankles, knees, and hips to constrain motion of the upper body—torso, head, and arms—between the feet, defined as the base of support (BoS). Each of these joints can elicit movement in three-dimensions in either independent or combined motions to maintain Center of Mass (CoM), the geometric center of the body, over the feet [5]. Muscle function fluctuates over time during quiet standing to maintain the CoM between the Center of Pressure (CoP) of the feet by tensing and relaxing muscles of the lower anatomy—this function can be modeled as a series of springs [6]. These fluctuations occur with a 6 ms lag between CoM and CoP in an

underdamped control system. There is an intrinsic sway, even when standing, to maintain an upright position. This natural process is exacerbated when walking, as the base of support changes over time and the torso accelerates constantly in both the medial-lateral (ML) and anterior-posterior (AP) direction, significantly increasing the challenge.

Reflex control of upright posture is mediated via the vestibulospinal reflex pathway, located in the Central Nervous System (CNS) extra-pyramidal pathway in the brain [7]. This pathway controls feedback between the vestibular system to the body position and is responsible for the righting reflex. The righting reflex corrects error between upright and displaced head position, which in turn guides body position back to upright via activation of the gastrocnemius and tibialis anterior muscles [8]. Use of external perturbation on this standing system likewise causes compensation—body tilt towards the center of the BoS via activation of both ankle and hip reflexes [9].

The ankle strategy of balance has been well documented in its relationship to increased fall risk [10]. The incidence of falling increases when ankle control is disrupted; this disruption can occur during the natural myelin breakdown during aging, or due to pathology in the case of stroke.

1.1.2 Quiet Balance: Eyes Open and Closed

Hip recruitment likewise plays a role in balance stability and compensation. Studies in quiet balance using independent component analysis of correlated force plate and EMG signals demonstrate that balance in the Anterior-Posterior (AP) direction is controlled by activation of the ankle joint, while Medial-Lateral (ML) balance is controlled by hip-flexor ab-adductor activation [11]. Combining these two control mechanism results in a periodic ML and AP sway over the force plate.

Control of sway is tantamount to preventing falls during quiet standing. Analysis of sway during different conditions demonstrates increased ML sway during narrow-stance BoS positions [12]. This increase in ML sway represents decreased ability to control the CoM relative to a neutral stance, indicative of a lower level of balance overall [13]. Closing the eyes causes an increase in CoP velocity and decreases balance. This indicates that the visual system coupled with the proprioceptive input modulates quiet balance. Quiet standing, with the eyes closed and feet apart, demonstrates that random fluctuations in muscle activity impact the control of CoM. There is an intrinsic sway, even when standing, to maintain an upright position against gravity. This process is exacerbated when walking, as the BoS changes over time.

Further breakdown of this motion in healthy adults using cross-correlational analysis elucidates a feed-forward system of control for quiet balance [14]. Gastrocnemius activation dominates AP sway, exhibiting a 250-300 ms delay between sway timing and activation. When standing in a Romberg stance (feet parallel, eyes closed), significant correlation between a wide stance versus narrowed stance demonstrates the domination of ankle-based vs hip based, respectively, activation to control CoM. A feed-forward predictive system is used to modulate this activation—CoM motion is permitted in this controller, allowing for a predictive model of the motion that allows ankle muscles to activate in the proper time frame. This feedforward system is implemented and adapted in walking balance, modulating more gross characteristics such as step length and step width to minimize AP and ML sway of the CoM during normal walking.

1.1.3 Perturbations and Quiet Balance—Planar Perturbations

Applying AP perturbations during standing trials reveals the efficacy of the ankle and hip strategy in quiet balance. When standing on support surfaces of different dimensions, humans

utilize varied control strategies to maintain balance under a forward-backward perturbation. A support surface with dimensions that accommodates the subject's foot yields an ankle-based response pattern: stabilization begins at the distal muscles of the gastrocnemius and tibialis anterior, propagating superiorly with a 92-113 ms delay to the proximal muscles of the quadriceps and hamstrings [15]. In contrast, when the support platform is smaller than the foot, subjects use a hip-based strategy: compensation begins with activation of the quadriceps and hamstrings with torso interaction, but fails to propagate down the kinetic chain, producing little activation of the ankle muscles [16].

By contrast, ML perturbation during standing balance produces a primarily hip-based strategy relative to AP perturbations. Compensation begins kinematically by the trunk shifting opposite to the direction of perturbation, with application of hip-adduction opposite the perturbation direction [17]. A three-stage compensation pattern occurs, with hips and ankles displacing in response to the perturbation, overcompensating via adduction, then settling to a steady-state value. Increasing the size of these perturbations increases the kinematic response, creating greater angular corrections and trunk motions relative to smaller perturbations, with increased overshoot of steady-state values [18].

1.1.4 Perturbations and Quiet Balance—Roll and Pitch Perturbations

Pitch perturbations, or “toes-up” and “toes-down” perturbations cause compensation in the sagittal plane with motion combined in the ML and AP directions, forcing greater balance correction. The tibialis anterior and soleus combine to compensate for pitch perturbations reflexively—the ankle-based strategy adopted during AP movement is mirrored when the support surface tilts [19] [20]. Co-contraction of the tibialis anterior and soleus muscles

primarily control the balance strategy at the ankles with a supraspinal reflex in the direction opposite the tilt that compensates for the perturbation.

Frontal plane rotation, by contrast, does not contribute to a primarily hip-based strategy. Ankle activation again occurs, combined with increased compensatory hip torque in the frontal plane. Stiffening at the trunk occurs immediately after the perturbation, which results in an overcorrection. A combined hip-ankle strategy more effectively describes balance compensation to frontal plane *rotations*, rather than either a primary ankle or hip strategy individually for translations [20] [21].

1.2 Walking Balance and Spatio-Temporal Control

1.2.1 *Mechanical Determinants of Gait*

Saunders et al. began the initial study of gait and control with their seminal paper “The Six Determinants of Normal and Pathological Gait” [22]. They posit that walking controls the CoM, minimizing its displacement in both the sagittal and frontal planes. This minimized motion, in turn, reduces energy expenditure; CoM motion can be optimized to simulate both normal and pathological gait. The respective six determinants are: pelvic rotation (frontal plane), pelvic tilt (transverse plane), knee flexion during stance, tandem ankle rotation and knee flexion, and lateral pelvic displacement. These segment and joint rotations were studied in both normal and pathological subjects, via kinematic observation, to determine impact on walking gait. Removal of any determinant resulted in decreased walking speed, decreased stability, and increased energy expenditure during the gait cycle.

When humans walk, several key factors are regulated: step length, step width, and stepping frequency (e.g. cadence). Walkers set a speed and use a combination of gait variables [23] to control this speed. Step length and width are used to affect gait on a step-by-step basis;

step frequency affects speed control over time [1]. There is an intrinsic variability in step length and width. Even in undisturbed walking, a baseline level of variability exists in the step length and step width which affects stability of the walking subject [24]. These variables are not completely random, but change on a step-by-step basis with past steps affecting future steps. These step parameters change to maintain the CoM over the base of support.

1.2.2 Machine Models of Walking Balance

The area of robotics has provided a great expansion for the field of dynamic balance. Control theory has been used to model each of the degrees of freedom at the ankle, knee and hip and optimize motion for humanoid robots. One promising area of research focuses on the *inverted pendulum*, wherein the CoM moves back-and-forth in a periodic motion pattern [25]. Findings indicate that step parameters change to maintain the CoM over the BoS.

Robotic models can simulate AP motion during walking: when placed on a downhill slope, these models exhibit passive forward progression without the need for compensatory balance mechanisms. Single-joint robotic toys can achieve this downhill motion easily on a sloped surface [26]. The inclusion of additional joints to the model may introduce failure regions; the model must therefore incorporate system constraints such as the aforementioned determinants of gait [27]. The McGreer machine, in particular, can be started with a range of initial conditions or joint angles; the model then settles into a convergence zone of dynamic stability such that walking down a slope may be simulated without any control corrections.

More robust models have been developed to simulate balance in the frontal plane during walking. Although passively stable on a downslope, ML perturbations cause these three degree of freedom models to diverge from the dynamically stable region to a failure region [28]. These marginal disturbances to ML balance may be mitigated through the inclusion of active

components (e.g. wheels or joint-stabilization at the ankle) or passive components (lateral step width and increased energy storage at the hip). Each active component reflects a feedback mechanism and strategy present in true human walking on level terrain.

1.3 Strategies for Walking Balance Compensation

1.3.1 Proactive Walking Compensation

During overground walking, the problems of quiet balance are amplified: the Center of Gravity (CoG, the projection of CoM) must be balanced between a continually changing BoS in both the ML and AP directions [11]. Walking has been described as a series of falls—the CoM of the body must be accelerated forward to achieve motion, then subsequently caught by the swinging limb at heel strike. Balance in this sense is *dynamic*, although the CoG is not located within the BoS during walking, the swinging limb moves towards a stable position during the subsequent heel strike. The steady-state version of this system creates a dynamically stable oscillator—it occupies similar positions over time without settling to a single controlled point in space [29].

The motion of the CoM follows a sinusoidal pattern in the sagittal plane between heel strikes. Brief instances of stabilization occur during double-limb support, when the CoM falls within the BoS, before being accelerated forward. Double-limb support time increases in pathologic populations, increasing stabilization timing for the walking motion [30]. However, double-limb support makes up less than one-tenth of the standard gait cycle—more sophisticated mechanisms must exist to explain balance throughout the gait cycle.

Similarly, balance in the frontal plane is also sinusoidal, but more intuitively controlled. Accelerations occur due to ML motion: during heel strike, the CoM sway velocity is transferred

from the ipsilateral to the contralateral foot. In the frontal plane, the inverted pendulum moves less than $\pm 2^\circ$ from vertical under typical walking conditions for healthy young adults [31].

The magnitude and velocity of CoM sway during walking varies with both age and walking speed. Older subjects exhibit a significantly larger ML displacement than young subjects [32]. Likewise, increasing walking speed by 2 to 2.5x produces an additional two degrees of CoM sway. This increase in ML sway has direct implications on joint motion in the frontal plane: to modulate the additional trunk sway, subjects reduce knee and hip adduction moment by 65 and 57.5 percent, respectively [33]. The increase in CoM sway might be compensatory or a preventative mechanism on knee wear, as in the case of osteoarthritis.

1.3.2 Reactive Walking Balance Control

The aforementioned content describes primarily *proactive balance*—a feedforward control system. This control system predicts the CoM position due to existing sway and velocity parameters and compensates by adjusting step length, step width, and trunk control [6]. Visual cues of environmental obstacles such as slope inform the subject's control of the CoM—subjects react with greater muscular activation at the hips and torso to stabilize the inverted pendulum due to the environment [5]. Trunk and hip-flexor muscles, including the erector spinae, hamstrings, and gluteus maximus activate at 5% of the gait cycle before heel strike and continue to activate halfway through stance phase, guiding the CoM through the gait cycle and preventing AP or ML trunk collapse [34].

1.3.3 AP Slip Compensation

By contrast, *reactive balance* is primarily initiated at the distal limbs. A reactive balance strategy compensates for a loss of balance by activating distal limb segments to restore continuous walking balance [5]. EMG recordings of lower anatomy muscles reveal that during a

reactive slip, where the terrain moves beneath the subject's foot, muscle activity overcompensation occurs. Muscular firing increases 2-3X baseline firing rate with a prolonged period of activation on a slip introduced by a moving platform [35]. Modulation of the gait cycle does not occur—instead, increased stabilizer and push-off forces are generated to complete the initial cycle as intended. The trunk muscles serve little in the way of compensation during a slip at heel strike, though the hamstrings and quadriceps do serve a significant correlation to initial balance recovery [36].

Regardless of slip timing, compensatory muscular activation remains similar across anterior-posterior single-foot slips. Experiments on treadmill walking during positive and negative acceleration impulses exhibit ipsilateral gastrocnemius or tibialis anterior activation with contralateral tibialis anterior or gastrocnemius activation, respectively, based on a deceleration or acceleration with 65-76 ms of activation latency [37]. Upper-body anatomy likewise plays a role during slipping: activation of deltoids tends to bring the arms up to compensate for a posterior CoM slip [38]. Reactive latency of the unperturbed limbs mimic the latency of the perturbed limb—there is not a significant timing advantage to phase lag of the unperturbed limb or upper anatomy, the body springs to action as quickly as possible to prevent a fall.

Work studying the effect of walking under ML force perturbations demonstrates that increased compensation and aftereffects occur during unpredictable ML perturbations in both healthy and incomplete spinal cord injury (iSCI) patients [2]. iSCI subjects exposed to unpredictable ML forces while walking on a wide-belt treadmill exhibit greater step width (SW) and medial-lateral margin of stability (ML-MoS), decreased step length (SL), and increased stepping frequency. However, healthy adults exposed to the perturbation field do not exhibit significantly different SW, SL, or ML-MoS over the course of the trial—but expressed a greater

variation in their stepping parameters. This implies a more reactive strategy is employed in continuous ML perturbations by healthy adults, mirroring the work in AP perturbations above.

As noted previously, the frontal plane is less stable than the sagittal plane during walking due to challenges in CoM control. Studies involving geriatric falls note that lateral balance recovery is more difficult due to problems in limb placement [39]. Investigating both AP and ML perturbations during walking represents a fertile field for new growth and understanding.

1.4 Quantifying Balance Control

1.4.1 Margin of Stability

One balance quantification method incorporates both CoM position and velocity relative to the BoS. This is known as the Margin of Stability (MoS). MoS is a metric which incorporates both the size (P_{CoM}) and speed (V_{CoM}) of the inverted pendulum motion, normalized to subject leg-length and gravity (l/g) relative to the BoS (Equation 1.1) [40].

$$XCoM = P_{CoM} + V_{CoM} * \sqrt{l/g}$$

1.1

$$MoS = BoS - XCoM$$

1.2

MoS quantifies the difference between BoS at heel strikes relative to both the displacement size and velocity of the CoM. MoS provides a quantifiable measure of balance used to assess changes in balance due to voluntary changes in step length and step width in

healthy subjects; subjects significantly increase ML-MoS from .09 meters to .16 meters during an increased step width [41].

1.4.2 *Double Limb Support Timing*

As mentioned previously, the level of balance impairment and disruption can be quantified by examining the amount of time spent in double-limb support. Double-limb support occurs between heel strike of the leading foot and toe-off from the trailing foot and represents approximately 10% of the gait cycle [30]. The purpose of this timing is two-fold: to rebalance during trunk sway between the BoS and to initiate swing of the trailing leg [1][42]. The former has a greater impact on subject balance confidence and more relevance to this work.

Elderly subjects exhibit a greater incidence of double-limb support compared to healthy young adult populations. For example, during self-selected walking trials, double-limb support duration is 20-35% of the gait cycle for elderly subjects [42]. Sedentary older adults also demonstrate increased double-limb support time compared to their healthy older peers [42]. This timing manifests differently between healthy and pathologic populations: pathologies such as Parkinson's disease and Stroke, which have increased ataxia, also exhibit increased in double-limb support duration [43], [44].

Double-limb support duration increases in young adults who experience difficulty walking. The majority of studies on young adult walking focus on dual-task conditions involving treadmills or overground while concurrently performing a memory, mathematic, or fine-motor task. These tasks may consist of remembering a sequence of numbers, subtracting by a sequence of numbers, or solving a knot-puzzle [45]. When concurrent tasks are presented to challenge to young adult walking, double-limb support duration also increases [46].

1.4.3 Kinematic Analysis using Motion Capture

Motion capture technology was born out of the direct linear transformation technique for two camera capture proposed in the early 1970's. This technique utilizes two cameras and a calibration technique to isolate 11 coefficients (L1 through L11) which define the orientation and internal parameters of two cameras. Data captured by each camera are then transformed from "Local Camera Space" positions (x,y) to "Object State Space" positions (X,Y,Z) with user-defined non-linear system noise ($\delta x, \delta y$), random error ($\Delta x, \Delta y$), and linear transformation constants (L₁ through L₁₁) and solved using Equations 1.3 and 1.4 [47]. Advances in this method progressed with increased computer power, allowing for multiple cameras to effectively be integrated and calculations for a larger object state space. More effective image processing techniques utilizing least squares quadratic algorithms can be used to more effectively calibrate capture volumes[48].

$$x + \partial x + \Delta x = \frac{X*L_1+Y*L_2+Z*L_3+L_4}{L_9+L_{10}+L_{11}+1}$$

1.3

$$y + \partial y + \Delta y = \frac{X*L_5+Y*L_6+Z*L_7+L_8}{L_9+L_{10}+L_{11}+1}$$

1.4

Motion capture gait analysis allows for quantitative analysis of gait patterns across different conditions, cadences, and patient groups. This facilitates statistical analysis across different gait patterns [49]. Joint angle quantification is referred to the standard 1-100% gait cycle determination, with heel strike and toe-off events calculated using the maxima and minima of ankle-marker position[50]. Motion capture is hampered by marker artifact caused by the shift of the marker over the landmarks on the skin due to its natural elasticity and pliability.

To minimize this effect, markers are placed on bony landmarks which run proximal to the surface, which minimizes movement to +/-5mm. Likewise, additional tracking markers and increasing camera triangulation provides better fidelity for motion analysis [51].

Motion capture models rely on established biomechanical data, fusing it to define mass and length properties with inertial components via proper marker placement. Development of this model began with the Golem Model, which utilizes anthropometric data to establish the mass of the pelvis, shank, femur, and foot segments by defining proximal and distal segment locations [50]. The Helen-Hayes marker system is a modification on the Golem system, changing the locations of markers and primarily defining the lower anatomy based on the definition of the hip [52]. Marker placement is contingent on the ability to accurately identify markers—trained technicians and physical therapists can adequately repeat the output of these models [53]. These models are often difficult to use on subjects with orthopedic pathologies or who exhibit excessive fat distribution, but modified Helen-Hayes marker sets are developed to adequately study this population [54].

1.5 Understanding Perturbation Compensation in Treadmill Walking

1.5.1 *Understanding Overground and Treadmill Walking*

Treadmill walking exhibits minimal differences compared to overground walking in healthy young adult populations. The advantages of treadmill walking are twofold in motion capture environments: it provides a large sample size for cross-subject variation analyses, and allows subjects to continuously walk in a straight line, rather than forcing a turn-around in the middle of a trial. There are minimal kinematic differences between treadmill and over-ground walking, mainly resulting in small (<3% from overground) changes in step length, width, and

joint angles [55]. These differences are minimized after an accommodation period of 5 minutes [56], [57], though this familiarization time may take longer in older populations less accustomed to treadmill walking [58]. Likewise, kinetic parameters of treadmill walking are marginally smaller in instrumented treadmills compared to overground walking, but remains within reported variability parameters [59], [60].

1.5.2 Medial-Lateral Perturbations in Treadmill Walking

The advent of treadmill mounted platforms have allowed for the study of balance reactions and slips in a controlled environment. Platforms such as these typically consist of a split-belt treadmill, motion-capture system, electromyogram, and six-degree-of-freedom motion base. These systems are commercially available from the MOTEK Corporation in the CAREN system [61]. Perturbations may be applied in the Medial-Lateral or Anterior-Posterior planes, or consist of rotations of the base in the Roll, Pitch, and Yaw directions (Figure 1.1). Real-time feedback allows researchers to study self-selected static speed vs real-time update speed feedback; when experiencing oscillations, subjects do not significantly alter walking speed when given the opportunity [3].

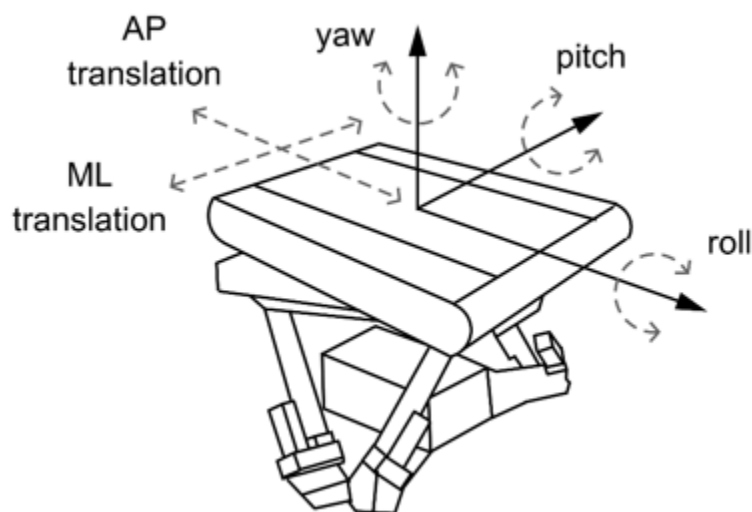


Figure 1.1: Motion base perturbations including ML and AP translations. Pitch rotation occurs about the X-axis, Roll rotation occurs about the Y-axis, and Yaw rotation occurs about the Z axis.

Perturbations may be applied to stimulate balance compensation, allowing

researchers to study the effects of perturbations on proactive and reactive balance. Subjects exhibit increased step width and decreased step length inherent to a disturbed walking balance, changing due to increased perturbation amplitude.

Unpredictable perturbations provide an increased challenge for healthy subjects, forcing greater compensation in step width, step length, and MoS [3]. Creating a pseudorandom perturbation signal allows researchers to challenge balance within the context of smooth disruptions using four superimposed sine waves with four different frequency parameters and a scaling factor, A, (Equation 1.5).

$$D(t) = A * (1.0 * \sin(0.16 * 2\pi t) + 0.8 \sin(0.21 * 2\pi t) + 1.4 \sin(0.24 * 2\pi t) + 0.5 \sin(0.49 * 2\pi t))$$

1.5

Although not a truly random signal, predictability of this signal for a human subject is lower than a pure sine wave and thus creates greater challenges to balance and compensation. Increasing the scaling amplitude significantly increases step width, decreases step length, and causes an increase in margin of stability on healthy young adult subjects [62][3]. The utility of the CAREN system allows for an exploration of walking speed—although the subjects slow their overall step length, stepping frequency increases to compensate and self-selected speed does not significantly change during any perturbation intensity [3]. This indicates that an initial self-selected pace is useful in measuring walking parameters during treadmill walking.

The majority of literature, does not exploit the Roll and Pitch Perturbations of the Motion base system because the visual flow software inside the CAREN system is poorly suited to the task; the screen-based system can make subjects nauseous. However, the changes in kinematics and kinetics of uphill and downhill graded treadmill walking have been probed.

1.5.3 Uphill and Downhill Changes in Body Kinematics during Treadmill Walking

Uphill walking is substantially more difficult to perform than level walking due to the increased positive mechanical work necessary to move up a gradient. During level walking, movement of the inverted pendulum causes the leading leg to produce negative work in the AP and ML directions to break and reverse the pendulum sway. Consecutively, the trailing leg must generate a positive force to drive the body forward[63]. In uphill and downhill walking, more net positive and negative (respectively) work is generated by the leading/trailing limbs during double limb support [64]. During downhill walking, ML CoM velocity remains negative for 7-27% longer in the gait cycle between -3 and -9 degrees of downhill walking relative to level walking. Subjects also typically adopt a lean-forward and to the left while walking uphill while righting themselves for downhill walking [65]. This suggests that greater challenges occur in the ML plane during consecutive uphill and downhill walking, more effectively challenging balance due to postural preference and kinematic adaptation than normal walking. The grade at which subjects walk has a direct effect on the amount of mechanical work performed as well as the compensatory posture adopted. Whether continuous uphill and downhill sinusoidal perturbations are more difficult than ML perturbations remains to be explored.

1.6 Summary and Statement of Purpose

An exploration of literature on standing balance perturbations, the mechanics of walking, and compensation techniques for balance perturbations demonstrates a deep utility in exploring the level of compensation provided by different continuously applied perturbations, or oscillations, to healthy young adult subjects. Pitch, Roll, and ML perturbations provide the most difficult challenge to quiet standing balance in healthy populations. Likewise, because

frontal plane stability is reduced intrinsically due to the increased CoM velocity and decreased base of support, perturbations within this plane will likely provide the greatest challenge to balance in healthy adults. Changes in grade of uphill and downhill walking significantly change kinematics and kinetics of CoM, challenging balance and forcing compensation—changing the grade of these values continuously will further destabilize subjects. Although there is a lack of literature on the effect of Roll perturbations, the same effect will likely be present in those perturbations.

Likewise, the speed at which perturbations occur will likely force more compensation due to a greater change in the vestibulospinal reflex. Differences in acceleration have already demonstrated varying levels of compensation in standing balance, forcing greater hip-and-ankle strategy adoption at higher platform accelerations. The degree to which oscillation acceleration impacts *walking* balance, however, has not been explored to great effect in the literature. The standard pseudorandom perturbation signal muddles both changes in amplitude and acceleration, obfuscating underlying mechanisms that impact walking balance. *This study aims to mechanistically isolate balance challenges due to changes in amplitude and frequency in Pitch, Roll, and ML oscillations during treadmill walking in young healthy subjects.* This will yield information on both proactive compensation changes in SL, SW, and MoS and any reactive changes due to an unpredictable signal changes. Specifically, the research hypotheses are that: 1) changes in amplitude will result in smaller changes in reactive balance than changes in frequency (if both are scaled to the same time and oscillation length), and 2) more proactive and reactive stepping will occur during the frequency than the amplitude perturbations. Study results will be used to define a graded challenge scale for balance in terms of proactive and reactive changes in step width, step length, and ML or AP MoS.

Chapter 2

The Effects of Random Oscillations on Balance Control in Healthy Young Adults

2.1 Introduction

The purpose of this study was to characterize gait responses to perturbation movements of a walking surface, incorporating different amounts of predictability, to provide the foundation for the development of walking balance therapies. The use of motion platforms to oscillate and perturb subjects while walking is emerging, with potential applications in the rehabilitation of gait [66]. There are many possible ways to manipulate a walking surface, including six possible degrees of freedom (3 rotations and 3 translations) and numerous waveforms, amplitudes and frequencies with which to. Prior work has focused primarily on pseudorandom oscillations in the medial-lateral (ML) direction [3], [62]. In the current study, we expanded on the types of walking surface movements, both in examining the movement directions and in varying the amount of predictability to the walking surface movements. We aimed to evaluate reactive and proactive compensation to various movements.

Reactive balance compensation is a change in the *variability* of spatio-temporal walking parameters in response to mild balance instability. Reactive balance compensation is a strategy favored by healthy young adults due to its energetic efficiency and use of short latency feedback systems to maintain balance during walking [63] [67]. Humans utilize a mixture of both feedforward and feedback control to maintain gait [68]. Feedback control is favored in healthy adults when undergoing mild balance destabilization in which the variability of spatio-temporal metrics increases due to occasional corrective stepping, but mean parameters remain constant [2], [69]. Neuropathic, ataxic populations such as incomplete Spinal Cord Injury and Multiple Sclerosis demonstrate decreased ability to engage in reactive compensation and favor a more feedforward strategy to maintain walking balance [70] [71].

Proactive balance compensation can be defined as a change in *mean* spatio-temporal walking parameters such as step width, in individuals experiencing balance instability. Ataxic individuals often exhibit these proactive changes, widening step width and decreasing step length to maintain balance [72], [73], [74]. External balance perturbations can also force proactive compensation in healthy adults. ML oscillations of walking platforms produce increases in proactive compensation during stepping in healthy adults [66],[74], [75]. Important to note in this work is that a change in both mean and variability occurs in healthy young adult SW, SL, and MoS [62]. By observing the response to walking surface perturbations, we can assess the proactive and reactive balance compensations in healthy adults. These observations have implications for treatment of neuropathic balance.

This study assessed reactive and proactive compensation to balance perturbations in healthy adults using walking surface oscillations with varying frequency or amplitude of movements to alter the level of predictability of the perturbations. Disruptions were administered in the Pitch, Roll, and ML directions to elicit both sagittal and frontal-plane balance challenges. Compensation to the walking surface oscillations were measured using step width (SW), step length (SL), medial-lateral margin of stability (ML-MoS), and anterior-posterior margin of stability (AP-MoS). Mean changes of these parameters was used to quantify proactive balance compensation, while changes immediately after an acceleration of the base were used to quantify reactive balance compensation. Characterization of the proactive and reactive compensation to different walking surface movements provide the basis for balance training paradigms in people with balance disorders.

2.2 Methods

2.2.1 Subject Population

Twelve healthy young adults (6 males) with an average age of 23.3 ± 2.6 years participated in this study. Average subject height was 1.72 ± 0.08 meters, and weight was 67.20 ± 8.83 kg. Subject self-selected walking speeds were 1.00 ± 0.10 meters per second based on treadmill speed. Subjects had no self-reported injuries nor prior orthopedic surgeries. All experimental procedures were approved by the Marquette Institutional Review Board, protocol number HR-3251.

2.2.2 Testing Equipment

Subjects walked on a custom motion platform system (Figure 2.1) that consisted of a split-belt instrumented Woodway treadmill (Woodway USA Inc., Waukesha, WI) mounted on top of a six degree of freedom motion base (Model MB-E-6DOF/12/1000KG, 170-140, Moog Inc., East Aurora, NY). This system allowed for the walking surface to be perturbed in six different degrees of freedom (Figure 2.1). The base was controlled using a custom LabVIEW program (National Instruments Corp., Austin, TX), which provided a command oscillation signal to the motion base and collected kinematic, kinetic and electromyogram data. To ensure subject safety, an overhead safety harness provided emergency fall-arrest, lateral handrails were available in the event of subject stumbling, and an emergency stop button was subject accessible in the event the subject became uncomfortable and wished to discontinue the study.

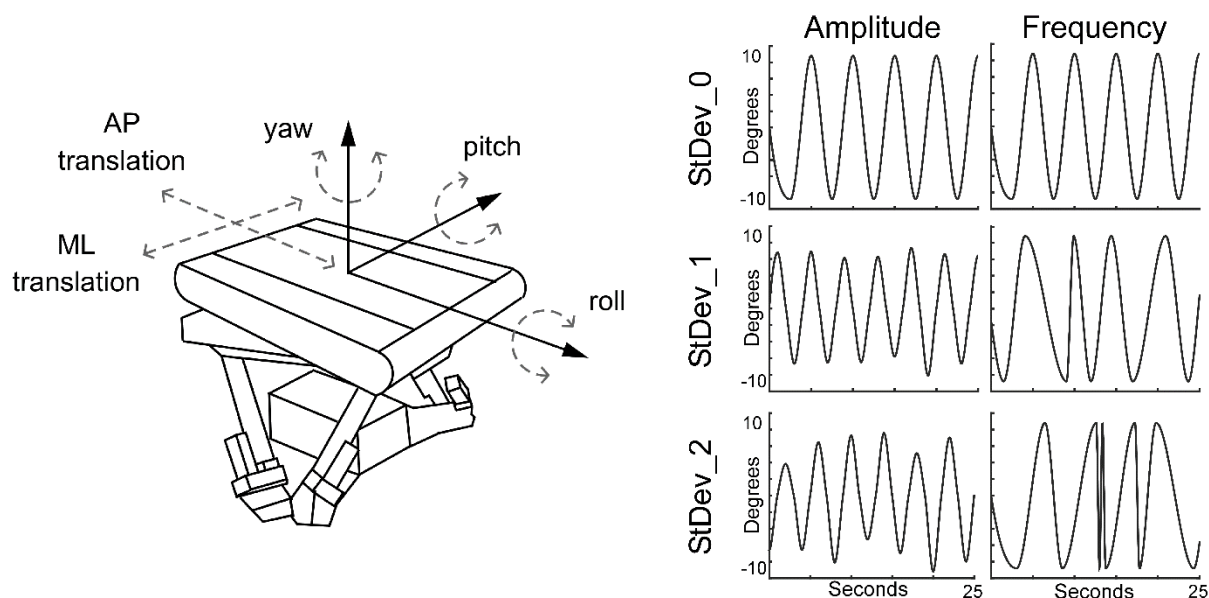


Figure 2.1: The motion base system (left) was capable of movements in 6 degrees of freedom. Of these, roll, pitch, and medial-lateral (ML) translation were used to perturb balance while walking. The perturbations (right) consisted of sinusoids with either changes to amplitude or frequency determined by a Gaussian-distributed random multiplier. Multipliers were constant (StD0), resulting in a pure sinusoid, a random multiplier based on a distribution that was one standard deviation about the mean amplitude or frequency (StD1), or a random multiplier that was two standard deviations about the mean amplitude or frequency (StD2). The randomization multiplier was derived from a Gaussian, rather than flat, distribution.

Thirteen motion capture cameras were used to collect kinematic data at 120 Hz (NaturalPoint Inc., Corvallis, OR). Subjects wore reflective markers attached to lower-anatomy anatomical landmarks: bilateral anterior superior iliac spines, sacrum, greater trochanter, medial and lateral femoral condyles, medial and lateral ankle malleoli, and second and fifth metatarsal heads bilaterally. Likewise, four markers were placed on the upper anatomy including the clavicle, C7, and left-and-right acromioclavicular joints. Locations of these landmarks were found through manual palpation. Three-marker clusters attached to plastic plates were placed on the thigh, shank, and heels of the subject's shoes.

2.3 Experimental Protocol

2.3.1 Walking Speed Selection

Subjects were familiarized with the treadmill system prior to the beginning of experimental trials. Treadmill belt speed was determined by increasing incrementally by 0.045 m/s until the subject reached a self-reported comfortable walking speed. This speed was maintained for experimental trials. An initial series of 7 level walking trials of 70 seconds were taken with no movement of the walking surface. This served as the Normal_A condition to which each of the 6 perturbation types were compared.

2.3.2 Experimental Trials

This experimental protocol consisted of 6 dynamic balance walking trials. Each trial was recorded continuously for 490 s and was composed of seven, 70 s segments. Each trial began with a 70 s period of level walking (Normal_A), followed by a sinusoidal oscillation with an amplitude 7 degrees (roll and pitch trials) or 17.8 cm (ML trials) at 0.2 Hz (StD0_A). The third segment was a sinusoid with a randomized amplitude/frequency based on a Gaussian distribution consisting of 1 standard deviation about the initial signal (StD1_A); the fourth segment was a sinusoid with a randomized amplitude/frequency based on a distribution consisting of 2 standard deviations of the initial signal (StD2). Fifth (StD1_B), sixth (StD0_B), and seventh (Normal_B) correspond to one standard deviation, pure sine wave, and level walking respectively. For each trial, motion was applied in the pitch, roll, or ML direction. Trials also differed in either the amplitude or frequency manipulations of the movement. Amplitude trials maintained the sinusoid frequency while varying movement amplitude. Frequency trials maintained the movement amplitude, but varied the timing of the sine wave. Changes in amplitude or frequency were made every one-half period of the sinusoid. Trials were labelled

either the Pitch Amplitude (PitchAmp), Pitch Frequency (PitchFreq), Roll Amplitude (RollAmp), Roll Frequency (RollFreq), ML Amplitude (MLAmp), or ML Frequency (MLFreq) depending in the movement direction and variation in amplitude or frequency. Subjects were given 120 s rest between the 490 s trials.

2.4 Data Analysis

2.4.1 Spatio-Temporal Parameter Calculations

Motion capture data were used to model body segments and determine gait parameters indicative of balance disruption. Kinematic data were low-pass filtered using a 4th order Butterworth filter with a cut-off frequency of 6 Hz in Visual 3D (C-Motion Inc., Rockville, MD). Local coordinate systems were used to define model segments for the torso, pelvis, thighs, shanks, and feet while kinematic data were used to determine heel strikes. Each heel strike was defined at the maximum AP-distance between the pelvis and the foot and verified by visual inspection. Center of mass (CoM) of the body was defined by the weighted average of each segment CoM based on anthropometric data from Winter et al (Winter 1990). Step length and width were calculated as the sagittal and frontal plane distances between ankle markers at heel strike, respectively. During roll and pitch perturbations, marker positions were corrected using an Euler transform applied to the coordinates of the base. Double-limb support was calculated as the sum of right heel strike to left toe off and left heel strike to right toe off, divided by the difference between right heel stride timing.

2.4.2 Margin of Stability Calculation

Margin of stability was calculated using the methods of Hof (Hof 2005) and Hak (Hak 2012). Briefly, maximum CoM displacement was found between each step during walking and subtracted from the position of the base of support (BoS), defined by the difference between

lateral ankle markers. Velocity of the CoM (V_{CoM}) was calculated by finding the first derivative of the calculated CoM. Similarly, velocity of the motion base, $V_{MotionBase}$ was calculated from markers on the base position and then subtracted from the CoM velocity. This difference was then normalized to the subject's leg length (l) divided by gravity (g) (equation 2.1). This value was added to the position of the CoM (P_{CoM}) to find the extrapolated center of mass (XCoM). This value is subtracted from BoS to find MoS (equation 2.2).

$$XCoM = P_{CoM} + (V_{CoM} - V_{MotionBase}) * \sqrt{l/g}$$

2.1

$$MoS = BoS - XCoM$$

2.2

2.4.3 Reactive Balance Calculation

To measure subject compensation to platform motion, the step width (SW), step length (SL), and MoS were correlated to walking surface movement peak acceleration events. Peak accelerations of the walking surface ranged from 0 to 1 m/s² during the walking trials.

Acceleration of the base was calculated by the second derivative of marker positions on the motion base. Peaks in acceleration were then found using the MATLAB Function peakfinder and values and timing of maxima greater than 0.1 m/s² were identified. Low accelerations were classified as events with peak accelerations from 0.1 to 0.5 m/s², while high accelerations were between 0.5 m/s² and 1 m/s².

2.5 Statistical Analysis

To determine significance differences between oscillation types, one-way repeated measures ANOVAs were conducted to compare the effect of oscillation type on SW, SL, and MoS. Sphericity was assured using Mauchly's test of Sphericity. Greenhouse-Geisser correction factors were used when the sphericity assumption was violated. α -level was set at 0.05 and data are reported as mean \pm standard deviation in the text and figures. Post-hoc tests were conducted with a Sidak correction for multiple comparisons. Main effects and contrasts between oscillation types are presented, unless otherwise stated. All statistics were performed using IBM SPSS Statistics 21.0 (IBM, New York, NY). All reported significances are from $p < 0.05$ or less and a power of 0.9 or greater.

2.6 Results

All participants completed each trial without falling or balancing themselves using the safety handrails. Subjects reported that the trials were difficult and frequently reported the frequency perturbation trials as more difficult than the amplitude perturbation trials.

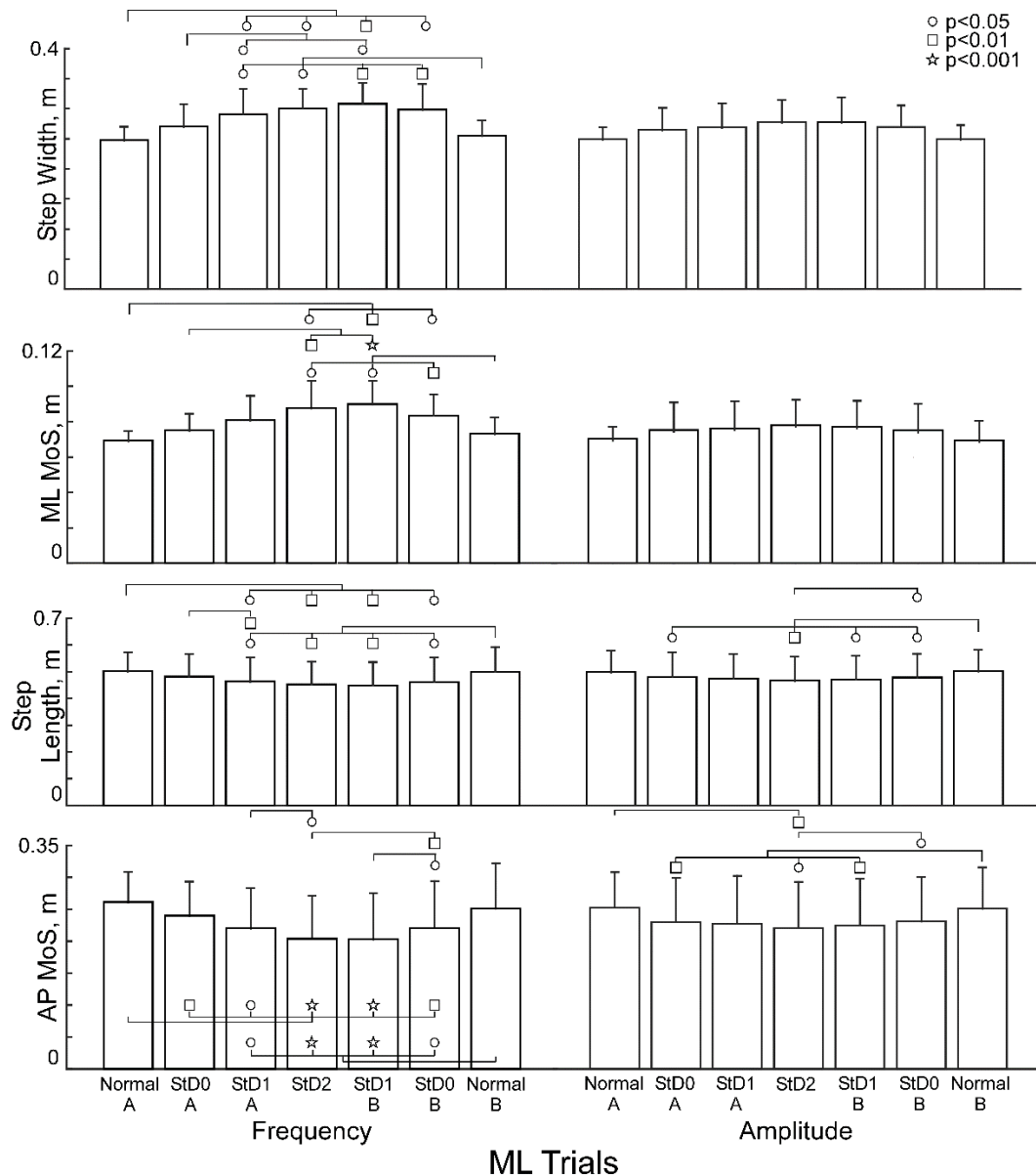


Figure 2.2: ML amplitude and frequency responses over the seven portions of each trial. Trials were subdivided into a beginning round of Normal walking (Normal_A), the first sinusoid (StD0_A), the one standard deviation multiplier (StD1_A), and the two standard deviation multiplier (StD2). A corresponding ramp-down occurred through StD1_B, StD0_B, and Normal_B.

MLFreq oscillations exhibited the greatest differences between oscillation types. Within ML Frequency trials (Figure 2.2), SL decreased significantly in StD1_A, StD2, StD1_B and StD0_B compared to both Normal_A and Normal_B. Likewise, a decrease between StD0_A and StD1_A was observed. SW increases were observed between StD1_A, StD2, StD1_B, and StD0_B

compared to Normal_A. Likewise, SW increased significantly in StD1_A and StD1_B compared to StD0_A. In StD1_A, StD2, StD1_B, and StD0_B SW was significantly larger than Normal_B. ML-MoS was significantly larger in StD2, StD1_B, and StD0_B compared to Normal_A; StD2, StD1_B, and StD0_B were significantly larger than Normal_B. AP-MoS was significantly smaller for StD0_A, StD1_A, StD2, StD1_B, StD0_B compared to Normal_A. StD1_A, StD2, StD1_B, and StD0_B were significantly smaller than Normal_B. StD2 was smaller than StD1_A; StD2 was likewise smaller than StD0_B; StD1_B was significantly smaller than StD0_B.

ML Amplitude trials exhibited the fewest significant differences in SL, SW, and ML-MoS compared to all other trials. ML Amplitude trials (Figure 2.2) exhibited significant decreases in SL for StD0_A, StD2, StD1_B, and StD0_B and Normal_B. There was also a difference between StD2 and StD0_B. No significant differences were detected in either SW or ML-MoS during oscillations. AP-MoS was significantly smaller for StD0_A, StD2, and StD1_A compared to Normal_B; StD2 was significantly smaller than Normal_A; and StD2 significantly smaller than StD0_B.

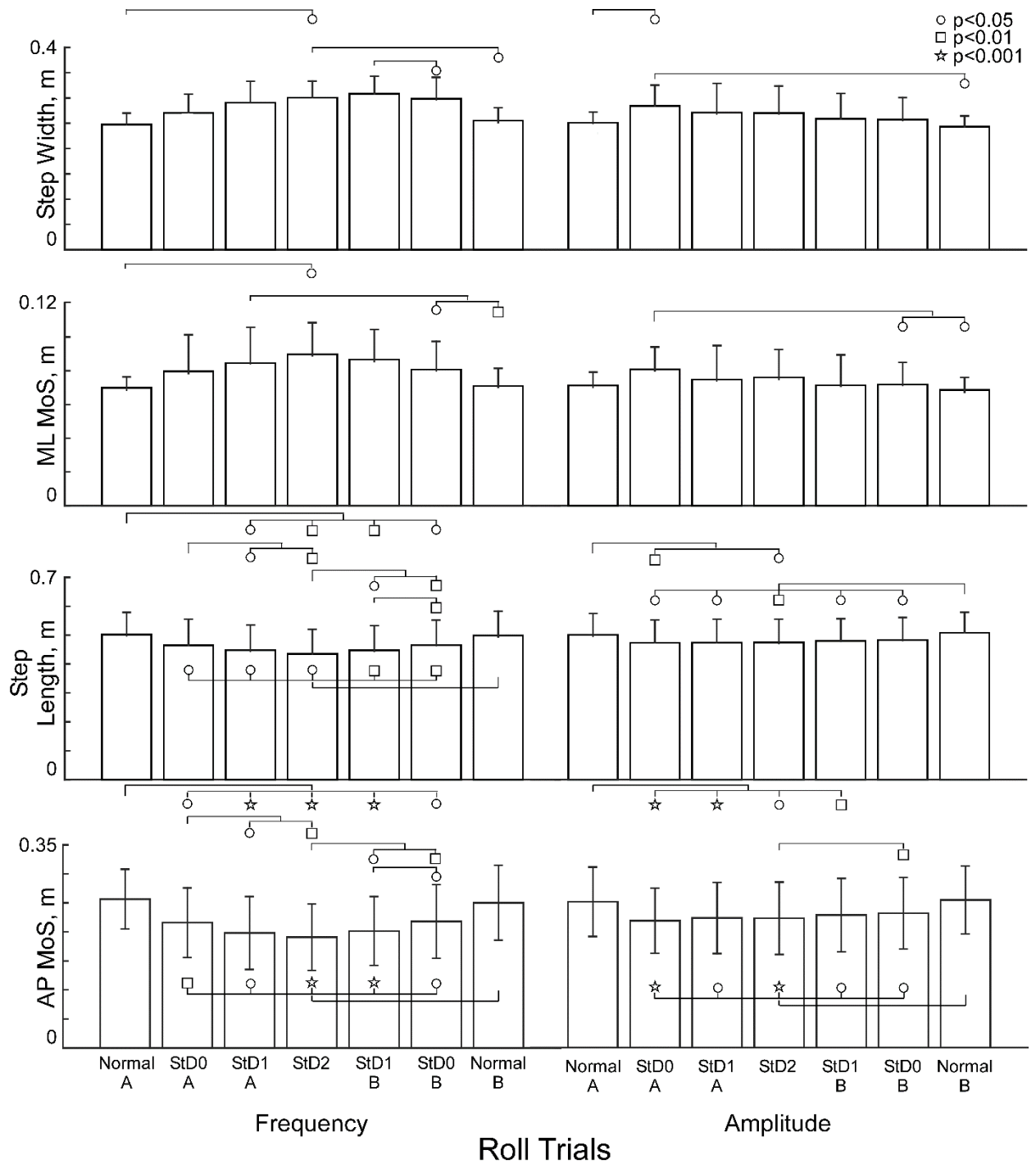


Figure 2.3: Roll Amplitude and Frequency responses over the seven segments of each trial. Trials were subdivided into a beginning round of Normal_A Walking (Normal), the first sinusoid (StD0_A), the one standard deviation multiplier (StD1_A), and the two standard deviation multiplier (StD_2). A corresponding ramp-down occurred through StD1_B, StD0_B, and Normal_B.

RollFreq trials exhibited significant differences in SL, SW, and ML-MoS (Figure 2.3) and demonstrated the most compensation differences in sagittal plane balance. RollFreq exhibited the greatest number of deviations to step length—StD1_A, StD2, StD1_B and StD0_B were significantly smaller than Normal_A. StD1_A and StD2 were shorter than StD0_A. StD1_B and StD0_B were both *longer* than StD2. StD0_B was likewise longer than StD1_B. StD0_A, StD1_A, StD2, StD1_B, and StD0_B were each shorter than Normal_B. For SW, RollFreq exhibited larger values in StD2 compared to Normal_A; StD2 was also larger than Normal_B. There was likewise a significant drop in SW from StD1_B to StD0_B. ML-MoS was significantly larger in StD2 compared to Normal_A; StD1_A was larger than StD0_B and Normal_B as well. AP-MoS was significantly smaller in StD0_A, StD1_A, StD2, StD1_B, and StD0_B were all significantly smaller than Normal_A; StD0_A, StD1_A, StD2, StD1_B, and StD0_B were all significantly smaller than Normal_B; StD1_A and StD2 were smaller than StD0_A; StD1_B and StD0_B were larger than StD2; StD0_B was larger than StD1_B.

RollAmp trials exhibited significant differences in SL, SW, and ML-MoS (Figure 2.3) and—similar to RollFreq—demonstrated greater compensation in sagittal plane balance compared to the frontal plane. StD0_A, and StD2 were shorter than Normal_A in SL. StD0_A, StD1_A, StD2, StD1_B, and StD0_B were each shorter than Normal_B. By contrast, significant increases in SW were only detected between StD0_A compared to Normal_B and StD0_A compared to Normal_A. This was mirrored in ML-MoS: StD0_A was significantly larger than StD0_B and Normal_B. AP-MoS demonstrated smaller values in StD0_A, StD1_A, StD2, StD1_B compared to Normal_A; StD0_A, StD1_A, StD2, StD1_B, and StD0_B were each significantly smaller than Normal_B; StD2 was significantly smaller than StD0_B.

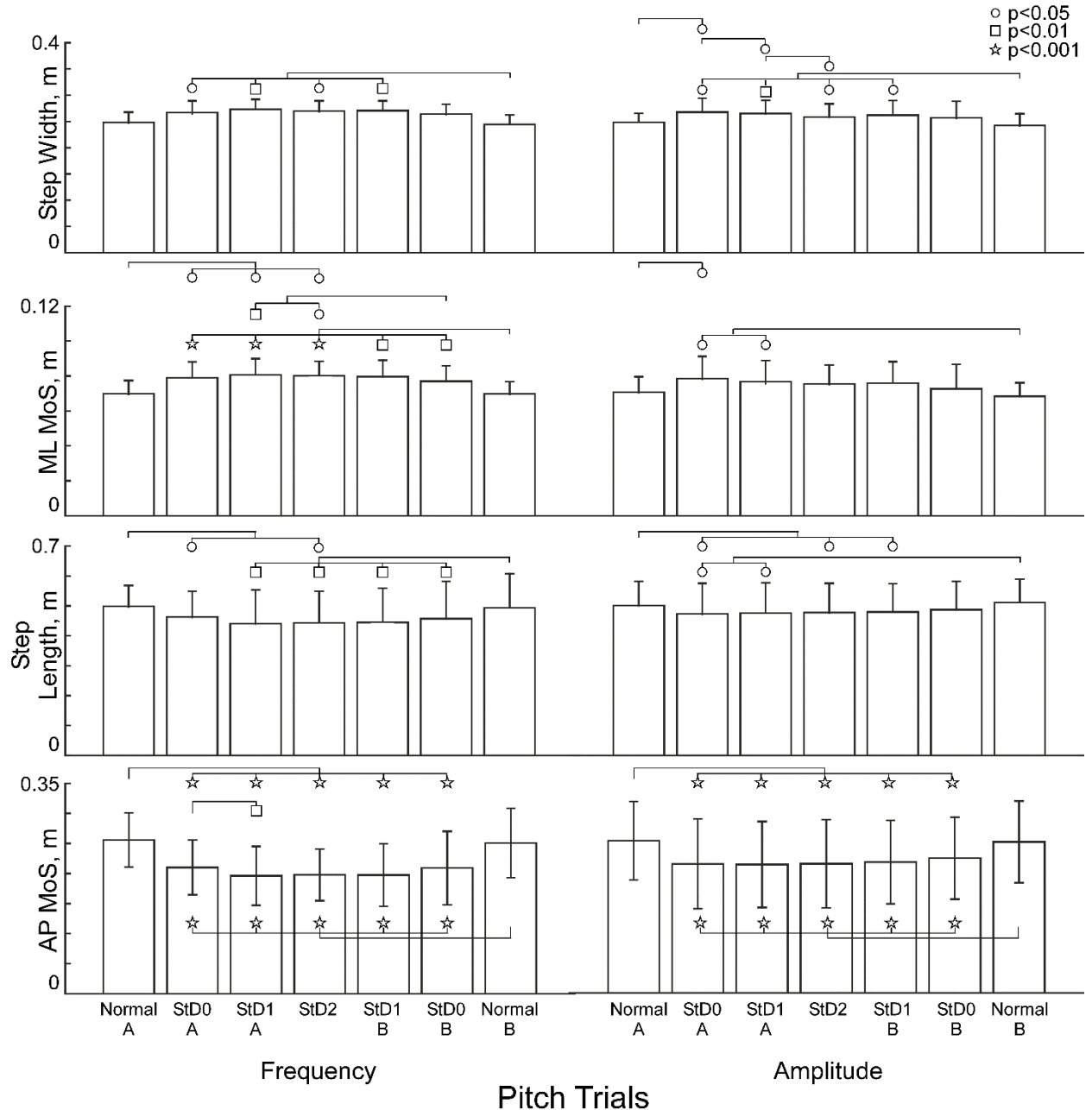


Figure 2.4: Pitch Amplitude and Frequency responses over the seven segments of each trial. Trials were subdivided into a beginning round of Normal_A walking (Normal_A), the first sinusoid (StD0_A), the one standard deviation multiplier (StD1_A), and the two standard deviation multiplier (StD2). A corresponding ramp-down occurred through StD1_B, StD0_B, and Normal_B.

PitchFreq trials demonstrated differences in SL, SW, and ML-MoS but demonstrated greater compensation in the Frontal Plane (Figure 2.4). In PitchFreq, SL was significantly smaller

in Std0_A and Std2 compared to Normal_A; Std1_A, Std2, Std1_B, and Std0_B were all significantly smaller than Normal_B. SW was significantly larger in Std0_A, Std1_A, Std2, and Std1_B compared to Normal_B. PitchFreq demonstrated the greatest number of differences in ML-MoS compared to any other trial: Std0_A, Std1_A, Std2 were larger than Normal_A; Std1_A and Std2 were each larger than Std0_B; Std0_A, Std1_A, Std2, Std1_B, and Std0_B were each larger than Normal_B. Within AP-MoS metrics, Std0_A, Std1_A, Std2, Std1_B, Std0_B were significantly smaller than Normal_A; Std0_A, Std1_A, Std2, Std1_B, Std0_B were significantly smaller than Normal_B; and Std1_A was significantly smaller than Std0_A.

PitchAmp trials demonstrated differences in SL, SW, and ML-MoS and mirrored the results of PitchFreq, demonstrating more compensation in Frontal Plane (Figure 2.4). SL was significantly smaller in Std0_A, Std2, and Std1_B than Normal_A; Std0_A and Std1_A were each smaller than Normal_B. PitchAmp demonstrated sequential changes in SW with an initial increase followed by decrease: Std0_A was larger than Normal_A; Std1_A smaller than Std0_A; Std2 smaller than Std1_A. Std0_A, Std1_A, Std2, and Std1_B were each larger than Normal_B. For ML-MoS, Std0_A was larger than Normal_A; Std0_A and Std1_A were both larger than Normal_B. For AP-MoS, Std0_A, Std1_A, Std2, Std1_B, Std0_B were significantly smaller than Normal_A; Std0_A, Std1_A, Std2, Std1_B, Std0_B were significantly smaller than Normal_B.

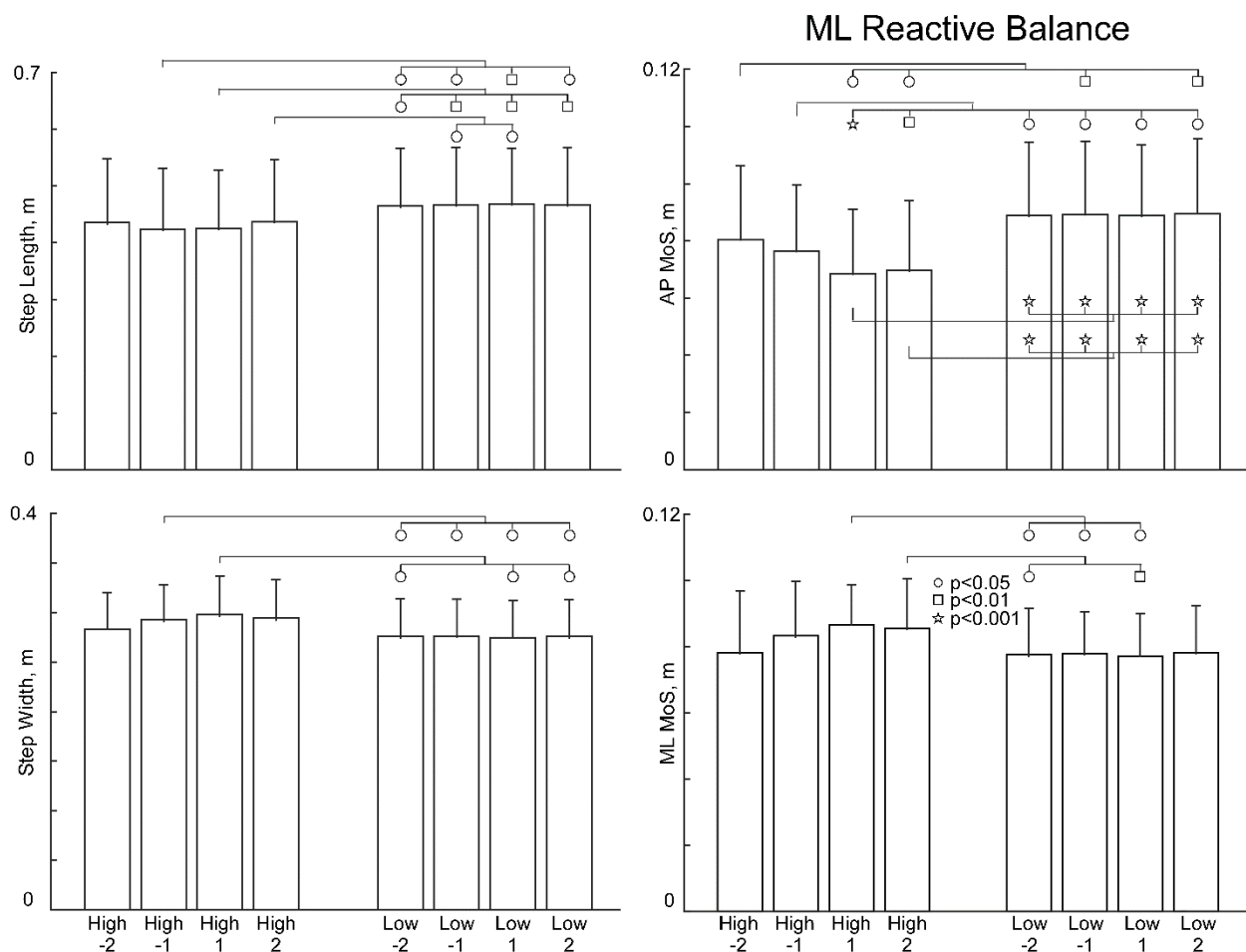


Figure 2.5: Medial-Lateral accelerations consisted of base motions in the Frontal Plane: Roll and ML oscillations. High corresponds to accelerations between 0.5 and 1 m/s^2 , while low corresponds to accelerations between 0.1 and 0.5 m/s^2 . Labels of -2 and -1 correspond with two and one steps before acceleration, while labels of 1 and 2 correspond to one and two steps after the acceleration peak. SL, SW, ML-MoS, and AP-MoS metrics are displayed.

Reactive balance to accelerations in the ML plane—during ML and RollFreq trials—

highlighted the difference between proactive balance strategies present during low acceleration and reactive strategies in response to high acceleration. SL, SW, and ML-MoS during high acceleration exhibited significant differences immediately before and after peak accelerations in data when compared to SL, SW, and ML-MoS during low acceleration. Low -2, Low -1, Low 1, and Low 2 were significantly smaller in SW than High -1; Low -2, Low -1, and Low 2 were significantly smaller in SW than High 1. Step Length was significantly smaller between Low -2, Low -1, Low 1, and Low 2 compared to High -1. Likewise, step length was significantly smaller in

Low -2, Low -1, Low 1, and Low 2 than High 1. Low -1 and Low 1 were significantly smaller than High 2. ML-MoS was significantly smaller in Low -2, Low -1, and Low 1 compared to High 1. Low -2 and Low 1 were likewise smaller than High 2. AP-MoS was significantly smaller in High 1 compared to all Low Accelerations; High 2 was smaller than Low Accelerations; High -2 was significantly larger than High 1 and High 2; High -1 was significantly larger than High 1 and High 2 as well.

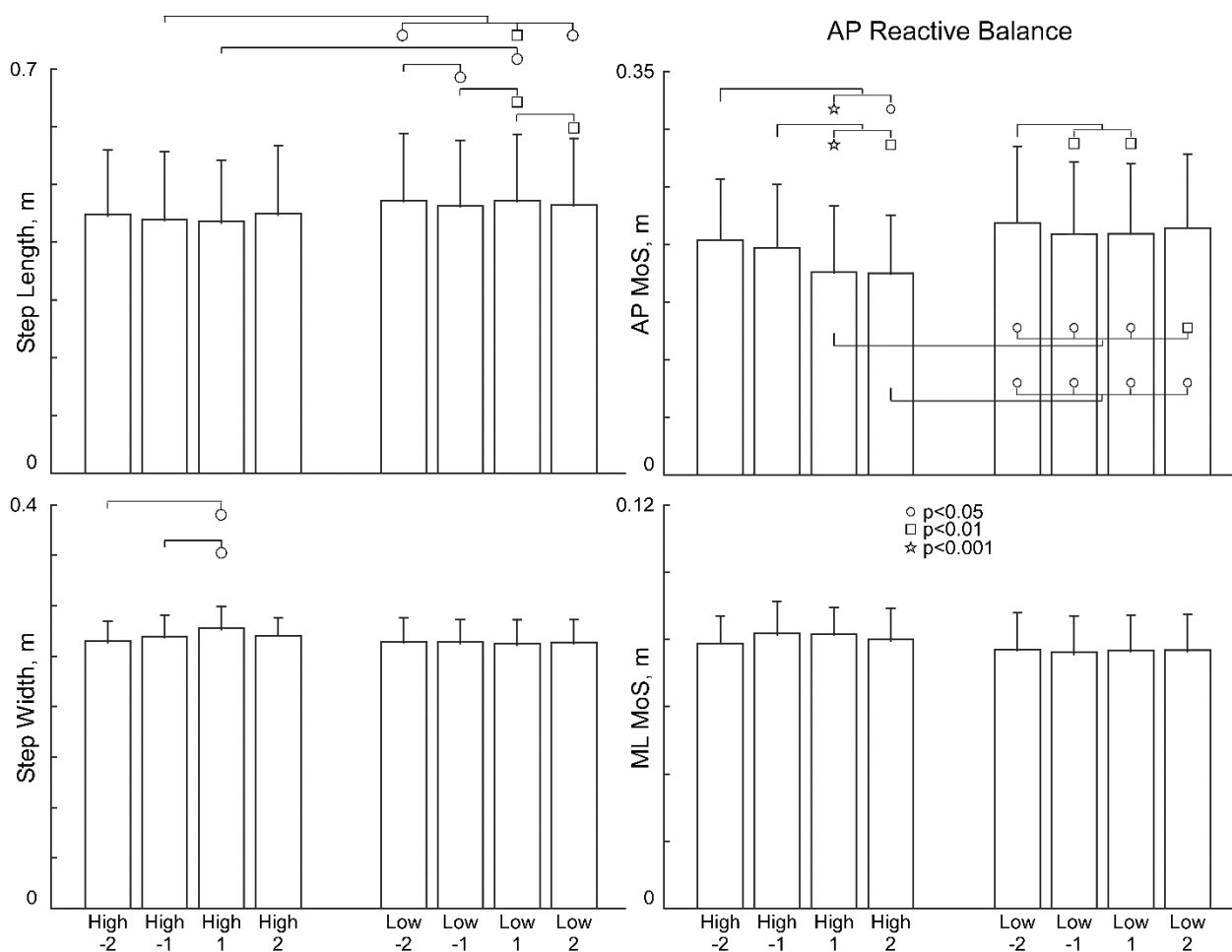


Figure 2.6: Anterior-Posterior (AP) reactive balance for accelerations in the sagittal plane for SL, SW, ML-MoS, and AP-MoS metrics. High corresponds to accelerations between 0.5 and 1 m/s², while low corresponds to accelerations between 0.1 and 0.5 m/s². Labels of -2 and -1 correspond with two and one steps before acceleration, while labels of 1 and 2 correspond to one and two steps after the acceleration peak.

Reactive balance to accelerations in the AP direction exhibited less compensation between high and low accelerations compared to ML direction accelerations; ML compensation techniques (SW and MoS) exhibited no change between high and low accelerations. SW was significantly larger in High 1 compared to High -2. SW was also significantly larger in High 1 compared to High -1. Step Length was significantly lower in High -1 than Low -2, High 1, and High 2. High 1 was significantly smaller than Low 1. Likewise, Low -2 was larger than Low -1. Low -1 was significantly smaller than Low 1. And Low 2 was smaller than Low 1. There were no significant differences between ML-MoS during AP-reactive balance. However, in AP-MoS was significantly smaller in High 1 compared to all low accelerations; High 2 was smaller than low accelerations; High -2 was significantly larger than High 1 and High 2; High -1 was significantly larger than High 1 and High 2 as well.

2.7. Discussion

In this study, we demonstrated that movements of the walking surface produce alterations in step parameters during treadmill stepping, comprised of predictive and reactive components. Movements of the walking surface generally caused a decrease in step length and increase in step width, with greater changes as the predictability of the walking surface movement was reduced. The alterations in step length and width had two components: a constant change associated with a predictive compensation and an acceleration-dependent change associated with a reaction to the movements. Randomization of frequency of oscillations in the frontal plane produced greater changes in gait parameters compared to randomization of amplitude in this study, presumably due to the increased acceleration in these trials. Using sinusoidal oscillations of a walking surface with randomized frequency or amplitude of movement might provide a means for training dynamic balance, enabling the therapist to target both predictive and reactive balance mechanisms.

2.7.1 *Using Walking Surface Movements to Improve Gait Training*

Movements of the walking surface might improve treadmill training of gait by increasing the difficulty of walking therapy. Current approaches to treadmill training utilize techniques such as body weight support [76], [77] or robotic devices [78], [79] that assist treadmill stepping to provide impaired patients access to gait training, effectively making the gait training experience easier. These approaches are effective [80] ; however, there appear to be no benefits of weight-supported treadmill training over standard home-based therapy [80]. Two approaches being investigated that might enhance gait training include 1) increasing the intensity of training and 2) using training approaches that target specific weakness in gait. The use of movements of the walking surface could both increase the intensity and target balance during walking.

Training intensity could be increased with a moving walking surface, which might improve the training effects of treadmill training. Treadmill training works better when physical load is increased: in spinalized rats, those that walked 1000 steps during training sessions were able to more effectively walk at different speeds and loads than those training at 100 steps per day [81]. In humans, applying speed-dependent treadmill training techniques normally reserved for athletes to post-stroke populations yields significantly better step length, cadence, and overground walking speed than traditional walking therapies [82], [83]. The engagement of neuromuscular, cardiopulmonary and metabolic components of exercise might all be important to the training effects associated with high intensity treadmill training [81], [82], [84]. Moving the walking surface creates a more intense exercise experience, engaging active corrections by the patient, which could be beneficial to therapeutic outcomes.

Another potential strategy to improve treadmill training is to implement an element of the training that challenges specific impairments to walking. To improve leg swing during gait, a cable-resistance training system to modify load to the paretic leg during swing improves ankle kinematics, and self-selected and fast walking speeds of post-stroke [79] and spinal cord injury subjects [78]. Further research into this resistance training in SCI subjects revealed that consistent resistance training leads to sustained after-effects in stride length after *resistive* load was applied, but not after *assistive* load was applied. Additional challenge forces overcompensation and more effective outcomes in gait parameters [85], similar to traditional motor adaptation paradigms. Another strategy is to minimize the assistance provided by a robotic trainer, thereby allowing gait training that would not otherwise be possible, yet maximizing the exercise benefits. For example, preliminary work using the REWALK exoskeleton demonstrates greater self-selected and fast walking speeds after an intervention which minimized robotic intervention; minimal corrections to the swing pattern provides better outcomes than full walking actuation [86]. The moving walking surface used in the current study could be used to target balance training during walking. Because balance is an important factor that limits walking in neurologic patients [87], [88], this approach could address unmet needs of current therapeutic approaches.

Improvements to balance confidence through therapeutic intervention are associated with gains in overground walking speed and improved walking kinematics in the elderly populations [89]. Balance confidence is the limiting factor to performance in elderly and neuropathic populations during walking tasks, improving this balance confidence through therapeutic intervention improves walking parameters and results in a better quality of life.

2.7.2 Proactive Balance Compensation

Various amplitude-banded oscillations caused one singular change in proactive stepping parameters relative to normal walking and did not change across the course of the trial. For Pitch (Figure 2.4), Roll (Figure 2.3), and ML (Figure 2.2) amplitude oscillations the spatio-temporal compensation jumped at StD0: SW and ML-MoS tended to increase while SL and AP-MoS tended to decrease. Of important note is the lack of change in either SW or ML-MoS in the ML Amplitude oscillations—preservation of frontal plane dynamics are important in maintaining balance and are often prioritized due to the inherent instability during walking [28]. By contrast, AP-MoS decreases voluntarily to decrease trunk velocity, mirroring work by Hak and Dingwell [3], [66] Instead, it was amplitude oscillations in the Pitch and Roll dimensions which demonstrated the greatest compensation effect. The change from level to non-level planes is likely the cause of this effect, forcing a switch from a purely ankle-based strategy of compensation to the combined ankle-hip strategy detailed in Horak's work on standing platform motion[90].

Frequency-based oscillations caused multiple changes in proactive stepping parameters across different oscillation types, indicating that they are the more effective at inducing balance challenges. PitchFreq trials demonstrated decreases in SL and AP-MoS with corresponding increases in SW and MoS relative to Normal *and* StD0; more variability occurs in the sagittal plane dynamics relative to the frontal plane. RollFreq and MLFreq oscillations demonstrate increasing compensation over the course of the trial, demonstrating less variability and more significant changes in corrective stepping in the sagittal plane.

Comparison of these results to work with scaled pseudo-random signals demonstrates the impact of higher-frequency, quick accelerations on gait instability. Previous work using

scaled pseudorandom signals in the ML plane yielded significant changes in proactive spatio-temporal parameters as the scaling factor increased [3], [62]. ML-MoS also demonstrated a tendency to increase as the scaling factor increased [66]. However, there were no significant changes between scaling factor trials reported within these studies, consistent with our oscillations. Our Amplitude-banded oscillations create similar balance disruption, but maintain mean velocity and acceleration profiles for future application in neuro-motor control response characterization.

2.7.3 Reactive Balance Compensation

Anterior-Posterior accelerations demonstrate fewer significant differences before and after acceleration, demonstrating differences mainly between high and low accelerations (Figures 2.5 and 2.6). SW and AP-MoS demonstrate significant compensation values immediately after a high acceleration. However, Step Length only exhibits difference one step before and one step after the accelerations—i.e. during the length of the perturbation. ML-MoS demonstrates that it is easily controlled proactively rather than requiring more reactive compensation after either a high or low acceleration in the sagittal plane, corroborating work on AP-translations of support surface, which saw minimal changes in MoS [62].

Medial-Lateral accelerations demonstrate pronounced differences between high and low accelerations. SL, SW, ML-MoS, and AP-MoS each exhibit significant differences between low acceleration and high acceleration responses. The ML MoS more thoroughly demonstrates a ramp-up and ramp-down phenomenon in response to these challenges. These are continuous oscillations with interspersed accelerations and exhibit fundamentally different reactions than work shown in AP slip trials where subjects do not modulate gait before reaching the potential fall hazard [91], [92].

Reactive balance compensation is an expected pattern of compensation in healthy young adults, while proactive balance compensation is a technique exhibited in low-balance-confidence populations, such as SCI and trans-tibial amputee patients. Work in dynamic low-force medial-lateral pulls on healthy young adults suggests that neuropathic individuals are more likely to use long-term proactive changes to gait, while healthy young adults are more likely to take short-term reactive changes to gait parameters in order to maintain balance [2]. Likewise, walking across a loose-rock surface by healthy young adults exhibited reactive changes in ML-MoS, increasing the variability of MoS, while trans-tibial amputees tended to take proactive changes in balance control, raising mean ML-MoS to preserve balance on an unstable surface [93]. However, these trials exhibit a layered proactive-reactive response, exhibiting proactive compensation during smaller accelerations, and adding additional reactive compensation during higher accelerations.

This layered proactive-reactive approach blends both feedforward and feedback based balance compensation systems together. In standing control during continuous platform motion, a change in oscillation frequency required four cycles in order to change feedforward adaptation to preserve balance [94]. During this transitional period, feedback-based balance control was used in order to facilitate balance, similar to work by Horak on unpredictable balance [95]. However, when changes in the platform motion are telegraphed to the user, EEG recordings note a predictive response for the event. This study is a walking-based analog to the feedforward predictive cycle—feedforward balance presents as modulations to SW, SL, and AP&ML MoS mean values while further reactive stepping occurs after high- accelerations.

2.7.4 *Study Limitations*

Limitations to this study include problems of comparing challenge across trials. Pitch, Roll, and ML Translation are three different types of oscillation in the base—although Pitch and Roll were equivalent in degrees, the displacement of treadmill was greater in the Pitch oscillation due to the shape of the treadmill itself. The displacement and velocity of the ML plane oscillation was matched to that of the Pitch and Roll, but as it occurs only in one axis movement rather than two, it is difficult to compare the difficulty side-by-side. Because oscillations were continuous, it is difficult to gauge how rapid changes in acceleration impacted gait mechanics. These quick changes were not administered at the same point in the gait cycle and thus limit the ability to assess reactive stepping when compared to other works.

2.8 Conclusions and Future Direction

There are distinct patterns of compensation which characterize changes in amplitude and frequency on during moving base oscillations. Healthy young adult subjects are particularly thrown off balance by sudden changes in acceleration, adopting more conservative spatio-temporal properties over the course of these trials as a whole; demonstrating that changes in signal frequency are more effective at disrupting balance than changes in amplitude. Disruptions in the Frontal plane cause more dramatic changes in temporal parameters overall, indicating that Roll and ML oscillations provide a greater challenge than Pitch balance disruptions.

This work is important in order to more fully understand what causes greater challenges to balance in healthy young adults. Motion base technology is becoming widely adapted across a variety of research and clinical settings, but there has always been a lack of clear guidelines for how to perturb subjects—it was very much dealer's choice. These results demonstrate a sliding

scale of oscillation types which force both proactive and reactive compensatory responses in SL, SW, AP-MoS, and ML-MoS while walking. This becomes a useful tool for using the base to investigate or rehabilitate neurologically impaired populations—chronic stroke can be investigated on a perturbation by perturbation climbing scale to force more compensation over time and strengthen balance in over ground walking.

Chapter 3

Conclusions & Future Directions

3.1 Proactive and Reactive Compensation for Continuous Unpredictable Oscillations

3.1.1 *Brief Summary of Results*

During human walking under difficult conditions, balance compensation occurs in both a proactive and reactive manner during the course of continuous oscillations. Subjects do exhibit an increased mean SW and MoS and reduced SL during the course of trials, but modulate that about that mean. Subjects change these parameters in anticipation of a change in acceleration of the base, proactively modulating on a step-by-step basis towards a more stable method of locomotion, then two-steps after the perturbation moving back toward a more energy efficient gait mechanisms. This proactive stepping mechanic was more prevalent in the frequency perturbations, where acceleration changed frequently, than the amplitude perturbations, where the absolute value of acceleration was constant.

3.1.2 *Step Frequency and Double Limb Support Time Variation and Balance Control*

Although glossed over in the central manuscript, step frequency indicates a significant change over the course of these trials. Step frequency increases uniformly across Amplitude trials to an average value of 1.83 Hz, while ramping upward from 1.73Hz to 1.93Hz during StD1_A and StD2 trials (Figure 3.1). This increased stepping is not indicative of entrainment to the base—subject's steps do not match an integer multiplier the 0.25Hz timing of the base. Significant differences exist between the Normal and Amplitude ($p < 0.01$, Bonferroni Correction of 2), Frequency ($p < 0.001$, Bonferroni Correction of 2). Likewise, there is a significant difference between the Amplitude and Frequency perturbations ($p < 0.001$, Bonferroni Correction factor of 2). This is confirmatory data that Frequency perturbations induce shorter step lengths than

Amplitude perturbations, requiring more steps overall to accommodate to the challenge. However, the fact that these steps occur with a greater frequency does not change the mean double limb support time.

Across all trials, including the normal walking trials, mean double limb support time remains nearly constant at 35% of the gait cycle (Figures 3.2 and 3.3). This is much larger than the 20% of gait cycle which characterizes over ground walking, suggesting an overall subject discomfort and instability with walking on the base. That subjects chose a lower walking speed of 1.00 ± 0.102 m/s compared to an average walking speed of 1.38 m/s for healthy adults demonstrates an initial trepidation with the system and cautionary planning before trials began [96]. Although the mean double limb support time for each oscillation time is essentially the same, the Coefficient of Variation for each oscillation varies significantly between perturbations. Variation increases from Normal to StD0_A to StD1_A to StD2 during Pitch, Roll, and Y Frequency perturbations. This increased variation in double limb support time is the result of greater reactive stepping rather than proactive stepping.

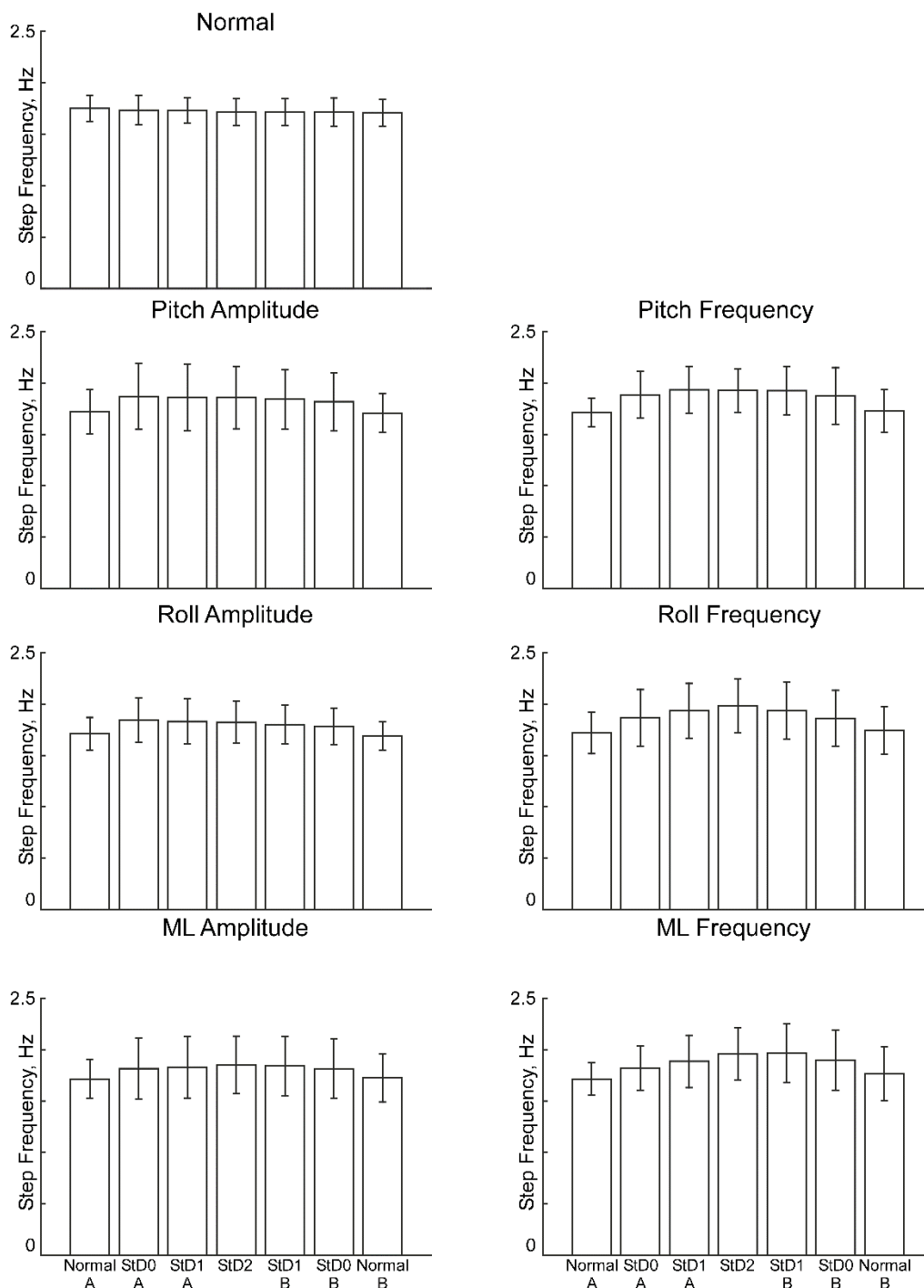


Figure 3.1 Step Frequency in Hz was measured as the inverse of time between each step for each oscillation during the seven trials. ML and Roll Frequency trials exhibited the greatest change, mirroring the changes in SL.

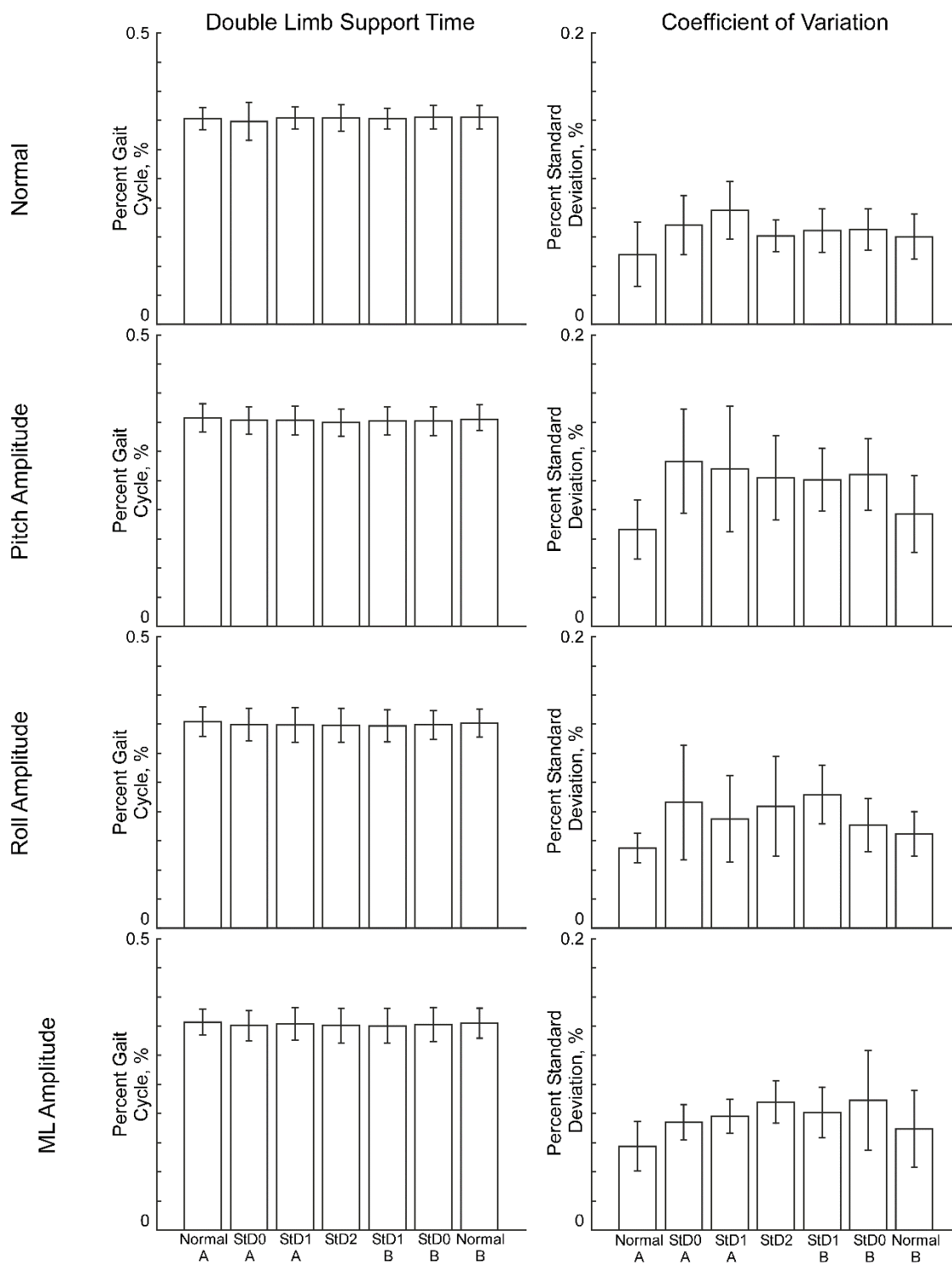


Figure 3.2: Double Limb Support Time was measured as the time between leading heel strike and trailing toe off as a percentage of standard gait cycle. Coefficient of variation was calculated as the Standard Deviation of the oscillation divided by the mean of the oscillation double-limb support time. Presented are the changes in double limb support time proper and the coefficient of variation for the Normal, PitchAmp, RollAmp, and MLamp Trials.

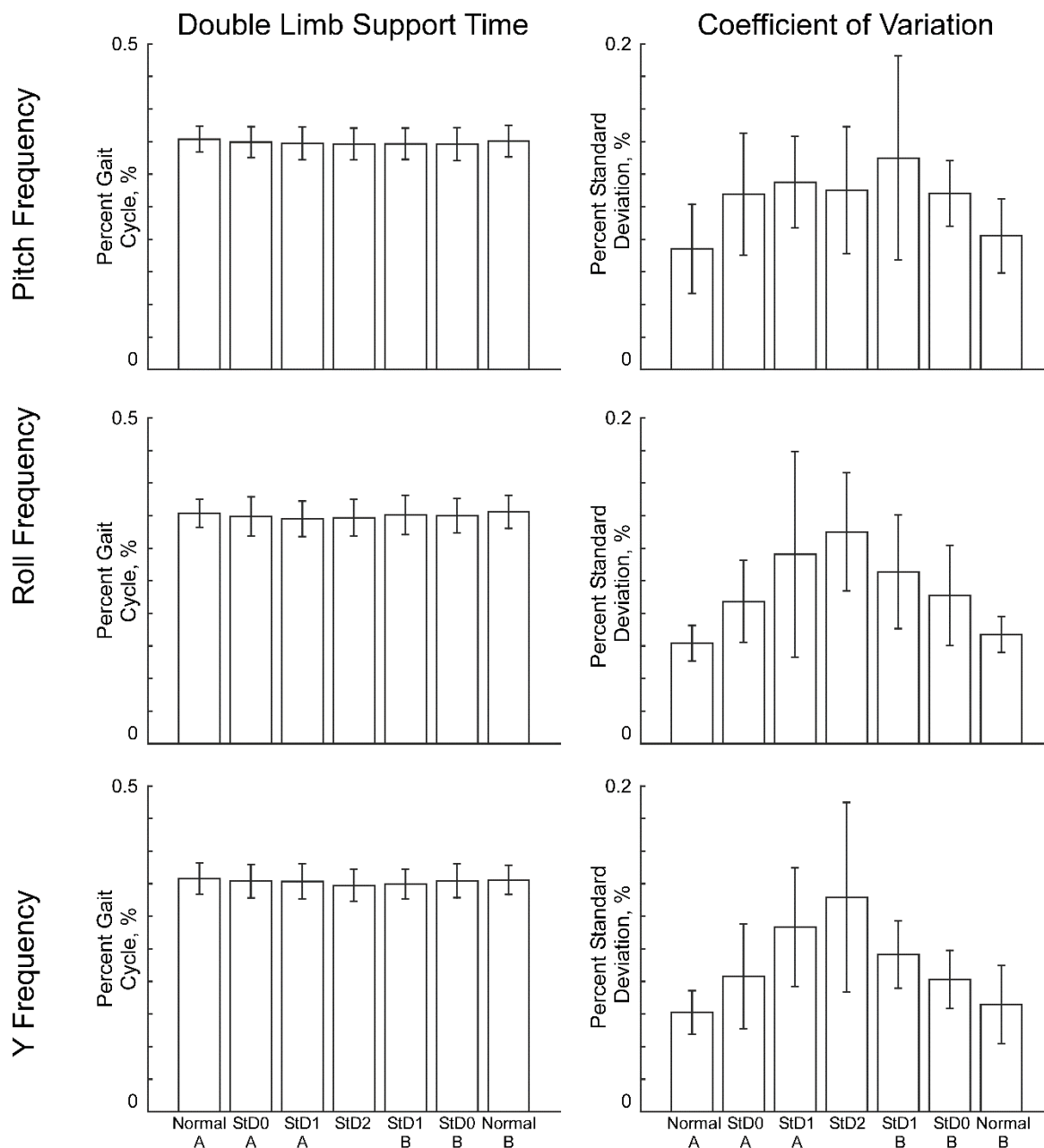


Figure 3.3: Double Limb Support Time was measured as the time between leading heel strike and trailing toe off as a percentage of standard gait cycle. Coefficient of variation was calculated as the Standard Deviation of the oscillation divided by the mean of the oscillation double-limb support time. Presented are the changes in double limb support time proper and the coefficient of variation for the PitchFreq, RollFreq, and MLFreq Trials.

3.2 The Effect of Added Visual Feedback on Walking Ability

A lack of visual feedback negatively impacts subject's ability to predict and compensate for future perturbations of the base—adding visual feedback to this system could either limit the reactive stepping potential or enhance it over time. Visual feedback allows participants to “see down the path” and predict future obstacles, which should allow for greater on-line modification of control parameters rather than triggering purely reactive stepping. However, visual flow information could also be exploited to enhance perturbation difficulty by encouraging complicated stepping patterns or increasing CoM sway.

3.2.1 The Effect of Medial-Lateral Visual Flow on Instability in Unperturbed Treadmill Walking

Work with simple over ground treadmills and visual flow information demonstrate that subjects adapt SW and SL to ML visual-flow perturbations [97]. These visual perturbations were applied at amplitudes of 20, 35, and 50 cm as a sum of three sine waves, similar to work done on pseudorandom perturbations in treadmill walking. Adaptation occurs at the onset of perturbation in the form of increased SW and decreased SL; both metrics exhibited an increase in variation as well. Over the course of the eight-minute trial subjects regressed toward unperturbed levels of walking, suggesting more reactive balance control. However, when pseudorandom oscillation is paired with treadmill walking researchers do not report a regression towards normal walking patterns over the course of 2 minute trials [3], [66]. This suggests that paired visual-feedback to base motion may provide a two-fold method of balance disruption, with visual-motor disruption increasing the disruption input by the base.

Mechanistically, disruption of balance by visual flow can be explained by entrainment of CoM sway to flow speed. Fran et al. again explored ML visual flow perturbations during treadmill walking, perturbing healthy young adult subjects using five different carrier

frequencies: 0.155 Hz, 0.31 Hz, 0.62 Hz, 0.93 Hz, and 1.24 Hz. CoM sway amplification was three times larger when carrying frequency matched to average stride frequency at 0.93 Hz [97].

Because CoM sway is associated with a greater need for spatio-temporal stepping measurement to correct and control balance during walking, matching the ML perturbation to stepping frequency can provide the greatest level of balance disruption while walking in healthy adults. Matching this paradigm dynamically to walking speed and frequency modulated by a moving treadmill could amplify this effect, forcing subjects to greater levels of balance compensation than unpaired combinations.

3.2.2 *The Effect of Visual Flow Information on Instability in Perturbed Treadmill Walking*

Very little work has been done on utilizing visual flow disruption in conjunction with motion base oscillation or perturbation: often, the visual flow presented matches the motion of the base. Pseudorandom oscillations bound between $\pm 10\text{cm}$ of ML perturbation have been studied and found to influence SW, SL, and MoS but have a marginal effect on self-selected walking speed [3]. Visual flow has been studied at 10x greater disruption in tandem with base disruption: this perturbation type elicits greater SW and SL variability and proactive step compensation that likewise characterizes work in this study [62]. However, the effect of inverting or time-delaying visual feedback when paired with base motion has yet to be studied and represents a rich vein of future material.

Visual cueing in the form of stepping stones and visual obstacles have been studied to disrupt balance. The CAREN system has been used to generate a flow of stepping stones which exhibit dramatic shifts and force new foot placement [98]. This study's major limitation is the ability of the subject to look down at the stones for stepping—visual flow information ends at the beginning of the treadmill in the CAREN environment. This could be rectified by conducting

experiments in new fully-immersive VR headsets, giving subjects the full freedom to look down and more effectively hit the targets. This methodology could be applied to ataxic individuals, such as persons with M.S., to induce greater balance challenges and train compensatory strategies more effectively. Altering the time delay of this flow information could likewise be used to reduce intention tremor, similar to the work currently focusing on upper-limb motion [99].

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Appendix A—Motion Base Communication Protocol

The Motion Base is controlled via one Host computer with a custom LabVIEW Program. This program sends command data from the Host Computer to the Motion Base Computer. The Motion Base Computer then interprets this data and sends it to the base via Fiber Optic Cable, see Figure A.1.

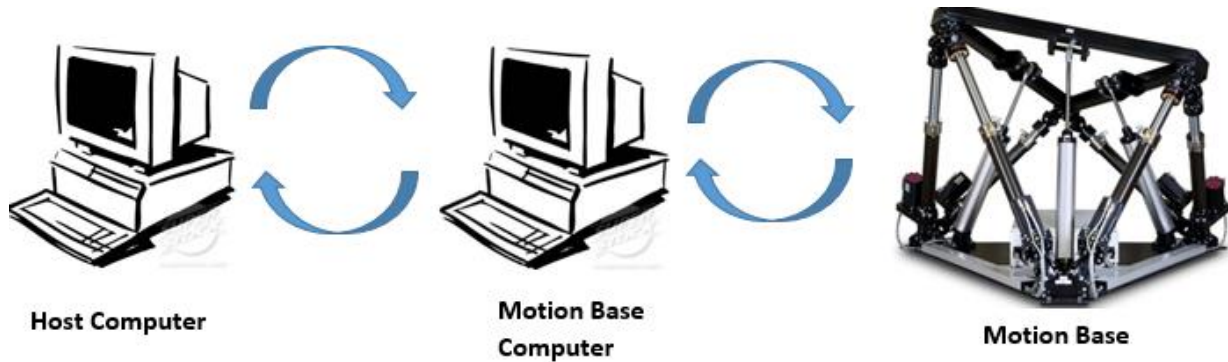


Figure A.1: The Host Computer is linked to the Motion Base Computer via an Ethernet Connection. Static Port 10.1.10.1 is set to the Host, Static Port 10.1.10.2 is set to the Motion Base Computer. Host Computer Subnet Mask is static at 255.255.255.0.

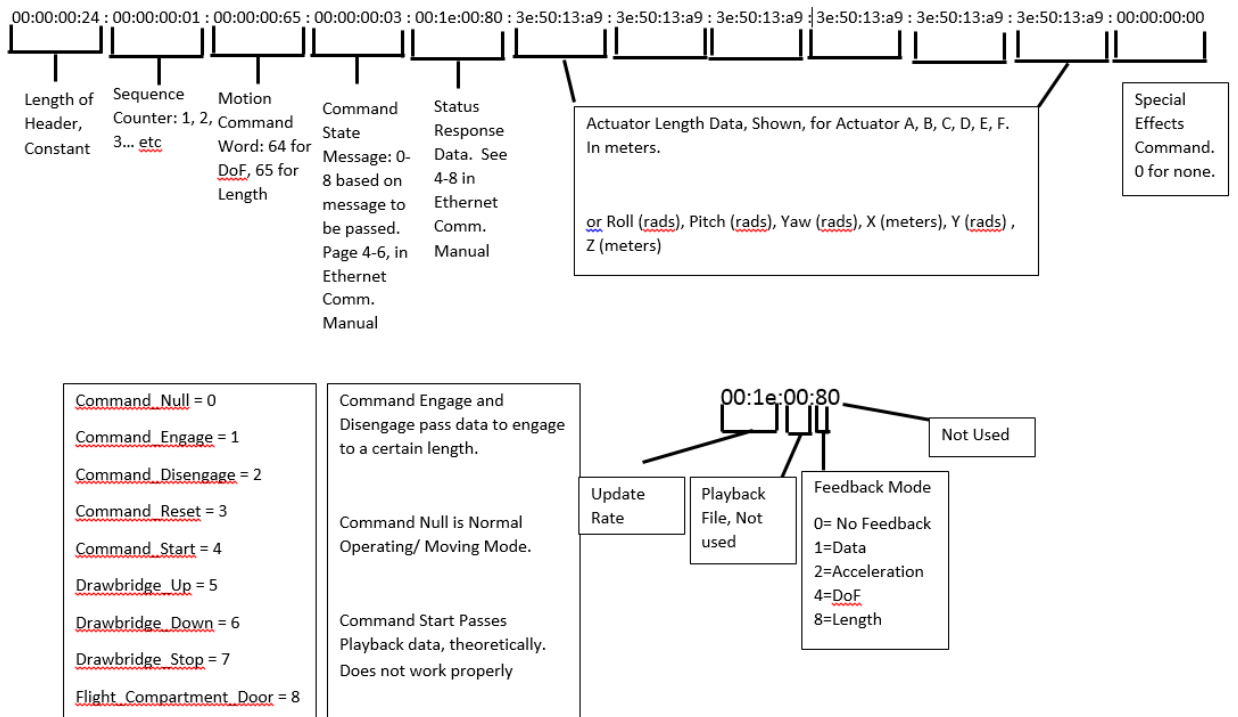


Figure A.2: The Host Computer message system Consists of several hexadecimal byte strings stitched together to denote the type of Motion, either Length or Degrees of Freedom, and

command message to Engage, Disengage, etc, as well as either Actuator Length or DoF Location Data.

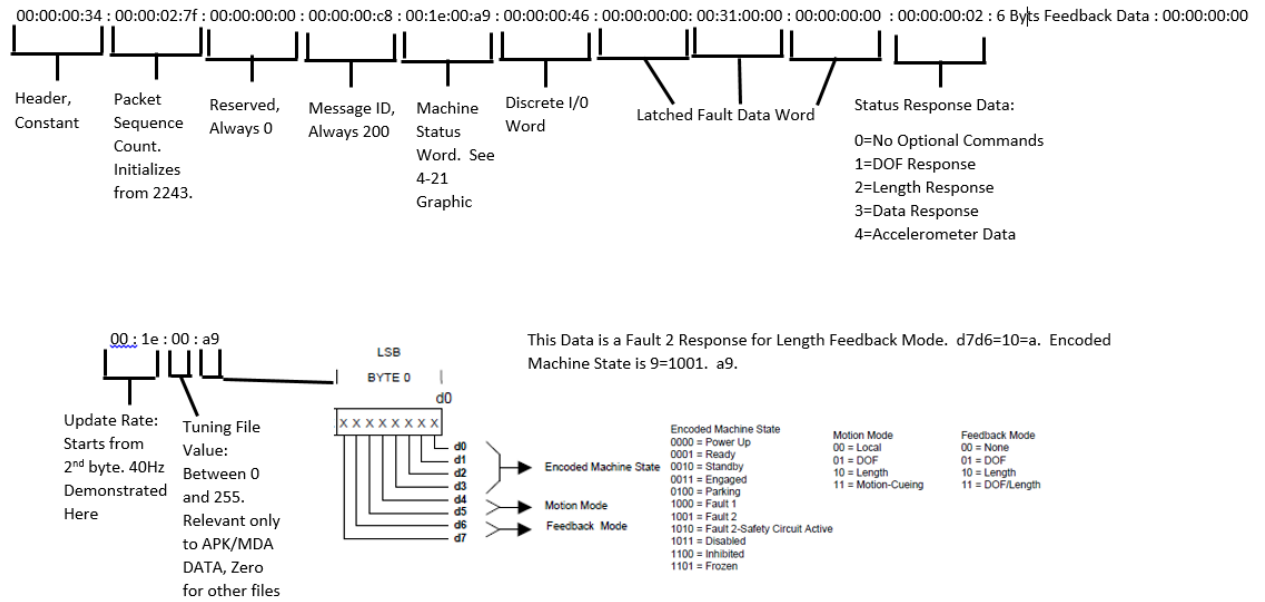
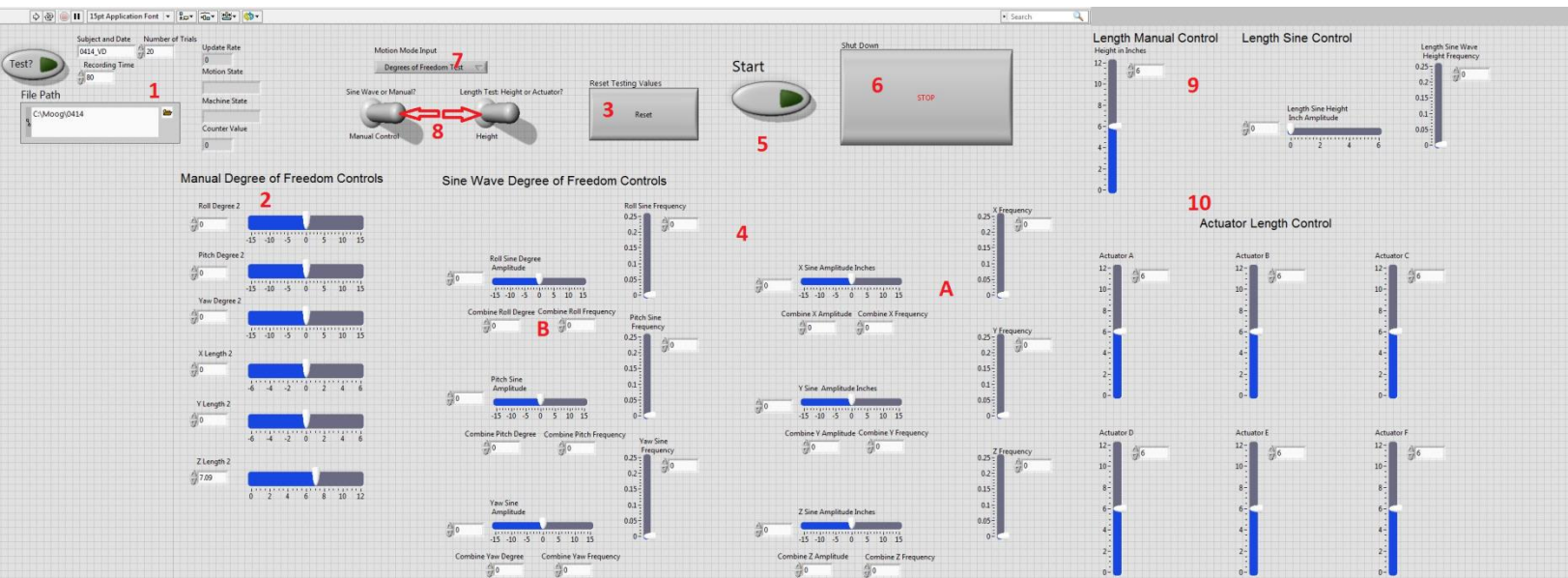


Figure A.3: The Motion Base Computer sends messages back to the Host Computer in Hexadecimal code.

Appendix B—Motion Base LabVIEW Controller Overview

The Current motion base control file allows for multiple different types of controls: a degree of freedom mode, length-of-actuator mode, File Replay Mode, and PsuedoRandom Mode. Each of these modes has two options: manual (base moves to position and stops) or sine-wave (base moves continuously through a sine wave).

The base can change between these modes by toggling the switches and hitting the 'RESET' button (3). Without clicking reset, the mode remains the same.



Testing SubVI cluster, used to name text files and used during CoP and EMG recordings

1. Manual Degree of Freedom Controls, used in manual mode to move the base to a static position in one of six degrees of freedom.
2. Reset button, used to change between different controls
3. Sine Wave Degree of Freedom controls, used to move the base in a sine wave fashion.
 - a. Control type A moves the base immediately after change, while B has a push button delay.
 - b. Type B allows for multiple sines to be input at once using (5).
4. Start button, used to pass multiple inputs from Sine Wave DoF Type B controls. Only functions when controller is in DoF Sine Wave Control.
5. Stop button. Provides an emergency stop to base movements if subject slips, trips, or is otherwise disturbed. Can be used at any time during testing.
6. Case Selector selects Degree of Freedom, length, or base disengage. To cycle between them, choose a new one and click 'RESET'.

7. Switches, left is used for controlling Sine Wave or Manual for both degree of freedom and length tests. Right controls whether Length test is a total-height of treadmill test or if manual control of actuators is requested.
8. Length test, controls Z length of treadmill in either manual or sine wave.
9. Individual actuator lengths, controls lengths of actuators when selected using both (7) and (8)
10. Feedback panel which displays current motion state.

Degree of Freedom Manual Control: (7) must be set to “Degree of Freedom Test” and the switch set to manual. This mode allows for static inputs of one degree of freedom by sliding the bar or entering the degree. The maximum degree of freedom input combination without interfering is 8 degrees for Roll, Pitch, and Yaw.

Degree of Freedom Sine Wave Control: (7) MUST BE SET TO “Degree of Freedom Test” and the switch set to Sine Wave. The maximum single degree of freedom is +/- 15 degrees for Roll, Pitch, and Yaw. The maximum single degree of movement is 9 inches for X and Y, 5 inches for Z. Maximum Frequency for these movements is .2 Hz

Maximum combined Roll, Pitch, and Yaw is 8 degrees each in combination. Maximum combined X, Y, and Z is 5 inches each. Negative inputs move the base in a different direction upon initial movement, creating a negative sine wave.

To use the A control, input a value to both the amplitude and frequency bars. As soon as both bars are filled, the base will move. Currently, the base is programmed to displace a random point on the sine curve and continue the motion.

To use the B control, input both values and click start. Multiple values can be input into multiple different blocks and the system can be “started” together by clicking (5).

Length Manual/Sine Wave Control: After selecting Length Test in (7) and either manual or Sine Wave control on the switch 1 and Height on Switch 2, this function adjusts the base height in either a static or sine wave.

Length of Actuator Control: (7) Set to length test, Switch 1 set to manual, and Switch 2 set to length of actuator. This allows for adjustment of the lengths of each actuator. Serves no practical purpose, but is a neat party trick.

Replay Mode—DoF or Length: (7) Change the dropdown menu to either Replay Mode—DoF or Replay Mode—Length. Under the respective Replay Mode tab, the file path is set to where the pre-generated replay files are stored. When launching the base, all the data will be read in before Engaging. Set the length of the file to replay in seconds. File to run begins at 0, set the file and click Start Replay to begin the replay of the file. Unclick the button to stop replay.

DoF replay files are formatted for Pitch (radians), Roll (radians), Yaw (radians), X (meters), Y (radians), and Z (meters) with 6 columns for each data stream to send. Length replay files are

formatted for Acuator A (meters), B (meters), C (meters), D (meters), E (meters), and F (meters) with 6 columns for each data stream to send.

Randomized DoF: (7) Change the dropdown menu to Randomized DoF and click reset. This allows for base movements following the ML pseudorandom motion but builds it for Pitch, Roll, Yaw, X, or Y motions (Hak 2012) [3].