The Effects of Varying Surface Translation Acceleration and Velocity on Compensatory Forward Stepping Responses

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Abstract

Introduction: Examination of balance control has often been accomplished via evocation of stepping responses through external perturbations. These external perturbations can take the form of sudden underfoot surface translations, which are often comprised of controllable parameters including acceleration, velocity, and displacement. Interestingly, the values of these parameters incorporated within surface translation perturbations vary substantially within the literature. While this variance is understandable based on researchers' research questions and infrastructure capacity, the systematic effect of perturbation characteristics on balance control responses during backwards surface translations is relatively understudied. Accordingly, the goal of this thesis was to improve the understanding how sudden posterior surface translation parameters affect spatial metrics of stepping responses and stability (study one) and to explore the effects of trial specific pre-perturbation participant-specific conditions on the same measures of balance control (study two).

Methods: Twenty-four young healthy adults (mean (SD): age 24.0 (3.61) years; height 1.71 (0.08) m; mass 73.2 (12.5) kg) with no history of balance impairment, recent musculoskeletal injury, or neurological disorder participated in the studies. Surface translations were initiated randomly during quiet stance in one of four directions (backward, forward, left, right). Platform acceleration values were varied from 1.0-3.5 m/s^2 (increments of 0.5 m/s^2) while two platform peak velocity values (low and high) were implemented at each acceleration level. Displacement (0.30 m) and deceleration (5.0 m/s^2) values were held constant across all perturbations. Backward translations (forward losses of balance) as well as single step responses were the focus of this thesis and thus the only trials analyzed. Dependent variables of normalized step length, maximum anteroposterior (AP) extrapolated centre of mass displacement (xCOM), and minimum AP extrapolated margin of stability (xMOS) were extracted from the kinematic data. Trial specific pre-perturbation values of underfoot weight distribution, AP centre of pressure (COP) location, ankle co-contraction index (CCI), AP COM location, AP COM velocity, and AP COM acceleration were extracted. In study 1, analysis of variance was used to analyze the effects of platform acceleration and velocity on the three dependent variables. In study 2, repeated measures stepwise linear regression was used to analyze the effects of the pre-perturbation factors on the predictive capacity of models predicting normalized step length and minimum AP xMOS.

Results: Study one demonstrated that increasing platform acceleration resulted in increased normalized step length and increased minimum xMOS (up to 30.7% and 90.4%, respectively), but only during high peak velocity trials. Increased platform velocity was also found to increase normalized step length and minimum xMOS by up to 26.8% and 127.6%, respectively. In contrast, participants' xCOM displacement demonstrated a max increase of only 9.2% across acceleration levels. Study two identified both AP COM and COP position prior to perturbation as being the most commonly statistically relevant factors across perturbations. In comparison to models that incorporated variables accounting for the repeated measures within participants and external platform perturbation characteristics, participant factors at the moment of perturbation onset only increased model adjusted r^2 values from 0.612 to 0.646 (low velocity trials) and 0.661 to 0.689 (high velocity trials) for normalized step length. Minimum xMOS adjusted r^2 values were increased from 0.375 to 0.419 (low velocity trials) and 0.466 to 0.507 (high velocity trials).

Discussion/Conclusion: Variation in platform parameters resulted in significant changes to measures of step length and minimum xMOS. The increase in overall perturbation magnitude resulted in theoretically more stable responses (increased minimum xMOS) which was driven by the increased step length. As the external surface translation parameters are such important drivers of dynamic stepping responses, their effects should be considered when comparing studies which utilize different perturbation parameters. The statistically significant associations between personal pre-perturbation factors (particularly AP COM and COP location) on step length and xMOS align with mechanical models which suggest they play important roles in balance control. Interestingly though, these pre-perturbation factors explained only a small degree of variance beyond that provided by factors such as repeated measures and external perturbation characteristics. These two studies provide insights for researchers to more appropriately compare previous literature as well as provide recommendations for future study design during sudden support surface translations.

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Table of Contents

List of Figures

List of Tables

General Introduction & Literature Review

1.1 Introduction

Humans' capacity for bipedalism is one of the defining features that separates them from nearly all other species on Earth. Although bipedalism has advantages, it also makes humans characteristically unstable as it raises the location of the whole body centre of mass (COM) and reduces the number of limbs used to generate the base of support (BOS) (Winter et al., 1990). This greatly increases the risk of losing balance which ultimately can result in falls.

Falls are of great concern, especially with Canada's aging population. In 2004, falls and fall related injuries accounted for $6.2 billion or 31% of the national injury costs (Smartrisk, 2009). Falls were the leading cause of hospitalization and the third leading cause of injury death (Smartrisk, 2009). The cost of falls continued to rise in years since then with a report showing that in 2010 the cost of falls and fall related injuries rose to $8.7 billion or 32% of the national injury costs. Falls remained as the leading cause of hospitalization and rose from the third leading cause of injury death to the number one cause of injury death in Canada (Parachute, 2015). This severe, and evidently growing, impact on the Canadian economy and health system raises the importance of addressing falls as a nation.

Addressing falls and their associated side effects should be a national goal and this thesis aims to improve researchers' capabilities of testing individuals balance responses. By improving understanding of the relationship between testing methods and outcome measures, identification of differences between populations of interest could potentially highlight the detriments in balance control that place these populations at higher risk of falls and fall related injuries.

1.2 Literature Review

1.2.1 Maintenance of Balance

Balance is often referred to as an individual's ability to maintain their centre of mass within their base of support. In static scenarios this seems like a relatively simple task, as the BOS is well established with an individual's feet maintaining a constant BOS. Centre of gravity (COG), COM without the vertical component, is tightly controlled within the BOS and therefore proper balance is achieved. This tight control is maintained using muscular contractions to manipulate the centre of pressure (COP). COP is the single point representation of force from all of the muscular outputs (Winter et al., 1990). This value is used as a point force with a moment arm length from the ankle axis of rotation to determine the net ankle moment, which is then able to directly affect the angular acceleration of the body about the axis of rotation (the ankle) (Winter et al., 1990). Essentially, the COP oscillates about the COM to continually prevent it from moving outside of the BOS. Figure 1-1 demonstrates the continual relationship between COM and COP over the course of five time points. In Figure 1-1, COM is depicted by the solid black dot located in the torso of the image, ankle point of rotation is denoted by the hallow dot located between the foot and shank, angular velocity and acceleration are represented with ω and α respectively. W represents the body weight of the individual and is equal and opposite of R which is the vertical ground reaction force. The variables g and p represent the moment arm lengths of W and R respectively. This process is continually performed during static balance. COP is controlled via muscular activation and this allows control over the location of the ground reaction force. This is how muscular activation allows for proper balance to be achieved and maintained in a static scenario.

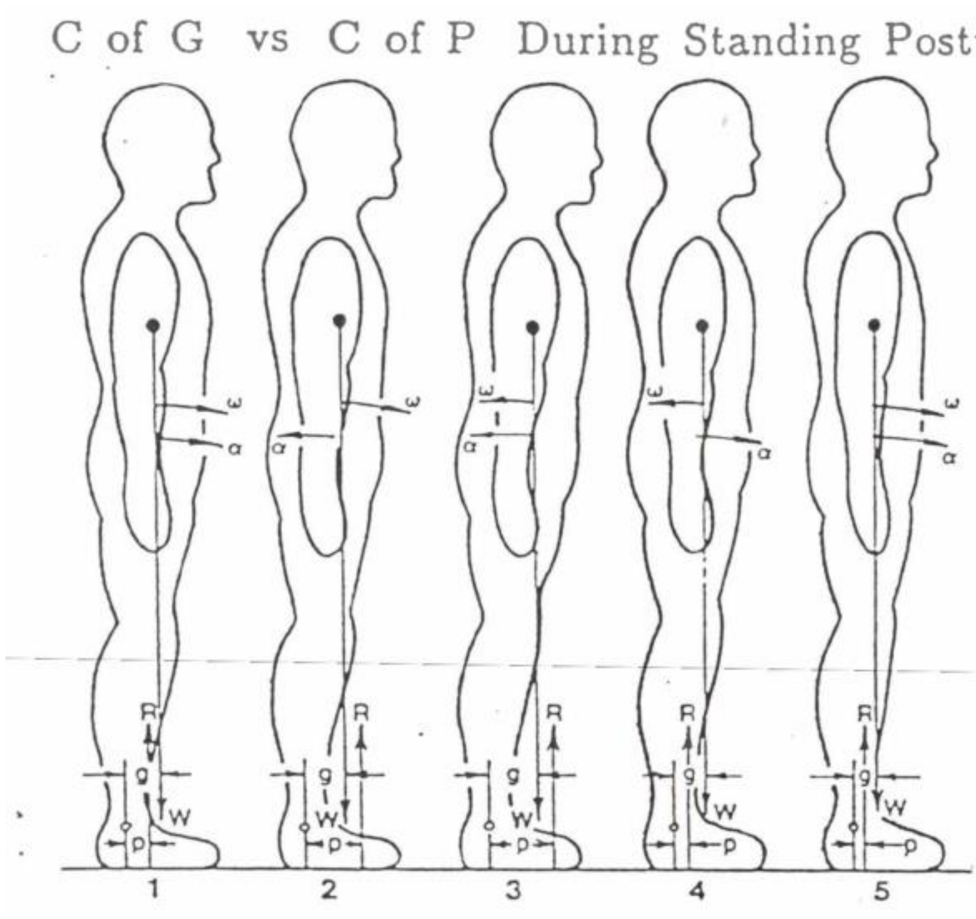

Figure 0-1: Regulation of centre of gravity (centre of mass) with the base of support by manipulation of the centre of pressure affecting the angular acceleration and velocity of the body about the axis of the ankles (Winter et al., 1990)**.**

The example shown in Figure 1-1 is based on the inverted pendulum concept, which assumes the body behaves like a rigid mass rotating about the ankle joint (Winter et al., 1990). This is a common simplification of the complex human system when assessing static balance. However, this system is challenged when dynamic tasks are performed or when either the COM or BOS are perturbed.

When an individual's balance is challenged and a postural response is required there are two common strategies adopted; change-in support or fixed-support strategies. Determination of postural strategy is based upon several factors including the size of the perturbation (McIlroy and Maki, 1993) as well as the individuals balance capabilities. Older adults tend to rely on change-in support strategies when experiencing a loss of balance. This was noted by Yang and colleagues when 42% of falls were accompanied by attempts to take compensatory or reactive steps (Yang et al., 2013). This furthers the concept that the postural response exhibited by an individual varies between subjects however, due to the increased prevalence of change-in support strategies, they will be the focus of this thesis.

Change-in support strategies are employed when the perturbed individual adjusts or manipulates their BOS to increase their BOS. By increasing the BOS, larger deviation of the COM is allowed while still maintaining it within the BOS. This strategy has been examined via grasping handrail supports or other external environmental objects(Allum et al., 2002; Bateni et al., 2004; Ghafouri et al., 2004; Sarraf et al., 2014) as well as reactive or compensatory stepping (Maki et al., 2000; McIlroy and Maki, 1996, 1993; Singer et al., 2016; Tripp et al., 2004).

Grasping is one method of increasing the BOS in an attempt to maintain or recover balance (Allum et al., 2002; Ghafouri et al., 2004; Maki and McIlroy, 2006). This strategy attempts to utilize the environment surrounding the individual to increase the BOS and therefore keep the COM with the BOS (Maki and McIlroy, 2006). Grasping of handrails or other objects is an area of focus especially as an intervention for older adults whose stepping responses may be impaired (Allum et al., 2002). As fall risk increases in older adults there is a greater need for fall prevention and modifying the external environment is a feasible solution. Studies have focused on the speed of arm movement initiation (Allum et al., 2002; Ghafouri et al., 2004), grasping inhibition due to physical interference (Bateni et al., 2004), training effects on grasping contact time (Mansfield et al., 2010) and hand forces (Sarraf et al., 2014).

Reactive or compensatory stepping is a dynamic balance response that is commonly adopted when participants are exposed to external perturbations (Maki et al., 2000; McIlroy and Maki, 1993; Tripp et al., 2004). Reactive stepping has been observed and studied in several dynamic perturbation paradigms and is generally accepted as a primary response to a perturbation. Although change-in-support strategies were originally thought to be last resort balance recovery methods to fixed support strategies (Horak and Nashner, 1986), it has been found that these strategies are often employed even when they may not be physically warranted by the perturbation (Maki and Mcilroy, 1997). It has even been noted that reactive steps will be taken even when the perturbation is of smaller magnitude and a step may not be physically warranted if the participants are not constrained (McIlroy and Maki, 1993). This has led to further exploration of these response mechanisms using various perturbation techniques.

1.2.2 Techniques to Study Dynamic Balance Control

Reactive balance control is often studied by examining individuals' responses to external perturbations. The majority of these external perturbations come in the form of either a tether release, waist attached cable pull or surface translation. These three methods are most commonly used and each have their own benefits (and drawbacks).

1.2.2.1 Tether Release

A tether release paradigm requires participants to lean against a harness passively (which will later be released) in an attempt to simulate a trip. The extent of lean is typically standardized by using surrogate measures such as angle of lean (Tripp et al., 2004) or a direct measure of harness force via percentage of the participant's mass on a force transducer (Singer et al., 2016). Varying the magnitude of force through the force transducer allows for different responses to be studied such as single step or multiple step (Singer et al., 2016; Thelen et al., 1997; Wojcik et al., 1999). A tether in series with a magnet or mechanical trigger is used to support the participant in the leaned position and the researcher has control over the magnet to disengage it as desired. This paradigm simulates a tripping scenario as the participant has their COM ahead of their feet so that it simulates catching the swing foot on an object. EMG can be used to ensure the participant is relying on the harness during the lean phase and not actively trying to control their posture (Singer et al., 2016). It is also desirable to ensure that the force distribution between feet is similar so that participants are not anticipating releases and preloading one leg. COP can also be used alongside the equal force distribution to improve repeatability and consistency between trials (Singer et al., 2016).

1.2.2.2 Cable Pulls

Cable pulls induce COM perturbations with the perturbation point of application being the participants' waist (Rogers et al., 2003, 2001; Schulz et al., 2006). Participants can be set up with a ring structure around their waist, which has various cables attached to it. Other methods of cable attachment consist of wearing padded belts around the participants waist (Schulz et al., 2005). A potential perturbation set up could consist of four cables that correspond with anterior, posterior, left and right cable pulls. In a region where the participant cannot see, a weight is attached to a cable that will correspond to one of the four positions on the ring. A magnet is often used to support the weight and the magnet can be disengaged whenever the researcher desires. This causes the weight to fall and pull the subject in one direction. By pulling the subject in a direction, their COM is rapidly accelerated in that direction and postural responses are required to maintain proper balance. The rate of change for the COM will be dependent on the mass of the weight compared to the mass of the participant. Figure 1-2 depicts the set up for the cable pull system with all aspects labelled and shows a four cable system as mentioned previously (Mansfield and Maki, 2009).

Similarly, cable pulls may also be performed using motors and pulley systems to allow for multidirectional perturbations (Mille et al., 2013; Sturnieks et al., 2013). These designs still involve the waist as point of perturbation but can vary perturbation magnitudes in a more controlled manner than relying on free falling weights. The perturbation is therefore more controllable and replicable between subjects and between studies. Whether using a motor or free weights, the perturbation is relatively comparable as the point of application is consistent and the force can be replicated between techniques.

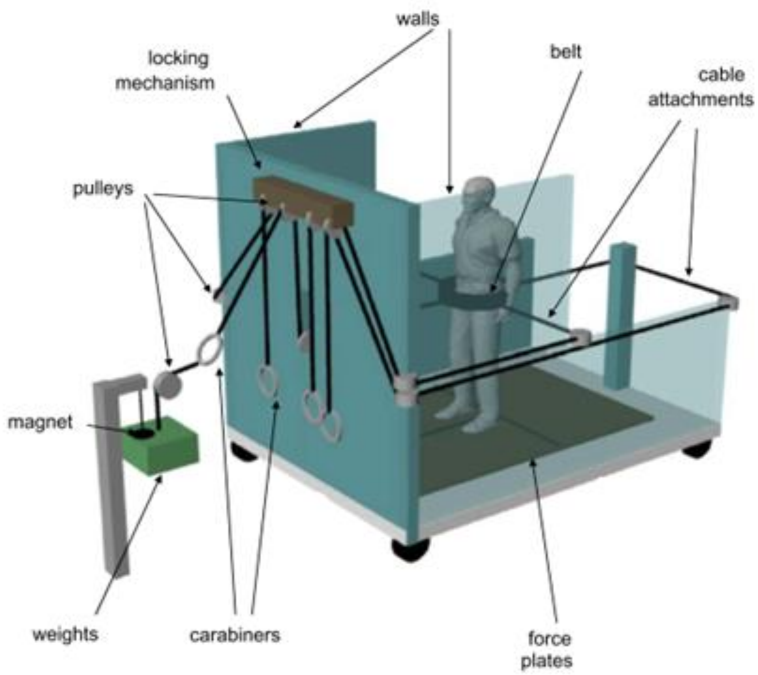

Figure 0-2: Schematic representation of a cable pull experimental set up demonstrating the location of the belt and therefore the point of perturbation application (Mansfield and Maki, 2009)**.**

1.2.2.3 Surface Translations

Surface translations are intended to simulate a loss of balance or a slip. This technique is unique as it involves the perturbation occurring at the level of the feet. Unlike real world slipping, there is not a loss of coupling between the foot and ground but rather a sudden movement in the ground causes the feet to move with the surface, while the inertial properties of the COM lag behind resulting in balance recovery to be necessary. The surface translations originally began as only anterior and posterior, but have recently been expanded to include lateral perturbations (Mansfield and Maki, 2009). Varying the overall distance of the translations and acceleration/deceleration rates and times also have been found to affect postural responses (Tokuno et al., 2010). This form of perturbation affords more control over more variables when compared to other methods, which unfortunately may reduce the comparability between studies. By being able to control the overall acceleration, peak velocity and displacement there is extensive variability found between study protocols.

1.2.3 Selection of Perturbation Technique

Having multiple methods of perturbation raises the concern of comparability between techniques chosen by different studies. In a comparison between the three previous outlined perturbation techniques done by Mansfield and Maki in 2009, they found that there were no major conflicting findings regardless of perturbation technique. However, the magnitude of the responses was found to vary between perturbation technique with surface translations yielding the largest differences between groups of young and older adults (Mansfield and Maki, 2009). This implies that a surface translation is more sensitive to postural control differences than the other two examined techniques. Therefore, the combination of the increased sensitivity and increased control make it an ideal approach to test balance and balance recovery.

Another benefit of the surface translation paradigm is the drastic improvement in freedom of the participant. By increasing the freedom of the participants' movements and actions there are more options for activities that could simulate real world experiences. Having space to move allows for gait to be studied and how a loss of balance would be reacted to in a more dynamic situation. Cable pulls and tether releases require the participant to be relatively static and minimize movement. A tether release has essentially no potential for dynamic activities prior to perturbation initiation as the participant must lean on the secure harness. Cable pulls allow for some movement but the mechanisms typically require little displacement from the starting point otherwise perturbation magnitudes may vary due to cable tension and therefore would allow tasks such as walking on the spot. Walking on the spot attempts to imitate walking in real life except it does not incorporate any of the inertial properties that accompany actual gait. With surface translations the participants are not required to stay in any spot in particular and therefore this allows for gait to be studied.

Although surface translations provide the opportunity for further dynamic studies to be performed there is still fundamental knowledge missing. A consensus of translational technique has not yet been established including any form on standardization of perturbation acceleration, peak velocity or displacement. By developing an established foundation first, the future research performed using surface translations will have increased merit and applicability as well as previous research will be more comparable to each other allowing more concrete conclusions to be made.

1.2.4 Lack of Consistency in Surface Translation Study Design Parameters

When examining the current literature that utilize surface translation perturbations there is substantial discrepancies between studies regarding the parameters used. Studies published from the same research group may use the same parameters but when comparing between various research groups there is no firm consensus. Table 1-1 outlines the published literature that use this method or a form of translating surface and the specific parameters they used (if they were provided). It is important to note that not all of the published studies included in this table utilize the same or similar system but all induce a surface translation. Some involve frictionless surfaces that can displace but do not involve any form of motor system. It is also important to note that only forward and backward translations were included as this thesis will focus on sagittal plane perturbations (some of the articles involved lateral translations as well).

As is made evident by the large variability shown in Table 1-1 for surface translation acceleration, velocity and displacement, there is a need for increased consistency within this testing paradigm to allow for more analogous comparisons to be made. Accelerations were found to range from 0.13 to 5 m/s^2 (only including balance recovery studies) (Wright and Laing, 2011; Zettel et al., 2008a, 2008b) or were not even available in some cases as the acceleration would be determined from the participants' gait parameters due to a frictionless surface (Yang et al., 2012, 2009; Yang and Pai, 2012). Peak velocities also varied drastically from 0.1 to 1.0 m/s (only including balance recovery studies) (Mansfield et al., 2007; Quant et al., 2005) or, again, were not available due to the presence of an uncontrolled frictionless surface (Yang et al., 2012, 2009; Yang and Pai, 2012). Overall displacements were also found to range from 0.04 to 1.5 m (only including balance recovery studies) (Bhatt and Pai, 2008; Quant et al., 2005). However, if the studies regarding gait balance recovery are excluded, the range of displacements changes to a max of 0.46 m (Tokuno et al., 2010) which is more relevant to the research being proposed for this thesis as the participants will be engaged in quiet standing at the time of perturbation.

The vast ranges found through this review of related literature clearly indicate variability in surface translation parameters in the literature. While each group may have selected parameters that were appropriate for answering their research question, a legitimate question is whether the conclusions drawn from individual studies are generalizable, or are perturbation-specific.

Table 0-1: Surface translation parameters provided by available published research studies. Contents inside of brackets relate to backward perturbations.

***Note: N/A=not applicable, N/P=not provided**

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Bateni et al., 2004	Forward (Backward)	2.0 (2.0, 3.0)	0.6 (0.6, 0.9)	0.18 (0.18, 0.27)	Standing	Holding a stability device (ie. cane) impairs ability to perform compensatory grasping.
Bhatt and Pai, 2008	Forward	N/A	N/A	0.9, 1.5	Gait	Observing slips provides tangible benefits in reducing falls but not to the extent of motor training.
Kam et al., 2016	Forward (Backward)	0.375-1 (0.875-1.5)	N/P	N/P	Standing	Weight-bearing asymmetry increased probability of stepping with unloaded leg.
Chen et al., 2014	Forward (Backward)	N/P	0.5 (0.5)	0.07 (0.07)	Standing	Surface translations are more destabilizing than surface sagittal rotations.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Diener et al., 1988	(Backward)	N/P	(0.1, 0.15, 0.25, 0.35)	(0.012, 0.036, 0.06, 0.12)	Standing	Degree of muscle activation appears to be modulated by sensory information based upon perturbation parameters.
Ferber et al., 2002	Forward (Backward)	N/A	0.4 (0.4)	0.1 (0.1)	Gait	Synchronized effort of the lower extremity joints is present to maintain dynamic balance during gait.
Henry et al., 1998	Forward (Backward)	0.135 (0.135)	0.35 (0.35)	0.09 (0.09)	Standing	Postural control responses specific to each perturbation direction depending on biomechanical constraints associated with each plane of movement.
Hsiao and Robinovitch, 1997	Forward (Backward)	4.2-9.7 (4.2-9.7)	0.91-2.5 (0.91-2.5)	0.21-0.52 (0.21-0.52)	Standing	Body exhibits series of responses which reduce risk of injury from fall and facilitate safer landing.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Inkol et al., 2018a	Forward (Backward)	0.73 (1.0)	0.22 (0.30)	0.07 (0.09)	Standing	Simplified COM calculations can be used without compromising substantial MOS accuracy.
Inkol et al., 2018b	Forward (Backward)	0.73-2.2 (1.0-3.0)	0.22-0.66 (0.3-0.9)	0.07-0.20 (0.09-0.27)	Standing	Generally, young adults prioritized balance response over cognitive task demonstrating the cognitive component of balance control.
Laing and Robinovitch, 2009	(Backward)	(5)	(0.2)	(0.265)	Standing	Low stiffness flooring can attenuate impact forces while having limited influence on balance.
Lin and Woollacott, 2002	Forward (Backward)	N/P	0.1 (0.1, 0.4)	0.1, 0.15 (0.05)	Standing	Temporal and spatial organization of postural muscle activation change due to age as well as individuals functional ability.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Maki et al., 1996	Forward (Backward)	0.73-2.2 (1.0-3.0)	0.22-0.66 (0.3-0.9)	0.07-0.20 (0.09-0.27)	Standing	Comparison between AP and ML translations demonstrating that responses to AP and ML perturbations are influenced by different constraints.
Maki et al., 2000	Forward (Backward)	0.65, 1.3 (1, 2)	0.2-0.4 (0.3-0.6)	0.05-0.16 (0.09-0.24)	Standing and walking in place	Impaired lateral-stepping reactions may be an early indicator of lateral fall risk.
Maki and Mcilroy, 1997	Forward (Backward)	1.5 (2.0)	0.45 (0.6)	0.14 (0.18)	Standing	Older adults appear to struggle more with lateral destabilization compared to younger adults.
Mansfield et al., 2007	Forward (Backward)	2.0 (3.0)	0.6 (1.0)	0.18 (0.27)	Standing	Proposed protocol for conducting a randomized control trial of perturbation based balance training.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
McIlroy and Maki, 1996	Forward (Backward)	1.5 (2.0)	0.45 (0.6)	0.135 (0.18)	Standing	Older and younger adults demonstrated similar step characteristics of the initial step but subsequent steps elicited age-related differences.
Nonnekes et al., 2013	Forward	0.5, 1.75	N/P	N/P	Standing	Startling auditory stimulus accelerate and strengthen postural responses.
Norrie et al., 2002	Forward (Backward)	0.75 (1.25)	0.23 (0.38)	0.068 (0.113)	Standing	Compensatory stepping is comprised of an initial "automatic" phase and a later "cognitive" phase.
Pai et al., 2011	Forward	N/A	N/A	Gait 1.5 YA, 0.9 OA, Sit-to-stand 0.24	Gait or sit-to-stand	Repeated slip exposure still results in developing fall-resisting skills in older adults.
Pai et al., 2006	Forward	N/A	N/A	0.24	Sit-to-stand	Young and older fallers had comparable weak limb support.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Pavol and Pai, 2007	Forward	N/A	N/A	0.24	Sit-to-stand	High incidence of falls in older adults related to deficient limb support.
Pavol et al., 2004	Forward	N/A	N/A	0.29	Sit-to-stand	Unsuccessful balance recovery was associated with diminished stepping response or an inappropriate reflexive step.
Quant et al., 2004	Forward (Backward)	0.5 (0.5)	0.15 (0.15)	N/P	Standing	Performing a cognitive task results in a decrease of early cortical activity.
Quant et al., 2005	Forward (Backward)	0.5	0.1	0.02	Standing	Perturbations with varying time between acceleration and deceleration do not affect later cortical potentials.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Tang and Woollacott, 1998	Forward	N/P	0.4	0.1	Gait	Older adults found to have inefficient balance strategy resulting in the use of secondary compensatory adjustments.
Tang and Woollacott, 1999	Forward (Backward)	N/P	0.4 (0.4)	0.1 (0.1)	Gait	Posture responses were differentially modulated to meet the needs of the step cycle.
Tang et al., 1998	Forward	N/P	0.4	0.1	Gait	Experience/exposure to perturbations results in fine-tuning of the nervous system's response to slips.
Tokuno et al., 2010	Forward (Backward)	1.2 (1.2)	0.2 (0.2)	0.06, 0.46 (0.06, 0.46)	Standing	Translations with increased displacement reveal more age-related postural control differences.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Tripp et al., 2004	Forward (Backward)	1.5, 2.5 (1.5, 2.5)	0.45, 0.75 (0.45, 0.75)	0.068, 0.112 (0.068, 0.112)	Surface translation following tether release	Step direction can be modulated during early stages of step reactions.
Weerdesteyn et al., 2012	Forward	15	1, 3	1.2	Standing	Body configuration at instant of foot contact accurately predicted successful or failed balance recovery attempts.
Wright and Laing, 2011	(Backward)	(5)	(0.2)	(0.265)	Standing	Study specific compliant floors had minimal influences on balance and supports the progression to clinical trials.
Yang et al., 2009	Forward	N/A	N/A	Gait 1.5, Sit-to-stand 0.24	Gait or sit-to-stand	Stability and limb support immediately prior to recovery step were highly effective at predicting falls.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Yang et al., 2012	Forward	N/A	N/A	1.5	Gait	Unilateral and bilateral slips have comparable likelihood of resulting in a fall.
Yang and Pai, 2012	Forward	N/A	N/A	0.75	Gait	Overhead harness loads can be reliably used as a predictor of falls in gait.
Zettel et al., 2005	Forward (Backward)	0.13-3.0 (0.13-3.0)	0.2-0.9 (0.2-0.9)	N/P	Standing	Visual fixation of the foot or floor is not required during obstacle avoidance or target landing during perturbation.
Zettel et al., 2002	Forward (Backward)	1.0, 3.0 (1.0, 2.0, 3.0)	0.3, 0.9 (0.3, 0.6, 0.9)	N/P	Standing or taking a single step	Hybrid postural control comprised of predictive and reactive control may be implemented to improve stability.

Author(s)	Direction of Translation	Acceleration (m/s^2)	Peak Velocity (m/s)	Displacement (m)	Task at Perturbation	Key Finding
Zettel et al., 2007	(Backward)	(3)	(0.9)	N/P	Standing	Balance recovery reaction based on visuospatial environmental information gathered prior to perturbation.
Zettel et al., 2008a	Forward (Backward)	0.13-3.0 (0.13-3.0)	0.2-0.9 (0.2-0.9)	N/P	Standing	Competing attentional demands compromised the control of lateral stability in older adults during compensatory stepping.
Zettel et al., 2008b	Forward (Backward)	0.13-3.0 (0.13-3.0)	0.2-0.9 (0.2-0.9)	N/P	Standing	Aging did not impair the predominate visual control strategy employed during reactive stepping.

1.2.5 Discrepancies Between Surface Translation Literature

Table 1-1 demonstrates the differences between study designs while utilizing the same or similar methods of perturbation. However, these discrepancies are of little consequence if they have no impact on the results from studies. Unfortunately, the large degree of outcome measures reported by surface translation studies makes direct comparisons difficult.

A relatively common measure that is reported is the onset of muscle activation or the activation latency. The method of reporting varies greatly though depending on the study design. Some studies report the differences between groups whereas others report individual muscle latencies and others report group onsets. The discrepancy between studies could vary as much as 161 ms (Tokuno et al., 2010) to 100 ms (Lin and Woollacott, 2002) for the same population but different perturbation parameters. The previous studies identified the time to muscle onset compared to the study by Quant et al. (2004) that identified the difference between various tasks to be 5 to 7 ms. If the focus is directed between the studies by Tokuno et al. (2010) and Lin and Woollacott (2002), both reported muscle onsets in the same style and can therefore be compared. Average muscle onset latency of 161 ms (Tokuno et al., 2010) to 100 ms (Lin and Woollacott, 2002) shows a substantial difference in magnitude with no originally apparent reason for this variation as both studies use similar populations and perturbation techniques. However, acceleration, velocity and displacement were not consistent between these study designs. The study in 2002 did not provide an acceleration value for the perturbations whereas in 2010 an acceleration of 1.2 m/s^2 was used. In a similarly inconsistent manner, the velocities and displacements were different between studies. This raises the concern that varying these parameters could have an effect on the measured response and be an underlying reason behind the measurable differences between the studies.

Another measure used in the literature is COM displacement. This measure is typically decomposed into anteroposterior (AP) and mediolateral (ML) components but both components may or may not be reported. If the studies by Henry et al. (1998) and McIlroy & Maki (1996) are compared for their COM displacements in the AP a substantial difference in values can be observed. Henry et al. (1998) found a mean backward COM displacement of 0.05 m and a forward displacement of 0.06 m. McIlroy & Maki (1996) found displacements of 0.09 and 0.15 m in the backward and forward directions respectively. This makes the differences observed larger than the magnitude of the values in the forward displacement scenario. However, these extensive differences in values may be attributable to the magnitude of the perturbations utilized by these studies. An acceleration of 0.135 m/s^2, velocity of 0.35 m/s and displacement of 0.09 m were used for both anterior and posterior perturbations by Henry et al. (1998). The study performed by McIlroy & Maki (1996) used different perturbation magnitudes for the anterior and posterior perturbations. Anterior perturbations had an acceleration of 1.5 m/s^2, velocity of 0.45 m/s, and displacement of 0.135 m. Posterior perturbations used 2.0 m/s^2, 0.6 m/s, and 0.18 m for the acceleration, velocity and displacement respectively. The differences found in the COM displacements are less confusing when all of the aforementioned information is considered. The individuals experienced a larger displacement of their COM because they were exposed to a substantially larger perturbation. Continuing the examination of COM displacement, more recent research has reported values ranging from 0.06 m (Chen et al., 2014) to extrapolated COM displacement of over 0.40 m (Inkol et al., 2018b). These variations in outcome measures relate directly to the magnitude of perturbation used and demonstrate how the range of test parameters implemented confounds comparison of results between studies.

The previously outlined examples dictate how a lack of understanding the underlying mechanical relationship between platform parameters and outcome measures jeopardizes the comparability of the previous literature. Without further understanding of the effects of perturbation parameters the lack of comparability between studies may continue.

1.2.6 Control of the Person

As outlined previously, there are multiple techniques used in research to perturb an individual's balance, but one aspect that was not compared between the paradigms was that of the state of the participant prior to perturbation. Pre-perturbation activity is most often controlled during the tether-release study design as the direction of perturbation and the timing can be more predictable than the other two described methods. This has led to monitoring and controlling various aspect of the participant prior to perturbation including muscle activity, weight distribution, and centre of pressure location (Singer et al., 2016, 2012; Weaver, 2017). However, these techniques are not typically implemented during surface translations due to the unpredictable nature of the perturbations. Controlling or monitoring some or all of these measures as well as others could potentially provide insights into the response outcomes being observed but it is unknown because these are not reported in the literature. While controlling these measures would reduce the freedom of the participant, simply monitoring them and accounting for them may provide valuable detail into the underlying mechanical mechanisms and provide context to the variability observed in the measured outcomes.

1.2.7 Research Objectives and Hypotheses

The global objective of this thesis was to determine factors that influence the magnitude of stepping responses during surface translation perturbations in healthy young adults. The specific goals were as follows:

1) to test the influence of two specific features of surface translation perturbations (peak velocity, acceleration) on spatial measures of stepping responses, and

2) to explore the influence of trial-specific pre-perturbation biomechanical measures on stepping responses.

In order to accomplish these objectives, participants' *step length*, *maximum extrapolated centre of mass displacement*, and *minimum extrapolated margin of stability* were examined to identify different responses expressed between unique translational perturbations. In addition to these outcome measures, participants' *ankle muscle co-activation*, *body weight distribution*, *centre of pressure* and *centre of mass characteristics* were examined to explore their potential effects on outcome measures. It was hypothesized that:

i) Participants would have larger step length and xCOM during trials with increased acceleration and increased peak velocity. These reductions would result in a decreased xMOS driven by a smaller increase in step length compared to xCOM (resulting in the smaller overall extrapolated margin of stability).

ii) Pre-perturbation trial specific factors would be associated with spatial metrics of stepping responses. Specifically, ankle co-contraction index, weight distribution between the feet, centre of pressure and centre of mass characteristics all measured prior to perturbation onset would significantly improve statistical model prediction of response outcomes compared to models containing only external platform factors.

Study 1 - Effects of varying translational platform characteristics on single step spatial stepping responses

1.3 Introduction

Researchers have been attempting to understand the humans' balance control systems for decades (Woollacott et al., 1980). The use of translating surfaces to perturb balance is a classic technique (Horak and Nashner, 1986) to elicit reactionary responses of balance control. Many studies have examined different response strategies including fixed-support and change-in support strategies, however change-in support strategies, specifically single step responses, will be the focus of this study.

Change-in support strategies are primarily achieved by using one's hands to grasp an object (Allum et al., 2002; Maki and McIlroy, 2006) or taking one or more steps (Maki et al., 2000; Singer et al., 2016; Tripp et al., 2004). Both of these strategies are used to increase an individual's base of support to prevent a complete loss of balance which could result in a fall. Although use of the hands has been studied, taking a step to maintain ones balance is the primary change-in support method implemented when individuals are free from constraints (McIlroy and Maki, 1993). As mentioned previously, examination of balance control and the responses observed is not a new concept and has been explored by many researchers. However, even though research groups have used surface translations relatively extensively, there is no consensus or consistency between research groups regarding the parameters of the surface translation used. Surface translation often vary in the magnitude of the displacement used as well as other primary factors such as the platforms acceleration and the peak velocity that the platform achieves during the perturbation. This uncertainty reduces the interpretability of this area of research as comparisons between studies and their outcomes can be convoluted based on a lack on symmetry in the study designs.

To addresses these gaps in the literature, the goal of this study was to examine the effects of translating surface peak velocity and acceleration on spatial parameters derived from reactive single step responses. Specifically, it was hypothesized that increases in peak platform velocity and platform acceleration would result in (1) greater normalized step length, (2) greater extrapolated centre of mass displacement, and (3) reduced extrapolated margin of stability.

1.4 Methods

1.4.1 Participants

Twenty-four young healthy adults (Mean (SD): age 24.0 (3.61) years; height 1.71 (0.08) m; mass 73.2 (12.5) kg) participated in this study; 12 were male and 12 were female. Exclusion criteria included: i) any form of balance impairment, ii) musculoskeletal injury or, iii) neurological disorder as their responses may have been atypical. Written informed consent was obtained prior to the experimental protocol. The Office of Research Ethics at the University of Waterloo (ORE #21988) approved this study.

1.4.2 Instrumentation

1.4.2.1 Translating Platform

Surface translations were elicited via a custom-built dual-axis servo-driven platform (4.87 m x 2.10 m) (Sidac Automated Systems Inc., Toronto, ON) (Figure 2-1). Translations occurred along both horizontal axes resulting in anterior, posterior, left and right translations (relative to participant). Surface displacements were held constant at 0.30 m in all perturbation directions while accelerations and velocities were varied depending on direction of translation and ranged from 0.5-3.5 m/s^2 and 0.5-1.0 m/s, respectively. Participants were provided with a visual target on the wall in an attempt to reduce variability of visual cues from the platform and environment.

Figure 0-1: Image of assembled translating platform with embedded force platforms and surrounding Optotrak cameras

1.4.2.2 Kinematics

Whole body kinematics were collected using a 12-camera active infrared optoelectronic system (Optotrak Certus, Northern Digital Incorporated, Waterloo, Ontario, Canada) collected at 64 Hz through First Principles software (Northern Digital Incorporated, Waterloo, Ontario, Canada). The collection space was calibrated and aligned prior to the participants' arrival. Multi-marker tracking clusters were placed on segments of interest including bilateral feet, shanks, legs, forearms and upper arms as well as the pelvis, thorax and head. End points of each segment were digitized in relation to the respective cluster. The number of markers per segment cluster as well as the associated digitization points are presented in Table 2-1.

Table 0-1: Kinematic tracking cluster locations and associated digitization landmarks.

Segment	Cluster Location (Number of Markers)	Digitization Landmarks
Foot (Bilateral)	Lateral aspect of foot below lateral malleolus (4)	Lateral Malleolus Medial Malleolus 1st Metatarsal Head 2nd Metatarsal Head 5th Metatarsal Head 1st Distal Phalange Calcaneus
Shank (Bilateral)	Mid shank facing laterally (4)	Lateral Tibial Condyle Medial Tibial Condyle Lateral Malleolus Medial Malleolus
Thigh (Bilateral)	Lower third of thigh facing laterally (4)	Greater Trochanter Lateral Femoral Epicondyle Medial Femoral Epicondyle
Pelvis	Belt with cluster fixed to sacrum (4)	Anterior Superior Iliac Spines Posterior Superior Iliac Spines Iliac Crests Greater Trochanters
Trunk	Chest cluster (4)	Iliac Crests Acromions C7 Spinous Process Xiphoid Process Suprasternal Notch
Head	Head band facing laterally (4)	Gonion Processes External Auditory Meatuses Vertex of Head
Upper Arm (Bilateral)	Mid upper arm facing laterally (4)	Acromion Lateral Humeral Epicondyle Medial Humeral Epicondyle
Forearm (Bilateral)	Mid forearm facing laterally (4)	Lateral Humeral Epicondyle Medial Humeral Epicondyle Styloid Process of Ulna Styloid Process of Radius
Hand (Bilateral)	Single marker on the 3rd metacarpal	N/A

1.4.2.3 Load Cell

A load cell (MLP-300-CO, Transducer Techniques, Temecula, CA) was placed in series with the participant's ceiling mounted safety harness which allowed for identification of 'failed' balance recovery trials. This data was monitored live using LabVIEW routines

(National Instruments Corporation, Austin, TX) with the outcome of "pass" or "fail" recorded for each trial. A criterion value of 18.5% of the participants body weight was used to identify a successful versus a failed trial (Thelen et al., 1997). If a trial was marked as a "fail" it was recollected to ensure a complete dataset for every participant.

1.4.3 Experimental Protocol

Participants visited the Injury Biomechanics and Aging Laboratory on one occasion and all data was collected. The collection required approximately two and a half hours from the time the participant signed the informed consent to the time when instrumentation was removed and the participant received their remuneration.

Upon completion of informed consent, basic anthropometric measurements of height, weight and age were taken and a health questionnaire completed to ensure participant eligibility. Whole body kinematic set up was performed which required the use of medical grade skin tape to adhere the optoelectronic markers to the participants' skin in the necessary positions. Table 2-1 outlines the tracked segments and the locations of the clusters as well as the landmarked digitization points for each respective cluster which allowed for segment endpoints to be identified. Kinematic data was collected for all trials performed.

The independent variables utilized included platform acceleration and peak platform velocity. Although four perturbation directions were employed, only backward translations were explored as part of this thesis. Backward translations consisted of seven different platform acceleration values along with two different peak platform velocities at a constant displacement. Platform acceleration values were selected as 0.5 m/s^2, 1.0 m/s^2, 1.5 m/s^2, 2.0

m/s^2, 2.5 m/s^2, 3.0 m/s^2, and 3.5 m/s^2. Peak velocities were employed at two different levels for all acceleration levels except the lowest (0.5 m/s^2) acceleration level which only included a single peak velocity. A peak velocity of 0.5 m/s was targeted at every acceleration level and will be referred to as the 'low' peak velocity for the levels of 1.0 m/s^2, 1.5 m/s^2, 2.0 m/s^2, 2.5 m/s^2, 3.0 m/s^2, and 3.5 m/s^2 acceleration. The 'high' peak velocity for each of the target acceleration levels was the theoretical maximum achievable velocity at the given acceleration level based on equations of motion. The lowest acceleration level (0.5m/s^2) resulted in a theoretical maximum achievable velocity of 0.5 m/s, which was used as the low velocity condition for all other acceleration levels. This resulted in the 0.5 m/s^2 acceleration level only having one level of peak velocity. The platform displacement and deceleration were held constant for all trials at 0.3 m and 5.0 m/s^2 respectively. Appendix A contains graphical depictions of all backward translations including the platforms position, velocity and acceleration over time. The data collection also consisted of trials in the forward and lateral (both left and right) directions. This data was collected but was not analyzed for the scope of this thesis. An outline of all trials and their parameters is presented in Table 2-2.

Participants completed five practice trials which consisted of one perturbation in each direction where the participant was informed prior to perturbation both the direction and the timing of the translation. The last practice trial was always a backward perturbation however the participant was not aware of the direction prior to perturbation, only the timing of the translation. This process was implemented in an attempt to improve participant's initial comfort while on the platform.

Each combination of parameters was implemented four times throughout the data collection (with the exception of the trials used as practice) resulting in 114 trials in addition to the five practice trials (119 trials total). The 114 trials were subsequently split into two blocks, the first block consisted of one of each combination and the second block consisted of the remaining three repetitions. Within each block, the trial order was completely randomized to minimize anticipation affects. Splitting the repetitions into two separate blocks was done to mitigate the learning effects observed during pilot testing.

Every trial was monitored live from the collection computers to ensure meaningful data was collected. The load cell data was monitored to classify successful and failed recovery attempts as outlined in section 2.2.2.3. When a trial was deemed a fail, the data was saved for subsequent analysis (not within the scope of this thesis) and the trial was recollected.

Table 0-2: Perturbation trial parameters including direction, acceleration, velocity and displacement. Bolded trials were assessed for this study.

Backward Translations			**Forward Translations**		
Acceleration (m/s^2)	Velocity (m/s)	Displacement (m)	Acceleration (m/s^2)	Velocity (m/s)	Displacement (m)
0.5	0.50	0.3	0.5	0.50	0.3
1.0	**0.50**	**0.3**	1.0	0.50	0.3
1.0	**0.65**	**0.3**	1.0	0.65	0.3
1.5	**0.50**	**0.3**	1.5	0.50	0.3
1.5	**0.75**	**0.3**	1.5	0.75	0.3
2.0	**0.50**	**0.3**	2.0	0.50	0.3
2.0	**0.85**	**0.3**	2.0	0.85	0.3
2.5	**0.50**	**0.3**	2.5	0.50	0.3
2.5	**0.90**	**0.3**	2.5	0.90	0.3
3.0	**0.50**	**0.3**	3.0	0.50	0.3
3.0	**0.95**	**0.3**	3.0	0.95	0.3
3.5	**0.50**	**0.3**			
3.5	**1.00**	**0.3**			
Left Translations			**Right Translations**		
Acceleration (m/s^2)	Velocity (m/s)	Displacement (m)	Acceleration (m/s^2)	Velocity (m/s)	Displacement (m)
1.0	0.5	0.3	1.0	0.5	0.3
2.0	0.5	0.3	2.0	0.5	0.3
3.0	0.5	0.3	3.0	0.5	0.3

Participants were given the following verbal instructions prior to the perturbations:

> “The platform will move either forward, backward, left or right. When movement occurs, do whatever is necessary to maintain your balance. However, if you are going to take a steps to maintain balance

> please try to do so in a single step. Avoid using the safety harness to maintain balance; it will prevent you from falling and impacting the surface but should not be used for support."

Once all of the trials were completed, the platform was shut down and the safety harness removed.

1.4.4 Programmed Platform Characteristics

Theoretical programmed time varying platform position, velocity, and acceleration can be found in Figures 2-2 – 2-14. Every combination of parameters used during backward perturbations are shown. The graphical representations depict a 0 order system for platform acceleration as programmed.

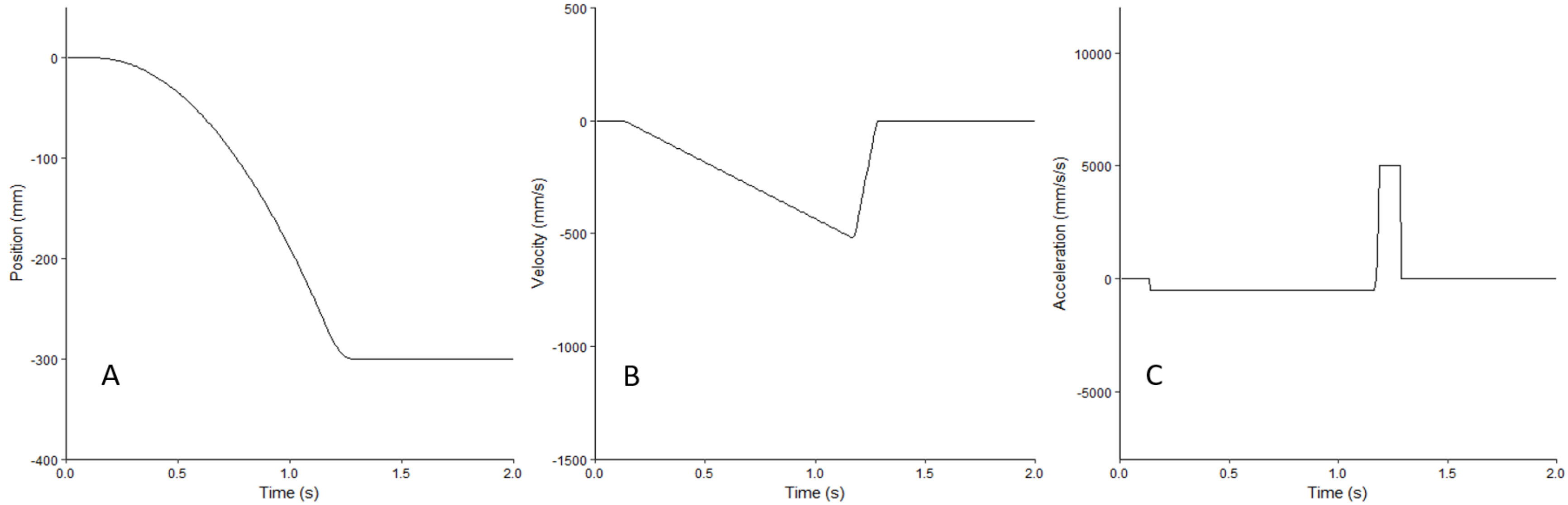

Figure 0-2: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 0.5 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

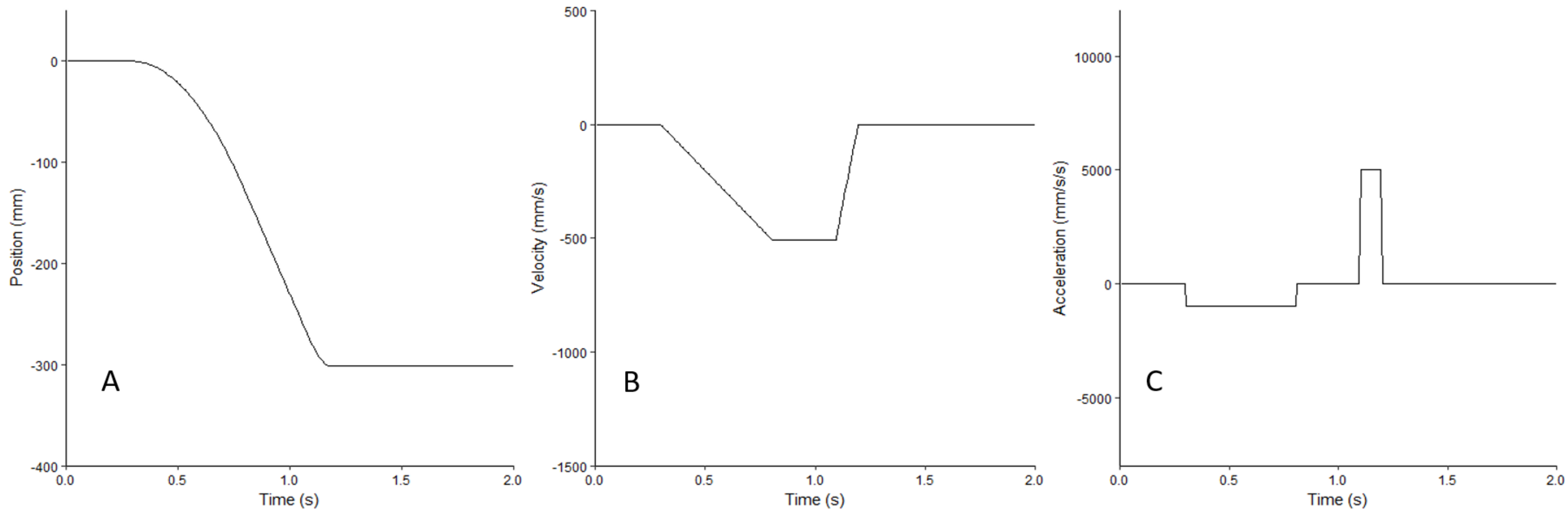

Figure 0-3: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 1.0 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

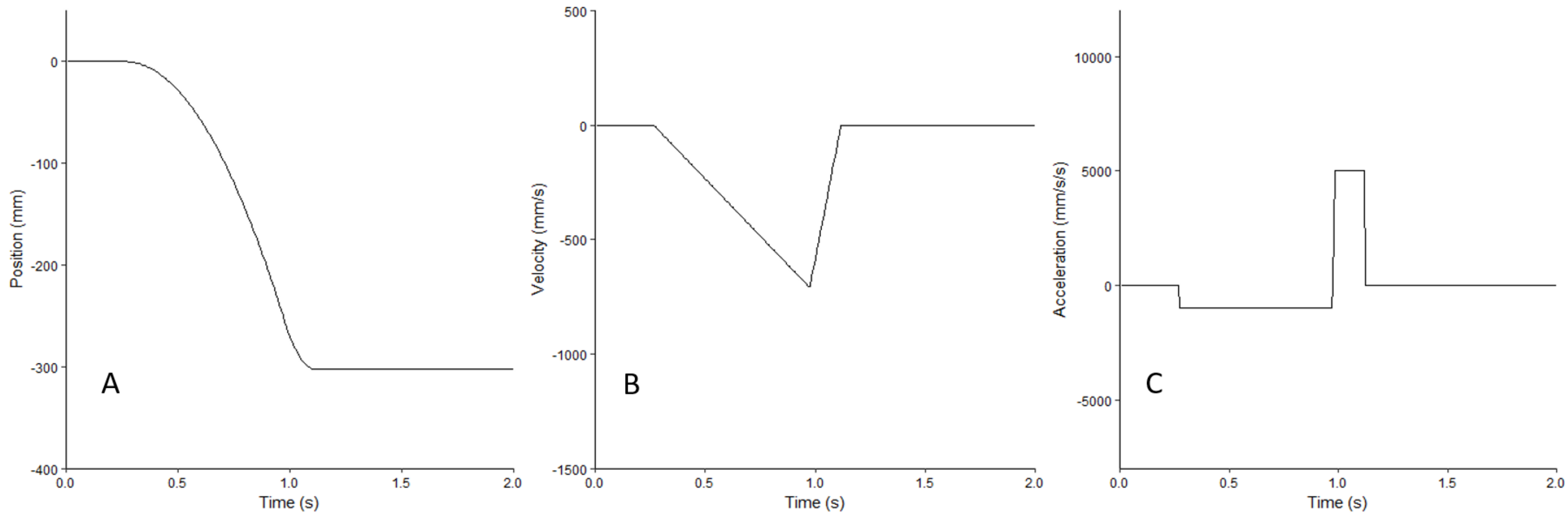

Figure 0-4: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 1.0 m/s^2 acceleration with a target peak velocity of 0.65 m/s and 0.30 m displacement.

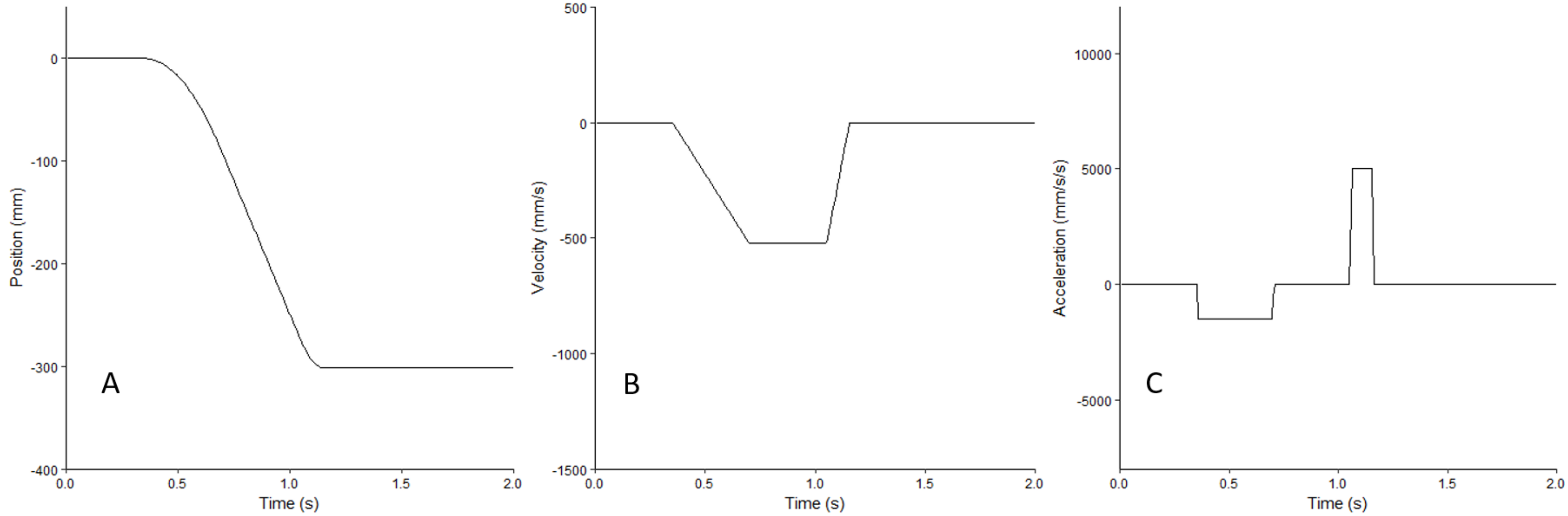

Figure 0-5: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 1.5 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

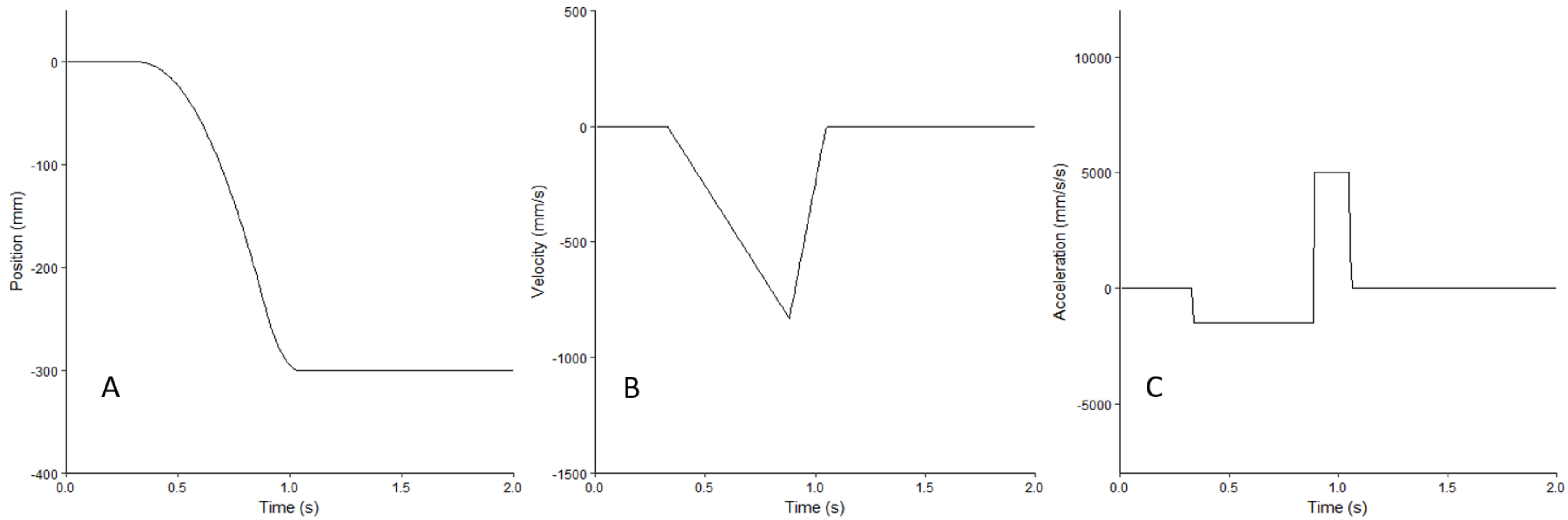

Figure 0-6: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 1.5 m/s^2 acceleration with a target peak velocity of 0.75 m/s and 0.30 m displacement.

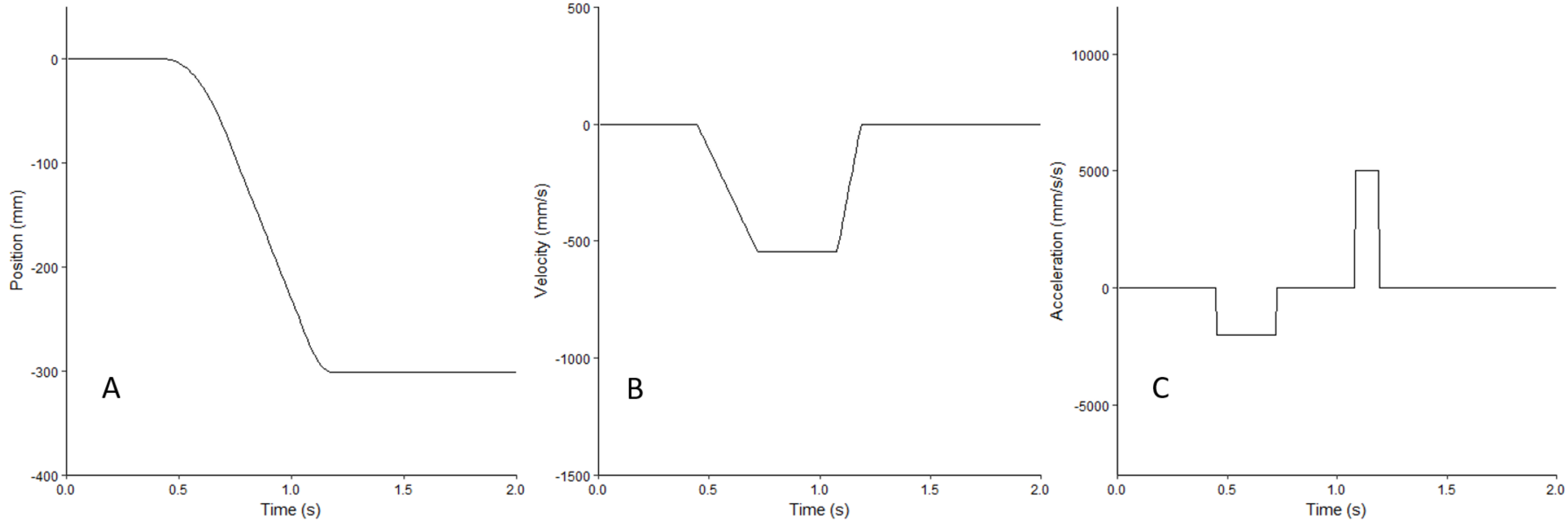

Figure 0-7: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 2.0 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

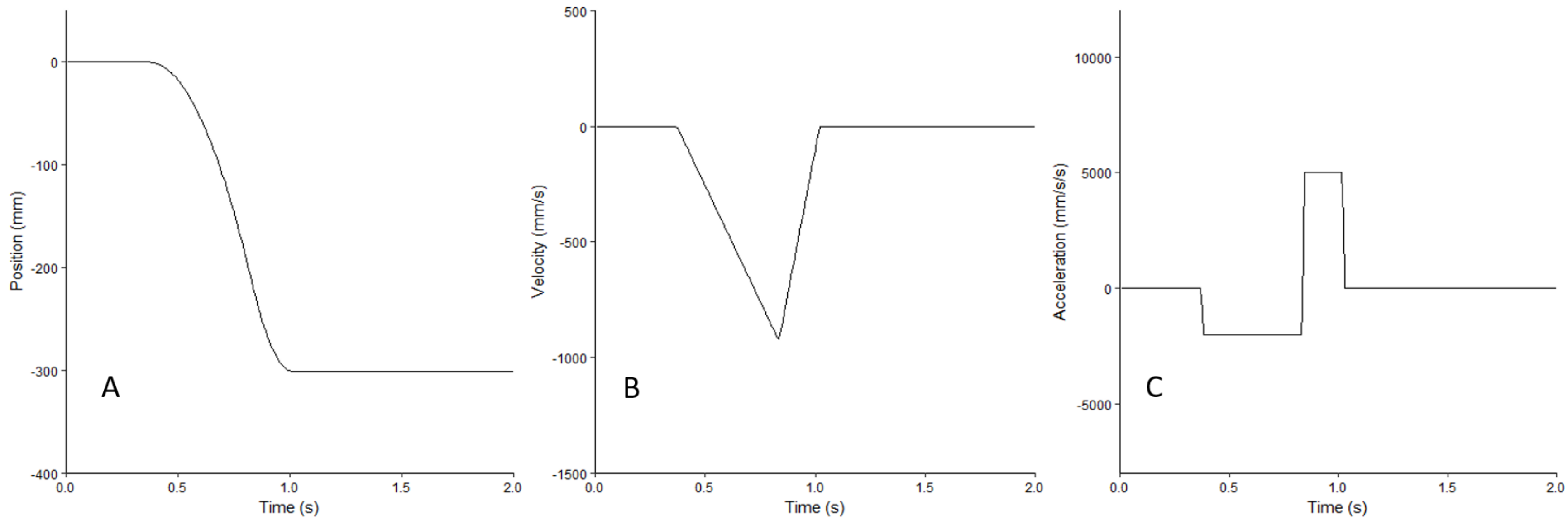

Figure 0-8: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 2.0 m/s^2 acceleration with a target peak velocity of 0.85 m/s and 0.30 m displacement.

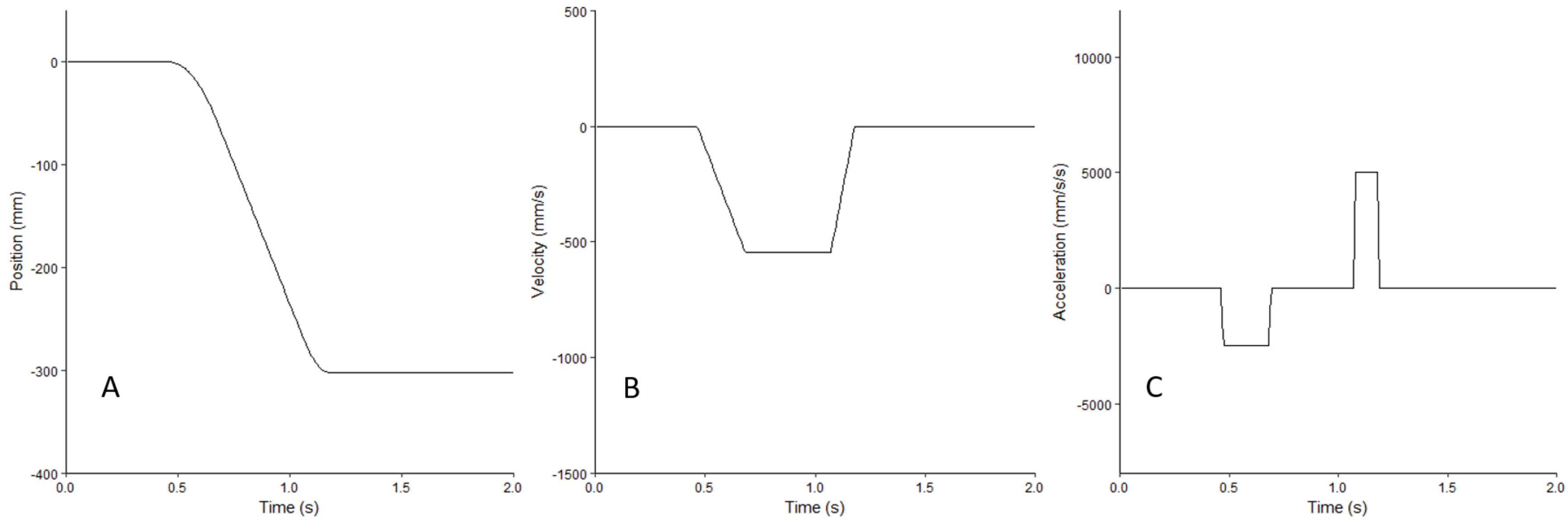

Figure 0-9: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 2.5 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

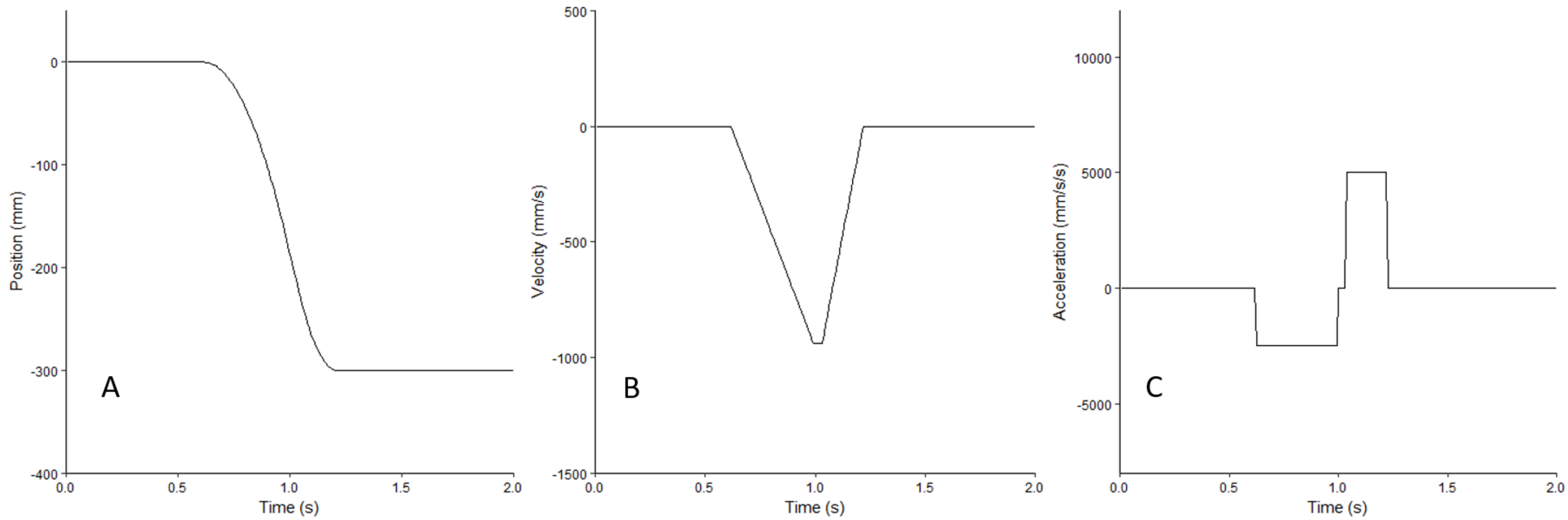

Figure 0-10: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 2.5 m/s^2 acceleration with a target peak velocity of 0.90 m/s and 0.30 m displacement.

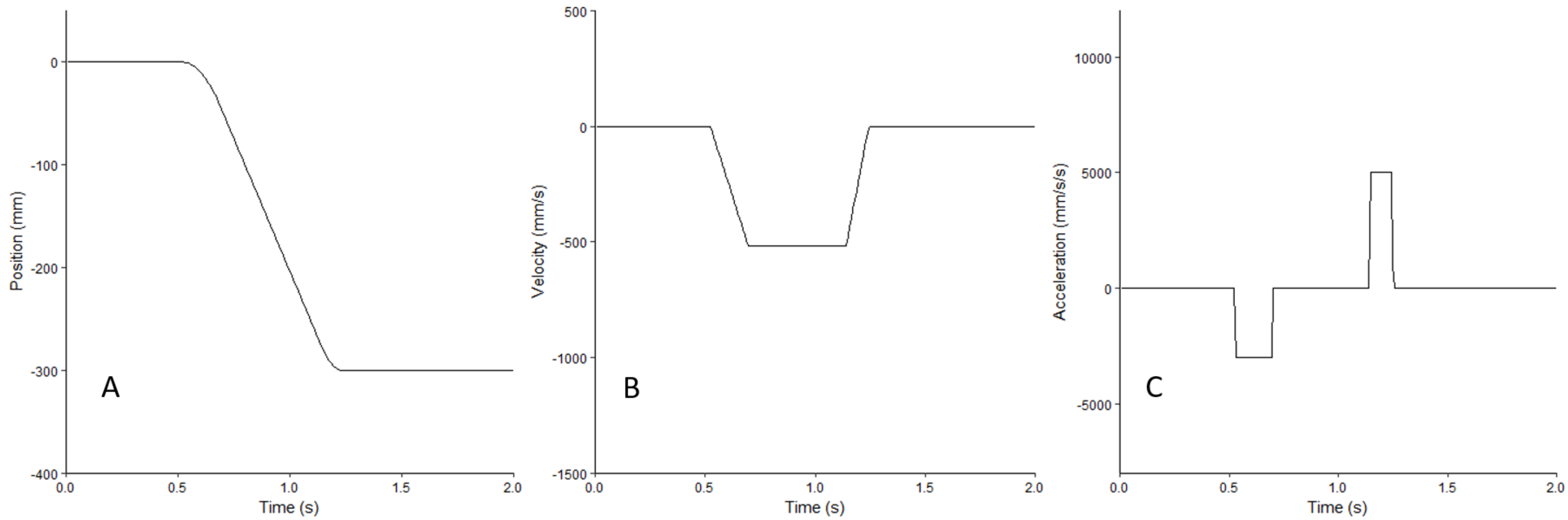

Figure 0-11: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 3.0 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

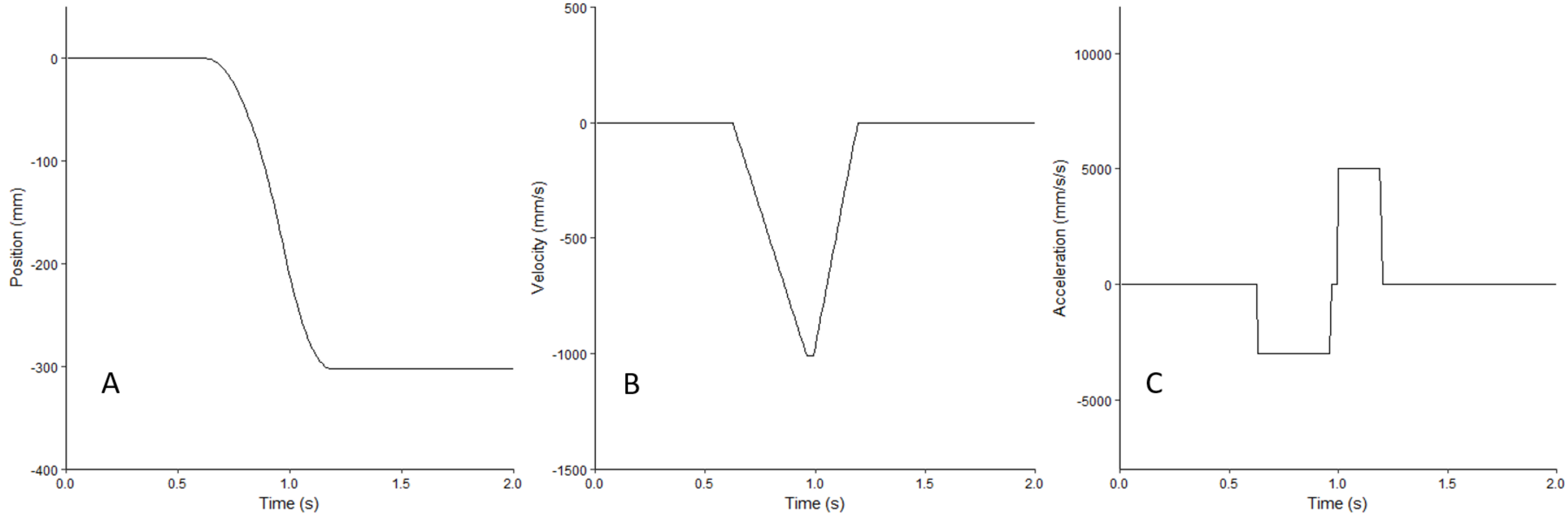

Figure 0-12: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 3.0 m/s^2 acceleration with a target peak velocity of 0.95 m/s and 0.30 m displacement.

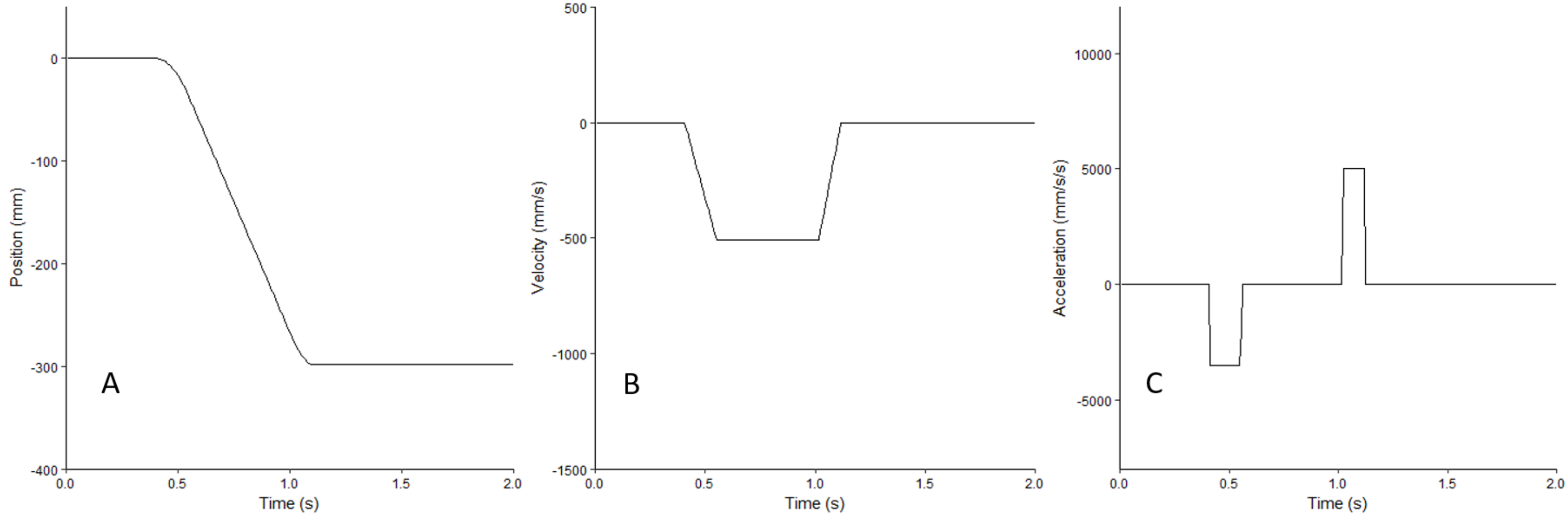

Figure 0-13: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 3.5 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

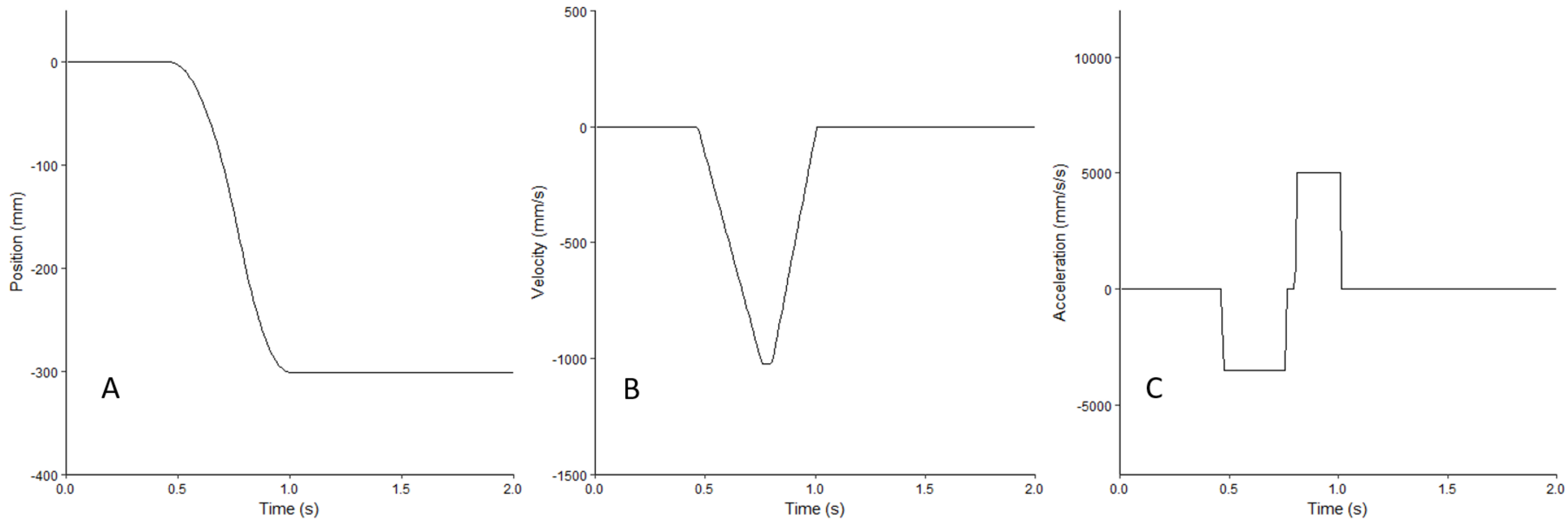

Figure 0-14: Comparison of platform position (A), velocity (B), and acceleration (C) of theoretical programmed values for the 3.5 m/s^2 acceleration with a target peak velocity of 1.00 m/s and 0.30 m displacement.

1.5 Data Analysis

Trials from block two were analyzed and backward translations were the focus of this study and subsequently the only trials analyzed. Primary analysis classified every trial as no step, single step, or multi step through visual inspection of data. Single step responses were analyzed further as single step responses were the focus of this study.

1.5.1 Kinematic Data Processing

Raw kinematic data was analyzed using custom MATLAB™ routines (version R2015a, Mathworks Inc., USA). Gaps in kinematic data were interpolated (<1000 ms) using a cubic spline routine (Warnica et al., 2014; Weaver, 2017). Data was dual pass filtered with a second order, low-pass Butterworth filter with a cut off frequency of 6 Hz as voluntary human movements rarely exceeds this frequency. Platform movement was accounted for by subtracting the coordinates of a marker that was rigidly attached to the platform from every data point following the process of filtering.

1.5.1.1 Platform Movement

A rigid cluster was attached to the platform and was used to track the platform's position as well as calculate velocity and acceleration. This data was processed using the same techniques as the rest of the kinematic data to ensure time synchronization. Position data was differentiated using a central-difference method to calculate velocity. The process of differentiation was then performed on the platform velocity data to calculate platform acceleration. The onset of platform movement was defined as the frame at which platform acceleration exceeded 0.1 m/s^2 (Bateni et al., 2004; Maki and Mcilroy, 1997; Norrie et al., 2002).

1.5.1.2 Step Length

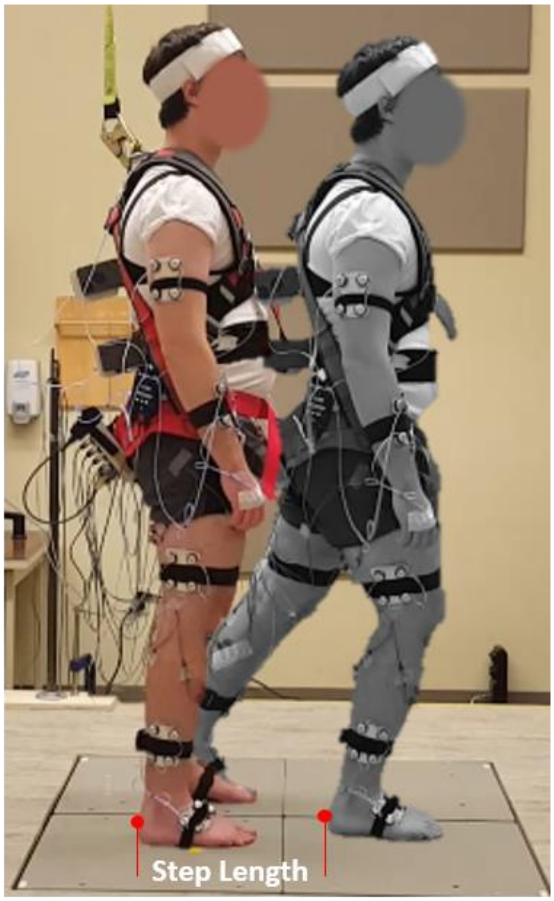

Figure 0-15: Participant step length based on heel displacement.

Step length was determined using the digitized point of the heel from the foot cluster (Figure 2-15). Initial AP position was determined during quiet stance prior to perturbation (mean of position during first 1000 ms of trial) and final AP position was determined following foot-contact (mean of position during last 1000 ms of trial). This process was performed bilaterally – the stepping limb was defined as the foot with the larger displacement. The difference between the final position and initial position of the stepping leg was identified as step length with normalized step length resulting from dividing the step length by the length of the participants' leg: greater trochanter to ipsilateral lateral malleolus.

1.5.1.3 Extrapolated Centre of Mass

Segment clusters as well as the corresponding digitization points were utilized to model individual body segments. The anthropometric tables used were from de Leva (1996) which are based off of the original anthropometric data set collected by Zatsiorsky-Seluyanov

(1983). This anthropometric data set was selected because the population used to generate the original data were young healthy Caucasians, which was more representative of the collected sample than other anthropometric data sets. However, the original data from Zatsiorsky-Seluyanov tables involved landmarks that were difficult to identify. De Leva modified the anthropometric data to allow more easily accessible landmarks to be utilized (de Leva, 1996). Overall, the de Leva modified anthropometric data set matches the population of interest and was more feasible to implement. After implementation of each individual segment's anthropometric properties, time-varying whole body COM was calculated.

Equation 1: $$xCOM = COM_{pos} + \frac{COM_{vel}}{\sqrt{g/COM_{vert}}}$$

Centre of mass was converted to extrapolated COM, or xCOM, by incorporating the COM velocity at every corresponding time point and was calculated in accordance with previous works of Hof et al, 2005. The xCOM was calculated as per Equation 1 where; COM_{pos} was the position of the COM, COM_{vel} was the velocity of the COM at that time point, COM_{vert} was the vertical height of the COM from the ground, and g is the acceleration due to gravity. Figure 2-16 demonstrates this relationship by plotting the AP COM position, AP COM velocity, and AP xCOM position. During the perturbation, the xCOM displacement will be larger than COM displacement unless a negative velocity is experienced (forward loss of balance resulted in a positive velocity as shown in the figure).

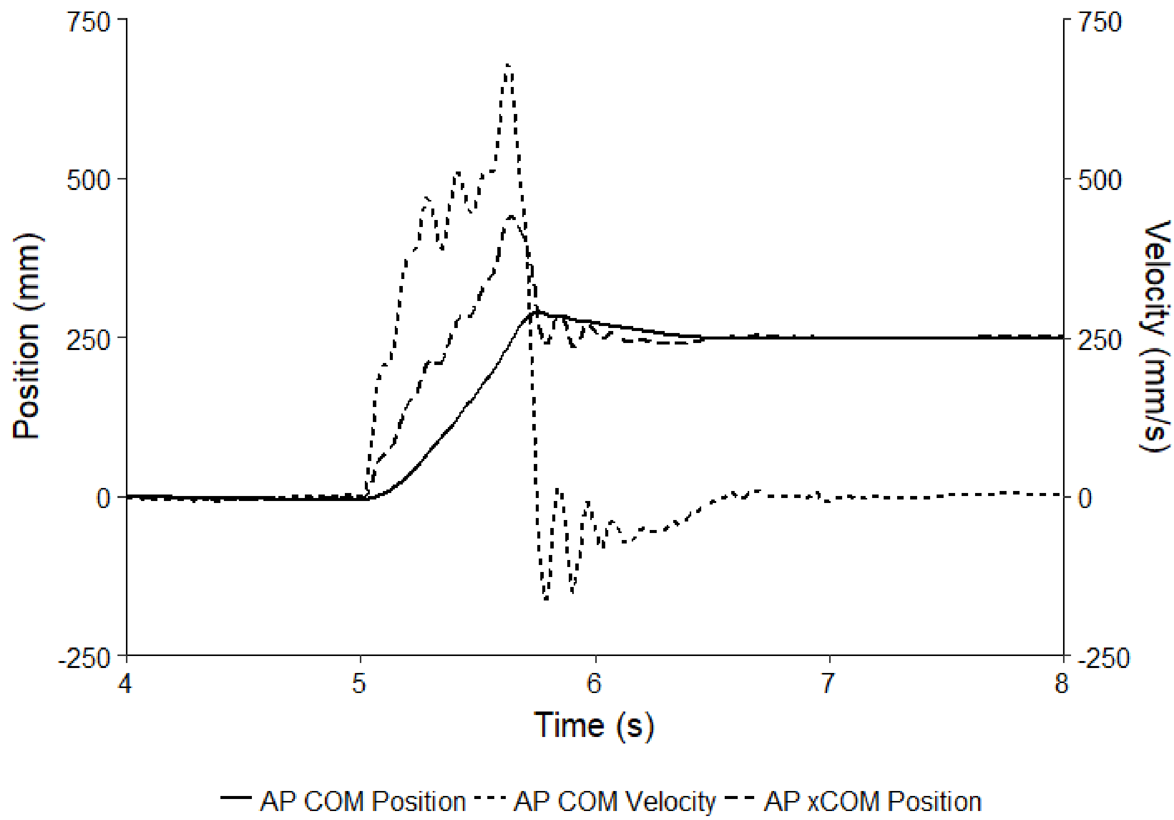

Figure 0-16: Representative trial of the relationship between AP COM (solid line), COM velocity (dashed line), and xCOM (longdashed line).

1.5.1.4 Extrapolated Margin of Stability

Once AP xCOM location was calculated, the relation to the BOS was established which allowed for the calculation of extrapolated margin of stability (BOS – xCOM). The step foot was determined using kinematic data and is described in section 2.2.4.1.2 Step Length. Once the step foot was determined, the AP xCOM component was compared to the 1st distal phalanx digital point of the step foot. The use of these digital points to determine the xMOS was due to the anatomical relevance of these markers and how it comprised the outer aspect of the BOS based on a backward translation. Extrapolated margin of stability was explicitly extracted at its minimum value following foot-contact.

1.5.2 Statistical Analysis

Data was analyzed based on within factors of platform acceleration (6 levels) and peak platform velocity (2 levels) and performed using IBM SPSS Statistics (V22, Armonk, NY). Step length, AP xCOM displacement and minimum AP xMOS after heel strike were the main dependent variables assessed using analysis of variation (ANOVA) statistical models. Post-hoc pairwise comparisons were performed when significant interactions or main effects were found. A significant alpha level of 0.05 was used while Benjamini and Hochberg corrections for multiple comparisons were implemented to mitigate the presence of false positive findings due to the number of comparisons made (Benjamini and Hochberg, 1995).

1.6 Results

1.6.1 Time-Varying Perturbation Responses

Detailed graphs representing time-varying perturbation characteristics (position, velocity, acceleration) and performance compared to theoretical programmed values can be found in Figures 2-17 – 2-29. These graphs demonstrate how the platform acceleration performed as a 2nd order underdamped system, with actual acceleration magnitudes over-shooting and oscillating around the target value. While oscillations were present, the correlation of the peak platform acceleration to the peak programmed values was very strong and consistent ($r^2 = 0.998$, $p < 0.001$).

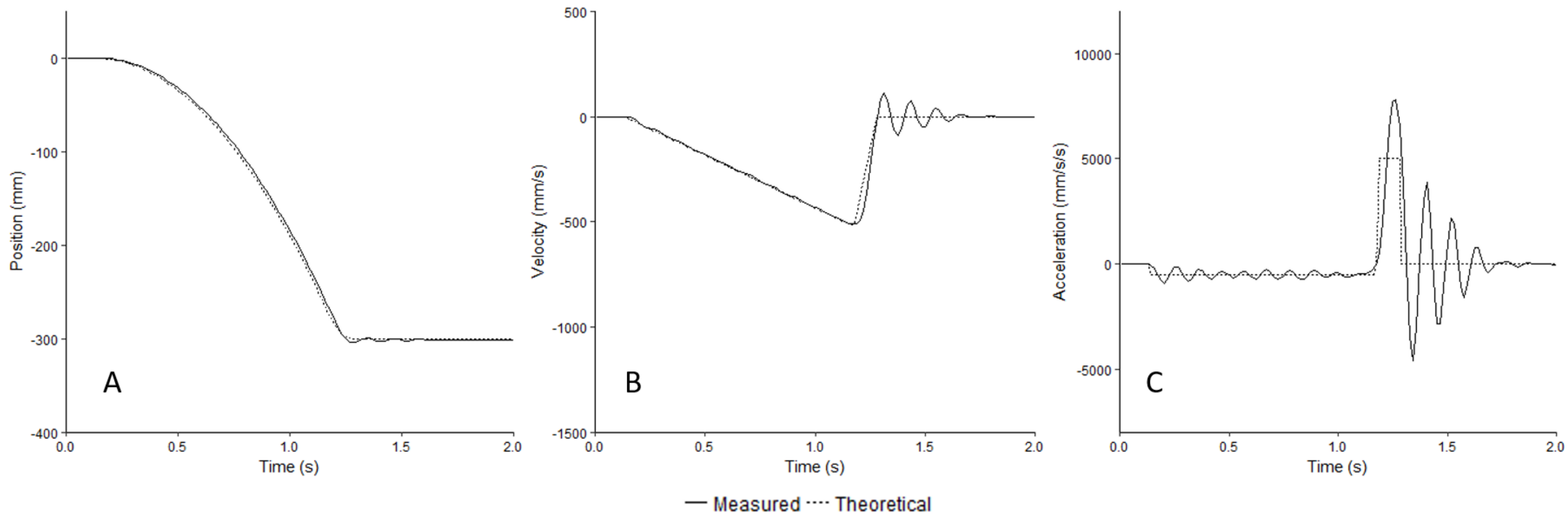

Figure 0-17: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 0.5 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

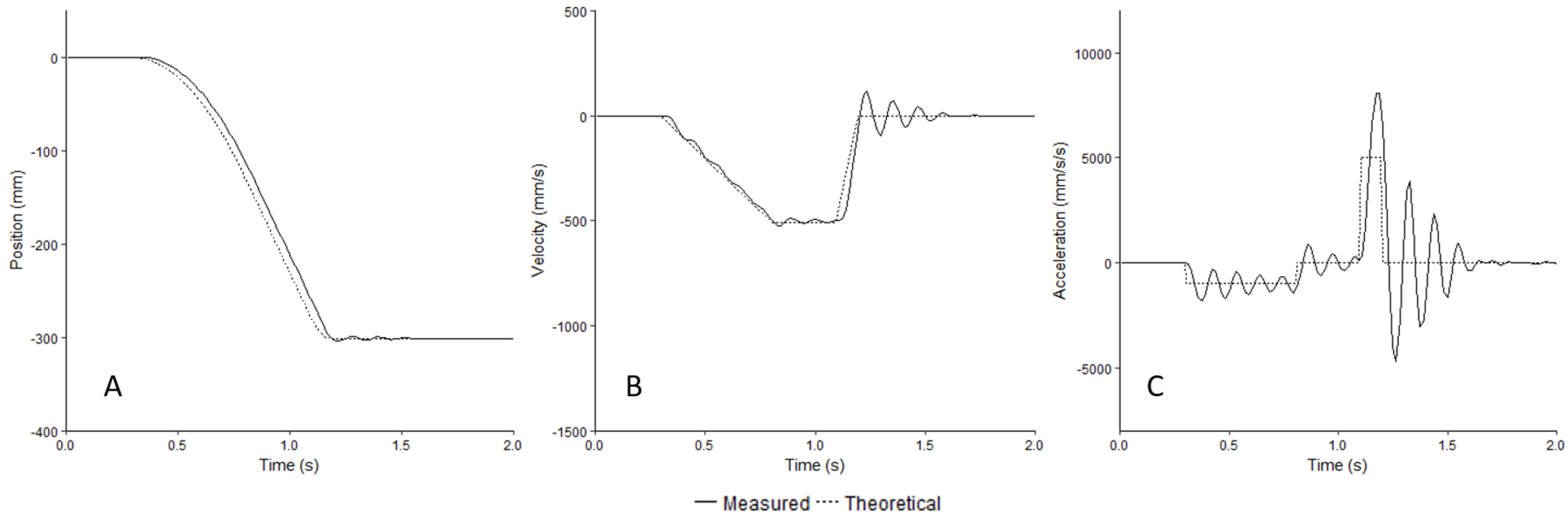

Figure 0-18: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 1.0 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

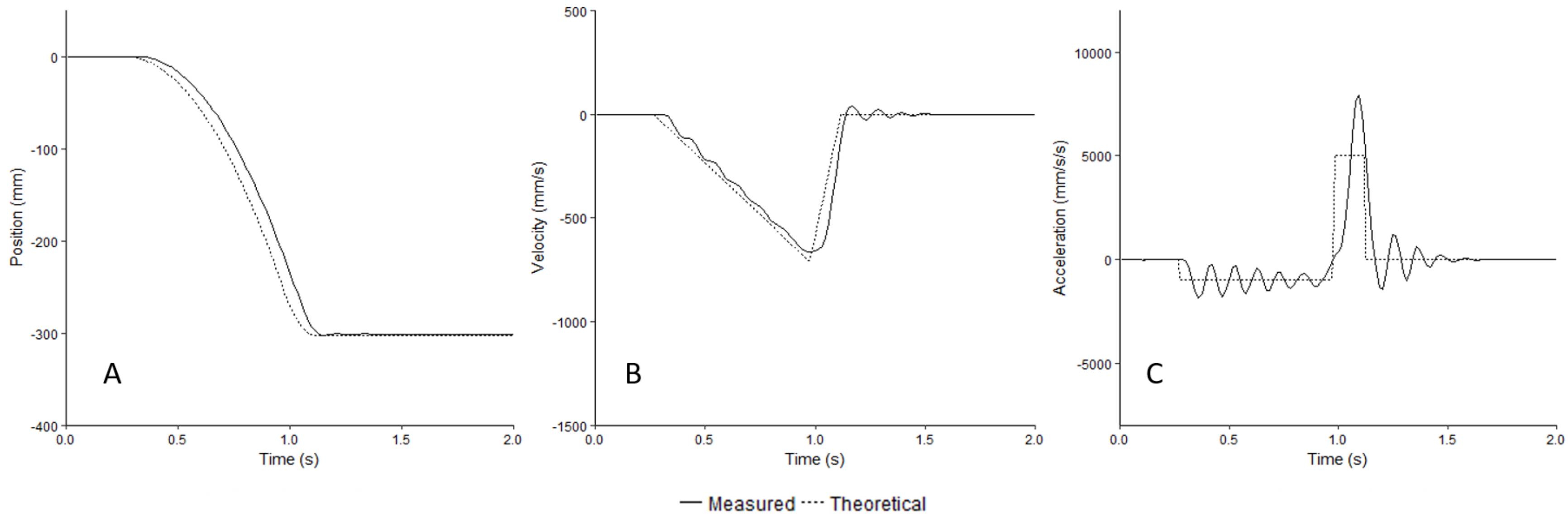

Figure 0-19: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 1.0 m/s^2 acceleration with a target peak velocity of 0.65 m/s and 0.30 m displacement.

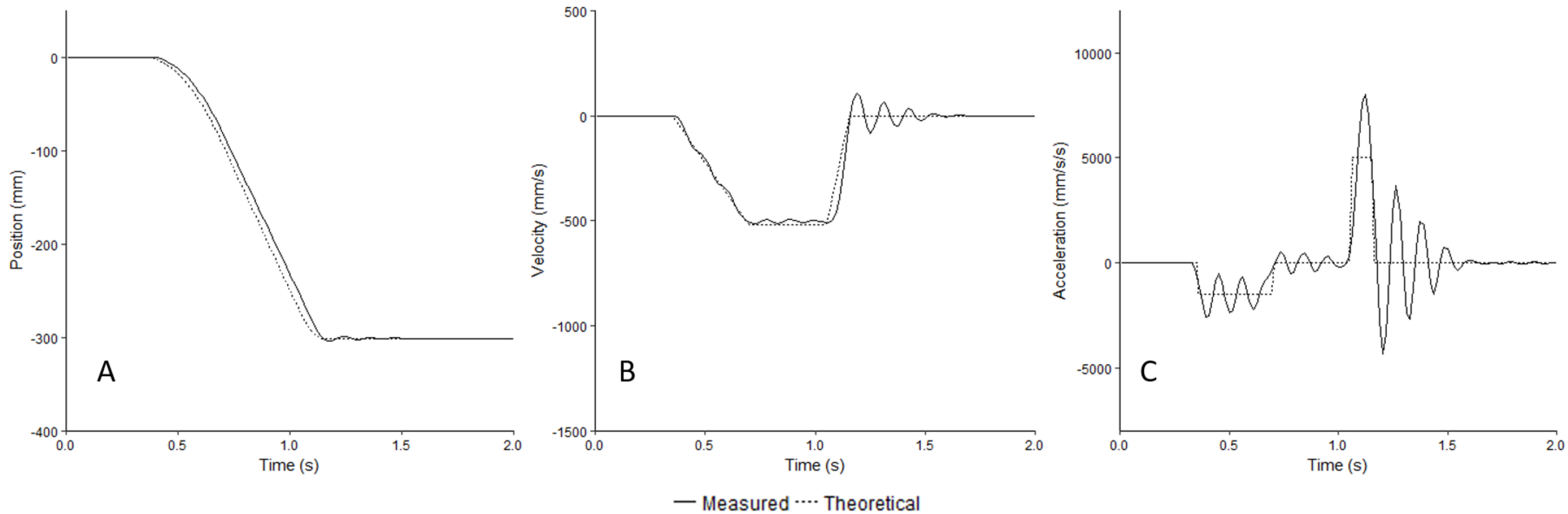

Figure 0-20: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 1.5 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

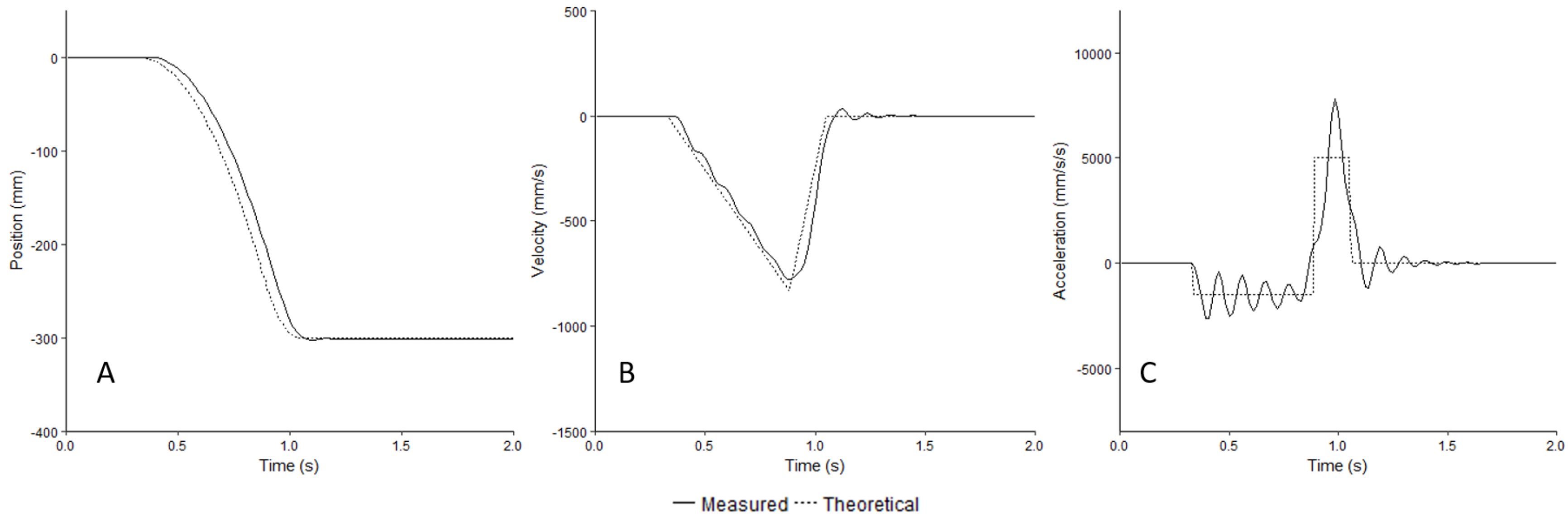

Figure 0-21: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 1.5 m/s² acceleration with a target peak velocity of 0.75 m/s and 0.30 m displacement.

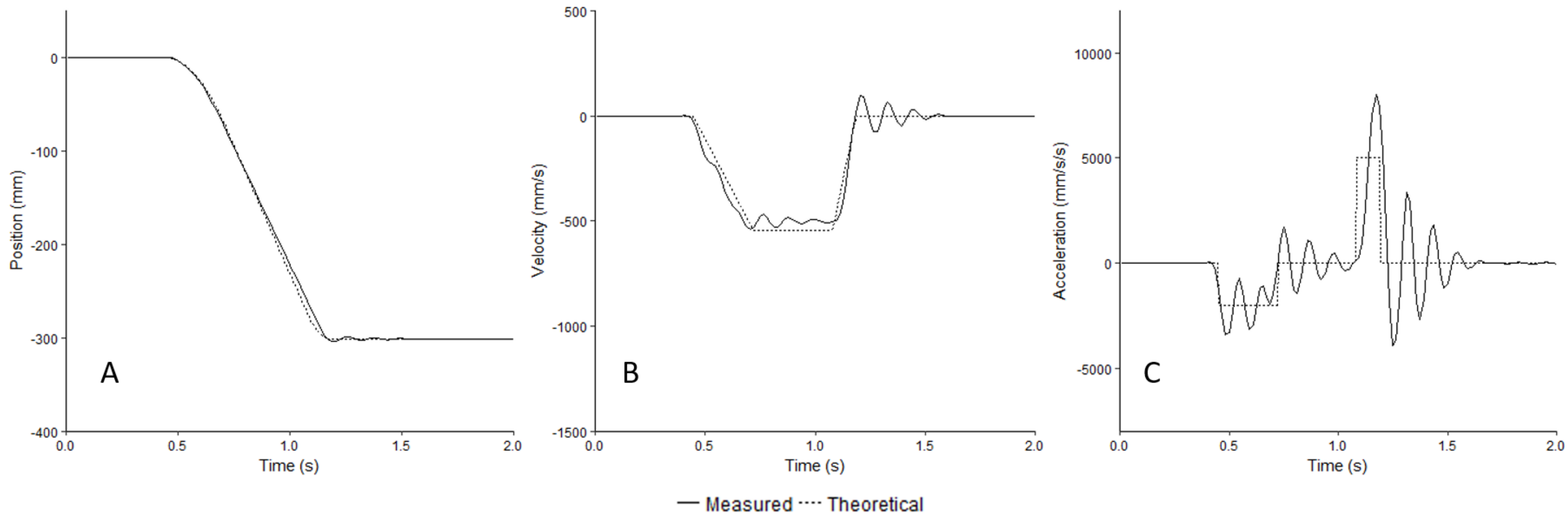

Figure 0-22: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 2.0 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

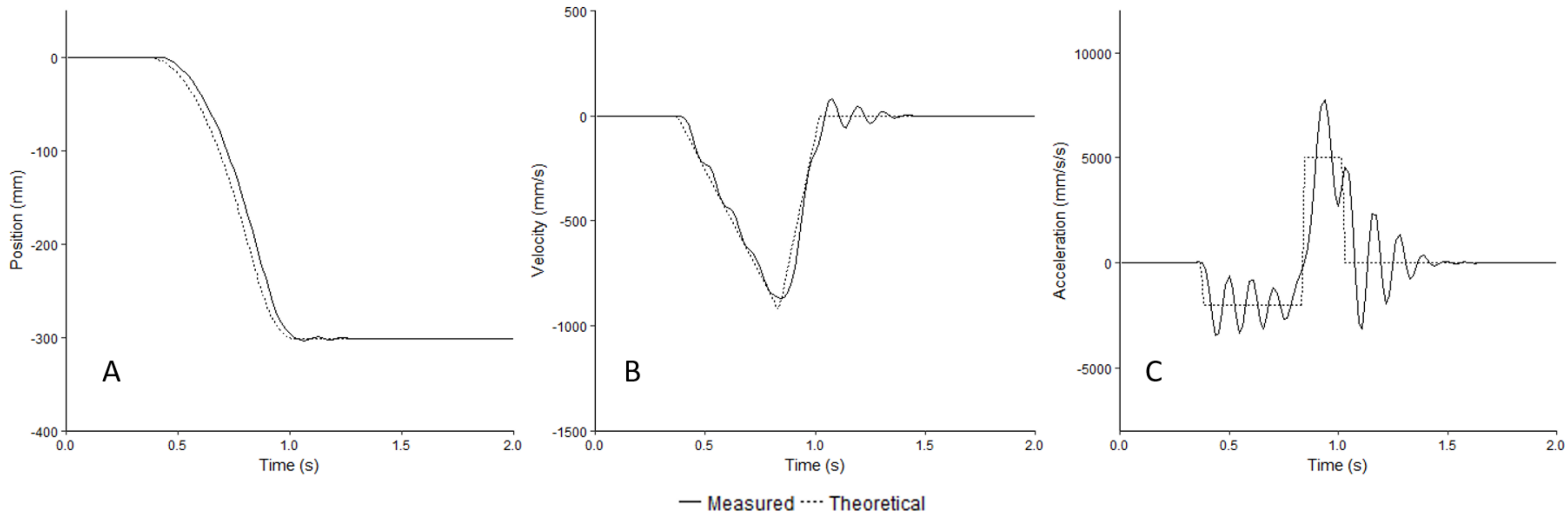

Figure 0-23: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 2.0 m/s^2 acceleration with a target peak velocity of 0.85 m/s and 0.30 m displacement.

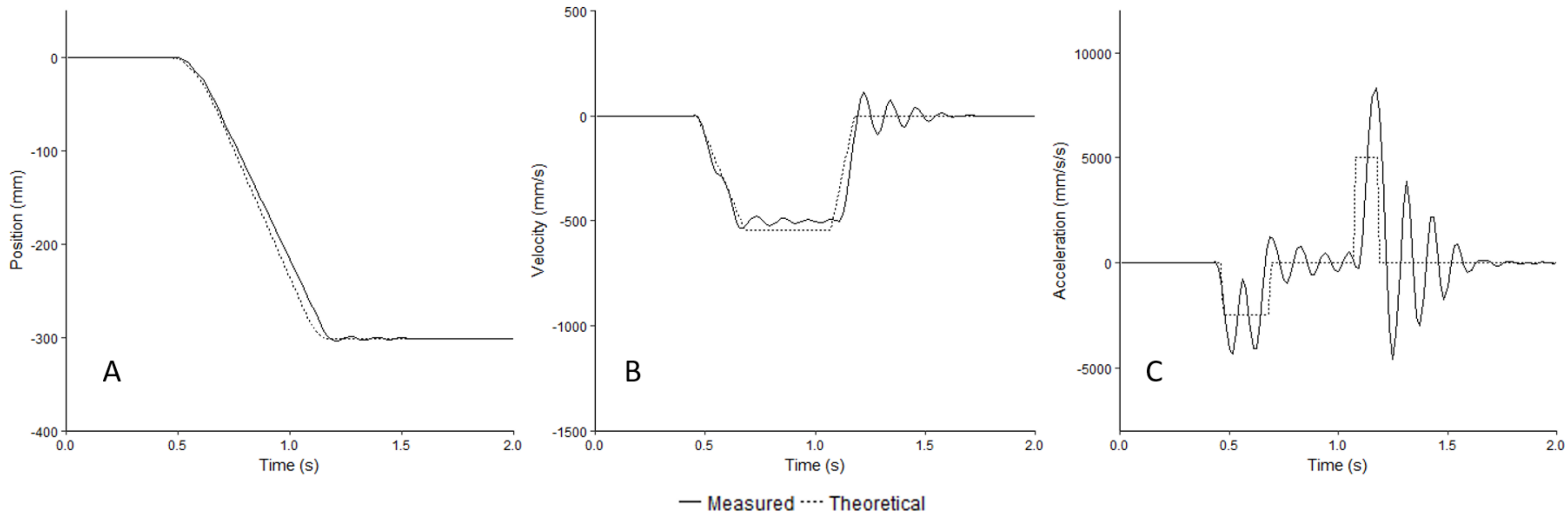

Figure 0-24: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 2.5 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

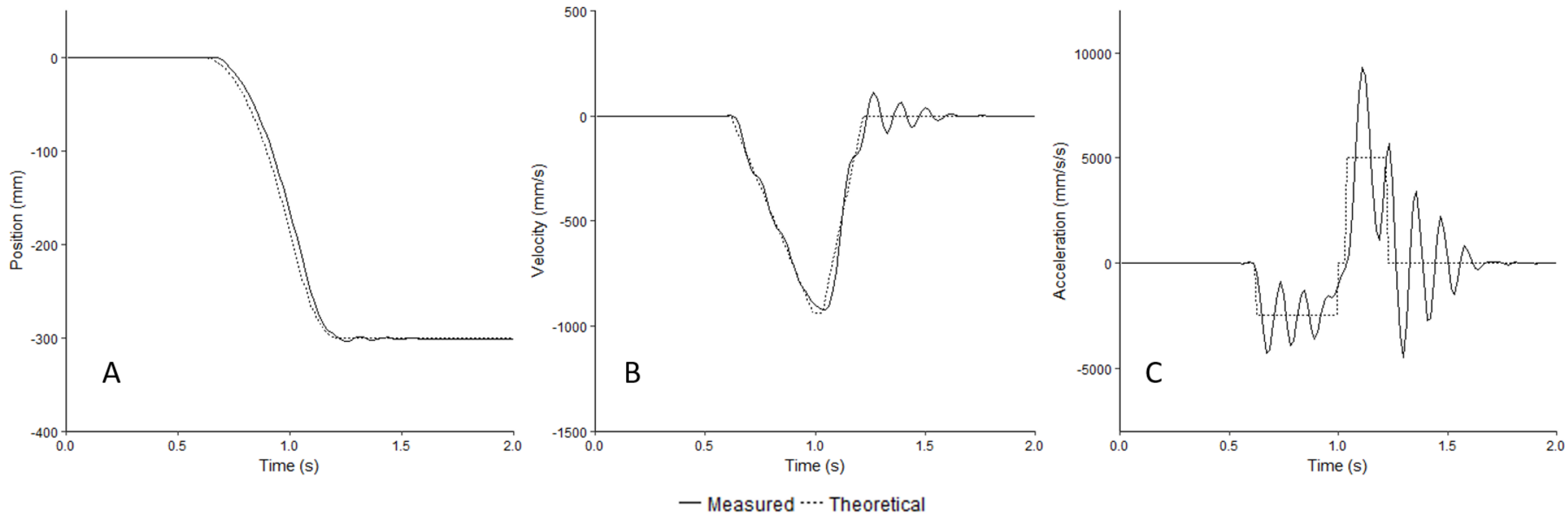

Figure 0-25: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 2.5 m/s^2 acceleration with a target peak velocity of 0.90 m/s and 0.30 m displacement.

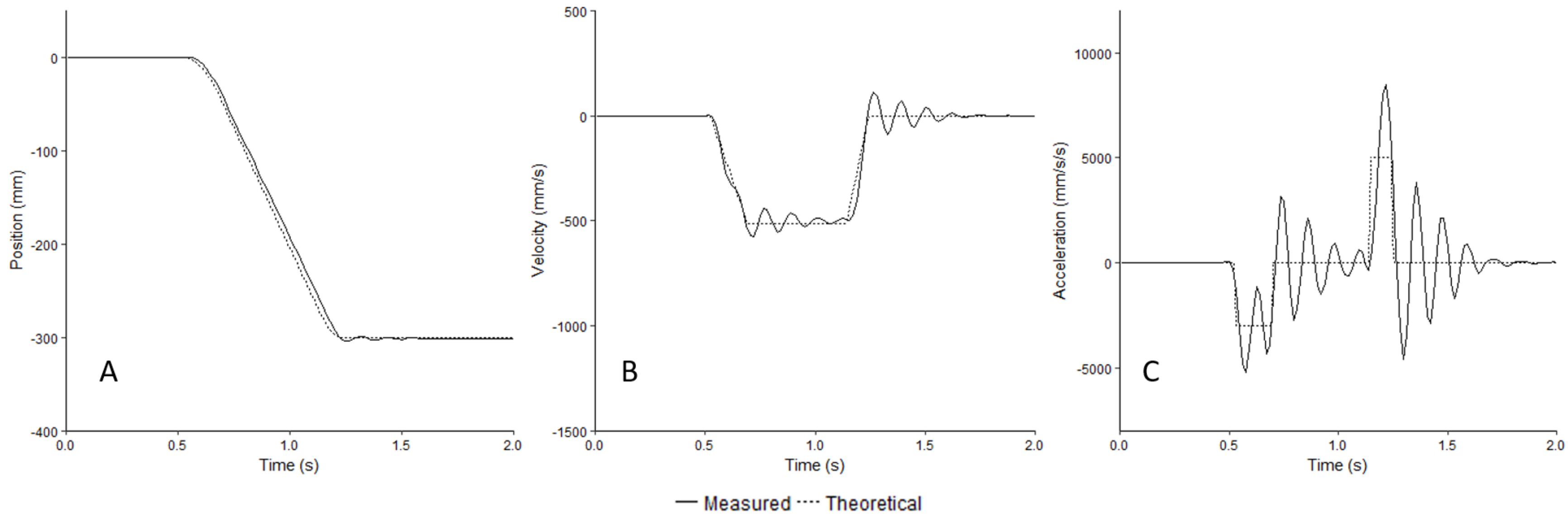

Figure 0-26: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 3.0 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

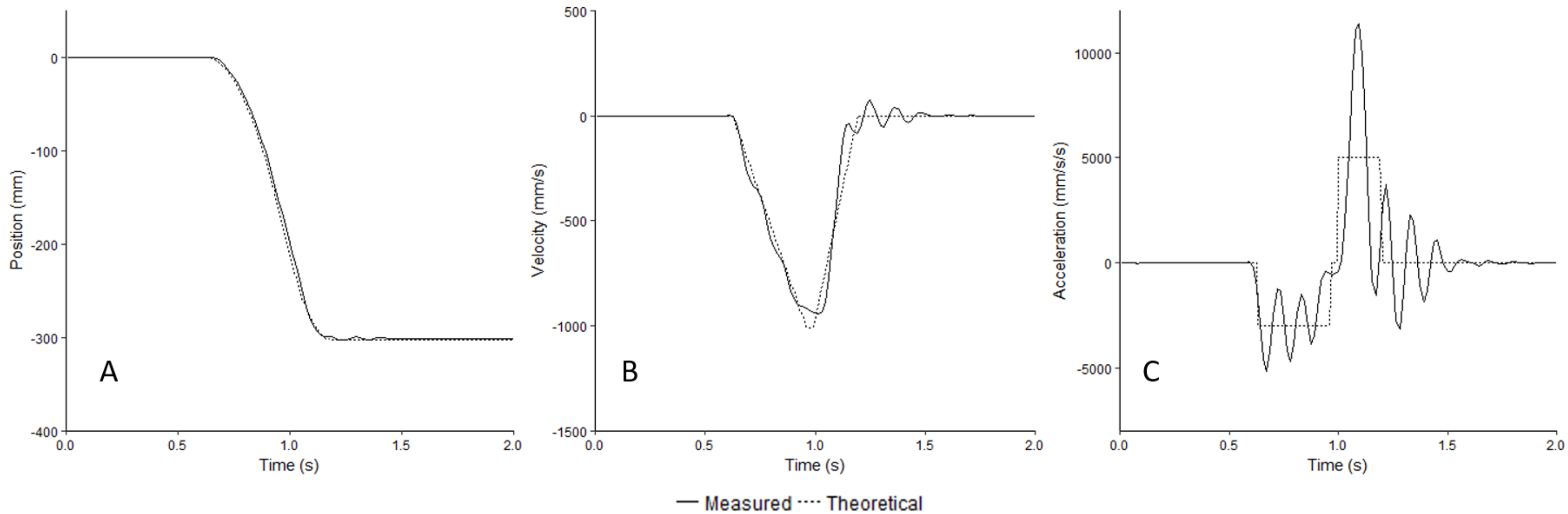

Figure 0-27: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 3.0 m/s^2 acceleration with a target peak velocity of 0.95 m/s and 0.30 m displacement.

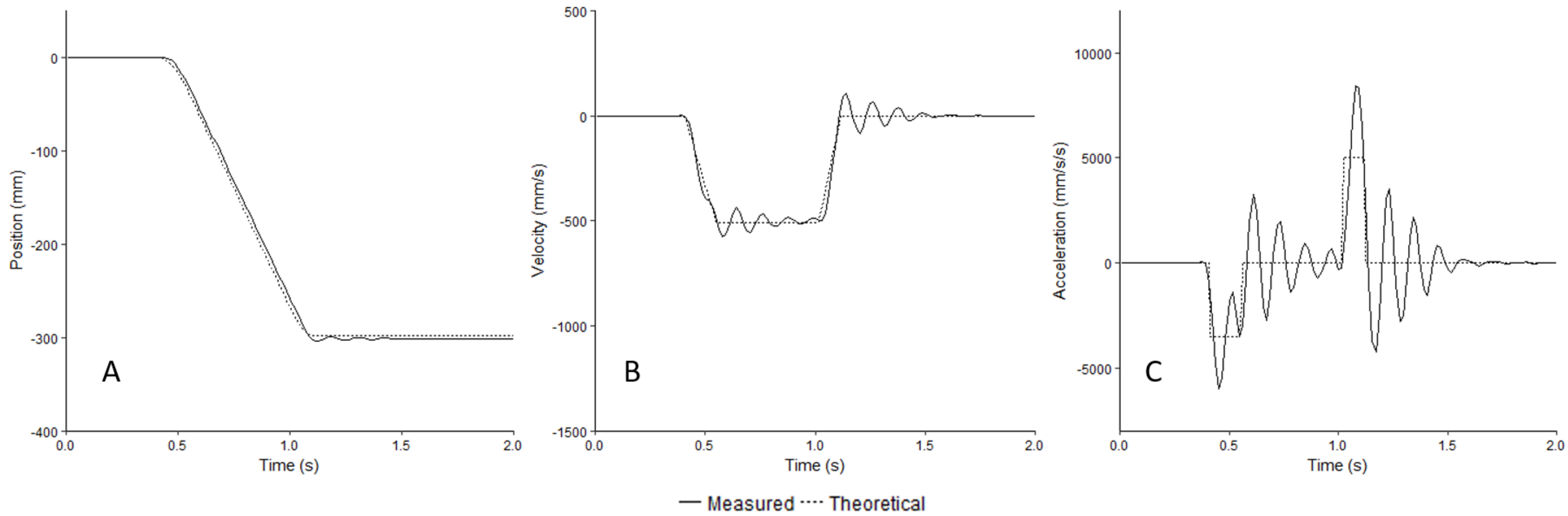

Figure 0-28: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 3.5 m/s^2 acceleration with a target peak velocity of 0.50 m/s and 0.30 m displacement.

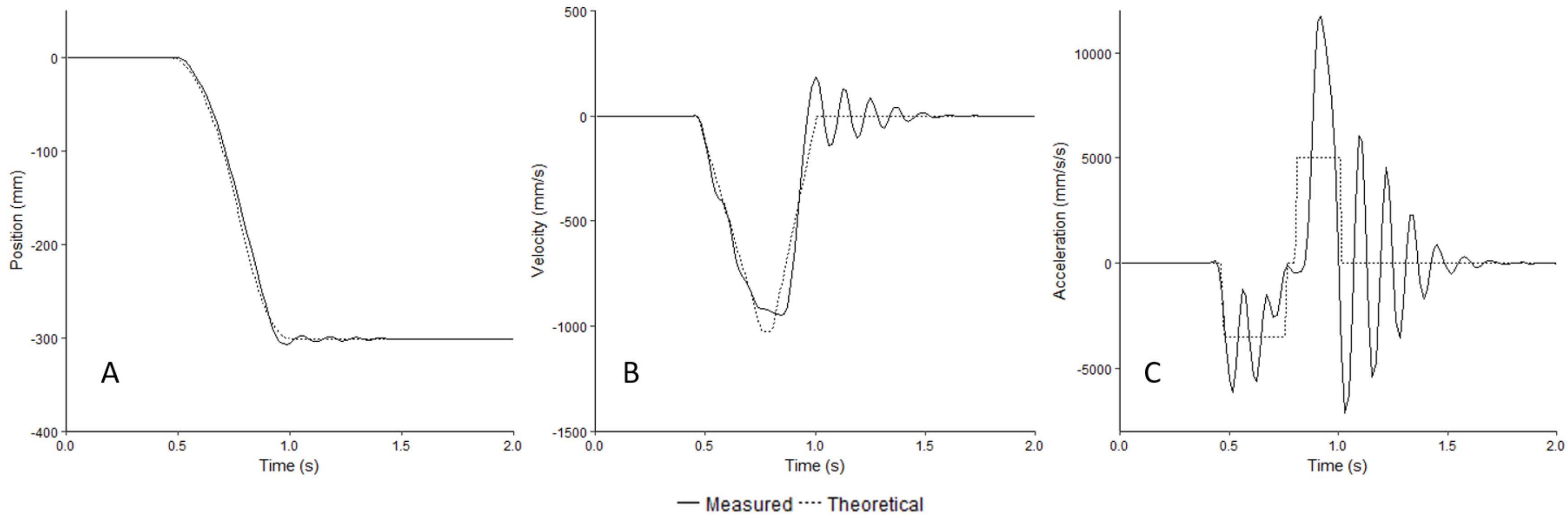

Figure 0-29: Comparison of platform position (A), velocity (B), and acceleration (C) of measured (solid line) to theoretical (dashed line) programmed values for the 3.5 m/s^2 acceleration with a target peak velocity of 1.00 m/s and 0.30 m displacement.

Appendix A contains graphs depicting some of the time varying results comparing AP COM position, AP stepping foot heel position, and raw stepping leg gastrocnemius EMG. This appendix demonstrates three different trials from three different participants and how similar responses were elicited in agreement with previous literature (Chen et al., 2014; Henry et al., 1998; Maki et al., 1996; McIlroy and Maki, 1993). The agreement between the current study and previous works provides confidence that the responses observed were not atypical.

1.6.2 Stepping Results

Across 896 backward perturbations, participants successfully maintained their balance using a single step response in 96.4% (864 trials) of trials. No steps and multiple steps accounted for 2.5% (22 trials) and 1.1% (10 trials) of trials, respectively. Twenty of the 22 no step and two of the 10 multistep responses occurred during the 0.5 m/s^2 acceleration level (this condition was not subsequently included in the statistical analysis).

1.6.3 Step Length

Mean normalized step length ranged from 45.5% leg length to 60.3% leg length across all conditions. A significant interaction was observed between platform acceleration and peak platform velocity ($F_{5, 105} = 21.6$, $p <.0001$) on step length. Statistically significant main effects for both platform acceleration ($F_{5, 105} = 18.2$, $p < .0001$) and peak platform velocity ($F_{1, 21} = 194.3$, $p <.0001$) were also observed. Figure 2-30 below depicts the changes in normalized step length as peak platform velocity and platform acceleration were varied.

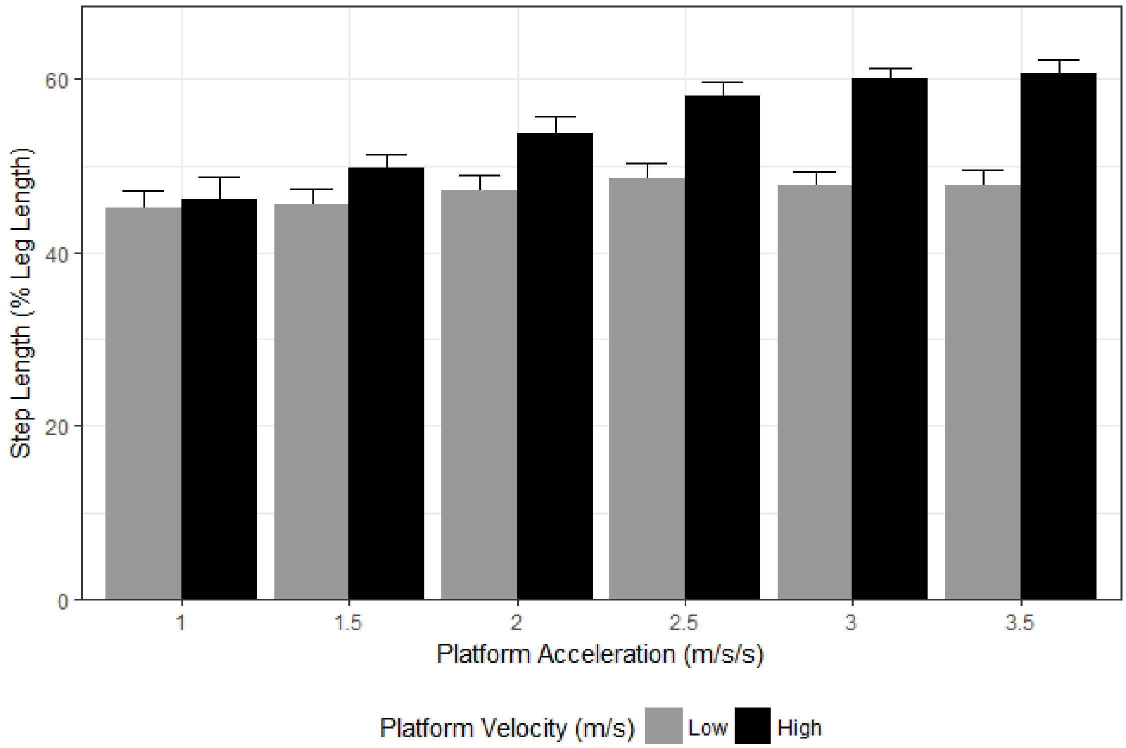

Figure 0-30: Mean (SE) values for normalized step length across platform acceleration and velocity.

Within Platform Acceleration Levels: Increasing peak platform velocity while maintaining platform acceleration level resulted in increased step length at platform accelerations of 1.5 m/s^2 (8.9% increase, $p = 0.001$), 2.0 m/s^2 (13.8% increase, $p < .001$), 2.5 m/s^2 (19.6% increase, $p < .001$), 3.0 m/s^2 (25.4% increase, $p < .001$), 3.5 m/s^2 (26.8% increase, $p < .001$). Increasing the peak platform velocity did not result in any change to participants step length when the platform acceleration was 1.0 m/s^2. All comparisons made within platform acceleration levels are outlined in Table 2-3 below.

Table 0-3: F (p) values for within platform acceleration comparisons of normalized step length with significant differences denoted*.

1.0 Low – 1.0 High	1.5 Low – 1.5 High	2.0 Low – 2.0 High	2.5 Low – 2.5 High	3.0 Low – 3.0 High	3.5 Low – 3.5 High
0.215 (0.648)	14.675 (0.001*)	40.337 (<0.001*)	65.646 (<0.001*)	200.581 (<0.001*)	206.975 (<0.001*)

Within Low Peak Platform Velocity Level: There was no significant effect of platform acceleration on step length during the low peak platform velocity conditions ($F_{5,\,105} = 1.9$, $p = .150$).

Within High Peak Platform Velocity Level: A significant effect of platform acceleration was observed while maintaining a high peak platform velocity ($F_{5,\,105} = 32.3$, $p < .001$). Increasing platform acceleration from 1.0 m/s^2 to 1.5 m/s^2 (7.5% increase, p = .027), 2.0 m/s^2 (15.7% increase, p < .001), 2.5 m/s^2 (25.9% increase, p < .001), 3.0 m/s^2 (29.6% increase, p < .001), or 3.5 m/s^2 (30.7% increase, p < .001) resulted in increased normalized step length. Likewise, increasing platform acceleration from 1.5 m/s^2 to 2.0 m/s^2 (7.7% increase, p = .002), 2.5 m/s^2 (17.1% increase, p < .001), 3.0 m/s^2 (20.6% increase, p < .001), or 3.5 m/s^2 (21.6% increase, p < .001) resulted in increased normalized step length. Increases from 2.0 m/s^2 to 2.5 m/s^2 (8.8% increase, p = .002), 3.0 m/s^2 (12.0% increase, p < .001), or 3.5 m/s^2 (12.9% increase, p < .001) and increases from 2.5 to 3.0 m/s^2 (3.0% increase, p = .045) or 3.5 m/s^2 (3.8% increase, p = .012) all resulted in increased normalized step length. All comparisons made within high peak platform velocity are outlined in Table 2-4 below.

Table 0-4: F (p) values for within high velocity comparisons of normalized step length with significant differences denoted*.

	1.5 High	2.0 High	2.5 High	3.0 High	3.5 High
1.0 High	5.687 (.027*)	34.324 (<.001*)	38.384 (<.001*)	41.483 (<.001*)	38.546 (<.001*)
1.5 High		12.53 (.002*)	87.766 (<.001*)	69.153 (<.001*)	74.891 (<.001*)
2.0 High			13.123 (.002*)	20.003 (<.001*)	23.231 (<.001*)
2.5 High				4.503 (.045*)	7.415 (.012*)
3.0 High					0.476 (.497)

1.6.4 Extrapolated COM

Mean AP xCOM displacement ranged from 0.384 m to 0.419 m across all conditions. A significant interaction was observed between platform acceleration and peak platform velocity ($F_{5, 105} = 3.4$, $p = .022$) as well as statistically significant main effect of platform acceleration ($F_{5, 105} = 5.4$, $p < .001$). No significant effect of peak platform velocity ($F_{1, 21} = 1.0$, $p = .320$) was observed. Figure 2-31 below depicts the changes in AP xCOM displacement as peak platform velocity and platform acceleration are varied.

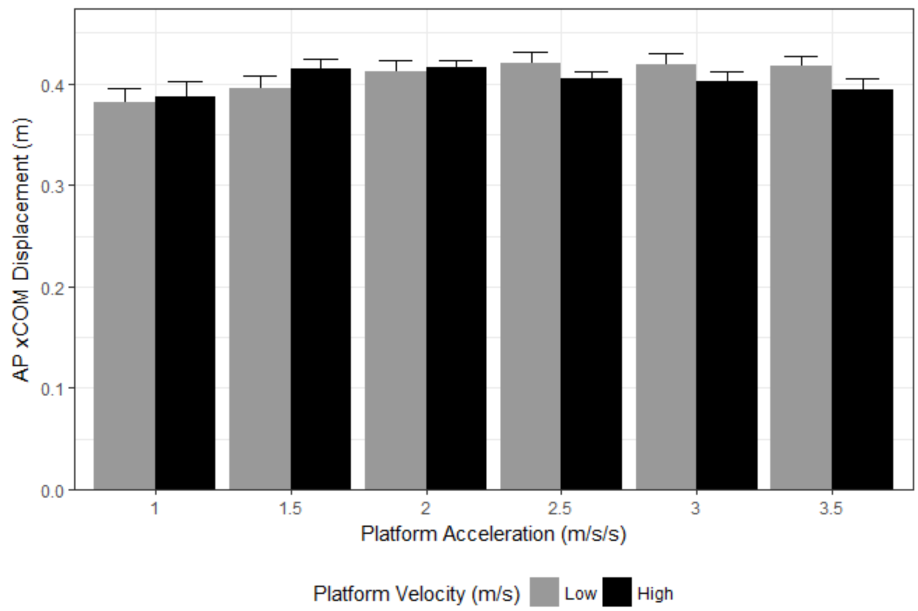

Figure 0-31: Mean (SE) values for AP xCOM displacement across platform acceleration and velocity.

Within Platform Acceleration Levels: Changing the peak platform velocity from low to high did not result in any significant changes in xCOM displacement regardless of platform acceleration level.

Within Low Peak Platform Velocity Level: A significant effect of platform acceleration was observed while maintaining a low peak platform velocity ($F_{5, 105} = 10.4$, $p < .001$). Maintaining a low peak platform velocity but increasing the platform acceleration from 1.0 m/s^2 to 2.0 m/s^2 (7.5% increase, $p < .001$), 2.5 m/s^2 (9.1% increase, $p < .001$), 3.0 m/s^2 (9.2% increase, $p < .001$) or 3.5 m/s^2 (8.7% increase, $p < .001$) resulted in increased AP xCOM displacement. Likewise, increasing platform acceleration from 1.5 m/s^2 to 2.0 m/s^2 (4.3% increase, $p = .009$), 2.5 m/s^2 (5.9% increase, $p = .002$), 3.0 m/s^2 (6.0% increase, $p = .002$) or

3.5 m/s^2 (5.6% increase, p = .001) resulted in increased AP xCOM displacement. All comparisons made within low peak platform velocity are outlined in Table 2-5 below.

Table 0-5: F (p) values for within low velocity comparisons of AP xCOM displacement with significant differences denoted*.

	1.5 Low	2.0 Low	2.5 Low	3.0 Low	3.5 Low
1.0 Low	4.739	18.126	21.69	17.019	23.419
	(.04)	(<.001*)	(<.001*)	(<.001*)	(<.001*)
1.5 Low		8.255	12.399	13.095	14.744
		(.009*)	(.002*)	(.002*)	(<.001*)
2.0 Low			2.007	1.228	0.747
			(0.171)	(0.28)	(0.397)
2.5 Low				0.001	0.306
				(0.977)	(.586)
3.0 Low					0.155
					(.697)

Within High Peak Platform Velocity Level: A significant effect of platform acceleration was not observed while maintaining a high peak platform velocity ($F_{5, 105}$ = 2.0, p = .128). Maintaining a high peak platform velocity while increasing platform acceleration resulted in no significant changes in the AP xCOM displacement.

1.6.5 Extrapolated MOS

Mean minimum xMOS ranged from 0.087 m to 0.201 m across all conditions. A significant interaction was observed between platform acceleration and peak platform velocity ($F_{5, 105}$ = 24.7, p <.001) as well as statistically significant main effects for both platform acceleration ($F_{5, 105}$ = 8.6, p = .001) and peak platform velocity ($F_{1, 21}$ = 52.7, p

<.001). Figure 2-32 below depicts the changes in minimum AP xMOS as peak platform velocity and platform acceleration are varied.

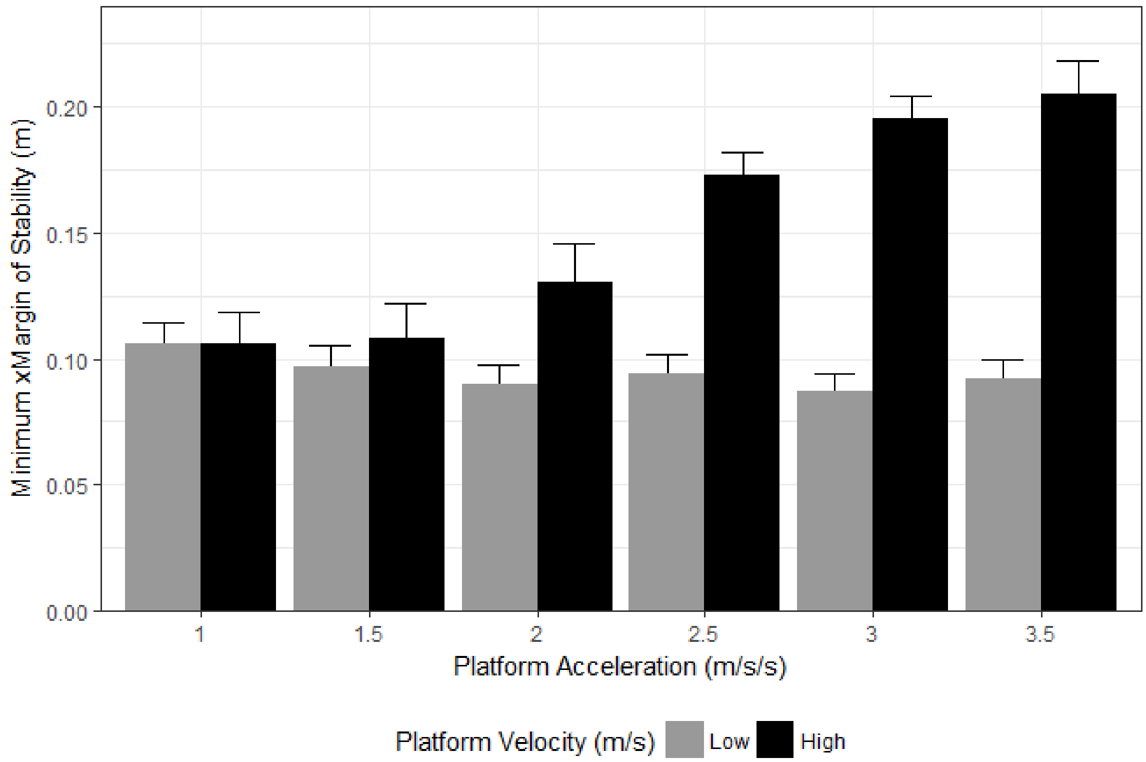

Figure 0-32: Mean (SE) values for minimum AP xMOS across platform acceleration and velocity.

Within Platform Acceleration Levels: Increasing target peak velocity from low to high resulted in significant AP xMOS increases at acceleration levels of 2.0 m/s^2 (52.1% increase, $p = 0.017$), 2.5 m/s^2 (84.8% increase, $p < .001$), 3.0 m/s^2 (121.7% increase, $p < .001$) and 3.5 m/s^2 (127.6% increase, $p < .001$). Remaining platform acceleration levels (1.0 m/s^2 and 1.5 m/s^2) resulted in no significant changes to minimum xMOS. All comparisons made within platform accelerations are outlined in Table 2-6 below.

Table 0-6: F (p) values for within platform acceleration comparisons of minimum AP xMOS with significant differences denoted*.

1.0 Low – 1.0 High	1.5 Low – 1.5 High	2.0 Low – 2.0 High	2.5 Low – 2.5 High	3.0 Low – 3.0 High	3.5 Low – 3.5 High
0.000 (.993)	0.78 (.387)	6.724 (.017*)	72.139 (<0.001*)	140.003 (<0.001*)	101.958 (<0.001*)

Within Low Peak Platform Velocity Level: A significant effect of platform acceleration was not observed while maintaining a low peak platform velocity ($F_{5, 105} = 1.7$, $p = .164$). Maintaining a low peak platform velocity but increasing the platform acceleration did not result in any statistically significant increases in AP xMOS regardless of the acceleration level tested.

Within High Peak Platform Velocity Level: A significant effect of platform acceleration was observed while maintaining a high peak platform velocity ($F_{5, 105} = 18.0$, $p < .001$). Maintaining a high peak platform velocity but increasing platform acceleration from 1.0 m/s^2 to 2.0 m/s^2 (25.4% increase, p = 0.033), 2.5 m/s^2 (62.7% increase, p <.001), 3.0 m/s^2 (83.6% increase, p <.001) or 3.5 m/s^2 (90.4% increase, p <.001) resulted in a statistical increase in minimum AP xMOS. Similarly, increasing the platform acceleration from 1.5 m/s^2 to 2.5 m/s^2 (56.8% increase, p <.001), 3.0 m/s^2 (76.9% increase.6, p <.001), or 3.5 m/s^2 (83.5% increase, p <.001) resulted in a statistical increase in minimum AP xMOS when peak velocity was high. Increasing platform acceleration from 2.0 m/s^2 to 2.5 m/s^2 (29.8% increase, p = .005), 3.0 m/s^2 46.5% increase, p <.001), or 3.5 m/s^2 (51.9% increase, p = .001), also significantly increased participants minimum xMOS as did increasing the platform acceleration from 2.5 m/s^2 to 3.5 m/s^2 (17.0% increase, $p = .003$). The remaining platform acceleration changes (1.0 m/s^2 to 1.5 m/s^2 and 3.0 m/s^2 to 3.5 m/s^2) did not result in a significant change in minimum xMOS. All comparisons made within high peak platform velocity are outlined in Table 2-7 below.

Table 0-7: F (p) values for within high velocity comparisons of minimum AP xMOS with significant differences denoted*.

	1.5 High	2.0 High	2.5 High	3.0 High	3.5 High
1.0 High	0.154 (.699)	5.199 (.033*)	27.351 (<.001*)	42.583 (<.001*)	28.102 (<.001*)
1.5 High		3.339 (.081)	22.607 (<.001*)	29.617 (<.001*)	23.445 (<.001*)
2.0 High			9.953 (.005*)	19.153 (<.001*)	15.219 (<.001*)
2.5 High				9.246 (.006*)	11.279 (.003*)
3.0 High					1.15 (0.295)

1.7 Discussion and Conclusions

The aim of this study was to identify potential effects of platform acceleration and peak velocity on measures of balance control and stability during dynamic stepping. Supporting the first hypothesis, step length significantly increased as peak platform velocity and platform acceleration increased (up to 26.8% and 30.7% respectively). Regarding increasing platform acceleration, only the high peak velocity group demonstrated changes in step length whereas the low peak velocity group did not. Contrary to the second hypothesis, peak platform velocity did not have an effect on xCOM displacement. Platform acceleration was found to have an effect on xCOM displacement (up to 9.2% increase) during low peak velocity trials but not during high peak velocity trials. Lastly, in disagreement with the third hypothesis, xMOS was increased (up to 127.6% increase) rather than decreased by an increase in peak platform velocity. Increasing platform acceleration was found to have no effect on xMOS during low peak velocity trials and significantly increased xMOS (up to 90.4% increase) during high peak velocity trials, again, contrary to the third hypothesis.

The outcomes observed were similar to previous research examining single leg stepping responses. Utilizing anthropometric data to convert all data to comparable units (de Leva, 1996), previous research using this paradigm has found participants step lengths to vary from approximately 34-42% of leg length (Inkol et al., 2018b; Maki et al., 1996; McIlroy and Maki, 1996) while the current study found a range of 45-60% leg length. While these ranges of magnitudes do not overlap, the process of creating comparable units results in error of the measures as the current studies values are based on participant specific values from digitized landmarks and the previous works values have been converted based on anthropometric tables. Previous works which report COM displacement range in values of 0.06-0.16 m (Chen et al., 2014; Henry et al., 1998; McIlroy and Maki, 1996). These values are substantially lower than the values observed in the current study which had a range of 0.38-0.42 m, however the current study used substantially larger perturbations and examined xCOM displacement rather than COM displacement. When comparing to more recent literature that also examined xCOM, values of approximately 0.3-0.35 m were observed (Inkol et al., 2018b). Based on the lack of consistency between COM and xCOM displacement from the literature, comparing xMOS values of the current study, 0.09-0.20 m, to previous work, 0.09 m (Inkol et al., 2018b), demonstrates that comparable responses were observed. Much of the discrepancies observed between the measures may be attributed to potential error in conversion of units for comparison via anthropometric data as well as different perturbation parameters used.

While the responses observed were comparable to previous literature, the performance of the platform characteristics, specifically platform acceleration, demonstrated unique profiles compared to reported characteristics of other researchers systems (Mansfield and Maki, 2009; Norrie et al., 2002; Quant et al., 2005, 2004; Rajachandrakumar et al., 2018). The system used for this study demonstrated greater inertial effects, and/or reduced damping, compared to the graphical representations provided by other researchers. While this does indicate the system provides a unique stimulus to the participant, the correlation between measured peak acceleration and programmed value produced an r^2 of 0.998 ($p<0.001$) indicating that while all measured peak accelerations were larger than intended, they were consistently larger. This strong correlation provides confidence that the platform perturbations were consistent between participants and between trials providing confidence that variability was not introduced due to perturbation system performance.

Step length changes are driven by an interaction between platform acceleration and velocity and generate thresholds where changes start and stop to occur. By supporting the corresponding hypothesis, step length follows the expected trend of increasing the base of support as the postural threat increases in magnitude. When comparing within acceleration levels the step length does not significantly change until the 1.5 m/s^2 level, demonstrating how peak velocity is not discriminatory at the lower acceleration. This threshold is important to note as a large portion of previously published literature utilizing an acceleration value at or below this threshold (Henry et al., 1998; Kam et al., 2016; Maki et al., 2000; Nonnekes et al., 2013; Norrie et al., 2002; Quant et al., 2005, 2004; Tokuno et al., 2010; Zettel et al., 2005). This indicates that their velocity selection does not appear to be of large concern when comparing these study results for normalized step length. However, being above this acceleration threshold raises the importance of the selected peak platform velocity. By increasing the acceleration, the selection of peak velocity becomes the discriminating factor, which can be seen when examining the differences in step length within an acceleration group in Figure 2-30 (comparing grey to adjacent black bars). However, unlike the majority of comparisons made between high peak velocity trials, no changes in step length were observed when comparing the 3.0 m/s^2 to the 3.5 m/s^2 when peak platform velocity was high. This may be attributed to participants approaching their functional maximum step length from a standing position during the most challenging perturbations. Combining anthropometric tables (determine average leg length) (de Leva, 1996) with previously reported step lengths during gait (Barreira et al., 2010) yields an average step length of over 85% leg length. While this value is still larger than the observed step lengths during the 3.0 m/s^2 acceleration level (59.8±1.3%) and the 3.5 m/s^2 acceleration level (60.3±1.5%), the ~85% step length during gait incorporates the inertial properties from previous movement as well as a stance limb which is ahead of the stepping limb. With all of these considerations, it is plausible that participants using step lengths of ~60% leg length are approaching their functional maximum based on the task performed.

Relative timing of stepping events provides further insights into the effects of platform acceleration and velocity on step length. Figure 2-33 depicts the timing of foot off (squares) and heel strike (plus signs) relative to the duration of platform acceleration (diamonds) and peak deceleration (triangles) with time of 0 seconds representing the onset of platform movement. This is a representation of the mean of each participant's data and shows how the event of foot off may be occurring before or after the acceleration phase has ended depending on the perturbation parameters. Likewise, heel strike timing is also influenced by the parameters selected as made evident by the general shape of the graph. Figure 2-33 demonstrates the differences seen between the low and high velocity trials relative to platform acceleration time. In the low velocity trials, with the exception of the 1.0 m/s^2 acceleration level, foot off consistently occurs after the end of platform acceleration. In contrast, in the high velocity trials (with the exception of the 3.5 m/s^2 acceleration level), foot off consistently occurs before platform acceleration has finished. However, while the relative timing of foot off and heel strike differ across perturbation condition, the relative timing between the two appear to remain relatively constant. Accordingly, the differences observed in step length across conditions do not appear to be a result of substantially different swing time. Another influential factor could be the onset of the deceleration phase and the potential re-stabilizing effect a predictable deceleration phase can produce (McIlroy and Maki, 1994). While having predictable deceleration timing can allow individuals to use the deceleration forces to their benefit, this study employed variable acceleration and velocity based on trial which subsequently resulted in variable onset of deceleration. The variability in deceleration onset likely mitigated the predictability of the deceleration phase and therefore reduced participant's ability to use the deceleration phase to help re-stabilize following perturbation. Investigating the sources of these differences (e.g. swing velocity, onset of platform deceleration) should be a focus of future work.

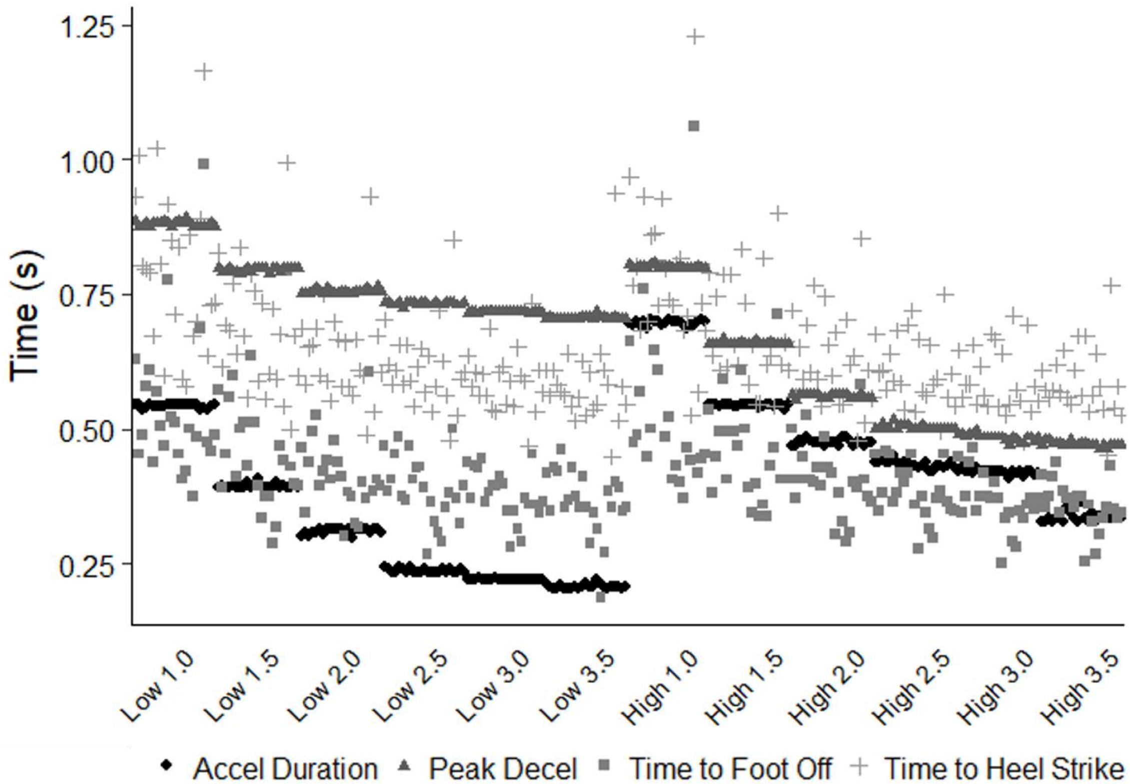

Figure 0-33: Comparison of the timing of foot off and heel strike to duration of platform acceleration and peak deceleration. Time is relative to the onset of platform movement, making time 0 s the onset of platform movement. Platform acceleration duration is depicted in diamonds, time of peak deceleration is depicted in triangles, time of foot off is depicted in squares, and time of heel strike is depicted in plus signs. All participant's mean responses across their trials are displayed and grouped into each perturbation condition. Low 1.0 represents the responses from the low velocity 1.0 m/s^2 acceleration trials; in contrast, the High 3.5 condition represents the responses from the high velocity 3.5 m/s^2 acceleration trials.

The relatively large effects of perturbation characteristics on step length likely drove the limited effects of perturbation parameters of xCOM. By increasing the step length, participants increased the moment arm available to themselves to generate a restabilizing torque via their resulting ground reaction forces. This increased torque could in turn have limited the xCOM displacement. This demonstrates how increasing step length serves to mechanically improve stability twofold: i) increased base of support, and ii) increase torque production to arrest COM movement.

By utilizing the step length results, some insight may be gained into the unexpected xCOM results. The xCOM demonstrates a nearly opposite response than the step length results with no significant changes occurring within acceleration levels. Also contrary to the step length results, for the high peak velocity trials there were no differences across acceleration conditions. There are also several observed significant differences when comparing low peak velocity groups between acceleration levels. This supports the concept that the step length increases previously discussed are aiding in the limitation of xCOM displacement. The only conditions in which xCOM displayed significant increases were within the low peak velocity between the 1.0 m/s^2 or the 1.5 m/s^2 acceleration conditions and all other tested acceleration conditions (2.0 m/s^2, 2.5 m/s^2, 3.0 m/s^2 and 3.5 m/s^2). These comparisons also demonstrated no significant differences between their step lengths but this demonstrates how if the step length were held constant and the perturbation acceleration was increased there is the expected response of an increase in xCOM displacement.

The general absence of change in the xCOM displacement and increase in the step length directly relate to xMOS. Interestingly, the hypothesized relationship was not observed, and in actuality, there was a significant effect in the opposite direction than that hypothesized. As explored previously, these unexpected results are driven by the significant increase in step length, and therefore base of support, and the unchanging xCOM displacement during the increased magnitude perturbations. While these findings are contrary to the hypothesized results they still provide valuable insights into the response mechanisms utilized when exposed to a greater magnitude perturbation. These findings suggest that as young healthy adults are exposed to greater postural threats their primary protective mechanism (step length) responds with a larger relative proportion. This increasing protective response is not present when peak velocity is held constant at a low magnitude but is observed during the high peak velocity trials. Examining the high peak velocity trials, there is no statistically significant increase in xMOS when the acceleration level increases from 1.0 m/s^2 to 1.5 m/s^2 as well as when increasing from 3.0 m/s^2 to 3.5 m/s^2. The plateau at highest end of the spectrum of the tested acceleration values leads to the possibility that the participants' xMOS could begin to follow the hypothesized trend of decreasing with increasing perturbation acceleration if more magnitudes were tested. While the platform acceleration would continue to increase, the normalized step length would remain constant as it has approached its functional maximum (as described previously). The same plateau is observed in the step length outcome with no statistical difference between 3.0 m/s^2 and 3.5 m/s^2 with a high peak velocity. Assuming the trend identified previously regarding maintaining a consistent step length continues, the participants xCOM displacement would begin to increase (as observed in the low velocity trials where step length was consistent).

This study provides important data to help future researcher's select surface translation parameters for balance control research. By examining an array of parameters this study identified the relationship between these parameters and commonly reported measures in the literature. For future research, platform accelerations of at least 3.0 m/s^2 and higher peak velocities appear to sufficiently challenge younger, healthy adults to employ their maximum single step response. This recommendation aligns with earlier works of Maki et al. who that identified that 3.0 m/s^2 acceleration, 0.90 m/s velocity, and 0.27 m displacement consistently elicited a successful single forward step response. It is important to note that these relationships have not yet been established in other populations who are frequently studied due to their higher risk of falling (i.e. older adults, Parkinson's patients, stroke patients, etc.). Similar trends could be present in the populations but at differing thresholds due to the changes in balance control experienced which place them at a higher risk of falling.

This study had several limitations. First, its ability to separate the effects of platform velocity and acceleration during the high peak velocity trials is limited as the peak velocity increased as the acceleration increased. This increased the range of values tested as well as maintained the same waveform of perturbation as the platform accelerated until it reached its maximum achievable velocity prior to decelerating within the constant displacement. Participants were also instructed to respond with a single step if they were going to step which could have increased the likelihood of stepping responses occurring and potentially falsely increasing their step lengths to ensure compliance with the one step outline given. As participants were aware they were engaging in a postural perturbation research study their postural responses may also be altered, as the element of surprise that is typically present during real-world losses of balance is absent and therefore systems may be primed for responses.

In conclusion, the measured response of the system is unique and different from both the theoretical step response of the programmed profiles and the measured responses from other systems described in the literature; however, the system was extremely consistent and repeatable and therefore did not introduce variability into the testing paradigm. Based on the results of this study and its inherent limitations, both peak platform velocity and platform acceleration play important roles in the balance response outcomes measured. The examined population demonstrated increased protective responses (normalized step length and minimum AP xMOS) to ensure successful balance recovery as perturbation magnitude increased, contrary to hypothesized results. Both of these controlled measures should be considered carefully during design of a surface translation protocol as the chosen parameters have the ability to drive the observed responses. Likewise, it is suggested that surface translation protocols use an acceleration of at least 3.0 m/s^2 and a high peak velocity as the results will be analogous to other studies using this magnitude of perturbation or larger as shown by the plateau of response variables (normalized step length and minimum AP xMOS). By utilizing a larger perturbation it also ensures that participants are sufficiently challenged by the perturbation and their responses are reactionary and necessary rather than due to observer effect or belief that a step response is the expected response. Based on these results, caution should be advised while comparing literature within this field especially during protocols that utilize relatively low peak velocities and low accelerations.

Study 2 - Characterizing the effects of participant-level pre-perturbation factors on stepping outcomes

1.8 Introduction

While the previous chapter explored the influence of external perturbation parameters on balance control, the ANOVA paradigm employed (while appropriate in testing the stated hypotheses related to the effects perturbation velocity and acceleration) did not allow an exploration of other factors that might influence balance control responses. In addition, the previous chapter employed the approach of comparing mean responses of multiple repeated trials observed across conditions. While this approach can help mitigate the influence of random noise, it does not leverage the natural variability in response outcomes towards better understanding the mechanisms underlying balance control. In particular, little is known about the influence of participant-specific state at the moment of perturbation onset on reactive balance control responses.

Several participant-specific factors have previously been identified as playing a role in standing balance, specifically through the inverted pendulum model of standing balance. Factors such as COP and COM position are primary components of the inverted pendulum model as outlined in section 1.2.1 (Figure 1-1) (Winter et al., 1990). Centre of mass acceleration and velocity are also included in the model and depicted in Figure 1-1 for their roles in quiet standing (Winter, 2009). Another factor that is not included in the image, but is explored in the literature, is the effect of ankle stiffness and its effects on the model (Winter et al., 2009). By providing the axis at which the pendulum oscillates, the ankle joint is a crucial component to the model and the stiffness of the joint must be accounted for to fully understand the interaction between factors. Ankle stiffness has also been found to affect postural sway when altered through active co-contraction (Warnica et al., 2014). Overall, this model incorporates a wide variety of variables and how they interact to produce a stable and balanced system during quiet stance.

Another participant-specific factor explored in the literature is the distribution of weight between the feet prior to voluntary step initiation. However, reactive stepping is often not accompanied by the observed shift in weight distribution seen in voluntary stepping (McIlroy and Maki, 1999). One study found this phenomenon to occur in a younger adult population but the magnitude was greatly reduced and demonstrated little associated functional benefit based on timing of stepping responses (McIlroy and Maki, 1996). However, this mechanism of weight shifting still has the possibility of influencing stepping responses and is therefore of interest for the current study.

While the aforementioned participant factors have not been examined thoroughly in the surface translation paradigm, some have been focused on during tether-release designs with the goal of improving trial consistency and unpredictability. The goal of improving trial consistency in tether release aligns with the concept of explaining trial variability during surface translations. Previous works have controlled individuals' weight distribution between the feet (Kam et al., 2016), centre of pressure (COP) location (Singer et al., 2016; Weaver, 2017; Wright et al., 2014) as well as electromyographic measures of ankle dorsi and plantar flexors (Singer et al., 2016, 2012; Weaver, 2017; Wright et al., 2014) during tether release protocols. These factors all relate to how an individual controls their centre of mass

(COM) and its current phase based on the inverted pendulum model outlined in Figure 1-1. These measures also provide an indication of how an individual could have prepared their response, as they were aware of an impending postural threat and prepared their responses accordingly. Therefore, assessing and controlling measures such as the ones outlined previously may provide valuable insights into strategies employed to maintain balance. While these measures have been used due to the predictable nature of tether release where perturbation direction and magnitude are typically known prior to release, they could be used to gather insights into the behavior of the individual prior to perturbation and may provide context to some change in the responses observed during surface translations.

The potential link between pre-perturbation measures and spatial metrics of forward stepping during surface translations forms the basis for the current study, which aims to explore the potential effect of monitoring and controlling person specific pre-perturbation conditions. This study was comprised of two research questions. First, are personal pre-perturbation trial specific factors significantly associated with spatial measures of single step balance responses during reactive forward stepping? Second, would the inclusion of these personal factors significantly improve the predictive capabilities of statistical models over the inclusion of only platform factors? It was hypothesized that: 1) pre-perturbation personal trial specific factors would be significantly associated with reactive stepping responses; and 2) provide clinically significant improvements in model predictions of stepping outcomes adjusted r^2 values.

1.9 Methods

This study utilized data from the experiment described in the previous chapter. Accordingly, specific details on the experimental protocol can be found in section 2.2 and additional instrumentation and data processing, relevant to the current study, are described below.

1.9.1 Instrumentation

1.9.1.1 Surface Electromyography

Two muscles, tibialis anterior and medial gastrocnemius (Table 3-1), were monitored via surface electromyography bilaterally on the lower extremities for a total of four muscles using a Bortec electromyography system (AMT-8, Bortec Biomedical, Calgary, AB). Electrodes were placed on the surface of the skin directly over the muscle belly of the desired muscle. The skin was shaved and cleaned using alcohol in an attempt to minimize impedance and improve adhesion. Bluesensor disposable bi-polar Ag-AgCl surface electrodes were used with an inter electrode distance of 2 cm. A ground electrode was placed on the right tibial tuberosity and the EMG data was collected at 2500 Hz via First Principles software (Northern Digital Incorporated, Waterloo, Ontario, Canada). Analog signals were conditioned through a differential amplifier with a hardware band-pass filter of 10-1000 Hz and a common mode rejection ratio of 115 dB at 60 Hz. Following placement of electrodes, maximum voluntary contractions (MVC) were performed to allow for normalization of EMG signal. Each muscle had a specific MVC with the position outlined in Table 3-1 (Konrad, 2006; Merletti et al., 2005) which allowed for normalization of the signal during processing (Lehman and McGill, 1999). Each MVC trial was 10 seconds in length and participants were instructed to ramp rather than burst their muscular effort. Participants were encouraged and motivated verbally by researchers to achieve a maximum effort contraction. The value selected to represent the MVC was the single highest peak from the trial during post processing and this value represented the participants' maximum voluntary contraction.

Table 0-1: Surface electromyography muscles including electrode placement and MVC description (Konrad, 2006; Lehman and McGill, 1999; Merletti et al., 2005)**.**

Muscle	Electrode Placement	MVC Position and Movement
Tibialis Anterior	Sensors placed at 1/3 of the distance along the line starting at the tip of the fibula and ending at the medial malleolus.	The participant performed separate trials for each leg. The participant stood upright. The ankle joint began in slight dorsiflexion and the foot in inversion without extension of the great toe. Pressure was applied against the medial side, dorsal surface of the foot in the direction of plantar flexion of the ankle joint and eversion of the foot by an RA/graduate student. The participant contracted their ankle into full dorsiflexion without extension of the toes.
Medial Gastrocnemius	Sensors placed at 1/3 of the distance along the line starting at the head of the fibula and ending at the heel.	The participant performed separate trials for each leg. Plantar flexion of the foot with emphasis on pulling the heel upward more than pushing the forefoot downward. For maximum pressure in this position, it was necessary to apply pressure against the forefoot as well as against the calcaneus.

1.9.1.2 Force Platforms

Four force platforms (Advanced Mechanical technology Inc., Watertown, MA), arranged in a square (visible in Figures 2-1 and 2-2), were utilized to record ground reaction forces and moments. This data was collected at 2500 Hz through First Principles software (Northern

Digital Incorporated, Waterloo, Ontario, Canada). These force platforms were embedded within the platform, and flush with the surrounding floor, to minimize any risk of tripping during the experimental protocol. The force plate configuration allowed for maximum coverage of the participant's two feet when reactive steps were taken.

1.10 Data Analysis

1.10.1 Surface Electromyography Data Processing

EMG data was processed and analyzed using custom MATLAB™ routines (version R2015a, Mathworks Inc., USA). EMG was down-sampled from the collection frequency of 2500 Hz to 2048 Hz to allow for time synchronization with force platform and kinematic data. Signal bias was removed from the EMG of each muscle by subtracting the mean of each trial from itself. Following the removal of signal bias, full wave rectification was performed. Next, a second order, low pass, single pass Butterworth filter (Winter, 2009) was applied using a cut-off frequency of 6 Hz (Chen et al., 2014; Weaver, 2017). Following the process of linear enveloping, EMG data was converted from a signal in Volts to a percentage of maximum voluntary contraction by dividing the signal by the peak of the associated muscle's MVC trial.

Electromyographic data was then used to calculate the co-contraction index (CCI) for each of the participants' ankles. Bilaterally, tibialis anterior and medial gastrocnemius muscles were used for the calculation. Co-contraction index, calculated similar to previous research (Hubley-Kozey et al., 2009; Lewek et al., 2004), was used as a metric of active ankle stiffness at the instant of perturbation onset (onset defined in section 2.3.1.1). Ankle CCI was calculated as per equation 2 for each ankle. Electromyographic data was analyzed for 100 ms prior to perturbation onset which was then normalized to 100 data points. These 100 data points were passed through equation 2 and resulted in a single representative value of co-contraction between tibialis anterior and medial gastrocnemius muscles bilaterally.

Equation 2: $$CCI = \frac{1}{100}\sum_{i=1}^{100} \left[\frac{lower\ EMG_i}{higher\ EMG_i} \times (lower\ EMG_i + higher\ EMG_i)\right]$$

1.10.2 Force Platform Data Processing

Data collected from the force platforms was used to determine both weight distribution between the participant's feet, and centre of pressure (COP) location. First, data was down sampled to 2048 Hz to synchronize with collected surface EMG and kinematic data. Data was then dual pass filtered using a second order low-pass Butterworth filter with a cutoff frequency of 50 Hz. Vertical force distribution between the participant's feet was calculated as a proportion of the total vertical force to determine weight distribution at the instant of platform movement onset. Centre of pressure was calculated based on both the ground reaction forces and moments from each of the force plates located underneath the participant's feet. These individual COP's were combined to produce an overall COP location using a weighted average based on the proportion of total body weight on each force plate. The location of the AP COP relative to the ankle joint centre (with anterior values as positive) was extracted at the instant of platform movement onset (section 2.3.1.1).

1.10.3 Kinematic Data Processing

Kinematic data was processed as described in section 2.3.1. Centre of mass position was differentiated to calculate COM velocity and COM velocity was differentiated to calculate

COM acceleration. Centre of mass position (relative to the ankle joint centre), velocity, and acceleration were extracted at the instant of platform movement onset, as calculated in section 2.3.1.1.

Kinematic data was also used to calculate the dependent variables of normalized step length section 2.3.1.2) and minimum AP xMOS (section 2.3.1.4).

1.10.4 Statistical Analysis

Following exclusion of no-step and multistep trials, 749 trials were analyzed. Data was analyzed using multiple backward elimination stepwise linear regressions (R Core Team, 2017). To reduce the number of total variables used, only the right ankle CCI was used in the models. Exclusion of the left ankle CCI was based on a strong correlation ($r^2 = 0.73$, $p = 0.011$) between the two variables and the advice of a statistician to reduce the number of overall variables. 'External' variables considered as model inputs were measured platform (1) acceleration and (2) velocity. 'Personal' variables at perturbation onset included: (3) proportion weight distribution between feet, (4) ankle CCI, (5) AP COP location relative to the ankle joint centre, (6) AP COM location relative to the ankle joint centre, (7) AP COM velocity and (8) AP COM acceleration. Dependent variables for the linear regressions were trial-specific normalized step length and minimum xMOS.

An initial model to identify the strength of the repeated measures design was performed followed by the addition of external factors and lastly the addition of personal factors (three unique models). Backward elimination stepwise regressions were performed on both the second and the third models. An apriori inclusion criteria of $p < 0.05$ was used as a threshold to determine whether to keep or exclude any given variable.

The process of analyzing external and personal conditions was also applied to subsets of the data. Specifically, the data was divided into low and high velocity trials based on the relevance of the platform velocity condition on spatial stepping measures as outlined in the previous chapter. As a product of this subset analysis, the platform velocity factor was accounted for, leaving platform acceleration as the sole external factor.

In total, 18 statistical models were run. Three data sets were assessed (all data, low velocity data, high velocity data) across two dependent measures (normalized step length, minimum AP xMOS) using the sequence of three models comprised of different possible factors (participant repeated measure design, addition of external factor(s), addition of personal factors). Model outputs were assessed based on which factors were maintained via the stepwise linear regression approach. Inclusion of factors identifies them as providing statistically significant value to the model. Resulting models were also compared based on adjusted r^2 values. An apriori r^2 improvement of 0.10 was used to identify clinically significant improvements in the models' predictive capabilities.

1.11 Results

1.11.1 Step Length

Entire Data Set: For normalized step length, the baseline model consisting of just the repeated measures factor resulted in a model with an r^2 of 0.433 (Figure 3-1A). Addition of external factors to the repeated measures design and performing a backward stepwise linear regression produced an r^2 of 0.671 while retaining both platform acceleration ($F_{1, 725}$ = 37.58, B = 1.02) and platform velocity ($F_{1, 724.37}$ = 352.30, B = 25.74) in the model (Figure 3-1B). The final model, including repeated measures, external, and personal pre-perturbation factors produced the best model (adjusted r^2 = 0.700) consisting of platform acceleration ($F_{1, 722.98}$ = 38.72,B = 1.00), platform velocity ($F_{1, 722.48}$ = 387.90, B = 26.02), weight distribution between the feet ($F_{1, 741.04}$ = 5.51, B = 0.31), and AP COP location ($F_{1, 742.61}$ = 54.16, B = 0.17) (Figure 3-1C). All final model results, including the regression coefficients, can be found in Table 3-3.

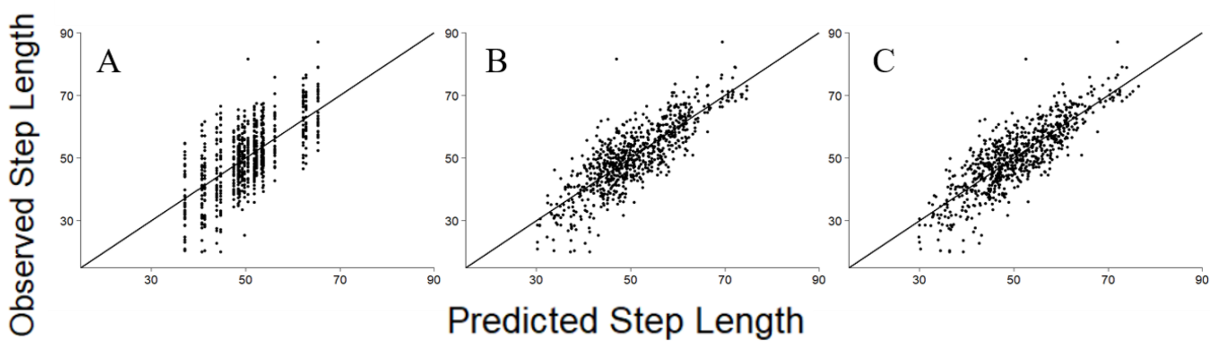

Figure 0-1: Comparison of observed normalized step length (% leg length) to model predicted normalized step length across the entire data set. A: model using only repeated measures design – each 'column' of data represents a distinct participant, B: significant model using repeated measures design and the inclusion of 'external' factors, C: significant model using repeated measures design, 'external' factors, and 'personal' factors. Angled line represents perfect agreement between predicted values and experimental data.

Low Velocity Data: Following stratification of the data by velocity, the low velocity data baseline model of repeated measures linear regression produced an r^2 of 0.602 (Figure 3-2A).

The addition of external factors (only platform acceleration) produced a model with an r^2 of 0.612 while keeping the external factor ($F_{1,\,354.11}$ = 9.63, B = 0.64) as a significant factor for the model (Figure 3-2B). Including pre-perturbation personal factors increased the adjusted r^2 to 0.646, with this final model including the repeated measures, platform acceleration ($F_{1,\,352.87}$ = 12.73, B = 0.71), and the AP COP location ($F_{1,\,374}$ = 26.88, B = 0.16) (Figure 3-2C). All final model results can be found in Table 3-3 including regression coefficients obtained from the final models.

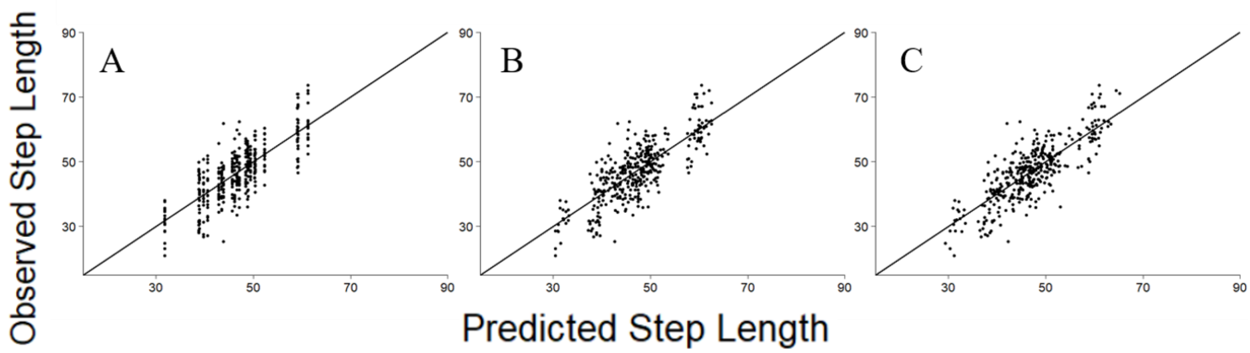

Figure 0-2: Comparison of observed normalized step length (% leg length) to model predicted normalized step length across the low velocity trials. A: model using only repeated measures design – each 'column' of data represents a distinct participant, B: significant model using repeated measures design and the inclusion of 'external' factors, C: significant model using repeated measures design, 'external' factors, and 'personal' factors. Angled line represents perfect agreement between predicted values and experimental data.

High Velocity Data: For the high velocity trials, the baseline regression model comprised of the repeated measures factor produced an r^2 of 0.422 (Figure 3-3A). The addition of external platform acceleration ($F_{1,\,349.21}$ = 244.07, B = 3.66) produced a model with an r^2 of 0.661 (Figure 3-3B). Including pre-perturbation personal factors increased the adjusted r^2 to 0.689, with this final model including the repeated measures, platform acceleration ($F_{1,\,348.32}$ =

239.72, B = 3.54), weight distribution between the feet ($F_{1, 365.09}$ = 4.11, B = 0.39), and AP COP location ($F_{1, 366.47}$ = 21.64, B = 0.16) (Figure 3-3C). All final model results can be found in Table 3-3 including regression coefficients obtained from the final models.

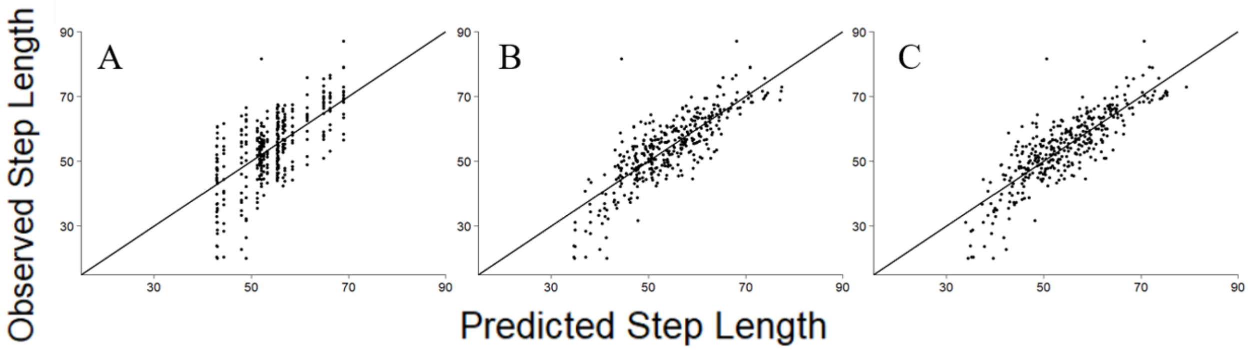

Figure 0-3: Comparison of observed normalized step length (% leg length) to model predicted normalized step length across the high velocity trials. A: model using only repeated measures design – each 'column' of data represents a distinct participant, B: significant model using repeated measures design and the inclusion of 'external' factors, C: significant model using repeated measures design, 'external' factors, and 'personal' factors. Angled line represents perfect agreement between predicted values and experimental data.

Table 0-2: Step length backward stepwise linear regression results. Regression intercept is mean of participant intercepts. Variable values are B coefficients (F value). Variables eliminated based on p > 0.05.

	Full Data Set			Low Velocity Data Set			High Velocity Data Set		
	Participant	Participant + External	Participant + External + Personal	Participant	Participant + External	Participant + External + Personal	Participant	Participant + External	Participant + External + Personal
r^2	0.433	0.671	0.700	0.602	0.612	0.646	0.422	0.661	0.689
Intercept	50.53	28.62	0.58	46.92	44.42	32.91	54.24	40.17	9.44
Platform Acceleration (m/s^2)	N/A	1.02 (37.58)	1.00 (38.72)	N/A	0.64 (9.63)	0.71 (12.73)	N/A	3.66 (244.07)	3.54 (239.72)
Platform Velocity (m/s)	N/A	25.74 (352.30)	26.02 (387.90)	N/A	N/A	N/A	N/A	N/A	N/A
Weight Distribution	N/A	N/A	0.31 (5.51)	N/A	N/A	Eliminated	N/A	N/A	0.39 (4.11)
Ankle CCI	N/A	N/A	Eliminated	N/A	N/A	Eliminated	N/A	N/A	Eliminated
AP COP Location (mm)	N/A	N/A	0.17 (54.16)	N/A	N/A	0.16 (26.88)	N/A	N/A	0.16 (21.64)
AP COM Location (mm)	N/A	N/A	Eliminated	N/A	N/A	Eliminated	N/A	N/A	Eliminated
AP COM Velocity (mm/s)	N/A	N/A	Eliminated	N/A	N/A	Eliminated	N/A	N/A	Eliminated
AP COM Acceleration (mm/s^2)	N/A	N/A	Eliminated	N/A	N/A	Eliminated	N/A	N/A	Eliminated

1.11.2 Minimum xMOS

Entire Data Set: For minimum AP xMOS, the baseline model consisting of just the repeated measures factor resulted in a model with an r^2 of 0.154 (Figure 3-4A). Addition of external factors to the repeated measures design and performing a backward stepwise linear regression produced an r^2 of 0.424 while retaining platform velocity ($F_{1, 726.14}$ = 336.29, B = 205.87) in the model (Figure 3-4B). The final model, including repeated measures, external, and personal pre-perturbation factors produced the best model (adjusted r^2 = 0.450) consisting of platform velocity ($F_{1, 723.65}$ = 347.88, B = 205.93), ankle CCI ($F_{1, 354.31}$ = 4.62, B = 11.25), AP COM location ($F_{1, 545.89}$ = 14.85, B = -0.75), and AP COM velocity ($F_{1, 732.96}$ = 7.21, B = -1.08) (Figure 3-4C). All final model results, including the regression coefficients, can be found in Table 3-4.

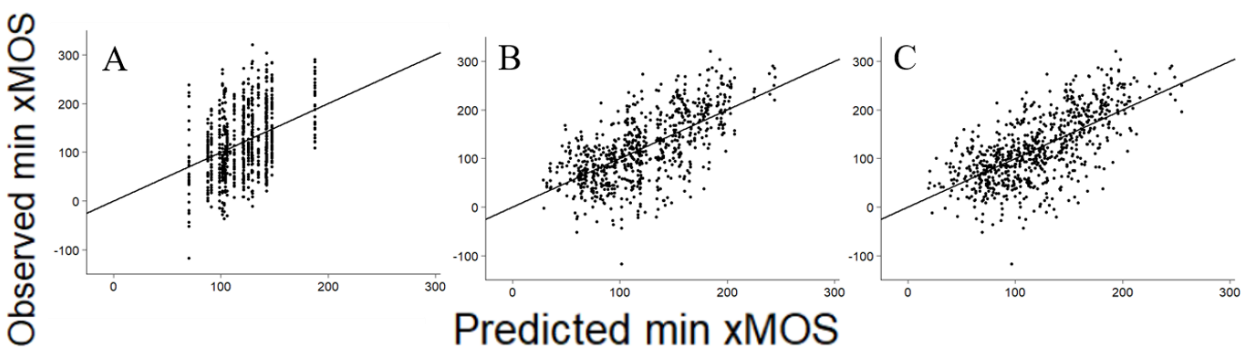

Figure 0-4: Comparison of observed minimum AP xMOS (mm) to model predicted minimum AP xMOS across the entire data set. A: model using only repeated measures design – each 'column' of data represents a distinct participant, B: significant model using repeated measures design and the inclusion of 'external' factors, C: significant model using repeated measures design, 'external' factors, and 'personal' factors. Angled line represents perfect agreement between predicted values and experimental data.

Low Velocity Data: Following stratification of the data by velocity, the low velocity data baseline model of repeated measures linear regression produced an r^2 of 0.361 (Figure 3-5A).

The addition of external factors (only platform acceleration) produced a model with an r^2 of 0.376 while keeping the external factor ($F_{1,\,354.71}$ = 7.44, B = -3.59) as a significant factor for the model (Figure 3-5B). Including pre-perturbation personal factors increased the adjusted r^2 to 0.419, with this final model including the repeated measures, platform acceleration ($F_{1,353.43}$ = 8.87, B = -3.83), and the AP COM location ($F_{1,\,343.06}$ = 18.06, B = -0.79) (Figure 3-5C). All final model results can be found in Table 3-4 including regression coefficients obtained from the final models.

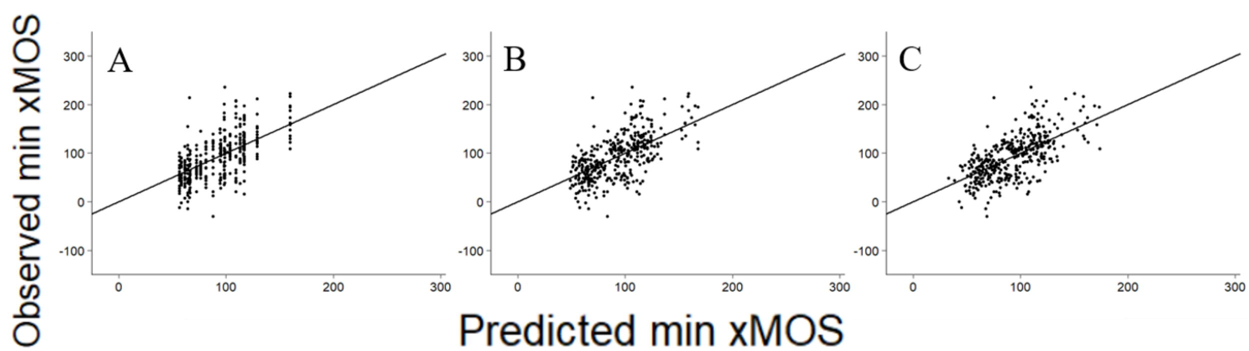

Figure 0-5: Comparison of observed minimum AP xMOS (mm) to model predicted minimum AP xMOS across the low velocity trials. A: model using only repeated measures design – each 'column' of data represents a distinct participant, B: significant model using repeated measures design and the inclusion of 'external' factors, C: significant model using repeated measures design, 'external' factors, and 'personal' factors. Angled line represents perfect agreement between predicted values and experimental data.

High Velocity Data: For the high velocity trials, the baseline regression model comprised of the repeated measures factor produced an r^2 of 0.235 (Figure 3-6A). The addition of external platform acceleration ($F_{1,\,349.71}$ = 149.57, B = 25.40) produced a model with an r^2 of 0.460 (Figure 3-6B). Including pre-perturbation personal factors increased the adjusted r^2 to 0.504, with this final model including the repeated measures, platform acceleration ($F_{1,\ 350.12}$ = 158.61, B = 25.66), ankle CCI ($F_{1,\ 260.55}$ = 4.41, B = 15.07), and AP COM location ($F_{1,\ 316.45}$ = 11.13, B = -0.79) (Figure 3-6C). All final model results can be found in Table 3-4 including regression coefficients obtained from the final models.

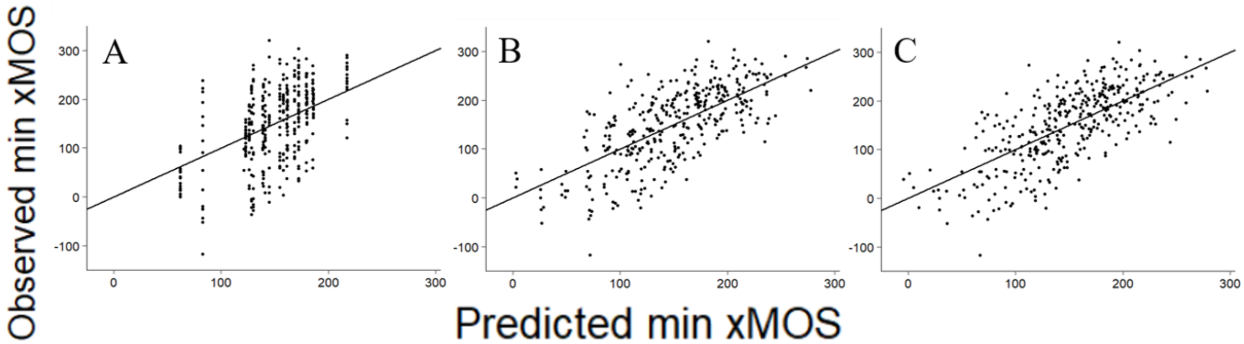

Figure 0-6: Comparison of observed minimum AP xMOS (mm) to model predicted minimum AP xMOS across the high velocity trials. A: model using only repeated measures design – each 'column' of data represents a distinct participant, B: significant model using repeated measures design and the inclusion of 'external' factors, C: significant model using repeated measures design, 'external' factors, and 'personal' factors. Angled line represents perfect agreement between predicted values and experimental data.

Table 0-3: Minimum xMOS backward stepwise linear regression results. Regression intercept is mean of participant intercepts. Variable values are B coefficients (F value). Variables eliminated based on p > 0.05.

	Full Data Set			Low Velocity Data Set			High Velocity Data Set		
	Participant	Participant + External	Participant + External + Personal	Participant	Participant + External	Participant + External + Personal	Participant	Participant + External	Participant + External + Personal
r^2	0.154	0.424	0.450	0.361	0.375	0.419	0.235	0.466	0.507
Intercept	120.77	-22.99	15.10	93.20	107.26	165.16	149.32	49.77	97.18
Platform Acceleration (m/s^2)	N/A	Eliminated	Eliminated	N/A	-3.59 (7.44)	-3.83 (8.87)	N/A	25.83 (149.57)	26.16 (158.61)
Platform Velocity (m/s)	N/A	205.87 (336.29)	205.93 (347.88)	N/A	N/A	N/A	N/A	N/A	N/A
Weight Distribution	N/A	N/A	Eliminated	N/A	N/A	Eliminated	N/A	N/A	Eliminated
Ankle CCI	N/A	N/A	11.25 (4.62)	N/A	N/A	Eliminated	N/A	N/A	15.64 (4.41)
AP COP Location (mm)	N/A	N/A	Eliminated	N/A	N/A	Eliminated	N/A	N/A	Eliminated
AP COM Location (mm)	N/A	N/A	-0.75 (14.85)	N/A	N/A	-0.79 (18.06)	N/A	N/A	-1.02 (11.13)
AP COM Velocity (mm/s)	N/A	N/A	-1.08 (7.21)	N/A	N/A	Eliminated	N/A	N/A	Eliminated
AP COM Acceleration (mm/s^2)	N/A	N/A	Eliminated	N/A	N/A	Eliminated	N/A	N/A	Eliminated

1.12 Discussion and Conclusions

The goal of this study was twofold: identify potentially relevant personal factors to spatial stepping responses and quantify their effects on predictive statistical models. In support of the first hypothesis, in all final regression models at least one personal factor at the moment of perturbation onset proved to be a significant predictor or step length and minimum xMOS (this held true across the full data set, and the subsets stratified by low and high velocity trials). However, the second hypothesis was not supported as the largest increase in adjusted r^2 observed due to the inclusion of personal factors was only 0.044, less than the criteria of 0.10 outlined previously. These results demonstrate that the improvements in model prediction by including personal pre-perturbation factors are modest in comparison to including a repeated measures factor representing each ‘participant’ and external perturbation characteristics (acceleration and velocity).

Addressing the first research question, two (out of six) personal factors appeared with the same consistency amongst all of the models (Figure 3-7). Specifically, AP COP location and AP COM location were retained as significant predictors in three models of the possible six (50%) models run which included personal factors. Other factors were limited to two instances (weight distribution and ankle CCI), one instance (AP COM velocity), or no instances (AP COM acceleration). If we focus on data stratified by low and high velocity trials (an approach supported by the Study 1 findings), the only personal factors retained were weight distribution, ankle CCI, AP COP location, and AP COM location. Within these subset analyses, AP COP location and AP COM location were again found to be the most common significant predictors (2 out of possible 4 final models). The statistical significance of these factors aligns with the inverted pendulum model (Figure 1-1) as the COP manipulates the COM to maintain it within the BOS and therefore the location of these factors provides context to the phase of postural control the participant is in during perturbation onset. During perturbation, the phase of postural control determines whether the current conditions of the individual provide an initial stabilizing (opposite direction of perturbation) or destabilizing (same direction as perturbation) force. While these factors were identified as providing statistically significant value to the model, their clinical significance could not be assessed using this method and thus comparisons of adjusted r^2 values were performed.

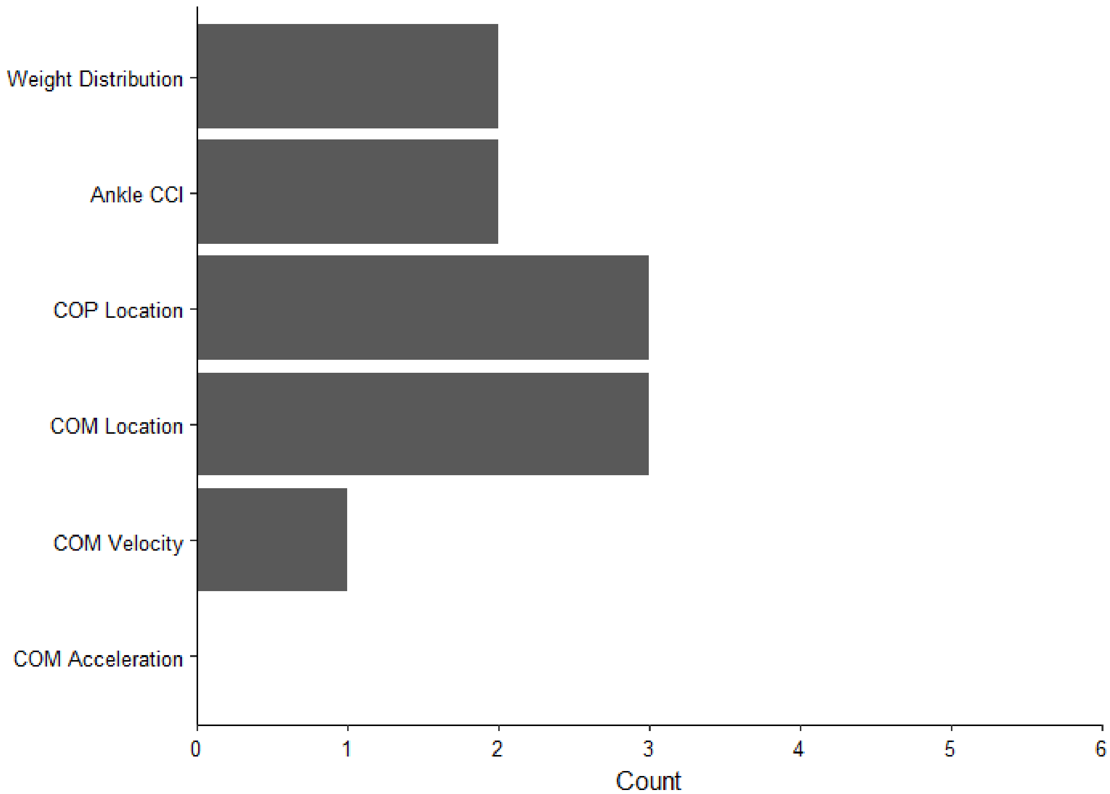

Figure 0-7: Count of each personal predictive variables number of occurrences through the backward stepwise regression process using an inclusion criteria of $p < 0.05$. Maximum count of six was possible if the variable was kept for each model.

While at least one trial-specific pre-perturbation factors was significantly associated stepping characteristics in all analyses, the relatively contribution of these factors was small (Figure 3-8). Examination of the identified best model using personal factors always improved the adjusted r^2 compared to external factor only models, this improvement was found to range from 0.02-0.04 across stepping and xMOS outcomes. Based on my clinically relevant threshold of an increase of 0.10, the inclusion of the personal factors did not provide a significant increase in any of the models predicative capabilities. However, while the observed improvements are relatively small, these are based on an ecological design where values were not systematically varied and therefore large variability within values were not observed (see Table 3-4). Testing a larger variety of values could potentially have resulted in a greater explained variance but this study aimed at assessing a real world range of values so input variables were not manipulated. The improvement of the adjusted r^2 values can also be visualized in Figure 3-8A for normalized step length and Figure 3-8B for minimum xMOS. These figures demonstrate the relative contributions of external factors and personal factors to the final models adjusted r^2 value. As the models increase in complexity, the adjusted r^2 continued to increase but at a reduced rate, which could be partially caused by shared variance between previously included variables.

Table 0-4: Descriptive statistics of personal input variables at the onset of platform movement. Data are presented as the mean of within-participant means across all conditions, and (in parentheses) the mean of within-participant standard deviations across all conditions.

Weight Distribution (% Body Weight)	Ankle CCI	AP COP Location (mm)	AP COM Location (mm)	AP COM Velocity (mm/s)	AP COM Acceleration (mm/s^2)
51.47	1.51	72.09	72.18	1.01	-30.01
(1.55)	(0.27)	(9.14)	(8.91)	(4.99)	(35.21)

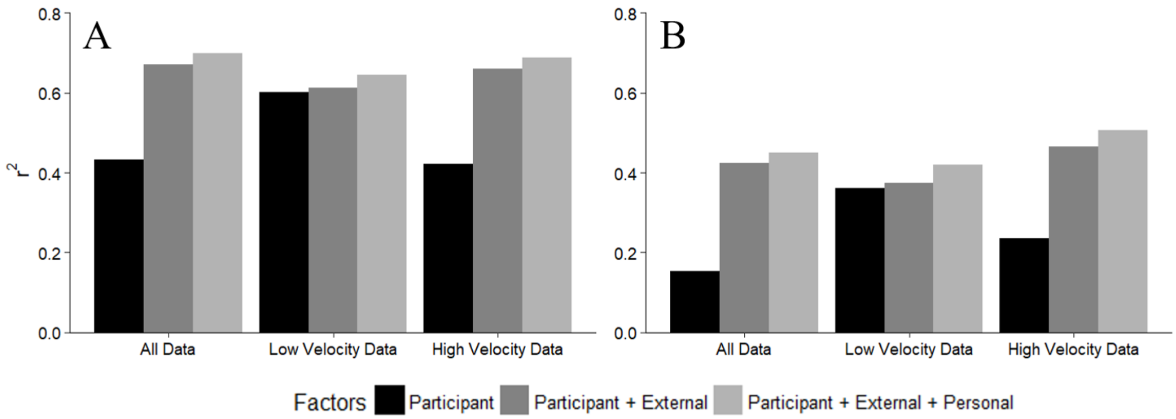

Figure 0-8: Comparison of the predicative capabilities, based on adjusted r^2, of the resulting best models based on input variables for A: normalized step length, and B: minimum AP xMOS.

Two main mechanisms used by the central nervous system to resolve destabilizing perturbations are anticipatory (APA) and compensatory postural adjustments (CPA) (Santos et al., 2010a). While the CPA is always present and is initiated by sensory feedback from the perturbation (Alexandrov et al., 2005; Park et al., 2004), an APA requires prediction of the impending perturbation (Bouisset and Zattara, 1987; Massion, 1992). Both of these mechanisms employ manipulation of muscle activation to successfully respond to the postural threat and maintain balance(Santos et al., 2010a, 2010b). While APA's have been found to provide significant contributions to the response following a predicted external perturbation (Santos et al., 2010b), by minimizing the predictability of the perturbation the role of the APA is also minimized. Due to the unpredictable nature of the perturbations used for this study, APA's should not play a significant role in affecting stepping outcomes, whereas CPA's are present regardless of the predictability of the perturbation. The CPA encompasses the stepping responses examined in this study and the pre-perturbation factors aimed at providing a context of the phase of postural sway. As APA's are observed less frequently during unpredictable perturbations, the pre-perturbation factors were hypothesized to relate to the stepping outcomes as they were independent of the APA.

Based on the analysis performed, future research using surface translations should consider monitoring and controlling AP COP location and AP COM location immediately prior to perturbation onset across a larger range of values. Previous works have also identified COM position as being related to stepping responses (Pavol et al., 2004) which supports these findings. Of the identified variables, AP COP and COM position maintain their relevance once the data had been stratified by platform velocity and are therefore more highly recommended to be considered during study design. Previous works have suggested a role of COP and COM movements during quiet stance to provide sensory feedback to the central nervous system (Carpenter et al., 2010; Murnaghan et al., 2013, 2011), and in connecting this feedback to improvements in balance control (Rajachandrakumar et al., 2018). These theories align with the findings of this study, which connect both the COP and COM locations to spatial measures of stepping responses. Between AP COP and COM location in the current study, both values had very similar means and SD as shown in Table 3-5 and the increased computational demand to monitor COM in real time may present challenges in some research settings. Based on all of these aspects, it would be recommended that future surface translation studies consider the possibility of controlling AP COP prior to perturbation to explore its potential role further across a larger range of values.

While this study provides insights into potential mechanisms for inter trial variability it is also accompanied by several limitations. First, while there are benefits to not controlling the variables used during the stepwise linear regressions, this also results in a limitation due to the potentially reduced range of values present in the variables. By reducing the range tested it can limit the ability of the variable to have any significant impact on the model. This could be mitigated by future research systematically varying some of the identified variables in an effort to determine a possible dose-response relationship. Second, examination of the effects of timing events (temporal aspects of platform acceleration and deceleration) were not included within the regression models. While beyond the scope of this thesis, there is potential value in exploring these factors in future work. Third, the generalizability of the resulting regression equations is limited due to the repeated measures linear regression approach employed. Specifically, while this approach accounts for any important source of variance due to the dependency of multiple trials completed by each participant, it uses an individualized intercept for each participant and therefore the equations generated cannot be applied to individuals who were not included in the statistical analysis. Finally, the population tested were young, healthy adults who likely demonstrated a smaller range of personal factors compared to populations at increased risk of falling (such as older adults, stroke recovery patients, or individuals with movement disorders). Accordingly, the relative importance of the personal factors may vary based on population and their individual capabilities.

In conclusion, this study provides substantial insights into some of the inter trial variability during support surface translations by identifying AP COP and COM positions as potential sources of variability between trials in a young healthy adult population. While in-vivo testing will always have inter trial variability due to the complexity of human balance control, being able to identify sources of the variability and reduce it allows researchers to more accurately identify potential mechanisms and associated mechanistic deficiencies within populations of interest, such as older adults or pathological populations. Future research should focus on continuing to improve the understanding of variability between trials and individuals to target and expand the knowledge surrounding specific mechanisms of balance control.

Summary of Contributions

The two studies presented as part of this thesis each contain their own novel contributions to the body of literature surrounding balance control, specifically as it relates to support surface translations.

The first study examined the effects of how different perturbation parameters effect spatial measures of single step balance control. By exploring this relationship in greater detail than previous literature, study one provides guidelines for future surface translation study design as well as insights into the comparability of previous research. The findings of this study regarding perturbation parameter recommendations generally aligns with some of the research groups who have implemented perturbations based on the works of Maki et al. (1996). However, by establishing the relationship between increases in platform acceleration and velocity, this study goes beyond what was previously done and allows for connections between literature regardless of parameters utilized.

Study two probed the underlying mechanisms of ecologically valid pre-perturbation factors and how they contribute to the gross balance response. Through identification of relevant factors, this study established a connection between key factors of standing balance control, COP and COM position, and their role in reactive balance control. While the strength of the relationship is relatively weak, as demonstrated through adjusted r^2 comparisons, this study provides a foundation for future research to be better equipped to answers questions pertaining to the individual mechanisms of balance control.

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Appendix A **Time-varying responses to perturbation onset**

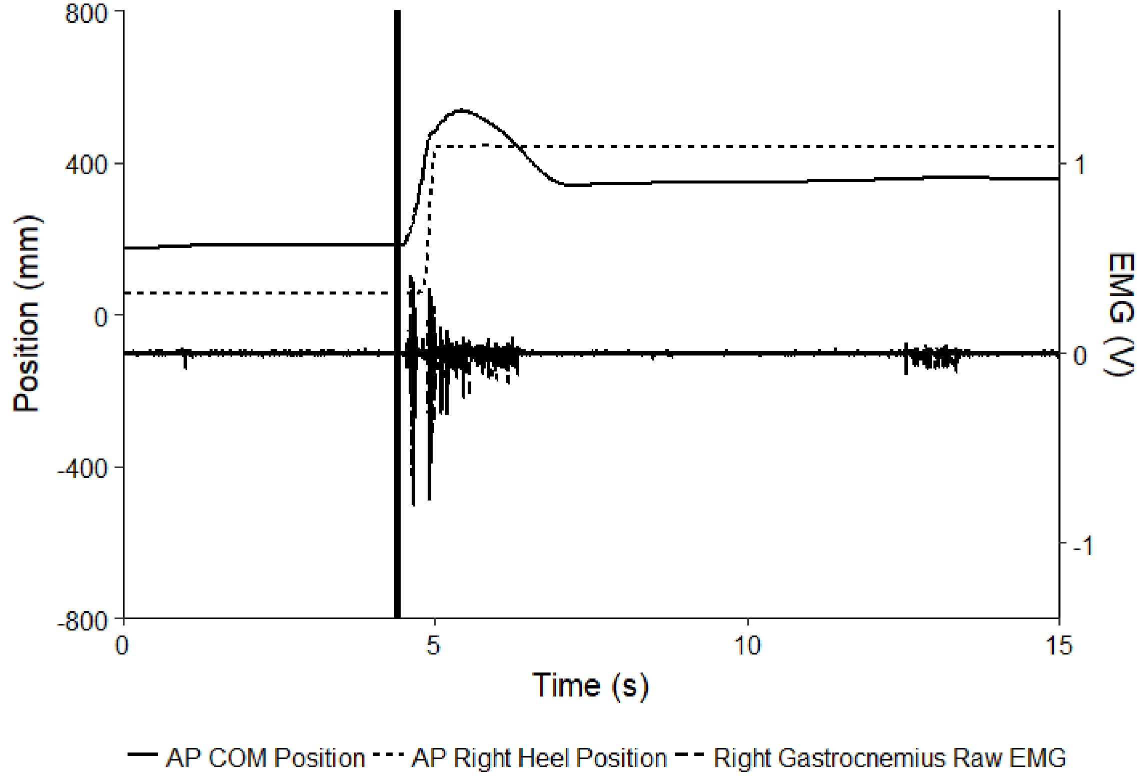

Figure Error! No text of specified style in document.**-1: Time varying responses of AP COM position (solid line), AP right heel position (dashed line), and right gastrocnemius raw EMG signal during a single step response with the right leg. Vertical line denotes onset of platform movement. Perturbation applied was 3.5 m/s^2 acceleration, 1.00 m/s velocity, 0.30 m displacement.**

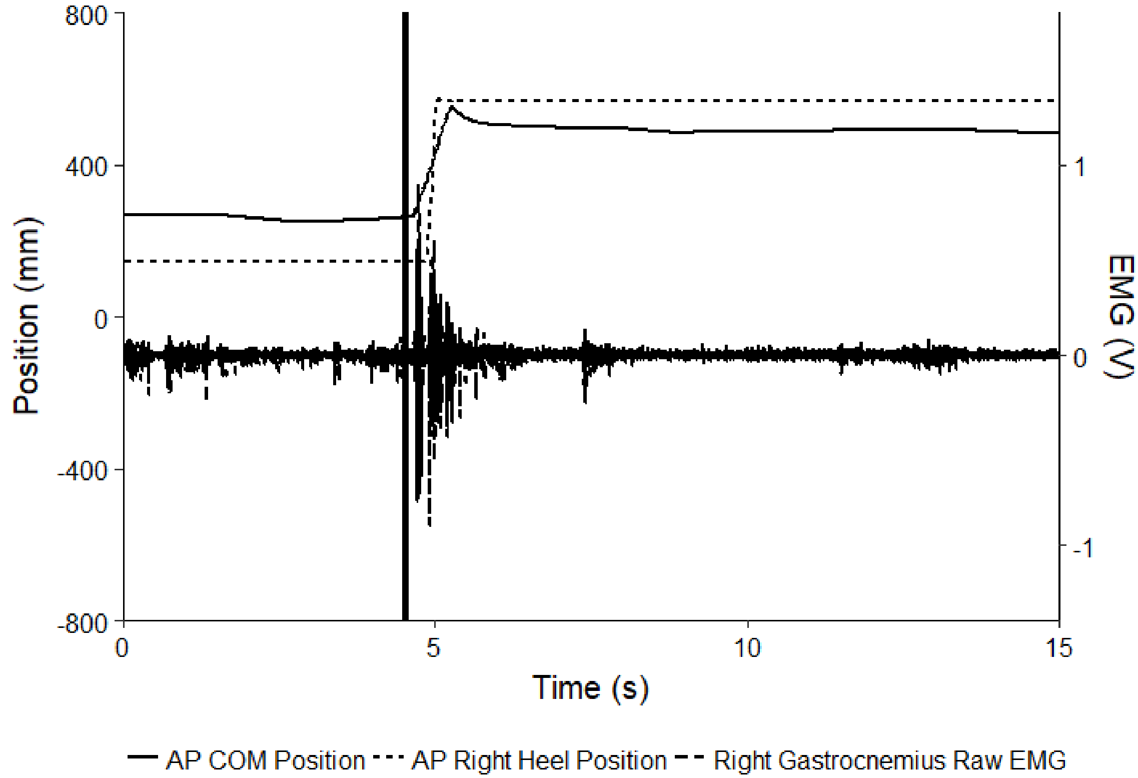

Figure Error! No text of specified style in document.**-2: Time varying responses of AP COM position (solid line), AP right heel position (dashed line), and right gastrocnemius raw EMG signal during a single step response with the right leg. Vertical line denotes onset of platform movement. Perturbation applied was 3.5 m/s^2 acceleration, 0.50 m/s velocity, 0.30 m displacement.**

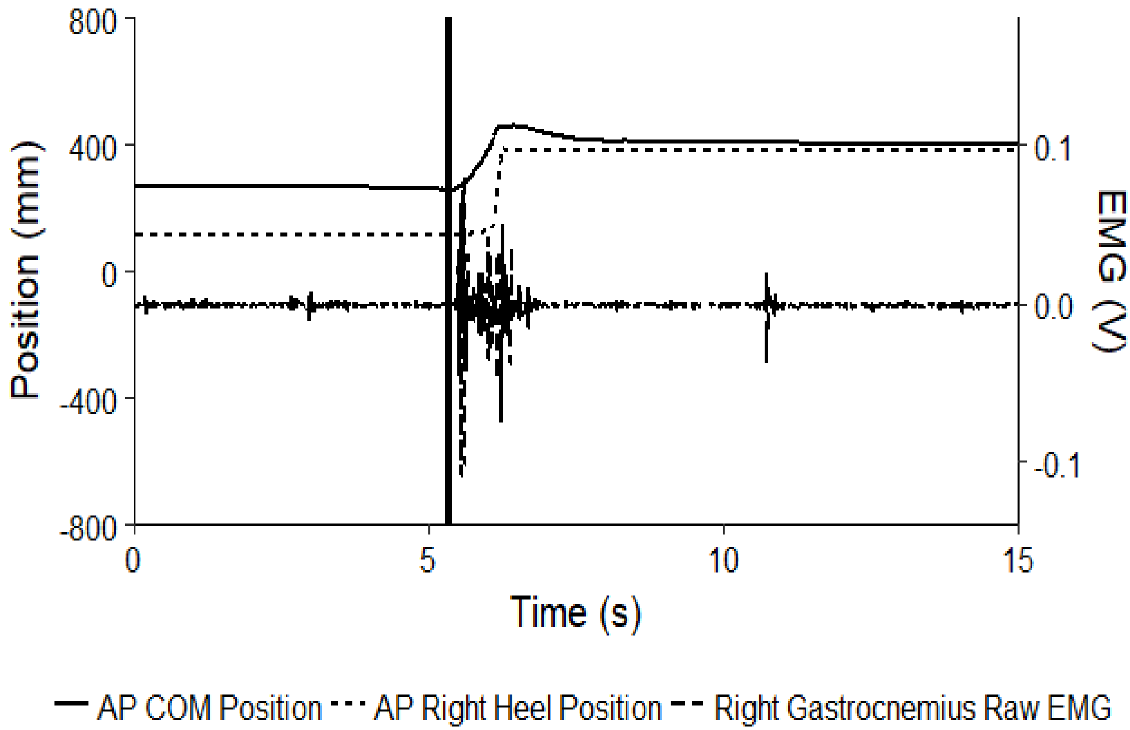

Figure Error! No text of specified style in document.**-3: Time varying responses of AP COM position (solid line), AP right heel position (dashed line), and right gastrocnemius raw EMG signal during a single step response with the right leg. Vertical line denotes onset of platform movement. Perturbation applied was 1.0 m/s^2 acceleration, 0.65 m/s peak velocity, 0.30 m displacement.**