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Strategies for the Control of Balance During Locomotion

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The neural control of balance during locomotion is currently not well understood, even in the light of considerable advances in research on balance during standing. In this paper, we lay out the control problem for this task and present a list of different strategies available to the central nervous system to solve this problem. We discuss the biomechanics of the walking body, using a simplified model that iteratively gains degrees of freedom and complexity. Each addition allows for different control strategies, which we introduce in turn: foot placement shift, ankle strategy, hip strategy, and push-off modulation. The dynamics of the biomechanical system are discussed using the phase space representation, which allows illustrating the mechanical effect of the different control mechanisms. This also enables us to demonstrate the effects of common general stability strategies, such as increasing step width and cadence.

Keywords: stability, inverted pendulum, feedback control

Being able to move from one place to another is an essential part of being an animal. Moving in a goal-directed way is a complex problem, and locomotion is no exception. It is especially hard for two-legged walkers such as us humans, because our center of mass (CoM) is relatively high above our base of support (BoS). Furthermore, our base of support is comparatively small, being determined by only two feet rather than four. Our brains solve this complex problem with deceiving ease: walking is a mundane activity that rarely receives conscious attention. Nonetheless, failures of this sophisticated neural control system happen, which causes humans to fall. In recent years, falls and the resulting injuries have become a major public health concern.

Maintenance of balance is a neural control problem. The standing or walking human body is mechanically unstable. It will usually fall over after a short time, unless actively stabilized by activating muscles in a controlled, goal-directed manner. This means that the central nervous system (CNS) has to detect threats to upright balance and then generate an appropriate motor action to counter that threat. Such a mapping of sensory input to motor output is referred to as a control law, and finding a suitable mapping is referred to as the control problem. Detection of threats to balance from the available sensory data, on the other hand, is referred to as the estimation problem. In order to understand how falls happen, we need to understand the structure of the neural controller for maintaining balance, which will allow us to pinpoint failures of balance control to a specific part of the system.

Both the estimation and the control problems of balance have received a lot of attention in standing. Biomechanically, standing is a relatively simple problem, because it can be mostly described by a single degree of freedom (DoF) at the ankle joint. Even considering the presence of substantial movement at other joints such as knees and hips in quiet stance (Hsu, Scholz, Schöner, Jeka, & Kiemel, 2007; Pinter, van Swigchem, Van Soest, & Rozendaal, 2008), modeling work has indicated that the high-level control problem can be reduced to mapping sensory data to a one-dimensional output related to the CoM, while the coordination between the multiple different degrees of freedom can be realized by predefined modules similar to muscle synergies (Reimann and Schöner, 2017). Peterka (2002) has provided a detailed model of the neural control system for balance in quiet stance, describing how different sensory signals are mapped to motor output in the form of torques around the ankle joint.

In walking, there are many more relevant degrees of freedom than in standing, which makes the problem much more complex. Furthermore, the role any given degree of freedom plays for the system varies greatly during the gait cycle. While standing can be well approximated by a linear system, the walking human body is at best locally linear in some configurations, and highly nonlinear at other critical points. The varying abundance of degrees of freedom implies that the CNS has multiple different options available to affect the body in space and solve the control problem. So where the control problem in quiet standing is a many-to-one mapping that is invariant in time, the control problem in walking requires a many-to-many mapping that is highly dependent on the phase of the gait cycle (Forbes et al., 2017; Yang and Stein, 1990).

In this paper, we approach this problem by describing the different motor actions available to the CNS to maintain balance during locomotion. For each of these different actions, or strategies, we briefly review the available evidence for humans actually utilizing them from the literature. We focus mainly on the medial-lateral direction of balance, because it is mechanically less stable in walking than the anterior-posterior direction and thus requires more active control (Bauby and Kuo, 2000).

The Control Problem

The behavioral goal of balance is simply to avoid falls. In upright stance, this translates into a control problem in a straightforward manner: keep the body CoM over the base of support. This has allowed modeling the control of quiet, upright stance with a set of simple feedback control laws, where a deviation from a set point, detected by one or multiple sensor signals, is mapped onto a counterforce that brings the CoM to the set point (Mergner, Maurer, & Peterka, 2003; Peterka, 2002). This counterforce is usually assumed to be generated by the ankle musculature. A

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slightly more sophisticated version of the model that has emerged in recent years is to consider not the raw CoM, but the extrapolated center of mass (XCOM), which is the CoM position plus its weighted velocity (Hof, Gazendam, & Sinke, 2005). When such a control scheme fails, however, the XCoM leaving the BoS does not necessarily lead to a fall. Rather, it implies that the standing human has to take a step to maintain balance. Having to take a step is thus used as a simple and straightforward criterion for loss of balance in standing.

Walking can be characterized as a sequence of controlled falls (Rosenbaum, 2010). We take steps continuously and the body CoM is regularly out of the BoS during normal gait, so this cannot be used as a criterion for loss of balance. There are several different criteria that researchers have used to quantify how well humans maintain balance during walking, ranging from simple variability measures of important variables (Winter, Patla, & Frank, 1990) to sophisticated techniques such as Lyapunov-exponent (Bruijn, Bregman, Meijer, Beek, & van Dieen, 2012) and entropy (Haran & Keshner, 2009). From a control perspective, the question is how the CNS detects threats to balance. One possibility is that the CNS predicts the outcome of the motor actions and then compares the available sensory signals against the expectation and treats deviations as threats to balance. We will avoid the discussion of this sensory component of balance control here and simply assume that the CNS is somehow capable of detecting a threat to balance in the form of an undesired movement of the CoM. To correct this CoM movement, a motor action has to be taken to accelerate the CoM in the opposite direction—or, rather, to decelerate the CoM—thus stopping the movement.

In the following sections, we will discuss several different motor actions available to the CoM to achieve this goal. To facilitate this discussion of how the CoM movement can be corrected in a goal-directed way, it is useful to understand how the CoM moves in the absence of any control action. In the simplest biomechanical approximation, the walking body can be modeled as a point mass on a massless leg with a single ground contact point (Kajita et al., 2003). The movement of that point in the horizontal plane is entirely defined by the velocity of the CoM and its distance from the ground contact point, or center of pressure (CoP). The relationship between these two points is that the horizontal acceleration of the CoM is proportional to the displacement between the CoM and the CoP (Hof et al., 2005). We will restrict our description to the medial-lateral direction here, but the anterior-posterior direction is analogous.

The dynamics of this system are illustrated by a phase plot in Figure 1. In this plot, the horizontal axis represents the position of the (medial-lateral) CoM relative to the CoP, and the vertical axis the weighted CoM velocity. The orange dot in the center represents the special case in which the CoM is located directly above the CoP and has zero velocity—that is, the person is standing still. Each quadrant corresponds to different types of CoM movement relative to the CoP, as indicated by the stick figures on the sides. In the upper left quadrant, the CoM is to the left of the CoP and moving right, toward the CoP, whereas in the lower left quadrant it is moving away. All movement of the CoM state flows along the blue lines, called flow lines or orbits, as indicated by the arrows. Any CoM state in the upper left quadrant is moving right toward the CoP, but whether it will reach it and cross over to the right side depends upon the velocity relative to the remaining distance. If the weighted velocity is larger than the remaining distance, the CoM will reach the CoP and continue moving to the right, passing into the upper right quadrant. If it is smaller, gravity will decelerate the CoM until it stops and then reverts its movement direction, passing into the lower left quadrant. Note that the flow lines of almost all states eventually leave the diagram, which means that the body will fall. The only exceptions to this are the special cases given by the purple diagonal, where the weighted velocity exactly matches the remaining distance, so the CoM will approach the CoP and come to a stop directly above it.

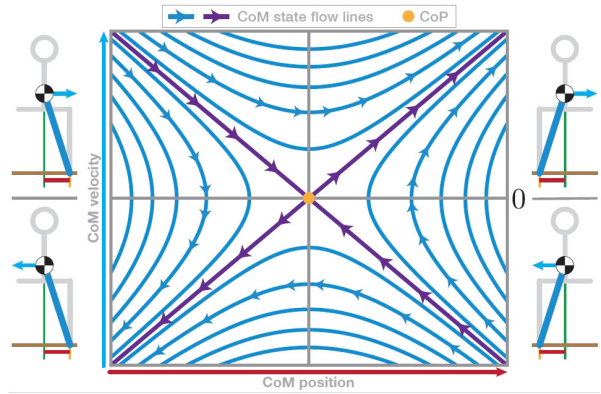

Figure 1 — Phase space representation of the simple inverted pendulum model dynamics in the medial-lateral direction. The horizontal axis represents the lateral position of the CoM relative to the CoP, the vertical axis the weighted CoM velocity. Each point in the plane corresponds to a kinematic state of the lateral CoM—that is, to a combination of position and velocity. The orange dot in the center is the special case in which the CoM is directly above the CoP and not moving. The blue flow lines indicate how the CoM state flows through this space under gravity, as indicated by the arrows. The purple diagonals are special case flow lines. The stick figures on the sides illustrate the relationship between the CoP and the CoM state, indicating whether the CoM is right or left of the CoP and moving toward or away from it.

Foot Placement

To avoid falling over and hitting the ground, the CoM has to be kept within the base of support on average over time. The most straightforward way to modify the CoP is to take a step. In our simplified model, we understand a step as an instantaneous shift of the CoP to a new location. The green lines in Figure 2 give an example of average CoM paths through the phase space when a person is taking regular steps. Assume that the CoP location is the left green dot, and that the CoM state starts out at the large green arrow. The CoM moves along the left half of the green flow line. At first, it moves toward the left, but gravity accelerates it toward the right. Eventually, it reaches zero velocity and inverts the movement direction, moving toward the right and increasing velocity. At this point, a step is taken, represented by the cusp in the green cycle. The CoP is instantaneously shifted to a new location on the right side of the phase space. Rather than falling *away* from the old CoP on the left, the body is now falling *toward* the new CoP on the right, so gravity now decelerates the CoM, and velocity is decreasing rather than increasing, forming the cusp. This leftward acceleration continues until the velocity reaches zero again, the movement direction reverts to the left, and another step is taken back to the left, which is represented by the second, lower cusp in the green cycle. At that point, a step is taken, shifting the CoP to the second green point on the right. This reverses the acceleration from gravity, and the velocity starts decreasing again

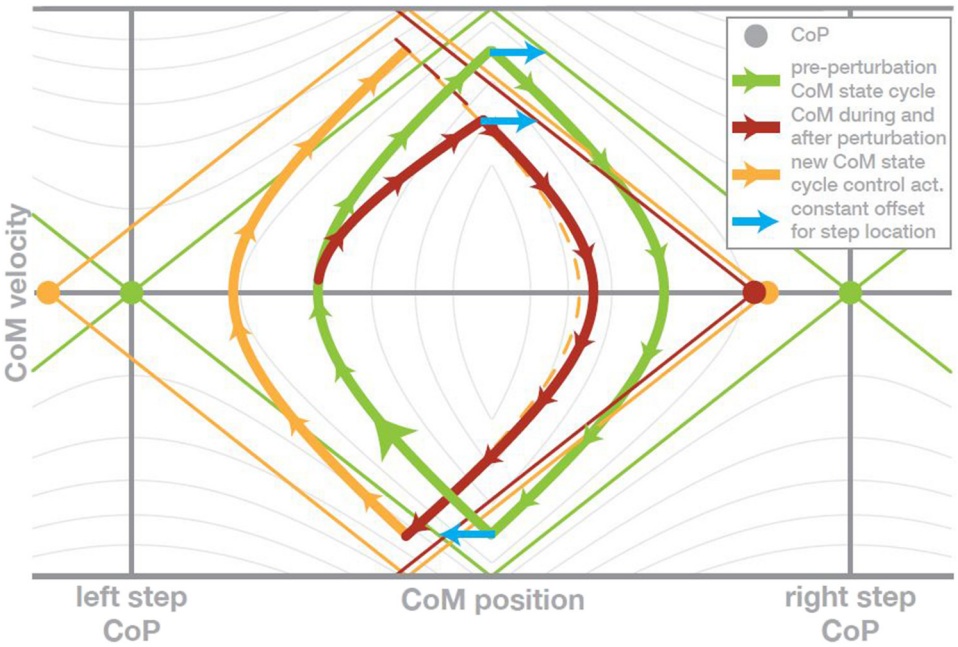

Figure 2 — Effects of the foot placement strategy in phase space. The colored dots on the left and right represent different CoP locations. Each CoP location defines a CoM flow through the phase space, as indicated by the light gray flow lines illustrating the flow determined by the green CoP locations. When a step is taken, the CoP switches from a point on the left to a point on the right, or vice versa. The location of the new step CoP is defined so that the new diagonal intersects the CoM state shifted by a constant offset as indicated by the blue arrow. The CoM flow changes accordingly, and the flow lines from each side join up in the center. The colored lines represent an example sequence of a perturbation followed by a control action. The green cycle shows the unperturbed movement, where the cusps represent (instantaneous) steps where the flow switches. The red line shows how the state trails off the green cycle due to a perturbation. The red dot is the step response to this perturbation, shifted leftward. The orange dots and lines represent the new stable cycle of the CoM after one successfully balances against the perturbation. Flow lines for the systems defined by the red and orange CoP locations are not shown.

until the second cusp is reached and a second step taken, which shifts the CoP back to the left green dot. Note that the cusp is sharp because we made the simplifying assumption that steps are instantaneous. In a more realistic model, with two legs and a double stance phase, these would be smooth.

Townsend (1985) has shown that this simple action of regularly shifting the CoP to a new location by taking a step is already sufficient to maintain upright balance in most situations. Hof et al. (2005) proposed a simple control scheme using only this mechanism, called *constant offset control*. In this control scheme, steps are taken at fixed intervals, and the new location of the CoP is set to the XCoM plus a constant offset. Setting the CoP exactly onto the XCoM would mean moving on the diagonals, which is shown in purple in Figure 1. This is not desirable, because small fluctuations will always perturb the state away from the diagonal. But the direction of this perturbation cannot be predicted, so the CoM could end up either under- or overshooting the CoP. Overshooting the CoP would mean that the next foot placement would need to cross over the midline, which is problematic (see the General Strategies section for details). Adding a constant offset adds a safety margin between the CoM state and the diagonal, which reduces the likelihood of such an overshoot. This offset is represented by the blue distance indicators in Figure 2. Instead of placing the new CoP directly onto the XCoM, which would make the diagonal intersect the current state, it is shifted toward the right by a fixed distance. This offset is added in the current movement direction, so in left stance, when the CoM is moving toward the right, the CoP at the new step is placed a bit more to the right than it is necessary. The same constant offset is applied toward the left at the next step, represented by the lower cusp. This safety margin ensures that the CoM state will remain on the same side of the diagonal under small fluctuations.

To understand how this control scheme maintains balance in the long run against perturbations, imagine that the walking body receives a perturbation in the form of a transient push to the left in late left stance. This results in the CoM being located more to the left and moving with lower rightward velocity at the time of the next step, as illustrated by the red path deviating from the green cycle in Figure 2. The new CoP location, represented by the red dot, is then also shifted to the left, with a constant offset from the XCoM (Hof et al., 2005). Since the rightward velocity was lower at the time of the step, the CoM is now on a different flow line, with less leftward acceleration from gravity. This reduced leftward acceleration effectively cancels the leftward push received earlier from the perturbation. At the time of the next step, the medial-lateral velocity is very close to normal again, and the CoM state is now on a cycle that is the same as the unperturbed one, only shifted to the left, as shown in orange in Figure 2. Thus the *constant offset control* scheme stabilizes the size and shape of the cycle, but not its location in absolute coordinates—that is, it allows lateral drift.

The underlying principle behind this control scheme is simple: when we sense our body moving to one side, we step toward that direction to bring the base of support back under the CoM in the long run, similar to maintaining balance when riding a bicycle. In recent years, researchers have found various pieces of evidence that this control scheme of "stepping in the direction of the perceived fall" is used by humans to control their balance during locomotion. Wang and Srinivasan (2014) showed that there is a strong correlation between the CoM position and velocity at midstance and the position of the following foot placement: 80% of the foot placement variability can already be explained by the CoM variability at midstance. Furthermore, the explanatory power of the CoM position and velocity rises relatively sharply up to midstance, but only

slower thereafter, which implies that the CNS decides the next foot placement around that time and then only makes smaller updates in the limited time before the actual step. Foot placement is actively modulated in response to mechanical perturbations at the hip (Hof & Duysens, 2013; Rankin, Buffo, & Dean, 2014) and galvanic perturbation of the vestibular system (Reimann et al., 2017). There is mounting evidence that active modulation of the foot placement exists, which possibly diminishes the notion that increased medial-lateral variation is a "bad thing." Foot placement is only one of many mechanisms contributing to bipedal balance during walking, but understanding the mechanisms that lead to "choosing" an appropriate foot placement location and the mechanisms the CNS uses to actively achieve that location may still prove to be beneficial in rehabilitating populations suffering from poor balance control, such as stroke patients (Dean and Kautz, 2015).

The Lateral Ankle Strategy

We can make the biomechanical model slightly more complex by adding a foot and an ankle joint to the leg. This allows the controller to act on the center of mass at any time, instead of only when taking a step, by applying torques at the ankle joint during the stance phase. Ankle torques are generated by muscles spanning the ankle joint pulling one side of the foot segment and the leg/body segment toward each other. The immediate effect of this pull is that it reduces the ground pressure under the foot on the side of the pull and increases it on the contralateral side. This difference shifts the CoP away from the pull. If the foot is a rigid, flat body in full contact with the ground, the effect of this pull over time is simply to change the roll angle of the body segment toward the direction of the pull. The human foot, however, is neither rigid nor flat, but rather a collection of smaller components in deformable, rounded padding. This segment can roll along the floor to some degree, and the pull from the ankle torque is expected to cause such a lateral roll. This lateral roll would shift the contact region in the direction away from the pull, along with the CoP.

Even with this additional motor action, the mechanical relationship between CoM and CoP captured in the phase portrait still largely holds up. Instead of letting the state run along the flow lines and having to reset the whole phase portrait at discrete points in time given by steps to control its evolution, the ankle torques provide a way to affect the state directly at any time. This takes the form of a small shift of the CoP—that is, the origin of the phase portrait—and the flow lines shift along, while the state of CoM position/velocity remains fixed. This is illustrated in Figure 3. Assume that the body is in left stance on the green flow line. An external perturbation pushes the body toward the right in early stance, as indicated by the red deviating line. Without any control action, the CoM state would follow the new flow line on the red path, leading to a substantial increase in velocity at the end of the step. The lateral ankle strategy allows shifting the CoP rightward, as indicated by the blue arrow, so that the state is back on the old flow line. The new CoP and flow line are indicated by the orange dot and line. Note that the flow lines are completely determined by the CoP location, so shifting the CoP essentially moves the flow lines under the CoM state and thus changes which flow line the CoM is on.

This lateral ankle strategy is conceptually equivalent to the ankle strategy studied in quiet stance. In walking, it has a much smaller functional range than the foot placement strategy, because it is limited by the relatively small size of the foot; consequently, it

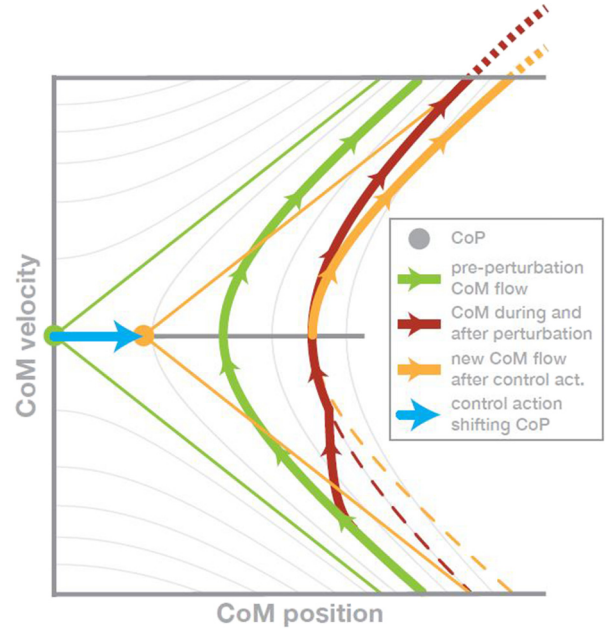

Figure 3 — Effects of the lateral ankle strategy in phase space. The colored dots on the left represent different CoP locations during left stance. Each CoP location defines a CoM flow through the phase space, as indicated by the light grey flow lines. The colored lines illustrate the effect of a perturbation and a control action. The green cycle shows the unperturbed movement of the CoM state along the flow line. The red line shows how the state trails off due to a perturbation and then continues on a new flow line. The blue arrow indicates an instantaneous shift of the CoP under the stance foot, generated by activation of lateral ankle musculature. The orange dot is the new CoP location, and the thick orange line shows the corrected flow after the control action.

has received less attention. Hof, van Bockel, Schoppen, and Postema (2007) first reported evidence of the lateral ankle strategy in an experiment with below-knee amputees. They observed that the CoP of non-amputees had a tendency to "cross over" the line of average CoP trajectories—that is, CoP trajectories that started a single stance phase more to the right than the average would tend to cross over the midline and end the stance phase more to the left, and vice versa. Amputees showed a similar effect in their healthy legs, but not in their amputated legs. This can be interpreted as an active control mechanism in the sense described above.

Additional evidence that humans use the lateral ankle strategy to maintain balance during locomotion comes from a study by Reimann et al. (2017). They applied galvanic vestibular stimulation during gait initiation to perturb the sensory system and induce the sensation of falling to one side (Bent, McFadyen, Merkley, Kennedy, & Inglis, 2000). In response to this stimulus, the subjects' CoP under the stance foot shifted laterally in the direction of the perceived fall, starting ~250 ms after stimulus onset, well before a corrective step is initiated. The authors also developed a neural control model using both the foot placement strategy and the lateral ankle strategy and fitted the model parameters to the experimental data. This allowed them to explore the relative importance of the two mechanisms. Removing the lateral ankle strategy resulted in an increase of foot placement change by 60%. This indicates that although not essential for balance control, the lateral ankle strategy complements the foot placement strategy, which can reduce overall foot placement variability and thus step width, which in turn may reduce the overall metabolic cost of walking.

Hip Strategy

The next addition to the biomechanical model is a hip joint. Similarly to the ankle joint, the hip joint allows application of torques to pull on the trunk, which contains most of the mass. Our assumption is that the single leg can be instantly moved to a new position, so the foot placement strategy remains unchanged. The ankle strategy, however, is complicated by adding a second internal degree of freedom to the body, due to interaction torques from the inertial properties. In the two-DoF model, an ankle torque that pulls clockwise across the ankle has an effect on *both* joints. The ankle is accelerated clockwise, but the hip joint is accelerated counterclockwise due to the inertial interaction torque. In order to accelerate only the ankle joint, the CNS has to account for this interaction torque by applying an active hip torque in the opposite direction (Winter, 1995). During normal walking, most of the modulation of the hip musculature is thus to stabilize against these interaction torques, both from active muscle torques at other joints and from changes in ground reaction forces, particularly at heel-strike and push-off. But just as there are combinations of ankle and hip torques that accelerate only the ankle joint, there are other combinations that accelerate only the hip joint. So while the underlying biomechanics are more complicated, this two-DoF model still allows us to make use of the ankle strategy in the same way as before, and add the hip strategy as an additional mechanism to move the CoM and control balance.

In practice, the range of possible CoM accelerations provided by the hip strategy is relatively small (Popovic, Goswami, & Herr, 2005). This is mostly because, in addition to the lateral acceleration, the hip strategy results in a sizeable angular acceleration of the trunk segment. The resulting angular motion of the trunk segment has to be stopped before the rotation becomes too large and reaches the limits of joint motion. The hip strategy is thus, in a way, a temporary loan of kinetic energy that has to be repaid relatively soon, limiting its long-term utility. In robotics, researchers have avoided this limitation by designing the hip joint as a flywheel, which can be accelerated to any steady-state angular velocity and effectively removes the joint range limits (Pratt, Carff, Drakunov, & Goswami, 2006). Although humans cannot do that, such a loan can be useful to cover a short-term need in some situations. On a narrow balance beam, for example, both the foot placement and lateral ankle strategies are severely limited. In such a situation, humans have to utilize even the small accelerations provided by the hip strategy to maintain balance, but they tend to end up with awkwardly rotated upper bodies as a result.

The hip strategy is well identified and an effective strategy in quiet standing (Kuo and Zajac, 1993; Winter, 1995), and is more recently a justifiable strategy to maintain balance during walking. Roden-Reynolds, Walker, Wasserman, and Dean (2015) used vibratory perturbations on the gluteus medius muscles to elicit a balance response, but they only describe resulting changes in foot placement, not hip joint angle or muscle activation. The bias toward foot placement as the primary balance mechanism may be hindering exposure of other mechanisms that play an important role in balance during walking.

Push-Off Strategy

A final addition to our biomechanical model is the second leg and a second degree of freedom at each ankle. This allows us to take a proper step, instead of the physiologically implausible instantaneous change of the contact point of a single leg. In normal walking, we now have a double stance phase, with two points of ground contact, instead of one. This second point of contact allows the CNS to modulate the relative force between the two legs during double stance by shifting weight between them. This is similar to shifting weight laterally in standing, which is a very effective balance control mechanism and a main reason why balance control in standing is substantially easier in the medial-lateral direction than in the anterior-posterior direction (Kuo, 1999). In the double stance period in walking, the leading leg is far in front of the trailing leg, but also slightly lateral, so shifting weight between the legs still affects lateral balance. In this configuration, an effective way to shift weight is to modulate the plantar-dorsiflexion angle of the trailing leg ankle. In other words, this changes how the trailing leg pushes off against the ground. Relative to the average, a stronger push-off with the left leg will accelerate the CoM forward and to the right, and a weaker push-off will accelerate the CoM backward and to the left.

Kim and Collins (2013) used a model to show that this mechanism can be used for balance control. In later work, they designed an ankle prosthesis controller based on this principle and showed that humans using this controller for an emulated prosthesis spent less metabolic energy during walking compared to a similar control approach without the push-off modulation (Kim and Collins, 2015). While there is no hard evidence that using such a mechanism for balance control can be beneficial, work by Vlutters, van Asseldonk, and van der Kooij (2016) suggests that the push-off mechanism could contribute to foot placement, at least in the anterior-posterior direction.

General Strategies

All strategies described in the previous sections are control strategies in the sense of a feedback law: a deviation from a desired state is detected by the sensory system and mapped to a motor action that brings the system back to the desired state—that is, when the body falls to the right, the CNS uses motor actions to accelerate it back to the left. There are other strategies available to the CNS that act in different ways. Instead of generating a directed motor response according to a feedback law, the CNS can increase the passive stability characteristics of the system. One well-established way to do this is to increase the average step width over multiple steps. The biomechanical rationale for this lies in the asymmetry between swing foot and stance foot at each step. The lateral CoM position oscillates between the legs during walking (see Figure 2). During early swing, the CoM is moving toward the stance foot. Then it is decelerated by gravity, reaching zero lateral velocity around mid-stance, and then moving toward the swing foot in late swing. In general, small deviations from the average trajectory have small effects. When a small perturbation moves the state to another nearby flow line on the phase plot, the system will usually be in a largely similar state after a short amount of time. This is different, however, when the new flow line lies on the other side of the diagonal divide. In such a case, the system will be in a vastly different state after some time. Rather than the velocity reaching zero and the movement direction reverting, the CoM position crosses over the CoP. At this point, gravity accelerates the CoM toward the stance foot rather than away from it. This generates a very large deviation of the CoM from the expected and planned state, which requires a large crossover of the next foot placement. In extreme cases, the stance leg will prohibit the swing leg from crossing over fast enough, leading to a fall.

To avoid this critical situation of the lateral CoM crossing over the CoP location, the state of the system is usually kept at a certain safety distance from the diagonal, so that it will not be pushed across by small perturbations. The size of this safety margin depends on the expected perturbation size, so in situations of a general threat to balance or where larger perturbations might be expected, the CNS might want to increase this safety margin. A simple and straightforward way to achieve this is to increase the average step width. With increased step width, the CoM is generally farther away from the CoP during each step, which means that distance between the CoM and the diagonal in phase space is larger, decreasing the likelihood that a perturbation will push the state across this divide, leading to a crossing of the CoP. The left side of Figure 4 illustrates the effect of increased step to stabilize against random perturbations. The trajectories shown are from simulations of a simple model using constant offset control (Hof et al., 2005) with three different offset parameters generating narrow (red), medium (blue), and wide (orange) average step width. The random perturbations are added in the form of a white noise error term on the foot placement (Reimann et al., 2017). Panel A shows the paths in the phase space, and panel B shows the trajectories of CoP and CoM. While the narrow average step width has smaller step-to-step oscillations, it is more affected by the perturbations in general, with larger overall lateral variability. In two instances, the perturbation leads to cross-over steps, represented by the two red curves in the lower center wedge of the phase space. The difference in overall variability between the medium and wide average step width is relatively small, in contrast, implying that the benefits of wide over medium in terms of balance are negligible. Although the strategy of increasing step width comes at a metabolic cost (Donelan, Kram, & Kuo, 2001), it may reduce the need for the CNS to intervene in actively controlling foot placements (Perry and Srinivasan, 2017).

Another mechanism to increase the general balance characteristics of the system and increase its safety features is to increase the step frequency, or cadence. In general terms, increasing cadence means that the system spends less absolute time simply falling between steps. Both the average accelerations and the resulting peak velocities of the CoM are smaller, meaning that the CNS has more time on average to react when something goes wrong before a critical velocity is reached. From a control perspective, the frequency of foot placement control actions is increased, meaning more opportunities to correct errors over a fixed time span using the foot placement strategy. Again, we can use simulations of the same simple model to illustrate the benefits of increased cadence for balance control. This time, we choose parameters to generate a fast (red), medium (blue), or slow (orange) cadence, but we keep the average step width the same in all three cases. The right side of Figure 4 shows the results of this model, with phase plot in panel C and trajectories in panel D. While all three models have similar average step widths, the slow cadence results both in higher step-to-step and long-term variability compared to the medium and fast cadences. As noted above, the peak velocities decrease with increasing cadence—that is, slower cadences reach higher velocity bands—while faster cadences are relatively

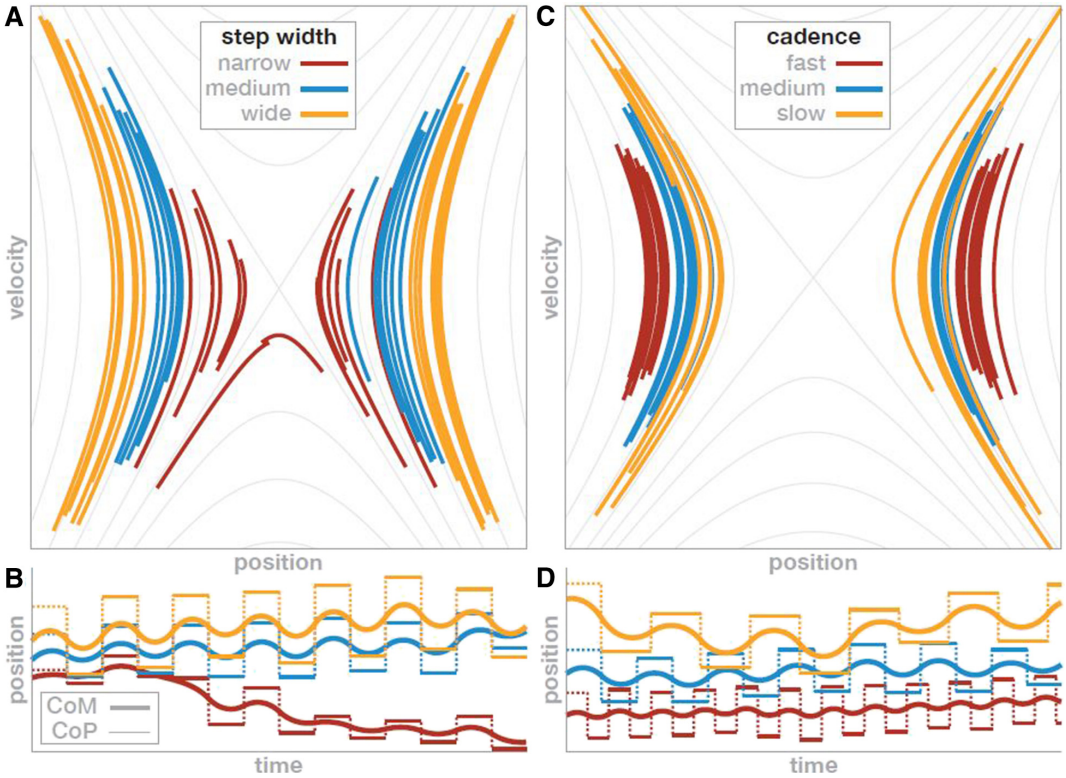

Figure 4 — Effects of different step width and cadence. Data shown are from model simulations with different parameter sets. Panels A and B show different average step widths but constant cadence. Panels C and D show constant average step width but different cadences. The upper graphs show paths of the medial-lateral CoM through phase space relative to the CoP at each step. Since the steps are instantaneous, the state jumps between left and right at each step. Each continuous curve corresponds to one step—that is, the time between two foot placement changes. The lower graphs show the medial-lateral CoP and CoM trajectories from the same simulation data. The jumps of the CoP at each step are indicated by dashed lines here. Between foot placement changes, the CoP is constant.

contained around the central horizontal axis in phase space. We can also see that the average distance from the diagonal increases with the cadence, indicating that all else being equal, faster cadences come with a larger "safety margin." It should be noted that the interaction between cadence and balance has been controversial (Bruijn, van Dieen, Meijer, & Beek, 2009; Dingwell & Marin, 2006; McAndrew, Dingwell, & Wilken, 2010). Intuitively, it might be thought that going slower is more stable. In many other movements, going slower certainly increases the accuracy, which can be understood as being similar to stability (Fitts, 1954). However, stability in the sense of balance during locomotion is different as a control problem than goal-directed movements of the upper extremity, leading to the counterintuitive effect that "faster is better" here. It should be noted that this effect is hard to isolate in human subjects, since cadence is strongly affected by things such as forward velocity, and effects that appeared to be related to cadence or speed have been shown to be rather covariants of other variables, such as leg strength and flexibility (Kang & Dingwell, 2008).

Understanding How the Brain Controls Balance During Walking

We have laid out the problem of balance during locomotion from a control point of view and presented a list of mechanisms by which a walking anthropomorphic body can be controlled. The presentation and explanation of some of these mechanisms borrowed heavily from the field of robotics, which deals with a very similar problem, but from a control engineer's perspective. Rather than studying and trying to understand how an existing system achieves the task of balance control, the designer of a robot is free to make use of whichever mechanism appears to be useful to solve the task. It should be pointed out that this task is very hard. Only very recently have engineers succeeded in designing and programming anthropomorphic robots that are capable of walking dynamically. The kinesiologist, on the other hand, has the luxury of being able to study an existing solution to the same problem, provided by evolution, but faces the problem that this is a complex system solving a complex task, and the available techniques to isolate and understand subsystems or study details of the inner states of the system are very limited. Partly due to these problems, researchers have focused on more accessible problems in the past, such as balance control during standing and the biomechanics of walking.

More recently, however, there has been vigorous research effort aimed to understand how the brain controls balance during walking. Several research groups have begun to generate a baseline understanding of how the brain uses some of the mechanisms detailed above, both in unperturbed walking and as a response to sensory or mechanical perturbations. Still, this body of research is in its infancy. Most of the focus has been on the foot placement strategy, with only very few studies including the other strategies. This is partly because the foot placement strategy is considered to be at once the most efficient and most important mechanism of balance control, and partly because it is substantially simpler to study, requiring only a motion capture system, which is relatively ubiquitous, and very few markers, in contrast to an instrumented treadmill, an EMG system, or the full-body motion capture required to process inverse kinematics or dynamics.

To gain a systematic understanding of the neural control of balance, we have to fill these gaps in our knowledge. One component is to study the other strategies and find out whether humans use them to maintain balance. For most of those strategies, there is preliminary evidence indicating that humans do use them. Once the phenomenology is established, we need to understand how the CNS combines the different strategies into a unified control scheme. Ultimately, this will require developing neural process models of balance control that are capable to reproduce and explain the observed data. Several such models have been developed for standing, the most prominent being that of Peterka (2002), which has been widely used as a basis to understand experimental data. For walking, Geyer and Herr (2010) have developed an impressive and promising model of the low-level control of muscle activity using spinal reflex arcs, which are already sufficient to generate human-like walking patterns in the sagittal plane with simplified dynamics. To generalize this model to three dimensions and the medial-lateral plane, Song and Geyer (2015) added a supraspinal control layer that modifies the set points of some of the spinal reflex arcs. In our own work on gait initiation, we used a model with very simplified biomechanics to understand how the foot placement and the lateral ankle strategy might complement each other over a single step (Reimann et al., 2017). In combination, different modeling approaches such as these could lead to a comprehensive model of neural control of balance during locomotion, including both spinal and supra-spinal components, and the necessary interface between those two. If capable of reproducing and thus explaining a wide range of experimental data, such a model would represent a systemic understanding of the neural control of balance during walking.

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