A Model and Experiment of Human Control of Lateral Balance During Walking

by

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Abstract

An essential component of human walking is the control of balance. Inadequate balance control can lead to falls and injuries, which are major concerns for older adults and those with walking impediments. Previous examinations of balance control during walking have largely been focused on stepping strategies, such as step placement control and stance ankle torque, that directly control the location of the center of pressure relative to the body center of mass. However, a relatively unappreciated approach is to indirectly affect the center of mass through inertial balancing strategies, where parts of the body like the trunk and arms are moved to induce a stabilizing reaction force. Inertial balance can significantly contribute to balance recovery during standing and on constrained surfaces such a tight-rope. However, the contribution of these inertial controls has scarcely been examined during walking, and particularly for lateral stability, which is thought to require more active control than forward stability. I intend to bridge this gap by demonstrating how inertial strategies can contribute to lateral balance control in response to perturbations during walking.

This work consists of a model of human walking and an experiment. A simple dynamical model is used to predict how both stepping and inertial strategies may be controlled to achieve lateral balance. The model walks in 3D largely through passive dynamics but also has active control for key degrees of freedom such as push-off, step placement, and trunk roll. I show how stepping and inertial balance controls may be designed using a once-per-step hybrid control scheme, and demonstrate how the two can be coupled operationally to counteract against lateral balance disturbances.

The experiment consists of lateral perturbations applied to human subjects as they walk. I designed a force feedback device to apply perturbations to a freely moving subject on an instrumented treadmill, with ability to modulate the timing and amount of force applied. The experiments were designed to induce a variety of balance strategies, both while walking normally and with constraints on available step placement, similar to a narrow walkway. This demonstrates

how humans modulate and select from a continuum of control between stepping and inertial strategies.

There were several model predictions that were validated by the experiment. One is that perturbations during different parts of the stride call for different amounts of foot placement, due in part to the time available before the next footfall. The model also predicts that multiple balance strategies should contribute simultaneously, and particularly so with narrow walking constraints. Experiments show that humans do modulate their foot placement according to perturbation direction and timing, with greater placement for earlier perturbations in a step. However, stance ankle torque and inertial balance strategies also contribute to a smaller degree throughout the stride, affecting the rate and timing of falling, and thus the amount of foot placement needed. These effects were amplified in the presence of narrow stepping constraints.

This dissertation shows how balance during walking is controllable by multiple strategies and quantifies how humans employ them in combination. Inertial balance strategies are employed in many cases, particularly when there is considerable time before the next footfall. This work also demonstrates how computational models can explain active balance control, and how humans can flexibly combine multiple strategies to accommodate various challenges and constraints.

Chapter 1 Introduction

It is essential to maintain balance during human walking. Inadequate balance control can lead to falls and injuries, which are major concerns for older adults and those with walking impediments. Previous examinations of balance control during walking have largely been focused on stepping strategies, which directly change the location of the base of support. This may be accomplished with mechanisms such as step placement control, ankle torque, push-off, and steering, and predominantly in the sagittal plane. Meanwhile, studies of human standing have shown that inertial balance strategies can significantly contribute to balance recovery. Inertial balancing refers to strategies where parts of the body like the trunk and arms are moved to indirectly induce a stabilizing reaction force. Such strategies have scarcely been examined during walking. They may be particularly important for lateral walking stability, which is thought to require more active control than forward stability. I intend to bridge this gap by demonstrating how inertial balance strategies can contribute to lateral balance control in response to perturbations during walking.

1.1 The challenges to walking balance

Falling is a major risk factor for injury especially among older adults (Masud and Morris, 2001), with more falls occurring during dynamic activities (Lord et al., 2004). As walking is the major source of activity, controlling and maintaining balance during walking is critically important for healthy living. Theoretical models suggest that lateral balance control during walking is especially challenging, requiring more active compensation versus fore-aft balance (Kuo, 1999; MacKinnon and Winter, 1993). It has also been shown clinically that lateral balance deficiencies are associated with fall risk in older adults (Hilliard et al., 2008). Thus, understanding how humans control lateral balance during walking could be an important component in better identifying risk factors for falling and strategies to mitigate those risks.

One way to improve balance control may be to constantly walk with a large base of support by widening the distance between the feet, sometimes called the step width. There is evidence that this can be useful to accommodate for some degree of lateral movement variability (Dean et al., 2007), but there are energetics and practicality limitations. Walking with larger-than-preferred step width increases metabolic costs of walking (Donelan et al., 2001). Meanwhile, there is no way to determine a wide-enough step width, as there could always be a larger balance perturbation that exceeds the stabilization potential of any given width. The width of the walking surface can also constrain the step width available. This seems to be corroborated by everyday observations, as humans do not normally walk with their legs spread especially wide. Thus, we would expect significant balance disruptions to require transient deviations from the nominal walking trajectory in order to regain balance.

1.2 Multiple stepping-related strategies can affect balance

Previous literature has identified stepping strategies, which I use to broadly categorize controls of behaviors related to the movement of the feet and legs within the usual stepping cadence of walking, as the key component to lateral balance control during walking. These behaviors include: (1) step placement, where the position of the landing foot is controlled (Hof et al., 2010; Townsend, 1985); (2) ankle torque, where the muscles of the ankle are used during stance (Best and Wu, 2020; King and Zatsiorsky, 2002); (3) push-off, where the force produced as the trailing foot leaves the ground is regulated (Kim and Collins, 2017); and (4) steering, where the yaw angle of the foot is turned (Rebula et al., 2017). These behaviors are direct alterations to where and how much force or pressure is applied to the ground.

The interpretation of stepping strategies is most often viewed through the lens of the inverted pendulum model. This is a common abstraction of standing and walking, because the center-of-mass (COM) of the human body is almost always near the pelvis (MacKinnon and Winter, 1993), and the overall inertia may be interpreted to be concentrated into a point mass at the end of a mass-less pendulum arm that represents the connection to the base of support (i.e., the supporting leg/foot). During normal walking, the center-of-mass moves along the pendulum arc (Cavagna et al., 1963) due to gravity until the other leg strikes ground. That contact creates a new pendulum arc for COM to move about, also under the influence of gravity. During the transition between arcs, termed the step-to-step transition (Kuo et al., 2005), some active modulation is possible with the work of collision from the leading leg and push-off from the trailing leg. However, this phase for modulation is sometimes overlooked to focus on the location of the center of pressure (COP), where the effective ground reaction force (GRF) vector is positioned on the ground.

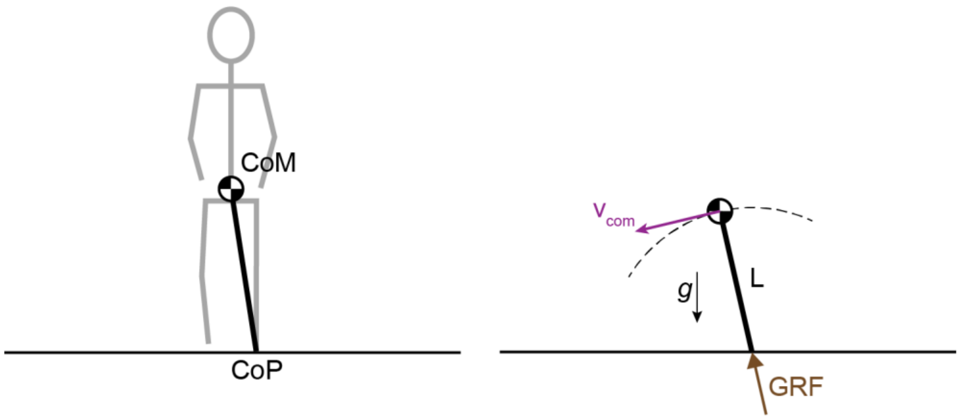

Figure 1.1: A simple inverted pendulum model of human walking as viewed in the frontal plane. The mass of the body is concentrated as a point mass the COM, and the stance foot forms the COP which is pinned to the ground. In this configuration, the GRF must point along the pendulum arm representing the leg with length L. Gravity *g* drives the inverted pendulum, which moves with velocity v_{com}. The resulting motion forms an arc on a sphere, with v_{com} perpendicular to the leg.

Step placement control involves controlling the landing position of the stepping foot to alter the lateral component of the GRF exerted on the COM. It is related to the concept of increasing the step width and thus the base of support to manage lateral movement variability (Bauby and Kuo, 2000). But rather than simply to keep the COM above the base, step placement control intends to move the landing foot in the direction of falling to a distance beyond the body COM in anticipation of its future motion, such that the COM never moves past the limits formed by the stepping foot. For perturbations towards the swing leg, this usually results in abducting the hips and moving the swing leg further away from the stance leg. For perturbations towards the stance leg, this often results in a cross-over step where the swing foot is moved towards the stance foot with hip adduction, and potentially past the stance foot such that the swing leg has "crossed over" the stance leg. The results of step placement control is a large-scale movement of the COP, with a more distance COP denoting the larger lateral component to the GRF vector which can work against undesirable movement of the COM. A preponderance of studies has focused on this strategy and shown that it is a powerful and significant contributor to lateral balance control (Hof et al., 2010; Joshi and Srinivasan, 2019; Townsend, 1985; Wang and Srinivasan, 2014; Winter, 1995).

The ankle strategy applies torque to the stance leg about the stance ankle to modulate COM velocity and is often marked by ankle inversion or eversion. The effect of this torque also moves the COP towards a lateral edge of the foot during stance, and can be viewed as another strategy to shift the COP. The effective movement of the COP is constrained laterally by the width of the foot (as the reaction force cannot be applied beyond the extents of the foot), so the impact of ankle strategy is generally less than that of step placement from perspective of the COP-COM relationship. However, the ankle strategy can be used and adjusted by any leg that is in contact with the ground, whereas step placement requires a foot to be off the ground for adjustment and cannot influence the COM until it lands, and thus it may be advantageous over step placement due to the immediate availability of its response. A number of studies have indicated that ankle strategy can impact lateral balance during walking but not necessarily to the extent of step placement (Hof et al., 2007).

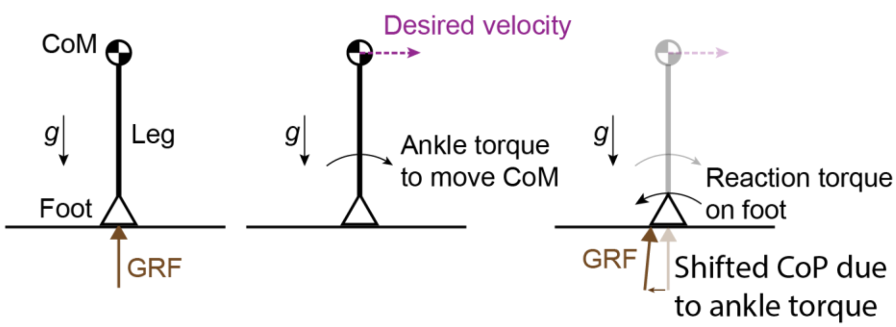

Figure 1.2: The effect of ankle strategy on COM velocity and COP in an inverted pendulum model on top of a mass-less foot. In order to accelerate the COM towards the desired velocity, a torque is applied to the leg at the ankle. The reaction torque then creates a moment on the foot, shifting the COP laterally to keep the foot stationary.

Push-off strategy describes using the force applied by the trailing foot before coming off the ground, to redirect COM velocity. Walking normally requires push-off to propel the body forwards via a significant torque at the ankle and smaller contributions from the knee and hip, with the additional effect of also propelling the body along the lateral pendulum arc of the other leg. This active work done during push-off comprises a significant amount of the total work done by the GRF during a step, and thus modulating the force can alter lateral velocity of the COM. Specifically, push-off force can be reduced on the leg away from the fall direction to minimize increasing COM velocity towards the fall direction and increased on the leg in the fall direction to raise the work performed to push back the COM towards balance. Several control models have demonstrated that such control of push-off force can significantly lateral balance during walking (Joshi and Srinivasan, 2019; Kim and Collins, 2017).

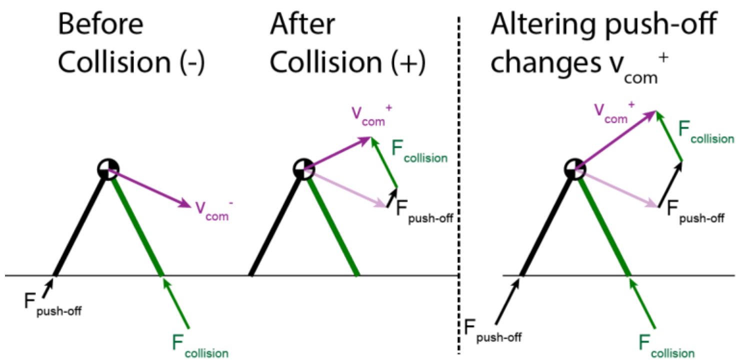

Figure 1.3: The effect of push-off and collision forces on an inverted pendulum model. The work performed by the push-off and collision forces effectively redirect and scale the COM velocity. The impulsive force of collision is determined by the pre-collision velocity of the COM and the angle of the colliding foot. Adjusting the angle and the magntiude of the force of push-off thus allows modulation of the COM velocity after collision. The model shown can be applied in three dimensions, and projected onto either the frontal or sagittal plane. Thus, push-off can affect COM velocity in the frontal plane.

Steering refers to changing the heading of the body, in part by twisting the foot in yaw before landing, which can adjust the resulting COP trajectory during the step and alter the walking direction to reduce the lateral component of a balance disturbance. Rotating the foot before landing can move the front of the foot laterally towards the direction of fall, thus increasing the lateral excursion of the COP through the normal back-of-foot to front-of-foot movement of the COP during a step. In addition, this steering can lead to a change in overall walking heading, shifting the fore-aft walking axis which has greater natural stability to be more aligned with the disturbed COM velocity, effectively allowing fore-aft balance mechanisms to help absorb some of the energy of the disturbance. It has been shown that humans tend to couple some external rotation of the foot with step placement and that a simple walking model with foot yaw can stabilize small lateral disturbances, suggesting that steering may work in conjunction with other stepping strategies to help lateral balance (Rebula et al., 2017).

The dominant ability of stepping strategies for modulating balance has led to the development of several measures and analyses to conceptualize how stepping can be performed. Of particular interest is the extrapolated center-of-mass (XCOM) proposed by Hof, which adapts the COM-base of support concept from standing balance to include a factor for the velocity of the COM using the inverted pendulum model. In that model, a non-moving mass will be perfectly (but unstable) balanced if the COP is precisely underneath it. For a moving inverted pendulum within the small angle approximation regime, the mass will come to a perfect stop if it is moving towards the COP and starts at a distance equal to the velocity divided by the natural frequency of the pendulum. This is the factor of velocity added to the position of the COM to produce the XCOM.

In practice, because moving the leg takes time, the instantaneous XCOM is an underestimate of ideal COP placement until the swing leg comes into contact with the ground. Furthermore, to maintain walking, the COM must move back and forth between the lateral COP positions set by the alternating feet, so the actual target position should be beyond the XCOM even upon ground contact to prevent the COM from coming to a complete stop. Experimental studies indicated that humans do tend to step such that the maximum excursion of their COP exceeds the maximum excursion of the XCOM by some margin.

$$\mathrm{XCOM} = x_{\mathrm{com}} + \frac{v_{\mathrm{com}}}{\sqrt{g/L}}$$
$$x_{\mathrm{com}} = \text{lateral position of COM}$$
$$v_{\mathrm{com}} = \text{lateral velocity of COM}$$
$$\sqrt{g/L} = \text{natural frequency of inverted pendulum with length L}$$

I will reference the XCOM throughout this work as it provides a simple to compute descriptor of balance from the stepping perspective, but it should be recognized that XCOM is not and was not meant to be a robust predictor of balance. Namely, there is no prediction for the actual margin between XCOM and COP required for lateral balance, only an identification that humans tend to keep a minimum margin in some experimental cases. One aspect it doesn't consider, for example, is the work performed by the ground reaction force in collision and push-off phases during the step-to-step transition. Several models that factor in this work have demonstrated stable predictive control of COP placement for balanced walking (Joshi and Srinivasan, 2019; Kim and Collins, 2017). That being the case, XCOM and its margin with COP still provide a strong correlative measure of stability margin obtained from stepping. Furthermore, none of these models provide a clear structure for understanding balance control mechanisms beyond stepping, a gap I seek to address.

1.3 Inertial balance strategies could supplement stepping but is poorly explored

Besides stepping, a separate inertial balance strategy is possible, using the angular momentum of the body by moving trunk and limbs to induce a stabilizing reaction force through the stance leg (Hof, 2007). This category includes mechanisms such as a trunk or hip strategy, and has been demonstrated in standing postural balance (Otten, 1999) and perhaps very low-speed walking (Best and Wu, 2020) but is discounted in normal walking (Hof et al., 2010) and generally sidelined in literature except for a few instances (Arvin et al., 2016; Martelli et al., 2015). One possibility is that inertial balance strategies operate only in the realm of standing or highly constrained walking such as walking on a narrow beam or at extremely low speeds. An alternative possibility is that angular momentum is continuously controlled for balance during walking but is less useful than stepping strategies generally, and thus is harder to identify except under conditions where stepping strategies are strained, such as larger perturbations from normal walking or during specific phases of the gait cycle. I propose that this alternative is indeed true, that angular momentum can play a significant role in lateral balance, and that it is present in normal walking control if we can identify the phases that favor its usage the most.

1.4 Using lateral perturbations to simulate balance conditions

Throughout this work, I will use lateral perturbations at or around a person's center-of-mass (COM) near the pelvis to generate conditions that required more active balance control. The goal of this location is to minimize rotational moments on peripheral parts of the body resulting from the perturbation itself in order to help distinguish any rotations that might be controlled by the individual. Meanwhile, the perturbation can serve as a proxy for some other forms of balance disruptions such as trips or slips: the perturbation results in an unexpected position and velocity of the center-of-mass of the body relative to the base of support, which can be analogous to the result of a trip (potentially a missed step and a disruption to COM trajectory and velocity due to external forces on the tripping foot/leg) or a slip once traction is regained (COM position and velocity has changed due to gravity and insufficient control during the slip). Thus, the results here could have clearer and broader applicability to balance than just the perturbation conditions presented.

1.5 Balance responses should diverge between perturbations to stance or to swing

There should be a significant difference in balance responses between perturbations towards the stance leg and perturbations towards the swing leg, due to the constraints of the body's geometry. This difference should apply to both stepping and inertial balance strategies, and may be better understood by considering two schematics of the walking human in the frontal plane. We begin with a stick-figure simplification of human geometry (Figure 4), seen walking into the page, with the right leg on the ground in stance and the left leg in the air in swing. If we perturbed this figure to each side to represent a fall in that direction, we can glean two results due to geometry: in a fall towards the swing-leg side, the swing foot will naturally be brought closer to the ground (Figure 4 left), while in a fall towards the stance-leg side, the swing foot will be further from the ground (Figure 4 right). A reasonable implication of this may be that a swing-side perturbation will lead to a quicker side-step to regain balance, while a stance-side perturbation will have a slower side-step due to the greater distance of movement required. Furthermore, the desirable foot placement in a stance-side fall may be further than the contralateral foot and require additional maneuvering to get around the contralateral leg, thus taking even more time. Considering this, we may intuit that a stance-side perturbation would be a more opportune situation for inertial balance strategies to be beneficial.

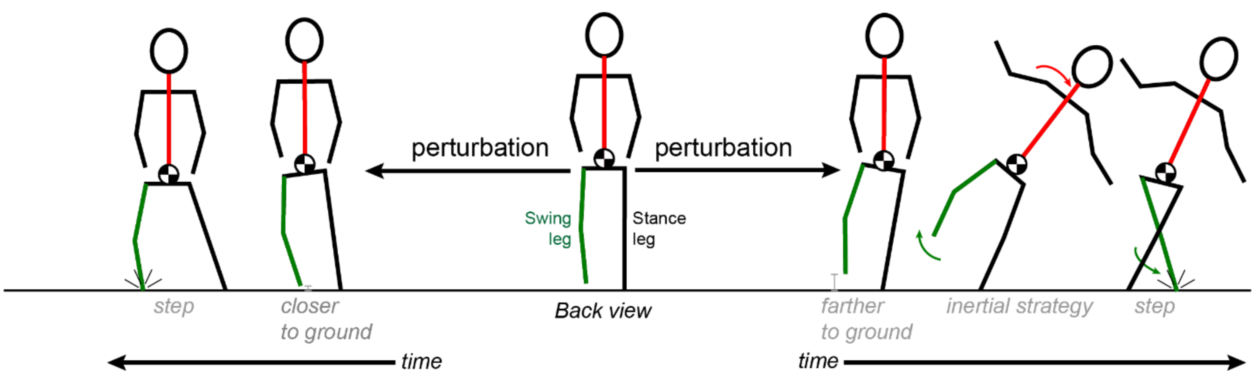

Figure 1.4: Back view of human figures in response to lateral perturbations. Perturbations towards the swing leg (left) brings the swing foot closer to the ground and allow for a quicker lateral step to reject the perturbation. Perturbations towards the stance leg (right) moves the swing foot further from the ground, increasing the time required to step. Thus, we might imagine stance-side perturbations are an opportune case for using inertial balance strategies in order to act on the falling velocity before stepping is available.

We can better understand the mechanical implications of angular momentum using the inverted pendulum model (Figure 1.5). We will shortly consider how motions of the other body appendages affects balance, but the dominant effect begins with the pendulum-like COM motion. During each pendular arc, a perturbation force can exacerbate natural falling due to gravity by increasing the velocity towards the swing-side or reducing or reversing the velocity towards the stance side, thus requiring the swing leg to be repositioned in order to properly “catch” the COM. The catch position should be somewhere beyond the XCOM to maintain the pendular arcs seen in walking. A larger perturbation would result in a more lateral velocity and a further XCOM, requiring a further step placement to regain balance.

The illustrated stepping response also highlights the geometric challenges facing a stance-side perturbation more clearly. First, the natural position of the swing leg means the end of the leg moves further from the ground as a result of the pendular arc movement of the COM. As a result, the swing leg must move a larger angular distance medially to make ground contact. Second, this movement of the swing leg must pass through the stance leg for perturbations beyond a magnitude that can push the COM over the top of the stance foot. In reality the stance leg is an obstruction that the swing leg cannot move through, and thus it must take a longer trajectory around the stance leg to move to the other side. A way to account for this greater distance may be to move the swing leg faster. However, muscles have limitations to how much and how fast they can produce force (Maffiuletti et al., 2016). Thus, it is likely that moving the swing leg will take more time to cover the distance, allowing gravity to pull the COM further along the pendular arc and increasing the lateral placement of the foot necessary to account for this additional fall time. In constrast, a swing-side perturbation naturally moves the swing leg closer to the ground, allowing it to quickly contact the ground while moving laterally and limit the time gravity can further drive the COM velocity.

These geometric differences could encourage augmenting step placement with other balancing actions in response to perturbations towards the stance leg. Specifically, we may expect that inertial balance strategies should be used during the movement of the swing leg to reduce the fall of the COM before step placement is ready. The mechanism for this is illustrated in the bottom right of Figure 1.5. The inertia of the peripheral parts of body (e.g. trunk, arms, swing leg) is embodied in a massless trunk segment (another possible representation could be a reaction wheel). The body can rotate the trunk towards the direction of the fall, thus generating a reaction torque in the stance leg. Due to horizontal contact constraints on the stance foot, this results in a reaction force that pushes the COM away from the falling direction.

To summarize, we predict two categories of balance responses distinguished between swing-side perturbations and stance-side perturbations. In swing-side perturbations, we expect a quick lateral change in step placement will be sufficient to reject the perturbation and relatively little contribution from inertial balance strategies. In stance-side perturbations, we expect inertial balance strategies to play a larger role, potentially limiting the rate of fall of the body until the swing leg can move into position for proper step placement.

1.6 Timing of perturbation during stride phase changes effective categories

One implied assumption in the categories proposed is that there is sufficient time during the perturbed step to perform the desired balance control strategy. However, the dynamic constraints of walking can limit the performance of any strategy if the perturbation occurs later in the step: momentum will already be carrying the body towards the unperturbed pose targets, and

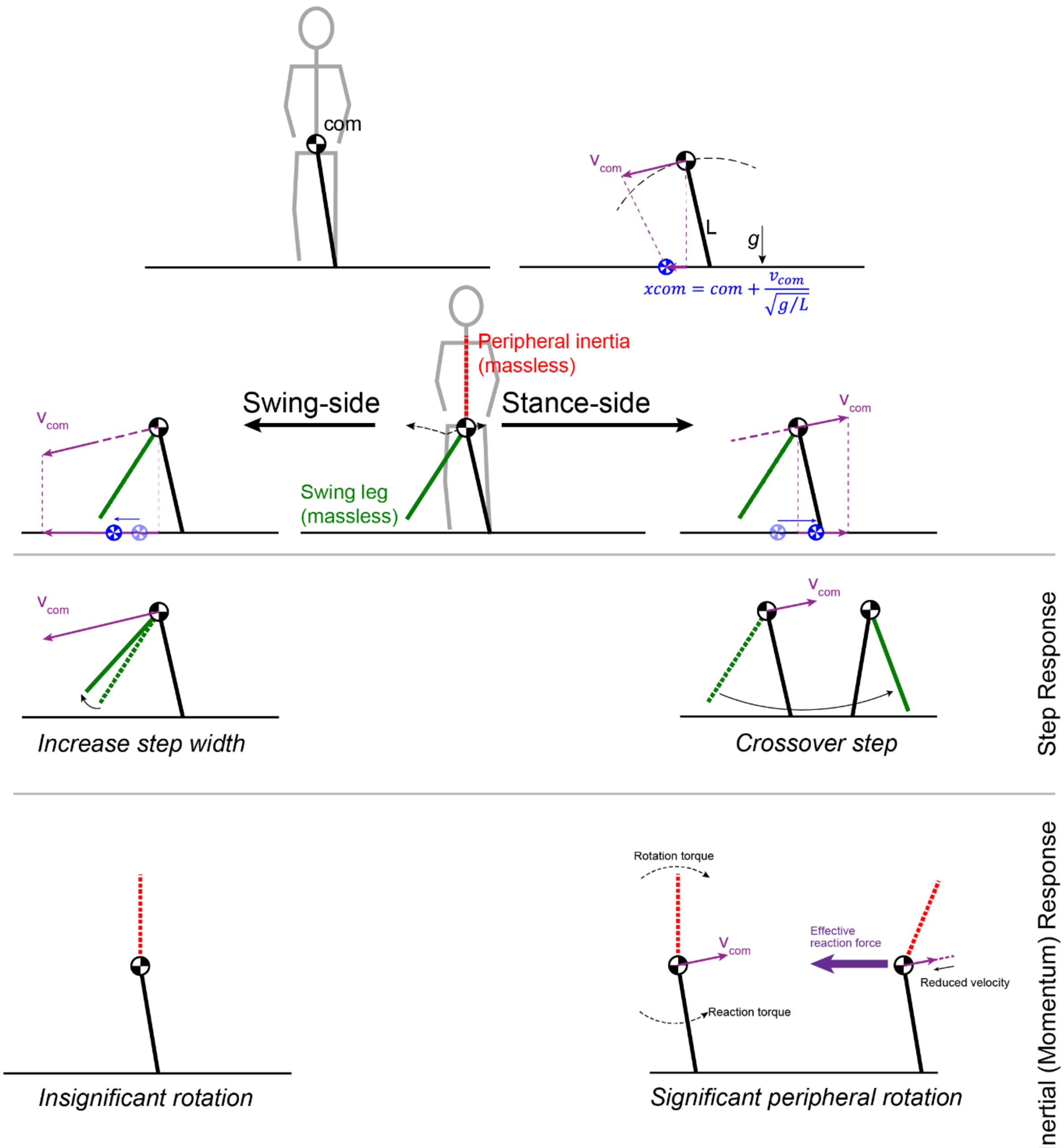

Figure 1.5 An inverted pendulum model of human walking in the frontal plane. The model includes massless stance (black) and swing (green) legs and a massless reaction wheel with inertia to represent peripheral inertia of the trunk, arms, etc. (represented by red dashed segment for convenience). Balance perturbations can be classified as swing-side (towards the swing leg) and stance-side (towards the stance leg). The result of the perturbation is increased COM velocity in that direction, which also shifts the XCOM, a measure combining the lateral position of the COM with factor of the COM velocity scaled by the natural frequency of the inverted pendulum (blue). Movement of the swing leg towards the perturbation direction to "catch" the COM is expected from both perturbation sides. The swing leg should need to move a relatively small distance towards swing-side perturbations since it is already located in that direction (left), thus producing a quick step that prevents the COM from gaining more velocity due to gravity and requiring little use of rotational inertia. However, the swing leg must move farther and perhaps crossover the stance leg if the perturbation is towards stance (right). The further movement may take longer, potentially allowing gravity to increase the rate of fall. Thus stance-side perturbations may be an opportunity to use inertial strategy: by rotating the peripheral inertia towards the perturbed direction, the reaction torque on the stance leg pushes against the ground, thus creating an effective reaction force that pushes the COM away from the perturbation direction.

reaction time and muscle limits prevent dramatic changes in trajectory. Thus we expect that late perturbations will be largely reacted to in the following step, with the leg in stance changing, and the response reflecting that change. In other words, we expect late swing-side perturbations to behave like stance-side perturbations in the following step, and late stance-side perturbations to behave like swing-side perturbations in the following step.

1.7 Chapters outline

To determine the contributions of direct and indirect strategies or lateral balance in human walking, I will use simple dynamical modeling to examine the expectations of what inertial balance strategies can do, followed by human subject experiments to validate the presence of expect behaviors. Chapter 2 describes a three-dimensional walking model with controllable inertial strategy, ankle strategy, and step placement mechanisms and looks at how they can be controlled separately and together, as well as their varying effects on balance and step durations. Chapter 3 explores the presence of inertial strategy behaviors in human subjects when laterally perturbed during treadmill walking and the dependence of that behavior on perturbation time relative to when it occurs during the step. Chapter 4 extends the experimental analysis by applying stepping constraints to the human subject on the treadmill, with the expectation that even small limitations to step placement effectiveness can increase reliance on inertial balance strategies to compensate.

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Chapter 2 A Model of Inertial Contributions to Lateral Balance During Walking

2.1 Introduction

Maintaining balance is a critical task during walking. The inverted pendulum dynamics in humans is inherently unstable, and stabilization is required in multiple axes to avoid falls (Kuo, 1999). This challenge is exacerbated during walking due to periodic changes to the base of support with each step, as well as movement of the body center of mass (COM). Understanding the dynamics of movement and how balance mechanisms interact with those dynamics could help to explain how humans actually balance.

At present, several stepping-related strategies have been characterized: step placement (Bauby and Kuo, 2000; Hof et al., 2007; Kuo, 1999), ankle torque (Hof et al., 2010), and push-off control (Joshi and Srinivasan, 2019). These strategies use and manipulate the center of pressure (COP) under the feet to restore balance. However, another possibility is to move the body's inertia, effectively manipulating angular momentum to affect balance, without necessarily using COP. Mechanisms of this inertial strategy could include tilting the trunk and hip, flailing the arms, and moving the swing leg in curved trajectories before the next step location. While some of these mechanisms have been studied in standing postural control (Nashner and McCollum, 1985; Kuo and Zajac, 1993), there has been limited investigations indicating impact to balance during walking.

Both stepping and inertial strategies should be able to stabilize balance in humans (Figure 2.1). Stepping strategies primarily exert control on the COP and the vertical ground reaction force acting through it on the body. This counteracts the force of gravity acting on the body COM, which in the inverted pendulum model generates a destabilizing moment whenever the COP is not positioned directly below the COM. Inertial strategies, meanwhile, affect the lateral ground reaction force, which indirectly moves the body COM.

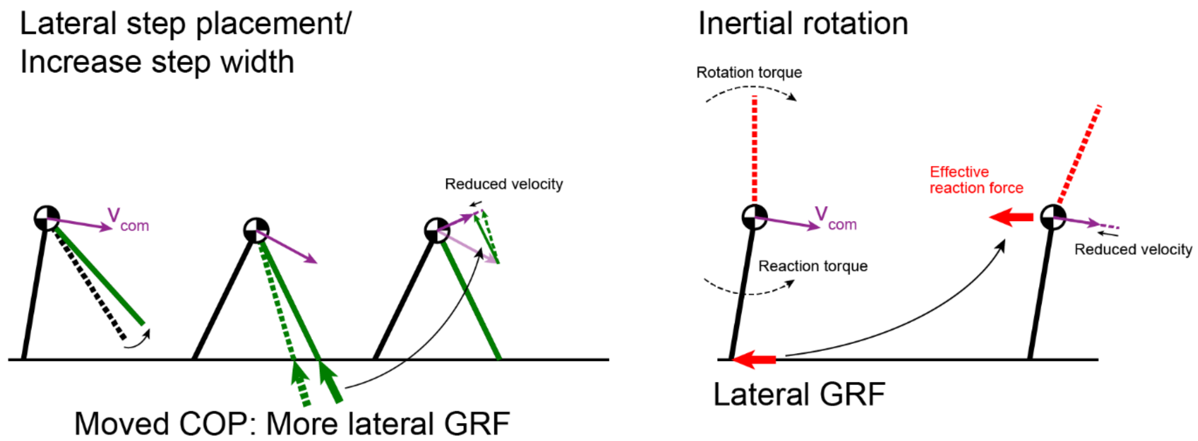

Figure 2.1 Examples of how stepping strategy and inertial strategy can influence stability. Assuming a balance perturbation that increased COM velocity (v_{com}) to the right of the figure, lateral step placement (left) can be used to move the COP further out for the following step, generating a more lateral GRF that will reduce v_{com}. Alternatively, rotation of inertial mass about the COM (represented as a massless inertial segment) generates a reaction torque on the stance leg that pushes against the ground, producing a lateral GRF that can also reduce v_{com}.

It may be more difficult for humans to control lateral than fore-aft balance during walking, and thus may require greater inertial strategy contributions. Modeling studies have shown that forward stability can be obtained passively while lateral stability required active control (Kuo, 1999; O'Connor and Kuo, 2009). Experimental studies have shown that stabilizing lateral movements on a treadmill reduces gait variability (Dean et al., 2007) and energy consumption (Donelan et al., 2004), as indications of control effort.

There have been a number of studies and proposals about how humans control lateral balance with stepping strategies (Eng and Winter, 1995; Hof et al., 2010; Joshi and Srinivasan, 2019; Townsend, 1985; Wang and Srinivasan, 2014). Of particular interest is the extrapolated center of mass (XCOM). In brief, the XCOM is the position where the COP of an inverted pendulum should be instantaneously placed to allow gravity to stop the COM in a balanced state above the COP (Hof et al., 2005), and is computed by adding the lateral position of the COM with the lateral velocity of the COM scaled by the natural frequency of the inverted pendulum:

$$\mathrm{XCOM} = x_{\mathrm{com}} + \frac{v_{\mathrm{com}}}{\sqrt{g/L}}$$

$$x_{\mathrm{com}} = \text{lateral position of COM}$$

$$v_{\mathrm{com}} = \text{lateral velocity of COM}$$

$$\sqrt{g/L} = \text{natural frequency of inverted pendulum with length L}$$

Some experiments have found that normal walking and stepping response to lateral balance perturbations is consistent in placing the COP beyond the XCOM (Hof et al., 2007; Hof et al., 2010). While the exact strategy that humans use for lateral balance may be different and more complex, a strategy targeting step placement beyond the XCOM could provide a reasonable approximation of human behavior, and the margin between the COP and the XCOM can provide an estimate of lateral stability, with a larger margin suggesting greater stability.

In contrast to stepping strategies, it is unknown whether and how much humans employ inertial strategies for lateral balance during walking. Inertial compensations have been demonstrated by humanoid robots (Atkeson and Stephens, 2007), which actively use the trunk and arms to facilitate acrobatic maneuvers (Guizzo, 2019). A tight-rope walker can also balance without lateral stepping by moving the rest of the body such as the trunk, arms, and swing leg. While some studies have described motions in the rest of the body during walking (Hof et al., 2010; Winter, 1995), few have found active use of inertial balance strategies. Here, we will demonstrate the feasibility of using inertial strategies in response to lateral balance perturbations during walking in a simple dynamical model. We will also examine the impact of inertial strategies on the use of stepping strategies such as lateral step placement through a one-step multi-strategy controller. Finally, we will consider some conditions which may help to quantify the use of inertial strategies in humans. We expect that in response to perturbations, increasing use of inertial strategies will increase the margin between the COP and the XCOM obtained from stepping strategies alone and allow for a decreased use of stepping strategies to obtain the same degree of stability.

2.2 Methods

We created a 3D walking model that generates a significant portion of its walking motion though passive pendulum-like dynamics. The model is based on (Kuo, 1999), which showed a system with two legs connected by a pelvis could be passively stabilized in forward motion but required active stabilization for lateral motion. We extended that model by adding a vertical trunk segment that can be commanded to tilt in the frontal plane about the pelvis to represent upper body rotation. We then examined the interaction between trunk tilt, hip abduction mid-swing, and final step width on lateral stability after the model is laterally perturbed.

The model is an extension of previous three-dimensional dynamic walking models (Figure 2.2). The motion is governed in large part by underactuated, passive dynamics, but with active control also available for critical degrees of freedom. Each foot is a portion of a cylinder that rolls forward on the ground. Each leg is attached to the foot via a freely-rotating hinge joint ankle that permits side-to-side movement about the roll axis and can also inject a commanded torque for control. The two legs are attached through the hip segment via a hinge that allows free swing in the pitch axis. A controlled degree-of-freedom regulates the abduction/adduction of the legs relative to the pelvis, with a single angular deviation applied to both legs symmetrically, such that the splay of the legs can be controlled. Finally, two controlled degrees-of-freedom control the pitch and roll of the trunk segment relative to the center of the pelvis, with both angles set to hold the trunk vertical to the world and the torques required propagated through the model via inverse dynamics. Additionally, the roll angle of the trunk relative to the world can also be commanded for inertial balance control. The walker is powered by an impulsive push-off from the stance foot ground contact through the model's center-of-mass, and swing is assisted by a hip spring. Length, mass, and inertia properties were chosen based on normalized human data (Table 2.1). The torques required to control the degrees-of-freedom are applied throughout the model during the step. An external force can be applied laterally to the midpoint of the pelvis to perturb the model.

To model human behavior, we selected parameter values that yielded approximately human-like walking and response to balance perturbations. For normal walking, the parameters produce a limit cycle when the controlled degrees-of-freedom are held at nominal values (producing no additional ankle torque, neutral hip splay, and a vertical trunk in pitch and roll relative to world). We characterized the stability of the model similar to (Rebula et al., 2017), by linearizing the one-step return map and analyzing the resulting discrete linear system. We then generated linear controllers to stabilize the system, and examined the signs of and changes in control gains to estimate the impact of inertial strategy relative to step placement for responding to balance disturbances.

Lateral balance perturbations were introduced to the model and manipulated to examine the effect of balance control on step parameters. Perturbations were introduced using a constant external lateral force at the pelvis for 0.1 dimensionless time (approximately 1/16 of the unperturbed stride duration), starting at various times throughout the stride and in either lateral direction (e.g. *stance-side* when towards the leg in stance connected to the ground and *swing-side* when towards the leg free to move). We examined how the one-step linear controllers would respond by back-integrating from the state of the uncontrolled model after the perturbation to find initial conditions that would produce the resulting disturbance without a discrete perturbation and then using the computed initial conditions to compute control inputs. We also compared resulting step parameters (lateral position and duration) of the perturbed step with the computed controls against a step controller based on placing the foot laterally beyond the lateral XCOM.

To understand the interaction between inertial strategies and step placement, we conducted a parameter study using the model to examine how varying intensities of trunk roll and mid-swing hip abduction controls versus lateral foot placement control impacts balance and step timing. We determined the balance margin of the model by computing the difference between the COP (defined by the point of contact of the rolling stance foot) and the XCOM, with a larger positive balance margin (when the CoP is more lateral than the XCOM) suggesting greater balance. Balance margin and step duration were evaluated around the control parameter space where lateral foot placement alone could obtain a positive balance margin after the model was laterally perturbed early in the step towards the stance leg.

2.3 Results

Using the walking model, we found that active control of either lateral foot placement, ankle torque, or trunk roll (as a representative of inertial strategy) could stabilize the lateral motion of walking. This was determined by assuming a nominal walking cycle and analyzing the resulting step response to a disturbance in the initial states. We started by finding a limit cycle of walking dynamically comparable to human walking using a shooting method, and linearized the system about the fixed point at heelstrike to create a matrix of partial derivatives representing initial state perturbation response, *A*. The resulting discrete linear system

$$\Delta x_{k+1} = A\Delta x_k$$

represents the one-step response of the full nonlinear system for sufficiently small perturbations to initial states, and its stability can be characterized by examining the poles of the open-loop system (Kuo, 1999). We found one unstable mode outside the unit circle corresponding to lateral falling about the stance ankle roll (-3.2591). We then linearized the system with respect to inputs to generate B_{ankle}, B_{splay}, and B_{trunk}, corresponding respectively to an ankle torque command (defined as a constant added torque at the joint), a leg splay command (defined as the angle deviation relative to nominal splay, changing from 0 to the commanded angle in the shape of a sinusoid over time from negative amplitude to positive amplitude) and a trunk roll command (defined as the angle relative to the world frame, changing from 0 to the commanded angle and then back to 0 in

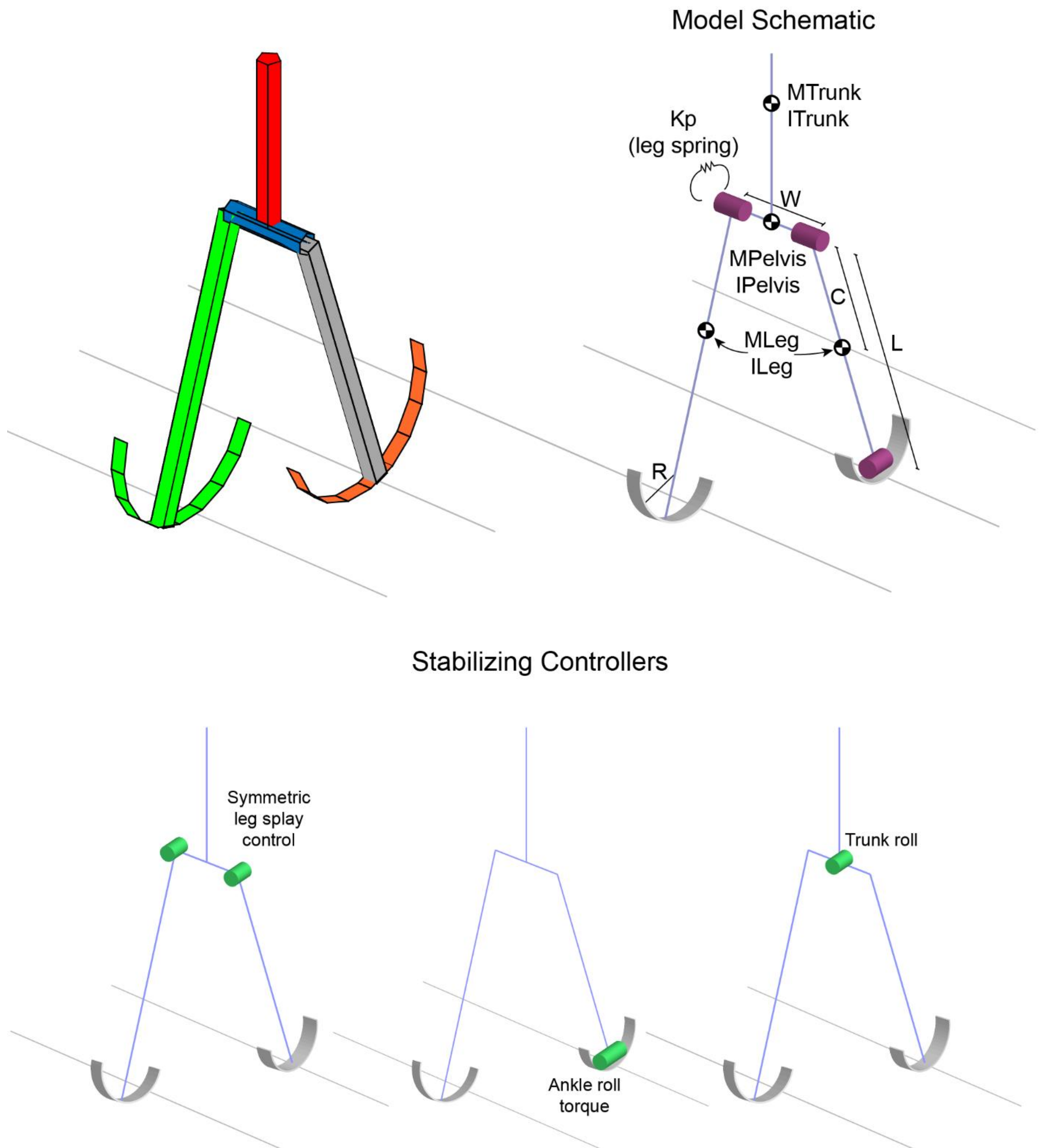

Figure 2.2: Schematic of the 3D walking model. The model can perform a roughly anthropomorphic gait limit cycle while the leg splay control is held fixed, the trunk roll control maintains a vertical trunk orientation, and the ankle roll control is held at zero.

Table 2.1 Walking model parameter values. All parameters are dimensionless and normalized with respect to gravity, leg length, and total mass.

Parameter	g	L	C	R	W	MPelvis	Ipelvis	Mleg	Ileg	Mtrunk	Itrunk	Kp
Value	1	1	0.645	0.3	0.3	0.14	0.0051	0.16	0.017	0.54	0.01	0.089

the shape of a sinusoid from negative amplitude to positive amplitude to negative amplitude). This allowed us to create linear systems of the form

$$\Delta x_{k+1} = A\Delta x_k + Bu_k$$

where u is either the commanded ankle torque, the change in leg splay angle, or the change in trunk roll angle, matched with their respective B matrix. We then created stabilizing controllers using pole placement, reflecting the unstable poles about the unit circle, of the form

$$u_k = K\Delta x_k$$

The resulting control gains K confirm that the ankle should apply torque to rotate the body away from the perturbation direction, the leg splay angle should change to move the stepping foot towards the perturbation direction, and the trunk should roll towards the perturbation direction. The relative size of the gains (K_{ankle} = 0.8414, K_{splay} = -2.0694, K_{trunk} = -7.7218) also suggests that given our commanded angular trajectories, trunk roll must undergo significantly more angular change to create a similar stabilizing effect.

2.3.1 Controlled step placement parameters are distinct between stance and swing sides

We found distinct differences in step placement and durations as a result of control. Using the linearized controllers for the one-step response, we determined the resulting step placements and step durations for lateral perturbations of 1 dimensionless force magnitude over 0.1 dimensionless time, ending at various times throughout the step (simulated by initial state deviations computed by back-integrating the post-perturbation state without the perturbation force), and to both stance-side and swing-side directions (Figure 2.3). We also compared the difference in step parameters with the step placement (hip splay) controller alone, with a B_{combined} matrix and input that included hip splay and trunk roll, and another B_{combined} that further included ankle torque. Step placement was notably in the direction of the perturbation in all case, with greater step placement deviation after perturbations earlier in the step than later in the step. This could be explained by the fact that earlier perturbations have longer to deviate under the effect of gravity before the swing foot lands, thus increasing the overall balance perturbation that must be corrected and thus increasing the lateral placement required. The combined step placement and trunk roll controller decreased the step placement deviation made in all circumstances, and the step, trunk, and ankle combined controller decreasing further still, suggesting that multiple strategies including inertial strategy can alleviate some of the control effort required of step placement alone.

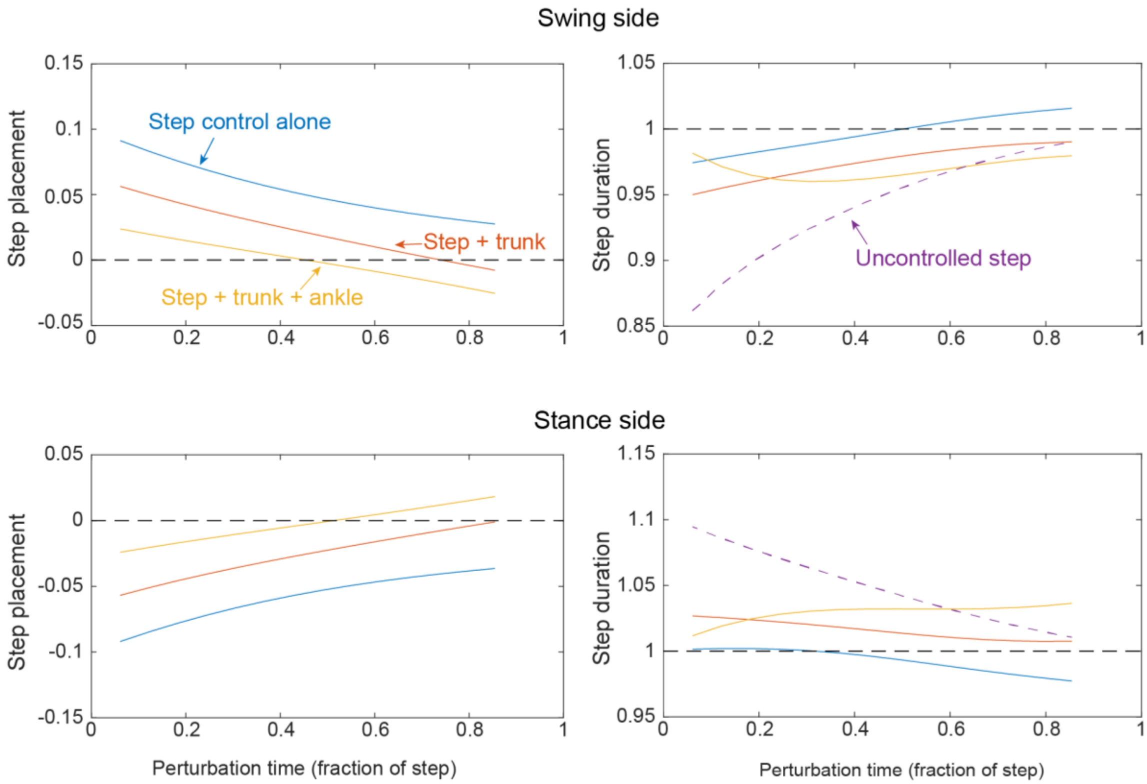

Figure 2.3 Lateral step placement (left) and step duration (right) for one-step return map controllers responding to perturbations throughout the step. With the addition of trunk control (red line) and ankle control (yellow line), step placement deviation (lateral for swing-side, medial for stance-side) decreased towards or passed nominal (left). In swing-side perturbations (top), this was coupled with a slight decrease in step duration, which was generally below nominal (1) throughout the step. In stance-side perturbations (bottom), step durations with step placement along were approximately nominal duration or shorter for later perturbations, but longer than nominal with added trunk and ankle controls.

Step duration behavior as a result of control reflected the duration of the uncontrolled perturbed step for swing-side perturbations but differed somewhat for stance-side. Across controllers, swing-side perturbations resulted in shorter-than-nominal-unperturbed step durations, with durations increasing for later perturbations. Combined control with step placement and trunk roll decreased step durations across all perturbation times, while adding additional ankle torque created a more complex duration behavior over perturbation times (Figure 2.3 upper right). Stance-side perturbations controlled through step placement alone resulted in nominal to shorter-than-nominal step durations, while adding trunk roll and ankle torque resulted in longer-than-nominal durations for perturbations throughout the step, similar to uncontrolled perturbed step behavior. Later stance-side perturbations generally resulted in shorter duration steps, with added ankle torque control creating a more complex duration behavior over perturbation time similar to a reflection of swing-side (Figure 2.3 bottom right).

While the results using the one-step controller is helpful for validating the ability of step placement, ankle torque, and trunk roll to respond to various scales of lateral perturbations, the required linearization, back-integration to use the controller, and reimplementation in the original perturbed step may mask intra-step behaviors that can alter step parameters. For example, the back-integration process and one-step control essentially allows the controller to predict the final magnitude and time of the perturbation and begin responding immediately as the step begins before the perturbation occurs. Practically, starting balance compensations after the perturbation reduces the time available to perform the desired controls and may increase torques used to actuate the joints. As these torques are moving inertial masses in the swing leg and trunk, there could be significantly greater inertial consequences when controlling for balance in real time after perturbations. Designing a controller to manage these additional effects becomes significantly more complex, however, and human-like control strategies for inertial movement are particularly not well understood.

As an alternative we considered an XCOM stepping controller to place the foot laterally at the XCOM after the perturbation. The controller is an optimization that finds the hip splay angle command necessary to place the foot on the XCOM after the next foot contact. In this case, the hip splay command begins immediately after the perturbation completes and set to complete by 1.5 dimensionless time. The result should be a step placement control that effectively mitigates the perturbation in one step, leaving the model in a state of quasi-stable balance at the beginning of the following step. This strategy resembles behavior found in human experiments, minus a margin that may facilitate next-step dynamics such as a return to normal gait (Figure 2.4).

The XCOM step controller resulted in similar step placement and duration behavior compared to the one-step controller (Figure 2.5). As with the one-step controller, commanded lateral step placement deviations were higher after early perturbations and reduced for later perturbations. However, the XCOM controller resulted in much smaller swing-side deviations than stance-side deviations. Step durations were also similar to the one-step controller for step placement alone, with both perturbation sides resulting in shorter-than-nominal step durations and early swing-side significantly shorter than stance-side. Later swing-side perturbations resulted in durations that consistently increased towards nominal, while later stance-side perturbations decreased step duration gradually but inconsistently, and generally changed less significantly than swing-side step durations.

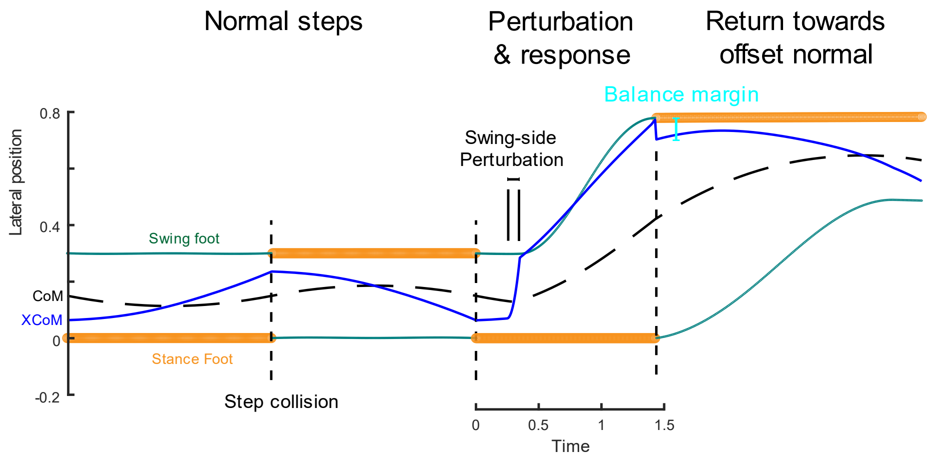

Figure 2.4 Model response to a swing-side perturbation in one step targeting a step placement with a balance margin beyond the XCOM after collision, with a return towards limit cycle gait in a laterally-offset position in the following step. This behavior more closely resembles previous human results when including the step following perturbation. The XCOM stepping controller optimization targets a balance margin of 0.

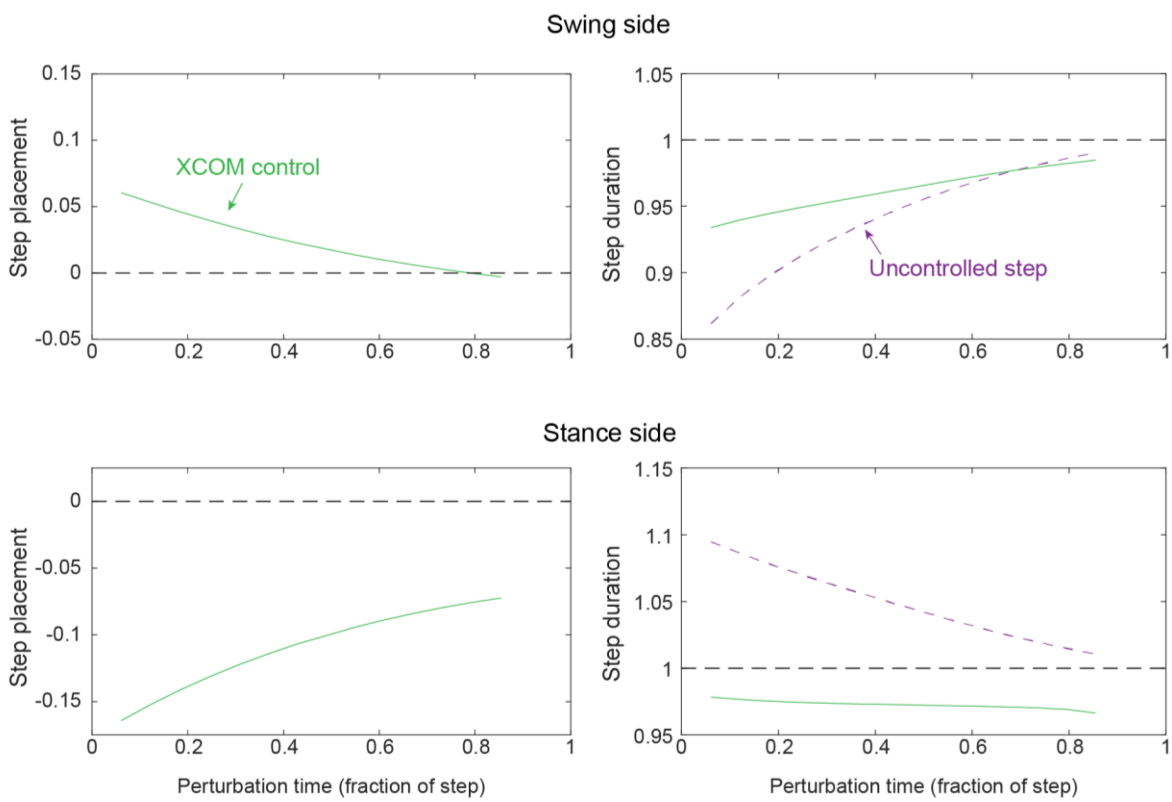

Figure 2.5 Lateral step placement (left) and step duration (right) for XCOM stepping controller in response to perturbations throughout the step. Step placement deviation from nominal (0) was lateral and lower in magnitude for swing-side perturbation than the medial placement magnitude for stance-side perturbations, and the deviation for both sides decreased towards nominal after perturbations later in the step. In swing-side perturbations (top), this was coupled with a step duration below nominal (1) throughout the step that gradually approaches nominal after later-step perturbations. In stance-side perturbations (bottom), step duration was shorter than nominal throughout and became increasingly shorter as a result of later perturbations. The overall rate of change in duration for stance-side was less than that for swing-side.

An important implication when examining XCOM step control is that moving the swing leg for step placement exacerbates the rate of falling. This can be clearly seen in an example of early perturbations towards both stance-side and swing-side (Figure 2.6). When the perturbation occurs without a commanded step placement (2.6 left), there is a sudden change in the XCOM as a result of the perturbation, then a more gradual excursion due to gravity as the model falls. When we command the swing leg to move to generate a step placement with zero balance margin (Figure 2.6 right), the rate of XCOM deviation after the perturbation is greater than that due to gravity alone. This is in part because moving the swing leg towards the desired position in the direction of falling also shifts the body center of mass further in that direction. Furthermore, the rotation of the swing leg in the frontal plane is also opposite of the rotation desirable for inertial balance strategy, and thus generates an undesirable reaction force that exacerbates the fall.

There is also an additional complication for stance-side perturbations not considered by our controllers: the naïve step placement command by both controllers previously does not consider contact between legs, but a cross-over movement early in a step could collide the swing leg into the stance leg. To avoid this collision, the swing leg must wait until it has swung past the stance leg to cross-over. We can model this by delaying the start of the step placement response (Figure 2.7). Here, we delay the onset of lateral movement until 0.7 dimensionless time, delaying the crossover between swing foot position and stance foot position from 1.1 to 1.25 dimensionless time into the step. The resulting lateral placement required does not change significantly, but the shorter time period in which it must occur means that the required movement speed of the leg, and thus the torque required to accelerate the leg, increases significantly (Figure 2.7 right).

2.3.2 Inertial balance strategy can improve balance margins and increase step duration

Given the greater challenges to step placement after stance-side perturbations, it is possible that inertial strategies could be useful to reduce the magnitude and rate of step placement required. The principal inertial strategies to be considered are active trunk motion (similar to a lateral hip strategy) alone, swing leg motion alone, and both together. In the case of stance-side perturbations (Figure 2.8), the appropriate inertial strategies are to rotate the trunk in the direction of the perturbation, and to abduct the swing leg to induce a similar inertial impulse to slow the fall. Consider the lateral position of the stance foot, swing foot, COM, and XCOM over two steps, and their deviations between nominal (Fig. 2.8A) and perturbed (Fig. 2.8B) at 0.25-0.35 dimensionless

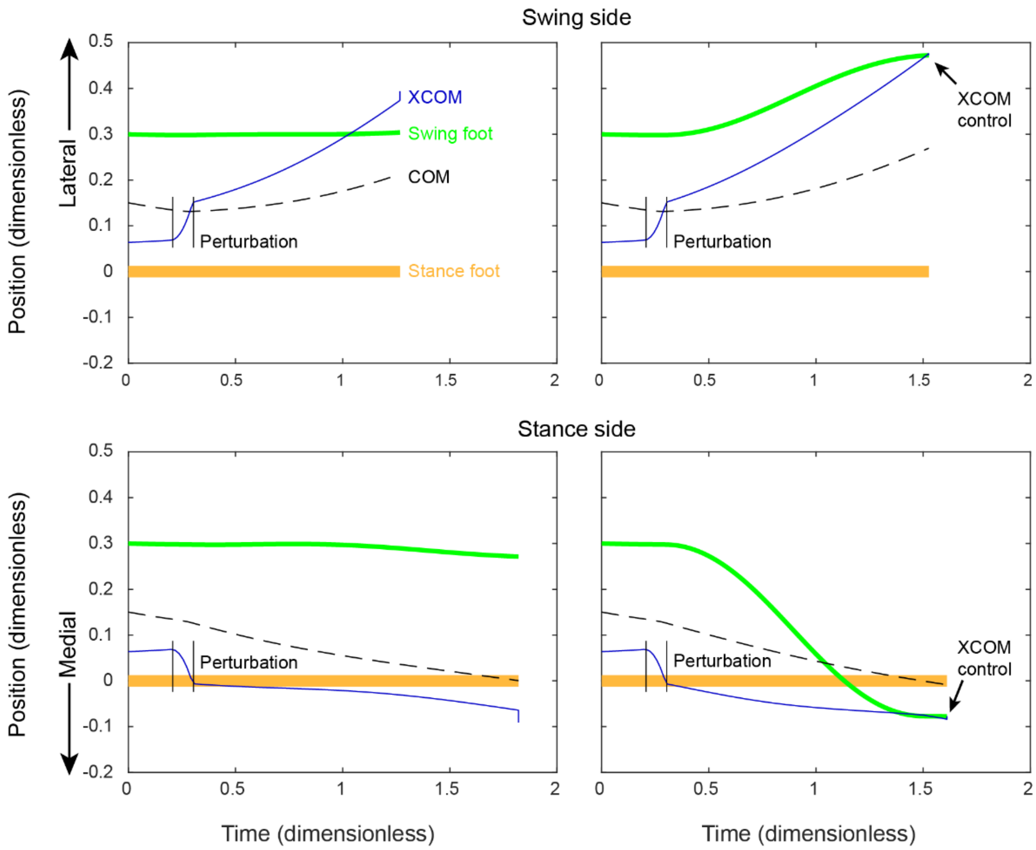

Figure 2.6 Lateral positions of the stance foot, swing foot, COM, and XCOM after a perturbation at 0.2-0.3 dimensionless time towards swing-side (top) and stance-side (bottom). Without a step placement command (left col), the perturbation causes a clear deviation in the XCOM and COM trajectory that continues more gradually after the perturbation. For swing-side perturbation, the swing foot naturally contacts the ground shortly after, while for stance-side perturbation, the step duration increases. In both cases, the XCOM ends beyond the foot position, suggesting the model will continue to fall in the following step. With a step placement command (right col), the XCOM deviates more quickly than without step placement, but the resultant placement puts the swing foot lateral position at the XCOM after ground contact.

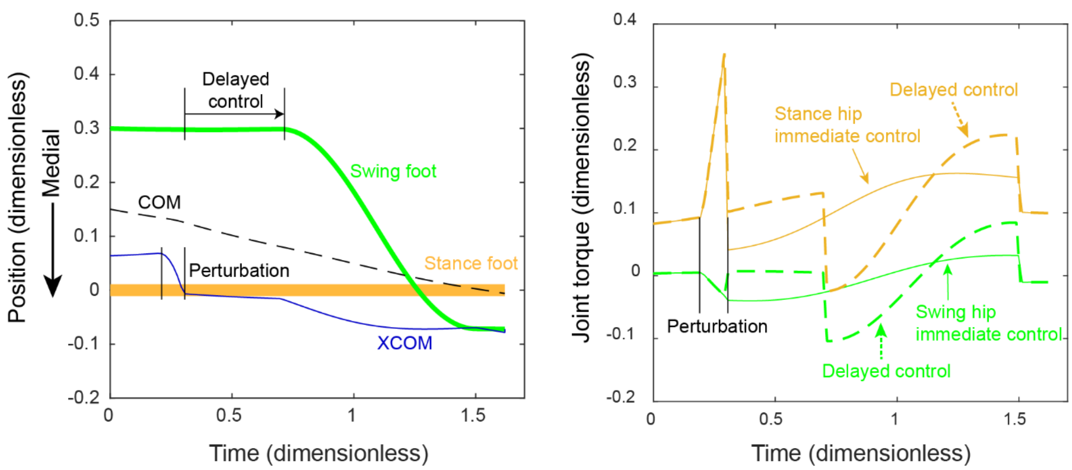

Figure 2.7 Step placement response for a stance-side perturbation accounting for avoiding contact between legs by delaying the start of step placement response. Onset of lateral movement is delayed from 0.3 until 0.7 dimensionless time to delay stance foot crossover from 1.1 to 1.25 dimensionless time into the step (left). The decreased time necessarily increases the torque required to perform the step placement (right), with greater torque to initiate and stop the crossover movement in both the stance hip and swing hip.

time during the second step, with the commanded step placement not sufficient by itself to produce a positive balance margin after collision.

Inertial strategies can augment foot placement. For example, a swing-leg abduction may be commanded (peaking at 0.75 dimensionless time) to induce an inertial reaction before the adduction needed for step placement (Fig. 2.8C). We can observe two major effects here. First, the resulting balance margin is still negative but the magnitude is smaller. Second, we can see the XCOM briefly moves positive towards the body before reversing back away from the body. The closest approach to the body appears to coincide with when we expect adduction torque to begin decelerating the abducting leg. Then as swing abduction reaches its maximum displacement, we see that XCOM has begun to move away from the body, approximately to the same lateral position as right after the perturbation, but still above where it would be in the absence of swing abduction (i.e. the same time point in 2.8B). Part of the reduced deviation of XCOM away from the body can be explained by the abduction movement shifting the mass of the leg, and thus the COM of the body, in the positive direction on the plot, which can be seen when comparing the COM position between 2.8C and 2.8B at the end of the abduction movement. Angular momentum can explain the rest of the behavior, with the abduction accelerating torque producing a stabilizing reaction force and thus an increasing XCOM, while the decelerating adduction torque generates a destabilizing reaction force and a decreasing XCOM. The final balance margin and this behavior suggests that swing abduction can help stabilize against perturbation, but the eventual need for adduction to fulfill step placement will eliminate a significant portion (though not all) of the stabilizing contribution.

Similar balance benefits can be seen when using active trunk roll (Fig. 2.8D). Here, the trunk is commanded to rotate towards the direction of the fall, reaching a maximum displacement at 0.75 dimensionless time, before returning to a vertical orientation by the end of the step, while the hip behavior performs only step placement as from 2.8B. At the end of the step, we see an increase in XCOM such that the balance margin is slightly positive. We can also see that the XCOM before hip adduction for step placement begins, while still sloping negatively, is at a higher position than for step placement without trunk roll. This can be attributed to the fact that rolling the trunk generates a similar angular momentum behavior as swing leg abduction, but it also significantly counteracted by the movement shifting the COM towards the direction of fall. The

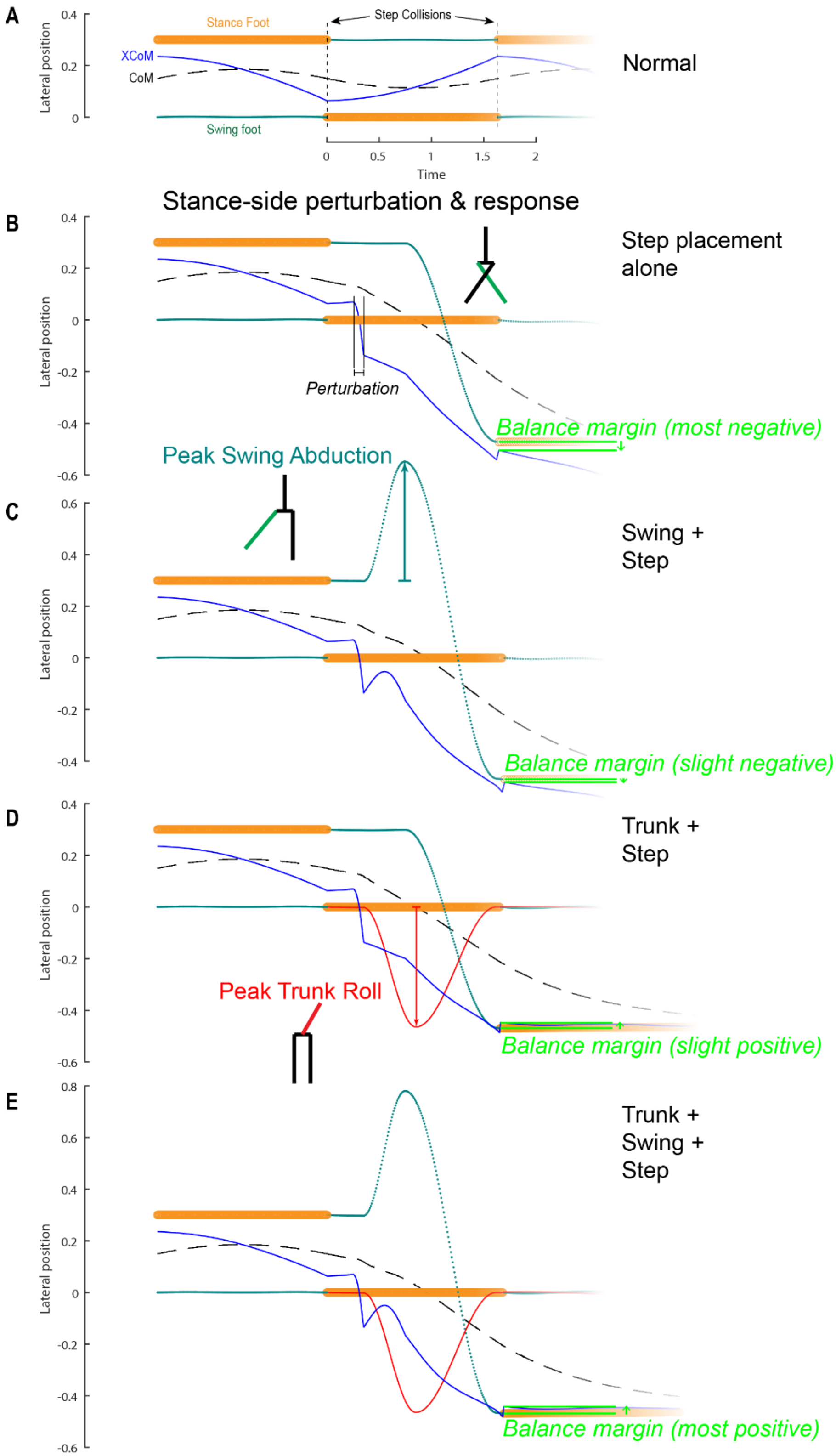

Figure 2.8 Model (A) response to a stance-side perturbation in one step, (B) with step placement alone; (C) with swing leg abduction and step placement; (D) with trunk roll and step placement; (E) with inertial strategies and step placement.

overall effect does still seem to be a decrease in the rate of falling by the time of step collision, however, as evidenced by the balance margin improvement.

Finally, it appears that using both inertial strategies roughly combines their effects (Fig. 2.8E). The final balance margin with both mechanisms is greater than that of either alone, and this seems to align with the XCOM trajectory appearing as a composite of the behaviors seen when the mechanism are used separately.

We then conducted a parameter study of varying magnitude of inertial balance and step placement mechanisms in combination to better understand the different effects on balance margin and step duration (Figure 2.9). In 2.9A, we see contours of equivalent balance margin over a range of peak trunk roll movements and final step placement position. Peak trunk roll increases from zero going up the vertical axis, while step placement position increases in extremity towards the right and is limited to a range near zero balance margin. The zero step margin contour is highlighted, with positive balance margins obtained above and to the right and negative balance margins below and to the left. This illustrates that for a given step placement, balance margin can be improved through greater peak trunk roll, and that for a desired balance margin, greater peak trunk roll can allow for less step placement.

These actions may be examined in combination. For example both mid-swing hip abduction and foot placement affect the balance margin (Figure 2.9C). Foot placement alone has a large effect on the balance margin, and swing hip abduction a much smaller effect. It is nonetheless helpful to add swing abduction to increase the balance margin or allow for a trade-off with less extreme step placement position while holding balance margin constant. The difference in the effects of trunk roll and swing leg abduction (compare slope of the contours between Figures 2.9C and 2.9A) suggests that, for similar angular rotations of the two body segments, increasing peak trunk roll is more effective at increasing balance margin than increasing peak hip abduction.

The comparative impact of the two mechanisms appears reversed when it comes to step duration (Figures 2.9B, 2.9D). Again, both trunk roll and swing abduction can affect step duration. In both cases, increasing peak displacement of the inertial mechanisms increased step duration, and such increases could trade off with decreasing step placement to obtain the same step duration. The comparative slopes, however, suggest that greater peak hip abduction is a more effective mechanism in increasing final step duration than greater peak trunk roll.

These results suggest a possible benefit from using inertial strategies in conjunction with step placement. Namely, because increasing inertial balance mechanisms can help achieve similar balance margins with less extreme step placement and help achieve increased step durations during which step placement can be performed, perhaps inertial strategies are used to reduce the rate of step placement movement, and the acceleration and thus force needed to produce that movement. In other words, inertial strategies may buy time for less rapid step placement and allow for less step placement to be performed.

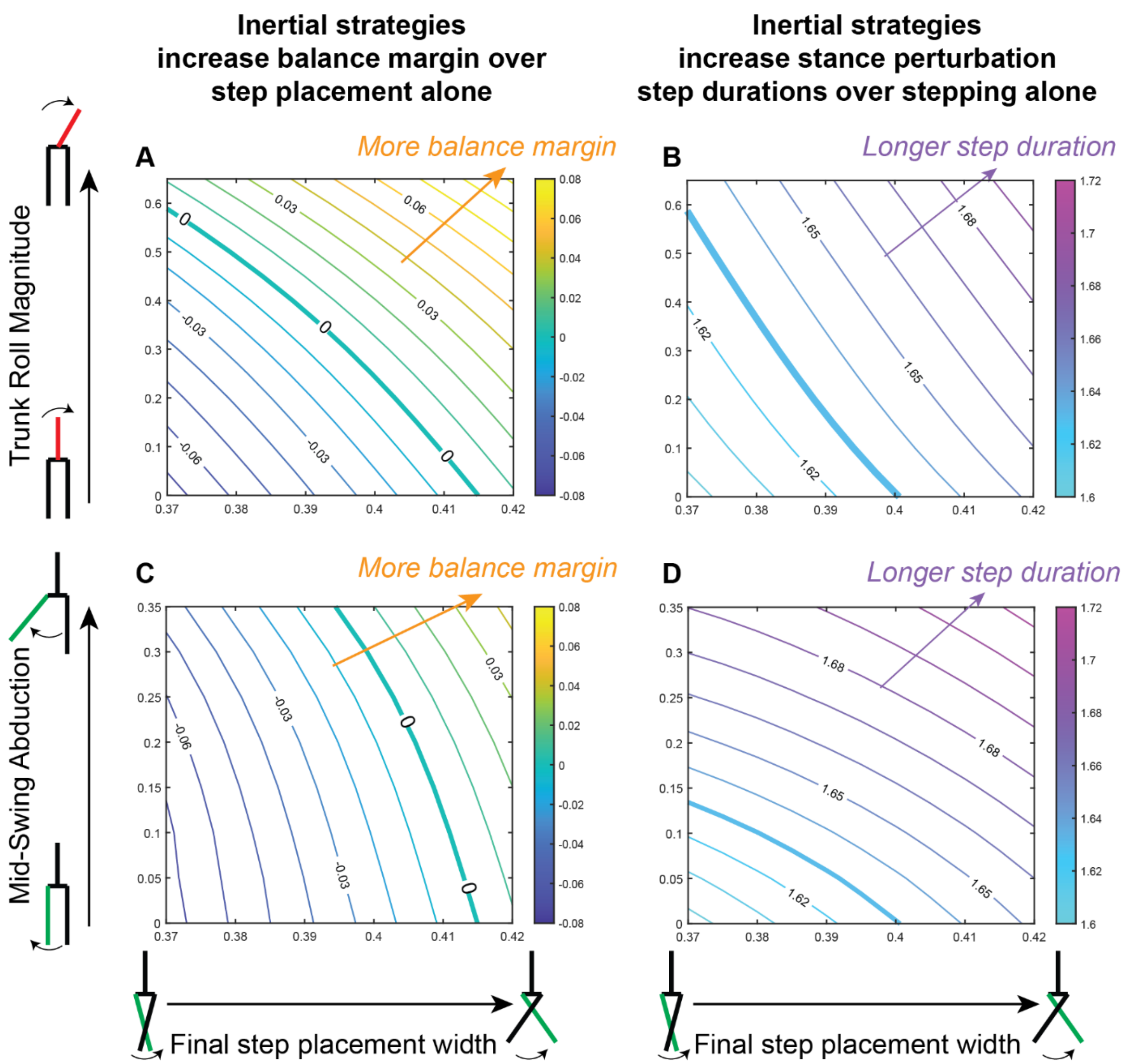

Figure 2.9 Interaction effects inertial strategies of trunk roll (top, vertical axis) and mid-swing abduction (bottom, vertical axis) and stepping strategy of final step placement width (all, horizontal axis) on contour lines of equal balance margin (left) and step duration (right), with zero balance margin and nominal step duration highlighted with thicker lines. Increased inertial strategy and step placement (towards the upper right) increased balance margin and step duration (across contour lines), and trade-offs could be made with increased inertial strategy reducing the step placement required (along contour lines).

2.4 Discussion and Conclusions

We sought to understand the potential contribution of inertial balance strategies to counteract lateral balance perturbations during walking and how various balance strategies could interact to effect balance. Our modeling suggests a number of balance strategies including inertial strategy, ankle strategy, and step placement can individually and in combination mitigate perturbations to balance. We made predictions about step parameters in lateral step placement and step duration as a result of such control. We also showed that inertial strategies such as trunk roll and swing abduction can contribute to reducing the severity of falling in conjunction with step placement, as well as increase the time available for step placement. The simulations suggest that while sufficient use of either mechanism can serve both purposes, increasing trunk roll may be more effective for increasing the balance margin.

Attaining a larger balance margin between foot position and XCOM isn't necessarily desirable, contrary to a possible simple interpretation of the margin. It takes more effort to move the body to more extreme configurations, and to return from them towards nominal. Furthermore, the XCOM can change based on the desired movement during the following step. For example, if the goal of the model were to move to a new lateral position after perturbation (i.e. continue walking forward while shifted to the right after being pulled to the right), then the XCOM desired after the step may be closer to or even exceed the foot position in anticipation of the following step. Actual human behavior may thus differ from this model, where we attempted to reject the lateral perturbation in one step based on the model's pose at the end of the step.

One observation is the discrete jump of the XCOM after collision, which we believe is an understated component of step placement highlighted here due to the impulsive nature of collision in our model. The ground reaction force from the collision, as well as the push-off force, changes the CoM velocity at the step transition instantaneously (Figure 2.10) as the model doesn't incorporate a double-support phase. However, we anticipate a similar but smoother behavior can be observed in humans, and thus illustrating that the COP does not have to always exceed the most lateral excursion of the XCOM to obtain a positive balance margin if collision and push-off are taken into account.

The simulated control strategies are intended to illustrate simple consequences for human actions. The complexity of the human allows for far more combinations of motions than explored

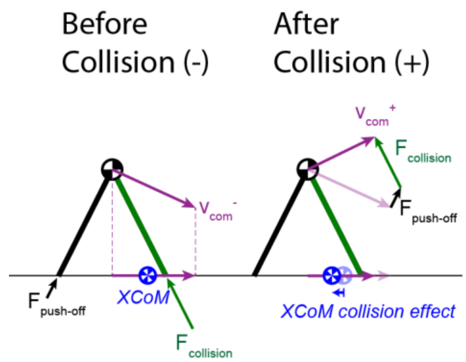

Figure 2.10 Collision and push-off are instantaneous impulses in the model and thus create an instantaneous change in XCOM.

here. For example, our model simplifies balance to attempt to reject the perturbation in one step, and to return trunk roll orientation to vertical at the end of the step. Humans may attempt to plan more steps ahead, and end one step in a transition pose that will be reset in a second step. The shape of the trajectories were also chosen for modeling convenience. Other factors may influence how humans shape trajectories, such as energy conservation, muscle strength, flexibility, and psychological effects. However, our overall model results seemed robust to at least two other trajectory variations: (1) a spline-based trajectory where the commanded angles were knots, and (2) a trajectory for the swing-leg that relied on collision to stop the lateral movement instead of controlling it to stop at a final angle.

This model illustrates how multiple balance strategies can be integrated and made operational with the dynamics of walking. I have shown that the hybrid dynamics can be accommodated with a combination of discrete-time balance command, and a continuous-time inner control that achieves the commanded movement. Thus, a trunk roll motion can be commanded by once-per-step feedback, and then the inner control fulfills that command in continuous time before the next step. This control scheme allows multiple strategies to be combined and work together, and makes predictions about how humans could control their balance in response to perturbations.

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Chapter 3 Inertial Balance Contributions to Lateral Balance During Perturbed Treadmill Walking

3.1 Introduction

Humans actively maintain their balance during walking through a number of compensatory actions. The most direct actions are to adjust the body's center of pressure (COP) location (Hof, 2007; Winter, 1995) relative to the body center of mass (COM). If the body is tipped in a particular direction, it is generally appropriate to move the COP in the tipping direction, for example by taking a step to move it a new discrete location (Kuo, 1999; Wang and Srinivasan, 2014), or by exerting ankle torque to move it to a different location underfoot (Hof et al., 2010). However, the body has many degrees of freedom, and all parts of the body could be employed to aid in balance. Although humans clearly step with active foot placement to maintain balance (Joshi and Srinivasan, 2019), they do not do so exclusively (Winter, 1995). This is most obvious in the case of a tight-rope walker, who has little choice with regard to medio-lateral COP but can nonetheless balance by moving the rest of the body. It is possible that humans also employ such motions even during normal walking on a flat surface, but to an unknown degree. Perhaps there is more to lateral balance during walking than stepping.

Balancing strategies have been described more fully in the literature for standing than for walking. One of the best understood approaches is to focus on sagittal plane balance, and to divide it into stepping, ankle, and hip strategies (Kuo and Zajac, 1993a; Nashner and McCollum, 1985; Nashner et al., 1989). Stepping refers to moving the foot to a new discrete location, and the ankle strategy to using continuous ankle torque against the rest of the body acting as an inverted pendulum. Both strategies can move the COP to ensure it is under the COM. In contrast, the hip strategy acts inertially (Kuo and Zajac, 1993b), using a combination of joint motions, say flexing the hips and (slightly) plantarflexing about the ankles to quickly move the COM. We use *inertial* to refer to actions that do not directly exert force or torque against the ground, but rather accelerate the COM rather than move the COP. Although the stepping and ankle strategies seem most evident, the higher frequencies of motion actually include considerable hip strategy (Kuo, 2005). It appears that normal balance involves some combination of all three strategies, along with other body articulations even if they are not as commonly measured in experiment.

In walking, active control of lateral balance appears to be more critical than fore-aft balance. Of particular interest is medio-lateral balance, which appears to be more unstable than in the fore-aft direction (Bauby and Kuo, 2000; Kuo, 1999; O'Connor and Kuo, 2009), and appears to account for most falls in older adults (Maki and McIlroy, 1996). One of the principal means of maintaining lateral balance is medio-lateral foot placement (Hof et al., 2010; Winter, 1995), which can be modulated as a function of body state to provide a discrete correction to COM motion once per step (Kuo, 1999). Evidence for such foot placement control includes its greater sensitivity to altered visual feedback in the medio-lateral than the fore-aft directions (Bauby and Kuo, 2000; Franz et al., 2015; O'Connor and Kuo, 2009). The location of foot placement has also been shown to depend systematically on body (particularly COM) state (Joshi and Srinivasan, 2019; Wang and Srinivasan, 2014). This discrete action is also accompanied by continuous inversion-eversion ankle torque, similar to the sagittal plane ankle strategy. Although stepping and ankle strategies are described well, they are also unlikely to be the sole strategies for walking.

Inertial compensations can also be performed for lateral balance during walking. Such actions have been illustrated by humanoid robots (Atkeson and Stephens, 2007), which actively use the trunk and arms to facilitate acrobatic maneuvers such as backflips (Guizzo, 2019). In the case of lateral balance, the example of the tight-rope walker suggests motion analogous to the postural hip strategy except employed in the frontal plane ("lateral hip strategy"). Here the upper body and even the arms and swing leg could potentially be actively used to accelerate the COM. Studies have described motions of the entire body (Winter, 1995), but few have directly found active use of inertial balance strategies, suggesting limited use during normal walking speeds and conditions.

It is possible that there are walking conditions where inertial balance strategies are more favorable or more evident. There may be particular parts of the gait cycle that are more challenging to control with foot placement, for example early in stance and long before the next footfall. Previous experiments have used no or relatively small perturbations, where inertial compensations may be small. Our modeling (Chapter 2) suggests several ways that rotation of body inertia could contribute to lateral balance control (Figure 3.1).

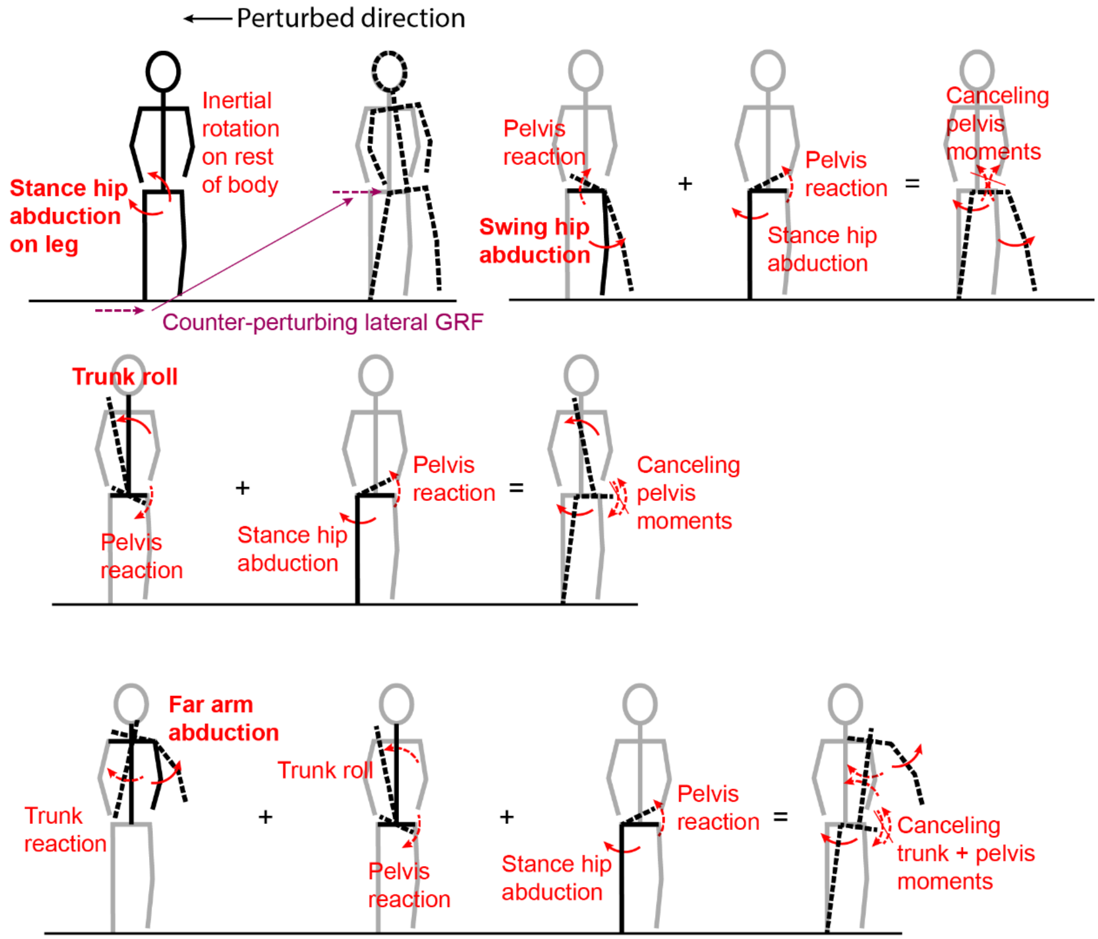

Figure 3.1 Examples of body inertia rotation to stabilize lateral balance after perturbation towards stance leg. Stance hip abduction (top left) can create a rotation of the rest of the body up and towards the perturbation direction, while pushing the stance leg out laterally towards the perturbation direction, creating a lateral GRF that pushes the COM away from the perturbation direction. Stance hip abduction could also be an indicator of several other body inertia rotations whose reactions may pass through the hip. The swing hip could be abducted (top right), the trunk could be rolled towards the perturbation (middle), or the arms could be rotated with the near arm in adduction and/or the far arm in abduction (shown, bottom).

Here, we endeavor to examine how humans control lateral balance after perturbations while walking on a treadmill. We use a custom perturbation system that can introduce relatively larger force disturbances to a walking individual, while still allowing arm swing to retain more normal walking characteristics. We will also examine various contributions of balance strategies throughout the gait cycle to understand possible phase-dependencies in balance response. Based on simple dynamical modeling (Chapter 2), we predict that in addition to medio-lateral step placement, there could also be a short burst of inertial balance strategy immediately after perturbations, and particularly after early stance perturbations. An early inertial response slows the rate of COM excursion, reducing the final magnitude of stepping response required.

3.1.1. Model Predictions

Perturbations are expected to cause changes in step width, duration, and length, as a function of perturbation timing. Such changes are expected as a result of the perturbed dynamics of walking, along with the active control compensations performed by the person.

We consider step width adjustments to be indicative of active step placement, where the lateral position of footfall is controlled. The goal is quite simple, in that the foot should be moved in the direction of the fall, so that the ensuing collision directs the body state toward nominal (Bauby and Kuo, 2000). However, this prescription is also affected by time, both too little and too much time. It takes time to move the foot, and so there may be insufficient motion if there is too little time before the foot contacts ground, namely when the perturbation occurs shortly before the swing foot's next ground contact. It also takes time for the body to fall, and too much time means there is ample opportunity for the fall to deviate from the nominal trajectory, and thus require more foot placement. The result is that lateral foot placement is expected to be greatest for perturbations early in a step, and close to zero for those occurring just before a footfall. Foot placement amplitude is also expected to be greater for stance-side perturbations, which are expected to extend step duration, and thus require more foot placement (Chapter 2, Figures 2.3 and 2.5).

The step duration effects should be relatively straightforward. The perturbation itself has a marked effect on step duration, because a swing-side perturbation will tend to roll the body toward that side, and thus cause earlier ground contact. Early swing-side perturbations are therefore expected to result in shorter step durations. This effect is diminished for perturbations later in the step, when the foot is already expected to make contact. On the other hand, stance-side perturbations should rotate the body away from swing leg contact, and thus extend step duration. (An exception is for larger perturbations, which may call for medial cross-over of the swing foot *and* speed up the fall in that direction.) In addition, active inertial compensations can potentially affect step duration, because a lateral hip strategy (rotating the trunk in the direction of the fall) will reduce the falling velocity and delay foot contact. Conversely, an opposing rotation (away from the fall) could be used to intentionally speed up foot contact.

Step lengths are also expected to be somewhat sensitive to perturbation. The effects should approximately match those for step duration, for example a swing-side perturbation should result in shorter step length because of the shorter step duration. Step length changes are expected to have relatively little effect on lateral balance, because there is relatively little coupling of fore-aft dynamics into lateral body states (Bauby and Kuo, 2000). However, longer or shorter step lengths are also expected to cause step durations to be longer or shorter, respectively, and thus affect the amount of lateral compensation necessary.

Perturbations also have effects that continue beyond one step. In some cases, the walking model is able to restore nominal gait through a single action such as lateral foot placement (ignoring a sideways shift in the walking track). But there are also clear cases where the restoration is incomplete. The most obvious such case is when a perturbation occurs immediately before a step, leaving too little time to actual perform the foot placement. In the case of a late stance-side perturbation, the appropriate compensation is a lateral foot placement of the following step, as with an early swing-side perturbation.

In addition to foot placement, inertial compensations can also contribute to balance. The lateral hip strategy can be performed immediately after perturbation, unlike foot placement which must wait until the next footfall. The principal body segment is expected to be the trunk, due to its relatively large mass. Inertial actions will be evidenced by changes in kinematic trunk angle, and in greater kinetic moment applied to rotate the trunk. There are additional degrees of freedom available from the pelvis and swing leg, and in particular it is possible to rotate the pelvis and trunk somewhat independently. We therefore expect moments applied to the stance hip to also contribute to inertial strategies, not necessarily identically to the trunk (Figure 3.1). Inertial strategies can only accelerate the COM to limited degree compared to stepping. It may nevertheless be advantageous to so to quickly reduce the rate of falling, and thus the amount of step placement eventually needed. Another factor for hip moment at the stance and swing hip is medial swing foot motion that should occur with stance-side perturbations and act opposite to a lateral hip strategy, except when obstructed by the stance leg. The stance hip moment is therefore expected to be greater in abduction (for lateral hip strategy) for perturbations early in stance-side, when stepping will take the longest to perform and the swing leg is obstructed. Trunk moment is expected to contribute throughout stance-side perturbations although priorities for keeping the trunk and head upright may inhibit usage when other mechanisms are available.

3.2 Methods

The experiment applied unexpected lateral perturbations to healthy human subjects while walking on a treadmill. There were eleven subjects (7 male, 4 female; ages 18 - 31) who participated in the experiment, after providing written informed consent according to University of Michigan Institutional Review Board procedures (HUM00020554). Some statistics for the subjects are listed in Table 1. Movement data was collected via active motion capture markers (PhaseSpace, San Leandro, CA, USA) on each participant's torso, pelvis, arms, and legs. Force data was collected from an instrumented split-belt treadmill (Bertec Corp, Columbus, OH, USA). Standard kinematics and inverse dynamics analysis was performed using models built in Visual3D (C-Motion, Germantown, MD, USA) and computed in MATLAB (MathWorks, Natick, MA, USA).

Table 3.1: Subject data. Leg length was measured from greater trochanter to ground.

SUBJECT	SEX	MASS (KG)	HEIGHT (M)	LEG LENGTH (M)	WALKING SPEED (M/S)
1	f	56.7	1.75	0.91	1.25
2	m	86.2	1.90	0.98	1.25
3	f	68.0	1.63	0.86	1.25
4	m	59.0	1.63	0.91	1.25
5	m	79.4	1.99	1.17	1.25
6	m	65.8	1.63	0.84	1.00
7	m	65.0	1.83	0.86	1.25
8	m	68.0	1.68	0.91	1.25
9	f	65.8	1.67	0.94	1.25
10	m	56.7	1.79	0.98	1.25
11	f	63.5	1.63	0.86	1.10
MEAN		66.73	1.74	0.93	
S.D.		9.05	0.13	0.09	

The perturbation system (Figure 3.2) was attached to the subject via rope cables extending to the left and right from a harness worn about the pelvis. The cables were each connected to a spool driven by an electric motor, which was controlled to regulate cable tension via a real-time feedback loop to a tension sensor along the cable's length. The harness was comprised of a waistbelt and a lightweight structure to allow the subject's arms to swing normally while lateral forces were exerted. The structure had two curved aluminum rods pointing fore-aft and passing to each side of the body, that were bent into a canoe-like shape, such that the subject was at the widest point and each pair of ends met in front of and behind the subject at a distance of approximately 60 cm. Each rope cable pulled laterally from each front and back joint to meet out at the subject's left and right sides, where they were attached to the pulling cables connected to the motors. The aluminum frame weighed approximately 3 kg and was stabilized from tilting via fabric suspenders worn over the participants shoulders. The frame's width was adjustable to the subject's waist width of each subject. The harness configuration allowed the motors to pull the participant laterally at the waist (approximately the body center of mass) while they walked with freely moving arms.

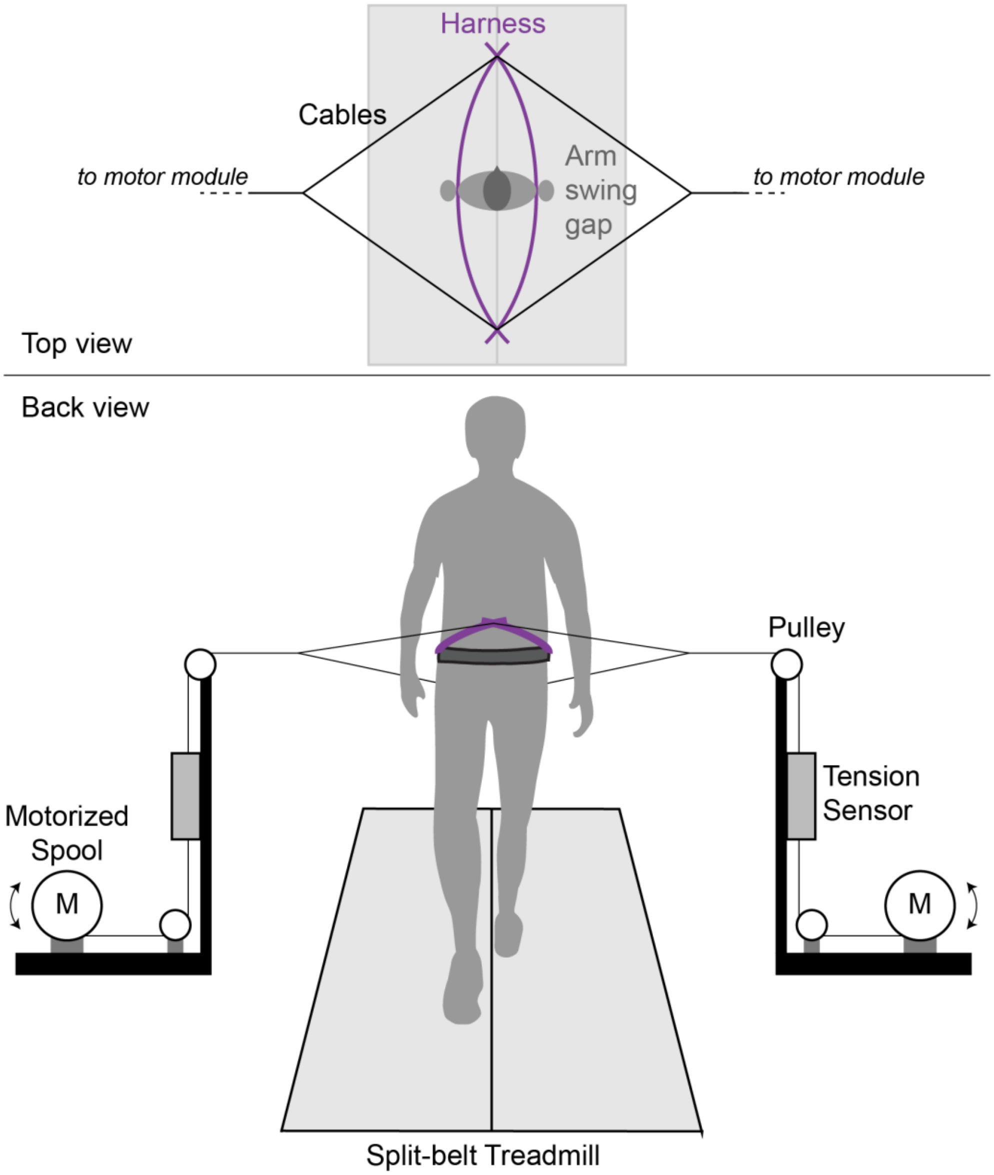

Figure 3.2 Schematic of the perturbation system. A human subject walks on an instrumented treadmill, with two lightly-tensioned cables pulling laterally and opposite each other at the waist. Force control ensures that the cables normally allow for natural body motions, and allows perturbations to be superimposed. A harness allows cables to apply force at the waist while still allowing for normal arm-swing. Perturbations were applied at random times during a stride, with a ramp profile with duration about 250 ms.

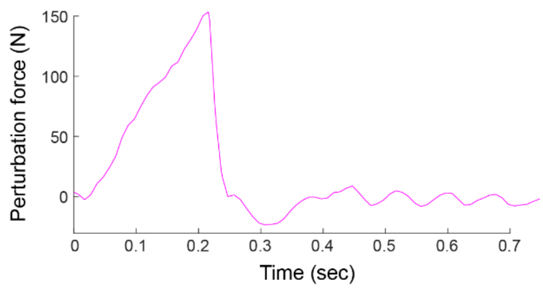

Figure 3.3: Example perturbation force measured at sensor. In this example, the perturbation consists of a ramp in force starting at 0 s and peaking at 150 N over 0.2 s.

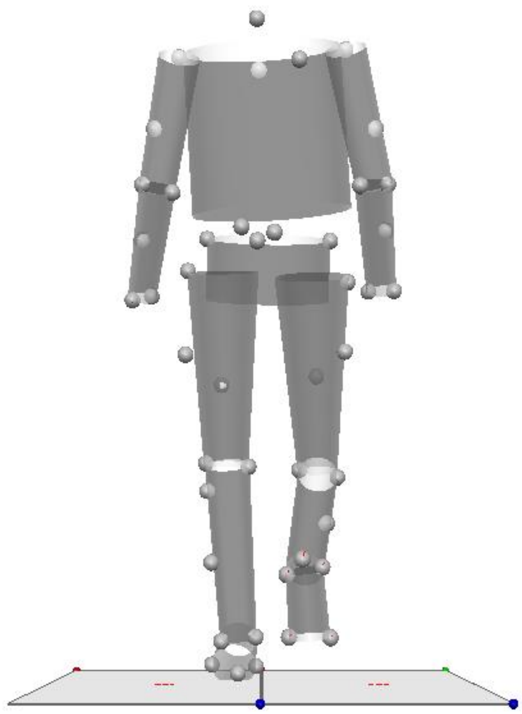

Figure 3.4 Example kinematic model built in Visual3D, also showing motion capture markers.

Perturbations were performed in a random lateral direction at randomly selected time points during a stride, and at three force levels. During unperturbed operation, both sides maintained approximately 20 N of tension to keep the cable taut and to balance the pull of the other side to cancel out effects on the participant. The perturbation force was generated by raising the commanded tension on the side of the desired pull in a sawtooth waveform (Figure 3.3), ramping linearly to a target low, medium, or high perturbation force (approximately 1.5 N, 2.25 N, and 3 N per kg of subject mass, or 105 N, 157.5 N, and 210 N for a 70 kg subject) over 0.25 seconds before dropping back to baseline tension (generating an approximate impulse of 13.125 kg m/s, 19.6875 kg m/s, and 26.25 kg m/s for a 70 kg subject, respectively). Perturbations were introduced randomly on either lateral side at one of 10 time points (5%, 15%, 25%, 35%, 45%, 55%, 65%, 75%, 85%, 95%) of average stride time. Stride time was measured between same-side heelstrikes on the instrumented treadmill via thresholding the real-time vertical force detection, with the average stride time computed as the mean of the last 10 heelstrike-to-heelstrike time intervals stored in a circular buffer. Stride time intervals were excluded from the buffer if fewer than 6 stride detections had passed since the last perturbation, as well as if the measured stride time exceeded 20% deviation from the mean of the times already in the buffer. A randomized wait of 8-14 strides occurred between perturbations. Perturbations were tested in groups of ten at the same force level while participants walked at 1.25 m/s (except subjects 6 and 11, see Table 1), and subjects were offered an opportunity to rest after each such bout. Bouts were conducted until there were two perturbations per side, per time point, per force level, for a total of 120 perturbations per subject.

We measured gait mechanics using standard procedures and computed a number of additional derived measures. Kinematics and kinetics were recorded with motion capture (PhaseSpace) using a standard lower-body marker set. In addition, torso markers were placed at the sternum, C7, shoulders, and mid-back, while upper arms and fore-arms markers were placed at the elbows, wrist, and mid-segment (Figure 3.4). Inverse dynamics (Visual3D and MATLAB) yielded lower body angles, moments, and powers. We measured frontal plane trunk angle and upper body moment estimates from the kinematic model. The trunk angle was calculated as the frontal plane angle about the vertical between the C7 marker and the center-of-mass (COM) of the pelvis model segment. The upper body moment was determined by computing top-down inverse dynamics from the forearms, upper arms and torso at the top end of the pelvis segment. Lateral step placement was computed as the change in lateral foot position (the average of the heel, first metatarsal, and fifth metatarsal marker positions on each foot), which was normalized to by dividing by mean unperturbed step width. Measures were set as positive in the direction that would produce a COM force to recover from the perturbation. For direct actions, this includes medio-lateral step placement towards the perturbation, and ankle inversion-eversion at the stance ankle away from the perturbation. For indirect inertial actions, this includes trunk rotations towards the perturbation, upper body moment (lateral hip strategy) to rotate the torso towards the perturbation, and hip abduction at the stance hip on the side of the perturbation. We quantified such forces and moments by their impulses, integrating the difference between the perturbed measure and the nominal unperturbed measure, over a brief duration of 10% of stride.

To evaluate balance after perturbations, we recorded the lateral position of the body center of mass (COM) and extrapolated center of mass (XCOM). COM position was calculated from masses and positions of the body segments generated from the Visual 3D motion capture model. The lateral XCOM position was calculated as defined by Hof (2008), as the sum of lateral COM position and COM lateral velocity divided by $\sqrt{g/L}$ (g gravitational acceleration, L leg length from ground to greater trochanter), as an estimate of the natural frequency of a linear inverted pendulum.

Measurements were normalized using the base units of body mass m (66.73 ± 9.05 kg), standing leg length L (0.93 ± 0.09 m), and gravitation acceleration g. Measures were computed for each subject, then normalized for averaging across subjects. We report the quantities as dimensionless measurements, as well as in dimensional units obtained by multiplying by the mean normalization factor. These mean factors were 3.2478 s^{-1} for $\sqrt{g/L}$ for frequency, 608.7978 Nm for MgL for moment, and 654.6 N for Mg for force. Time was reported as a fraction of stride time (between ipsilateral heelstrikes). Mean step lengths and durations were derived from halving respective stride parameters.

3.3 Results

The perturbation system produced lateral perturbations as observed through kinematic, force, and balance measures. We refer to the perturbations as swing-side or stance-side, based on whether the perturbation force was directed in the direction of the swing leg or stance leg, respectively. We treated strides as beginning with foot contact for a swing-side perturbation, followed by a stance-side perturbation. Thus, we ignore whether perturbations were to the left or right, and only consider the direction relative to the person’s gait phase. The swing-side perturbations therefore always occurred within 0 – 50% of a stride, and stance-side in 50 – 100% of stride. We also defined measures such as medio-lateral foot placement, ankle inversion-eversion moment, and hip abduction moment as positive if their deviations acted to restore the COM opposite to the perturbation direction. Each perturbation may be examined in terms of its deviation from the nominal, unperturbed case (Figure 3.5, 3.6). For example, a stance-side perturbation would cause the XCOM and COM to deviate towards the stance leg, and swing-side towards the swing leg. It would then be appropriate to place the next step more laterally for swing-side or medially/contralaterally for stance-side (positive position) to counteract the perturbation. That foot placement would also be expected to be accompanied by more positive ankle moment for the stance leg.

These initial actions were usually sufficient to counteract most perturbations, with smaller additional compensations after footfall and thus during the following step. A general observation was that subjects nearly always recovered the nearly nominal gait after the second footfall following perturbation, although sometimes displaced slightly in lateral position.

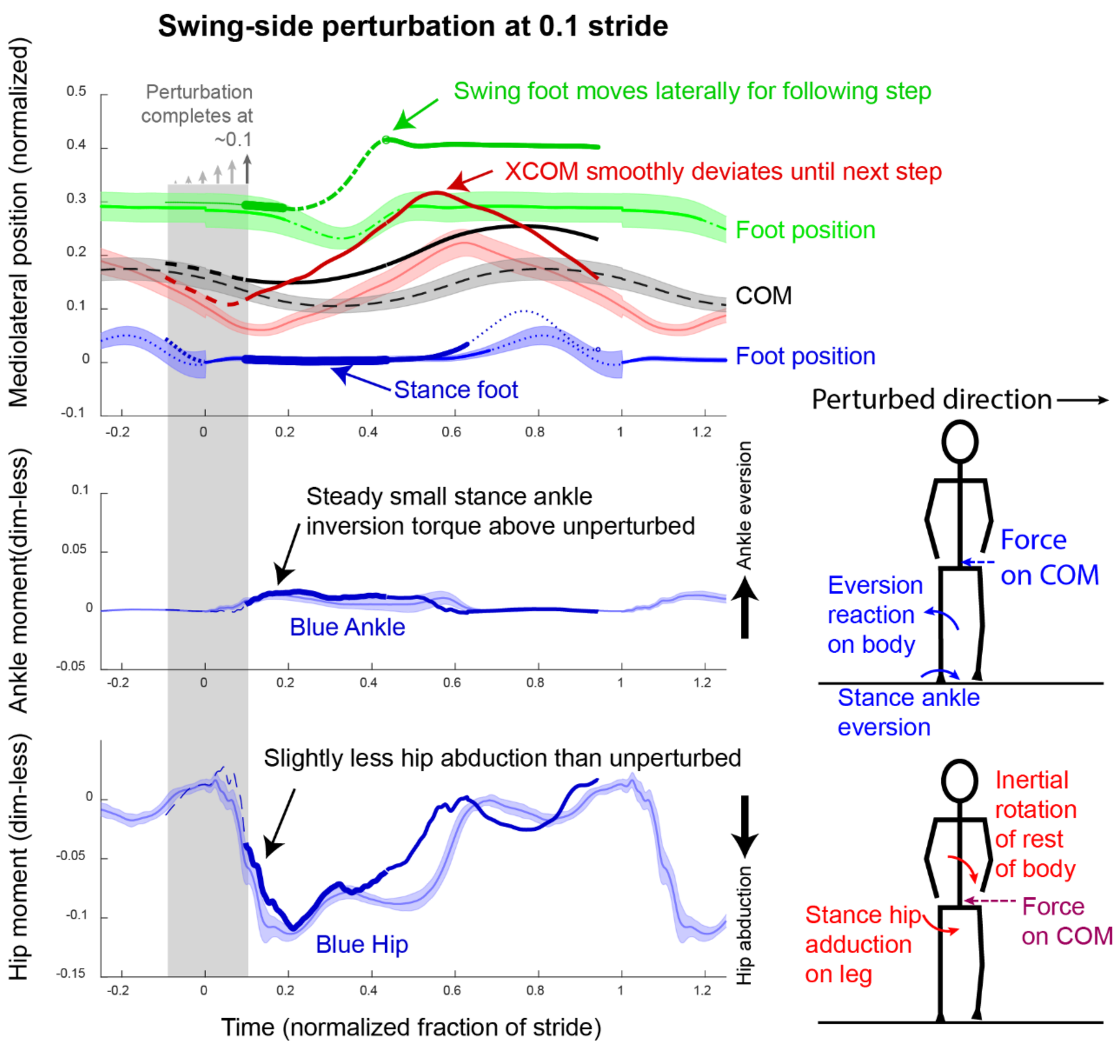

Figure 3.5: Representative subject swing-side perturbation response at 0.1 stride. Measures during the perturbation (thick dashed lines in shaded region) and after the perturbation (dark thick lines) are plotted against mean ± standard deviation band of nominal unperturbed walking to highlight deviations. (Top) Lateral positions of the XCOM and COM show immediate deviation towards the swing leg direction after perturbation, and swing foot position moved laterally after leaving the ground (dot-dashed region) before setting down for the next step. (Middle) The stance ankle produced a slightly higher-than-nominal eversion torque, which should exert a force on the COM opposing the swing-side perturbation. (Bottom) The stance hip produced slightly less abduction moment than nominal, although the inertial implications are unclear (ideally, an adduction torque on the stance hip would produce a force on the COM that pushes against a swing-side perturbation.

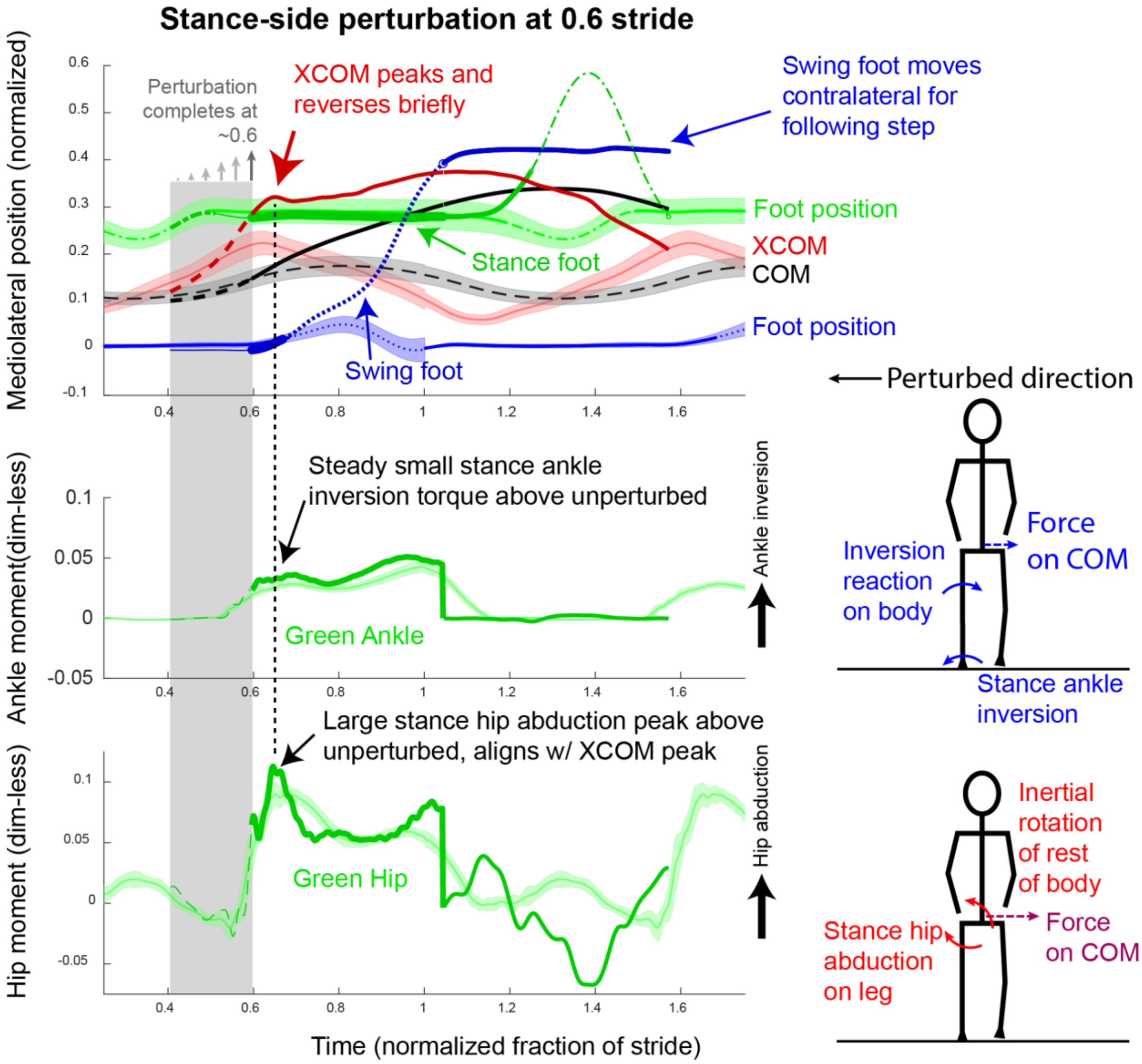

Figure 3.6: Representative subject stance-side perturbation response at 0.6 stride. Measures during the perturbation (thick dashed lines in shaded region) and after the perturbation (dark thick lines) are plotted against mean and +/- one standard deviation band of nominal unperturbed walking to highlight deviations. (Top) Lateral positions of the XCOM and COM show immediate deviation towards the stance leg direction after perturbation, and swing foot position moved medially after leaving the ground (dot-dashed region) before crossing past the stance foot contralaterally for the next step. (Middle) The stance ankle produced a slightly higher-than-nominal inversion torque, which should exert a force on the COM opposing the stance-side perturbation. (Bottom) The stance hip produced peak in abduction moment above nominal, which should push against body inertia and produce a force on the COM that opposes the stance-side perturbation.

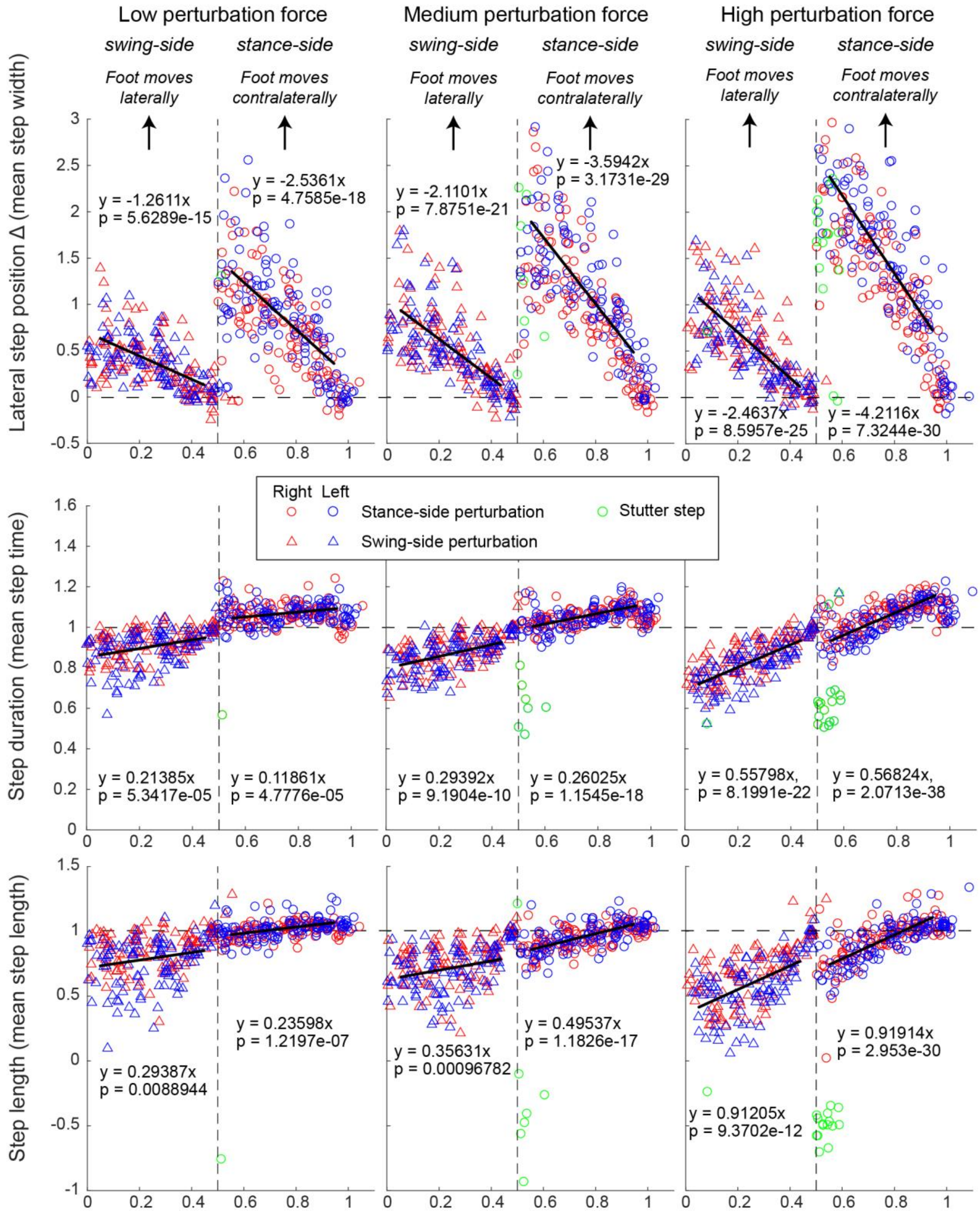

Figure 3.7 Perturbed step parameters after low (left col), medium (middle col), and high (right col) perturbations at varying times in the stride (horizontal axis). Swing foot lateral step placement (top row) moved laterally for swing-side perturbations (time 0-0.5, triangles) and contralaterally for stance-side (time 0.5-1, circles). Step durations reduced for swing and increased for stance (middle row). Step length (bottom row) shortened for swing. Higher perturbation forces increased deviations. All measures were normalized by mean unperturbed step parameters and referenced from nominal (dashed line). Green stutter steps are outliers explained later.

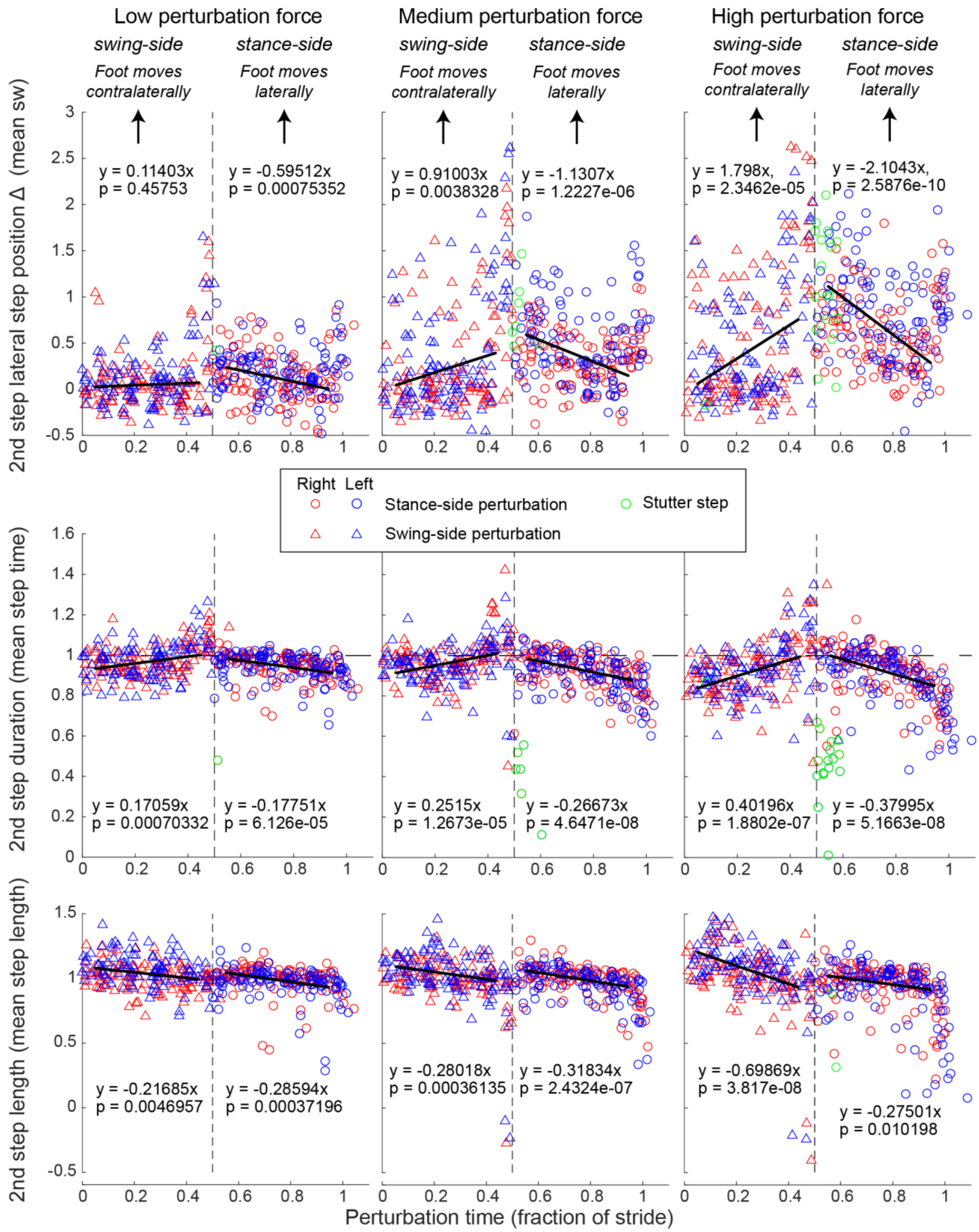

Figure 3.8 Parameters of step after low (left col), medium (middle col), and high (right col) force perturbations at varying times in the stride (horizontal axis). The second step's swing foot (top row) moves contralaterally to place the foot towards swing-side perturbations (time 0-0.5, triangles) and laterally after stance-side perturbations (time 0.5-1, circles). Of note, second step behaviors after late perturbations of each side behaves similarly to first step behaviors of early perturbations of the opposite side.

3.3.1 Stance-side and swing-side foot placement responses were distinct

There were clear distinctions between how subjects moved their swing foot in response to stance-side and swing-side perturbation. As expected from the lateral foot placement model, the swing foot was generally moved laterally in the direction of the perturbation force, but the amount of motion and duration of swing were unequal. The most straightforward case was swing-side perturbations, which were followed by lateral foot placement in slightly shorter than nominal unperturbed time. Stance-side perturbations were also followed by foot placement toward the perturbation direction, and thus moved medially toward the stance leg and in some cases crossed past the stance foot (Figure 3.6). The amount of movement and timing were different, however, with an average of more movement and longer than nominal step times (Table 3.2).

Table 3.2: Perturbed step parameters after swing- and stance-side perturbations, for medio-lateral swing foot displacement and step duration. Swing foot displacement is defined as positive in the direction of perturbation force a fraction of nominal step width, reported as mean ± s.d. Step duration is defined as a fraction of nominal step duration. P-values shows significant difference in response between the two perturbation sides.

Perturbation Level	Medio-lateral displacement					Step duration				
	Swing-side		Stance-side			Swing-side		Stance-side		
	Mean	SD	Mean	SD	p-value	Mean	SD	Mean	SD	p-value
Low	0.35	0.28	0.78	0.54	1.08E-22	0.91	0.08	1.07	0.05	6.36E-78
Medium	0.50	0.41	1.03	0.67	2.89E-21	0.88	0.08	1.05	0.05	1.08E-91
High	0.55	0.42	1.37	0.80	3.80E-33	0.84	0.10	1.05	0.08	1.84E-78

These responses also changed systematically with the timing of the perturbation. As swing-side perturbations occurred later in the step, the lateral step placements decreased and the durations increased, both approaching nominal unperturbed values (Figure 3.7). The lateral adjustments decreased approximately linearly with perturbation time with coefficient -2.1101 step widths/stride (-2.4993 – -1.7209 c.i., 95% confidence interval; p=7.8751e-21, linear regression from 0.05 to 0.45 stride) for medium force perturbations, for example. The durations increased approximately linearly with perturbation time with coefficient 0.29392 mean step times/stride

(0.2044 – 0.3834 c.i., 95% confidence interval; p= 9.1904e-10, linear regression from 0.05 to 0.45 stride) at medium perturbations.

Even though responses to stance-side perturbations were also in the direction of the perturbation, they were quite different in their dependence on perturbation timing. There was a larger step placement that decreased more sharply with later timing, and a duration closer to nominal and increasing less with later timing. At medium force perturbations, the coefficient for stance-side foot placement was -3.5942 mean step times/stride (-4.1138 – -3.0745 c.i., 95% confidence interval; p= 3.1731e-29, linear regression from 0.55 to 0.95 stride) and the coefficient for stance-side duration was 0.26025 mean step times/stride (0.2085 – 3120 c.i., 95% confidence interval; p=1.1545e-18, linear regression from 0.55 to 0.95 stride).

Perturbations of one side late in a step resembled an early-step perturbation of the opposite side in the following step. This can be seen in the step placement and step duration behaviors of the second step (Figure 3.8). Thus, there is stride-level continuity to perturbation response types despite the discrete step-level behaviors outlined thus far, and any references to perturbations throughout the stride should be interpreted as meaning both stance- and swing-side perturbations.

There is also an outlier cluster of short duration steps in early stance-side perturbations, most prominently at high perturbation force. These "stutter" steps will be examined later and are excluded from statistical analyses unless otherwise noted.

3.3.2 Extrapolated COM shows early reaction against perturbation across most perturbation times

As expected, the perturbations caused a disturbance in the XCOM trajectory (Figure 3.9). The XCOM trajectory was disturbed in the direction of the perturbation, indicating a need for compensation. There was also a brief early reduction or reversal in deviation within approximately 0.1 stride durations following the perturbation, across many perturbation times and regardless of the timing to the next step. This suggests that non-stepping compensations, including ankle strategy and possible inertial compensation, may be particularly active in the period immediately after perturbation. The compensations reduce deviation of the XCOM after perturbation throughout the stride before the next step.

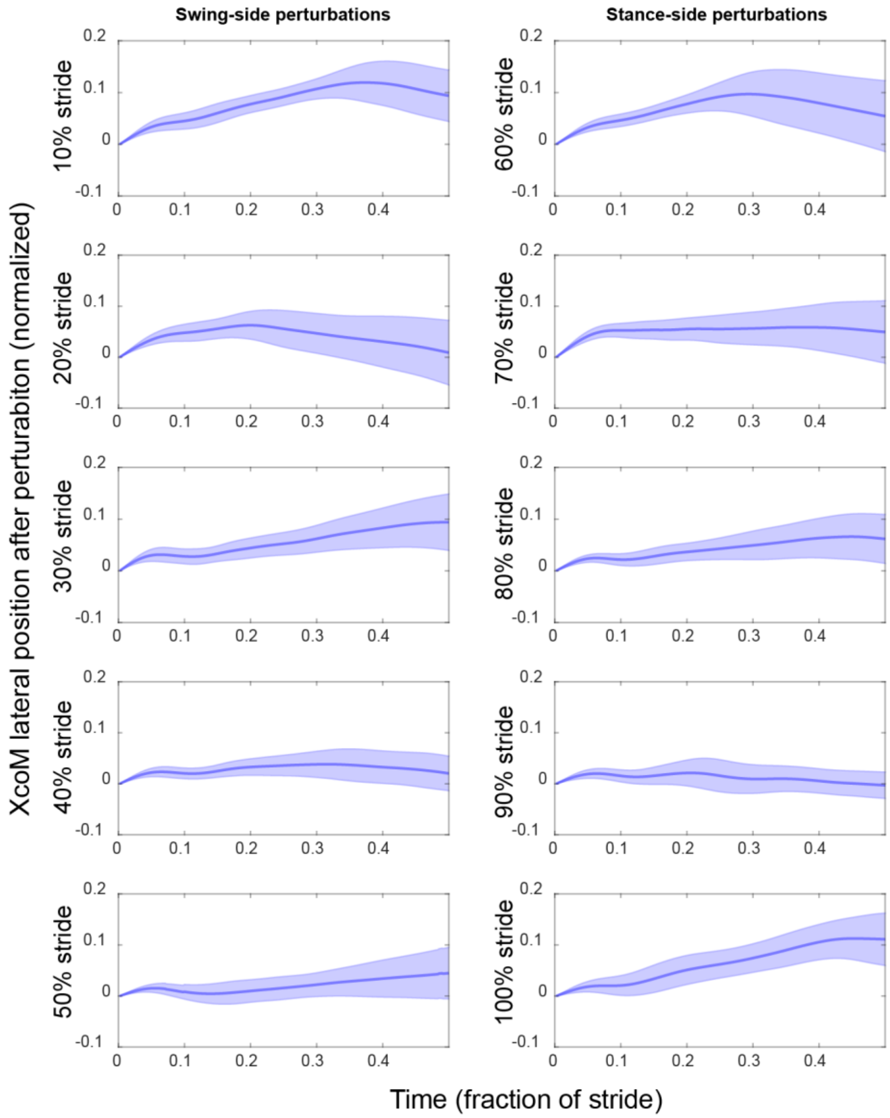

Figure 3.9 Mean and standard deviation of XCOM immediately after perturbations for medium perturbations throughout stride.

3.3.3 Frontal plane ankle moment contributes immediately mid-step

We observed inversion-eversion ankle torques consistent with a lateral ankle strategy, but in relatively small magnitude (Figures 3.6 and 3.10). The COP was moved in the direction of the perturbation, so that ankle torque was modulated appropriately to reduce the falling of the inverted pendulum stance leg, for perturbations throughout the stride. In particular, we were interested in possible ankle-strategy contributions in the 0.1 strides after perturbations as a possible factor to immediately reducing XCOM deviations. We found that ankle torques during these periods were higher during the middle of the step for both swing-side and stance-side perturbations, and lower during early and late step perturbations (Figure 3.10, 3.11). Stance-side perturbations also tended to elicit about double the response of swing-side across all force levels (*Low:* p = 0.0358; *Medium:* p = 0.00002; *High:* p = 0.0045). This suggests that the lateral ankle strategy may be a significant contributor to modulate XCOM velocity after mid-step perturbations, but that other contributions may be significant after early and late step perturbations.

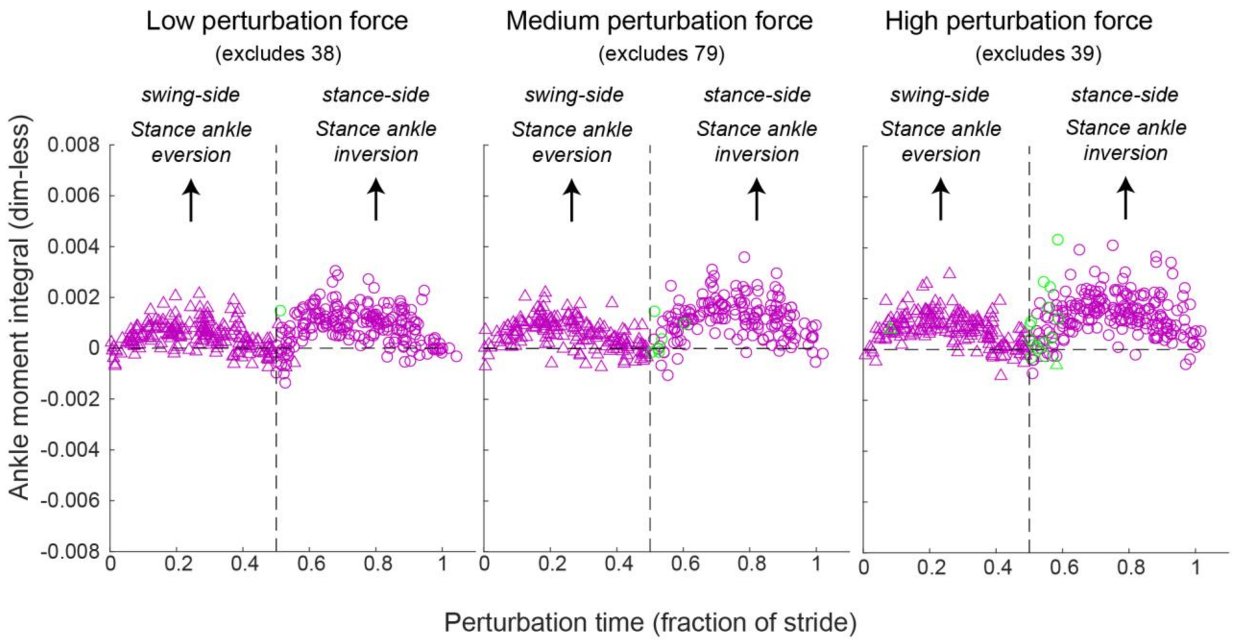

Figure 3.10 Integral of ankle moment for 0.1 strides after perturbation. Values are positive throughout much of swing and stance. Some trials were excluded due to subjects stepping on the same belt with both feet during the step, leading to unresolable forces with inverse dynamics (38 for low force trials, 79 for medium force trials, 39 for high force trials).

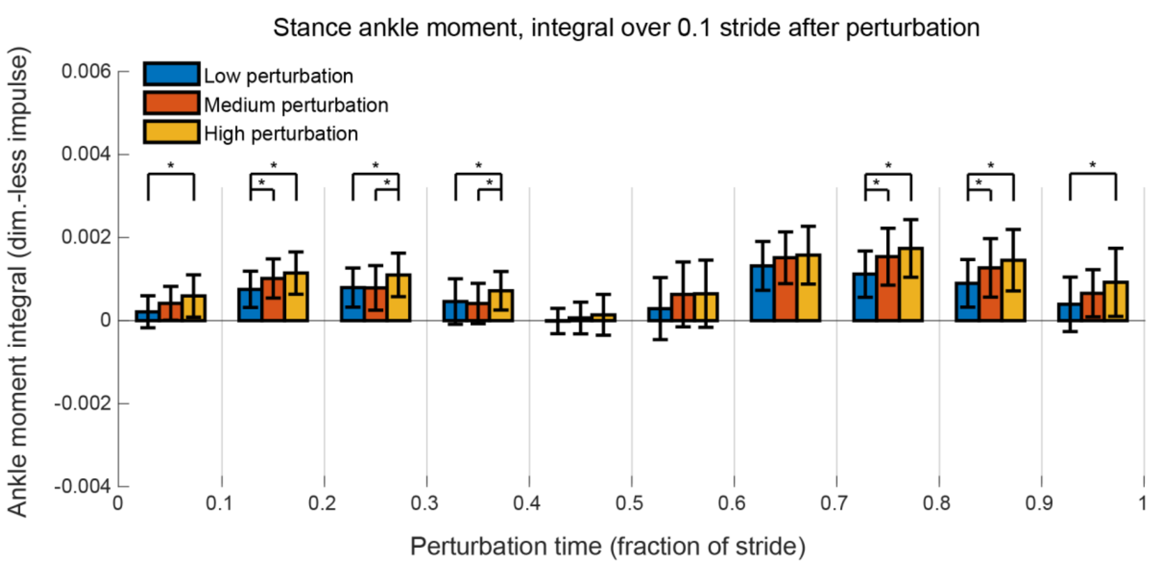

Figure 3.11 Mean and standard deviations at every 0.1 periods of stride of stance ankle moment integral from Figure 3.10. Mean impulse are all significantly different from 0 with α = 0.05 except between 0.4 to 0.5 stride. Stars (*) denote significant difference between perturbation force levels at α = 0.05.

3.3.4 Hip moment suggests inertial balance strategy in early stance-side/late swing-side

The hip moment on the stance leg was significantly higher than nominal immediately after late swing-side and early stance-side perturbations (Figures 3.12 and 3.13). The timing of the change in magnitude seemed to correspond well with time-derivative changes in the XCOM, specifically the early reversal in the first 0.1 strides (Figure 3.6 bottom). Based on the inertial hip strategy, positive hip moment means a combination of the pelvis, upper body, and soon-to-be-swing leg are rotated up and towards the perturbation direction. The reaction forces are such that the stance leg is being pushed against the ground towards the perturbed direction, and thus the COM away from the perturbed direction. This would cause the XCOM to be pulled negatively toward its nominal trajectory.

The hip abduction moment's impulse after the perturbation was significantly positive beyond nominal after stance-side perturbations during 50-60% of the stride (mean +/- sd: *Low* 0.0026+/-0.0017, $p=2.6*10^{-12}$; *Medium* 0.0028+/-0.0018, $p = 8.0*10^{-9}$; *High* 0.0028+/-0.0017, $p=5.1*10^{-10}$) and late swing-side perturbations during 40-50% of the stride (mean +/- sd: *Low* 0.0018+/-0.0016, $p=4.2*10^{-7}$; *Medium* 0.0026+/-0.0015, $p = 9.8*10^{-12}$; *High* 0.0027+/-0.0012, $p=2.1*10^{-14}$). There was also a smaller but significantly positive deviation after early swing perturbations during 0-10% stride (mean +/- sd: *Low* 0.0011+/-0.0014, $p=5.8*10^{-5}$; *Medium* 0.0006+/-0.0014, p = 0.0467; *High* 0.0009+/-0.0012, p=0.0015). This was computed excluding trials where there were cross-over steps that precluded accurate inverse dynamics for the individual legs. The impulse becomes negative after perturbations towards the middle of step. This suggests two phases in the stance hip response. After early and late step perturbations, there is little ability to adjust foot placement as the foot needs to take an entire step duration to contribute (early) or there is too little time to move foot placement effectively (late), and thus using the stance hip for inertial strategy may be more advantageous for balance. Meanwhile after mid-step perturbations, the swing leg is in motion and has time to move laterally, and thus the stance hip may contribute to assisting step placement: the stance hip should produce a moment opposite to inertial strategy to facility movement of the swing leg for step placement.

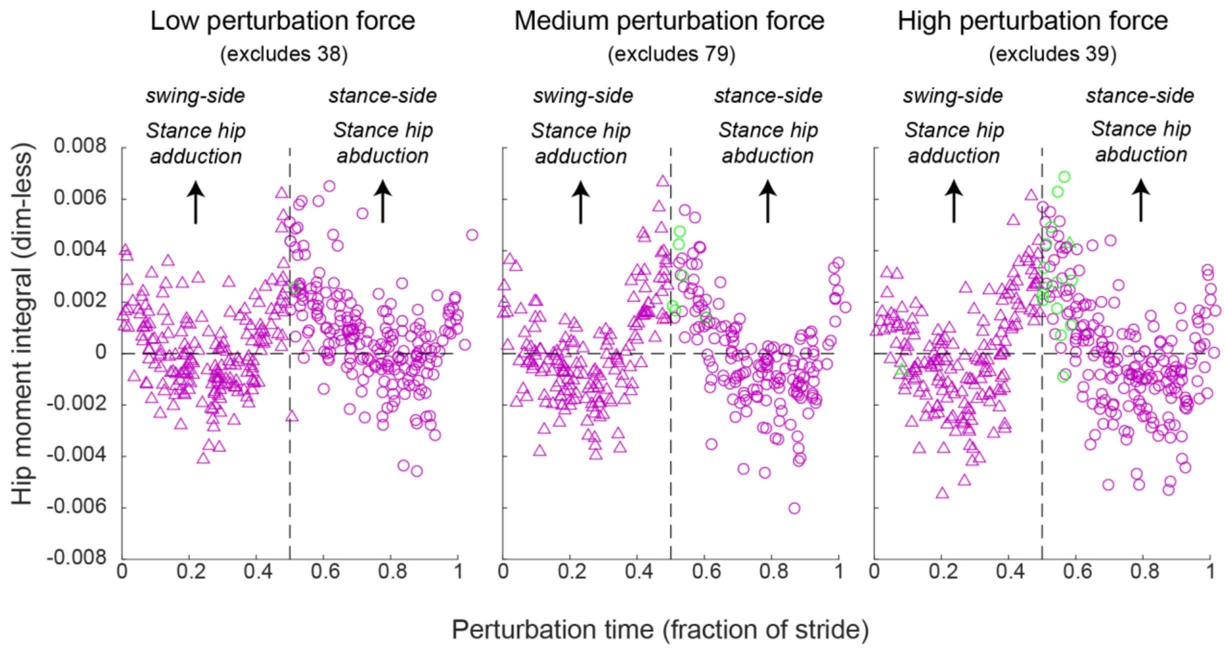

Figure 3.12 Stance hip moment integral (impulse) over 0.1 strides after the perturbation, at low (left), medium (middle), and high (right) perturbation force, for varying perturbation times during the stride (horizontal axis).

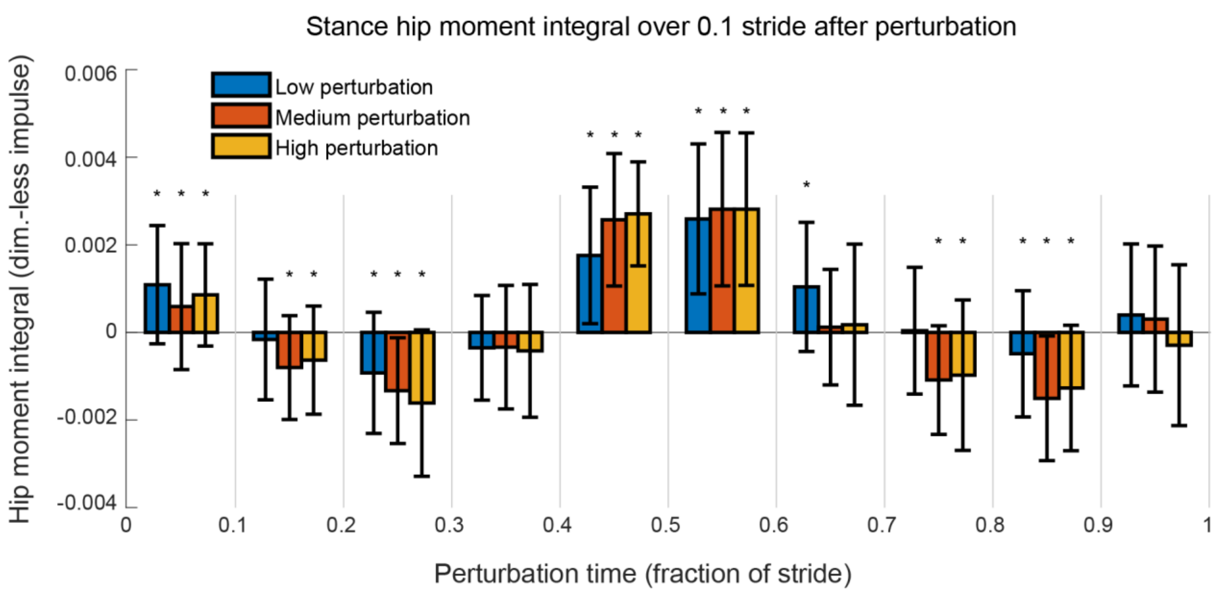

Figure 3.13 Mean and standard deviations at every 0.1 periods of stride of stance hip moment integral from Figure 3.11. Stars (*) denote significant difference from 0 at α = 0.05.

3.3.5 Mid-to-late step perturbations caused increases in trunk angular acceleration and upper body moment suggesting inertial balance use

We also quantified other indicators of inertial balance strategies, in the form of upper body moment and trunk angular acceleration. Although stance hip abduction suggests that substantial body rotation was performed, it does not indicate which body segments actually moved. Quantification of trunk angular acceleration shows good correlation with the upper body moment, which combines contributions of the torso and arms (Figure 3.14). The moment impulse showed

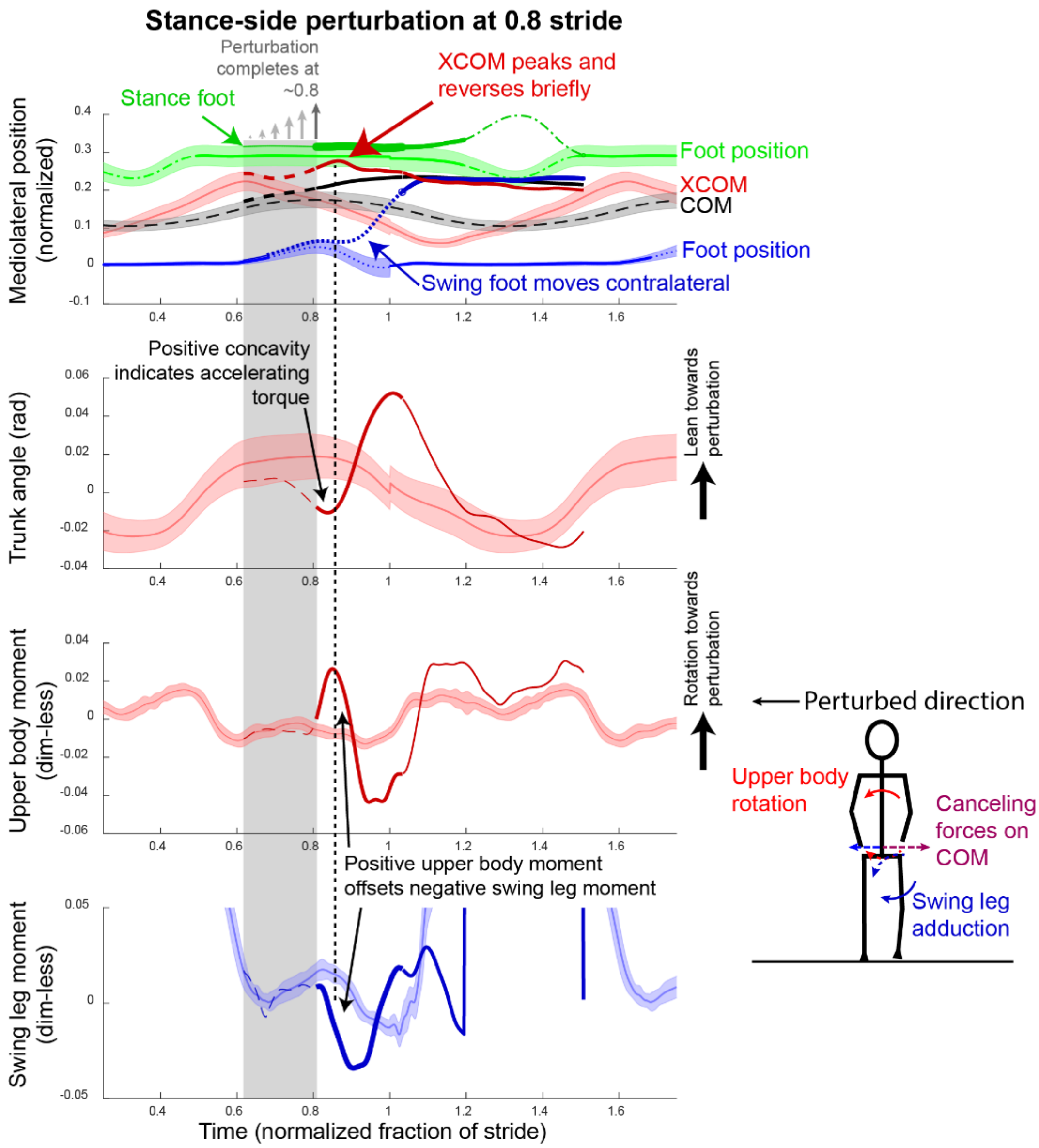

Figure 3.14 Example late stance-side perturbation. Trunk angular acceleration and upper body moment are well aligned with the reversal in XCOM. One further explanation is that the upper body moment may help offset negative inertial contributions from swing leg movement for step placement

positive contribution to inertial balance throughout the stride (Figure 3.15, 3.16) in the 0.1 strides after perturbation. This suggests that some of the positive hip abduction moment is carrying the upper body rotational moment through to the ground, and thus that the upper body moment is contributing to opposing XCOM deviation during that period.

Upper body moment also increased throughout the step (Figure 3.15). At medium perturbation force, the upper body impulse increased approximately linearly over 5 – 45% time (swing-side), with coefficient 0.0012 (0.0025 to 0.0040 95% c.i.; p = 0.01316, linear regression), and similarly for 55 – 95% time (stance side), with coefficient 0.0029 (0.0005 to 0.0023 95% c.i.; p = 7.017e-7, linear regression). Interestingly, the increase in upper body moment during the middle of the step does not correspond to an increase in hip abduction moment, which we would expect if the reaction were carried through to the ground. This would suggest that the upper body moment after mid-step perturbations is not contributing directly to reducing XCOM deviation, but may be useful for something else (see Discussion).

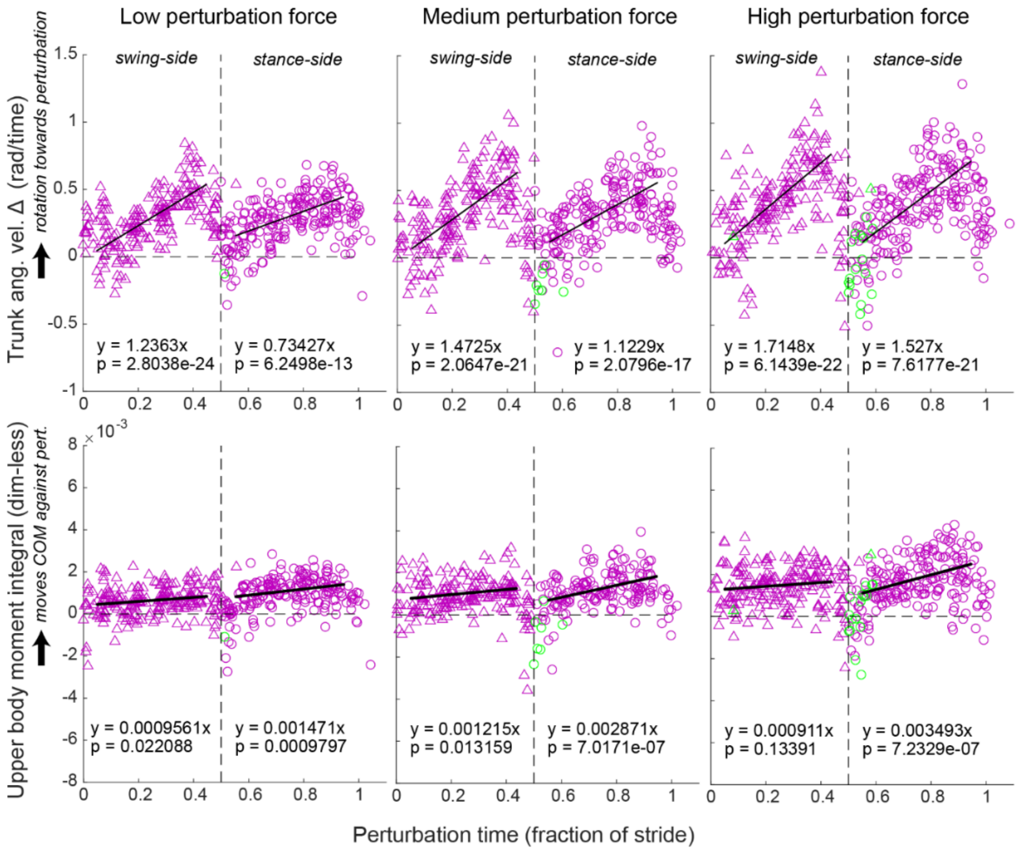

Figure 3.15 Trunk angular velocity change towards perturbation direction (top) and integral of upper body moment (bottom) over 0.1 stride after perturbation, versus perturbation time. Both measures show an increase over time towards late step on both stance and swing side, supporting the notion that the upper body moment is measuring actual inertial rotation.

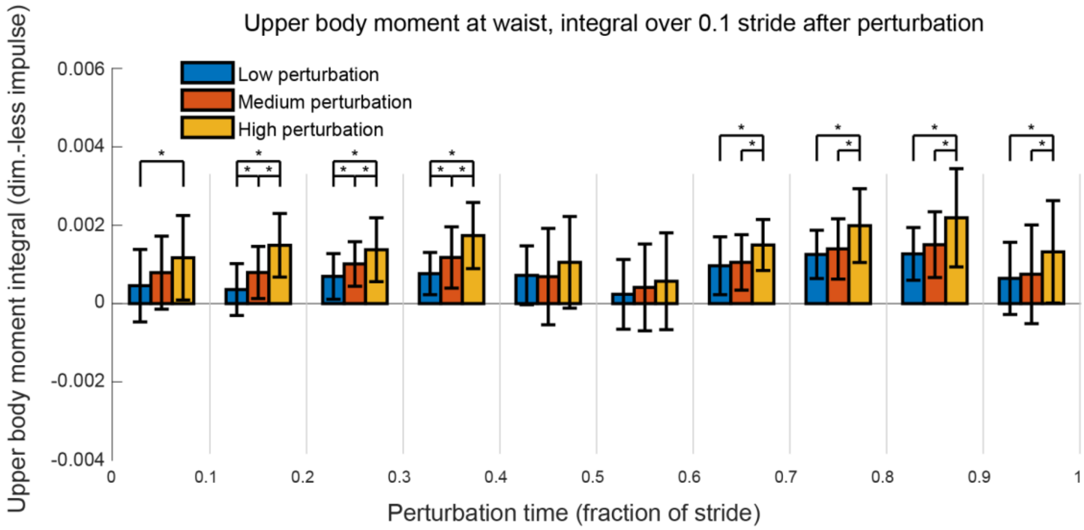

Figure 3.16 Mean and standard deviations at every 0.1 periods of stride of upper body moment integral from Figure 3.13. Mean impulse are all significantly different from 0 with α = 0.05 except low force perturbations at 0.5-0.6, with p = 0.0821. Stars (*) denote significant difference between perturbation force levels at α = 0.05.

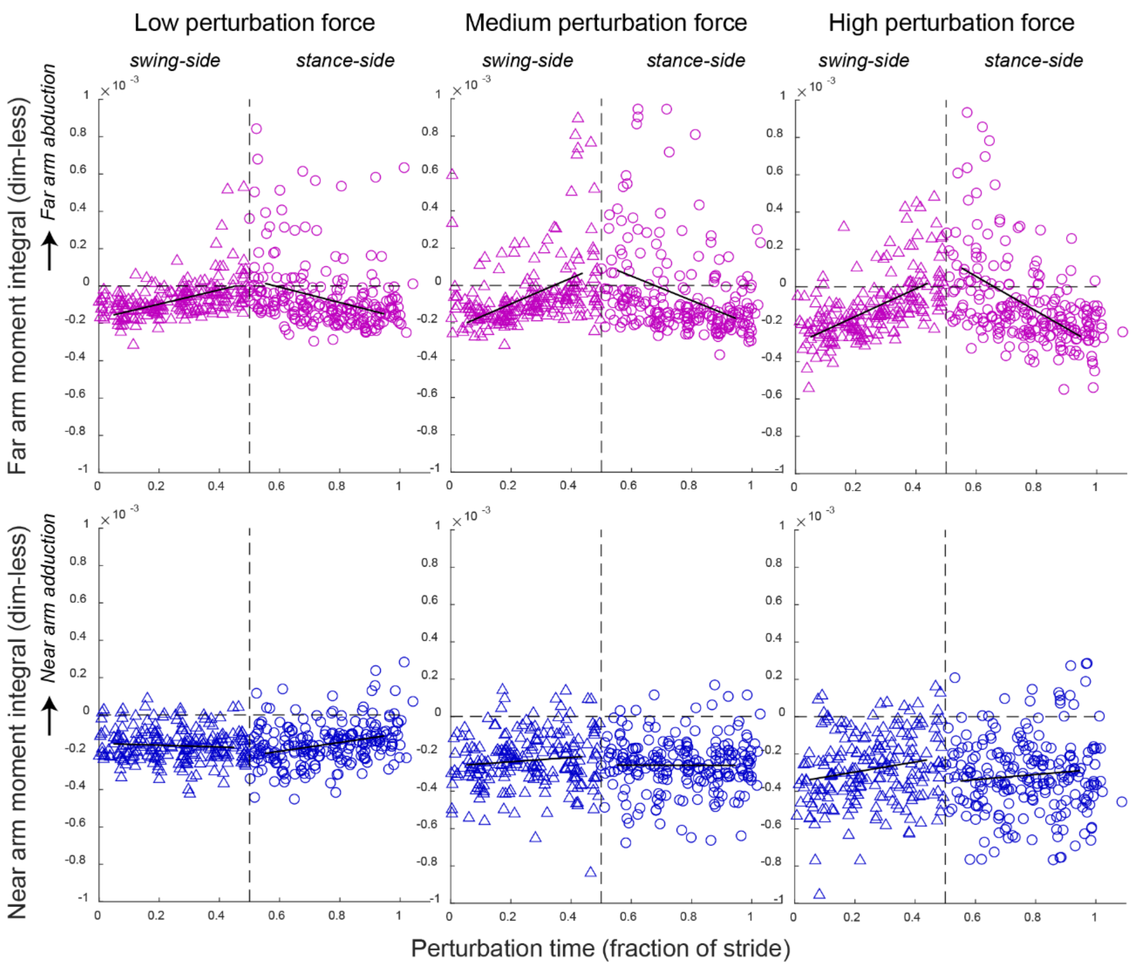

Figure 3.17 Shoulder moment impulse of the far arm opposite from the perturbation direction (top) and near arm one the side of perturbation direction (bottom).

3.3.6 Arm away from perturbation may move for limited inertial strategy contribution

We estimated the inertial contribution of arm movement in response to perturbations. Although the arms sometimes exhibited substantial kinematic motion, the resulting inertial moments about the shoulder immediately after perturbations were quite small and were generally negative in terms of contributing to inertial balance strategy (Figure 3.17). However, there was high variability, and in particular, the far arm away from the perturbation direction was sometimes moved in abduction towards late swing-side/early stance-side perturbations, in line with inertial strategy. This was most statistically significant at high perturbation force levels, where a linear regression of far arm moment showed an increase from 5% to 45% time with coefficient 0.00074 (0.00056 to 0.00092 95% c.i., $p = 1.886*10^{-13}$) and a decrease from 55% to 95% time with coefficient -0.00093 (-0.00121 to -0.00065 95% c.i., $p = 6.681*10^{-10}$). That being the case, any positive inertial contribution from the far arm would at best offset the overall negative contribution from the near arm (*High:* mean +/- sd -0.0002978 +/- 0.0001955, $p=8.6*10^{-115}$). A possible explanation is that the arm moment contribution is normally quite small due to the comparative inertia of the arms versus the trunk, and has even less effect on the COM velocity due to the 1.5 times increased distance from the ground of the shoulders compared to the pelvis where the trunk and hip moments approximately act (Winter, 2009), such that the arms might normally be moved towards the perturbation opposite of inertial strategy, perhaps to brace for a potential fall. However, when balance response is most challenged at high force perturbations and at late swing/early stance when step placement is least immediately effective, it becomes beneficial to mitigate even this small negative inertial contribution by moving the far arm in accordance with inertial strategy.

3.3.7 Swing leg doesn’t contribute to inertial strategy during swing phase

The moment at the swing hip during swing phase is consistently opposite of the direction desired for inertial strategy. For swing-side perturbations, the swing hip moment tends towards abduction, while for stance-side perturbations, the swing hip moment tends towards adduction (Figure 3.18). This is in line with using the swing leg primarily for foot placement, as an abduction moment during swing-side perturbations moves the swing leg laterally, and an adduction moment during stance-side perturbations moves the swing leg medially, which is the foot placement behavior we saw previously (Figure 3.7). The moment also increases in magnitude as perturbation force increases (Table 3.3), again in line with greater lateral foot placement behavior.

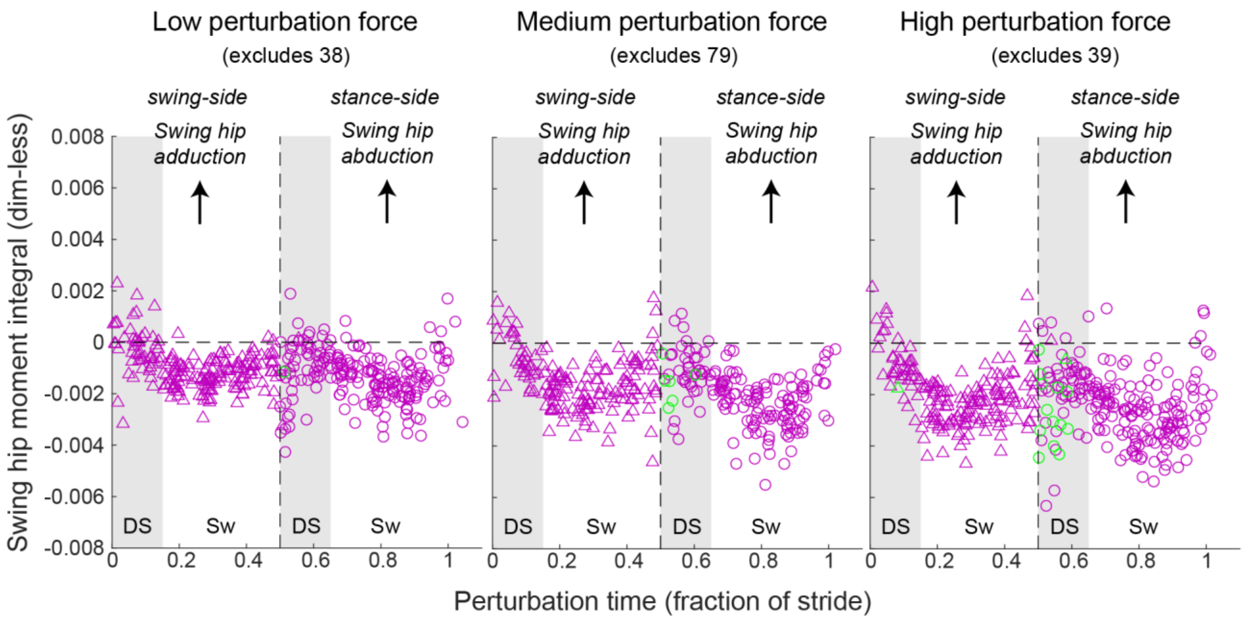

Figure 3.18 Swing hip moment impulse after perturbation for low (left), medium (middle), and high (right) force perturbations. Shaded double support phase (DS) highlights approximately when the swing foot is still on the ground and thus doesn't necessarily desire to move according to inertial balance strategy, while swing phase (Sw) highlights approximately when the swing foot is off the ground and we would expect a positive moment integral if the swing leg is used in accordance with modeled inertial balance strategy.

Table 3.3 Swing hip moment impulses during swing phase, mean +/- standard deviation, across perturbation force levels for swing-side and stance-side perturbations. Results are significantly different between force levels and between sides at all force levels for $\alpha = 0.05$.

	SWING-SIDE		STANCE-SIDE	
FORCE	*mean*	*sd*	*mean*	*sd*
LOW	-0.00126	0.00060	-0.00145	0.00090
MEDIUM	-0.00180	0.00094	-0.00232	0.00099
HIGH	-0.00230	0.00098	-0.00267	0.00123

3.3.8 Stutter steps suggest configurations where balance response is challenging

There were a few side perturbations that resulted in unusually short step durations and lengths. In most of the cases, this involved subjects moving their swing foot laterally behind the stance foot and taking a short step, followed by another short duration step by the original stance foot. There were also a few instances where subjects briefly lifted the foot that recently landed on the ground and shifted its mediolateral position before landing again, effectively taking a double step with the same foot. More of these instances occurred as perturbation force increased (1 instance at low force, 7 instances over 5 subjects at medium force, 18 instances over 8 subjects at high force). All but one instance occurred after early-stance perturbations. Some subjects who performed stutter steps informally mentioned feeling unable to complete a normal step without falling.

3.4 Discussion

We observed multiple contributions to lateral balance in response to medio-lateral perturbations. As previously recognized in the absence of perturbations, foot placement and ankle inversion-eversion contribute to maintenance of upright body orientation. Of the two, foot placement has a much more pronounced effect on the COP, with ankle torque responsible for about 2-3 cm of COP movement while foot placement could move upwards of 40 cm at the highest lateral displacement.

But in addition to these effects, we also found that humans perform inertial balance strategies that also contribute to balance. This was similar to the hip strategy observed in sagittal plane standing posture, but applied here to frontal plane motion during walking. The strongest indicator was stance leg hip abduction, which was modulated as a function of perturbation timing. It was most pronounced for early, stance-side perturbations, perhaps because swing foot placement alone is impeded by the stance leg, and must first swing past it before deviating medially to counteract the perturbation. It is therefore helpful to initially rotate the upper body accordingly. If the perturbation is clockwise (counter-clockwise), the upper body should be rotated clockwise (counter-clockwise). We found that the upper body, which accounts for nearly 70% of body mass (Winter, 1995) accounted for most of the inertial motion. The swing leg did not seem to contribute, with its movement seemingly dominated by step placement priorities. The arm towards the perturbation did not seem to contribute, while the arm away from the perturbation may sometimes contribute during late swing-side and early stance-side.

Interestingly, we found evidence that inertial balance strategies also contribute to swing-side perturbations. Such perturbations may be considered easy to counter, because there are no obstacles to lateral foot placement, which should therefore be sufficient to retain balance. Even though such foot placement was clearly performed, it was also accompanied by significant inertial action. The upper body was rotated in the direction of the swing-side perturbation, particularly if applied in the middle-to-latter portion of the step. Though there was a lack of increase in hip moment to carry that contribution to the ground until very late-step perturbations, another possibility is that positive upper body moment offsets the negative inertial contribution from moving the swing leg for foot placement (Figures 3.14, 3.19).

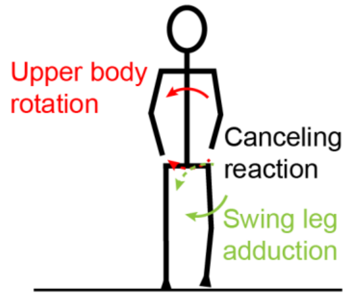

Figure 3.19 Schematic showing how the reaction from rotating the upper body generates a reaction at the pelvis that offsets the pelvis reaction from moving the swing leg for step placement, thus limiting or eliminating the negative inertial effect from step placement.

One interpretation is that the entire body acts in concert to contribute to balance. In standing posture, the ankle and hip strategies are understood to not to be independent, and to occur in combination (e.g., Park & Kuo; Kuo; Horak, etc.). The ankle strategy can move the COP, but only to limited degree due to the finite length of the foot, and resisted by the considerable inertia of the body as an inverted pendulum. The hip strategy is particularly effective for fast balance reactions, with less effective inertia about the hip than about the ankle (Kuo & Zajac). And once the COM motion has been arrested, there may be ample time to restore upright posture. Applying such concepts to walking, the most straightforward way to counteract a perturbation is with foot placement, which can move the COP substantially, but only upon footfall, and most directly in the lateral direction. Inertial strategy can act more quickly than foot placement, and can even delay when the next footfall occurs. In addition, inertial strategy can offset some of the negative COM effect generated by the act of step placement.

Stepping strategies seem to be the preferred mechanism for lateral balance control during normal walking. While inertial balance strategies do seem to contribute significantly across all force levels and on a magnitude similar to ankle strategy in the period immediately after a perturbation (Figure 3.20, Table 3.4), the total amount of contribution seems not to increase much, whereas stepping strategies, especially lateral step placement and step timing, change dramatically. One reason may be that the magnitude of balance correction that can be produced by stepping is far greater than that of inertial movement. Inertial strategies may also be disincentivized as it requires deviating from upright posture, which may be uncomfortable or unfamiliar in normal walking situations.

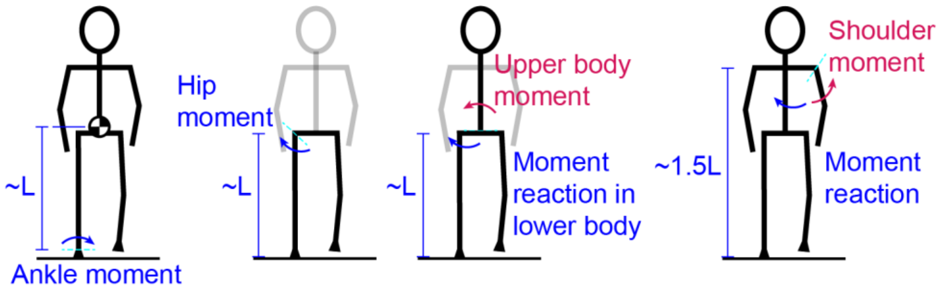

Figure 3.20 Approximate effective moment arms for each joint moment's force impact on the body COM. The ankle moment acts directly on the COM with a length of approximately leg length L. Stance hip moment and upper body/swing hip reaction moment act through the stance leg to push against the ground, also approximately with moment arm L. Shoulder moment acts through the torso and stance leg to push against the ground, with a moment arm of approximately 1.5 L.

Table 3.4 Comparison of estimated force impulse magnitude on COM immediately after high force perturbations. Foot placement impact, which occurs after foot contact and on a different time scale, is included for reference and has significantly greater balancing impact on the COM than the other mechanisms. Ankle and upper body moments immediately after perturbations are constantly positive and of similar magnitudes, though for perturbations at different phases of the gait cycle, and could be considered to have similar overall effects. Hip moment is particularly contributing with a greater overall effect around early stance-side perturbations and to a lesser extend around early swing-side perturbations, with negative contributions relative to nominal unperturbed during other times. Shoulder moment impulse is generally negative except around early stance-side, but the magnitude of the moment and the larger moment arm makes the balancing force on the COM approximately an order of magnitude smaller than the other mechanisms.

	Ankle	*Hip*	*Upper body*	*Shoulder*	*Foot placement*
Force equation	Moment/L	Moment/L	Moment/L	Moment/1.5L	∆COP×GRF
Early stance	0.001	0.006	0.0005	0.00005	0.05
Mid stance	0.002	-0.001	0.002	-0.0001	0.025
Early swing	0.0005	0.001	0.001	-0.0001	0.02
Overall effect	+++	++++++/---	+++	+/-	++++++++++ ++++++++++

It is further possible that walking on the treadmill at fixed speed, which enforces taking steps to maintain forward walking, biases the response towards stepping. If one must step anyways to avoid falling off the treadmill, then it may make sense to put more effort into maximizing the balancing abilities of stepping. Inertial balance strategies can interfere with forward walking in later steps: rotating the upper body shifts the center-of-mass, and ideal rotation of the swing leg for inertial balance is in the opposite direction of rotation required for stepping. It is possible that in a normal over-ground walking environment without enforced speeds, humans increase their preferences for using inertial balance strategies coupled with slowing or stopping.

These findings add new dimension to the understanding of lateral balance during walking. Previous studies have emphasized the importance of foot placement (Bauby and Kuo, 2000) and ankle inversion/eversion (Hof and Duysens, 2018), particularly in the absence of explicit perturbations. But lateral perturbations cause greater disturbance to COM motion, and necessitate greater responses. A number of other studies have described the kinematics and other features of the balance responses (Afschrift et al., 2019; Batcir et al., 2020; Hof et al., 2010), but few have considered the inertial ramifications. Here we have shown that motion of the rest of the body, particularly the trunk, are consistent with inertial balance control. These actions appear more important when reacting to quite substantial perturbations, and are less evident during normal, unperturbed walking, where we still consider lateral foot placement to be the primary compensation. Nevertheless, our results suggest that the entire body can act to maintain balance.

Additional experimental conditions could potentially help test trade-offs between inertial balance strategies and stepping even on a treadmill. For example, a narrower walking surface could constrain lateral foot placement and establish a continuity between partly-limited step placement and inertial strategies. Or a constrained step frequency could alter the time the body is unfavorable poses that present challenges to stepping (such as early stance perturbations that induced skipping steps). Perturbation studies at lower speeds or on larger walking surfaces could help identify if the drawbacks from walking disruption impact inertial strategies preferences.

The perturbation system used here reaches peak forces over 250 ms due to physical limitation of the available hardware. This duration represents a relatively large period of an average step, and there was some opportunity for subjects to respond to perturbations before it completed. Our analysis here ignores behaviors during the perturbation force ramp since it is difficult to distinguish what is an active effect opposing the perturbation and what is a direct result of the perturbation on the body. By analyzing only after the perturbation system force peak, resulting forces/moments should all be part of the compensation strategy and not a direct effect of inducing the perturbation. This may underestimate/miss behaviors that occur before the perturbation force ramp completes.

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Chapter 4 Step Placement Constraints Increase Inertial Balance Strategy Usage During Perturbed Treadmill Walking

4.1 Introduction

Humans contend with a variety of challenges to their balance during walking. One of the principal challenges is with lateral balance control, which is associated with falls in older adults (Maki, 1997). Perhaps the most direct means to maintain stability is to actively adjust lateral foot placement when stepping, or adjust the forces or torques exerted by the foot already on the ground. But there often environmental constraints during daily living that could limit such compensations, such as when walking on a narrow path or avoiding a puddle. There are nonetheless clearly other ways to maintain balance, as illustrated by a tightrope walker, who cannot directly adjust their stepping but must instead move other parts of the body such as the torso or arms. It is unknown to what degree such compensations are normally performed, and how they are modulated by environmental constraints. If these control actions could be quantified, insight might be gained on how humans select from the options available for maintaining balance during walking.

Balance compensations may be categorized as direct and indirect actions. Direct strategies exert force or torque against the ground to move the body's center of pressure (COP) relative to the center of mass (COM) (Hof, 2007; Kuo, 1999; Wang and Srinivasan, 2014). For lateral balance, possible actions are lateral step placement and inversion-eversion ankle torque, which are apparent in normal human walking (Hof et al., 2010; Joshi and Srinivasan, 2019). Indirect actions use the body's inertia to restore balance, by rotating body segments. For lateral balance, the angular momentum of the body only changes slowly, and so a fast rotation of the torso in the direction of the fall induces an opposing rotation of the stance leg that accelerates the COM away from the fall. The most obvious case is rotation of the torso about the hip (Hof, 2007), but it is also possible to move the arms or swing leg to similar effect. Such actions have mostly been observed in standing or very low speeds (Best and Wu, 2020; Otten, 1999), or very constrained walking such as balance beam or tightrope walking. However, it is possible that indirect, inertial balance strategies are modulated, so that they contribute little at normal speeds but significantly more when environmental constraints limit the applicability of direct balancing actions. This may be the case when the lateral walking surface is narrow or has insufficient room for natural step placement. This suggests that the degree of direct and indirect balance control should change continuously with walking conditions.

We have previously used a dynamic walking model to demonstrate direct and indirect strategies during gait. The model shows that step placement is by far the most effective means of directly moving the COP, except that this motion does not occur until the next footfall. Direct step placement thus occurs discretely in time, and has no immediate effect on the body state. In contrast, indirect actions move the body's inertia continuously and immediately in time. They have comparatively smaller effect on the COM, but the immediacy may be advantageous. This could potentially explain why they are more evident during very slow walking (Best and Wu, 2020; Otten, 1999), when the delay until the next step placement is longer. Another drawback is that medial step placement can sometimes be physically obstructed by the stance leg. The swing leg's medio-lateral motion may thus be delayed until mid-stance, which could constrain the achievable amount of step placement. In addition to environmental constraints, the body itself can also obstruct step placement and increase the need for inertial control actions.

The purpose of this study was to characterize the combination of direct and indirect control actions that humans employ as a function of constraints on normal balance control. The challenges were due to lateral balance perturbations applied to walking, along with artificially imposed environmental step constraints similar to a narrow walking path. We expected that perturbations that occur earlier in the stride should require larger control actions, both direct and indirect, due to the greater time before step placement takes effect. We also expected that indirect control actions would have greater contribution with the addition of step constraints. Even though step placement is the dominant response during normal walking (Chapter 3), it is also possible that inertial balance strategies are modulated continuously as a function of walking conditions and will be more prominent when walking balance is most challenged.

4.1.1 Model predictions

With constrained stepping regions, we expect decreased lateral foot placement in response to perturbations. Assuming there is coupled control of foot placement and inertial movement, this decrease should be accompanied by greater inertial balance strategy as measured by moments at the hip and trunk, and manifested as a sharp acceleration of the trunk in the direction of the perturbation.

A lateral foot placement constraint is expected to have greatest effect for early perturbations, which usually call for the most placement. This may be accompanied by small reductions in step duration and step length, because less placement can mean earlier ground contact. Constraints effects may also spill over into the second step, if the ability to compensate in the first footfall is incomplete. A spill-over means some additional foot placement may occur on the next step, particularly so for early stance-side perturbations, which usually call for the most (first step) foot placement.

There may be an increase in lateral ankle strategy, using ankle inversion/eversion torque to act on the COM. This is expected to act in the positive (correcting) direction, although also expected to have relatively small effect compared to foot placement. As with other cases, the largest (of a small) effect is mostly expected for stance side, which is most challenging.

We expect to humans to employ inertial balance strategies to greater degree when foot placement is most restricted. The inertial strategies may be quantified by the stance hip and trunk motions, where the trunk should be rotated in the direction of the perturbation. Foot placement is particularly restricted during double support in either direction, and balance may be regained in part from appropriate inertial action. In addition, stance side perturbations are expected to induce more inertial control due to the greater amount of placement (compared to swing-side) needed and the attendant constraints on that placement. We would also expect larger perturbations to require more placement, and thus require more use of inertial strategy given stepping constraints.

The body's appendages may contribute more to inertial balance given stepping constraints. The arms have relatively little inertia, and so they're not expected to have much effect on their own beyond being carried by the trunk. But if they are used, they (or far arm) should at least rotate consistent with a lateral hip strategy, albeit to small effect. The stance leg would be expected to move away from the perturbation under inertial balance strategy. Though we have previously found that the swing leg may prioritize foot placement over inertial strategy (Chapter 3), it is possible that constraining foot placement may change the priority in favor of more inertial strategy.

4.2 Methods

The experiment applied unexpected lateral perturbations to healthy human subjects while walking on a treadmill and subject to constraints on where to step. There were four subjects (2 male, 2 female, ages 24-31) who participated in the experiment, after providing written informed consent according to University of Michigan Institutional Review Board procedures. Some statistics for the subjects are listed in Table 4.1. Movement data was collected via active motion capture markers (PhaseSpace, San Leandro, CA, USA) on each participant's torso, pelvis, arms, and legs. Force data was collected from an instrumented split-belt treadmill (Bertec Corp, Columbus, OH, USA). Standard kinematics and inverse dynamics analysis was performed using models built in Visual3D (C-Motion, Germantown, MD, USA) and computed in MATLAB (MathWorks, Natick, MA, USA).

The perturbation system is the same as described in Chapter 3, summarized briefly here. The perturbation system comprises of two motorized spools pulling cables to the left and right of a harness worn by the subject (Figure 4.1). The harness comprises of a lightweight adjustable aluminum structure around a waist belt, with cable attachments in front and behind the subject that meet the motor-driven cables out to the left and right of the subject. The cables form a gap area that allows the subject to freely move their arms during walking while still allowing lateral perturbations to be applied at the subject's waist. The motors are force-controlled to regulate the tension on the cable, normally holding with equal force to allow natural body motions and superimposing a sawtooth-shaped increase in force over approximately 250ms to produce perturbations. Perturbations were applied in a random lateral direction at randomly selected time points during a stride (as measured by instrumented treadmill ground reaction force), and at several discrete force levels designated as low (approximately 1.5 N/kg of subject mass at the peak of the sawtooth) and medium (approximately 2.25 N/kg of subject mass). Perturbations were tested in groups of ten at the same force level, with randomized wait of 8-14 strides between perturbations. Bouts of perturbations were conducted until there was two perturbations per side, per time point in the stride (ten equally spaced through the stride), per force level, for a total of 40 perturbations per condition-force level per subject.

Table 4.1 Subject data. Leg length was measured from greater trochanter to ground.

SUBJECT	SEX	MASS (KG)	HEIGHT (M)	LEG LENGTH (M)	WALKING SPEED (M/S)
1	f	56.7	1.75	0.91	1.25
2	m	86.2	1.90	0.98	1.25
3	m	68.0	1.68	0.91	1.25
4	f	65.8	1.67	0.94	1.25
MEAN		69.2	1.75	0.94	
S.D.		12.4	0.11	0.03	

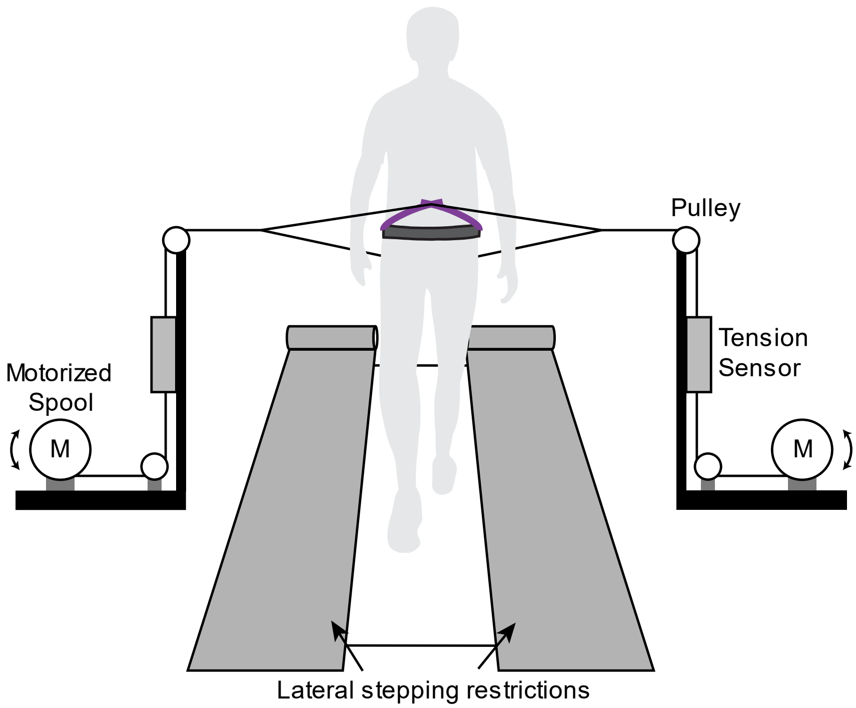

Figure 4.1: Perturbation system with constraints on lateral step placement. A human subject walks on an instrumented treadmill, with two lightly-tensioned cables pulling laterally and opposite each other at the waist. Force control ensures that the cables normally allow for natural body motions, and allows perturbations to be superimposed. Perturbations were applied at random times during a stride, with a ramp profile with duration about 250 ms. Constraints on foot placement were applied by restricting the allowable stepping area, from unconstrained to narrow (about 1.5 times nominal step width) to narrowest (1 times nominal step width).

Subjects were asked to walk normally as well as with constraint regions to restrict their mediolateral step placement. The regions were formed by draped rolls of paper covering distal portions of each treadmill belt. This restricted the subject to a strip about the centerline of the treadmill, with the allowed strip width set to two condition levels: a *narrow* condition equal to approximately one-and-a-half times their normal step width, and a *narrowest* condition equal to approximately one times their normal step width. Each subject's step width was measured by the experimenter as the distance from the lateral edge of one foot to the lateral edge of the other foot while the subject walked normally in the tensioned experimental setup without experiencing perturbations or constraints. Perturbations were performed randomly at the low and medium force levels, with the low force level in conjunction with both condition levels, and the medium force level at the narrow level only.

We measured gait mechanics using standard procedures and computed a number of derived measures as described in Chapter 3. Measurements were normalized in the same manner as previously described.

4.3 Results

The perturbation system produced lateral perturbations and subjects altered their walking parameters in response to constraints as observed through kinematic, force, and balance measures. We refer to perturbations as swing-side or stance-side based on whether the perturbation force was directed in the direction of the swing leg of stance leg, respectively. We treated strides as beginning with foot contact for a swing-side perturbation from 0-50% of a stride, followed by a stance side perturbation from 50-100% of stride. Measures were defined as positive when such deviations acted to restore the COM opposite to the perturbation direction, and each perturbation can be examined in terms of deviation from the nominal, unperturbed case.

There were some changes to unperturbed step width as a result of the constraint conditions. Notably, the average step width may have decreased under the narrowest constraint condition at low perturbation force (Table 4.2). This may slightly decrease the stability of the walking stance, but nevertheless should still limit foot movement for step placement more than the narrow condition, so the comparison between lateral step conditions levels should still reflect increased constraint.

Table 4.2 Unperturbed step parameters as a result of lateral stepping restrictions. Step width (meters), step duration (seconds), and step length (meters) are reported with mean (μ) and standard deviation (sd) for each subject as well as the mean and standard deviation across subjects, for unconstrained (Unc), narrow (Nar), and narrowest (Nar+) condition levels.

	STEP WIDTH (M)						STEP DURATION (SEC)						STEP LENGTH (M)					
COND:	**Unc**		**Nar**		**Nar+**		**Unc**		**Nar**		**Nar+**		**Unc**		**Nar**		**Nar+**	
SUB	*μ*	*sd*	*μ*	*sd*	*μ*	*sd*	*μ*	*sd*	*μ*	*sd*	*μ*	*sd*	*μ*	*sd*	*μ*	*sd*	*μ*	*sd*
1	0.15	0.02	0.16	0.02	0.16	0.02	0.99	0.02	0.98	0.02	0.98	0.02	1.24	0.04	1.23	0.04	1.24	0.03
2	0.17	0.03	0.18	0.03	0.15	0.02	1.08	0.02	1.07	0.02	1.07	0.02	1.36	0.03	1.35	0.03	1.34	0.03
3	0.23	0.03	0.21	0.02	0.18	0.02	1.13	0.02	1.13	0.04	1.15	0.03	1.41	0.04	1.44	0.05	1.42	0.06
4	0.18	0.03	0.19	0.03	0.17	0.02	1.09	0.02	1.08	0.02	1.08	0.02	1.38	0.03	1.36	0.05	1.35	0.03
MEAN	0.18		0.19		0.16		1.07		1.07		1.07		1.35		1.34		1.34	
SD	0.03		0.02		0.01		0.06		0.06		0.07		0.07		0.09		0.08	

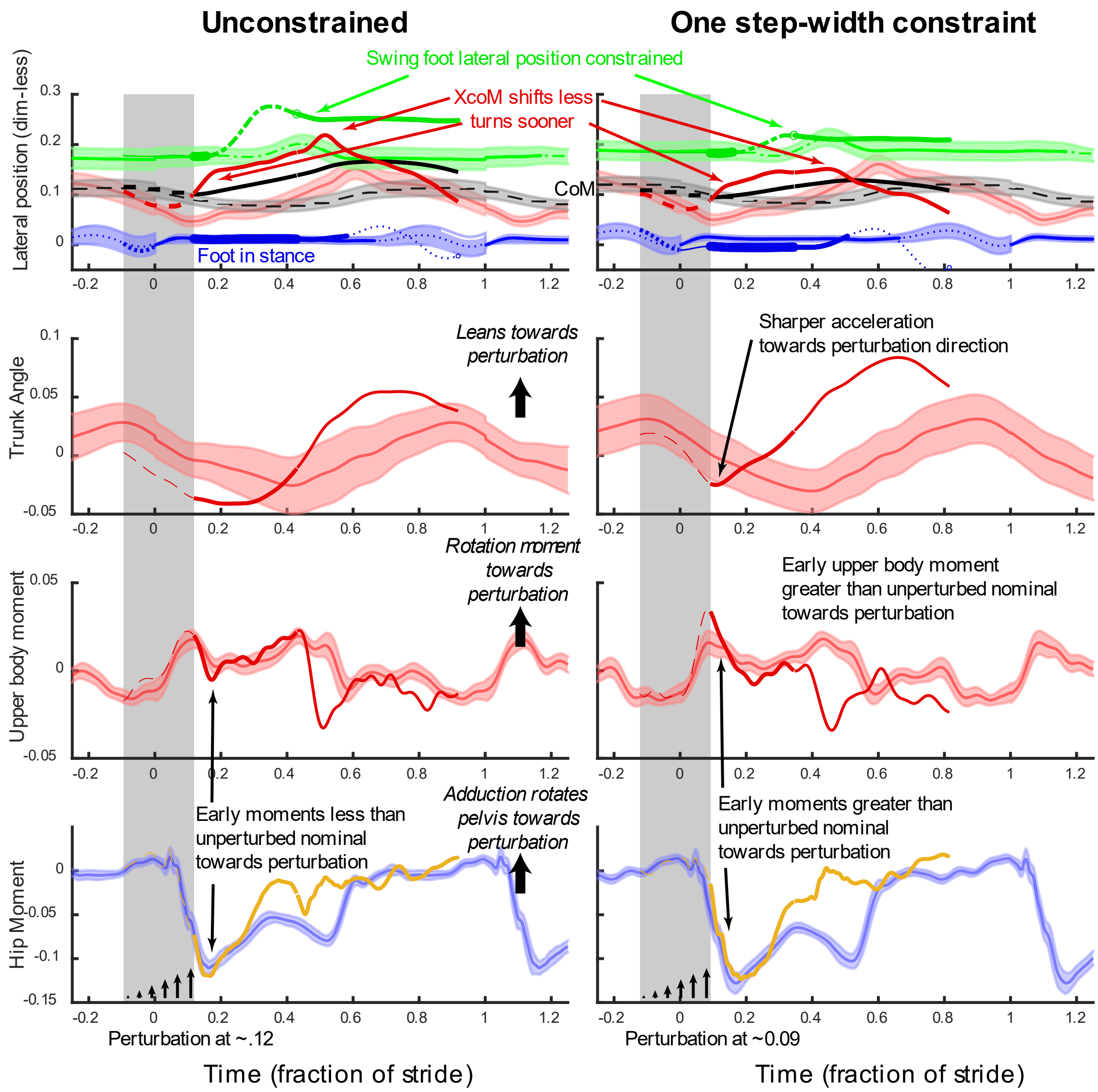

Figure 4.2 Representative data for early swing-side perturbations without constraints (left) and with one step-width lateral stepping restrictions (right). Perturbed measures (thicker, darker) are plotted over the nominal unperturbed measures (shown as mean line with +/- 1 standard deviation shaded curve). Perturbations occurred during the vertically shaded time, pushing the body COM (black line, top) towards the green foot entering swing. The swing foot moves laterally in response to the perturbation, significantly in the unconstrained condition while only slightly in the constrained condition. The XCOM (red line, top) also moves laterally towards the swing foot before its rate of excursion decreases, with this change in rate occurring earlier in the constrained condition. There was also a sharp acceleration of the trunk angle to lean towards the swing leg direction in a similar time frame (row 2). The upper body moment (row 3) also reflected an increase in rotation, beginning significantly above the unperturbed nominal mean before dipping below in the first 0.1 stride of the constrained condition as opposed to within the standard deviation band and dipping below the nominal. The stance hip moment (bottom row) appeared to be mostly within the 1 standard deviation band of nominal during the first 0.1 stride, though notably below in the unconstrained condition (signifying more abduction that would increase the rate of XCOM and COM excursion) and above in the constrained condition (signifying less such abduction).

4.3.1 Lateral step constraints reduced foot placement and altered step duration and length

Narrower lateral step placement constraints led to decreased lateral foot movement towards the perturbation force direction and changes in step durations and lengths (Figure 4.2). In the narrow condition, there were reduced deviations in lateral step placement in both early swing-side and early stance-side at both perturbation force levels (Figure 4.3, 4.5). For medium force perturbations, as the unconstrained step placement was generally greater than for low force perturbations, the equivalent constraint meant that perturbations later into the mid-step also had reduced step placement. For low force perturbations, subjects also walked under the narrowest constraint condition, which corresponded with further reduced step placement deviation on both sides but especially in swing-side, where lateral step placement was essentially restricted for perturbations throughout the step. The changes in step placement were reflected by changes in the duration of the perturbed step and sometimes the step length. Notably, for low-force perturbations, the most constrained condition resulted in significantly reduced step times in early swing perturbations compared to unconstrained, while constraints in early stance perturbations appeared to reduce step length (Figure 4.3, right column). For medium-force perturbations, constrained step placement seemed to correspond with decreased step duration and step lengths (Figure 4.5, right column).

The behavior of the second step after perturbation generally reflected compensations that suggest a return towards nominal stepping for early-to-mid step perturbations and making up for incomplete compensations for late step perturbations. For low-force perturbations, lateral step placement was generally around nominal except for very late step perturbations, whose second step behavior resembled first step behavior of early perturbations of the other direction (suggesting an incomplete compensation from the first step) and thus were impacted by step placement restrictions (Figure 4.4). For medium force perturbations, the incomplete compensation behavior seemed to extend to somewhat earlier perturbations, and thus more mid-to-late step perturbations were impacted by the stepping restrictions (Figure 4.6). The resulting effect for later-step perturbations was similar to early step perturbations in the first step, with reduced step duration and step length. For early-step perturbations in the original step, especially towards swing-side, the second step behavior seemed to in part mitigate the stepping behavior of the first step, with a still reduced but closer to nominal step duration but a much larger step length, perhaps to maintain walking speed on the treadmill.

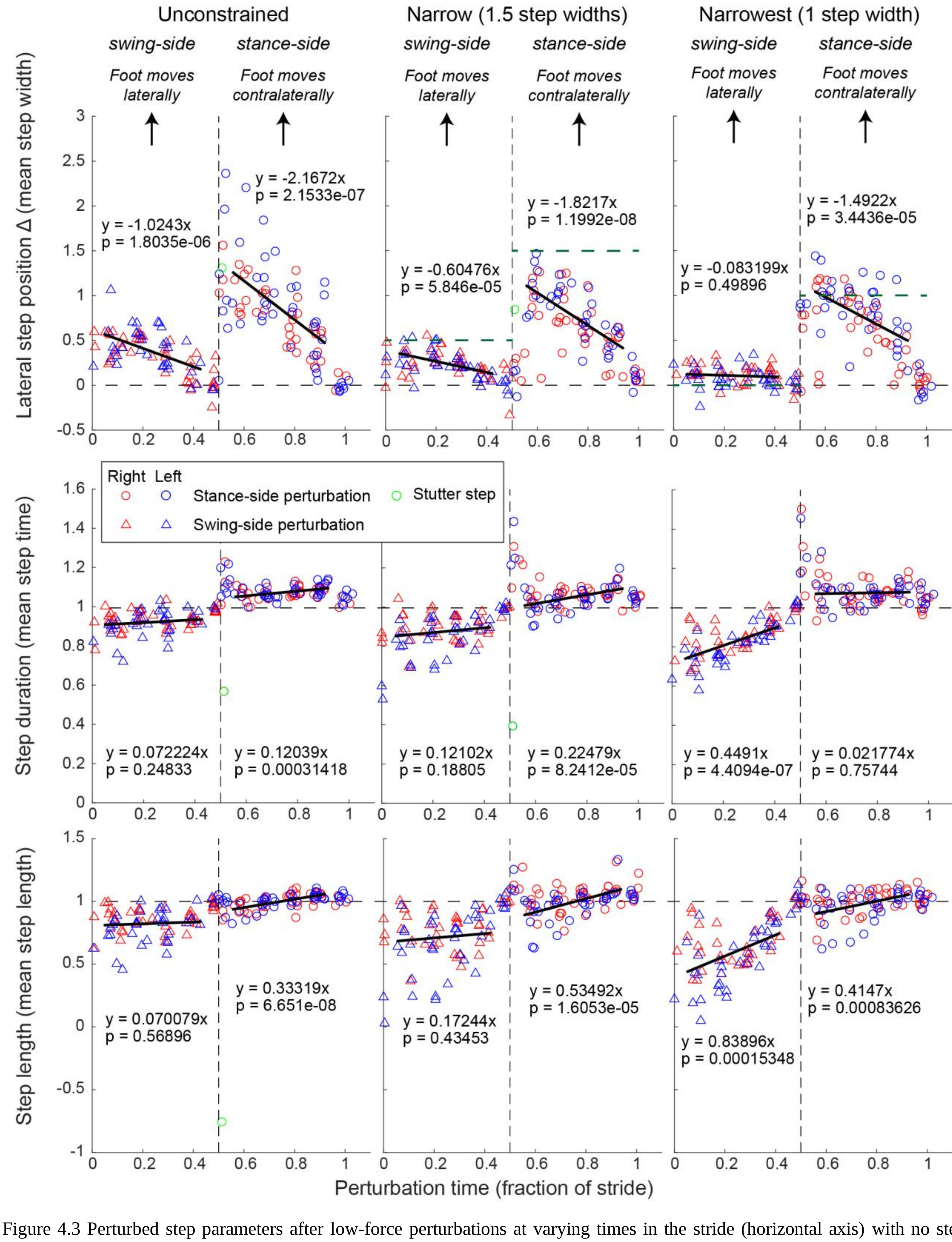

Figure 4.3 Perturbed step parameters after low-force perturbations at varying times in the stride (horizontal axis) with no step placement constraints (left col), narrow width (1.5 nominal step widths, middle col), and narrowest width (1 nominal step width, right col). Approximate lateral constraints (green dashed line) reduced lateral placement deviation (top row) early in both sides for narrow condition and increasing later in the step for narrowest condition. Early stance-side step duration increased with constraints while early swing-side decreased (middle row). Swing-side step lengths decreased with constraint (bottom row).

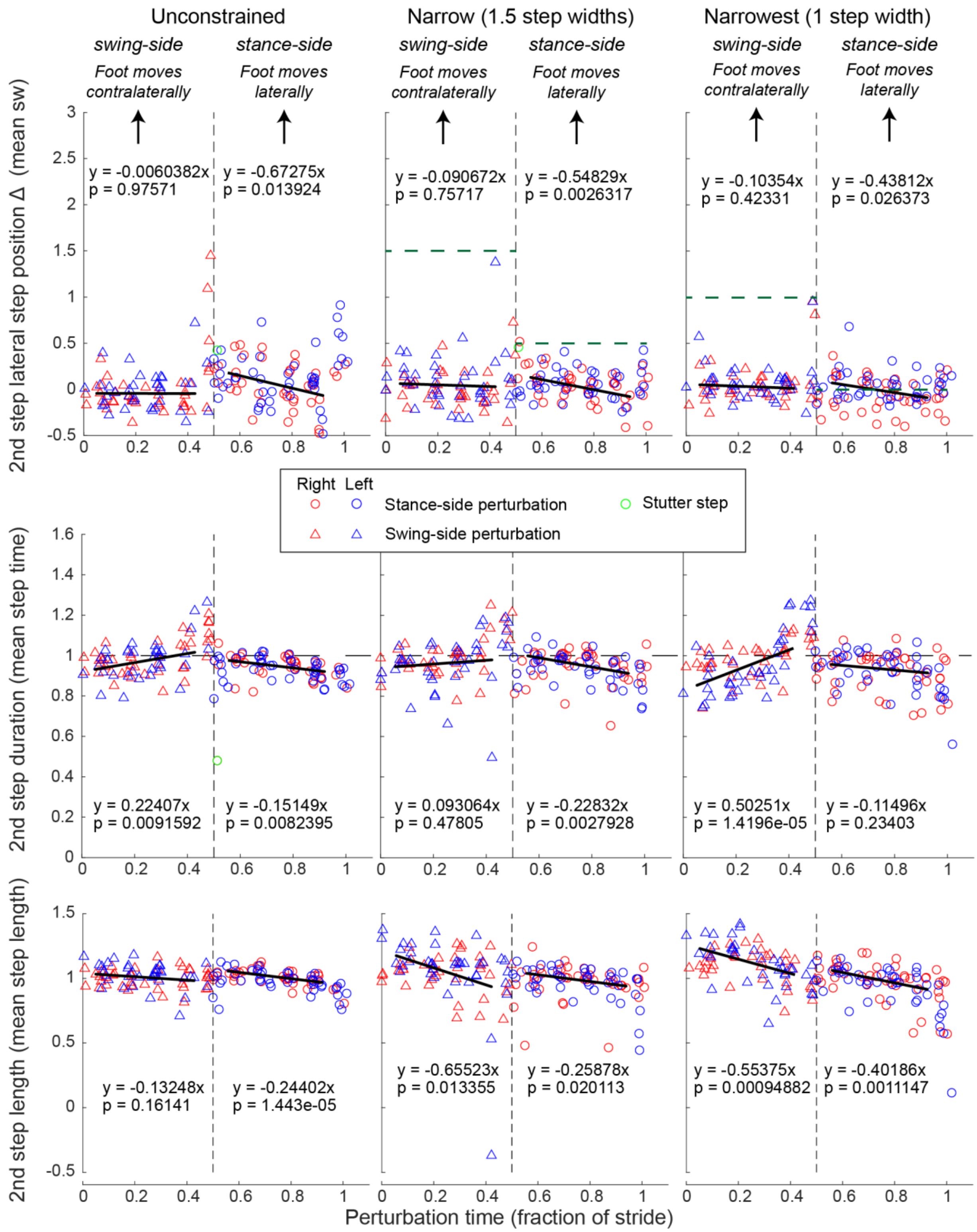

Figure 4.4 Parameters of step after low force perturbed step (2nd step) with no constraints (left col), narrow width (middle col), and narrowest width (right col), versus original perturbation time in the stride (horizontal axis). The swing foot of 2nd step moves contralaterally after originally swing-side perturbations and laterally after originally stance-side (top row), with constraints decreasing placement for very late-step perturbations on both sides. The narrowest constraint decreased 2nd step durations after early perturbations towards either side (middle row). Early swing-side had longer 2nd step lengths with constraints (bottom row).

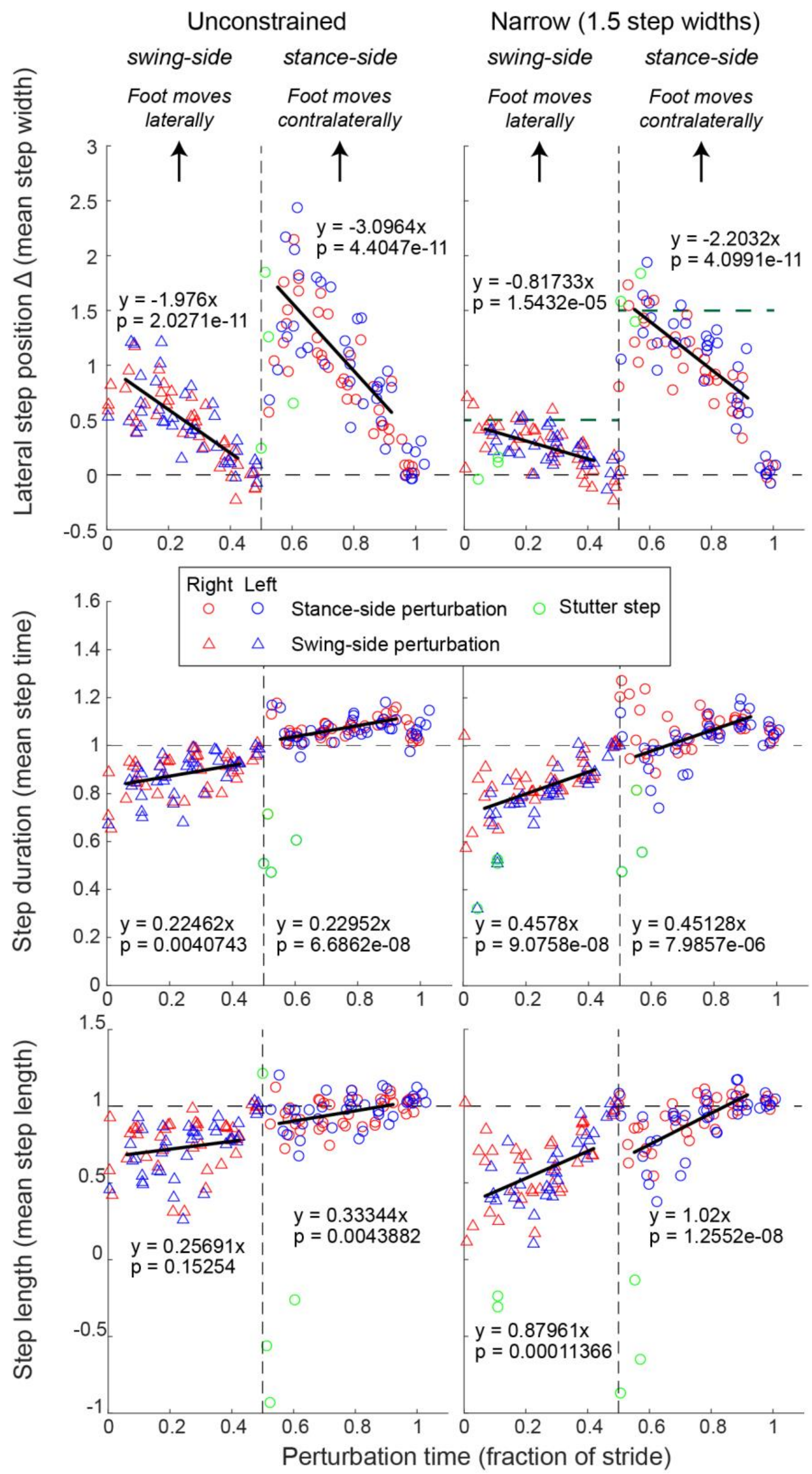

Figure 4.5 Perturbed step parameters after medium-force perturbations at varying times in the stride (horizontal axis) with no step placement constraints (left col) and narrow width (1.5 nominal step widths, right col). Approximate lateral constraints (green dashed line) reduced lateral placement deviation (top row) early through mid-step in both sides for the narrow condition. The reduced step placement seemed to correspond with reduced step duration (middle row) and step lengths (bottom row).

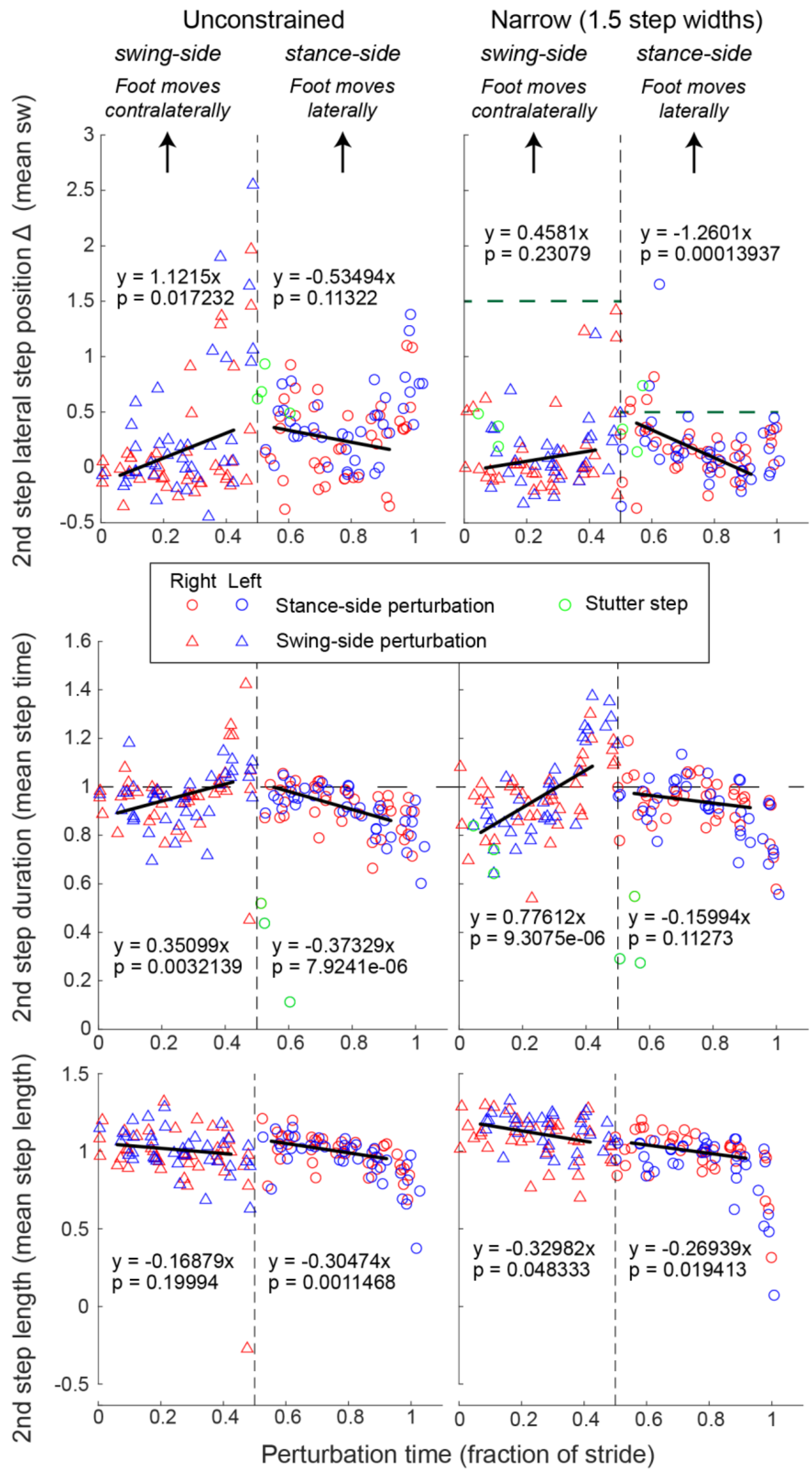

Figure 4.6 Parameters of step after medium force perturbed step (2nd step) with no constraints (left col) and narrow width (right col), versus original perturbation time in the stride (horizontal axis). The swing foot of 2nd step moves contralaterally after originally swing-side perturbations and laterally after originally stance-side (top row), with constraints decreasing placement for very late-step perturbations on both sides. The narrow constraint decreased 2nd step durations most significantly for early swing-side perturbations (middle row). Swing-side also had longer 2nd step lengths with constraint (bottom row).

4.3.2 Stance ankle moment acts against perturbation limited changes due to constraint

We observed inversion-eversion ankle torques consistent with lateral ankle strategy in the 0.1 strides immediately after the perturbation, but there was not a consistent change in the impulse of the ankle moments as a result of constraints (Figure 4.7). There was a statistically significant increase in stance side moments between unconstrained and narrow conditions at low perturbation force (p = 0.0246, unconstrained: mean +/- s.d. of 0.000668 +/- 0.000692; narrow: 0.000912 +/- 0.000666), but this behavior did not continue for the narrowest condition, nor was the narrowest condition significantly different from unconstrained. Examining the result of the perturbations at every 10% of stride, we found one instance of statistically significant difference between unconstrained and narrow at 50%-60% stride corresponding to early stance-side (p= 0.0330, figure 4.8, but again the behavior did not extend to narrowest constraint, nor was there significant different at any other time division. For medium force perturbations, there was no significant different between the unconstrained and narrow condition for perturbations grouped by side or by 10% of stride.

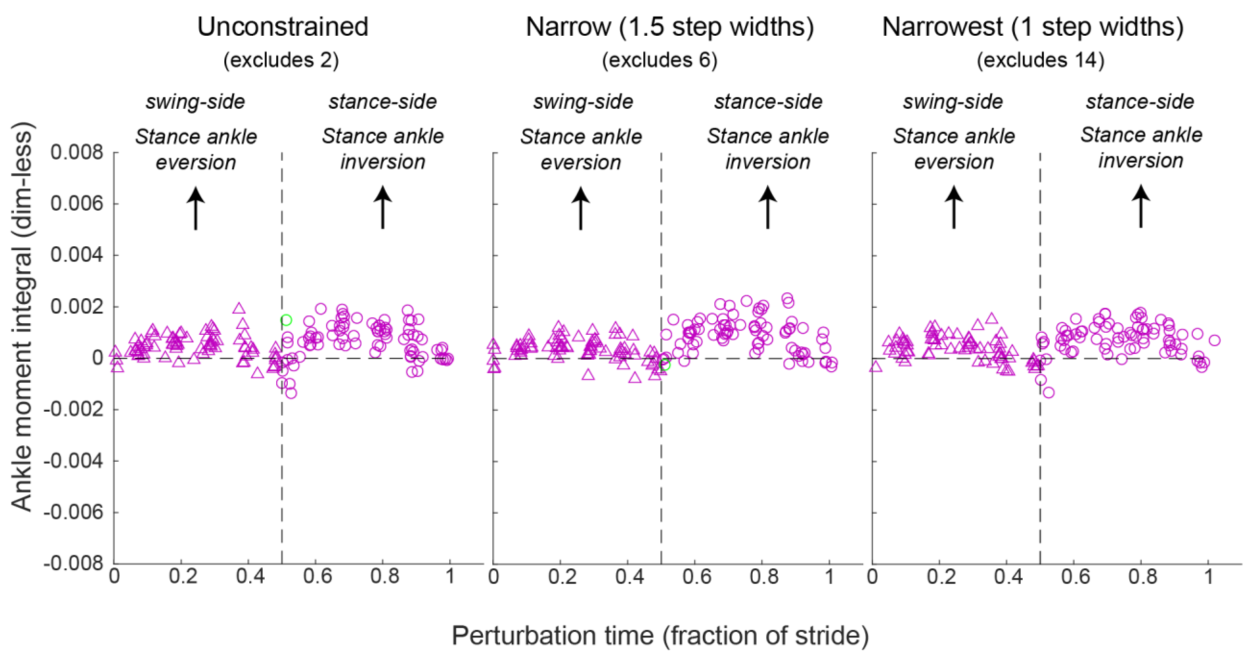

Figure 4.7 Integral of ankle moment for 0.1 strides after perturbation at low perturbation force. Values show relatively little change after perturbations throughout the gait cycle due to increasing constraint.

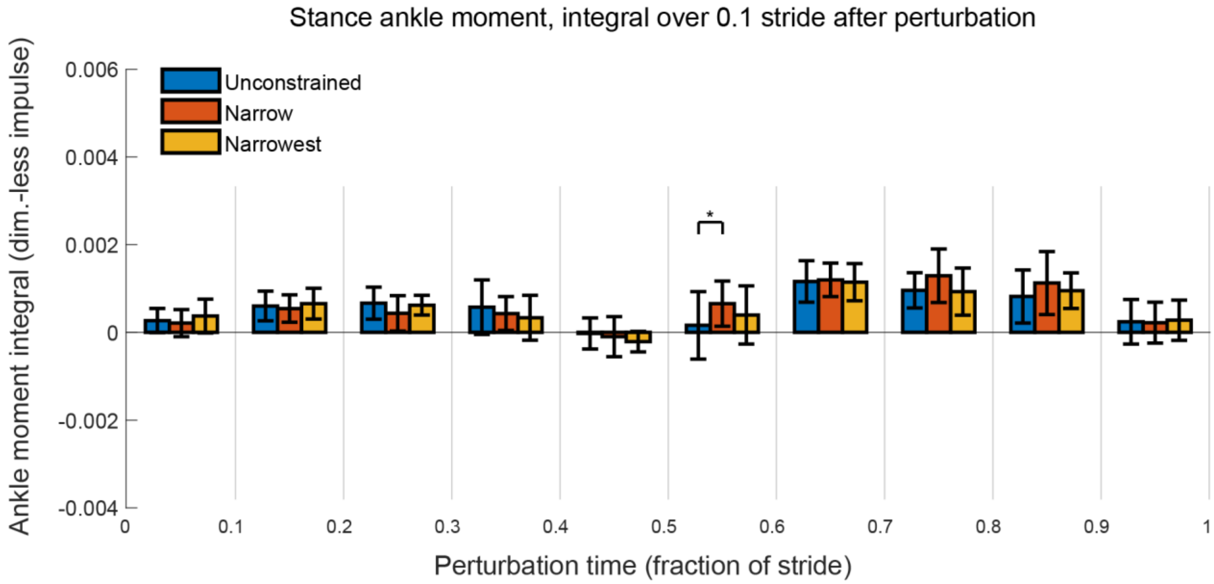

Figure 4.8 Mean and standard deviations at every 0.1 periods of stride of stance ankle moment integral from Figure 4.4. Mean impulse are all significantly different from 0 with α = 0.05 except between 0.4 to 0.5 stride under unconstrained and narrow conditions and, 0.5 and 0.6 under unconstrained condition, and between 0.9 and 1 under all conditions. The only significant difference between condition levels was between unconstrained and narrow at 0.5 to 0.6 with α = 0.05.

4.3.3 Stance hip moment impulse increased in response to step constraints

As an indicator of inertial balance strategies, we observed significant increases in stance hip moment changes immediately after the perturbation throughout swing-side perturbations with step constraints (Figure 4.9). For low force perturbations, the difference was most significant comparing unconstrained to narrowest (Table 4.3, p = 0.0008). There was also nearly significant difference for stance-side perturbations (p = 0.0889). There was a similar difference between unconstrained and narrow for medium force perturbations, although with reversed statistical significance (p = 0.0600 for swing-side, p = 0.0228 for stance side). This suggests that humans use the hip in a manner aligned with inertial strategy to response to restrictions to step placement.

In terms of specific phases of the step, low-force swing-side perturbations showed more significant increases in stance hip moment between unconstrained and narrowest conditions for perturbations during the single-support phase (0.1 to 0.4 stride). Stance-side perturbations appeared to increase throughout the step but did not reach the level of significance (Figure 4.10). For medium force perturbations between unconstrained and narrow, stance hip moment increased throughout the stride except for between 40% to 60% stride, where it remained at a similar high level (Figure 4.11).

Table 4.3 Mean and standard deviation of hip moment impulse (normalized) across condition levels. Swing-side was significantly different between unconstrained (Unc) and narrowest (Nar+), p = 0.0008. Stance-side was nearly significant, p = 0.0889.

	Swing-side		Stance-side	
Cond	*Mean*	*SD*	*Mean*	*SD*
Unc	8.67E-05	1.73E-03	8.52E-04	1.95E-03
Nar	5.72E-04	1.57E-03	9.44E-04	1.77E-03
Nar+	1.01E-03	1.47E-03	1.37E-03	1.82E-03

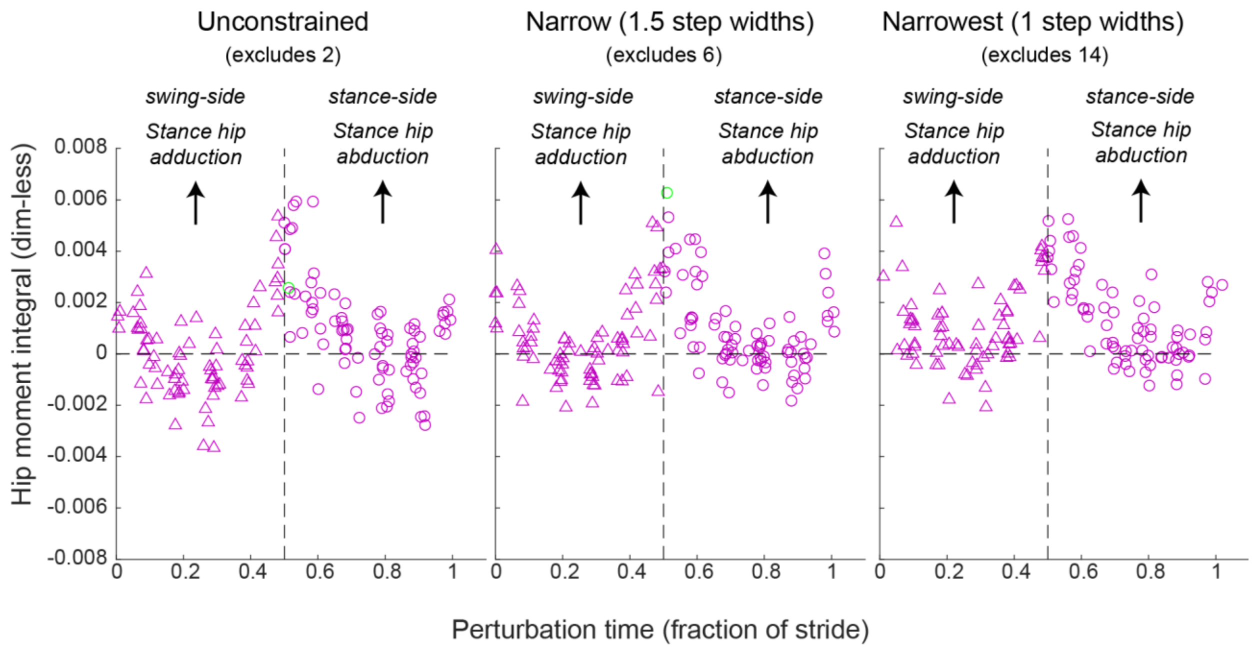

Figure 4.9 Stance hip moment impulse over 0.1 strides after perturbation at unconstrained (left), narrow (middle), and narrowest (right) constraint conditions. Impulse shows significant variation as function of perturbation time (horizontal axis), with mean increases throughout the stride as constraints tighten (See Table 4.3).

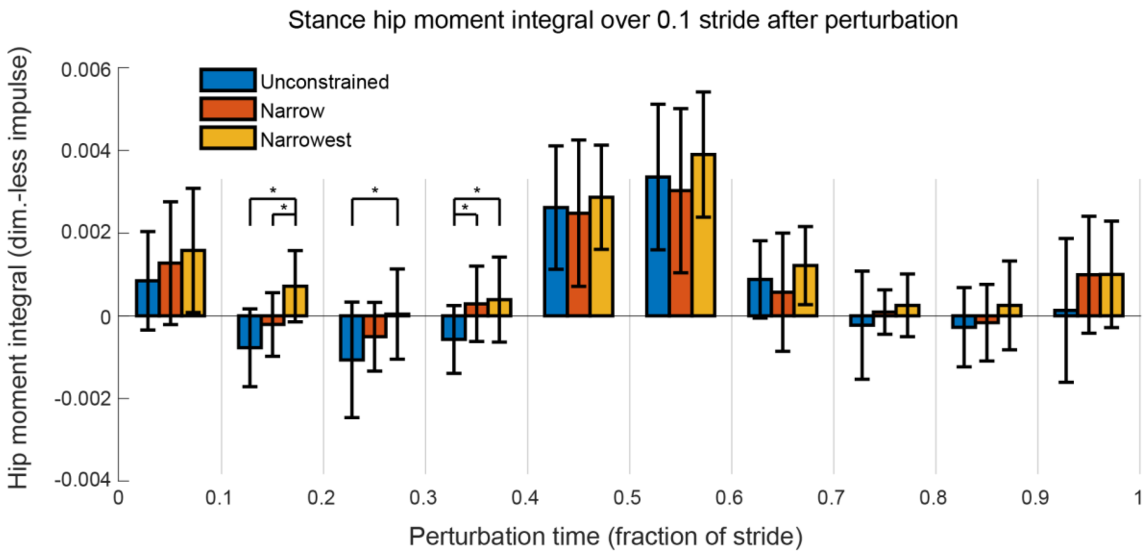

Figure 4.10 Stance hip moment impulse mean and standard deviations over 0.1 strides after low-force perturbation grouped for perturbations for every 0.1 periods of stride for width constraint conditions. Stars (*) denote significant differences at α = 0.05. Swing-side showed significant increases around single support phase (0.1-0.4 stride). Stance-side showed increasing behavior with constraints but did not reach significance threshold.

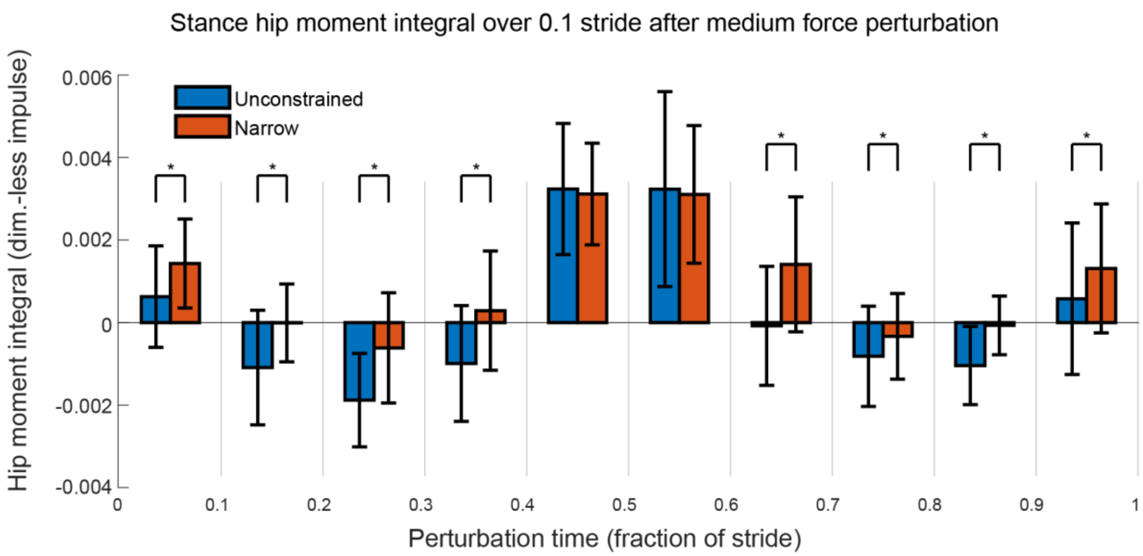

Figure 4.11 Stance hip moment impulse mean and standard deviations over 0.1 strides after medium-force perturbation grouped for perturbations for every 0.1 periods of stride for width constraint conditions. Stars (*) denote significant differences at α = 0.05. Narrow constraint condition caused statistically significant increase throughout the stride except between 0.4 to 0.6, where the hip moment was very high to begin with.

4.3.4 Tighter lateral step constraints increase upper body moment throughout stride

There upper body moment pushing the COM away from the perturbation direction increased significantly throughout the stride with tighter step placement constraints. The moment aligns in time with accelerations of trunk rotation towards the perturbation and reductions in

XCOM deviation towards the perturbation in the 0.1 stride period after the perturbation (Figure 4.2), suggesting that movement of the trunk is in part responsible for the increase in moment, and that such a moment plays a part in slowing the COM movement. For low force perturbations, there was significant increases throughout all but the middle of stance-side perturbations and early- and late-middle period of swing-side perturbations (Figure 4.12). For medium force perturbations, stance-side perturbations under narrow constraint increased significantly, and most particularly during early- and late-middle periods of the step (Figure 4.13).

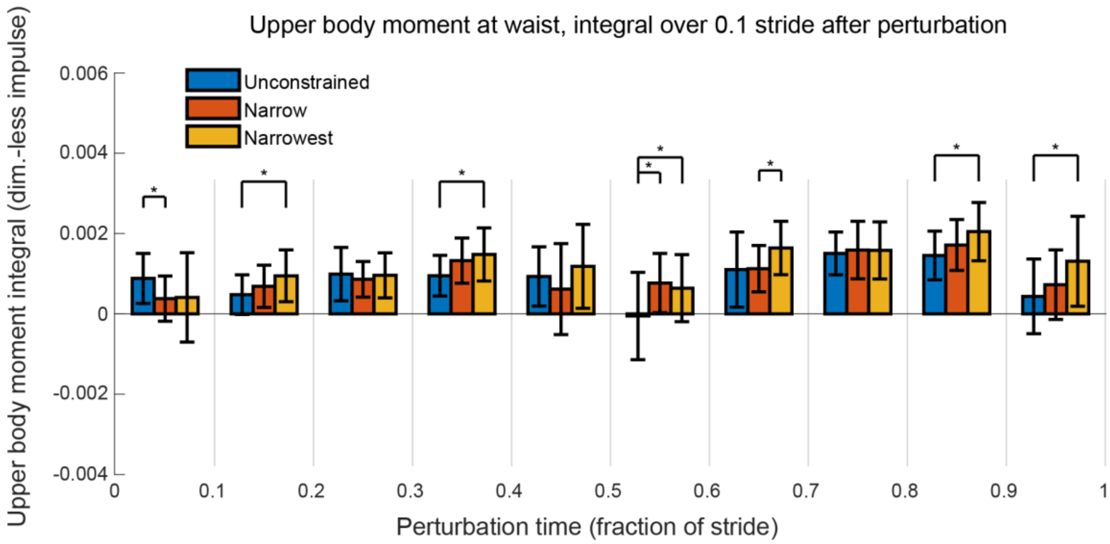

Figure 4.12 Mean and standard deviations of upper body moment that produces a reaction that pushes against COM perturbation, integrated over 0.1 strides after low-force perturbation, grouped for perturbations at every 0.1 periods of stride. Stars (*) denote significant differences at α = 0.05. Periods of both early and late step perturbations to both sides (0.1-0.2, 0.3-0.4, 0.5-07, and 0.8-1) seem to exhibit increases in moment as a result of increasing constraints.

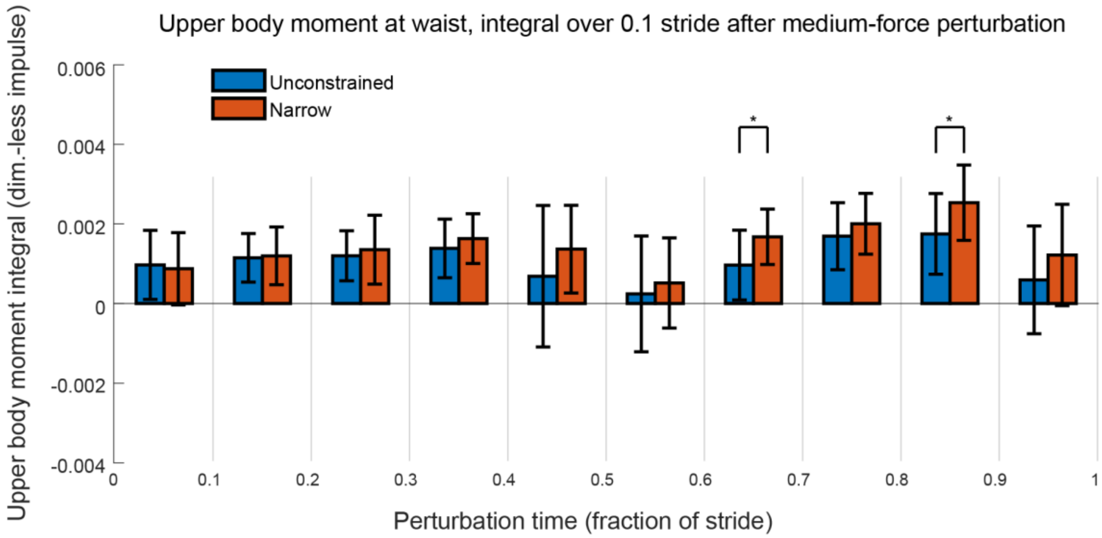

Figure 4.13 Mean and standard deviations of upper body moment that produces a reaction that pushes against COM perturbation, integrated over 0.1 strides after low-force perturbation, grouped for perturbations at every 0.1 periods of stride. Stars (*) denote significant differences at $\alpha = 0.05$. Periods of stance-side perturbations (0.6-0.7 and 0.8-0.9) show significantly higher moment as a result of step placement constraint. Stance-side perturbations also show a significant difference when evaluated as a whole with p = 0.0036, unconstrained: mean +/- s.d. 0.0010 +/- 0.0012; narrow: 0.0016 +/- 0.0012

4.3.5 Arm away from perturbation increased inertial strategy contribution after low force perturbations with narrowest stepping constraints

The arm opposite the perturbation direction seemed to increase movement in accordance with producing an inertial strategy moment in response to the narrowest constraint level at low perturbation force level (Figure 4.14, 4.15). Stance-side perturbations showed the most significant increase in the far shoulder moment impulse for 0.1 strides after perturbation, with a mean of 0.3414e-4 towards abduction for the unconstrained condition (s.d. 2.5838e-4, not significantly different from 0 with p = 0.2322) and 1.4009e-4 towards abduction for the narrowest condition (s.d. 2.7124e-4, significantly different from 0 with p = 7.008e-6), with a t-test difference p-value of 0.0102. This seemed to be driven by significant increases in moments after early- to mid-stance perturbations (Figure 4.11), with the moment contribution decreasing towards late-stance perturbations with coefficient -0.0010 (95% ci -0.0016 to -0.0005, p = 0.00030). Meanwhile, the arm on the side of the perturbation did not seem to significantly contribute via rotation to inertial balance strategy, and this behavior did not change with constraints. There was not a significant change in arm contributions for medium force perturbations between unconstrained and narrow conditions (Stance side: unconstrained: mean +/- s.d. 0.8889e-4 +/- 0.3213e-4; narrow: 0.9457e-4 +/- 0.3353e-4). It is possible that the narrow condition did not create enough restrictions to necessitate recruiting more arm contributions for medium perturbation.

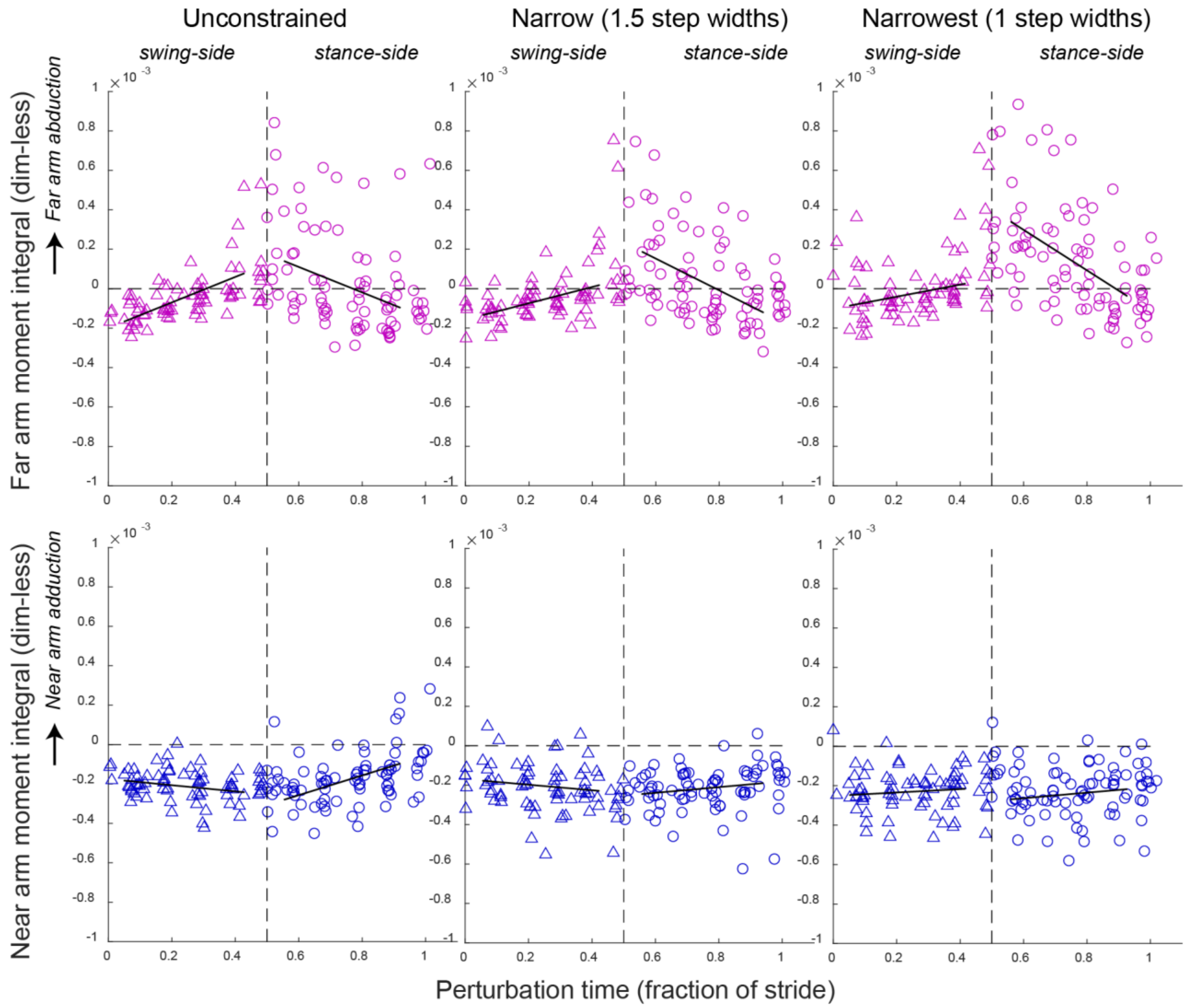

Figure 4.14 Shoulder moment impulse of the far arm opposite from the perturbation direction (top) and near arm on the side of perturbation direction (bottom) for low force perturbations.

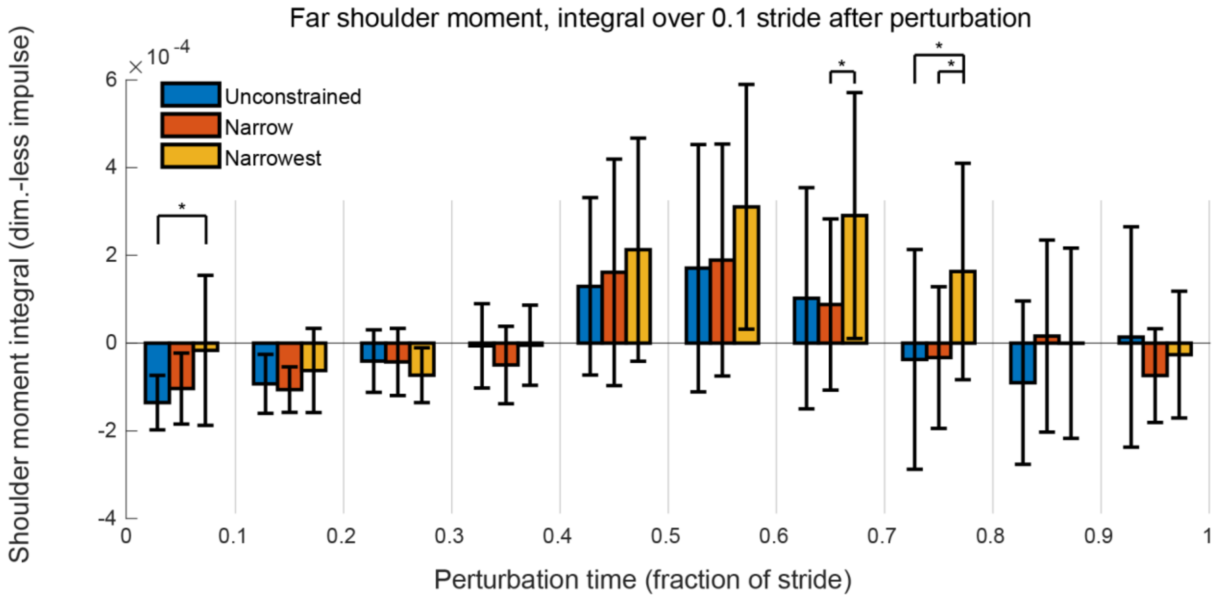

Figure 4.15 Mean and standard deviations of far shoulder moment from Figure 4.12 grouped for every 0.1 stride.

4.3.6 Swing hip moment impulse reflects step placement priority and constraint effects

The swing hip moment during swing does not reflect movement in accordance with inertial balance strategy (Figure 4.16). Notably, it does significantly change as a result of increasing constraint on swing-side perturbations, with swing hip moment becoming less negative in abduction (Table 4.4). Medium force perturbations also produced similar results. This potentially reflects a decrease in moving the swing foot laterally as the constraints limit the lateral placement allowed on swing-side perturbations the most.

Table 4.4 Swing hip moment impulses during swing phase, mean +/- standard deviation, across lateral restraint conditions for swing-side and stance-side perturbations. Results are significantly different between swing-side and stance-side at all condition levels and between unconstrained (UNC) and both narrow (NAR) and narrowest (NAR+) conditions on swing-side for α = 0.05.

	SWING-SIDE		STANCE-SIDE	
FORCE	*mean*	*sd*	*mean*	*sd*
UNC	-0.00138	0.00053	-0.00172	0.00088
NAR	-0.00113	0.00067	-0.00168	0.00092
NAR+	-0.00102	0.00064	-0.00151	0.00104

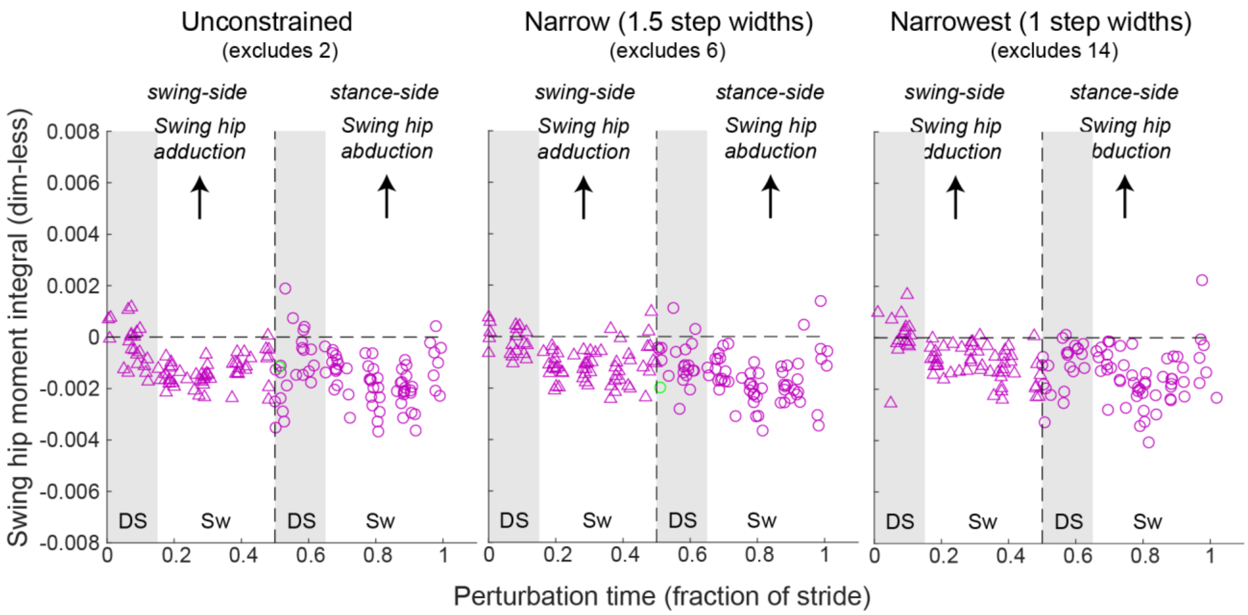

Figure 4.16 Swing hip moment impulse after low force perturbations for unconstrained (left), narrow (middle), and narrowest (right) step restriction conditions. Shaded double support phase (DS) highlights approximately when the swing foot is still on the ground and thus doesn't necessarily desire to move according to inertial balance strategy, while swing phase (Sw) highlights approximately when the swing foot is off the ground and we would expect a positive moment integral if the swing leg is used in accordance with modeled inertial balance strategy.

4.4 Discussion

We observed multiple changes in responses to perturbations as a result of walking constraints. When foot placement is constrained, we found that humans increased performance of inertial balance strategies that can contribute to balance. In particular, the upper body, consisting of the torso, head, and arms, tends to rotate about the pelvis towards the perturbation direction, creating a reaction moment that pushes the COM away from the perturbation. We next discuss how these compensations contribute to overall maintenance of balance during walking.

Step placement was modulated based on perturbation timing and step constraints. For most perturbations, we observed substantial foot placement in the direction of the perturbation, consistent with previous literature (Hof et al., 2010; Wang and Srinivasan, 2014). Step placement was used the most in the unconstrained condition, and particularly after perturbations early in the step. As expected, there was significantly less step placement for late perturbations (Figure 4.3 top), apparently due to the limited time available to actively respond and move the swing foot before ground contact. As a result, compensatory balance control may be completed in the following step. A late stance-side perturbation may thus be regarded as fairly similar to an early swing-side perturbation (Figure 4.3, compare perturbation times near 0 and 1 stride time). And on narrower paths where step placement is more limited, the actions were scaled downward, particularly for early perturbations (Figure 4.3, compare narrower path constraints). Of particular interest are the cases where there was less step placement, thus suggesting the need for additional compensations to restore balance. Perturbations late in a step may not to be fully compensated by foot placement, and would be expected to be accompanied by greater use of inertial strategies, and/or an additional foot placement adjustment in the following step.

We observed the greatest evidence of inertial strategies for stance-side perturbations. The upper body moments were of higher magnitude during stance-side (and late swing-side) perturbations, and higher still on narrower paths (Figures 4.12, 4.13). This is suggestive of a lateral hip strategy to quickly move the COM prior to the next step. Although this is not necessarily sufficient to fully restore balance, it may delay falling enough to allow the next foot placement. The lateral hip strategy is also a fast initial action to move the COM, but then requires the trunk to eventually be restored to upright. This appears to be similar to sagittal plane posture, where the hip strategy is quite fast, but restoration to upright much slower (Park et al., 2004).

Additional insight may be gained from the stance hip moment, which accompanied upper body moment for some perturbations. An overall inertial strategy is not confined to moving the upper body, and so stance hip moment is indicative of other actions to help rotate the entire body other than the stance leg towards the perturbation. An overall inertial strategy is not confined to moving the upper body, and so stance hip moment is indicative of other actions to help rotate the entire body other than the stance leg towards the perturbation. We observed greatest use of hip moment entering into and during early stance-side (40-60% of stride) and swing-side perturbations (90-100% and 0-10% of stride). These perturbations occur with either limited time to adjust the landing foot or after the foot has landed and with long wait before the next footfall, and may thus require greater inertial compensation. To interpret these actions, it may be instructive to consider the opposite possibility of stance hip moment. This could rotate the pelvis and rest of body away from the perturbation, placing the swing leg closer to the ground for a more immediate step. If quicker stepping were the only priority, we would expect upper body rotation away from the perturbation, along with a briefer and more lateral step (Figure 4.17). This is not what we observed, because the greatest hip moments were toward the perturbation, tending to slow the step and allow for less lateral step placement. Although hip and upper body moments were both consistent with inertial compensation, they did not occur in equal magnitudes and at equal times. The body’s many degrees of freedom appear to act inertially, but in a complex combination.

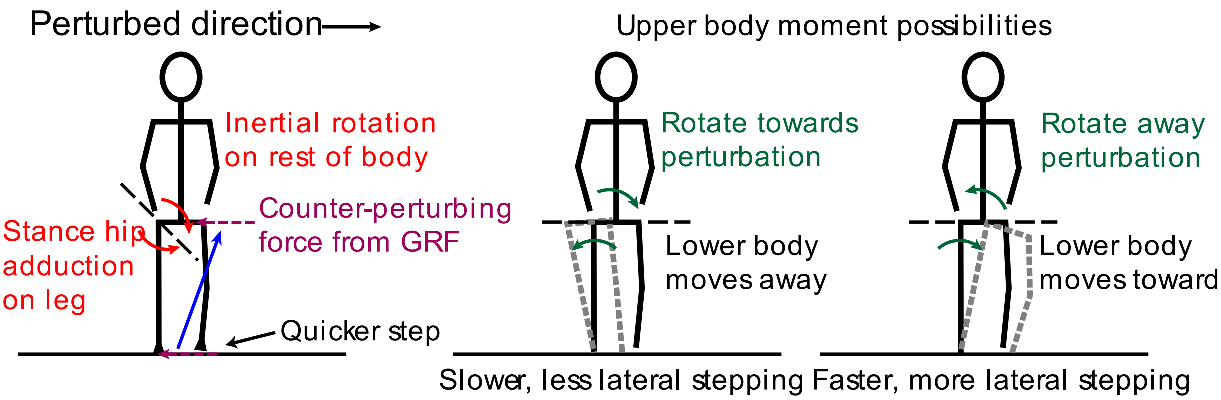

Figure 4.17 Schematic showing two possibilities for upper body rotation moment in response to swing side perturbations.

The narrow step constraints tended to increase use of inertial strategies. The constraints had unequal effect depending on perturbation direction, causing greatest inertial compensation for stance-side perturbations. This may be due to the relative ease by which swing-side perturbations may be compensated, with a relatively fast (due to briefer step duration, Fig. 4.3 bottom) and modest placement of the foot laterally. The narrow path constraint caused a modest reduction in step placement, but not obviously enough to necessitate large inertial contributions. For stance-side perturbations, the constraint actually has less effect, because it only limits the ability for the swing foot to cross-over medially beyond the stance foot. Nevertheless, we still found foot placement to be reduced with the constraint (Fig. 4.3, compare narrow path constraints) and inertial strategies to be increased. Stance-side perturbations may be inherently more challenging, in part because they tend to increase step duration and allow for more time to fall, thus leading to proportionally larger step placement deviations. When medial foot placement was restricted for stance-side perturbations, subjects tended to exert greater upper body moment, and at times greater hip moment as well (for earlier perturbations, 50-60% of stride). It is also possible that late stance-side perturbations will transition into a swing-side response in the following step (Figure 4.4, compare late stance placement during second step), which is particularly restricted by constraint, thus necessitating more inertial compensation.

We have observed the combination of foot placement and inertial compensation with a few simple indicators. We summarized overall effects into foot placement distance for step placement, and upper body and hip moments for inertial strategies. We found some evidence through shoulder moments that the arm away from perturbation may move relative to the trunk in a manner consistent with inertial compensations, particularly in the narrowest constraint condition. However the magnitude of the moment coupled with its greater distance from the ground suggests that it has a comparatively small impact relative to the overall movement of the upper body via the trunk and hip. We found no evidence to suggest the swing leg moves consistently with inertial compensation, and instead found evidence that the swing leg prioritizes foot placement even under stepping constraints. We do not discount the possibility of inertial actions beyond the ones we have identified. However, we expect their contributions may be relatively small compared to the inertia available at the trunk.

Another limitation is that we implemented relatively crude perturbations and constraints on walking. Humans can encounter quite complex challenges to their balance, such as slippery surfaces, obstacles to step on or over, and puddles to avoid. These are not necessarily well modeled by lateral perturbations and narrow path constraints. Rather than attempt to replicate the challenges faithfully, we attempted to apply simple perturbations to deterministically affect balance, and path constraints to limit and alter the compensations available to the person. Even with this simplification, we observed some unusual responses (e.g., stutter steps where the swing foot was quickly placed medially and behind the stance foot instead of in front, possibly to respond more quickly to stance-side perturbations). These occurred too infrequently to characterize well, and so we anecdotally acknowledge that humans perform compensations in a greater range than described here.

It would be ideal to establish an operational model of balance during walking. We have previously shown how foot placement (Kuo, 1999), trunk motion (Chapter 2), and even steering (Rebula et al., 2017) can contribute to balance. The human can employ many more than these compensations, and so we find it helpful to broadly summarize control responses here. The challenge for modeling is that many combinations of control strategies can potentially be used together to maintain balance. The present data may inform what combinations humans tend to use, which might ultimately be determined by a variety of factors such as energy expenditure, maintenance of body posture, and muscle strength. By first considering how humans combine foot placement and inertial compensations, the results may eventually inform an integrative model with multiple options that act together similar to humans.

Humans use multiple balance strategies during walking. The simplest and most direct is foot placement (Bauby and Kuo, 2000; Hof et al., 2010; Kuo, 1999), which appears to be employed in all cases where time and space allow. There are also indirect, inertial compensations that use the trunk and other body segments to act quickly, prior to the next step. This may be regarded as a lateral hip strategy, analogous to the sagittal plane hip strategy observed in standing posture control. The strategies appear not be organized in a strict hierarchy, but rather in combination that is modulated dependent on the degree and direction of perturbation. Inertial strategies are most evident when balance is challenged the most. Humans seem to maintain balance with a complex repertoire of possible actions, the simplest of which include direct step placement and indirect inertial compensation.

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Chapter 5 Conclusion

In this work I explored the mechanisms for lateral balance control during human walking. Using a simple dynamical model of human movement, I made predictions about how stepping and inertial balance strategies could be used together during walking. I then examined the validity of those predictions through a perturbation study on human subjects during treadmill walking. To further understand trade-offs between stepping strategies and inertial strategies, I conducted a perturbation study with varying constraints on stepping and observed corresponding stepping adaptations along with changes in inertial balance contributions. Here, I will summarize the contributions of this work and potential implications.

I created a simple dynamical model to simulate human walking with the potential for upper body dynamics and applied straightforward controls to predict how balance control could be performed. There were two types of controllers designed, based on control systems principles for small perturbations, and another selected ad hoc to deal with larger perturbations. In the first controller, I treated heelstrike states as Poincare sections to model step-to-step dynamics as a discrete time system, and linearized the system to allow for standard pole-placement control design to inform how a mix of discrete (e.g., final step placement) and continuous time (e.g., inertial trunk rotation) controls should respond to feedback. This approach allowed for simpler interpretation of how the feedback controls could be coupled. It was also used to make some predictions about control inputs in response to continuous time disturbances between heelstrike events, by integrating a post-disturbance state in the middle of a step backwards in time to a disturbed heelstrike state that could be input to the controller. A drawback of this approach is that it yields only local stability for small deviations about the nominal gait.

The second controller I implemented was a stepping control based on previous experimental observations about human behavior and the XCOM. I used optimization to determine the amount of foot placement needed to counteract a perturbation, and place the foot laterally according to the XCOM. An advantage of this approach is that it is able to produce large step placements that compensate for substantial perturbations that the local controller cannot. However, a drawback is that there is no formal or principled means to verify stability, other than the note that lateral falling has been prevented. Nevertheless, this approach provided a starting point for a parameter study how inertial mechanisms could influence balance around the stepping strategy.

My approach differs from more complex computational modeling and control approaches. While such methods may be able to more accurately and optimally simulate human walking and balancing behaviors, the additional degrees of freedom and states would not necessarily improve intuitive understanding about how stepping and inertial balance strategies can be coupled. As demonstrated previously (Kuo, 1999), there are methods to design local controllers that stabilize within a single step, but it is challenging to design controllers that operate across both single and double support phases, and no solution using control system principles. My approach is to adopt relatively simple, once-per-step commands for foot placement, trunk motion, and swing leg abduction, and to analyze their effects on balance. It may also be helpful to have a more intuitive understanding simple models when designing or working with robots or assistive devices that have more limited degrees of freedom for control.

I then tested my predictions using a custom-built perturbation system for human subjects walking on a treadmill focused on minimizing influence on balance mechanisms that might be present during normal walking. The system notably uses force-feedback control to allow for normal lateral movement, and features a harness that allow for free arm swing while still applying lateral perturbations at the pelvis near the body center of mass. While the expected contributions of arm movement to inertial balance was relatively small, the amount of movement can be significant and may have other implications for normal walking gait (Collins et al., 2009; Gholizadeh et al., 2019). This also allowed us to quantify the behavior of the arms as a result of force perturbations at the hip in contrast to many other previous studies.

I conducted a rigorous examination of human balance response to lateral perturbations over two studies (Chapters 3 and 4). I attempted to quantify contributions from step placement, ankle strategy, and various forms of inertial strategy including the stance hip, trunk, arms, and swing leg. Perturbations were performed at various levels and at various times through the gait cycle to tease out possible force- and phase-dependent behaviors. The resulting analysis showed notable potential inertial contributions from the hip, trunk, and arms at particular phases of the gait cycle, as well as a record of the step response. Using models predictions, I offered potential explanations for why inertial contributions are sensible in relation to stepping strategies at those phases. I contend that the results demonstrated that inertial balance strategies are important to lateral balance during normal walking phases when effective step placement will take a comparatively longer time. This is highlighted by perturbations towards the leg in stance that requires long crossover movement of the swing leg, and by perturbations near the end of a step where the time remaining is insufficient to place the step at the ideal location.

To further confirm this result, the second study directly applied varying levels restriction on lateral step placement while human subjects experienced perturbations. Indeed, the results showed that even small restrictions on step placement at the right time and perturbation level can significantly increase use of inertial balance strategies. This is in contrast to a possible expectation that inertial strategies are insignificant unless step placement is severely constrained, such as walking on a tight-rope or narrow beam. The conclusion instead may be that humans are actively controlling both stepping strategies and inertial strategies together for balance, and while stepping is dominant during walking, inertial strategies are employed whenever stepping may fall short.

This work could have implications for engineering and clinical research applications. For engineering, bipedal robots and assistive devices may seek increased coupled control of inertial strategies and stepping, or consider different stepping responses when inertial control is unavailable. For clinical research, quantification of inertial balance abilities could form a more complete understanding of an individual's walking stability, and could reveal additional reasons for balance failures and areas to target for rehabilitation. The results of this work should help dispel the notion that balance during walking is a job for stepping alone.

5.1 References

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