Foot placement in balance recovery

complex humans vs simple model

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FOOT PLACEMENT IN BALANCE RECOVERY

COMPLEX HUMANS VS SIMPLE MODEL

DISSERTATION

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the degree of doctor at the University of Twente,
on the authority of the rector magnificus
Prof. dr. T. T. M. Palstra
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CHAPTER 1

Introduction

“Let's-a-go”

Human balance control

Maintaining balance in daily life is so common to us, and so embedded in the developed human body, that watching someone accidentally fall can even be considered funny. A fall is simply not supposed to happen, and the major contradiction between balance loss and a daily life situation can make it humorous (Suls, 1972). Healthy humans can stand and walk while interacting with the environment, avoid walking into obstacles, deal with uneven or slippery terrains, and quickly recover from unexpected disturbances, all while staying upright on our two feet.

To do this, we need information regarding the state of our body. Multiple sensors are at our disposal to provide a variety of feedback, such as the linear and angular accelerations of the head through the vestibular system, muscle length and muscle velocity through muscle spindles, tendon forces through Golgi-tendon organs, environmental contact through touch, and the distant environment through vision. Prior knowledge regarding the situation is utilized in balance control as well. When walking on slippery surfaces we adapt our gait based on our expectations of a possible slip (Oates et al., 2010). When we know that an unavoidable perturbation is coming, preparations are made to minimize the possible negative consequences (Brown and Frank, 1997). Even expected effects of our own voluntary movements are taken into account with anticipatory postural adjustments (Bouisset, 1987).

With information from both feedback and feedforward mechanisms, mediated by cortical, sub-cortical, and spinal pathways, our muscles must contract to move the body appropriately, see Figure 1. Muscles cannot, however, provide any desired force, or contract at any desired velocity. Not every movement can be executed within a given time. The constraints on our movement generation therefore also influence the way we maintain balance (Horak, 1987). For example, falls occurring in elderly is often considered a consequence of decreased muscle capabilities, such as muscle strength (Skelton et al., 2002; Pijnappels et al., 2008). Even the way we perceive and possibly misjudge the constraints on our own movements can have an effect (Kluft et al., 2017). Considering these many factors influencing and contributing to balance control, it is perhaps not surprising that it is not fully understood how we can stay upright with so little effort.

Impaired balance control

It is no longer amusing if a person has permanent issues staying upright. Various neurological conditions can lead to difficulties in maintaining balance, and with it affect a person's mobility. One example is a stroke or cerebrovascular accident, in which a blood clot or a blood vessel rupture in the brain can cause damage to the brain tissue. This often leads to hemiparesis, involving difficulties in muscle control on one side of the body. Another example is a spinal cord injury (SCI), in which the nerves in the spinal cord are damaged, leading to poor conductivity of the control signals from the brain to the muscle. Various muscles might fail to contract depending on the completeness and the height of the lesion. The leg muscles are especially susceptible to SCI, as these are innervated through lower regions of the spinal cord, far from the brain. Finally, also for healthy people there exist various conditions where maintaining balance is difficult. Examples include carrying heavy loads, traversing difficult terrain, or simply through the effects of aging.

Figure 1. Systems in the human body that play a role in balance control. Balance control is mediated through 1) cortical and sub-cortical neural pathways as well as 2) spinal pathways, which rely information from 3) the vestibular system, 4) vision, 5) cutaneous receptors, and 6) muscle spindles and Golgi-tendon organs in a muscle-tendon complex. This information is used to contract our muscles and move 7) the musculoskeletal system. Damage to the neural pathways or to a sensory system can lead to difficulties maintaining balance.

Balance support through powered orthoses

Robotic devices might provide an outcome in scenarios where balance is difficult by providing assistance during standing and walking. A growing field in robotics are powered orthoses. Such an orthosis might be considered as a machine that can be worn around the body. It is often referred to as an exoskeleton, especially when the construction contains rigid linkages. In case of a lower-extremity exoskeleton, a construction is worn around the legs to provide the user with support during standing and especially during walking, see Figure 2. This can be to various purposes, such as to reduce the energetic costs of walking or load carrying during walking (Kawamoto et al., 2003; Kazerooni and Steger, 2006; Mraz, 2008; Collins et al., 2015), to assist in gait rehabilitation (Jezernik et al., 2003; Banala et al., 2007; Veneman et al., 2007; Meuleman et al., 2016), or to function as a “permanent” walk-assist to help regain walking ability (Strausser et al., 2011; Esquenazi et al., 2012; Hartigan et al., 2015; Wang et al., 2015).

Despite their different purposes, most actuated lower-extremity exoskeletons have one thing in common: they have no sense of balance. The user must take the lead in balance recovery whenever an unexpected disturbance occurs, because the exoskeleton has no clue what to do. This is problematic for various reasons. First, if the user is only poorly able, or not at all able to maintain balance on his or her own, a fall might be inevitable. To prevent falls, crutches are often required when a person with SCI

Figure 2. Example of two commercially available lower extremity exoskeletons. Left: the Ekso developed by Eksobionics. Right: the Hybrid Assistive Limb (HAL) for medical use, developed by Cyberdyne.

uses an ambulatory exoskeleton to regain walking functionality. Second, if the user *is* able to initiate balance recovery, the exoskeleton might only impede the user in execution of the recovery movements. The device adds additional mass and inertia to the user's legs. These have to be dragged along by the user if there is no support from the exoskeleton during the balance recovery.

Controllers implemented in powered lower-extremity exoskeletons require a sense of balance to tackle such issues. This way the exoskeleton might assist the user in maintaining balance rather than the other way around. Especially in the case where the user is (partially) able to recover balance him- or herself, it is important that the exoskeleton provides balance support in a human-like way. If not, the exoskeleton might collide with what the user would naturally try to do. However, to realize balance controllers that allow a powered lower-extremity exoskeletons to assist in maintaining balance in a human-like way, we must first gain an understanding of what human balance is and how humans regain balance when it is lost.

What is balance?

There exists no general accepted definition of what human balance is (Pollock et al., 2000). It has been described as the body dynamics that prevent a fall (Winter, 1995), but this raises questions about what a fall is. For example, walking might be seen as a sequence of interrupted falls. Every step the body moves toward the ground and is subsequently redirected using the leading leg (Donelan et al., 2002). Yet, we do not consider walking to be a loss of balance.

It is possible to intentionally fall down if we want to, but perhaps in a controlled way. Springboard

diving, gymnastics, and other sports with flight elements involve highly controlled falls. Considering this example, human balance control might be seen with respect to some intended objective. It is okay to let gravity pull you down if that is what you want. In this view balance is not related to falling per se. It just happens to be that most of the time the intended objective is to stay upright, while gravity causes increasing deviations from that objective if no immediate action is taken. External forces acting on the body might therefore oppose, align, or be invariant with respect to some objective, and could therefore have a perturbing or an assisting effect. As a result, two physically equivalent situations might be considered both balanced and unbalanced depending on what the objectives are. To have a lower-extremity exoskeleton assist a user in maintaining or regaining balance, specifically during walking, the control of the exoskeleton must therefore align with the intended objective(s) of the user.

The problem is that we often do not exactly know what the objectives are, or on what level they are defined. In walking, for example, we generate consistent vertical forces (Toney and Chang, 2013) possibly to keep the body's center of mass (COM) at a certain height, while moving in a certain direction at a certain speed, and while also keeping the rotation of the body in check (Herr and Popovic, 2008; Thielemans et al., 2014). Further objectives might depend on environmental constraints. When walking on a treadmill, an implicit objective might be to stay in the center to prevent rolling off the belt. When asked to walk over stepping stones there is an explicit foot placement objective, which is perhaps not present during unconstrained walking.

We can probe human balance control by threatening it through the application of perturbations. When the perturbation is not too large a balance recovery will likely occur. This provides insight in what exactly humans are trying to recover, and how this recovery occurs. The latter is of major interest for the support of human balance through a lower-extremity exoskeleton in a human-like way. If the exoskeleton could execute the same balance recovery strategies as a healthy human, a user who possibly suffers from reduced balance capabilities might regain function through the device.

Investigating balance control

Balance is often investigated by disturbing it and observing the recovery response. Literature provides a massive range of perturbation experiments. These include perturbations applied to the stance leg(s) through support surface translations (Maki et al., 1996; Oddsson et al., 2004; Brady et al., 2009; Hak et al., 2013; Sari and Griffin, 2014), rotations (Rouse et al., 2014), and elevations (Nashner et al., 1979; Af Klint et al., 2009), through loss of support surface area (Horak and Nashner, 1986; Otten, 1999) and loss of support surface friction (Pai and Iqbal, 1999; Cham and Redfern, 2002), through perturbations applied to the swing leg (Cordero et al., 2003; Pijnappels et al., 2005; Rankin et al., 2014), to the pelvis (Hof et al., 2007; Qiao and Jindrich, 2014), to the shoulders (Engelhart et al., 2015), and to the head (Horak et al., 1994), through sudden release of a supporting force (Do et al., 1982; Hsiao-Wecksler, 2008), and through sensory disturbances (Hayashi et al., 1981; Fitzpatrick et al., 1994), just to name a few, see Figure 3. On top of all the possible ways to apply a perturbation, other factors such as the magnitude and direction (Maki et al., 1996) of the perturbation, the subject's expectations about the perturbation (Bhatt et al., 2006), the instruction to the subject (Do et al., 1999), as well as the subject characteristics (healthy, impaired, young, elderly) can affect the observed balance recovery response.

The various experimental paradigms come with advantages and disadvantages. Considering the application of human-like strategies to exoskeletons, it is useful to use perturbations that could occur

during daily use of the device. A Stewart motion platform used for support surface translations can induce sudden accelerations to perturb balance. The range of motion of such a platform is often limited, however, such that an acceleration has to be shortly followed by a deceleration. These decelerations can have a stabilizing effect (Carpenter et al., 2005; van Asseldonk et al., 2007), and might hamper analysis of the destabilizing effects caused by the acceleration. Applying forces to the swing leg can cause tripping. This is a major cause of falls in the daily life of elderly (Robinovitch et al., 2013), which motivates the use of such perturbations to investigate balance. Trips induce changes in linear body motion, but can also have a effects on the rotation of the body because the perturbation force is applied at a distance from the body's COM. Because this affects multiple variables simultaneously, analysis of how humans respond to a change in a specific variable might be complicated. In contrast, sensory disturbances such as stimulation of the vestibular organ allows investigating specific effects of a sensing modality on human balance. Yet, such isolated stimuli are rarely encountered in a daily life situation. This could make it challenging to translate such experimental findings to ways of assisting humans in maintaining their balance through a lower-extremity exoskeleton. This motivates the use of an experimental setup that can apply perturbations through physical interaction with the body, as most often occurs in a daily life, while keeping the manipulation controlled and simple.

Figure 3. Examples of balance perturbation methods. Left: mediolateral perturbations at the pelvis through an actuated rod. Right top and bottom: tether release. A subject takes a lean angle while suspended from a cable. A fall can be initiated by suddenly releasing the support force in the cable.

Balance recovery strategies

The various ways to perturb balance can give rise to many different balance recovery responses. We might look for interaction with the environment for stability, as even minor tactile cues with a stable surrounding can already provide support (Johannsen et al., 2017). A stable environment is often out of reach, however, such that we have to reject disturbances on our own by executing certain balance recovery movements. The recovery actions that we take can be divided in three main “balance strategies”. These are the ankle strategy, the hip strategy, and the stepping strategy (Nashner, 1982; Horak, 1987; MacKinnon and Winter, 1993), see Figure 4.

In the ankle strategy the ankle muscles rotate the ankle joint(s) of the stance leg(s) to affect body movement. If the foot remains flat on the floor this results in the body rotating around the ankle joint. Because of the ankle joint moments, the force distribution beneath the foot or feet on the ground changes. This distribution can be captured in a single point termed the center of pressure (COP), which can be considered as the origin of the net reaction force beneath the foot or feet. As a result, the COP is constrained to the area of the base of support made up by the foot or feet. The shifts in COP accompanied by the changes in ground reaction force (GRF) that occur during walking as a result of the ankle strategy assist in continuous stable walking (Gruben and Boehm, 2014).

In the hip strategy, the hip joints influence linear body motion through upper body rotation. Angular acceleration of the relatively large upper body inertia can influence the ground reaction forces originating from the COP. Forward angular acceleration of the trunk about the hip can help accelerate the COM backward, whereas backward angular acceleration of the trunk has the opposite effect. The hip joints are also involved in keeping the upper body upright, but this is not strictly part of a hip strategy. For example, a perturbation might linearly accelerate the body forward, while also causing the upper body to rotate forward. In this case, moving the upper body back upright requires a backward angular acceleration of the trunk, which would contribute to forward acceleration of the COM, acting in the same direction as the perturbation. Because this upper body rotation does not help to restore the linear motion, we do not consider it a hip strategy. Influencing linear motion through rotation can also be achieved with other body segments, such as arm movement (van Asseldonk et al., 2007) or leg movement (MacChietto et al., 2009). The hip strategy might therefore be considered part of a larger set of “inertial strategies”.

Finally, in the foot placement strategy or stepping strategy the location of the foot is changed. This creates a new base of support within which the COP can move, and allows modulation of the ground reaction forces that would not be possible without displacing a foot. Foot placement is therefore often considered the most relevant strategy in human walking (Patla, 2003). Adjustments might be made not only in terms of location, but also in terms of time. A faster step at the same location relative to the body might be sufficient to recover from a perturbation during walking, without adjusting the location of the foot compared to normal walking. The ability to adjust both the location and the timing of foot placement leads to a tremendous range of possible options for the foot placement strategy. Understanding and predicting those combinations used by humans is of major importance for a natural, human-like way of controlling balance in lower-extremity assistive devices.

Modeling balance

Although literature already provides a broad range of perturbation experiments and conditions, it is next to impossible to investigate all possible forms of perturbations, applied to various parts of the body, during various circumstances. It is therefore of importance to converge towards models, and control strategies applied to such models, which can mimic and explain the observed human behavior when balance is threatened. Such models and control strategies might subsequently serve as a basis for more human-like control of lower-extremity orthoses.

Figure 4. Various balance strategies that humans can address. The ankle and hip strategies involve contraction of the ankle and hip muscles to influence linear and angular body motion. The stepping strategy involves adjustments to the location and/or timing of foot placement. Red lines indicate the effects on the ground reaction forces.

Figure 5. The inverted pendulum analogy for human standing and walking, and basic concepts associated with the inverted pendulum. COM: center of mass. COP: center of pressure. XCOM: extrapolated center of mass, directly proportional to the horizontal COM velocity. In a linear inverted pendulum model, placing the COP in the XCOM through foot placement brings the pendulum to an upright movement stop.

One issue that complicates modeling human-like balance is the dimension of the problem. Humans can act in many different ways to maintain balance, and address a multitude of joints to deal with disturbances. Unfortunately, with increasing degrees-of-freedom the control complexity of a model rapidly increases as well. On the other hand, a model that does not contain the dynamical properties similar to those of a human is unlikely to fully explain human-like balance because balance control is dependent to the constraints and capabilities of the system at hand. Consequently, an as simple as possible model with an as high as possible explanatory value is always preferred.

On one end of the model spectrum are highly simplified, low dimensional approaches. One of the most simple ways to model standing and walking is the inverted pendulum model (Cavagna et al., 1976; Townsend, 1985; Kajita and Tani, 1991), which only consists of a point mass swinging on top of a single leg with a foot of infinitesimal size. Though simple, the pendulum swing has similarities with both quiet standing and the single support phase of human walking (Winter, 1995). It can be used to study the effects of foot placement on the motion of the model (Townsend, 1985) and to derive fundamental concepts related to such motion. One such concept is the extrapolated center of mass (XCOM) (Hof et al., 2005) or capture point (Pratt et al., 2006) derived from a linear inverted pendulum model. The XCOM can be regarded as a point on the floor at a distance from the COM proportional to the horizontal COM velocity, see Figure 4. If the COP is placed in the XCOM the model will convert all its kinetic energy into potential energy, which will result in an upright motion stop. If the COP cannot reach the XCOM the model cannot terminate movement, hence the concept's potential in maintaining balance. Because inverted pendulum models are simple, yet contain the most basic requirements for bipedal walking, they have served as a basis for studies of passive gait stability such as in limit cycle walking (McGeer, 1990; Geyer et al., 2006; Hobbelen and Wisse, 2008). It has even functioned as an underlying concept for foot placement in bipedal models that are far more complex than the inverted pendulum model itself (Khadiv et al., 2016).

On the other end of the model spectrum are high degrees-of-freedom models underlying the control of humanoid robots. In the field of bipedal robotics there exist various strategies to make complex two-legged robot stand and walk. A straightforward way to generate a walking motion is to pre-program every step to realize walking. If any unexpected disturbance occurs, however, the model will continue to execute its motion without reacting to the disturbance, which can easily lead to a fall. More sophisticated control methods, such as those based on the zero-moment point (ZMP) (Vukobratović and Borovac, 2004), attempt to constrain the movements based on conditions related to body rotation. A ZMP-based control of walking often leads to the typical “bent-knee gait” that robots such as Asimo display (Hirai et al., 1998). While this method can deal with minor perturbations, it is known that humans do not use to this form of control in which the changes in angular momentum of the body are kept zero (Herr and Popovic, 2008). Other methods, such as model predictive control, can result in human-like motion through minimization of joint torques time to achieve a certain control objective (Van der Kooij et al., 2003). Unfortunately, these methods are computationally intensive and require one or multiple control objectives, such as a foot placement location and time, which are generally unknown following a disturbance. Hence, complex models and bipedal robots might successfully maintain balance during standing and walking, but often miss a more fundamental link to human-like balance control.

There are many perturbation studies that show how humans maintain their balance in various

conditions. There are also various ways to realize balance control during standing and walking in models that represent humans. However, the link between these two is rarely present in walking conditions involving disturbances. Only a few studies have attempted to mimic observations made in perturbed human gait using model approaches, for low dimensional (Hof et al., 2007) and high dimensional models (Mombaur et al., 2010; Park and Levine, 2013). Possible causes of the limited number of studies include the need of specific experimental equipment to investigate human subjects, as well as the challenges that are involved with modeling the human walking dynamics. Decreasing this gap between observations of balance control made in humans and models to mimic these observations can lead to a better understanding of human-like balance.

Thesis aim and outline

Human balance control is a complex, not fully understood process involving many mechanisms. Deterioration of any of the involved processes as well as environmental factors might impede balance control during standing and especially during walking. In such cases a powered lower-extremity orthosis might provide balance assistance. To prevent conflict between the device and its user it is important that the device can provide balance support in a human-like way. Realizing such balance controllers requires an understanding of what human balance is, and how it works. Insight in how humans maintain their balance can be obtained by experimentally applying perturbations to healthy subjects. This will lead to balance recovery responses involving various balance strategies. Foot placement adjustments are expected to be a major, crucial strategy in balance control during gait. Foot placement strategies might be replicated using simple inverted pendulum models of walking. Using experimental outcomes to validate simple models of human balance could provide a basis for the design of balance controllers for lower-extremity orthoses. To make comparison with such point-mass models more straightforward, perturbations might be applied to the approximate location of the whole-body center of mass of the human, which is the pelvis. The main goal of this thesis is therefore to investigate balance strategies in response to horizontal pelvis perturbations during human walking, with a specific focus on foot placement strategies. The thesis deals with the following questions:

- I)** How do humans adjust their foot placement in terms of location and time when subjected to horizontal pelvis perturbations during walking? (Chapter 3-6).
 - Ia)** How do physical constraints affects the foot placement strategy for these perturbations? (Chapter 4-6).
 - Ib)** How is the foot placement strategy complemented by other balance strategies for these perturbations? (Chapter 3, 4, 5, 7, 8).
- II)** Can the foot placement strategy be predicted using simple low dimensional models of walking, and concepts derived from these models? (Chapter 3-6, 9).

Though the focus of the thesis is on balance during walking, we start experimenting in standing to show that we require balance control even when we try to stand still.

Chapter 2 shows that the intrinsic properties of our ankle joints are not sufficient to counter the gravitational pull on our body. This implies that humans do not passively maintain upright balance, but must use a control strategy instead.

Chapter 3 presents balance responses following horizontal pelvis perturbations during walking, with a specific focus on how we adjust our foot placement with the various disturbances. It becomes clear that foot placement adjustments are not always the primary strategy to recover from perturbations during walking, even if it is an available option. Other strategies might be addressed instead, such as an ankle strategy following perturbations in the sagittal plane. Furthermore, relations are presented that allow the prediction of the COP location after foot contact based on the velocity of the COM. These relations are in line with the XCOM concept derived from a simple inverted pendulum model of walking.

Chapter 4 replicates the experimental conditions presented in Chapter 3, but also introduces an additional constraint on the stepping strategy by applying perturbations closer to the instance of foot contact to limit the available adjustment time. The most prominent foot placement modulation occurs for mediolateral perturbations, given that there is sufficient time to realize foot placement adjustments. The closer in time a disturbance occurs to foot contact, the less the step will be modulated. As a result, various strategies are employed in the subsequent gait phases to deal with the perturbation effects.

Chapter 5 replicates the experimental conditions presented in Chapter 3, but also introduces an additional physical constraint to the ankle joints to limit the available recovery strategies. This constraint elicits foot placement adjustments that were otherwise not present, suggesting that the ankle provides an alternative recovery. It furthermore shows that humans prefer to stabilize the upper body, while maintaining the relation between COP location and COM velocity found in Chapter 3.

Chapter 6 presents the stepping responses of stroke survivors to sudden sideways perturbations applied during walking. Though there are clear differences between steps made with the paretic and the non-paretic legs of these subjects, both legs have a similar ability to modulate foot placement in response to mediolateral perturbations, despite the impairments resulting from the stroke. A possible cause is that the balance responses are partially involuntary, bypassing regions affected by the stroke.

Chapter 7 provides an inverse kinematics and inverse dynamics analysis of the results obtained in Chapter 3, as well as responses of various leg muscles. These results provide insight in balance control on a joint level, and can function as a benchmark for orthoses and controller design in terms of actuator degrees-of-freedom and output.

Chapter 8 provides insight in unexpected muscle activation patterns in the lower limbs during the experiments with blocked ankle joints presented in Chapter 4. It reveals that rapid changes in ankle joint muscle activation occurs in the ankle joint muscles in response to perturbations, even if these muscles are not subjected to muscle length changes, nor have a functional contribution to maintaining balance. The results therefore suggest a centralized regulation of balance control involving supra-spinal neural structures, without the need for changes in ankle muscle proprioceptive information.

Chapter 9 uses a simple model approach in an attempt to explain the simultaneous modulation in step location and step time as observed in humans in Chapter 3. By combining costs for leg swing and step-to-step transition suggested in previous studies, and by applying an XCOM constraint to the location of the foot, both the location and the time of foot placement can be made to modulate simultaneously in patterns that resemble those in the collected experimental data in Chapter 3.

Chapter 10 provides a general discussion of the acquired results, as well as several future directions.

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CHAPTER 2

Direct measurement of the intrinsic ankle stiffness during standing

"You're too slow!"

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Direct measurement of the intrinsic ankle stiffness during standing

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Abstract

Ankle stiffness contributes to standing balance, counteracting the destabilizing effect of gravity. The ankle stiffness together with the compliance between the foot and the support surface make up the ankle-foot stiffness, which is relevant to quiet standing. The contribution of the intrinsic ankle-foot stiffness to balance, and the ankle-foot stiffness amplitude dependency remain a topic of debate in the literature. We therefore developed an experimental protocol to directly measure the bilateral intrinsic ankle-foot stiffness during standing balance, and determine its amplitude dependency. By applying fast (40 ms) ramp-and-hold support surface rotations (0.005–0.08 rad) during standing, reflexive contributions could be excluded, and the amplitude dependency of the intrinsic ankle-foot stiffness was investigated. Results showed that reflexive activity could not have biased the torque used for estimating the intrinsic stiffness. Furthermore, subjects required less recovery action to restore balance after bilateral rotations in opposite directions compared to rotations in the same direction. The intrinsic ankle-foot stiffness appears insufficient to ensure balance, ranging from 0.937 ± 0.09 to 0.447 ± 0.06 (normalized to critical stiffness 'mgh'). This implies that changes in muscle activation are required to maintain balance. The non-linear stiffness decrease with increasing rotation amplitude supports the previous published research. With the proposed method reflexive effects can be ruled out from the measured torque without any model assumptions, allowing direct estimation of intrinsic stiffness during standing.

Introduction

Human standing balance is continuously challenged by gravity, which imposes a negative stiffness on the upright equilibrium posture. This “critical stiffness” must be compensated to maintain upright stance. The ankles' muscle–tendon structures provide stiffness at multiple levels. First, stretch reflexes can lead to changes in muscle activation levels and affect joint stiffness (Sinkjaer et al., 1988). Second, the muscle–endon complex provides a direct mechanical torque response to stretch (Rack and Westbury, 1974). This intrinsic stiffness depends on the tonic activation level, which influences the muscle's mechanical properties through cross-bridge formation. Cross-bridges are thought to cause a high short-range stiffness due to elastic stretch (Rack and Westbury, 1974; Morgan, 1977), and a lower long-range stiffness by detaching and sliding muscle filaments (Campbell and Lakie, 1998). Separating the reflexive and intrinsic contributions to the overall ankle stiffness during standing might give insight into neuromuscular disorders, and could help in the assessment of balance control in a clinical setting.

Ankle stiffness can be estimated by applying a rotation to the foot and measuring the torque response. In the current literature there are various definitions of ankle stiffness, which are here distinguished as: (1) The actual ankle stiffness, which can be estimated using the rotation between the lower leg and the foot. (2) The ankle-foot stiffness, which can be estimated using the rotation between the lower leg and the contact surface of the device used to apply a rotation to the foot. This includes both the ankle stiffness and possible foot compliance. (3) The pseudo anklefoot stiffness, which can be estimated using only the rotation angle of the foot contact surface, assuming no lower leg movement.

Ankle stiffness, in general, has been investigated using a wide variety of conditions. Stiffness varies with muscle contraction level (Hunter and Kearney, 1982), mean joint angle (Gottlieb and Agarwal, 1978; Weiss et al., 1986) and rotation amplitude (Kearney and Hunter, 1982). In the latter study, pseudo-random binary sequence rotations varying from 0.01 to 0.25 rad were applied to the left foot in supine subjects. The pseudo ankle-foot stiffness decreased with increasing rotation amplitude, and both intrinsic and reflexive mechanisms contributed to the results. Later, in (Kearney et al., 1997) system identification methods were applied to separate intrinsic and reflexive components. It was concluded that reflexive contributions depend strongly on the conditions, and that the generated reflexive torques can be of the same magnitude as those from intrinsic mechanisms. Similar proportions were reported in hemiparetic patients (Sinkjær and Magnussen, 1994), where nerve stimulation was used to suppress reflexive activity.

In a limited number of studies ankle stiffness was estimated in upright stance, where stiffness is often expressed as “relative stiffness”, i.e. normalized to the critical stiffness. In (Peterka, 2002), subjects mimicked a single-link inverted pendulum during backboard supported stance. A pseudo-random ternary rotation sequence of 0.009–0.14 rad was applied to the support surface. Parametric estimates resulted in a relative ankle-foot stiffness of approximately 0.15. The work of (Loram and Lakie, 2002) described the use of a piezo-electric element to apply 0.001 rad rotations to the left foot during both free and backboard supported stance. Cosine waves with a rise time of 70 ms were used. A relative pseudo ankle-foot stiffness of 0.91 was found by using parametric estimates. In a subsequent study, values of 0.67 and 0.54 were found for slow (41 s) 0.003 and 0.007 rad rotations respectively, using a similar setup (Loram et al., 2007a, b). Transient rotations of 0.02 rad and a rise time of 150 ms were used in (Casadio et al., 2005). Subjects were freely standing on a footplate capable of perturbing both feet simultaneously. Various estimation methods were attempted to minimize potential effects of short

latency reflex activity and lower leg movement, leading to a relative ankle-foot stiffness of 0.64. Short latency reflex activity in human soleus muscle occurs approximately 40 ms after stretch onset (Grey et al., 2001). Until now, reflex activity has not been ruled out from the ankle stiffness estimates by applying sufficiently fast rotations during stance. Although several previous studies suggest that the relative (pseudo) ankle-foot stiffness is lower than 1, the intrinsic contribution remains uncertain. Here the goal is to directly quantify the rotational amplitude dependency of the intrinsic (pseudo) ankle-foot stiffness in healthy subjects during stance. By applying 40 ms ramp-and-hold plantar- and dorsiflexion rotations to both ankle joints simultaneously, reflex activity will be removed from the stiffness estimates. What remains is the intrinsic stiffness that can be estimated directly, without model assumptions. Furthermore, simultaneously applying a plantar flexion to one ankle and a dorsiflexion to the other might prevent disturbing the subject's balance during the experiment.

Methods

Participants

Eight healthy volunteers with no known history of neurological or muscular disorders participated in the study (7 men, age 23 ± 1 years, weight 75 ± 8 kg, height 1.85 ± 0.07 m, mean $\pm$ sd). All subjects gave prior written informed consent in agreement with the guidelines of the local ethical committee, and in accordance with the Declaration of Helsinki.

Apparatus

Rotations were applied to both ankle joints using the bilateral ankle perturbator (BAP) as shown in Figure 1. A detailed description of the apparatus can be found in (Schouten et al., 2011). The device consists of two lightweight platforms, each connected to an electromotor (HIWIN, IL; type TMS3C) via a lever arm. The lever arms can be adjusted to align the subject's ankle joints with the rotational axis of the motors. As a safety measure, an ultrasound sensor was incorporated in each platform to check for heel contact. Rotations could not be applied if the subject did not make heel contact with the sensor. Furthermore, a safety harness connected to the ceiling with a belt and locking retractor was worn around the chest to prevent injury in case of a fall. The harness did not provide any support while standing on the BAP.

Force transducers (Revere Transducers Inc, CA; type ALC-C2) between each lever arm and motor were used to measure the torque exerted on each platform. Platform angular displacement and velocity were measured using rotary encoders (2.5×10^{-4} rad accuracy). All BAP data was captured at 10 kHz using a DAQ-card (HUMUSOFT, Czech Republic, MF624) running xPC-target (The Mathworks, Natick, US). Kinematic data was captured at 120 Hz using a 6-camera VICON system (Oxford Metrics, Oxford, UK) and 20 reflective markers. Markers were placed at the acromion, femur head, lateral epicondyle, tibia, lateral malleolus, calcaneus and metatarsal 1 head on the left and right side of the body, as well as on top of each lever arm, and on the front and back of each platform. Activity patterns of the tibialis anterior (TA), soleus (SO), gastrocnemius medialis (GM) and gastrocnemius lateralis (GL) muscles were recorded using surface EMG electrodes (Delsys Inc, Natic, MA). EMG data was amplified (1000x) and captured at 1560 Hz using the AD converter of the Vicon.

Figure 1. The Bilateral Ankle Perturbator (BAP). Rapid 40 ms plantar- and dorsiflexions were simultaneously applied to both feet using the BAP. The lever arms can be adjusted to align the subject's ankle joints with the motor axis.

Experimental protocol

Subjects stood on the BAP platforms, and were instructed to keep knees and hip in an extended position. Arms were held over the chest. Subjects leaned slightly forward to reduce effects of natural sway and achieve a consistent ankle angle at perturbation onset. A screen in front of the subject gave visual feedback on a target ankle torque and the exerted ankle torques. The target torque for each ankle was derived from a simple linearized inverted pendulum equation:

$$T_{target} = \frac{(m g h \phi)}{2}$$

where m is the subject's mass (kg), g the earth's gravitational constant (ms^{-2}), h the subject's estimated center of mass (COM) height (m) and ϕ the desired subject lean angle from the vertical (rad). The term $m * g * h$ is equal to the critical stiffness (Casadio et al., 2005), being the minimum intrinsic ankle-foot stiffness required for stabilization without changes in muscle activity. The COM height was estimated using a weighted average of body segments (Winter, 2009). For all subjects ϕ was set to 0.07 rad (4 degrees). This led to a subject average target torque of 26 ± 3 Nm per ankle.

A ramp-and-hold rotation was applied when the torque exerted on each platform was held within 10% of the target torque for a random time interval of 2–4 s. To allow non-parametric intrinsic stiffness estimation, 40 ms minimum-jerk profiles (Burdet et al., 2000) were used. These ensure (near) zero velocity and acceleration at the start and end of the perturbation, such that damping and inertial effects of the platforms and feet are minimized. Both plantar- and dorsiflexion rotations of 0.08, 0.04, 0.02, 0.01 and 0.005 rad were applied. Either unidirectional rotations (UR) or bidirectional rotations (BR) were used, for which the absolute amplitude within one condition was equal for both platforms. Each condition was repeated eight times, resulting in 160 trials per subject. All rotations were applied in a randomized order to prevent anticipation or learning effects. Two seconds after rotation onset the platforms returned to their neutral position in one second. The subject could subsequently try to reach the target torque again for the next perturbation. To prevent muscle fatigue, subjects were instructed to sit down and rest for a short period after every 6 min of measurements. Baseline trials were collected by applying the rotations without a subject standing on the BAP.

Data analysis

All data were processed using Matlab (R2012b, The Mathworks, Natick, US). Subject trials for which the average torque before onset deviated more than 10% from the target torque were disregarded. The torque responses of the baseline trials were averaged over the eight repetitions. These were subtracted from all corresponding subject torque responses to correct for gravitational effects of the platforms. Baseline corrected torques were subsequently filtered with a 2nd order 150 Hz zero-phase low-pass Butterworth filter.

Raw marker data were used to get an indication of the platform, feet and lower leg segment rotations. All markers that moved more than 10 cm in any direction during a window of 100 ms before to 200 ms after perturbing were considered outliers. The corresponding trials were removed from further analysis. The toe, heel and malleolus data were interpolated between 30 and 70 ms after perturbation onset to deal with (skin) movement artefacts resulting from the perturbation. Segment rotations were calculated using the marker displacements averaged over the eight repetitions. Rotations of the feet could not be calculated accurately for the smaller perturbations (< 0.04 rad), given the accuracy of the motion capture device. Consequently, the actual ankle stiffness was not further analyzed. Marker data were filtered with a 2nd order 20 Hz zero-phase low-pass Butterworth filter to estimate the COM position and its deflection angle from the vertical passing through the ankle joint markers.

The intrinsic pseudo ankle-foot stiffness for each ankle was estimated by dividing the difference in torque by the difference in platform encoder angle. The intrinsic ankle-foot stiffness for each ankle was estimated by dividing the difference in torque by the difference in the angle between lower leg and platform. For both estimates, differences were calculated for each trial using the average signal values before (–15 to –5 ms) and after (45–55 ms) rotation onset. The relative intrinsic (pseudo) ankle-foot

stiffness was computed from the sum of the left and right ankle joint stiffness, divided by the subject's critical stiffness. For each subject and each condition an average relative intrinsic stiffness was calculated over the eight repetitions. These averages were used to compute a between subjects standard deviation.

A statistical linear mixed model with fixed effects for amplitude, direction (plantar- and dorsiflexion) and similarity (UR and BR) was used to investigate their effects on the intrinsic (pseudo) ankle-foot stiffness. Subject effects were specified as random. Effects of rotation direction were only investigated for UR, since the total stiffness calculated for BR already consisted of both plantar- and dorsiflexions. Given the test results, plantar- and dorsiflexions were considered equal and were pooled for further analysis. Subsequently, differences in stiffness between UR and BR were investigated, as well as the effect of the rotation amplitude on the pooled UR and BR stiffness. A significance level of $\alpha = 0.01$ was used and a Bonferroni correction was applied to correct for multiple comparisons. Finally, a least squares fit of the form $y = a * 10^{\log(x)} + b$ was made to the rotation amplitude and the pooled UR and BR intrinsic (pseudo) ankle-foot stiffness per amplitude.

EMG data were detrended and filtered with a 1st order 48–52 Hz bandstop Butterworth filter. These data were subsequently rectified, filtered with a 1st order 40 Hz low-pass Butterworth filter, and cut into sequences from 1 s before to 2 s after perturbation onset. EMG averages were calculated over the eight repetitions of each condition, as well as over all subjects for each condition.

Results

Subject responses

The 40 ms minimum-jerk ramp-and-hold rotations were completed before short latency reflex activity occurred. This can be seen in Figure 2, where the subjects' average response to the UR and BR conditions is shown for left 0.08 rad dorsiflexions. The GM muscles showed short latency reflex activity, starting approximately 45 ms after perturbation onset. The amplitude of the reflex response is slightly lower for UR compared to BR. The GL and SO muscles showed similar effects as those of the GM. The TA muscles showed little to no short latency responses.

The difference in torque before and after perturbing was found slightly higher for all BR compared to UR. For all 0.08–0.005 rad BR the average torque differences ranged from $13.70 \pm 2.87 - 1.23 \pm 0.51$ Nm respectively, compared to $11.60 \pm 2.78 - 1.12 \pm 0.48$ Nm respectively for UR. Subjects required a larger balance recovery response for UR compared to BR. This is reflected in the torque and EMG signals, as well as in the COM angle with the vertical. After approximately 150 ms, an increase in TA EMG can be observed for UR but not for BR. This is followed by a dorsiflexion torque, which is higher for UR compared to BR. Furthermore, the COM shows more deflection from its initial position for UR compared to BR.

Finally, the segment rotation angles as calculated from the marker data revealed lower leg rotations, and foot rotations smaller than those of the platforms. The average rotation angle between platform and lower leg is 5 ± 2 , 10 ± 2 , 16 ± 6 , 23 ± 6 and $26 \pm 9\%$ less than the applied platform rotations of 0.08–0.005 rad respectively. Consequently, the encoder angle is not an accurate representation of the angle between platform and lower leg over the full perturbation range.

Figure 2. Time series averages over all subjects' bidirectional rotations (BR, solid black) and unidirectional rotations (UR, dashed gray) for left 0.08 rad dorsiflexion trials. The top row shows the rotation angles of the platforms (arrow up), feet (arrow down) and lower legs (star). The next four rows show the applied rotation angle, measured torque, and GM and TA EMG responses for the left and right ankle. The vertical line indicates perturbation onset. A negative torque corresponds to a plantar flexing torque exerted on the BAP platforms. The middle three rows show the same torque and EMG responses on a longer time scale. The bottom graph shows the COM deflection angle with the vertical axis from the ankle joints. For each trial the average signal values over the shaded areas were used for stiffness estimation.

Intrinsic ankle stiffness

Of the total of 1280 trials, 188 were disregarded because the average torque before perturbing deviated more than 10% from the target torque. For the intrinsic ankle-foot stiffness estimates, another 148 trials were disregarded for inaccurate marker data. Figure 3 shows the amplitude dependency of the relative intrinsic (pseudo) anklefoot stiffness for plantar- and dorsiflexions of the UR. Plantar- and dorsiflexions for UR tested significantly different for the 0.005 rad ($p < 0.001$) rotations, with a mean difference of 0.13 and 0.37 for the pseudo ankle-foot stiffness and ankle-foot stiffness respectively. No significant differences between plantar- and dorsiflexions were found for the other amplitudes ($p = 0.040$ for 0.01 rad in the pseudo ankle- oot stiffness, $p > 0.116$ for all others).

Figure 4A shows the relative intrinsic pseudo ankle-foot stiffness for the BR and the UR with pooled plantar- and dorsiflexions. Several values found in the literature are also included in the figure. Loram et al. (Loram and Lakie, 2002; Loram et al., 2007a, b) mainly applied smaller rotations than in this study. Their results are in proximity of the fit, but suggest a steeper descending slope. On average, the relative intrinsic pseudo ankle-foot stiffness ranged from 0.67 ± 0.05 for 0.005 rad rotations to 0.42 ± 0.06 for 0.08 rad rotations. Parameter values of the fit $y = a * 10^{\log(x)} + b$ are $a = -0.21$ and $b = 0.21$. The fit has a coefficient of determination (R^2) of 0.97 and a root mean square error of 0.02. The fit extrapolates to 1 for approximately 10^{-4} rad rotations (not shown).

Figure 3. Rotation amplitude dependency of the subject average relative intrinsic pseudo ankle-foot stiffness (gray) and ankle-foot stiffness (black) for UR, shown separately for plantar- and dorsiflexions. Errorbars indicate the between-subject standard deviation.

Figure 4B shows the relative intrinsic ankle-foot stiffness calculated using the marker segment rotation angles. Especially for the smaller rotations the stiffness is higher compared to those found using the encoder angle, leading to a steeper decrease in stiffness with increasing rotation amplitude. The result at 0.02 rad found by Casadio et al. (Casadio et al., 2005) is in accordance with the fit. On average, the intrinsic ankle-foot stiffness ranged from 0.93 ± 0.09 for 0.005 rad rotations to 0.44 ± 0.06 for 0.08 rad rotations. Parameter values of the fit are $a = -0.38$ and $b = 0.01$. The fit has an R^2 of 0.99 and a root mean square error of 0.01. The fit extrapolates to 1 for $2.5 \cdot 10^{-3}$ rad rotations.

For both estimates, the intrinsic stiffness during standing decreased non-linearly with increasing rotation amplitude. With the exception of the ankle-foot stiffness for the 0.005 rad rotations, the UR yielded a consistently lower stiffness than the BR. These tested significantly different with a mean difference of 0.06 ($p < 0.001$) and 0.05 ($p = 0.007$) for the pseudo ankle-foot stiffness and ankle-foot stiffness respectively.

For pooled UR and BR, the relative intrinsic pseudo ankle-foot stiffness at each applied rotation amplitude tested significantly different from the stiffness at all other amplitudes ($p < 0.003$). For the relative intrinsic ankle-foot stiffness these differences tested significant as well ($p < 0.002$), with the exception of that between the 0.08 and 0.04 rad rotations ($p = 0.024$).

Figure 4. Rotation amplitude dependency of the subject average relative intrinsic psuedo ankle-foot stiffness (A) and ankle-foot stiffness (B) for BR, and for UR with pooled plantar- and dorsiflexions. Other values from the literature are shown for comparison. Errorbars indicate the between-subject standard deviation. The x-scale is logarithmic. The UR and BR data in panel A were slightly shifted on the x-axis for clarity, to prevent overlap. The fits were established by a linear least squares estimate on the $10\log$ of the (average) rotation amplitude and the average of the UR and BR relative stiffness.

Discussion

Reflexive activity

Rapid 40 ms minimum-jerk profiles of various amplitudes were applied to the separate ankle joints in order to directly estimate the intrinsic (pseudo) ankle-foot stiffness during standing. Muscle reflex activity could not influence the torques used in the stiffness estimation. Although stretch reflex onset starts within the time windows used for the estimates, it will not bias the torque in those windows due to the muscle's electromechanical delay. This delay was shown to be in the order of 30 ms for knee extensor muscle (Häkkinen and Komi, 1983).

Intrinsic ankle stiffness

On average, the relative intrinsic (pseudo) ankle-foot stiffness was found to be lower than 1 for pooled plantar- and dorsiflexions. Assuming that no velocity effects would occur, extrapolating the current findings suggests that the intrinsic (pseudo) ankle-foot stiffness could counteract rotations that are smaller than those occurring in the ankle during natural sway (Aramaki et al., 2001). Hence, a 40 ms externally applied disturbance of nearly any magnitude cannot be counteracted by the passive ankle structures under these tonic muscle activation levels. These results suggest that the intrinsic ankle-foot stiffness alone is insufficient to maintain upright balance. This is in accordance with previous findings (Morasso and Sanguineti, 2002; Loram et al., 2005). Consequently, changes in ankle muscle activation are required, or another method of control such as a hip strategy must be applied.

The differences with the findings in (Loram et al., 2007a, b) might be explained by the different experimental conditions. In their work subjects were supported at the waist to ensure minimal muscle activity. Here, subjects maintained a small forward lean angle. The higher muscle contraction levels in this experiment can explain the higher stiffness.

The lean angle and corresponding muscle contraction levels are not expected to differ greatly from normal quiet standing. In (Winter et al., 2001), an average sway angle during quiet standing of 0.064 rad (3.67 degrees) is reported. This angle was estimated using the whole body COM deflection from the ankle joint. The accompanying average total exerted torque was 55 Nm. This is in the same range as our average onset torque (26 Nm per ankle).

The actual ankle stiffness could not be accurately estimated for the smaller rotations due to the limited spatial and temporal resolution of the motion capture device. Nevertheless, it is the ankle-foot stiffness that is relevant to standing balance. The higher ankle-foot stiffness found for the 0.005 rad plantar flexion compared to the dorsiflexion is mainly caused by the marker data, as the difference is less pronounced in the pseudo ankle-foot stiffness. A similar high stiffness is not found for the BR. Given the inter-distance of the markers, these rotations are near the accuracy of the motion capture system, possibly causing these effects.

Rotational amplitude dependency

The observed decrease in stiffness with increasing rotation amplitude is likely caused by the contractile tissue. A Hill-type muscle model would suggest mainly tendon stretch at high velocities due to the viscous properties of the muscle. However, in our protocol the different rotations had a fixed duration, hence the largest are also the fastest. A Hill-type model does therefore not apply to the data, because it cannot predict a decrease in stiffness for increasing stretch velocity. Furthermore, tendon stiffness does

not decrease with increasing stretch (Maganaris, 2002), hence cannot explain the observed decrease. In (Loram et al., 2007a, b) it was shown that the contractile tissue is stiffer for smaller rotations, while the stiffness of the series elastic element remained largely constant over various rotation amplitudes. This suggests that the observed decrease relates to the muscle. In (de Vlugt et al., 2011), model fits to the human wrist revealed that the short-range stiffness of the total muscle-tendon complex did not vary with angular velocities between 1 and 4 rad s⁻¹. Their data also showed that short-range stiffness lasted for 30 ms regardless of the stretch velocity. Therefore, the short-range stiffness is also independent of the rotational amplitude within this velocity range. Assuming this property holds in our velocity range (0.125–2 rad s⁻¹), the observed decrease might be attributed to changes in muscle stiffness occurring after short-range stiffness effects. Initially the entire muscle–tendon complex might behave as a spring with constant stiffness, after which the muscle stiffness decreases due to detaching crossbridges.

Effects of rotational velocity

The rotational velocity decreased with decreasing rotation amplitude, possibly introducing a velocity effect. In cat soleus muscle the stiffness was shown to be independent of stretch velocity for stretches faster than 8 mm/s, and decreased for slower movements (Rack and Westbury, 1974). Here, velocity effects would lead to a lower stiffness for smaller rotations. Consequently, applying all perturbations with the same angular velocity could lead to an even steeper decrease in stiffness with increasing rotation amplitude.

Uni- and bidirectional rotations

The systematic differences in stiffness found for UR and BR might be caused by body movement in response to the rotation. The mechanical coupling of both legs through the pelvis might explain the observed differences in torque response between UR and BR. The UR could lead to slight COM movement in the same direction as the perturbation, whereas BR could lead to rotation about the body's longitudinal axis. Hence, effects of the COM on the measured torque might not be completely disregarded.

Recommendations

Our results stress the need for accurate measurement of the differences between applied and actual ankle(-foot) angle. This might be improved by using dedicated equipment, such as a laser as in (Loram and Lakie, 2002). Unwanted effects caused by the rotations, such as a loss of balance, might be reduced by using BR.

Although not the purpose of this study, a further reduction in rotation duration might be required to solely estimate short-range stiffness effects using the current method. We believe applying sufficiently fast rotations to rule out unwanted effects can be a valid method to investigate muscle–tendon properties and their contributions to standing balance. With the proposed method reflexive effects can be ruled out from the torque response, without any model assumptions.

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CHAPTER 3

Center of mass velocity-based predictions in balance recovery

"up up down down left right left right B A"

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**Center of mass velocity-based predictions in balance recovery
following pelvis perturbations during human walking**

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Abstract

In many simple walking models, foot placement dictates the center of pressure location and ground reaction force components, whereas humans can modulate these aspects after foot contact. Because of the differences, it is unclear to what extent predictions made by models are valid for human walking. Yet, both model simulations and human experimental data have previously indicated that the center of mass (COM) velocity plays an important role in regulating stable walking. Here, perturbed human walking was studied to determine the relation of the horizontal COM velocity at heel strike and toe-off with the foot placement location relative to the COM, the forthcoming center of pressure location relative to the COM, and the ground reaction forces. Ten healthy subjects received mediolateral and anteroposterior pelvis perturbations of various magnitudes at toe-off, during 0.63 and 1.25 ms⁻¹ treadmill walking. At heel strike after the perturbation, recovery from mediolateral perturbations involved mediolateral foot placement adjustments proportional to the mediolateral COM velocity. In contrast, for anteroposterior perturbations, no significant anteroposterior foot placement adjustment occurred at this heel strike. However, in both directions the COM velocity at heel strike related linearly to the center of pressure location at the subsequent toe-off. This relation was affected by the walking speed and was, for the slow speed, in line with a COM velocity-based control strategy previously applied by others in a linear inverted pendulum model of walking. Finally, changes in gait phase durations suggest that the timing of actions could play an important role during the perturbation recovery.

Introduction

Humans are currently unparalleled when it comes to bipedal walking. Despite a relative high located center of mass (COM) and small base of support (BoS), movement can be maintained or altered at will. Various strategies such as adjustments to the location and timing of foot placement, and adjustments to ankle and hip torques can be addressed to control balance during unconstrained walking. These strategies affect the magnitude, direction and point of application of the ground reaction force, with the point of application being the center of pressure (COP). The force components affect the COM acceleration. Together with the COP location relative to the COM, they also determine the angular acceleration of the whole body about the COM. Predicting how healthy humans shift the COP and modulate the ground reaction force could aid in fall prevention in humans, exoskeletons and bipedal robots, as faulty weight shifting could easily lead to a fall (Robinovitch et al., 2013).

Simple models such as the inverted pendulum have been extensively used to describe human locomotion (Townsend, 1985; Kajita and Tani, 1991; Winter, 1995; Garcia et al., 1998; Kuo, 2001). These models incorporate foot placement, which is considered a major strategy in directing locomotion in both mediolateral (ML) and anteroposterior (AP) movement directions (MacKinnon and Winter, 1993; Patla, 2003). However, these models do not capture several other important aspects of walking (see Figure 1A). First, the double support phase is often neglected. Consequently, the COM does not move between heel strike and toe-off. Second, the area of the foot is often infinitesimal, such that foot placement fully determines the COP location during single support. Third, if no inertia properties are present in the model, the ground reaction force will always pass exactly through the COM. This way, foot placement also fully determines the ground reaction force components. In humans, the COP makes a continuous shift from the trailing foot at heel strike to the leading foot at toe-off (Jian et al., 1993), during which the COM continues to move. Furthermore, humans have mechanisms other than foot placement to alter the COP location and ground reaction force. Hence, many inverted pendulum models might not correctly represent the spatio-temporal location of the COP as well as the ground reaction forces following foot placement.

Despite the differences, both model simulations and data collected in humans suggest an important role of the COM velocity in regulating stable walking. The horizontal position and velocity of the COM have predictive properties in human foot placement. Using the pelvis as an approximation of the COM, a linear function of the ML pelvis position and velocity relative to the stance foot at mid-stance could be used to predict over 80% of the variance in ML foot placement during unperturbed human treadmill walking (Wang and Srinivasan, 2014). Pelvis predictive power was lower for AP foot placement, explaining just over 30% at midstance. In a 3D spring-loaded inverted pendulum model, stable running (Peucker et al., 2012) and walking (Maus and Seyfarth, 2014) could be realized by setting the swing leg angle of attack proportional to the angle of the COM velocity vector with the vertical. Both studies reported that this leads to increased stability compared with strategies that did not take into account COM velocity. In a planar bipedal robot, stable running could be achieved by setting the swing leg angle of attack proportional to the horizontal COM velocity (Hodgins and Raibert, 1991). Foot placement strategies directly proportional to the horizontal COM velocity were also derived from a linear inverted pendulum model's energy orbits, which allowed a low-dimensional robot to walk for several steps (Kajita et al., 1992).

Figure 1. Schematic overview of concepts in human and inverted pendulum model walking. (A) At heel strike, the center of pressure is near the trailing foot in humans. It takes until the subsequent toe-off for it to fully shift to the leading leg. In many inverted pendulum models, the instances of heel strike and toe-off coincide, and so do the foot and COP location. Furthermore, the ground reaction force often passes exactly through the COM in these models. (B) The XCOM is a concept derived from a linearized inverted pendulum model, that can be considered as a point on the floor at a distance from the COM that is directly proportional to the COM velocity. Moving the COP beyond the XCOM makes the pendulum fall back in the direction it came from. Moving the COP before the XCOM makes the pendulum topple over the COP. Placing the COP exactly in the XCOM brings the pendulum to an upright stop.

A special case of these energy orbits (zero energy) can be used to obtain the extrapolated center of mass (XCOM) (Hof et al., 2005) or capture point (Pratt et al., 2006). This concept can be conceived as a point on the floor at a horizontal distance from the COM that is directly proportional to the horizontal COM velocity (see Figure 1B). The proportionality constant is $\omega_0^{-1} = \sqrt{l/g}$, in which g is the Earth's gravitational constant and l is the pendulum (leg) length. It is the reciprocal of the eigenfrequency of a linear inverted pendulum model. This model can come to an upright stop by placing the COP in the XCOM. In simulations, stable walking could be achieved by placing the COP at a fixed offset from the XCOM in both the ML and AP directions, using a fixed step time (Hof, 2008). This 'constant offset control' allowed the model to return to a stable gait after perturbing the COP location at heel strike. This concept is supported by experimental data, suggesting that humans also apply ML constant offset control in both normal (Hof et al., 2007) and ML perturbed walking (Hof et al., 2010). In the former work, it was concluded that foot placement is the primary strategy for realizing the ML COP offset, and that an ankle torque allows minor COP adjustments through feedback after the foot has been placed. This also means that the offset can be realized in ways other than foot placement alone, not captured by the linear inverted pendulum model from which the XCOM concept is derived. This is especially the case in the AP direction, not investigated in Hof's work, where COP shifts are most feasible because of the dimensions of the foot.

Simple walking models and the concepts derived from them can give insight into human balance control, but might also fail to accurately describe human balance during walking because of their simplicity. In many inverted pendulum models, foot placement is directly linked to the COP location

and the ground reaction force components. As these concepts are not strictly linked in humans, it is unclear to what extent predictions made by these models are valid for human walking. Yet, both model simulations and human experimental data suggest some proportionality of one or more of these concepts with the COM velocity. In this study, we investigate relations between the horizontal COM velocity and (1) the location of the foot relative to the COM, (2) the location of the COP relative to the COM and (3) the ground reaction force components. Only the instances of the first heel strike and toe-off following ML and AP perturbations are chosen for analysis. These are often a single key instance in inverted pendulum models, at which the model state determines subsequent ballistic motion. Variables 2 and 3 will only be investigated at toe-off, because these are not influenced yet by foot placement at the instance of heel strike.

Methods

Participants

Ten healthy volunteers with no known history of neurological, muscular or orthopedic problems participated in the study (five men, age 25 ± 2 yr, weight 67 ± 12 kg, height 1.80 ± 0.11 m, means $\pm$ s.d.). The setup and experimental protocol were approved by the local ethics committee. All subjects gave prior written informed consent in accordance with the Declaration of Helsinki.

Apparatus

Subjects walked on a dual-belt instrumented treadmill (custom Y-Mill, Motekforce Link, Culemborg, The Netherlands). A force plate beneath each belt was used to measure 3 degrees-of-freedom ground reaction forces and moments. To perturb subjects in both ML and AP directions, two motors (SMH60, Moog, Nieuw-Vennep, The Netherlands) were located adjacent to the treadmill, one to the right and one at the rear. The motors were bolted onto a steel support structure that was tightly clamped to the exterior frame of the treadmill, without influencing the force plates. Each motor had an aluminum lever arm (0.3 m) attached to its rotational axis, onto which a load cell (model QLA131, FUTEK, Los Angeles, CA, USA) was located for torque sensing. A ball joint was located at the end of each lever arm, to which an aluminum rod (0.8 m in length, 0.3 kg) could be attached. The other end of each rod could be attached to the right or rear of a modified universal hip abduction brace (Distrac Wellcare, Hoegaarden, Belgium), also using a ball joint. The brace (0.9 kg) could be tightly worn around the pelvis by the subject. With the lever arms in neutral position (vertical), the ball joints of the lever arm were 1 m above the walking surface of the treadmill, such that the rods were approximately horizontal when the brace was worn by a subject. The maximum possible excursion of each motor was 1.1 rad in each direction of the neutral position, allowing up to 0.55 m pelvis excursion. A schematic overview of the setup is shown in Figure 2. Motor control signals were generated at 1000 Hz using xPC-target (MathWorks, Natick, MA, USA) and sent to the motor drivers over ethernet (User Datagram Protocol), using a dedicated ethernet card (82558 ethernet card, Intel, Santa Clara, CA, USA).

Data collection

Motor torque and encoder angle were collected at 1000 Hz using the ethernet card. Kinematic data were acquired using a 12-camera motion capture system (Visualeyez II, Phoenix Technologies, Burnaby, Canada). In total, nine three-LED marker frames were placed on the subject. Frame locations were on

the feet, lower legs, upper legs, the front of the pelvis below the strap of the brace, the sternum and the head. Additional single LEDs were placed on the lateral epicondyles of the femur and on the lateral malleoli. Ground reaction force data were captured at 1000 Hz using a PCI-6229 AD card (National Instruments, Austin, TX, USA), also using xPC target software. The same card was used to generate an analog signal for synchronization with the motion capture system.

Protocol

Prior to the experiment, several kinematic measurements were taken during which bony landmarks were indicated using an LED-based probe, as described in (Cappozzo et al., 1995). Captured landmarks were the calcaneus, first and fifth metatarsal heads, medial and lateral malleoli, fibula head, medial and lateral epicondyles of the femur, greater trochanter, anterior and posterior superior iliac spines, xiphoid process, jugular notch, seventh cervical vertebra, occiput, head vertex and nasal sellion (Dumas et al., 2007).

Figure 2. Schematic overview of the perturbation setup. Two motors adjacent to a dual belt instrumented treadmill can be used to perturb the subject at the pelvis during walking. Colored arrows indicate the direction of the different perturbation magnitudes of 0.04, 0.08, 0.12 and 0.16 * body weight (positive: yellow-red, negative: green-blue). An inward perturbation is regarded as a perturbation toward the (left) stance leg, an outward perturbation away from the stance leg.

During the experiment, subjects were instructed to walk on the treadmill with their arms crossed over the abdomen. A safety harness was worn to prevent injury in case of a fall. The brace was tightly worn around the pelvis. Subjects walked four blocks of three trials each. The first trial of each block was a 2 min unperturbed walking trial, the second and third were perturbation trials. In two blocks, subjects were attached to the right motor, and in the other two blocks to the rear motor. Subjects were never attached to both motors simultaneously to minimize restraints. The attachment order was randomized. For each motor attachment site, subjects walked one block at a slow speed ($0.63 * \sqrt{l} \text{ ms}^{-1}$) and one at a normal speed ($1.25 * \sqrt{l} \text{ ms}^{-1}$), where $\sqrt{l}$ is the square root of the subject's leg length; (Hof, 1996). Subjects walked the slow trials first, followed by the normal trials for the same motor. In addition to the required rest after two blocks, subjects were free to take breaks between trials.

During perturbation trials subjects randomly received perturbations at toe-off right (TOR), detected using the vertical ground reaction force (threshold 5% body weight). Toe-off was chosen for perturbation onset to maximally allow foot placement adjustment, while preventing push-off modulation in response to the disturbance. A random interval of approximately 6–12 s was given between perturbations. Perturbation signals consisted of 150 ms block pulses resulting in force magnitudes equal to 4, 8, 12 and 16% of the subject's body weight. Perturbation force direction was either inward (negative sign, leftward for right swing leg) and outward (positive sign, rightward for right swing leg) or backward (negative sign) and forward (positive sign), depending on the motor in use (see Figure 2). Each condition was repeated eight times, giving 256 perturbations in total (32 per trial). All perturbations were randomized over magnitude and direction within each block. When no perturbation was being applied, the motors were admittance controlled, actively regulating the interaction force between subject and motor to (near) zero.

Data processing

All data were processed using MATLAB (R2014b, MathWorks). Raw perturbation forces were integrated to obtain the impulse delivered by the motors. Ground reaction force and moment data were filtered with a 4th order 40 Hz zero-phase low-pass Butterworth filter before calculating a COP location. Marker data were filtered with a 4th order 20 Hz zero-phase low-pass Butterworth filter. Local landmark positions (relative to their respective marker frames) were extracted from the probe measurements. In each trial, the global landmark positions were reconstructed using least squares estimation of a rotation matrix and a displacement vector between the local and global marker frame coordinates (Söderkvist and Wedin, 1993).

Landmark data of the feet were used to detect the gait phase, comparable to (Zeni Jr et al., 2008). The maximum backward excursion of the metatarsal head I was used to detect toe-off. Heel strike was detected as the instance at which the AP calcaneus velocity stopped decreasing following its largest forward excursion. Furthermore, landmarks were used to estimate the locations of the ankle, knee, hip, lumbar and cervical joints, as well as the COM locations of the feet, lower legs, upper legs, pelvis, torso and head (Dumas et al., 2007). The available segment COM locations were used to calculate a weighted total COM location.

Unperturbed walking data were used as baseline for the trials with a corresponding walking speed. All data were made dimensionless according to (Hof, 1996). For each subject, the baseline average Euclidean distance between the COM of the feet at heel strike was used as the length scaling value (l_0).

Subject mass was used to scale forces. Perturbation onsets were identified from the motor reference signals. All perturbation data were cut into sequences from 0.5 s before to 2.5 s after perturbation onset, and were sorted by perturbation type and walking speed. All position and velocity data were expressed relative to those of the COM. The velocity of the COM itself was expressed relative to the walking surface by adding the belt speed to the AP COM velocity.

Linear least squares fits of the form $y = a\dot{x} + b$ were made to the subject average data. The independent variable $\dot{x}$ was the ML or AP COM velocity at HSR or TOL. The dependent variable y was one of the following: the distance between the COM and the COM of the leading foot; the distance between the COM and the COP; a horizontal ground reaction force component; ratio RF; or ratio RD (each in the ML or AP direction at HSR or TOL). For comparison, a dimensionless XCOM proportionality constant (ω_0^{-1}) was calculated for each subject. These were subsequently used to find a subject average proportionality constant and a subject average ML or AP $XCOM = \omega_0^{-1} * \dot{x}$, where the XCOM is relative to the COM, and $\dot{x}$ corresponds to the horizontal ML or AP COM velocity at any given instance.

Linear mixed models were used to assess the effects of the perturbation (fixed factor, with intercept) and walking speed (fixed factor, with intercept) on the distance between the COM and the COM of both feet at HSR and TOL, the distance between the COM and the COP at TOL, the ground reaction force components at TOL, as well as the duration of the single and double support phase following the perturbation. Subject effects were included as a random factor (intercept) to account for the correlation between repeated measurements within a single subject. A significance level of $\alpha = 0.050$ was used and a Bonferroni correction was applied to correct for multiple comparisons during post hoc analysis. In the latter, the perturbed conditions were only compared with the unperturbed walking condition and not mutually. SPSS statistics 21 (IBM Corporation, Armonk, NY, USA) was used for the statistical analysis.

Results

Subject balance responses were assessed following ML and AP perturbations during both slow and normal walking. Here results will only be visualized for the slow walking speed. The normal walking speed yielded mostly comparable results (see Figures S1–S3). Statistical values apply to both slow and normal walking speeds unless indicated otherwise. Subject average data are shown dimensionless. Subject average scaling values for the slow walking speed are $l_0 = 0.44 \pm 0.04$ m for distances, $\sqrt{(g * l_0)} = 2.08 \pm 0.10$ m s⁻¹ for velocities, and $\sqrt{(l_0 / g)} = 0.21 \pm 0.01$ s for durations, where l_0 is the average Euclidean distance between the COM of the feet at heel strike during unperturbed walking. For the normal walking speed, scaling values are $l_0 = 0.63 \pm 0.06$ m, $\sqrt{(g * l_0)} = 2.48 \pm 0.11$ m s⁻¹, and $\sqrt{(l_0 / g)} = 0.25 \pm 0.01$ s.

Perturbations

Perturbations of various magnitudes (± 0.04 , ± 0.08 , ± 0.12 , ± 0.16 * body weight) were applied to the subject's pelvis at TOR using two admittance-controlled motors. Although the motors cannot exactly track the reference block pulses, the integrals of the reference and measured perturbation force are similar. Effects of the different perturbations on the horizontal COM velocity can be clearly distinguished (see Figure 3).

Balance responses

Various balance responses were observed to recover from the perturbation (see Figure 4). At HSR, the leading foot was placed further inward (leftward for right swing leg) or outward (rightward for right swing leg) with increasing inward (ML, negative sign) or outward (ML, positive sign) perturbation magnitude, respectively. The ML distance between the COM and the leading foot was significantly affected by the ML perturbation magnitude ($F_{8,153}=363.005$, $P<0.001$), walking speed ($F_{1,153}=71.916$, $P<0.001$) and their interaction ($F_{8,153}=9.300$, $P<0.001$). For slow walking, the distance between the COM and the leading foot was significantly different from that in unperturbed walking for all but the lowest magnitude perturbations ($P\leq 0.001$). For the normal speed, this was the case for all but the lowest magnitude outward perturbations ($P\leq 0.025$).

Figure 3. Typical single subject anteroposterior perturbation profile. (A) Reference (dashed) and measured (solid) motor force. (B) Reference (dashed) and measured (solid) motor impulse, obtained by integrating the motor forces. (C) AP COM velocity relative to the walking surface. Colors indicate the various perturbation magnitudes as a ratio of body weight. Lines are within subject averages for a single subject. Shaded area's indicate the within subject standard deviation.

At TOL, the ML distance between the COM and the leading foot was decreased compared with that at HSR, but was still significantly affected by ML perturbation magnitude ($F_{8,153}=351.252$, $P<0.001$), speed ($F_{1,153}=15.283$, $P<0.001$) and their interaction ($F_{8,153}=3.899$, $P<0.001$). This distance was significantly different from that in unperturbed walking for all but the lowest inward perturbation magnitudes for both slow ($P\leq 0.033$) and normal ($P\leq 0.009$) walking speeds. The ML distance between the COM and the COP showed similar effects of ML perturbation magnitude ($F_{8,153}=399.611$, $P<0.001$), speed ($F_{1,153}=20.970$, $P<0.001$) and their interaction ($F_{8,153}=5.225$, $P<0.001$). This distance was significantly different from that in unperturbed walking for all but the lowest magnitude inward perturbations, for both slow ($P\leq 0.006$) and normal ($P\leq 0.002$) walking speeds.

Figure 4. Positions of the COM of the feet and the COP relative to the COM. (A) At HSR, the location of the COM of the leading and trailing foot relative to the COM in (0,0), for ML perturbations. (B) Same as (A) for AP perturbations. (C) At TOL, location of the COM of the leading foot relative to the COM for AP perturbations. (D) At TOL, location of the COP relative to the COM for AP perturbations. Triangles show subject averages and indicates the perturbation direction. Ellipses represent the subject standard deviation. Colors indicate the various perturbation magnitudes. Data shown is dimensionless and for slow walking only.

The ML ground reaction force at TOL also changed significantly with perturbation magnitude ($F_{8,153}=489.051$, $P<0.001$), speed ($F_{1,153}=9.849$, $P=0.002$) and their interaction ($F_{8,153}=26.742$, $P<0.001$). With the exception of the lowest magnitude perturbations for slow walking, all ML perturbations led to ML forces at TOL significantly different from that in unperturbed walking (slow: $P\leq 0.001$, normal: $P\leq 0.043$). Although the vertical force at TOL was also significantly affected by ML perturbation magnitude ($F_{8,153}=10.506$, $P<0.001$), speed ($F_{1,153}=401.749$, $P<0.001$) and their interaction ($F_{8,153}=3.440$, $P=0.001$), it was not significantly different from the vertical force in unperturbed walking for any ML perturbation during slow walking ($P\geq 0.209$). For the normal walking speed it was significantly different for the larger (-0.12 , ± 0.16) perturbation magnitudes ($P\leq 0.002$).

For AP perturbations, subjects barely adjusted the AP distance between the COM and the leading foot at HSR. Although this distance was significantly affected by the AP perturbation magnitude ($F_{8,153}=2.650$, $P=0.009$), speed ($F_{1,153}=50.985$, $P<0.001$) and their interaction ($F_{8,153}=5.094$, $P<0.001$), it was not significantly different from that in unperturbed walking for any AP perturbation for either slow ($P\geq 0.124$) or normal ($P\geq 0.324$) walking speeds. The AP distance between the COM and the trailing foot was also significantly affected by AP perturbation magnitude ($F_{8,153}=65.671$, $P<0.001$), speed ($F_{1,153}=86.310$, $P<0.001$) and their interaction ($F_{8,153}=4.658$, $P<0.001$). For slow walking, this distance was significantly different from that in unperturbed walking following all but the lowest magnitude AP perturbations ($P\leq 0.017$). For the normal walking speed, it was different for the larger magnitude (± 0.12 , ± 0.16) AP perturbations ($P<0.013$).

Figure 5. Gait phase duration following perturbations. Gait phase duration directly following ML (A) and AP (B) perturbations. Single and double support phase durations are indicated by the open and filled markers respectively. Triangles show subject averages and indicate the perturbation direction. Error bars indicate the subject standard deviation. Colors indicate the various perturbation magnitudes. Asterisks (*) indicate significant differences from the corresponding unperturbed phase duration. Double asterisks (**) indicate that there was no significant interaction effect between slow and normal walking, such that the corresponding p -values represent both slow and normal walking speeds. Data shown is dimensionless and for slow walking only.

At TOL, the AP distance between the COM and the leading foot showed more of an effect of the AP perturbation magnitude than at HSR. This distance at TOL was significantly affected by AP perturbation magnitude ($F_{8,153}=20.149$, $P<0.001$) and speed ($F_{1,153}=139.137$, $P<0.001$), but not by their interaction ($F_{8,153}=1.563$, $P=0.140$). This distance was significantly different from that in unperturbed walking for the larger magnitude (0.08, 0.12 and 0.16) forward and -0.16 backward perturbations ($P\leq 0.030$). The AP distance between the COM and the COP at TOL shows more variation in the means than the AP distance between the COM and the leading foot at TOL. This distance between COM and COP was significantly affected by AP perturbation magnitude ($F_{8,153}=65.583$, $P<0.001$), speed ($F_{1,153}=64.175$, $P<0.001$) and their interaction ($F_{8,153}=3.517$, $P=0.001$). For slow walking, this distance was significantly different from that in unperturbed walking for the larger magnitude (0.08, 0.12 and 0.16) forward and -0.16 backward perturbation ($P\leq 0.001$). This is similar for the normal walking speed, which also showed a significant difference for the -0.12 backward perturbation ($P\leq 0.013$). Subjects adjusted the AP ground reaction force at TOL significantly with AP perturbation magnitude ($F_{8,153}=122.686$, $P<0.001$), speed ($F_{1,153}=677.983$, $P<0.001$) and their interaction ($F_{8,153}=11.086$, $P<0.001$). For slow walking, AP forces were significantly different from that in unperturbed walking for the larger magnitude (0.08, 0.12 and 0.16) forward and -0.16 backward perturbations ($P\leq 0.011$). For the normal walking speed, all AP perturbations led to significant differences ($P\leq 0.043$). The vertical force component was significantly affected by AP perturbation magnitude ($F_{8,153}=79.415$, $P<0.001$), speed ($F_{1,153}=583.701$, $P<0.001$) and their interaction ($F_{8,153}=32.201$, $P<0.001$). However, with the exception of the largest forward perturbation magnitude, none of the AP perturbations led to vertical forces significantly different from that in unperturbed slow walking ($P\geq 0.875$). In contrast, for the normal speed, all but the smallest forward perturbation led to significant differences ($P\leq 0.019$). Finally, both the ML and AP perturbations had a significant effect on the single support duration during which the perturbation was applied (ML: $F_{8,153}=47.370$, $P<0.001$, AP: $F_{8,153}=7.581$, $P<0.001$), as well as on the following double support duration (ML: $F_{8,153}=8.941$, $P<0.001$, AP: $F_{8,153}=51.762$, $P<0.001$). Walking speed also significantly affected these single (ML: $F_{1,153}=715.091$, $P<0.001$, AP: $F_{1,153}=1354.447$, $P<0.001$) and double support durations (ML: $F_{1,153}=1313.883$, $P<0.001$, AP: $F_{1,153}=2073.293$, $P<0.001$). Interaction effects of perturbation magnitude and walking speed only occurred for ML perturbations in both single ($F_{8,153}=12.833$, $P<0.001$) and double support durations ($F_{8,153}=4.412$, $P<0.001$). Durations significantly different from that in unperturbed walking and the corresponding P-values can be found in Figure 5 and Figure S1 for slow and normal walking, respectively.

Relation with COM velocity

The relation between the horizontal COM velocity and (1) the location of the foot relative to the COM, (2) the location of the COP relative to the COM and (3) the ground reaction force components were investigated at the instances of the first HSR and TOL following perturbation onset at TOR. Combinations of instances (HSR, TOL) were also investigated, analogous to walking models without a double support phase. The coefficients of determination (R^2) of the linear least squares fits made to these data are shown in Table 1. Results for the COP, forces, RF and RD at HSR were omitted, as these are not yet affected by foot placement at this instance. For several fits, corresponding data are shown in Figures 6 and 7 for slow walking, and in Figures S2 and S3 for normal walking speed.

Table 1) Coefficient of determination (R^2)* of the linear least squares fits made to the data for slow and normal walking speeds. Underlined values correspond with a fit of which the root mean square error is less than 5 percent of the range of the dependent variable.

ML Perturbations				
	ML COM velocity at HSR		ML COM velocity at TOL	
	Slow	Normal	Slow	Normal
ML distance foot-COM at HSR	0.983	<u>0.996</u>	<u>0.986</u>	0.978
ML distance foot-COM at TOL	<u>0.989</u>	<u>0.996</u>	0.966	0.968
ML distance COP-COM at TOL	<u>0.993</u>	<u>0.997</u>	0.964	0.968
ML ground reaction force at TOL	<u>0.998</u>	0.998	0.957	0.968
ML R_F at TOL	<u>0.997</u>	<u>0.996</u>	0.955	0.968
ML R_D at TOL	<u>0.994</u>	<u>0.997</u>	0.964	0.967
AP Perturbations				
	AP COM velocity at HSR		AP COM velocity at TOL	
	Slow	Normal	Slow	Normal
AP distance foot-COM at HSR	0.281	0.917	0.069	0.941
AP distance foot-COM at TOL	0.851	0.916	0.426	0.847
AP distance COP-COM at TOL	<u>0.982</u>	0.974	0.672	0.920
AP ground reaction force at TOL	<u>0.988</u>	<u>0.992</u>	0.696	0.963
AP R_F at TOL	<u>0.983</u>	0.973	0.668	0.923
AP R_D at TOL	<u>0.984</u>	0.973	0.681	0.916

For ML perturbations, the ML distance between the COM and the leading foot changed directly proportional with the ML COM velocity at HSR. Similar effects can be observed for the ML distance between the COM and the COP (see Figure 6, Figure S2). At TOL, there are only minor differences in the ML distance between the COM and the leading foot, and between the COM and the COP. This gives rise to approximately the same linear relations. For distances in the ML direction, the strongest linear relations were found between the COM velocity at HSR and the distance between the COM and the COP at TOL. For slow walking, the slope of the fit to this data ($y=1.54\dot{x}+0.03$, $R^2=0.993$) corresponds well with the subject average dimensionless ω_0^{-1} (1.46 ± 0.04). For the normal speed, this slope ($y=1.48\dot{x}+0.04$, $R^2=0.99$) shows more deviation from the corresponding ω_0^{-1} (1.23 ± 0.04). Similarly, the COM velocity at HSR had the strongest linear relations with ML forces and ratios at TOL. For slow walking, the vertical ground reaction force at TOL is approximately 1 for most perturbations. Consequently, the ML force at TOL ($y=-0.74\dot{x}-0.01$, $R^2=0.998$) and the ML RF at TOL

($y = -0.70\dot{x} - 0.01$, $R^2 = 0.997$) have similar relations with the ML COM velocity at HSR. This also holds for RD at TOL ($y = -0.70\dot{x} - 0.01$, $R^2 = 0.994$), see Figure 7, such that the total ground reaction force in the frontal plane at TOL points approximately toward the COM for all ML perturbations. For the normal walking speed, these similarities between RF ($y = -0.97\dot{x} - 0.02$, $R^2 = 0.996$) and RD ($y = -0.97\dot{x} - 0.02$, $R^2 = 0.997$) also exist (see Figure S3). However, the ML force shows a different relation ($y = -1.21\dot{x} - 0.02$, $R^2 = 0.998$), as the vertical force component tends to increase with ML perturbation magnitude.

Figure 6. Various relations between COM velocity and both leading foot and COP. (A) ML COM velocity at HSR vs ML distance between COM and leading foot at HSR, for ML perturbations. (B) Same as (A) in the AP direction for AP perturbations. (C) ML COM velocity at HSR vs ML distance between COM and COP at TOL for ML perturbations. (D) Same as (C) in the AP direction for AP perturbations. Triangles show subject averages and indicates the perturbation direction. Ellipses represent the subject standard deviation. Colors indicate the various perturbation magnitudes. Pink solid and dashed lines indicates the subject average and standard deviation of the XCOM relative to the COM. This XCOM line has slope 0.0-1 and no intercept. Black dashed line is a linear least square fit to the data. R2 indicates the coefficient of determination of the fit. Data shown is dimensionless and for slow walking only.

For AP perturbations, the AP distance between the COM and the leading foot at HSR shows only minor changes with AP COM velocity at HSR (slow: $y=0.27\dot{x}+0.43$, $R^2=0.281$, normal: $y=-0.34\dot{x}+0.66$, $R^2=0.917$; see Figure 6, Figure S2). As in the ML direction, the strongest linear relations were found between the AP COM velocity at HSR and the AP distance between the COM and the COP at TOL. Again, for slow walking, the slope of the fit to this data ($y=1.49\dot{x}-0.31$, $R^2=0.982$) corresponds well with ω_0^{-1} . This is less the case for the normal speed ($y=0.87\dot{x}-0.26$, $R^2=0.97$). Also for AP forces and ratios at TOL, the strongest linear relations were found with the AP COM velocity at HSR. The vertical ground reaction force at TOL tends to change with AP perturbation magnitude. Consequently, for slow walking, the relations of both the AP force ($y=-0.84\dot{x}+0.19$, $R^2=0.988$) and the AP RF ($y=-0.73\dot{x}+0.17$, $R^2=0.983$) with the AP COM velocity at HSR are less similar compared with those in the ML direction following ML perturbations. However, comparison of the fit to the AP RF with the fit to the AP RD ($y=-0.68\dot{x}+0.14$, $R^2=0.984$; Figure 7) suggests that the total ground reaction force in the sagittal plane points approximately toward the COM at TOL, for all perturbations. For the normal walking speed, similar comparisons can be made between the relations of the AP COM velocity at HSR with the AP force ($y=-1.19\dot{x}+0.48$, $R^2=0.992$), AP RF ($y=-0.74\dot{x}+0.27$, $R^2=0.973$) and AP RD ($y=-0.57\dot{x}+0.17$, $R^2=0.973$). Comparing RF and RD for normal walking speed (Figure S3) suggests that the total ground reaction force in the sagittal plane at TOL tends to point above the COM for backward perturbations and below the COM for forward perturbations.

Figure 7. Relations between COM velocity and both force and distance ratios. (A) ML COM velocity at HSR vs ML force ratio RF at TOL. For distance ratio RD only a fit to the data is shown for comparison. (B) Same as (A) in the AP direction for AP perturbations. Triangles show subject averages and indicates the perturbation direction. Ellipses represent the subject standard deviation. Colors indicate the various perturbation magnitudes. Black and gray dashed lines are linear least square fits to the data, for RF and RD respectively. The R^2 indicates the coefficient of determination of the fits. Data shown is dimensionless and for slow walking only.

Discussion

Walking human subjects were perturbed in the horizontal plane at the start of the single support phase. The distance between the COM and the COP at toe-off, as well as the horizontal ground reaction force, increased linearly with increasing horizontal COM velocity at the preceding heel strike, in both ML and AP directions. In the ML direction, foot placement is crucial to realize these COP relations given the limited possibilities for ML COP displacement within the foot. In the AP direction, other contributions such as an ankle torque are key in adjusting the COP location and the ground reaction force. Furthermore, gait phase durations varied following the perturbations, especially for ML perturbations. In the following sections, the subject responses will be discussed in order of occurrence following the perturbation.

Single support duration

Humans show variations in foot placement timing for the first recovery step. By controlling the swing leg, humans can choose from a tremendous amount of spatio-temporal options for foot placement. Yet all subjects show similar consistent spatial and temporal responses, suggesting a preferred recovery strategy among all possible options. This could arise from a trade-off between the energetic costs of leg swing against the expected cost for recovery after foot placement, in a way similar to that predicted for a preferred step frequency during normal walking (Kuo, 2001).

The single support duration might shorten with increasing deviation of the COM from the desired walking direction. This is mainly supported by the durations following ML perturbations. When the COM is pushed away from or over the BoS of the stance foot, the need for lateral corrections increases. This leads to a decreased single support duration. For the lower magnitude inward perturbations, the COM is laterally pushed toward but not over the stance foot. This way there is no direct need to correct for lateral imbalance, which can even increase the single support duration. After completion of any AP perturbation, the COM is still moving in the desired forward direction, possibly leading to little need to adjust the single support duration. Furthermore, effects of the AP perturbations can be partially counteracted by modulating the ankle torque of the left stance foot directly after the perturbation has been applied (not shown). A possible explanation for the increased single support duration following larger backward perturbations is that subjects wait to regain forward velocity.

In (Hof et al., 2010), especially inward perturbations led to a decrease in single support duration with increasing perturbation magnitude. Although this appears to contradict the current results, Hof's perturbations were applied shortly before heel strike, mainly affecting the subsequent swing phase. Hence, temporal results for inward perturbations at HSR in Hof's work are most comparable with results for outward perturbations in this work. Significant increases in single support duration were not reported in (Hof et al., 2010), most likely because of the walking speed of 1.25 m s^{-1} . In the present study, no significant increase in single support duration was found following ML perturbations during the normal walking speed.

Foot placement

The ML COM velocity at HSR has a strong predictive value for ML foot placement. This is in line with previous findings in ML foot placement (Hof et al., 2007; Wang and Srinivasan, 2014). Findings in unperturbed walking have suggested that the AP COM velocity at mid-stance also significantly

contributes to predictions of the next AP foot placement location (Wang and Srinivasan, 2014). This location was expressed relative to the trailing foot, and therefore contained effects occurring between the COM and both the leading and the trailing foot. Our results suggest that the findings for AP foot placement in (Wang and Srinivasan, 2014) are mainly caused by changes between the COM and the trailing foot. Here, none of the AP perturbations led to a distance between the COM and the leading foot that was significantly different from the distance in unperturbed walking. Humans might choose not to adjust the AP distance between the COM and the leading foot, as increasing this distance is energetically costly. The work rate required to redirect the COM from one single support phase to the next increases with the fourth power of the step length, in both an inverted-pendulum-based collision model and human experimental data (Donelan et al., 2002). Humans might prefer a less costly recovery option, possibly provided by adjustments in ankle torque of the leading foot. Modifying the available recovery options, for example, through applying a constraint to the ankle joint of the subject, could give insight into why humans make this choice. Not adjusting the AP distance between the COM and the leading foot contradicts with COM velocity-dependent foot placement strategies that were previously used in simple inverted pendulum models (Kajita et al., 1992; Hof, 2008; Peuker et al., 2012), although these footless models have no option other than foot placement to displace the COP.

Double support phase duration

Changes in double support duration might be caused by actions both preceding and during the double support phase. First, when falling forward during the single support phase, the trailing leg extends to provide time and clearance for positioning of the leading foot (Pijnappels et al., 2005). The more extension occurs before the double support phase, the earlier the trailing leg will have to leave the floor during the double support phase, simply because it cannot extend any further. Second, the double support phase might be actively shortened or lengthened. The trailing leg cannot contribute well to horizontal forces required to slow down COM motion away from the trailing foot. A safer option could therefore be to initiate swing earlier, creating more time to prepare for the next step, which can reduce excessive velocity. Conversely, the double support phase might be lengthened when the trailing leg has to deliver additional force to regain velocity. Significant changes in the double support duration were not reported in (Hof et al., 2010) following ML perturbations during walking at 1.25 m s^{-1} . In the present study, fewer changes in double support duration were observed for the normal walking speed compared with the slow walking speed following ML perturbations, although significant changes were still present.

COP location

Using simple linear relations, the COM velocity at HSR can be used to predict the distance between the COM and the COP observed at TOL, in both ML and AP directions, for both slow and normal walking speeds. For the slow walking speed, these relations are similar in both ML and AP directions. Although humans cannot directly sense COM velocity, underlying proprioceptive and vestibular sensing systems could be used to make an estimate. The strong linear relations support that such an estimate could be used to generate a proportional recovery response. However, a causal relation between the COM velocity and the observed responses cannot be inferred from the data. Further perturbation studies, possibly combined with sensory perturbations (Kiemel et al., 2011), could be used to infer a

neurological cause of these responses.

The double support phase plays an important role in establishing these relations. They result from foot placement, COP displacements by a weight shift to the leading leg, changes in double support duration, as well as specific joint torques. Following AP perturbations, the larger range in distance between the COM and the COP at TOL compared with that between the COM and the leading foot at TOL can only be caused by effects other than foot placement, most likely an ankle torque. Hence, both passive dynamics and controlled actions prior and during the double support phase contribute to the observed linear relations.

Most effects that play a role in establishing these relations are not captured by simple inverted pendulum models. Yet, the relation between the COM velocity at HSR and the COP at TOL for slow walking is in line with constant offset control (Hof, 2008). If the fit to the data has the same slope as that of the XCOM line (ω_0^{-1}), the distance between the COP and the XCOM is approximately equal for all perturbations. This distance is then given by the intercept of the fit. Similarities are further supported by applying the offsets found in the data to foot placement in the linear inverted pendulum model. This would result in model movement that is in the same direction as that of the subjects. The model would topple over the COP in the AP direction for all AP perturbations. In the ML direction, the model would return in the direction it came from following most ML perturbations. Exceptions are the larger (-0.12 , -0.16) inward perturbations, for which the COP is located between the COM and the XCOM. This would make the model topple over the COP in the ML direction. Subjects likely also do this after making a cross-step, to undo the crossing of the legs in the subsequent step.

Although the model can mimic the observed relations, it does not explain the relations. The data violate several assumptions made in the model. Constant offset control only makes the linear inverted pendulum model converge to some stable gait as long as the swing time can be kept constant (Hof, 2008). Subjects showed adjustments to single and double support durations; hence in this scenario the linear inverted pendulum model provides no explanation for COP adjustments directly proportional to the COM velocity.

The relations might differ for other perturbation magnitudes and types. For sufficiently large AP perturbations magnitudes, the AP COP required to satisfy the relation is no longer obtainable without changing the foot location relative to the COM at heel strike. This could lead either to an increased distance between COM and leading foot to further expand the BoS, or to a recovery over multiple steps without adjusting this distance. Furthermore, it is unclear whether these relations hold for other perturbation types such as tripping, which has a major effect on the body's angular momentum (Pijnappels et al., 2005).

Ground reaction force

The horizontal ground reaction force components, the distance between the COM and the COP, RF and RD all co-vary at TOL in both the ML and AP directions. In a similar way, co-variation between AP COP location and ground reaction force direction has previously been shown to occur throughout the unperturbed gait cycle (Maus et al., 2010; Gruben and Boehm, 2012), in which the ground reaction force appears to be directed toward a point above the COM. Although not representative of the complete gait cycle, here the ground reaction force mostly points toward the COM at TOL. The main exception is in the AP direction for the normal walking speed, where it tends to point above the COM

for backward perturbations, and below the COM for forward perturbations.

Such co-variation could have advantages for balance control during walking. A change in horizontal force, for example, to modulate horizontal COM velocity, would alter the direction of the ground reaction force, and with it the angular acceleration of the body. By co-variation of either the vertical force magnitude or the COP location relative to the body, effects of a changing horizontal force on the body's angular acceleration can be prevented. Changing the vertical force would lead to fluctuations in vertical COM acceleration. Moreover, uncontrolled manifold analysis in unperturbed human walking suggests that creating a consistent vertical force is an implicit goal of walking, whereas creating a consistent AP force component is not (Toney and Chang, 2013). Simultaneous changes in horizontal force and COP location can be achieved through ankle torque modulation. It could therefore play an important role in simultaneously regulating horizontal and angular body accelerations. Previous work has suggested that an ankle torque is involved in regulating the body's angular acceleration in the sagittal plane during gait, reflected in changes in AP COP location and ground reaction force direction (Gruben and Boehm, 2014). Hence, ankle torque modulation could provide an alternative to increasing the AP distance between the COM and the leading foot during recovery.

Conclusions

The present work revealed simple linear relations between the COM velocity at heel strike and the COP location and horizontal ground reaction forces at toe-off during perturbation recovery. These relations are the result of passive dynamics as well as controlled actions during the single and double support phases. For slow walking, the relation between COM velocity and COP location is comparable between both ML and AP directions, possibly indicating a similar underlying objective. However, actions taken to realize these relations differ between the ML and AP directions. Although foot placement adjustment is crucial in the ML direction, other actions such as ankle torque modulation contribute to the relations in the AP direction. Furthermore, changes in gait phase duration suggest that the timing of actions could play an important role in the recovery phase. A further challenge is to unravel why humans choose one recovery strategy over another, and to what extent variables such as foot placement location, COP shift, and gait phase duration are actively regulated.

Many aspects that contribute to the observed relations are often not represented in simple inverted pendulum models. Although these simple models might mimic the relations through foot placement only, they do not necessarily provide an explanation of the observed human behavior. Using models to gain insight into why humans prefer a certain strategy requires modeling the involved degrees of freedom. Our study motivates the modeling of a double support phase, for instance, using a spring-loaded inverted pendulum (Geyer et al., 2006) that can mimic the double support through compliant legs. Our study furthermore suggests the inclusion of feet in walking models, such as in (Kim and Collins, 2013), where ankle control was used to stabilize a walking model. Such extended models are required to investigate the underlying costs and constraints that determine the balance strategies employed by humans during walking. Further mining of human data for simplified expressions of balance in walking can support making such models more human-like.

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Supplementary material

The figures below correspond to Figures 5, 6, and 7 but for the normal walking speed.

Figure S1. Gait phase duration following perturbations. Gait phase duration directly following ML (A) and AP (B) perturbations. Single and double support phase durations are indicated by the open and filled markers respectively. Triangles show subject averages and indicate the perturbation direction. Error bars indicate the subject standard deviation. Colors indicate the various perturbation magnitudes. Asterisks (*) indicate significant differences from the corresponding unperturbed phase duration. Double asterisks (**) indicate that there was no significant interaction effect between slow and normal walking, such that the corresponding p -values represent both slow and normal walking speeds. Data shown is dimensionless and for the normal walking speed only.

Figure S2. Various relations between COM velocity and both leading foot and COP. (A) ML COM velocity at HSR vs ML distance between COM and leading foot at HSR, for ML perturbations. (B) Same as (A) in the AP direction for AP perturbations. (C) ML COM velocity at HSR vs ML distance between COM and COP at TOL for ML perturbations. (D) Same as (C) in the AP direction for AP perturbations. Triangles show subject averages and indicates the perturbation direction. Ellipses represent the subject standard deviation. Colors indicate the various perturbation magnitudes. Pink solid and dashed lines indicates the subject average and standard deviation of the XCOM relative to the COM. This XCOM line has slope 00-1 and no intercept. Black dashed line is a linear least square fit to the data. R2 indicates the coefficient of determination of the fit. Data shown is dimensionless and for the normal walking speed only.

Figure S3. Relations between COM velocity and both force and distance ratios. (A) ML COM velocity at HSR vs ML force ratio R_F at TOL. For distance ratio R_D only a fit to the data is shown for comparison. (B) Same as (A) in the AP direction for AP perturbations. Triangles show subject averages and indicates the perturbation direction. Ellipses represent the subject standard deviation. Colors indicate the various perturbation magnitudes. Black and gray dashed lines are linear least square fits to the data, for R_F and R_D respectively. The R^2 indicates the coefficient of determination of the fits. Data shown is dimensionless and for the normal walking speed only.

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CHAPTER 4

Foot placement modulation diminishes for perturbations near foot contact

"Time passes, people move..."

In preparation for submission as:

Foot placement modulation diminishes in the first recovery step for perturbations near foot contact

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Abstract

Whenever a perturbation occurs during walking we have to maintain our balance using the recovery strategies that are available to us. Foot placement adjustment is likely a poorly available strategy if the foot is close to foot contact at the instance a perturbation occurs, making adjustments more difficult. The main goal of this study is to gain a better understanding of how humans deal with balance perturbations during walking if foot placement adjustments are constrained by time.

Ten healthy subjects walked on an instrumented treadmill and received mediolateral and anteroposterior pelvis perturbations at various instances during the single support phase. The results show that foot placement modulation in the first recovery step following mediolateral perturbations diminishes when the perturbation onset is closer to the instant of foot contact. For anteroposterior perturbations, foot placement is fairly invariant of the perturbation magnitude and direction, regardless of the onset instance. For most perturbations the first recovery step was consistent across subjects for all onset instances. However, in the second step various strategies arose that were not consistent across subjects, nor within subjects, especially for perturbations applied close to foot contact. Despite these different strategies, previously reported relations between the COM velocity and the COP location following foot contact remained intact. The results show that humans have various ways to compensate for poor availability of a foot placement strategy. This makes balance responses highly dependent on the instant during the gait phase at which the perturbation is applied.

Introduction

Human balance control is highly flexible, with a multitude of strategies that can be addressed to reject disturbances and allow continuation of walking. One example is the modulation of ankle joint moments to affect the movement of the body. Another are inertia-based strategies such as the hip strategy, in which changes in angular momentum are used to affect linear body motion. Furthermore, foot placement modulation can change the base of support area, allowing adjustments to be made to the walking cycle. This might be achieved by adjusting both the location and timing of foot placement. To gain more insight in human balance control and the preferred ways of balance recovery, it is helpful to understand how humans maintain balance when one or multiple strategies are only limited available.

The way balance is controlled depends on physical capabilities and constraints. For example, experiments in standing balance show that humans no longer utilize an ankle strategy if the size of the support surface is decreased (Horak and Nashner, 1986). This not only makes the ankle strategy ineffective, but also potentially threatening to balance. In walking, we have previously shown that foot placement adjustments are elicited in response to anteroposterior (AP) perturbations after physically blocking the ankle joints to make an ankle strategy ineffective (see Chapter 5). Such adjustments in foot placement were not observed following AP perturbations in normal walking (Vlutters et al., 2016). Other changes to the physical capabilities of the body also modify balance control, such as increased body sway in unilateral amputees (Geurts et al., 1992), or enhanced lateral balance performance through the use of a powered ankle device (Kim and Collins, 2015).

Physical constraints on balance control can also be in the form of time. Especially for foot placement modulation, time is required to make adjustments to the swing leg (Hof et al., 2010). The instance of the gait cycle at which a disturbance occurs is therefore an important factor in determining how balance will be maintained. If a disturbance occurs short before foot contact, there is little time to make foot placement adjustments. As a result, adjustments might have to be postponed to the subsequent step, or other balance strategies have to be addressed to compensate. Especially mediolateral (ML) disturbances given briefly before foot contact are expected to be challenging, given the already limited availability of other strategies, such as ML ankle control.

In an attempt to make predictions of balance control strategies during gait, the center of mass (COM) velocity has previously been shown to relate to the center of pressure (COP) location following foot contact, in the first recovery step following both ML (Hof et al., 2007; Hof et al., 2010; Vlutters et al., 2016) and AP (Vlutters et al., 2016) perturbations. Specifically, the COP shift after foot contact was in line with the velocity-based extrapolated center of mass (XCOM) concept derived from a linear inverted pendulum model (Hof et al., 2005). For AP perturbations, this COP shift was realized during the double support phase without the need to strongly adjust the location of the leading foot as compared to the unperturbed condition. For ML perturbations however, this COP shift was made possible mainly through foot placement adjustments in response to the perturbations. If there would be insufficient time for foot placement adjustments to occur, the COM velocity could lose its predictive value for that step. It is unclear if such predictions would hold for the subsequent step, especially if subjects cannot counteract the disturbances during the double support phase after the first step.

The main goal of this study is to gain a better understanding of how humans deal with balance perturbations during walking if foot placement adjustments are constrained by time. Specifically, we investigate balance responses of walking subjects following both ML and AP perturbations applied at

various instances within the single support phase. There will be less time to use foot placement adjustments as a recovery strategy if perturbation onset is close to an instance of foot contact. Foot placement adjustments are therefore expected to diminish for such perturbations. However, modulation of the COP with the COM velocity might still occur, but perhaps through strategies other than foot placement modulation.

Methods

Participants

Ten healthy young adults without known history of neurological, muscular, or orthopedic problems participated in the study (3 male, age: 21 ± 2 yr, height: 1.76 ± 0.1 m, weight: 65 ± 9 kg). The local ethics committee approved the experimental setup and protocol. All participants gave written informed consent prior to the experiment, in accordance with the Declaration of Helsinki.

Apparatus

Here only a brief description of the experimental setup is given. A more detailed description is provided elsewhere (Vlutters et al., 2016). A dual-belt instrumented treadmill (custom Y-mill, Motekforce Link, Culemborg, The Netherlands) and two motors (SMH60, Moog, Nieuw-Vennep, The Netherlands) adjacent to the treadmill were used to deliver ML and AP perturbations during walking in a controlled way. Each motor had a vertical lever arm linked to a horizontal aluminum rod through a ball joint, which in turn was connected to a universal hip abduction brace (Distrac Wellcare, Hoegaarden, Belgium) worn by the subject, also using a ball joint. Control signals for the motors were generated using xPC-target (MathWorks, Natick, MA, USA) at 1000 Hz.

Data collection

Subject kinematic data of the feet, lower legs, upper legs, pelvis, upper body, and head were captured using a 9-camera motion capture system (Visualeyez II, Phoenix Technologies, Burnaby, Canada). To this purpose a three-LED cluster was attached to each of those body segments. Additional single LEDs were placed on both lateral malleoli, and both lateral epicondyle of the femur. Motor torque and angle were collected over UDP using an Ethernet card (82558 Ethernet card, Intel, Santa Clara, CA, USA). Ground reaction force data of the treadmill were also collected at 1000 Hz using an AD card (PCI-6229, National Instruments, Austin, TX, USA). Both cards were part of the xPC-target hardware. The AD card was also used to generate an analog signal to synchronize the motion captures system with the xPC-target hardware.

Protocol

Before the start of the experiment, several kinematic measurements were performed during which the locations of the bilateral first and fifth metatarsal heads, calcaneus, medial and lateral malleoli, fibula heads, medial and lateral epicondyles of the femur, greater trochanter, anterior and posterior superior iliac spines, xiphoid process, jugular notch, 7th cervical vertebra, occiput, head vertex, and nasal sellion were indicated using an LED-based probe (Cappozzo et al., 1995).

During the experiment, subjects walked on the treadmill with their arms crossed over the abdomen. The walking speed was 0.63 ms^{-1} multiplied with the square root of the subject's leg length (Hof, 1996).

Subjects walked two blocks of five trials each. The first trial of each block was a two-minute baseline trial in which no perturbations were applied. The first baseline trial was furthermore used to determine the single support duration during unperturbed walking. The remaining four trials were perturbation trials. Perturbations consisted of sudden block pulses with a duration of 150 ms. Perturbation onset occurred at toe-off right at the start of the left single support phase (SS_0), at one one third of the single support phase ($SS_{1/3}$), and at two thirds of the single support phase ($SS_{2/3}$). The interval between subsequent perturbations was random between 6 and 12 seconds. The delivered force magnitudes were equal to 8 and 16 percent of the subject's body weight. Perturbations were directed either inward (negative sign, leftward for left single support) and outward (positive sign, rightward for left single support), or forward (positive sign) and backward (negative sign). In one block only ML perturbations were applied, in the other block only AP perturbations. Block order was randomized across subjects. Within a block, all perturbations were randomized over onset instance, magnitude, and direction. Each condition was repeated 8 times, yielding 196 perturbations in total. When no perturbation force was being delivered, the interaction force between subject and motor was regulated to (near) zero using admittance control. A safety harness was worn at all times to prevent the body from hitting the treadmill in case of a fall.

Data processing

Data were processed using Matlab (R2016b, MathWorks, Natick, MA, USA). Marker data were filtered with a 4th order zero-phase 20 Hz low-pass Butterworth filter. Landmark positions were subsequently reconstructed using the probe measurements through least squares estimation (Söderkvist and Wedin, 1993). Using a method comparable to that in (Zeni Jr et al., 2008), the calcaneus and first metatarsal head landmarks on both feet were used to detect gait phase events of toe-off right (TOR), heel strike right (HSR), toe-off left (TOL) and heel strike left (HSL). All landmark data was used to estimate the location of the COM of each segment, as well as that of the whole body COM (Dumas et al., 2007). The COM position was differentiated to find COM velocities.

The unperturbed walking data from the baseline trials was used to find the average Euclidean distance between the COM of the feet at heel strike. This value was used as a scaling factor (l_0) to make all position and velocity data dimensionless following (Hof, 1996). Next, all position and velocity data were expressed relative to those of the whole body COM. The velocity of the whole-body COM itself was expressed relative to the treadmill belt. All data were sorted on perturbation magnitude, direction, and onset. The COM velocity data was cut into sequences using the gait phase events. Each sequence was resampled to 50 samples to allow averaging. All data at gait events were averaged over repetitions to obtain average data per subject. These were used to obtain subject averages and standard deviations. In line with previous analysis (Vlutters et al., 2016), linear least squares fits were made to the distance between the COP and the COM at TOL as a function of the COM velocity at the preceding HSR. Such fits have previously shown to correspond well to the XCOM concept, with the COP-COM distance increasing with a slope $\omega_0^{-1} = \sqrt{l/g}$, where l is the subject's leg length and g is the Earth's gravitational acceleration. A dimensionless XCOM proportionality constant (ω_0^{-1}) was calculated for each subject, and averaged over all subjects for comparison with the fits.

Linear mixed models were used to assess the effect of the perturbation (fixed factor, with intercept) and the onset timing (fixed factor, with intercept) on the ML and AP distance between the COM and the

COP at TOL, as well as on the duration of the single and double support phases during and after the perturbation. Subject effects were included as a random factor (intercept) to account for correlation effects from repeated measures within the same subject. A significance level of $\alpha=0.05$ was used and a Bonferroni correction was applied to correct for multiple comparisons during *post hoc* analysis. The perturbed conditions were only compared to the unperturbed condition and not mutually to reduce the number of comparisons. Finally, the analysis was performed separately for ML and AP perturbations. SPSS statistics 21 (IBM Corporation, Armonk, NY, USA) was used for the statistical analysis.

Results

Foot placement adjustments in terms of location and time were assessed following both ML and AP perturbations in walking subjects. All data is shown dimensionless. Subject-average scaling factors to make the data dimensionless were $l_0 = 0.41 \pm 0.03$ m for distances, $\sqrt{(g * l_0)} = 2.02 \pm 0.07$ m s⁻¹ for velocities, and $\sqrt{(l_0 / g)} = 0.21 \pm 0.01$ s for durations, where l_0 is the average Euclidean distance between the COM of the feet at unperturbed heel strike.

For various ML perturbations one subject showed stepping strategies that were not consistent with the other subjects. These special cases are shown separately in Figures 3, 4, and 5, and were removed from the statistical analysis. For the 0.16 magnitude outward perturbation applied at SS_{2/3} the responses were not consistent across subjects, nor within several subjects. As a result, the data cannot be pooled subject-wise to represent a specific strategy. Corresponding data were omitted from Figures 3 and 5 to prevent image cluttering, and were also removed from the statistical analysis. However, all data were included when determining the relations of the COP data with the COM velocity. As we have previously demonstrated in (Vlutters et al., 2016), the underlying COP might still modulate with the COM velocity throughout different balance strategies.

Perturbation effects on COM velocity

Both the ML and AP perturbations affected the subject's COM velocity, see Figures 1 and 2. The velocity profiles following ML perturbations appear dependent on the onset timing. Deviations from the unperturbed case obviously start later for later perturbation onset, but the way the velocity progresses changes with the onset timing as well. The effects of various AP perturbations on the COM velocity appear less dependent on the onset timing. Though later onset leads to later deviations from the unperturbed case, the velocity profiles between the different onset conditions start to appear similar again at HSL, at the start of the second step.

The disturbances move the subjects in the direction of the perturbation, such that subjects had to return to the center of the treadmill during their recovery. This can be derived from the velocity data, corresponding to the instances where the perturbed velocity data crossed the unperturbed velocity data. The point at which this return occurs following ML perturbations shifts with the perturbation onset, becoming later for perturbations that are applied later. This is less the case for AP perturbations.

Finally, the COM velocity perpendicular to the perturbation direction remains relatively unaffected by the perturbation itself, but may change through subject actions following HSR. This is mainly the case for ML perturbations, especially for those with a SS_{2/3} onset. Subjects speed up in the walking direction for inward perturbations, and slow down for outward perturbations.

Figure 1. COM velocity profiles following ML perturbation. (A) ML COM velocity in response to ML perturbations. (B) AP COM velocity in response to ML perturbations. Top, middle, and bottom rows correspond with perturbation onset instances of SS_0 , $SS_{1/3}$, and $SS_{2/3}$, respectively. Data is shown as a function of the gait phase. Colors indicate the various perturbation magnitudes. Shaded areas indicates the subject-standard deviation for the unperturbed condition. It is not shown for perturbed conditions to prevent image cluttering. Data is shown dimensionless.

Foot placement location following ML perturbations

Subjects modulated their foot placement in terms of location and/or timing following the perturbations. Especially ML perturbations with a SS_0 onset lead to adjustments in foot placement location in the first recovery step, see Figure 3. This might be expected given that these perturbations are perpendicular to the walking direction, and that there is sufficient time to adjust the foot. In the first step, at HSR, subjects generally placed their foot in the direction of the perturbation at an increased ML distance from the COM with increasing ML perturbation magnitude. The ML distance between the COM and the leading foot was significantly affected by the ML perturbations ($F_{4,126}=114.410$, $p<0.001$), the onset timing ($F_{2,126}=7.605$, $p=0.001$), and their interaction ($F_{8,126}=126.000$, $p<0.001$). However, the *post hoc* analysis revealed that the leading foot was not placed significantly different from the unperturbed case

Figure 2. COM velocity profiles following AP perturbation. (A) ML COM velocity in response to AP perturbations. (B) AP COM velocity in response to AP perturbations. Top, middle, and bottom rows correspond with perturbation onset instances of SS_0 , $SS_{1/3}$, and $SS_{2/3}$, respectively. Data is shown as a function of the gait phase. Colors indicate the various perturbation magnitudes. Shaded areas indicates the subject-standard deviation for the unperturbed condition. It is not shown for perturbed conditions to prevent image cluttering. Data is shown dimensionless.

for any ML perturbation with a $TOR_{2/3}$ onset ($p=1.000$). Other exceptions were the -0.08 inward perturbations with a TOR_0 and $TOR_{1/3}$ onset ($p \geq 0.589$). For the second step, at HSL, modulation of the foot location occurred, but the changes in ML distance between the leading foot and the COM tend to diminish for perturbations with a later onset. A possible explanation is that part of the recovery occurred during the double support phase, and the single support phase prior to foot contact of the second step. This way there is less need for adjustments in the location of the foot. Note that that the 0.16 magnitude outward perturbations with a $SS_{2/3}$ onset are disregarded here, for which lateral foot adjustments did occur. For perturbations with a SS_0 and $SS_{1/3}$ onset the ML distance to the leading foot changes in the second step for various reasons, such as uncrossing the legs following a cross-step, and to return to the center of the treadmill. The ML distance between the COM and the leading foot in the

second step was significantly affected by the perturbations ($F_{4,114,101}=23.251$, $p<0.001$), the onset timing ($F_{2,114,068}=9.262$, $p<0.001$), and their interaction ($F_{7,114,058}=2.825$, $p=0.009$). Visual inspection of Figure 3 suggests that the modulation pattern as seen in the first step in response to ML perturbations with a SS_0 onset do not clearly re-appear in the second step for ML perturbations with a $SS_{2/3}$ onset.

Figure 3. Foot placement locations following ML perturbations. (A) Top-down view of the locations of the COM of the leading and trailing foot relative to the whole body COM in (0,0), at the first step (HSR) and second step (HSL) after the perturbation. Perturbation onset was at SS_0 . For the second step, one subject deviated from the other subjects. The repetition-average data of this subject is given by a cross. (B) Same as A, but for onset at $SS_{1/3}$. (C) Same as A, but for onset at $SS_{2/3}$. The subject averages for the 0.16 magnitude $SS_{2/3}$ perturbation are not shown as it is not representative of any specific stepping strategy, but contains a mixture of three different strategies. Triangles show subject-averages and indicate the perturbation direction. Ellipses represent the subject-standard deviation. Colors indicate the different perturbation magnitudes. Data is shown dimensionless.

Figure 4. Alternative stepping strategies. Alternative strategies used by one or multiple subjects for specific perturbations. The number (1,2) corresponds to the step number in Figure 3. (A) Cross-uncross, making a double step with the right leg. (B) Near-cross step with the right leg, followed by a short left step to prevent the body from falling leftward. (C) Left cross-step. (D) Rear cross-step. (E) Foot pivoting, by first rotating about the toes, then shifting the COP back toward the heel and rotating about the heel. Accompanied by left leg abduction during the swing to prevent toppling over the right stance leg. Several subjects showed both (D) and (E) throughout the repetitions of the 0.16 perturbations at $SS_{2/3}$.

Alternative foot locations following ML perturbations

The aforementioned statistical results do not include the alternative strategies performed by some subjects in the second step, deviating from the rest of the population, see Figure 4. For perturbations with a SS_0 and $SS_{1/3}$ onset, only one subject showed alternative stepping responses, for one specific perturbation. Subject 4 consistently performed a double right step following the -0.16 inward perturbations with a SS_0 onset, first crossing the legs in step 1 like all other subjects, but then uncrossing the legs with a second right step (Figure 4A). Furthermore, subject 4 made a consistent short compensatory step with the left leg during the second step after the -0.16 inward perturbations with a $SS_{1/3}$ onset, to further counteract the induced inward COM velocity (Figure 4B). This short step was also of shorter duration than that of the other subjects (Figure 5).

For perturbations with a $SS_{2/3}$ onset more varying responses occurred, mainly for the 0.16 outward perturbations. Subjects 1, 3, 4, and 10 performed left leg abduction, opposite of the perturbation direction, as well as foot pivoting (Figure 4E). Moving the heel laterally by pivoting on the forefoot, and subsequently moving the forefoot laterally by pivoting on the heel allows changes in the base of support using only a single foot, without actual stepping. Subjects 7 and 9 performed a rear cross-step (Figure 4D), stepping behind the leading leg without the body fully toppling over the leading foot in the sagittal plane. Subjects 2, 5, and 8 performed both of these strategies, and subject 6 even performed three different strategies, including a cross-step using the left leg (Figure 4C). Furthermore, subject 6 was the only subject to perform such a cross-step for the 0.08 outward perturbations with a $SS_{2/3}$ onset. Other subjects dealt with this perturbation through a relatively long lasting right single support phase during which the left leg was abducted, sometimes combined with foot pivoting as in Figure 4E. The leg abduction is not directly clear from the foot placement locations, but is in line with the long lasting single support duration of the second step following 0.08 outward perturbations with a $SS_{2/3}$ onset in Figure 5.

Foot placement timing following ML perturbations

Aside from adjustments in foot placement location, the gait phase durations following the perturbations were affected as well, see Figure 5. For the first step, inward perturbations tend to increase the single support duration, whereas outward perturbations tend to decrease it. These effects diminish with later perturbation onset. In contrast, for the second step inward perturbations tend to decrease the single support duration, whereas outward perturbations tend to increase it. These effects become stronger with later perturbation onset. Specifically, major deviations occur for the -0.16 inward and the 0.08 outward perturbations with a $SS_{2/3}$ onset. For the -0.16 inward perturbation a fast step with the left leg is used to correct in the second step. For the 0.08 outward perturbation the duration increases because of the earlier mentioned leg abduction strategy that occurs during this single support phase.

All gait phase durations were affected by the ML perturbations ($F_{4,144.053} \geq 5.248$, $p \leq 0.001$), but there was no main effect of the onset timing on any of the gait phases ($F_{2,114.032} \leq 1.585$, $p \geq 0.209$). This is likely because the effects of inward and outward perturbations tend to cancel out in the average duration over all perturbations. A significant interaction effect was found only for the single support phases of the first and second step ($F_{7,114.025} \geq 9.069$, $p \leq 0.001$), but not for any of the double support phases ($F_{7,114.127} \leq 1.269$, $p \geq 0.166$). For the single support phase of the first step after the perturbation, deviations in duration from the unperturbed case diminish with increasing perturbation onset delay. For

none of the ML perturbations with a $SS_{2/3}$ onset was the duration of this single support phase significantly different from the unperturbed case ($p \geq 0.421$). This is consistent with the findings for the foot locations following $SS_{2/3}$ onset perturbations. It is likely that subjects cannot make major adjustments to their foot placement if the remaining time to the intended (unperturbed) foot contact is short.

Foot placement location following AP perturbations

Changes in the AP distance between the COM and the leading foot during both the first and second step after the AP perturbations are generally small, see Figure 6. For forward perturbations there is a tendency for the first step to be longer and the second to be shorter, while the opposite is the case for backward perturbations. The leading foot AP distance from the COM was significantly affected by the AP perturbations ($F_{4,126} \geq 9.252$, $p < 0.001$) and the onset timing ($F_{2,126} \geq 7.432$, $p = 0.001$), but not by their interaction ($F_{8,126} \leq 1.600$, $p \geq 0.131$), for both at HSR (step 1) and HSL (step 2). Though significant (unlike as in Chapter 3), the mean difference is typically 1 cm between the various onset conditions, as well as between the various AP perturbations and the unperturbed condition. The foot locations in step 2 appear similar regardless of the onset timing. This suggests that the AP perturbations are mostly rejected during the double support phase following the disturbance. No aberrant strategies were observed in response to AP perturbations.

Foot placement timing following AP perturbations

For AP perturbations, the largest changes in gait phase duration appear to occur in the double support phases rather than in the single support phases. For the first double support phase, backward perturbation leads to increases in duration, and forward perturbations to decreases. The second double support phase has the tendency to show opposite effects. This is possibly related to the distance to the trailing foot at heel strike. If this distance is larger, the trailing foot will have to leave the floor earlier during the subsequent double support phase, making it of shorter duration. All gait phase durations were affected by the perturbations ($F_{4,126} \geq 5.168$, $p \leq 0.001$). The perturbation onset timing only affected the single support durations ($F_{2,126} \geq 4.827$, $p \leq 0.010$) and not the double support durations ($F_{2,126} \leq 0.310$, $p \geq 0.712$). Most gait phase durations were also affected by the interaction effect ($F_{8,126} \geq 2.112$, $p \leq 0.039$), with exception of the double support phase of step 2 (HSL-TOR). This indicates that the perturbation responses in this double support phase are independent of the onset timing of the perturbation. The effects of the perturbation onset must therefore have been negated in an earlier gait phase. Finally, for the first step, backward perturbations with a $SS_{1/3}$ onset appear to yield swing durations that are distinct from those with a SS_0 and $SS_{2/3}$ onset. However the latter two last significantly longer than the unperturbed case ($p \leq 0.019$), whereas the single support durations in the first step following backward perturbations with the $SS_{1/3}$ onset are not significantly shorter ($p \geq 0.055$). Hence, unlike for ML perturbations, the swing time does not strictly show more similarities to the unperturbed case with increasing perturbation delay.

Figure 5. Gait phase durations. (A) Gait phase durations following ML perturbations. (B) Gait phase durations following AP perturbations. The single subjects with an alternative strategy for step 2 are shown separately with a cross. Triangles show subject-averages and indicate the perturbation direction. Larger marker size corresponds with later perturbation onset. Error bars represent the subject-standard deviation. Open and filled markers correspond to the single and double support phases, respectively. Colors indicate the different perturbation magnitudes. Data is shown dimensionless.

Figure 6. Foot placement locations following AP perturbations. (A) Top-down view of the locations of the COM of the leading and trailing foot relative to the whole body COM in $(0,0)$, at the first step (HSR) and second step (HSL) after the perturbation. Perturbation onset was at SS_0 . (B) As in (A), but for onset at $SS_{1/3}$. (C) As in (A), but for onset at $SS_{2/3}$. Triangles show subject-averages and indicate the perturbation direction. Ellipses represent the subject-standard deviation. Colors indicate the different perturbation magnitudes. Data is shown dimensionless.

Relations with the COM velocity

In line with previous studies (Hof et al., 2010; Vlutters et al., 2016) we investigated the predictive power of the COM velocity for the location of the COP after foot contact. Note that the relations presented here span two instances of the gait cycle: heel strike for the velocity, and the subsequent toe-off for the COP. We have previously found this to provide the best correspondence with the XCOM concept, which is also dependent on the COM velocity (Vlutters et al., 2016). The slopes, intercepts, and coefficients of determination of the fits to the data are presented in Table 1. If the data modulates with the same slope as that of the pink XCOM line (ω_0^{-1}), then the COP shifts to a constant offset from the XCOM on average over all perturbation magnitudes. The subject-average dimensionless XCOM proportionality constant ω_0^{-1} was 1.49 ± 0.05 , for comparison with the slopes in Table 1.

For ML perturbations, both COP and XCOM modulate in a similar way with the COM velocity within the first recovery step if the perturbation is given early, at SS_0 , see Figure 7 and Table 1. Perturbations with later onset diminish the similarities, in line with the reduced foot location modulation (Figure 3). For the second step similarities are highest if the perturbation is given late, at $SS_{2/3}$. This is possibly

because the return to the center of the treadmill begins to play a role for the second step if the perturbation was given early. Furthermore, a major contributor to the fit to the data in the second step is the mixture of strategies for the 0.16 outward perturbations at $SS_{2/3}$, see Figure 4B. Even though the standard deviation is large due to the various strategies, its variation is aligned with the XCOM line. Note that for the second step the data is located on the other side of the XCOM line, as the step is made with the other leg. For AP perturbations, both COP and XCOM modulate in a comparable way with the COM velocity within the first step for perturbations with a SS_0 or $SS_{2/3}$ onset. The likeness is less for perturbations with a $SS_{1/3}$ onset. For the second step, too, the modulation in COP and XCOM is less similar compared to step 1, irrespective of the perturbation onset.

Table 1. Slope, intercept and coefficient of determination of the linear least squares (LLSQ) fit made to the subject-average data at specific instances of the gait cycle after the perturbation. Distance COP-COM is the independent variable, COM velocity the dependent variable. Underlined values correspond with fits of which the root mean square error is less than 5% of the range of the dependent variable.

ML perturbations			
	ML COM velocity, at HSR (step 1)		
ML distance COP-COM at TOL (step 1)	Slope	Intercept	R ²
SS_0 onset	<u>1.425</u>	<u>0.055</u>	<u>0.994</u>
$SS_{1/3}$ onset	1.284	0.079	0.986
$SS_{2/3}$ onset	0.090	0.159	0.974
	ML COM velocity, at HSL (step 2)		
ML distance COP-COM at TOR (step 2)	Slope	Intercept	R ²
SS_0 onset	1.081	-0.146	0.962
$SS_{1/3}$ onset	0.853	-0.165	0.896
$SS_{2/3}$ onset	<u>1.444</u>	<u>-0.110</u>	<u>0.931</u>
AP perturbations			
	AP COM velocity, at HSR (step 1)		
AP distance COP-COM at TOL (step 1)	Slope	Intercept	R ²
SS_0 onset	1.348	-0.314	0.959
$SS_{1/3}$ onset	1.020	-0.212	0.966
$SS_{2/3}$ onset	1.268	-0.255	0.998
	AP COM velocity, at HSL (step 2)		
AP distance COP-COM at TOR (step 2)	Slope	Intercept	R ²
SS_0 onset	1.058	-0.204	0.979
$SS_{1/3}$ onset	1.000	-0.192	0.885
$SS_{2/3}$ onset	0.960	-0.162	0.632

Figure 7. Relation COM velocity and COP location in comparison to the XCOM. (A) The ML COP location relative to the COM at TOL (step 1) and TOR (step 2) as a function of the ML COM velocity at the preceding heel strike. Perturbation onset was at SS_0 . (B) Same as A, but for perturbation onset at $SS_{2/3}$. (C,D) Same as A and B, but for AP distances and velocities in response to AP perturbations. The pink line corresponds to the XCOM position relative to the COM, as a function of the COM velocity. It has slope ω_0 and no intercept. The dashed line is a linear least squares fit to the data. Triangles show subject-averages and indicate the perturbation direction. Ellipses represent the subject-standard deviation. Colors indicate the different perturbation magnitudes. Data is shown dimensionless.

Discussion

Walking subjects received both ML and AP pelvis perturbations at various onset instances throughout the single support phase to investigate the effect of a time constraint on foot placement adjustments. For ML perturbations, adjustments in foot placement location and time in the first recovery step clearly diminished with increasing perturbation onset delay. For AP perturbations the first step is relatively insensitive to the onset timing. Most adjustments in foot placement were consistent across subjects, with several exceptions in the second step. Mainly the largest magnitude (0.16) outward perturbations with an onset at two-thirds of the single support phase ($SS_{2/3}$ onset) resulted in inconsistent and varying responses across and within subjects during this second step.

Balance responses are gait phase dependent

Gait phase dependent responses arise because foot placement modulation takes time. Hof and colleagues (Hof et al., 2010) reported at least 0.28 s to be required for a “correct” lateral positioning of the foot to occur, though it is unclear how this was defined. Here, the foot placement location relative to the COM did not significantly alter if ML perturbations were given at $SS_{2/3}$, but did alter if the onset was at $SS_{1/3}$ or earlier. As a result, more than 0.15 s (0.73 dimensionless time units) are required for significant adjustments to be made, but less than 0.3 s. Adjustments might also occur mechanically without active involvement of the subject. For AP perturbations, the single support duration was significantly longer than the unperturbed condition for -0.16 backward perturbations at $SS_{2/3}$. For these perturbations the single support duration during the first step might increase because the body is pulled backward, which could postpone foot contact resulting from a forward fall due gravity.

The instance of the gait cycle during which the perturbation is applied affects how threatening a perturbation is to balance. This is clearly demonstrated by the 0.16 outward perturbations with a $SS_{2/3}$ onset. Balance control should therefore be evaluated with respect to the specific instances of the gait cycle at which a perturbation is applied. Several studies have suggested that slow walking is overall more stable than fast walking based on the analysis of kinematic variability in unperturbed gait (Dingwell and Marin, 2006; England and Granata, 2007). However, a specific perturbation such as the 0.16 outward perturbation might be additionally challenging if it is applied short before the weight transfers to the leading foot, near the end of the swing phase or start of the double support phase. As the loading of the leading foot takes more time in slow walking compared to fast walking (Hebenstreit et al., 2015), it is possible that slow walking is more prone to this specific perturbation. Conclusions on walking stability based on inherent variability are therefore unlikely to extrapolate to large magnitude disturbances, nor can a specific walking speed be declared strictly more stable than another.

Lack of foot placement adjustments in the first recovery step elicits other strategies

Though balance control is sometimes divided in ankle, hip, and foot placement strategies (Horak and Nashner, 1986; Maki and McIlroy, 1997), it is certainly not limited to these strategies. For example, when adjustments to the foot location are not possible, foot pivoting allows one-legged base of support corrections without making an actual step. It is therefore not always straightforward to group the responses in specific strategies. This is further illustrated by changes in foot placement location and/or time that are not a result of a foot placement strategy. Foot placement adjustments can be a way to alter body motion following foot contact (i.e. a stepping strategy), but it can also be a result of another

strategy that occurred before foot contact. The increased single support duration in the second step following the 0.08 magnitude outward perturbations with a $SS_{2/3}$ onset is an example of this. Subjects abducted their left swing leg possibly as an inertial strategy, or to provide a counter-weight and shift the whole-body COM in the direction of the perturbation. Because subjects spend time abducting their leg, the change in step time is not strictly because it was required for a specific foot placement adjustment. Hence, possible interaction with other strategies must not be disregarded.

It remains unclear why some subjects prefer one recovery strategy over another, and why alternative strategies mainly occur at higher magnitudes. In (Hof et al., 2010) no differences between or within subjects were reported for perturbations delivered at various instances of the gait cycle. This is possibly due to the lower perturbation magnitudes. In general, subjects might have less experience dealing with large magnitude disturbances compared to lower ones. Large magnitude perturbations occur less in daily life, such that differences are more likely to arise when the perturbation magnitude is high. The fact that different strategies occurred not only across subjects but also within subjects suggests it is not simply a matter of subject preference. Perhaps minor differences in onset conditions could provide an explanation, but this would require an extensive correlation analysis between the onset conditions and the resulting strategy. Still, biomechanical constraints could provide some insight on the underlying causes. Whether a cross-step occurs behind or in front of the other leg following outward perturbations at $SS_{2/3}$ likely relates to the AP velocity relative to the ML velocity. That is, if the ML disturbance is so large that it would take too long to traverse over the stance foot in the AP direction to make a corrective cross-step, a backward cross-step might be a preferred option instead. Future experiments might point out if this is indeed the case, for example by combining ML and AP perturbations.

Various responses contribute to the same relation

The relations between the COM velocity at heel strike and the COP distance from the COM at the subsequent toe-off, previously described in (Vlutters et al., 2016), remained intact throughout various conditions. If the perturbations are given early, at SS_0 , the results are comparable with those in our previous study. However, the relation tends to diminish in the second step, possibly because the return to the center of the treadmill begins to play a role. If the perturbation is given late, at $SS_{2/3}$, the relation did not occur in the first step for ML perturbations due to the lack of foot placement adjustments, but re-appeared in the second step, even throughout the varying strategies for the 0.16 outward perturbation. Only for the $SS_{1/3}$ perturbations the relation appears less in both the first and the second step. For these perturbations there might have been insufficient time to expand the base of support to realize the COP – COM velocity relation in the first step, and it might no longer be required for the second step, because recovery actions occur during single and double support phases before the foot contact as well (see Chapter 7).

Conclusions

To summarize, foot placement modulation takes time and therefore diminishes in the first recovery step if little adjustment time is available after a perturbation. Foot placement adjustments do occur for the second step, but the degree of modulation is dependent on the perturbation magnitude, direction, onset timing, and preceding actions. If foot placement modulation is not an option in the first step, actions during subsequent gait phases are addressed as an alternative. This can lead to peculiar balance

strategies such as foot pivoting. These strategies can be inconsistent both across and within subjects. Though it remains unclear what causes the use of the various strategies, subject preferences are unlikely given that differences also occur within subjects. Despite the varying strategies, previously observed relations between the COM velocity and the COP location relative to the COM persist. These might disappear in the first step and re-appear to the second step if the perturbation is given late in the preceding single support phase. These results suggest that foot placement, like any other balance strategy, is a way of achieving some underlying objective, possibly reflected in the COP location. Further probing human balance through perturbations might help reveal these objectives.

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CHAPTER 5

Reduced center of pressure modulation elicits foot placement adjustments

“Any objections, lady?”

Under review as:

Reduced center of pressure modulation elicits foot placement adjustments, but no additional trunk motion during anteroposterior-perturbed walking.

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Abstract

Understanding balance during human gait is complicated by the abundance of recovery options. Among all possible recovery options, three main strategies are often considered for human balance control, being the ankle, hip, and foot placement strategies. All can be addressed when balance is threatened during walking, but their relative importance remains uncertain. We have previously shown that healthy human subjects did not significantly adjust their foot placement relative to the body's center of mass (COM) in the first recovery step following anteroposterior pelvis perturbations, as compared to unperturbed walking. An ankle strategy could have contributed to the recovery instead. Here the goal is to further elucidate balance strategy preferences by investigating the stepping and hip strategies following these anteroposterior perturbations, but with an ankle strategy made ineffective. This was achieved by physically blocking each ankle and minimizing the support area of each foot through a pair of modified ankle-foot orthoses. These “pin-shoes” enabled stilt-like walking and ensured that foot placement adjustment was the only way to modulate the center of pressure location, comparable to “footless” inverted pendulum models of walking. Despite the pin-shoes, subjects did not additionally address a hip strategy compared to normal walking, but relied on foot placement adjustments instead. The observed foot placement adjustments were furthermore in line with concepts derived from a linear inverted pendulum model of walking. These results suggest low hip strategy priority when a foot placement strategy is available, while the latter can be predicted with concepts derived from a simple walking model.

Introduction

Balance during walking can be maintained in numerous ways, given the abundance of motor control options. For example, the body's center of mass (COM) motion can be influenced by foot placement adjustments ('stepping strategy') (MacKinnon and Winter, 1993), ankle moment modulation ('ankle strategy') (Pijnappels et al., 2005; Kim and Collins, 2015) and hip moment modulation ('hip strategy') (Horak, 1987). In the latter case, hip moments either oppose or aid gravity in angular accelerations of the trunk. The net trunk angular acceleration, together with the trunk's inertial properties, leads to shear-forces under the foot or feet on the floor which affect horizontal COM motion.

Foot placement adjustment is often considered the most important strategy during walking, especially in the frontal plane (MacKinnon and Winter, 1993). In the sagittal plane, however, we recently found that healthy subjects did not significantly adjust their foot placement relative to the COM in the first step following anteroposterior (AP) pelvis perturbations applied at toe-off (Vlutters et al., 2016). Instead, subjects modulated subsequent gait phase durations, and the center of pressure (COP) location within the new base of support (BoS) after foot placement. The COM velocity at heel strike after the perturbation was furthermore a strong predictor of the COP location at the subsequent toe-off. This relation between COM velocity and COP location was in line with the extrapolated center of mass (XCOM) concept that can be derived from a linear inverted pendulum model (Hof, 2008).

In addition to unexpected perturbations, another way to investigate compensatory actions during walking is by changing the walking dynamics (Thielemans et al., 2014; Selinger et al., 2015), and inferring to what purpose a compensation occurs. This way, underlying objectives of balance control during walking it might be extracted. Fully eliminating a balance recovery strategy might lead to more pronounced effects in other strategies. A clear example is provided in (Horak, 1987), where subjects utilize a hip strategy when the availability of both the ankle and stepping strategies become severely limited when standing on a narrow ridge.

In this study, we examine balance strategy preferences by investigating whether subjects adjust the stepping and hip strategies in the recovery from AP perturbations during walking, while the ankle strategy is made ineffective. By eliminating the ankle strategy, the stepping and hip strategies might need to be additionally addressed as a compensation. Conversely, if in such a challenging condition a strategy is not additionally addressed while it is an available recovery option, it is reasonable to assume that this strategy has low priority in normal walking as well. This can provide insight in strategy preferences. The ankle strategy is made ineffective by physically blocking each ankle joint and reducing the BoS of each foot through a pair of 'pin-shoes'. These ensure a gait that resembles stilt-walking. Second, we investigate whether under such conditions the stepping strategy will be in line with predictions based on the XCOM concept. Due to the pin-shoes the COP can no longer modulate within the BoS of each foot. This way a stepping strategy becomes the only way to realize COP modulation in response to variations in COM velocity, in line with (Hof, 2008). Such comparisons might further validate concepts derived from simple inverted pendulum models of walking, and aid in the development of controllers for balance assisting devices such as lower-extremity exoskeletons.

Methods

Apparatus

Set-up details can be found in (Vlutters et al., 2016). Briefly, an instrumented treadmill (MotekForce Link, Culemborg, Netherlands) and a motor (Moog, Nieuw-Vennep, Netherlands) at the rear of the treadmill were used to perturb subjects forward and backward during walking. The motor could be connected to the subject's pelvis through a rod and a brace (Distrac Wellcare, Hoegaarden, Belgium). Additionally, in this study a pair of ankle-foot orthoses (Airselect Elite, Aircast Inc., Summit NJ, USA) were used to restrain ankle movement, and reduce the area of the BoS through a modification, see Figure 1. An aluminum pin was attached perpendicular to the sole, with a solid, aluminum hemisphere located at the end of the pin to ensure a minimal BoS. The weight of one pin-shoe was approximately 1.5 kg.

Protocol

The experimental protocol was approved by the local ethics committee. All participants gave written informed consent prior to the experiment, in accordance with the Declaration of Helsinki. Ten subjects (5 men, age 20 ± 1 years, weight 71 ± 7 kg, height 1.76 ± 0.08 m, mean $\pm$ s.d.) walked on the treadmill while wearing the pin-shoes. For each subject, the pins were fixated at a position approximately halfway the longitudinal arches of each foot. We used this position based on a pilot study, in which subjects found it difficult to walk with the pin shifted too far towards the toes or toward the heel, making them either fall backward, or make the knee prone to buckling during the stance phase.

Subjects were given time to adapt to the pin-shoes until they could walk with their arms crossed over the abdomen. Walking speed was $0.63 * \sqrt{l} \text{ ms}^{-1}$, where $\sqrt{l}$ is the square root of the total leg length, including pin length, following (Hof, 1996). This speed corresponds to the 'slow' walking speed in (Vlutters et al., 2016), and is considered reasonable for walking on the pin-shoes. Subjects randomly received 150 ms forward (+) and backward (-) pelvis perturbations at the instance of toe-off right (TOR) using the motor behind the treadmill. Perturbation magnitude was 4, 8, 12, or 16% of the subject's body weight. Each perturbation condition was repeated 8 times, yielding 64 perturbations per subject.

Figure 1. Schematic drawing of one pin-shoe: an ankle-foot orthosis that blocks the ankle joint, modified to strongly reduce the support area of the foot. The vertical pin can be positioned along the foot.

Kinematic data of landmarks on the feet, lower legs, upper legs, pelvis, trunk, head, and the pin-shoes were collected at 100 Hz using a motion capture system (Visualize II, Phoenix Technologies Inc, Vancouver, Canada). The landmark on each shoe was the pin-tip. Ground reaction forces and moments were collected at 1000 Hz.

Data processing

For comparison, previously collected data in perturbed walking without pin-shoes at a similar speed (Vlutters et al., 2016) were also processed as described below. In each data set, with and without pin-shoes, the trunk data of one subject was removed from analysis due to motion artifacts of the trunk cluster. Landmark data were used to estimate the total-body COM position (Dumas et al., 2007). The COM position was differentiated, and the belt velocity was added for a COM velocity relative to the walking surface. Trunk kinematic data were used to find sagittal plane trunk angular accelerations and angular excursions. Trunk angular accelerations were used as an indication of the hip strategy, containing both passive (gravitational) and active (human) contributions. Trunk angular excursions indicate the changes in trunk angle relative to that at the instance of perturbations onset.

Instances of TOR, heel strike right (HSR), toe-off left (TOL) and heel strike left (HSL) were detected using vertical ground reaction forces (threshold 20 N). All data were cut, and re-sampled to five 50-sample sequences spanning one gait cycle: (1) TOR (perturbation onset) to TOR+150 ms (perturbation end), (2) TOR+150 ms to HSR, (3) HSR to TOL, (4) TOL to HSL, (5) HSL to TOR. Sequence duration was calculated before re-sampling. Sequences and their corresponding durations were sorted, and averaged over the 8 repetitions per perturbation type to obtain repetition-average sequences per subject. As the pin-shoes change the way of walking, even without perturbations, a direct comparison of the perturbation effects between walking with and without pin-shoes can be difficult. We therefore quantified the effects of the perturbations on the COM velocity, trunk angular acceleration, and trunk angular excursion data relative to the unperturbed case, for both the with and without pin-shoes conditions. For each subject, the repetition-average unperturbed sequences were subtracted from all repetition-average perturbed sequences. This makes the data relative to the baseline in both the with and without pin-shoes conditions. These differences with the baseline were integrated for a 'deviation area'. The integration time step size was obtained by dividing the repetition-average duration of each perturbed sequence by its 50 samples, therefore yielding a different time step size for every perturbed condition. The total integral for all conditions spanned from the end of the perturbation to time $t_A = 1.108$ s, which was the longest subject-average gait cycle duration among all perturbed conditions with pin-shoes. The obtained areas quantify how much a subject deviates from unperturbed walking up to time t_A as a result of the perturbations, and can be compared between the with and without pin-shoes conditions. To obtain the deviation areas for the trunk angular excursion, we integrated over the absolute difference between perturbed and unperturbed conditions instead, as the integral of an angle over time has little physical meaning while positive and negative deviations can cancel out. Finally, the repetition-average data and deviation areas per subject were averaged across subjects for subject-averages and standard deviations.

Linear mixed models were used to assess the effect of the pin-shoes (fixed effect, with intercept), and its interaction with perturbation magnitude (fixed effect, with intercept) on the deviation areas. Tests were done separately for forward and backward perturbations. To account for correlated data resulting

from repeated measures within a single subject, a random subject factor (intercept) was included. The significance level was set to 0.05, and a Bonferroni correction was applied during *post hoc* analysis to correct for multiple comparisons. In the *post hoc* analysis we focus on the effect of the pin-shoes on the areas, and do not mutually compare the area sizes for the different perturbation magnitudes and directions. Statistical analysis was performed with SPSS statistics 21 (IBM Corporation, Armonk, NY, USA).

To investigate COM-velocity effects on foot placement adjustments, linear least-squares fits were made to the subject-average data at previously investigated gait events (Vlutters et al., 2016), see Table 1. For comparison with the fits, an XCOM proportionality constant $w_0^{-1} = \sqrt{l/g}$ and an offset $b_{AP} = S_c / (e^{T_c \sqrt{g/l}} - 1)$ were calculated for each subject following (Hof, 2008), where l is the subject's leg length plus pin height, g is the Earth's gravitational constant, S_c is some desired step length, and T_c is some desired single support duration. In a linear inverted pendulum model of walking, proportionality constant w_0^{-1} relates the model's horizontal COM velocity to a stepping location that terminates model movement (i.e. the XCOM). Each subject's w_0^{-1} was used to find a subject-average w_0^{-1} , and a subject-average AP XCOM = $w_0^{-1} * \dot{X}$ where the XCOM is relative to the COM, and $\dot{X}$ corresponds with the horizontal AP COM velocity. Offset b_{AP} can be used in 'constant offset control' proposed in (Hof, 2008), to make a linear inverted pendulum model converge toward a desired gait with a step length S_c and single support duration T_c . Each subject's average step length (AP pin-pin distance at heel strike) and single support duration (time between toe-off and subsequent heel strike) during unperturbed pin-shoe walking were used to calculate this offset.

Results

COM velocity

During unperturbed walking, the step frequency was higher in subjects wearing pin-shoes (1.88 ± 0.11 Hz) than in subjects without pin-shoes (1.40 ± 0.08 Hz) at similar speeds. The perturbations lead to additional movements that temporarily deviate from the unperturbed case. The deviation areas quantify the degree to which subjects deviated from the unperturbed walking condition within the chosen time window, for both the with and without pin-shoes conditions.

The pin-shoes affect the COM velocity the most for backward perturbations. Backward perturbed subjects with pin-shoes showed significantly more deviations from the unperturbed walking velocity than subjects without pin-shoes ($F_{1,18}=146.200$, $p<0.001$), as reflected by the deviation areas in Figure 2I. These areas were furthermore significantly affected by the interaction between pin-shoes and perturbation magnitude ($F_{3,54}=18.639$, $p<0.001$). Post hoc comparison showed that the areas were significantly larger (more negative) in the pin-shoes condition for all backward perturbations ($p \leq 0.019$). In contrast, for forward perturbations there was no significant effect of the pin-shoes ($F_{1,18}=1.951$, $p=0.179$), unless when interacting with perturbation magnitude ($F_{3,54}=3.368$, $p=0.025$). Only for the two smaller magnitude forward perturbations were the deviation areas significantly larger in the pin-shoes condition ($p \leq 0.040$).

Figure 2. Subject-average responses to the perturbations during walking, with (A,C,E,G) and without (B,D,F,H) pin-shoes over one gait cycle. (A,B) AP COM velocity. (C,D) Sagittal-plane trunk angular acceleration. (E,F) Sagittal-plane trunk angular excursion. Positive corresponds with backward trunk angular accelerations and excursions. (G,H) Duration from perturbation onset at TOR to subsequent gait events. Horizontal bars indicate the subject standard deviation. Vertical dashed lines indicate the integration time window for the deviation areas. (I,J,K) Deviation area for COM velocity, trunk angular acceleration, and trunk angular excursion per perturbation. Vertical bars indicate the subject standard deviation. Dotted line corresponds to data with pin-shoes, solid line to data without pin-shoes. Colors indicate the perturbation magnitudes as fraction of the subject's body weight. In (A-F) standard deviations (shaded area) of the perturbed conditions are not shown to prevent image cluttering.

Trunk response

During the perturbations, the trunk initially obtains a backward (+) angular acceleration for forward perturbations and a forward (-) angular acceleration for backward perturbations, attributed to a passive response. Subsequently, subjects must actively accelerate the trunk in the opposite direction to prevent it from toppling over. Although in terms of magnitude these trunk accelerations tend to deviate more from those in unperturbed walking in the pin-shoe condition, the sequence within which these deviations occur (P.End – HSR) is generally also of shorter duration in the pin-shoe condition. Integrating the angular acceleration deviations leads to deviation areas which are significantly smaller in the pin-shoe condition for backward perturbations ($F_{1,16}=12.108$, $p=0.003$), but no different for forward perturbations ($F_{1,16}=0.025$, $p=0.876$). Neither for forward nor for backward perturbations there was an interaction effect between pin-shoes and perturbation magnitude ($p>0.658$). Similarly, the pin-shoes had no significant effects on the trunk angular excursion areas for both forward ($F_{1,16}=0.006$, $p=0.940$) and backward ($F_{1,16}=0.865$, $p=0.366$) perturbations, nor where there any interaction effects with perturbation magnitude ($F_{3,48}\leq 2.048$, $p\geq 0.120$). Overall, the results suggests that COM velocity is affected by the pin-shoes for various perturbations, while the trunk response remains similar between both conditions, or even tends to decrease in the pin-shoe condition.

Foot placement

Following perturbations at TOR, subjects made the first recovery step with the right leg. On average, subjects increased the AP distance between the COM and the tip of the right pin-shoe with increasing forward COM velocity at HSR, and decreased it with decreasing COM velocity, eventually placing the pin behind the COM. This foot placement modulation did not occur in subjects that walked without pin-shoes (Vlutters et al., 2016). Following contact, subjects shifted their weight to the newly placed foot, resulting in a COP location at TOL that is beneath the right pin. The relation of the COM velocity at HSR with both the distance between the pin and the COM at HSR, and with the distance between the COP and the COM at TOL, can be quantitatively explained with a linear relation, see Table 1 and Figure 3. The slope of the latter relation in Table 1 is about equal to the subject-average ω_{θ}^{-1} (0.327 ± 0.008 , mean $\pm$ std). Under the condition that the slope of this fit is equal to that of the XCOM line in Figure 3 (i.e. ω_{θ}^{-1}), the intercept of the fit provides a constant offset between the COP and the XCOM of -0.158 m. For comparison, the offset b_{AP} calculated according to (Hof, 2008) was -0.129 ± 0.021 m, using the step length (0.311 ± 0.04) and single support duration (0.402 ± 0.036) during unperturbed pin-shoe walking. From the perspective of a linear inverted pendulum model this negative offset would make the pendulum topple over the foot, which is indeed required in the sagittal plane to continue walking forward.

Discussion

We investigated balance recovery responses to AP pelvis perturbations during walking with constrained COP modulation within the feet to examine preferences in balance recovery strategies, and to compare the foot placement strategy with concepts derived from a simple inverted pendulum model. The results suggest that the trunk is not additionally addressed despite the more challenging pin-shoe condition. Foot placement adjustments are used in the recovery instead, in line with the XCOM concept.

Table 1. Slope, intercept and coefficient of determination of the linear least squares (LLSQ) fit made to the subject-average data at specific instances of the gait cycle after the perturbation.

Dependent variable →	AP COM velocity, at HSR		
Independent variable ↓	Slope	Intercept	R ²
AP distance right pin – COM, at HSR	0.421	-0.148	0.926
AP distance right pin, COM, at TOL	0.409	-0.188	0.986
AP distance COP – COM, at TOL	0.325	-0.158	0.986

Figure 3. Subject-average foot placement responses. (A) AP COM velocity at HSR against the distance between the COM and the right pin-tip at HSR. (B) AP COM velocity at HSR against the distance between the COM and the COP at TOL. Triangles show subject averages and indicate the perturbation direction. Ellipses represent the subject standard deviation. Black dashed line is a linear least square fit to the data. R^2 indicates the coefficient of determination of the fit. Pink solid line has slope ω_r^{-1} and no intercept, and indicates the subject average of the XCOM relative to the COM. Pink dashed lines indicate the corresponding standard deviation. The two vertical dotted arrows have the same length, and indicate the magnitude of the intercept of the fit. The intercept corresponds to a constant offset between the fit and the pink XCOM line, assuming the slopes of both lines are equal. Colors indicate the perturbation magnitudes as fraction of the subject's body weight.

Velocity recovery

Step frequency was higher in subjects with pin-shoes, possibly due to the inability to deliver a push-off, and to reduce the collision costs at heel strike (Kuo, 2002). This higher frequency could partially explain why it takes more steps to counteract the perturbation-induced velocity changes. Based on the deviation areas, deviations from the unperturbed walking velocity up to t_A tend to be of larger magnitude for subjects with pin-shoes. This is presumably caused by the inability to deliver an ankle moment. There is no intrinsic stiffness or damping about the point of rotation with the floor that affects the passive response to the perturbation, nor can an active moment be generated to influence body movement during single support. Hence, in the pin-shoe conditions the perturbations tend to cause larger deviations in COM velocity, while at the same time the ankle strategy is ineffective in

counteracting these deviations. Two alternative recovery options are trunk motion and foot placement adjustment.

Trunk response

A hip strategy has low priority in perturbation recovery when foot placement adjustments are possible, and maintaining an upright upper body appears to be an important objective even under destabilizing conditions. Subjects cannot instantly react to the perturbations, hence the trunk response during the perturbation likely contains passive effects. The lower peak accelerations during the perturbations for subjects with pin-shoes compared to those without pin-shoes suggest an increased trunk impedance prior to perturbation onset. This possibly results from proactive trunk stabilization (Winter, 1995) in response to wearing the pin-shoes. Trunk accelerations directly after the perturbations must contain active contributions from hip moments. These both help the trunk return toward its unperturbed state, and initially help mitigate the perturbation effects on the COM velocity through inertial effects. While it is difficult to discriminate between these two purposes, it appears that subjects try to prevent deviations from the unperturbed trunk state, rather than using a hip strategy to influence COM motion. This is because subjects with pin-shoes do not show increased trunk angular acceleration or angular excursion areas, even though they cannot use an ankle strategy to counteract the perturbations. Similarly, findings in (Oddsson et al., 2004) suggest that humans tend to stabilize the trunk, and mainly use foot placement to recover from diagonal support surface perturbations. If instead the perturbation would have been given briefly before foot contact, there is little time to correct through foot placement adjustments (Hof et al., 2010). In such a scenario subjects might be more inclined to use the trunk for maintaining balance, or transfer the recovery to the subsequent step. The trunk findings therefore hold under the condition that at least foot placement adjustment is an available recovery option.

Velocity-related foot adjustment

Contrary to our previous findings in subjects without pin-shoes (Vlutters et al., 2016), subjects with pin-shoes actively adjusted their AP foot placement relative to the COM in the first recovery step after the perturbations. Still, linear relations previously observed between the AP COM velocity at HSR and the AP COM-COP distance at TOL remain intact. Without pin-shoes this can be partially realized through ankle moment modulation, which provides a window for COP displacements. With pin-shoes, this is only possible through foot placement adjustments. Hence a similar relation is therefore observed through strategies that differ between the with and without pin-shoes conditions.

The observed foot placement adjustments resemble velocity-based strategies previously applied in low-dimensional models of walking (Hof, 2008) and running (Peuker et al., 2012). The slope of the aforementioned relation is about equal to Ω_0^{-1} , which is the proportionality constant between COM velocity and the XCOM (Hof, 2008). Furthermore, the intercept of the fit provides a (near) constant offset from the XCOM, which is comparable to the calculated offset b_{AP} . Both of these findings support the use of the linear inverted pendulum model and constant offset control for predictions in the first recovery step.

It must be noted, however, that there are some important differences between human and model. In humans, it takes until the end of the double support phase for the COP to shift toward the leading foot, which is required for the relation presented in Figure 3b. In the linear inverted pendulum model, a

double support phase does not occur and the COP shifts instantaneously at heel strike. To calculate b_{AP} we used the average unperturbed single support duration, which does not include the time that is required to shift the COP forward during double support. If this time would be included, T_c increases and b_{AP} would decrease to deviate more from the intercept presented in Table 1. Furthermore, b_{AP} is only meaningful in constant offset control proposed in (Hof, 2008) if the single support time is kept constant, while humans will modulate this time. The model therefore does not provide a complete explanation of the data.

Implications

The presented findings have implications for the design and control of assistive devices, such as lower-extremity exoskeletons and prostheses. Humans might adopt different balance recovery strategies if no ankle control is present, such as when using passive prostheses, or in gait with impaired control of the distal joints. In unilateral amputees, shifts in AP COP were significantly lower in the prosthetic limb compared to the intact limb in six different walking conditions (Kendell et al., 2010). The inclusion of ankle actuation could increase the similarities with an intact limb, improve balance control during walking (Kim and Collins, 2015), and possibly result in balance control that is more similar to that of a healthy human. When considering an able-bodied user of a lower-extremity exoskeleton, such as a rescue or construction worker, a conflict might occur between the natural response of the user and the control of the device if the device has not the same degrees of freedom as its user. As demonstrated in this work, balance control is dependent on the constraints on the human body. Having such an exoskeleton assist in maintaining balance therefore requires at least human-like capabilities of the device, without impeding the user.

Conclusions

Foot placement adjustments are essential for perturbation recovery when ankle contributions are ineffective. Considering that a hip strategy was available but not additionally addressed in a challenging pin-shoes condition, we might conclude that this strategy has low priority during walking in general when foot placement adjustments are an option. Hence, humans prefer to keep their trunk upright when facing a balance threat.

By introducing the pin-shoes we furthermore showed that an earlier presented relation between the COM velocity and the COP is still established by the subjects, but now with the help of foot placement adjustments rather than with ankle moment modulation. This relation was in line with a concept derived from a linear inverted pendulum model, and therefore motivates the use of low dimensional models as a basis for mimicking concepts of reactive balance during human walking.

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CHAPTER 6

Paretic versus non-paretic stepping responses following pelvis perturbations

"It's dangerous to go alone..."

Published as:

**Paretic versus non-paretic stepping responses following
pelvis perturbations in walking chronic-stage stroke survivors**

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Abstract

Introduction: The effects of a stroke, such as hemiparesis, can severely hamper the ability to walk and to maintain balance during gait. Providing support to stroke survivors through a robotic exoskeleton, either to provide training or daily-life support, requires an understanding of the balance impairments that result from a stroke. Here, we investigate the differences between the paretic and non-paretic leg in making recovery steps to restore balance following a disturbance during walking.

Methods: We perturbed ten chronic-stage stroke survivors during walking using mediolateral perturbations of various amplitudes. Kinematic data as well as gluteus medius muscle activity levels during the first recovery step were recorded and analyzed.

Results: The results show that this group of subjects is able to modulate foot placement in response to the perturbations regardless of the leg being paretic or not. Modulation in gluteus medius activity with the various perturbations is in line with this observation. In general, the foot of the paretic leg was laterally placed further away from the center of mass than that of the non-paretic leg, while subjects spent more time standing on the non-paretic leg.

Conclusions: The findings suggest that, though stroke-related gait characteristics are present, the modulation with the various perturbations remains unaffected. This might be because all subjects were only mildly impaired, or because these stepping responses partly occur through involuntary pathways which remain unaffected by the complications after the stroke.

Introduction

Stroke survivors often experience problems with maintaining their balance. A variety of neurological deficits can hamper balance control during walking, such as hemiparesis, sensory impairments, as well as cognitive problems such as fear of falling. As a consequence, fall rates in stroke survivors are 2-8 times higher than those in healthy age-matched subjects (Weerdesteyn et al., 2008). In general, especially balance control in the frontal plane is often considered challenging, requiring adequate foot placement to continue walking (MacKinnon and Winter, 1993). This might be an additional challenge when suffering from hemiparesis following a stroke, which could lead to differences in recovery steps made with the paretic and the non-paretic leg in response to a disturbance during walking. To reduce the fall risk of stroke survivors and make their rehabilitation more effective, it is required to characterize how balance control is affected by a stroke. Such knowledge might be used to provide limb-specific support in robot-assisted gait rehabilitation.

Stroke survivors typically show differences in gait characteristics between the paretic and non-paretic leg during unperturbed walking, for example as a result of decreased motor control in the paretic leg. In (Balasubramanian et al., 2010) subjects placed the paretic leg at an increased lateral distance from the pelvis compared to the non-paretic leg in mediolateral (ML) foot placement during unperturbed walking. However, no differences in step width were found with regard to the leg used for stepping. It is therefore of importance to consider both legs individually, in a body referenced frame such as that of the center of mass (COM). Dean et al. studied the relation between gluteus medius muscle activity in the swing leg and both the ML position and velocity of the COM relative to the stance foot (Dean and Kautz, 2015). For low fall-risk subjects the results suggest a stronger activity modulation in the non-paretic swing leg than in the paretic swing leg, though it did not show how both legs respond to actual destabilizing conditions such as external perturbations.

Perturbations can be used to affect the body state, such as the position and velocity of the COM relative to the stance foot. This may lead to adjustments in foot placement location and timing to maintain balance. In (Krasovsky et al., 2013) perturbations were applied by unexpectedly arresting the ankle of the leg at early swing. Stroke survivors showed shorter step lengths and shorter swing times compared to healthy controls when stepping in response to perturbations applied to the non-paretic swing leg. Furthermore, in (Hak et al., 2013) continuous ML support surface translations were used to assess if and how low-fall risk stroke survivors change their base of support (BoS) during walking through foot placement adjustments. It was found that stroke survivors more strongly shortened their step length and step time, and increased their step width, as compared to healthy controls. However these studies did not compare the perturbation responses of the individual paretic and non-paretic limbs.

In healthy subjects, ML foot placement adjustments in response to ML perturbations relate to the ML COM velocity (Hof, 2008; Vlutters et al., 2016). This is reflected in a concept called the extrapolated center of mass (XCOM), which has been previously used to indicate fall risk in relation to foot placement (McAndrew Young et al., 2012; Hak et al., 2013). The XCOM can be derived from a linear inverted pendulum model (Hof et al., 2005; Pratt et al., 2006) and can be regarded as a point on the floor at a horizontal distance from the COM. This distance is directly proportional to the horizontal COM velocity through a proportionality constant ω_0^{-1} . If the inverted pendulum model would place its point-foot onto the XCOM, the model will stop moving in an upright position. Placing the foot further away from the COM than the XCOM will cause the model to fall back in the direction from which it

came. The further the model places its foot beyond the XCOM, the sooner the model is redirected in the opposite direction. Such movement is relevant to the ML direction of walking, where toppling over the leg is often not desired. Hence, investigating paretic and non-paretic foot placement in relation to the XCOM allows analysis from a simple model perspective.

The main purpose of this study is to characterize differences between the paretic and the non-paretic legs in walking chronic-stage stroke survivors in terms of ML stepping following ML pelvis perturbations. Rather than applying continuous disturbances, we will use transient perturbations directed inward and outward. We will also investigate how the steps are made in relation to the XCOM. Only low fall-risk subjects will be considered, to reduce the risk of actual falls in response to the perturbations.

Methods

Participants

Ten stroke survivors in the chronic stage (7 male, age: 52 ± 16 years, weight: 82.5 ± 13.6 kg, height: 1.75 ± 0.06 m, mean $\pm$ std) were recruited. All gave written informed consent before participation. All participants had a Functional Ambulatory Category score of 4, meaning that these subjects can walk independently on even surfaces, but might require assistance in more challenging situations such as uneven terrain (Holden et al., 1984). Related clinical measures can be found in Table 1. The experimental setup and protocol were approved by the local ethics committee, and was in accordance with the Declaration of Helsinki.

Apparatus

Subjects walked on an instrumented treadmill (custom Y-mill, Motekforce Link, Culemborg, the Netherlands), used to collect 3 degrees-of-freedom ground reaction forces and torques per foot. Subjects could be perturbed in the ML direction using a motor (SMH60, Moog, Nieuw-Vennep, the Netherlands) bolted to a support structure clamped to the treadmill's exterior frame. A brace (Distrac Wellcare, Hoegaarden, Belgium) worn by the subject could be connected to the motor through an aluminum horizontal rod and a vertical lever arm. The rod was connected to both the brace and the lever arm through ball joints, to allow freedom of movement in the anteroposterior (AP) direction. Given the motor range of motion, the maximum lateral excursion from the center of the treadmill was approximately 0.55 m in each direction. The motor was admittance controlled over Ethernet (User Datagram Protocol) at 1000 Hz, using xPC-target (The Mathworks, Natick, US). A schematic overview of the experimental setup can be found in Figure 1A. Details can be found in (Vlutters et al., 2016).

Data collection

Subject kinematic data were collected at 100 Hz using a 9 camera motion capture system (Visualeyez II, Phoenix Technologies Inc, Burnaby, Canada). Marker clusters, each consisting of 3 LEDs, were placed on the feet, lower legs, upper legs, pelvis, sternum and head. Additional single LEDs were placed on both lateral malleoli, and both lateral epicondyles of the femur. Gluteus medius EMG activity in both legs was recorded using a Bagnoli Desktop EMG system (Delsys, Natick, MA, USA) at 1000 Hz. Finally, subjects were filmed from the rear using a video camera to be able to detect any instances during which the subject grasped the handrails of the treadmill.

Figure 1. Experimental setup and single-subject perturbation profiles. (A) Schematic overview of the experimental setup. In the schematic the left leg is the stance leg. For the right stance leg the leftward and rightward arrow colors interchange. (B) Reference (dashed) and measured (solid) interaction force between motor and subject. (C) Reference (dashed) and measured (solid) motor impulse, obtained by integrating the interaction forces. (D) ML COM velocity relative to the walking surface. (E) EMG profile of the swing leg gluteus medius. Colors indicate the various perturbations. Lines represent within-subject averages of a single subject. Shaded areas indicate the within-subject standard deviation. The subject shown was the subject with the highest step frequency among subjects. The square on the time axis indicates the average time of heel strike after perturbation onset at toe-off (paretic and non-paretic pooled).

Protocol

Various bony landmarks were indicated using an LED based probe (Cappozzo et al., 1995), prior to the experiment. These landmarks were the 1st and 5th metatarsal heads, calcaneus, medial and lateral malleoli, fibula head, medial and lateral epicondyles of the femur, greater trochanter of the femur, anterior and posterior superior iliac spines, xiphoid process, jugular notch, 7th cervical vertebra, occiput, head vertex and nasal sellion (Dumas et al., 2007). Subjects wore a safety harness at all times to prevent injury in case of a fall. The preferred walking speed was determined by gradually increasing the speed of the treadmill, until the subject felt comfortable. This was also used to have subjects familiarize with walking on the treadmill.

During the experiment subjects wore the brace around the pelvis and were instructed to walk at the center of the treadmill. First, a two minute baseline measurement at the preferred walking speed was recorded without applying perturbations. When no perturbations were applied the motors were admittance controlled such that the interaction force between the subject and the motor was as low as possible, see Figure 1B. In subsequent trials, perturbations could be randomly applied at the instance of

toe-off of either leg. This way subjects were required to use both the paretic and the non-paretic leg in the first recovery step. The toe-off instances were detected using the vertical ground reaction forces. Perturbations were 150 ms block pulses of magnitudes equal to 4, 7, and 10% of the subject's body weight. The direction could be inward (e.g. leftward for left stance) or outward (e.g. rightward for left stance). Though the motors cannot track the desired reference force, the delivered impulse is approximately as desired, see Figure 1C. Each perturbation condition was repeated 5 times, yielding 60 perturbations per subject. The perturbations were randomized over magnitude, direction, and onset instance (paretic or non-paretic toe-off). The minimum time between successive perturbations was also randomized, and varied between 6 and 12 seconds. Finally, Berg Balance Scale (BBS), Dynamic Gait Index (DGI), Mini-Mental Scale Examination (MMSE), Fall Efficacy Scale International (FES-I) and 10-meter-walking-test (10MWT) measures were collected within one week after the experiment.

Data processing

Data were processed using Matlab (R2013b, Mathworks, Natick, US). EMG data were detrended, filtered with a 1st order 48-52 Hz Butterworth bandstop filter, then rectified, and filtered with a 1st order 20 Hz Butterworth low-pass filter. Marker data were filtered with a 4th order 20 Hz zero-phase Butterworth low-pass filter before reconstructing the global bony landmark positions from the probe measurements. Using the landmarks, the COM positions of the captured segments were estimated following (Dumas et al., 2007). The total body COM was calculated as a weighted average of the segment COM positions, and was differentiated for a COM velocity. All other position and velocity data were expressed relative to those of the COM. Gait events of toe-off and heel strike were detected using landmarks of the feet (Zeni Jr et al., 2008). Instances of perturbation onset were detected from the motor input signals. The recorded video data were visually inspected for perturbations during which subjects touched the hand-rails of the treadmill. Corresponding trials were removed from analysis.

For each subject, the position of the COM of both feet relative to the total-body COM, as well as the COM velocity, were extracted at the instances of heel strike and toe-off after the perturbation. The gait phase durations between these instants were also calculated. The EMG data of each individual muscle was scaled to the median of the maximum values occurring every gait cycle during the unperturbed walking condition. For each subject and in each condition, the scaled EMG time-averages between toe-off and heel strike were calculated, representing the averaged muscle activity level during the swing phase. All these data were sorted on perturbation magnitude, perturbation direction (inward or outward), and leg used for stepping (paretic or non-paretic). Since five right-sided and five left-sided paretic subjects participated in the study, all position, velocity, and EMG data resulting from the left side of the body were laterally mirrored for all subjects. All data in this paper is therefore presented as if the first recovery step was made with the right leg, for both the paretic and the non-paretic leg of all subjects. Next, for each subject all data were averaged over the repeated conditions to obtain repetition averages. These were in turn averaged for subject-averages and standard deviations. Finally, for each subject an XCOM proportionality constant was calculated following $\omega_0^{-1} = \sqrt{l/g}$ (Hof, 2008), in which l is the subject's leg length, and g the earth's gravitational acceleration. These ω_0^{-1} were averaged across subjects for a subject-average proportionality constant.

Statistical analysis

Linear mixed models were used to determine the effect of all perturbations (fixed factor, with intercept), the effect of the leg used in the first recovery step (fixed factor, with intercept), and their interaction effect on the outcome measures. Outcome measures were the ML and AP distance between the leading foot and the COM, the ML and AP distance between both feet (step width and step length, respectively), the ML COM velocity at heel strike, the gait phase durations following the perturbations, and the EMG signal averages. To account for correlated repeated measures within the same subject, a random subject factor (intercept) was included in the model. The significance level was set to $\alpha = 0.05$, and a Dunn-Šidák correction was applied during post hoc analysis. In the post hoc comparison, perturbation data were only compared to the unperturbed condition and not mutually. The Statistical Package for the Social Sciences version 19.0 (IBM Corporation, Armonk, NY, USA) was used for the statistical analysis.

Results

All subjects completed the experiment without falling. Subjects 1, 6, and 7 did not complete the DGI and the 10MWT. Subjects 1, 2, and 3 grabbed the handrails 4, 4, and 6 times respectively, mainly in response to the largest magnitude inward perturbations. The corresponding data were removed from analysis. Subject characteristics are presented in Table 1.

Foot placement location in ML and AP directions

Perturbations resulted in foot placement modulation. Here, we specifically focus on the location of the leading foot. Perturbations were applied such that each subject had to make the first recovery step with both the paretic- and the non-paretic leg. The locations of the feet relative to the COM at the instant of the first heel strike after the perturbation are presented in Figure 2. Note that because the data is presented relative to the COM, the stance foot data changes with the perturbations because the body is pushed away from it or towards it. Foot placement relative to the COM as a result of the different perturbations was not found to differ between the paretic and non-paretic leg, for both ML ($F_{6,117}=0.981$, $p=0.441$) and AP ($F_{6,117}=0.068$, $p=0.999$) distances between the COM and the leading foot. This indicates that no interaction effect was present between factors perturbation and leg used for stepping.

When considering the effects of the leg used in the stepping response, it is important to regard the foot location relative to the COM rather than relative to the trailing foot. In the frontal plane, the foot of the paretic leg is always laterally placed further away from the COM than the foot of the non-paretic leg ($F_{1,123}=26.956$, $p<0.001$). Instead, when considering step width, i.e. the ML the distance between the feet, no significant differences were found between the legs used in the first recovery step ($F_{1,123}=2.086$, $p=0.151$). For the AP distance between the COM and the leading foot no significant effects of factor leg were found ($F_{1,123}<0.001$, $p=0.999$). However the step length, i.e. the AP distance between the feet, was larger when stepping with the paretic leg compared to stepping with the non-paretic leg ($F_{1,123}=9.294$, $p=0.003$). The latter effects are therefore caused by the trailing leg, and are likely related to the time spent standing on that leg.

Mediolateral adjustments with the leading foot mostly occur in the direction of the perturbation. Especially for outward perturbations the foot appears to be placed further in the direction of the perturbation with increasing perturbation magnitude. Significant effects of factor perturbation were found for the ML distance between the COM and the leading foot ($F_{6,123}=7.215$, $p<0.001$), as well as for the AP distance between the COM and the leading foot ($F_{6,123}=24.799$, $p<0.001$). Post hoc analysis revealed that this AP distance decreased significantly for all outward perturbation compared to unperturbed walking ($p<0.002$), but did not change significantly for all inward perturbations ($p>0.160$).

Figure 2. Positions of the COM of the feet relative to the whole body COM. (A) Locations of the COM of both the leading and the trailing foot relative to the whole-body COM, at the instant of heel strike, for steps made with the paretic leg. (B) Same as in A, for steps made with the non-paretic leg. Triangles show subject averages and indicate the perturbation direction. Ellipses represent subject standard deviations. Colors indicate the perturbation magnitudes as a fraction of the subject's body weight.

Table 1. Subject characteristics

ID	Gender (M/F)	Age (yrs.)	Time post-stroke (yrs.)	Affected leg (L/R)	Weight Distr. Stance (%)	Weight Distr. Walking (%)	BBS (pts)	DGI (pts)	MMSE (pts)	FES-I (pts)	10MWT (km/h)	Treadmill speed (km/h)
1	M	66	10	R	54/46	58/42	51	-	28	16	-	0.4
2	M	30	7	R	48/52	56/44	55	22	30	30	4.5	1.0
3	F	70	9	R	51/49	59/41	55	22	26	17	1.9	0.7
4	M	63	10	L	54/46	54/46	55	22	27	19	4.6	2.0
5	M	38	5	R	40/60	50/50	48	21	28	29	3.3	1.6
6	M	67	7	L	55/45	57/43	53	-	30	22	-	2.2
7	M	64	4	L	56/44	54/46	56	-	26	16	-	5.0
8	M	45	3	R	46/54	43/57	56	23	27	34	4.7	2.2
9	F	29	8	L	54/46	61/39	56	22	28	17	3.7	2.2
10	F	46	29	L	61/39	58/42	56	24	30	17	4.7	2.4
Mean $\pm$ std	7M, 3F	52 $\pm$ 16	9 $\pm$ 7	5R, 5L	52/48	55/45	54 $\pm$ 3	22 $\pm$ 1	28 $\pm$ 2	22 $\pm$ 7	3.9 $\pm$ 1	2.0 $\pm$ 1.3

BBS = Berg Balance Score, DGI = Dynamic Gait Index, MMSE = Mini-Mental State Examination, FES-I = Falls Efficacy Scale International, 10MWT = 10 Meter Walking Test

Relation between ML foot placement and ML COM velocity

We investigated the mediolateral stepping location in relation to the ML COM velocity at the instant of heel strike, specifically in relation to the location of the XCOM at heel strike, see Figure 3. If the ML distance between the COM and the leading foot (y-axis) would increase with increasing mediolateral velocity (x-axis) at the same rate (ω_0^{-1}) as that of the pink XCOM line, then the foot would be placed at a fixed distance from the XCOM (Hof, 2008). This would provide a simple explanation for lateral foot placement, based on a concept derived from a linear inverted pendulum model.

Factors perturbation and leg had a significant interaction effect on the ML COM velocity following the perturbations ($F_{6,114,068}=2.587$, $p=0.022$). As a result, steps made with the non-paretic leg show more variability in the COM velocity between the various perturbations, as compared to that of steps with the paretic leg. That is, for steps with the non-paretic leg outward perturbations lead to more positive COM velocities, whereas for inward perturbations it leads to smaller, or even a slightly negative COM velocity. Factor leg also provided a significant main effect on the ML COM velocity, with a higher velocity for steps made with the non-paretic leg ($F_{1,114,058}=7.011$, $p=0.009$). Furthermore, the various perturbations significantly affected the ML COM velocity at heel strike ($F_{6,114,068}=82.830$, $p<0.001$), increasing in the direction of the perturbation with increasing perturbation magnitude (post hoc: step with paretic: $p\leq 0.027$ for all perturbations except for 0.04, 0.07 outward, step with non-paretic: $p\leq 0.007$ for all perturbations except for 0.04 outward). While at heel strike the ML distance between the COM and the leading foot showed modulation with the COM velocity, the average distances do not appear to modulate with the COM velocity in the same way as the XCOM does. That is, the ML distance of the foot relative to the COM does not relate to the ML COM velocity through a factor ω_0^{-1} . Instead, both the paretic and the non-paretic leg show an increase in foot-COM distance with increasing outward perturbation magnitude, but remain approximately the same with increasing inward perturbation magnitude. Therefore, for the presented results the XCOM alone is not a sufficient predictor of the lateral stepping location, though this might be different in a more homogeneous group.

Gait phase durations

For all presented gait phase durations, modulation with the perturbations is invariant of the leg with which the step is made. No significant interaction effects were found between factors perturbation and leg on the duration of the swing phase in which the perturbation was applied ($F_{6,117}=0.163$, $p=0.986$), nor on the subsequent double support phase duration ($F_{6,115,993}=1.045$, $p=0.400$).

The most prominent gait phase duration modulation occurred in the swing phase during which the perturbation was applied. The duration of this swing was significantly affected by the perturbations ($F_{6,123}=26.035$, $p\leq 0.001$), significantly increasing with increasing inward perturbation magnitude ($p\leq 0.045$), and decreasing with increasing outward magnitude ($p\leq 0.002$, except 0.04 outward), see Figure 4. In contrast, the subsequent double support durations were not significantly affected by factor perturbation ($F_{6,121,993}=1.864$, $p=0.092$).

The gait phase durations also reflect the differences between steps made with the paretic and the non-paretic leg. Subjects preferred to spend more time on the non-paretic leg than on the paretic leg during the single support phase, regardless of which perturbation was applied ($F_{1,123}=16.448$, $p\leq 0.001$). Furthermore, the double support phase during which the weight is shifted from the paretic leg toward the non-paretic leg also last longer compared to the opposite shift ($F_{1,121,993}=23.365$, $p\leq 0.001$).

Figure 3. Distance between the COM of the leading foot and the whole body COM against the COM velocity. (A) For recovery steps made with the paretic leg, at the instant of the first heel strike after the perturbation. (B) Same as in A, but for steps made with the non-paretic leg. Triangles show subject averages and indicate the perturbation direction. Ellipses represent subject standard deviations. Colors indicate the perturbation magnitudes. The pink dashed line indicates the position of the XCOM relative to the COM. This line has a slope of ω_0^{-1} and no intercept.

Gluteus medius activity

Subjects modulated the gluteus medius activity in the swing leg with the various perturbations, for both the paretic and non-paretic leg. Though more varied modulation appears to occur in the non-paretic leg over all perturbations, no interaction effects were found between factor leg and perturbation magnitude ($F_{6,88,978}, p=0.452$).

However, a significant effect of the perturbations was found on the gluteus medius activity levels ($F_{6,94,917}=18.993, p<0.001$), increasing on average. Especially for outward perturbations where the gluteus medius participates in foot placement adjustments through leg abduction, increased activity levels can be observed with increasing perturbation magnitude ($p<0.001$, except for 0.04 outward), see Figure 5. For inward perturbations the post hoc comparison provides no significant effects, though the non-paretic swing leg appears to show more EMG modulation than the paretic swing leg. Differences between both legs were not tested as activity levels of both legs were scaled to baseline values of each leg.

Figure 4. Gait phase durations. (A) The single support duration during which the perturbations were applied, and the subsequent double support duration, for recovery steps made with the paretic leg. (B) Same as in A, but for recovery steps made with the non-paretic leg. Triangles show subject averages and indicate the perturbation direction. Vertical bars represent subject standard deviations. Colors indicate the perturbation magnitudes.

Figure 5. (A) For the paretic leg, modulation with the perturbation magnitude of the time-average EMG signals between toe-off (perturbation onset) and heel strike. (B) As in A, for the non-paretic leg. Triangles show subject averages. Vertical bars represent subject standard deviations. Colors indicate the perturbation magnitudes.

Discussion

We investigated the first recovery step following mediolateral pelvis perturbations in chronic-stage stroke survivors who walked at a self-selected speed. The transient perturbation were randomly applied at the instant of toe-off of the paretic and non-paretic leg, such that every subject had to make recovery steps with both legs. The aim was to evaluate and compare stepping responses regarding the use of the paretic and non-paretic leg. In accordance to our hypothesis, we found various differences in gait characteristics between steps made with the paretic and non-paretic leg. However, foot placement modulation in response to the various perturbations did not differ between the legs within the included group of stroke survivors.

Foot placement with respect to the COM

Our findings in perturbed walking support the analysis of foot placement measures with respect to the COM, similar as previously applied in unperturbed walking (Balasubramanian et al., 2010). As in the work by Balasubramanian et al., in the current results the mediolateral distance between the feet is independent of the leg with which the step is made. However, this does not imply that there are no differences between both legs in terms of foot placement, as the paretic leading leg is placed at a consistent increased lateral distance from the body's COM. It has been suggested that this might be related to strength limitations in the paretic leg (Balasubramanian et al., 2010; Hall et al., 2011), leading to the typical weight bearing asymmetry during stance (Bujanda et al., 2003). These limitations could in turn lead to limited control over the COM motion with the paretic leg (Genthon et al., 2008; Roerdink et al., 2009), especially during the single support phase in which the paretic leg is the stance leg. This might also explain the larger variability in ML COM velocity following the perturbations applied during the paretic single support phase compared to those during the non-paretic single support phase. The results suggest that classical gait parameters such as step length and width might not fully capture balance and walking impairments in stroke survivors. Analysis of the feet locations relative to the COM might identify leg specific aspects of the impairment, and help specify gait-phase specific support in robotic gait training and rehabilitation.

Leg-independent step modulation in response to perturbations

The results imply that subjects attempt to execute similar stepping strategies with both legs, and that the modulation with the various perturbations is not hampered by the effects of the stroke. Though the paretic leg is laterally placed further away from the COM than the non-paretic leg when making a step, the way in which the foot placement location alters with the various perturbations does not statistically differ between both legs. Consequently, subjects do not modulate their stepping location worse with the paretic leg than with the non-paretic leg, in response to the various perturbations. Similar effects were observed in modulation of the gait phase duration following the perturbations. Although subjects spent more time standing on the non-paretic leg, they showed comparable changes in gait phase durations with both the paretic and the non-paretic legs in response to the different perturbations. One potential cause might be that the subjects were only mildly affected by the stroke, as classified by their clinical scales. Subjects with lesser scores on clinical outcome measures might have shown larger deviations. For instance, in (Dean and Kautz, 2015) high fall risk subjects showed different stepping strategies compared to low fall risk subjects. Another potential reason might be that the modulation is partly

automatic. For example, (Nashner and Cordo, 1981) showed that balance perturbations during standing can trigger rapid postural adjustments that take precedence over voluntary movements. It might be that these responses involve neural pathways that bypass the brain areas damaged by the stroke.

Gluteus medius activity modulation

It has previously been suggested that swing leg gluteus medius activation for ML foot placement might be based on sensory information from the stance leg (Rankin et al., 2014; Dean and Kautz, 2015). Our outward perturbations move the COM further away from the stance foot with increasing perturbation magnitude. Therefore, as in (Dean and Kautz, 2015), gluteus medius activity increased with increasing ML distance between the COM and the stance foot. These effects occurred in both the paretic and the non-paretic swing legs, without significant interaction effects between factors leg and perturbation. These findings therefore match those for step location and step time modulation in this study, suggesting that modulation with the perturbations is not different between both legs in the current subject group.

With the current setup in stroke survivors, it is not possible to confirm whether this relation between the stance leg and the gluteus medius activity of the swing leg is a causal one. As stroke survivors often suffer from sensory impairments in the paretic leg (Carey, 1995), one would expect information from the stance leg to be less reliable if that leg is paretic. Consequently, if gluteus medius activity modulation would primarily depend on the state of the stance leg, one would expect the paretic swing leg to receive the best information. However, in (Dean and Kautz, 2015) the gluteus medius of the paretic swing leg showed poorer modulation with the state of the stance leg than that in the non-paretic swing leg. It is therefore likely that effects of the stroke on the paretic leg dominate the relations in Dean et al., rather than how well information is conveyed from one leg to the other.

Absence of cross-stepping

In a similar study with healthy subjects, inward perturbations of sufficient magnitude lead to cross-steps (Vlutters et al., 2016). These were not observed in the current study. Though the maximum perturbation magnitudes were not as high as in (Vlutters et al., 2016), subjects visually appeared to explicitly avoid crossing the legs. In case of larger magnitude inward perturbations, subjects tended to move the arms, or abduct the swing leg, possibly using angular momentum strategies (Hof, 2007; Herr and Popovic, 2008) to recover from the perturbations. Similar leg abduction responses were reported in (Dean and Kautz, 2015) for unperturbed high fall-risk subjects. Subjects might be reluctant to make cross-steps, as entangling the legs would hamper gait progression. These effects might be amplified by walking on a treadmill, as it requires the subjects to keep walking to prevent falling off the belt.

Analysis from a model perspective

Evaluating foot placement in relation to the XCOM as derived from a linear inverted pendulum model might aid in understanding why a subject places the foot at a certain location. It can be deduced how such a model would move depending on where its foot is placed relative to the XCOM. In general, subjects laterally placed their foot beyond the XCOM, further away from the COM than the XCOM itself. The foot was placed further beyond the XCOM for steps made with the paretic leg as compared to steps made with the non-paretic leg. Doing so in the model would make the COM fall back toward

the other leg, but at a higher rate for steps made with the paretic leg. This would require the subsequent step with the non-paretic leg to be of shorter duration, therefore spending less time standing on the paretic leg. This is in line with our findings, as well as with other studies in unperturbed walking (Jeon et al., 2013; van Meulen et al., 2016).

Previous studies have investigated the XCOM as a linear predictor for a mediolateral stepping location in healthy subjects (Hof, 2008; Vlutters et al., 2016). In (Vlutters et al., 2016) the COM velocity related linearly to the mediolateral stepping location after the perturbation. While the ML stepping location in our stroke subjects does modulate with COM velocity, and therefore with the XCOM, the data shows substantial variability. The strong linear trend observed in healthy subjects does therefore not hold here. This implies that the COM velocity holds some predictive power for a mediolateral stepping location in this group of stroke survivors, but not as apparent as in healthy subjects, or perhaps not through a linear relation.

Effects of walking speed

The subjects in this study were allowed to walk at a self-selected comfortable walking speed, resulting in a treadmill speed range from 0.4 to 5.0 km/h among subjects. This range leads to variability in spatio-temporal gait parameters such as step length, width, and frequency, which all alter with gait speed. However, our findings focus on the differences between the paretic and the non-paretic legs of the subject population as a whole. The effects of walking speed are likely to be equal for both legs within a single subject, allowing comparison between both legs throughout the subject population.

Limitations

The perturbation device was controlled such that the interaction force between the subject and the motor was as low as possible. The total reflected inertia from the motor to the pelvis is about 1 kg. In (Meuleman et al., 2013) it was concluded that up to 5.3 kg of inertia can be added to mediolateral pelvis movement without significantly affecting the gait. Interaction with the motor is therefore not expected to affect the subjects' gait when no perturbation is applied. However, the rod used to connect the motor to the pelvis brace partly obstructed the right arm sway. Subjects were not able to fully swing their arm backward during the measurement trials, thereby possibly affecting the gait kinematics. Several subjects indicated that the rod was not a problem, or preferred walking with their arms in front of their body to hold the paretic hand. Little effects of this constraint on the lower limb responses are therefore expected.

It has previously been shown that self-selected overground walking speed as assessed by the 10MWT relates to the degree of walking impairment (Perry et al., 1995). Based on this classification, all subjects in our study are within a group of community walkers, and are mildly affected. The subject data shows that the gait speed on the treadmill is often lower than that recorded in the 10MWT. Even though subjects were given a period of time to familiarize with the measurement set-up, subjects might adapt their gait while walking on a treadmill (Bayat et al., 2005) due to being unfamiliar with the measurement setup, or due to fear of falling. However, in Zadavec et al. it was demonstrated that responses of healthy subjects to horizontal pelvis perturbations show high degrees of similarity between overground and treadmill walking (Zadavec et al., 2016). Hence, despite the differences in treadmill and overground walking speed in our subjects, the responses to perturbations are expected to be similar

in both scenario's given a walking speed.

Finally, the variety between subject characteristics such as the walking speed results in more variable data than what would be obtained with a more homogeneous subject group, or with more subjects. The variability in the data could affect the interpretation of the results, such as whether or not the XCOM provides predictive value for lateral foot placement in stroke survivors. Studying more consistent subject groups could reveal to what extent these relations hold for reactive stepping in response to perturbations.

Conclusions

The current work investigated paretic and non-paretic mediolateral stepping responses following ML pelvis perturbations in walking stroke survivors. Stepping responses were investigated for the first step taken after the perturbation. Following a perturbation, subjects preferred to spend more time standing on their non-paretic leg. Analysis in a body-referenced frame identified that the paretic leading leg was placed at a consistently increased ML distance from the body's COM. Despite these differences in gait characteristics, our results suggest that the modulation of the paretic and non-paretic leg with the various perturbations is not greatly hampered by the effects of the stroke. Foot placement and gluteus medius EMG activity imply that subjects attempted to execute similar stepping strategies with both legs. This might be explained because all subjects were classified as having low-fall risk, or because the responses might be partially involuntary, with the modulation not directly affected by the effects of stroke. Though modulation occurred in both legs, mediolateral foot placement in the presented group of stroke survivors could not be predicted using a linear predictor based on the COM velocity. If the subject is able to modulate steps using the paretic leg one might refrain from assisting leg swing during training using a robotic device or exoskeleton, and only provide support after foot contact in weight bearing.

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CHAPTER 7

Lower extremity joint-level responses to pelvis perturbations during human walking

"Do a barrel roll"

Under revision as:

Lower extremity joint-level responses to pelvis perturbation during human walking

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Abstract

Humans can address various strategies to maintain balance during walking, in which the leg joints often play a major role. The aim of this study is to gain a better understanding of how disturbances during walking are rejected through joint-level control in the stance and swing legs. We therefore investigated the bilateral responses of the ankle, knee, and hip joints to sudden perturbations during walking, in terms of joint angles, moments, power, and activation of several major leg muscles acting on those joints. In an experiment, subjects walked on an instrumented treadmill while randomly receiving both anteroposterior and mediolateral pelvis perturbations at the instance of toe-off right. The open-source modeling tool OpenSim was used to perform inverse kinematics and inverse dynamics analysis. The resulting data show that responses in the investigated joints often simultaneously scale with the perturbation magnitude and direction. Ankle and hip control reveal that not only the swing leg, but also the stance leg plays an important role during the recovery, for example to decrease the movement speed in the perturbation direction. We furthermore show that the swing leg is actively accelerated and then halted to prepare for foot placement, suggesting that foot placement is actively controlled. Finally, the recorded muscle activity suggests both spinal and supra-spinal mediated contributions to balance recovery, also scaling with perturbation magnitude and direction. These findings provide a better understanding of the actions that lead to the remarkable resilience of human walking.

Introduction

Reacting to unexpected disturbances during walking is often a necessity to continue the gait cycle. A perturbation might lead to a variety of balance responses. This depends on various factors, such as the magnitude and direction of the perturbation (Hof et al., 2010; Vlutters et al., 2016), the point of application of the perturbation to the body (Engelhart et al., 2015), the instance during the gait cycle at which the perturbation occurs (Eng et al., 1994; Cordero et al., 2003), and the objective the subject is trying to achieve (Todorov and Jordan, 2002; Potocanac et al., 2014). Often, foot placement adjustments are considered as a crucial strategy to recover balance during walking (MacKinnon and Winter, 1993; Patla, 2003). Adequate foot placement, however, does not guarantee balance; if the leg joints do not provide the appropriate moments, the body might move in an undesired direction, or the leg might collapse resulting in a fall. Indeed, faulty weight bearing was shown to be a major cause of falls in elderly residing in long-term care (Robinovitch et al., 2013). Providing balance support to people prone to falls, for example through a powered lower-extremity exoskeleton, therefore requires an understanding of which joints are involved in balance recovery strategies, and what the output of these joints should be.

In the field of humanoid robotics, too, there is an increasing demand for an understanding of human balance and perturbation recovery, as well as data to compare robot controller output to human balancing behavior. Data collected in humans can be used to synthesize robot motions, for example through inverse optimal control (Mombaur et al., 2010; Park and Levine, 2013), in which an underlying optimality criterion for a walking motion is identified based on collected data. This criterion can subsequently be used to generate optimal movement in new walking conditions. Another approach are neuromuscular walking models, which synthesize gait based on neural feedback loops from the muscles driving the leg joints (Geyer and Herr, 2010; Song and Geyer, 2015). Optimizing such models to mimic data collected in human perturbation recovery could provide insight in the underlying muscular and neural parameters involved in reactive balance, and help drive powered lower extremity assistive devices.

To gain a better understanding of balance during walking, several studies have focused on adjustments in step location and step duration following perturbations. The experimental results in (Hof et al., 2010; Vlutters et al., 2016) suggest that lateral stepping modulates linearly with the mediolateral (ML) COM velocity, as a result from ML pelvis perturbations. Furthermore, (Rankin et al., 2014) showed that hip abductor muscle activity in the swing leg correlated with the ML distance between the COM and the stance foot during unperturbed walking. These findings suggests some form of proportional swing leg control, but it is unclear whether the swing leg is simply accelerated stronger in the direction of the fall with increasing perturbation magnitude, or if the foot is actively positioned following a perturbation. While instruction on foot positioning can be provided through targets or stepping stones (Hoogkamer et al., 2015; Maeda et al., 2017) to make subjects explicitly control the location of their foot, such control does not have to apply when reacting to unexpected perturbations without a distinct foot placement objective. Furthermore, we previously found that wider steps as a result of ML perturbations (i.e. increased ML foot-foot distance) were generally steps of shorter length (i.e. decreased foot-foot anteroposterior distance), possibly as a consequence of the finite leg length (Vlutters et al., 2016). This indicates that disturbances in one plane can also influence control in another other plane.

Strategies other than foot placement adjustments are involved in stable walking as well (MacKinnon

and Winter, 1993; Winter, 1995). The importance of stance-leg ankle responses in maintaining balance after a trip has been shown to prevent the body from gaining excessive angular momentum (Pijnappels et al., 2005). Furthermore, we previously argued that ankle moment modulation in the stance leg might assist in counteracting AP perturbations, leading to reduced need for changes in foot placement (Vlutters et al., 2016). In the study, we found that applying anteroposterior (AP) perturbations at toe-off did not lead to significant changes in the position of the leading foot relative to the COM, at the subsequent heel strike. After foot contact, however, the center of pressure (COP) location did modulate with the perturbations, possibly indicating ankle involvement.

The purpose of this study is to gain a better understanding of how individual joints contribute to the recovery of balance during walking, following both ML and AP pelvis perturbations. Specifically, we investigate 1) how the joints of the stance leg are involved in restoring the walking cycle, 2) how leg swing is regulated following perturbations and whether the leg is actively positioned for foot contact, 3) to what extent joint modulation occurs in the plane perpendicular to the perturbation direction. We use inverse kinematics and inverse dynamics analysis to obtain joint angles, moments, and power of the ankle, knee, and hip joints, and provide muscle activity responses of various muscles acting on those joints. We investigate how these outcome parameters modulate with the applied perturbations. The presented results provide insight in human balance control during walking, and inspiration for the design and control of devices to support of human balance.

Methods

For the presented analysis we used data from 9 subjects collected in (Vlutters et al., 2016). The former also provides details regarding the experimental setup and protocol. A brief description is provided below. Both the setup and protocol were approved by the local ethics committee. All subjects gave written informed consent before participation, in accordance with the Declaration of Helsinki.

Apparatus

Subjects walked on a dual-belt instrumented treadmill (MotekForce Link, Culemborg, Netherlands), and received both ML and AP perturbations at the pelvis through a motor (Moog, Nieuw-Vennep, Netherlands). The motor could be connected to the subject's pelvis through a horizontal aluminum rod and a hip-brace (Distrac Wellcare, Hoegaarden, Belgium). Motor control was done over Ethernet (User Datagram Protocol) using xPC-target (MathWorks, Natick, MA, USA) with a dedicated network card (82558 Ethernet card, Intel, Santa Clara, CA, USA). The same xPC-target was used to collect 3 degrees-of-freedom ground reaction forces and moments of each of the two force plates in the treadmill, using a PCI-6229 AD card (National Instruments, Austin, TX, USA). A Bagnoli Desktop system (Delsys Inc, Natick, MA, USA) with an internal bandpass filter (20-450 Hz) was used to collect the bilateral EMG activity of the tibialis anterior (TA), gastrocnemius medialis (GM), rectus femoris (RF), biceps femoris (BF), gluteus medius (GLM), and adductor longus (AL) muscles. These EMG signals were digitized using the PCI card. All motor, force plate, and EMG signals were collected at 1 kHz. Finally, a 12-camera motion capture system (Visualeyez-II, Phoenix Technologies Inc, Burnaby, Canada) and LED clusters were used to collect kinematic data of the subject's feet, lower legs, upper legs, pelvis, trunk, and head. All kinematic data were collected at 100 Hz.

Protocol

Prior to the experiment, bony landmarks were captured using an LED based probe, relative to the LED clusters on each segment. Landmarks were the calcaneus, 1st and 5th metatarsal heads, medial and lateral malleoli, fibula head, medial and lateral epicondyles of the femur, greater trochanter, anterior and posterior superior iliac spines, xiphoid process, jugular notch, seventh cervical vertebra, occiput, head vertex and nasal sellion (Dumas et al., 2007).

Next, subjects were asked to walk on the treadmill with their arms crossed over the abdomen. Each subject walked four blocks of three trials each. In each block, the first trial was an unperturbed walking trial, while the second and third were perturbation trials. In two blocks only ML perturbations were applied, and in two blocks only AP perturbations were applied. For each direction (ML or AP), subjects walked one block at a slow speed (2.25 km/h) and one at a normal speed (4.5 km/h), with each speed scaled to the square root of the subject's leg length (Hof, 1996) to normalize to leg length. During perturbation trials, perturbation onset could randomly occur at the instance of TOR, detected using the vertical ground reaction force. Perturbations were implemented as 150 ms block pulses of force magnitude equal to 4, 8, 12, and 16% of the subject's body weight. For ML perturbations, the direction of the force could be inward (leftward for left stance leg) or outward (rightward for left stance leg). For AP perturbations, the directions were forward and backward. Each combination of perturbation direction, magnitude, and walking speed was repeated 8 times, yielding 256 perturbations per subject. When no perturbation was applied, the motors were admittance controlled such that the interaction force between subject and motor was as small as possible (reflected inertia < 1 kg).

Data processing

Data were processed using Matlab (R2016b, Mathworks) and OpenSim v3.3 (Delp et al., 2007). In Matlab, marker data were filtered using a 4th order 20 Hz zero-phase low-pass Butterworth filter before reconstructing the bony landmark positions through singular value decomposition (Söderkvist and Wedin, 1993). The same filter was applied to the ground reaction forces and moments, which were subsequently resampled to 100 Hz to match the marker data. Finally, the EMG recordings were filtered using a 1st order 48-52 Hz Butterworth bandstop filter, then detrended by subtracting the signal average, rectified, and filtered with a 2nd order 20 Hz low-pass Butterworth filter.

In OpenSim the default 23 degrees-of-freedom gait2354 model was used for the inverse kinematics and inverse dynamics. In this model the arms are assumed to be kept over the chest, and the head and trunk together make up one segment. For each subject, the model was scaled to subject dimensions using the landmark data. Joint angles were calculated using inverse kinematics, by matching the model markers to those in the recordings. The head marker data were not used in the inverse kinematics, as the head is part of the trunk segment in the model. The inverse kinematics results were filtered with OpenSim's default 6 Hz low-pass filter prior to the inverse dynamics analysis. In the inverse dynamics, the ground reaction forces and moments recorded using the left and right force plates were applied to the left and right calcaneus segments of the model, respectively. Furthermore, the measured force exerted from the motor on the subject was modeled as a horizontal force applied to the center of mass of the model's pelvis.

In Matlab, for each subject the joint angles from a static measurement were subtracted from all kinematic data, such that the subject's upright standing position corresponds to joint angles of 0 radian.

Joint power was calculated by multiplying the joint velocities with the joint moments. Joint moments and power were subsequently made dimensionless following (Hof, 1996). The EMG data of each muscle was scaled to the maximum activity of that muscle occurring within the median of the activity over all unperturbed gait cycles.

Next, gait phases were detected using landmarks on the feet, in line with (Zeni Jr et al., 2008). Using the instances of perturbation onset at TOR, perturbation end, and the gait events of toe-off left and right (TOL, TOR, respectively), and heel strike left and right (HSL, HSR, respectively), all data were cut into sequences, together spanning one gait cycle after perturbation onset. The duration of each of these sequences was determined, before resampling them to 50 samples each. For each subject, all sequences and their durations were sorted on perturbation magnitude and direction. These were then averaged over the 8 repetitions per condition, to obtain within-subject averages and standard deviations. In turn, these data were averaged over all subjects.

Results

We highlight specific subject-average joint-level and EMG responses, resulting from transient pelvis perturbations applied randomly at the instance of TOR during slow walking. We present slow walking data as it sometimes show more pronounced effects as compared to data from the normal walking speed, for example in the knee moments. This is possibly because the perturbations cause larger shifts in movement velocity relative to the intended walking speed. The complete set of results for both slow and normal walking speeds can be found in the supplementary materials.

All data is shown dimensionless, with joint angles presented in radians, and with joint moment scaled according to (Hof, 1996), and power scaled according to (Pinzone et al., 2016). Subject-average scaling factors for the presented data are $1215 \pm 259 \text{ Nm} (m \cdot g \cdot l_0)$ for moments and $525 \pm 123 \text{ W} (m \cdot g^{1/2} \cdot l_0^{3/2})$ for power, where m is the subject mass, g is the Earth's gravitational acceleration, and l_0 is the subject height. Finally, the joint moments and power are not shown for inward perturbations, as these often lead to cross-stepping in which subjects stood with both feet on the same force plate during the recovery. This yields incomplete data for the inverse dynamics.

Direct mechanical effects

The left leg is the stance leg during the perturbations. As the foot remains stationary on the treadmill belt and the body moves in the direction of the perturbations, direct mechanical effects of the perturbations are expected to occur in the stance leg joints. These effects are best reflected in the hip angles of the stance leg after the perturbations, and prior to HSR, see Figure 1 and 2.

As expected from a pure mechanical point of view, completion of outward perturbations makes the left hip move more toward abduction (Figure 1). Inward perturbations result in opposite effects, yielding more left hip adduction as compared to the unperturbed condition. Forward perturbations lead to little deviation in left hip flexion-extension angle compared to the unperturbed condition (Figure 2), possibly because the perturbation has the same direction as the desired movement. Backward perturbations, however, lead to increased left hip flexion, which can be partially attributed to pelvis tilt (Figure S13).

Figure 1. Subject-average angles, moments, and power for left and right hip abduction-adduction after ML perturbations. Data is shown for one gait cycle after perturbation onset at TOR, split into sequences. The horizontal length of each line in a sequence reflects the duration of that sequence relative to the other conditions. Angles are in radians. Moment and power are dimensionless, with subject average scaling factors of 1215 ± 259 Nm and 525 ± 123 W, respectively. Colors indicate the perturbation magnitudes. Yellow-red corresponds with outward perturbations, green-blue with inward perturbations. The gray shading is the subject-average standard deviation, only shown for the unperturbed condition to prevent image cluttering. In case of inward perturbations, joint moments and power are left out due to cross-stepping with both feet on the same force plate.

In addition, the right swing leg might also experience direct effects of the perturbations. For example, directly after completion of the perturbations the right hip joint first tends toward adduction following outward perturbations, and toward abduction following inward perturbations. This is likely caused by the body dynamics and swing leg inertia. Since this moves the leg opposite of what is required for foot placement in the direction of the fall, it indicates the necessity for active swing leg control.

Stance leg control during the ongoing step

The perturbations change the velocity of the body such that, movement deviates from the desired direction, with exception of the forward perturbations. Though foot placement adjustments could help redirect the body back toward the desired movement, the stance leg already intervenes before right heel strike occurs after a perturbation. This is reflected in the left hip joint, see Figure 1 and 2, as well as in the left knee joint, see Figure 3, and the left ankle joint, see Figure 4 and 5, all prior to HSR.

As the outward perturbations already induce left hip abduction angles, less left hip abduction moment is delivered as compared to the unperturbed condition (Figure 1). However, no adduction moment is generated in an attempt to oppose the perturbation-induced left hip abduction. Possible reasons are that not sufficient moment can be delivered to counter the sideways fall anyway, or because such moments could result in unwanted pelvis or trunk rotations. For the left ankle joint, outward perturbations yield a minor increase in eversion angle (Figure 4). The ankle moment also tends toward eversion, especially those of larger outward magnitude. This allows the COP to move more medially, in the direction of the fall. In contrast, inward perturbations lead to strong left ankle inversion angles, likely moving the COP laterally, also in the direction of the fall.

In the AP direction, the hip and ankle joints of the stance leg help oppose the AP perturbation effects. In response to forward perturbations, subjects delivered an increased left hip flexion moment (Figure 2). This slows down forward movement, and at the same time facilitates forward swing when the left leg transfers from stance to swing between HSR and TOL. These moments could furthermore assist in keeping the trunk upright, which will tend to flex backward after the pelvis is pushed forward. The left ankle delivers more plantar flexion moment (Figure 5), which also slows down forward movement as long as the foot is flat on the floor and does not pitch over the toes.

For backward perturbations the opposite occurs; subjects delivered more left hip extension moment, and negate the perturbation-induced left hip flexion (Figure 2). Meanwhile, the left ankle shows a strong reduced plantar flexion moment, or even a dorsiflexion moment (Figure 5). Dorsiflexing the stance foot helps the body fall forward, allowing it to pick up speed again.

Finally, subjects appear to maintain the unperturbed left knee angle following AP perturbations (Figure 3). They generated less extension moment or even a flexion moment to prevent the knee from over-extending following forward perturbations. Conversely, they generated more extension moment to prevent excessive knee flexion following backward perturbations. As for the ankle and hip joints, deviations in the knee from the unperturbed condition increase with increasing perturbation magnitude.

Figure 3. Subject-average angles, moments, and power for left and right knee flexion/extension after AP perturbations. Yellow-red corresponds with forward perturbations, green-blue with backward perturbations. Please refer to the caption and legend of figure 1.

Slow walking, ML perturbations Ankle inversion-eversion

Figure 4. Subject-average angles, moments, and power for left and right ankle inversion-eversion after ML perturbations. Yellow-red corresponds with outward perturbations, green-blue with inward perturbations. Please refer to the caption and legend of Figure 1.

Slow walking, AP perturbations Ankle plantarflexion-dorsiflexion

Figure 5. Subject-average angles, moments, and power for left and right ankle plantar/dorsiflexion after AP perturbations. Yellow-red corresponds with forward perturbations, green-blue with backward perturbations. Please refer to the caption and legend of Figure 1.

Swing leg control ensures foot positioning

Swinging the right leg to adjust foot placement at HSR must involve joint moment modulation, as not taking action in response to one of the perturbations will likely result in a fall. Subjects changed their swing leg control in response to the perturbations, as can be seen in the right hip joint, see Figure 1 and 2, as well as in the right knee joint, see Figure 3.

In the ML direction following outward perturbations, subjects delivered a right hip abduction moment to swing the leg outward and realize a more lateral foot placement (Figure 1). This moment increased in amplitude with increasing perturbation magnitude. However before HSR this abduction turns into an adduction moment, halting further right hip abduction. After HSR, the right leg becomes the new stance leg and subjects again delivered an abduction moment. This pushes the body back in the direction from which it came, and prevents it from laterally toppling over the leg.

Similar moment reversals can be observed in the AP direction. Subjects delivered more right hip flexion moment to facilitate forward leg swing with increasing forward perturbation magnitude (Figure 2), even though right hip flexion-extension angles initially do not deviate strongly from the unperturbed condition. Then, before HSR the moment is reversed into a right hip extension moment, which halts forward leg swing. The opposite occurs for backward perturbations; subjects initially delivered less right hip flexion moment as compared to the unperturbed condition, or even generate an extension moment. Subsequently, for all backward perturbations a hip extension moment is delivered prior to HSR to halt forward leg swing.

Finally, also the right knee shows such moment fluctuations as compared to the unperturbed condition (Figure 3). Following forward perturbations first less flexion moment is delivered, only to deliver more flexion moment slightly later within the same gait phase, before HSR. This might be related to movement of the upper leg, which can influence knee flexion-extension through inertial effects of the lower leg. Again, a similar but opposite response occurs for backward perturbations.

Ankle control after first heel contact

After HSR the right leg further contributes to the perturbation recovery. Here we highlight the right ankle joint, see Figure 4 and 5. Following outward perturbations, subjects delivered more right ankle inversion moment just before HSR, and continuing after HSR (Figure 4). Such moments shift the COP laterally in the direction of the perturbation, and help the body return toward the direction from which it came. Forward perturbations result in additional ankle plantar flexion moments between HSR and TOL to counteract the gained forward velocity (Figure 5). Prior to HSL, however, less right ankle plantar flexion moment is generated as compared to the unperturbed condition, possibly to generate less push-off. For backward perturbations no major changes occur directly after HSR, but some additional right ankle plantar flexion moment is generated between TOL and HSL, possibly to generate forward velocity through additional push-off.

Perturbations also evoke perpendicular responses

The perturbations not only affect movement in the direction in which they are applied, but also evoke responses perpendicular to the perturbation direction. These responses also scale with perturbation magnitude. This can be seen in the hip flexion-extension responses following ML perturbations, see Figure 6, but also in the ankle responses (Figures S1 and S9).

For the left stance leg during and directly after the perturbations, subjects delivered an increased left hip flexion moment following all inward perturbations. This flexion moment slows down forward progression, and might be in preparation for the decreased AP distance to the (right) leading foot at HSR that accompanies wider steps (Vlutters et al., 2016). This additional left hip flexion moment might help prevent toppling over the leading foot too quickly once it is placed at a shorter AP distance from the body.

As in the left leg, the right hip flexion-extension moments also modulate with ML perturbations. Subjects delivered an increased hip extension moment prior to HSR to slow forward leg swing. This helps move the foot toward the floor in preparation for an earlier HSR. Finally, for inward perturbations, increases in right hip flexion angle can be observed, likely required to move the right leg over the left leg when making a cross-step.

Muscle activity reveals spinal and supra-spinal mediated responses

All recorded muscles, in both legs, are affected by both the ML and the AP perturbations. Activity modulation mainly occurs before the HSR following perturbation onset. Responses often depend on the perturbation direction, and scale with its magnitude, see Figure 7 for ML perturbations and Figure 8 for AP perturbations. The earliest response onsets are in the order of 65-75 ms, such as those in the left gluteus medius (GLM) muscle following ML perturbations, starting just before the second half of the 150 ms perturbation window (TOL-P.End). Such latencies suggest a reflexive origin, most likely caused by perturbation-induced changes in muscle length and tendon force. Even muscles which are unlikely to experience direct mechanical effects of the perturbations show modulation with the perturbation magnitude, starting approximately 100 ms after perturbation onset. Examples are the right tibialis anterior (TA), rectus femoris (RF) and biceps femoris (BF) muscles following ML perturbations. Supra-spinal neural structures can be involved in these responses, given their latency (Petersen et al., 1998). Most often the activity increased, but the left GLM muscle and the left gastronemius medialis (GM) muscle show that activity decreases can be induced as well.

Various perturbations lead to co-contractions. The TA and GM, the RF and BF, and the GLM and Adductor longus (AL) muscle pairs in the left leg all show activity increases following larger inward perturbations. These suggest an increased impedance in all the stance leg joints, keeping the leg stable while the body laterally topples over it. Finally, in many cases the muscle activity corresponds with the moments from the inverse dynamics, such as the strong increase in the left TA muscle and the activity drop in the left GM muscle following backward perturbations, causing the observed dorsiflexion ankle moments.

Slow walking, ML perturbations Hip flexion-extension

Figure 6. Subject-average angles, moments, and power for left and right hip flexion/extension after ML perturbations. Yellow-red corresponds with forward perturbations, green-blue with backward perturbations. Please refer to the caption and legend of figure 1.

Figure 8. Subject-average EMG responses for various muscles of the left and right leg after AP perturbations. Yellow-red corresponds with forward perturbations, green-blue with backward perturbations. Please refer to the caption and legend of figure 1.

Discussion

We aimed to identify joint-level contributions to countering unexpected pelvis perturbations during walking, for joints in both the stance leg and the swing leg. Healthy human subjects were perturbed during walking at the instance of toe-off right. Inverse kinematics and inverse dynamics analysis were used to investigate joint-level responses to the perturbations. Shortly after a perturbation the joint angles and moments, as well as the activity levels in various leg muscles, often scaled with perturbation magnitude and direction. This occurred in multiple joint and muscles simultaneously. Not simply a single joint is affected, nor is a single joint primarily responsible for rejecting the perturbation-induced effects. Instead, a multitude of actions ensures a return to the unperturbed walking cycle. This data set reveals the complexity of the responses, and provides a unique, multi-joint insight in both the frontal and sagittal planes of balance control during walking.

Though foot placement adjustments are often considered the most important strategy to maintain balance during walking, it is complemented by an abundance of other recovery strategies, especially in the sagittal plane. This is demonstrated by the stance leg, which intervenes before the first heel strike occurs after the perturbations. Such actions reduce the need for foot placement adjustments. For example, the left hip extension moments paired with the dorsiflexion moments in the left ankle joint allow the body to pick up forward speed following backward perturbations. Other stance leg actions could include preventing the COM from losing height, for example by timely stance leg extension through ankle plantar flexion following forward perturbations. Preventing the COM from losing height could be beneficial as redirecting the COM back upward is costly (Donelan et al., 2001). Other recovery options, available after foot contact, might also reduce the need for foot placement adjustments. Given the dimensions of the human foot, ankle joint moments delivered after foot contact are suitable for further slowing down forward movement and rejecting forward perturbations. These actions might explain why we previously found little adjustment in foot placement location relative to the COM following AP perturbations (Vlutters et al., 2016).

The swing leg is actively positioned to realize foot placement adjustments, as indicated by the hip joint moments that halt the leg swing, as well as by the corresponding EMG activity. The leg is not simply swung in the direction of the fall until foot contact occurs due to the fall itself. It remains inconclusive, however, whether humans explicitly regulate the location and timing of foot placement, or operate in some other control space. While it is possible to control the foot location, for example when stepping towards targets (Hoogkamer et al., 2015; Maeda et al., 2017), accurate foot placement is not a prerequisite for balance, nor guarantees balance. It merely provides a window of opportunity for the leading leg to redirect body motion after foot contact. This might be made easier by putting the foot on the floor at a certain location and time. Rather than operating in a foot endpoint space to step toward a specific location, we perhaps learn coordinated muscle activation patterns that facilitate balance recovery. As a toddler, humans learn to balance and walk through trial and error, play, and practice, involving more than 2000 steps an hour and a fall every 3.5 minutes (Adolph et al., 2012). This could help us train coordinated muscle activation that minimizes variability in certain objectives related to balance control during walking (Todorov and Jordan, 2002; Dingwell et al., 2010). It seems however unlikely that these objectives must involve stepping toward specific locations. Instead, it is expected that reactive stepping is generally based on objectives related to whole-body balance, similar as has been suggested for strategies in standing balance (Horak et al., 1990; Savavynia and Ting, 2013).

The responses occurring in the plane perpendicular to that of the perturbation demonstrate that human balance control is a three-dimensional problem. Though a disturbance might occur only in the frontal plane or only in the sagittal plane, both planes remain linked in time as there is only a single instance of foot placement. Furthermore, a subject might have to compromise between the mediolateral and the anterior distance to the leading foot at foot contact due to the finite length of our legs. A hip abduction or adduction might therefore require a paired change in hip flexion or extension, for example to ensure that wider steps are also steps of shorter length (Vlutters et al., 2016). Consequently, findings in two-dimensional models of walking are not guaranteed to successfully transfer to the three-dimensional case. For example, in (Allen and Neptune, 2012) it was hypothesized that a neuromuscular controller previously used to mimic the sagittal-plane biomechanics of human walking could also drive a three-dimensional model. Though the experimental data-based controller indeed assisted in the ML direction, additional mechanisms were required to contribute to frontal plane balance. Simulations and prototype studies have furthermore shown that dynamic walking stability can be enhanced through motion coupling between different planes (Wisse, 2004; Wisse and Schwab, 2005).

The observed balance recovery responses can involve spinal and supra-spinal neural pathways. This can be deduced from the latencies observed in the muscle responses, as well as from the magnitude and direction dependency of the responses arising in multiple muscles simultaneously. The latter suggests that sensory integration is involved in generating the responses (Horak et al., 1990), therefore likely involving supra-spinal processing.

Many muscles show a functional change in activity within 100 ms after perturbation onset. For example, sudden increases in GLM activity following outward perturbations facilitate outward leg swing. Such activity is phase dependent, and not strictly dependent on the background activity (Hof and Duysens, 2013). In (Rankin et al., 2014) it was proposed that humans modulate the GLM activity of their swing leg based on the distance between the COM and the stance foot during unperturbed walking. Our outward perturbations lead to increases in this distance, and also lead to increases in GLM activity in the right leg during swing. However, theoretically, for a given distance between the COM and the stance foot there could be various COM velocities, potentially requiring different adjustments in lateral foot placement. It is therefore unlikely that lateral foot placement, and with it the GLM activity, depends solely on the distance between the COM and the stance foot, and also depend on the COM velocity (Hof, 2008; Vlutters et al., 2016). Furthermore, the ab- and adductor muscles also respond to AP perturbations, suggesting that there are additional influences from movement in the direction perpendicular to the perturbation direction.

Decreases in muscle activity, such as those in the GLM and GM muscles following various perturbations, might be caused by reciprocal inhibition or an unload response. The latter was previously shown in the soleus muscle following plantar flexion perturbations (Grey et al., 2007) or a sudden drop in ground support (Af Klint et al., 2009), both during walking. Such activity decreases occur after removing the load from a muscle, suggesting a reduction in the contribution of spinally mediated force-sensitive afferents to the muscle activation. In our results, however, the reduction in activity does not modulate with the perturbation magnitude, such that an unload response is less likely.

In future studies we will use these perturbed gait data to validate control principles in walking models that aim to replicate stable human-like walking. For example, the reflex-based walking models developed in (Geyer and Herr, 2010; Song and Geyer, 2015) might be optimized using the collected

data to not only have the muscles encode for locomotion, but also facilitate reactive balance recovery in response to large perturbations. In turn, such models can be used to drive devices that aim to provide balance and walking support, including orthoses such as lower extremity exoskeletons (Pratt et al., 2004; Neuhaus et al., 2011; Wang et al., 2015), support backpacks (Lemus et al., 2017), as well as (neuro)prostheses (van den Brand et al., 2012; Kim and Collins, 2015). Such models might furthermore be used to extrapolate and predict balance responses to disturbances not present in the data, as it is a cumbersome task to experimentally map all responses to the tremendous amount of different perturbations that could occur during gait. Nevertheless, experimental data remains of crucial importance to provide steps toward understanding and mimicking human-like balance control during walking.

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Supplementary material

Figure S1 – S7 : Slow walking, mediolateral perturbations

Figure S8 – S14 : Slow walking, anteroposterior perturbations

Figure S15 – S21 : Normal walking, mediolateral perturbations

Figure S22 – S28 : Normal walking, anteroposterior perturbations

For completion, all figures resulting from the inverse kinematics and inverse dynamics analysis are presented in the supplementary materials. In all figures, data is shown for one gait cycle after perturbation onset at TOR, split into sequences. Angles are in radians. Moment and power are presented dimensionless, with subject average scaling factors of 1215 ± 259 Nm and 2739 ± 562 W, respectively. Colors indicate the different perturbation magnitudes. Yellow-red corresponds with outward perturbations (ML) or forward perturbations (AP), green-blue with inward perturbations (ML) or backward perturbations (AP). The gray shading is the subject-average standard deviation, which is only shown for the unperturbed condition to prevent image cluttering. In case of inward perturbations, joint moments and power are left out due to cross-stepping, standing with both feet on the same force plate. The horizontal length of each line in a sequence reflects the duration of that sequence relative to the other conditions.

Fig. S1

Slow walking, ML perturbations Ankle plantarflexion-dorsiflexion

Fig. S2

Slow walking, ML perturbations Ankle inversion-eversion

Fig. S3

Fig. S4

Fig. S5

Slow walking, ML perturbations Hip abduction-adduction

Fig. S6

Fig. S7

Slow walking, ML perturbations
EMG

Fig. S8

Fig. S9

Slow walking, AP perturbations Ankle inversion-eversion

Fig. S10

Fig. S11

Slow walking, AP perturbations Hip flexion-extension

Fig. S12

Fig. S13

Fig. S14

Fig. S15

Normal walking, ML perturbations
Ankle plantarflexion-dorsiflexion

Fig. S16

Fig. S17

Normal walking, ML perturbations Knee flexion-extension

Fig. S18

Fig. S19

Normal walking, ML perturbations
Hip abduction-adduction

Fig. S20

Fig. S21

Fig. S22

Fig. S23

Fig. S24

Fig. S25

Normal walking, AP perturbations
Hip flexion-extension

Fig. S26

Normal walking, AP perturbations
Hip abduction-adduction

Fig. S27

Fig. S28

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CHAPTER 8

Ankle muscle responses during perturbed walking with blocked ankle joints

“It's not very effective...”

Under review as:

Ankle muscle responses during perturbed walking with blocked ankle joints

M. Vlutters, E.H.F. van Asseldonk, H. van der Kooij

Abstract

The ankle joint muscles can contribute to balance during walking by modulating the center of pressure and ground reaction forces through an ankle moment. This is especially effective in the sagittal plane through ankle plantar- or dorsiflexion. If the ankle joints would be physically blocked to make an ankle strategy ineffective, there would be no functional contribution of these muscles to balance during walking, nor would these muscles generate afferent output regarding ankle joint rotation. Consequently, ankle muscle activation for the purpose of balance control would be expected to disappear.

To investigate human balance control, we have performed an experiment in which subjects received anteroposterior pelvis perturbations during walking while the ankle joints could not contribute to the balance recovery. The latter was realized by physically blocking the ankle joints through a pair of modified ankle-foot orthoses. Here, we present the lower-limb muscle activity responses in reaction to these perturbations. Of particular interest are the tibialis anterior and gastrocnemius medialis muscles, as these could not contribute to the balance recovery through the ankle joint, nor encode muscle length changes caused by ankle joint rotation. Yet, these muscles showed long-latency responses approximately 100 ms after perturbation onset, even though the ankle joints were blocked. The response amplitudes were dependent on the perturbation magnitude and direction, as well as the state of the leg. The results suggest a centralized regulation of balance control involving supra-spinal neural structures, without the need for changes in ankle muscle proprioceptive information.

Introduction

Humans are able to stay upright and continue walking without continuously thinking about how to move. Even under challenging conditions involving unexpected perturbations, rapid and possibly unconscious reactions can occur to facilitate balance recovery. For example, in (Hof, 2013) it was shown that gluteus medius muscle activity changes can occur within 100-150 ms after lateral perturbation onset during walking. Such activity changes facilitate lateral foot placement adjustments, to negate the perturbation effects.

The neural origin of such balance responses remains inconclusive. A perturbation can cause muscle length and tendon force changes. These can trigger muscle spindles and Golgi tendon organs to generate afferent signals that can affect muscle activity. In the work of Hof and Duysens it was argued that the observed activity changes were involuntary but not caused by muscle stretch, given the absence of a short latency reflex. Effects of muscle stretch could not be ruled out, however, such that afferent information might be integrated on a supra-spinal level to contribute to the observed responses. Furthermore, another study has shown the existence of inter-limb pathways that might affect muscle activity. It was shown that externally applied knee extension in one limb can elicit activity increases in knee flexor muscles of the other limb (Stevenson et al., 2013). These responses, with a latency of 76 ms, likely involve trans-cortical pathways as shown through trans-cranial magnetic stimulation.

The vestibular system is also considered to play an important role in balance control. Approximately 75 ms after suddenly dropping subjects suspended above the floor ankle muscle responses occurred possibly in preparation for landing (Jones and Watt, 1971). In baboons, these responses disappeared after cutting off vestibular input (Lacour et al., 1978). During stance, responses in ankle muscles occurred 65-75 ms after onset of a forward or backward acceleration of the head (Horstmann and Dietz, 1988; Horak et al., 1994). Though these perturbations cannot prevent proprioceptive input from neck muscles, such responses were absent in patients with adult-onset vestibular loss. The vestibular system can furthermore influence ongoing muscle activity. In walking subjects it was shown that the ankle muscles receive leg-specific input from the vestibular system, depending on the velocity of that leg in a dual-belt walking condition (Forbes et al., 2017). Such vestibular contributions typically diminish if the investigated muscles provide no functional contribution to maintaining balance (Fitzpatrick et al., 1994; Luu et al., 2012).

The ankle joint muscles can contribute to balance during walking by generating ankle moments, which induce changes in the center of pressure (COP) and the ground reaction forces originating from that point (Gruben and Boehm, 2014). We have previously performed an experiment to investigate balance control during anteroposterior (AP) perturbed walking, while eliminating contributions of an ankle moment to the balance recovery (see Chapter 5). This was achieved by physically blocking the ankle joints and minimizing the base of support area using a pair of modified ankle-foot orthoses, to enable stilt-like walking (“pin-shoes”, see Figure 1). The pin-shoes blocked ankle movement, disrupting the relation between ankle muscle activation and modulation of the COP and ground reaction forces that exists during normal walking. Consequently, the ankle muscles had no functional contribution to balance through the ankle joint, nor were subject to length changes caused by ankle rotations. Because of this, contributions from the vestibular system to ongoing ankle muscle activity, as well as the contributions from bilateral ankle muscle afferents to muscle activity, are both expected to diminish. However, ankle muscle activity still modulated with both the magnitude and the direction of the

perturbations, as well as with the state of the leg, starting approximately 100 ms after perturbation onset. We discuss possible causes as well as the implications of these findings in relation to human balance control.

Methods

A brief description of the experimental setup and protocol are given below. More details regarding the experiment and data collection can be found in Chapter 5.

Apparatus

A motor (Moog, Nieuw-Vennep, Netherlands) was fixated to the rear of an instrumented treadmill (MotekForce Link, Culemborg, Netherlands) to apply anteroposterior (AP) pelvis perturbations to walking subjects. Subjects could be connected to the motor through an aluminum rod and a brace (Distrac Wellcare, Hoegaarden, Belgium). A pair of ankle-foot orthoses (Airselect Elite, Aircast Inc, Summit NJ, USA) were used to minimize ankle movement and the base of support of each foot, see Figure 1. The latter was achieved through a modification, by attaching an aluminum pin to the sole of the orthoses.

Protocol

The protocol was approved by the local ethics committee. Prior to the experiment all participants gave written informed consent, in accordance with the Declaration of Helsinki. Ten subjects (5 men, age 20 ± 1 years, weight 71 ± 7 kg, height 1.76 ± 0.08 m, mean $\pm$ s.d.) wore the pin-shoes and walked on the treadmill. Subjects were given time to adapt to the shoes until they could walk while keeping their arms crossed over the abdomen. The walking speed was $0.63 * \sqrt{l} \text{ ms}^{-1}$ (Hof, 1996), where $\sqrt{l}$ is the square root of the total leg length, including the additional height gained by wearing the pin-shoes.

Subjects received 150 ms forward (+) and backward (-) pelvis perturbations from the motor at the instance of toe-off right, with a random 6-12 second interval between subsequent perturbations. The perturbation magnitudes were 0.04, 0.08, 0.12, and 0.16 times the subject's body weight. Each perturbation condition was repeated 8 times. The in total 64 perturbations per subject were applied in a randomized order. The total time spent walking on the pin-shoes during the experiment, including the adaptation prior to the measurements, was approximately 30 minutes.

Bilateral surface EMG of the tibialis anterior (TA), gastrocnemius medialis (GM), rectus femoris (RF), biceps femoris (BF), gluteus medius (GLM), and adductor longus (AL) were collected at 1000 Hz (Bagnoli desktop EMG, Delsys, Natick, MA, USA). Sensor placement was according to the Seniam guidelines (Freriks and Hermens, 2000). Finally, a 9-camera motion captures system (Visualeyez II, Phoenix Technologies, Burnaby, Canada) was used to track the subject's leg kinematics at 100 Hz.

Figure 1. Schematic drawing of the pin-shoes: a pair of ankle-foot orthoses that block ankle joint rotation which were modified to strongly reduce the support area of the foot. The vertical pin can be positioned along the foot.

Data processing

To investigate potential rotation of the pin-shoes with respect to the lower legs, for each leg the Euclidean distance between the toe-tip of each pin-shoe and the caput fibula of the corresponding leg were computed from the motion capture data. This distance captures translation and rotation of each shoe with respect to the lower leg, and was expected to remain unchanged between perturbation conditions. The AL EMG of one subject was removed from analysis due to bad electrode contact in one of the legs during the experiment. All EMG data were detrended by subtracting the average, then filtered with a 48-52 Hz bandstop Butterworth filter. The data were then rectified, and subsequently filtered with a 4th order 20 Hz low-pass Butterworth filter. For each subject, the data were averaged over the repetitions and used to calculate a within-subject standard deviation. These were subsequently averaged over all subjects to obtain subject averages and subject-average standard deviations. To investigate the effects of repeated perturbation exposure on the observed responses, EMG averages were also calculated using only the first four and only the last four repetitions of each perturbation condition.

Results

We perturbed subjects in the anteroposterior direction during walking while the ankle joints were physically blocked by a pair of modified ankle-foot orthoses, and observed EMG activity in various leg muscles such as the tibialis anterior and gastrocnemius medialis. Though there is some movement of the orthoses with respect to the lower legs, possibly due to compression of the leg and the sole, the orthoses do not systematically move with the 150 ms perturbations, see Figure 2. The distance shown is on average similar across all perturbation conditions, and does not exceed the boundaries of the standard deviation in the unperturbed condition. It is therefore unlikely that the perturbations triggered systematic afferent signals associated with ankle joint rotation.

Figure 2. Distance between the caput fibula (CF) on the lower leg and the toe-tip of each corresponding pin-shoe during the perturbation window (0-0.15 s). This distance should remain approximately constant if the orthoses do not rotate or translate with respect to the corresponding leg. Colors indicate the various perturbations. Yellow-red: forward. Green-blue: backward. Gray shading corresponds to the subject standard deviation during the unperturbed condition. Standard deviations are not shown for the perturbed conditions to prevent image cluttering.

Nearly all recorded muscles show modulation with the perturbation magnitude and direction, most often starting approximately 100 ms after perturbation onset at toe-off right, see Figure 3. The modulation is furthermore leg-dependent, related to the state of the leg at perturbation onset. Surprisingly, even though the ankle joints are fixated and an ankle strategy is ineffective, the TA and GM muscle activity is modulated in relation to the magnitude and direction of the perturbations. Though the GM also spans the knee joint, and could potentially be affected by knee joint rotation, the TA only spans the blocked ankle joint. Still, the observed activity changes in TA and GM muscles are comparable to those observed during AP perturbed barefoot walking, collected in (Vlutters et al., 2016), see Figure 3.

These responses are functional in the scenario where the subject is not wearing the pin-shoes. For example in the left ankle, which corresponds to the stance leg during the perturbations, the GM muscle increasingly activates following forward perturbations of increasing magnitude. This would lead to plantar flexion of the foot, and would shift the center of pressure (COP) forward and help counter the effects of forward perturbations. In contrast, the TA muscle increasingly activates following backward perturbations of increasing magnitude. This would lead to dorsiflexion of the foot, and a backward shift of the COP to counteract the effects of backward perturbations. In the current pin-shoe condition, however, muscle activation does not result in COP shifts within the area of the foot.

Figure 3. Subject-average EMG activity of the bilateral tibialis anterior (TA), gastrocnemius medialis (GM), rectus femoris (RF), biceps femoris (BF), gluteus medius (GLM), and adductor longus (AL) muscles in response to the forward and backward perturbations during walking with pin-shoes. Top two rows show TA and GM activity of perturbed walking subjects without pin-shoes from (Vlutters et al., 2016) for comparison. Colors indicate the various perturbations. Yellow-red: forward. Green-blue: backward. Gray shading corresponds to the subject standard deviation during the unperturbed condition. Standard deviations are not shown for the perturbed conditions to prevent image cluttering.

Also for the right ankle, which corresponds to the swing leg during the perturbation, muscle activity changes occur that would be functional when the ankle is free, but not when wearing the pin-shoes. Increasing right-ankle TA activity following forward perturbations would facilitate dorsiflexion of the swing foot, and could normally assist in preventing the toes from scuffing over the floor during forward leg swing. For backward perturbations, activating the GM to facilitate plantar flexion of the right ankle would allow a toe-first landing of the foot. This is convenient for backward steps that occurred for the larger magnitude backward perturbations (see Chapter 5). This was not observed when walking without pin-shoes, where subjects were better able to continue the gait cycle following a backward perturbation. The other recorded muscles functionally contribute to the recovery and help keep the body upright. In the left stance leg the RF and BF muscles might prevent knee over-extension or collapse after the body above it is perturbed forward or backward, respectively. In the right swing leg, both these muscles appear to assist in leg swing. For example, following backward perturbations the BF assists in knee flexion, which is paired with hip extension with possible contributions from the GLM. This is followed by knee extension through the RF muscle, likely to place the foot on the floor. Only the AL in the left stance leg shows little modulation with any perturbation, possibly due to little involvement in recovery from perturbations in either direction.

Figure 4. Change in average peak EMG amplitude occurring for the first four and last four repetitions of each perturbation condition. Peak amplitude was determined in a 0.5 second window after perturbation onset. Triangles indicate subject averages and indicate the perturbation direction. Errorbars represent subject standard deviations. Colors indicate the various perturbations. Yellow-red: forward. Green-blue: backward.

Finally, the responses showed minor habituation effects. The average maximum amplitude in a 0.5 s window after perturbation onset decreased in the last four repetitions compared to the first four repetitions per perturbation condition, see Figure 4. These effects occurred in all recorded muscles, with an average decrease of approximately 8% over all muscles and all perturbation conditions. In the left and right TA muscle the average decrease over all perturbations was 3%. In the left and right GM muscle this was 3% and 9%, respectively.

Discussion

The perturbations applied in this study affect whole-body movement and could therefore trigger a variety of sensory systems. This resulted in activity changes in TA, GM, and various other leg muscles, occurring approximately 100 ms after perturbation onset. With the ankle-joints fixated by the pin-shoes it is unlikely that the perturbations evoke TA and GM muscles proprioceptive responses. This is further supported by the absence of short latency (~50 ms) responses in these muscles, such that a pure spinally mediated response is unlikely. Hence, afferent information from TA and GM proprioception is unlikely to contribute to the observed muscle responses.

Because of the pin-shoes, the TA and GM muscles cannot functionally contribute to balancing the body through the ankle joints. The vestibular contributions to the ongoing activity of these muscles is therefore expected to be lower than in walking without the pin-shoes. In (Fitzpatrick et al., 1994; Luu et al., 2012) electrical vestibular stimulation was used to show that vestibular contributions to ongoing muscle activity diminish when those muscles do not have functional contributions to maintaining balance. This can even occur unconsciously, as was shown in (Luu et al., 2012) where the postural control of subjects was taken over using a computer-controlled balance mechanism unknown to the subjects. It is however unclear whether this reduced input also applies to sudden whole-body perturbations.

A mechanism that could explain the observed responses are “automated postural adjustments”. In (Nashner and Cordo, 1981) it was demonstrated that perturbations during standing can elicit such adjustments, which were found to be distinct from reaction-time task movements. These automated responses are independent of the subject instruction, have a higher priority in execution than reaction-time task movements when balance is threatened, and have no strict relation to the joint kinematics as is the case with a stretch reflex. Even healthy subjects with disrupted somatosensory feedback from the feet through blocked blood flow at the ankles, as well as subjects with bilateral vestibular loss, show such postural adjustments (Horak et al., 1990). These responses altered depending on which sensory information was disrupted. This lead to the suggestion that the available sensory information is used to select a set of balance strategies prior to a disturbance, whereas the implementation of a specific strategy is dependent on afferent information signaling the effects of the magnitude and direction of a disturbance on the body.

With regard to the prior selection of a set of strategies, there might be no urgent need to make adjustments based on the perceived sensory information during our experiment. There is no sensory deprivation that would limit the strategy selection as in (Horak et al., 1990). Consequently, an ankle muscle response might still be implemented, despite being ineffective. Prominent frameworks of human movement control, such as optimal feedback control (Todorov and Jordan, 2002), suggest that sensory feedback is used to correct for deviations that interfere with task-related goals, in this case

related to balance. In (Horak and Nashner, 1986) for example, an ankle strategy disappeared in response to support surface perturbations when the support surface size was decreased. Rotating the ankle joints would likely lead to an inappropriate response, and might interfere with maintaining balance. However in our case there are no major adverse effects for balance by contracting the ankle muscles, perhaps only in terms of energy expenditure. Consequently, there might be little immediate need to adjust the selection of balance strategies. More time might be required for the responses to fade than the 30 minutes of (un)perturbed pin-shoe walking during the experiment.

With regard to the implementation of a specific strategy, the results suggest that this originates from more central information about the magnitude and direction of the perturbations. Since the ankle muscle responses are unlikely to arise from changes in ankle muscle afferent information, they must have a different origin. Specifically, a task-level error related to balance rather than a joint-level error might trigger the responses, such as proposed in (Safavynia and Ting, 2013). In the study, sequential perturbations were used to generate uncorrelated center of mass and joint kinematic trajectories during standing. Both trajectories were subsequently used to predict TA and GM muscle activity patterns resulting from the perturbations. This worked best using delayed (~100 ms) feedback of the center of mass kinematics, while the predictions based on joint kinematics yielded poorer results and physiologically impossible delays (~10 ms). This implies that multiple muscles can be driven by a common, centralized signal.

To conclude, we believe the observed responses to be reminiscent of the “automated postural adjustments” observed in (Nashner and Cordo, 1981). The origin of such responses remains inconclusive. Given the response latency, involvement of sub-cortical (Stapley and Drew, 2009), or even cortical (Petersen et al., 1998) pathways is possible, and a pure spinal mediated response is unlikely. The responses can involve input from a range of sensors, though vestibular and somatosensory contributions are not strictly required (Horak et al., 1990). The presented results show that changes in ankle muscle proprioceptive information are not strictly required either. Instead, the results support a more centralized, task-level regulation of balance control involving supra-spinal neural structures. The occurrence of these responses despite the lack of a functional contribution indicate how deeply balance is embedded in the control of our movements.

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CHAPTER 9

Costs for leg swing and step transition can explain basic foot placement modulation

“Show me ya moves!”

In preparation for submission as:

**Costs for leg swing and step transition can explain
basic foot placement modulation in perturbed walking**

M. Vlutters, E.H.F. Van Asseldonk, H. Van der Kooij

Abstract

There are many possible combinations in step location and step time from which one can choose when making a recovery step in response to a perturbation. Still, humans modulate their foot placement fairly consistent when a specific perturbation is given, provided there is sufficient time to make such adjustments. Preferences in recovery steps might arise from energetic costs related to walking. Two prominent costs that help explain human walking are the step-to-step transition cost to redirect the center of mass velocity during the transfer from the trailing leg to the leading leg following foot contact, and a leg swing cost to swing the leg in the desired direction. In this study we investigate whether these costs allow a simple walking model to replicate human-like step location and step timing when making balance recovery steps in response to perturbations.

Both costs were implemented in a linear inverted pendulum model of walking. The model was made to converge to a steady-state gait over 100 steps before perturbing it with various force magnitudes in different directions. Despite its simplicity, the model showed foot placement location and time modulation in line with perturbed human walking data from a previous experimental study. The results therefore suggests that step-to-step transition and leg swing costs not only play a role in steady-state walking, but could also provide directions for understanding balance recovery and control during walking.

Introduction

When foot placement adjustment are used to recover balance during bipedal walking, there exists a large range of possible combinations of step location and step time to choose from. Multiple of these combinations might allow for a successful balance recovery. Yet, experiments in human walking have shown that the foot placement location and timing are often modulated in a similar and consistent way among different subjects, when a specific perturbation is presented at a certain point in the gait cycle (Oddsson et al., 2004; Hof et al., 2007; Hof et al., 2010; Vlutters et al., 2016), (Chapter 3-5). This suggests that humans often prefer one way of recovery over another. If these preferred strategies could be predicted, it could aid in a more natural and human-like control of bipedal robots, as well as lower extremity powered orthoses that aim to assist the user in maintaining balance during walking.

Human behavior is often considered to arise from some form of cost minimization. Both walking and running gaits can occur in a simple inverted pendulum model when the work performed by the telescopic legs on the center of mass is minimized to achieve a given speed and step length (Srinivasan and Ruina, 2006). These two parameters also determine the step time, as a given speed can only be obtained through traveling a given distance in a certain time. Humans furthermore select a step frequency that minimizes metabolic expenditure for a given walking speed (Zarrugh et al., 1974), and are able to minimize this in an online fashion (Selinger et al., 2015). Various costs can contribute to the total metabolic expenditure that potentially governs such behavior.

One cost is related to the transition from one single support phase to the next (Donelan et al., 2001). Every step the body's center of mass (COM) has to be redirected upward when transferring it from the trailing leg to the leading leg. Inverted pendulum models of walking predict that this step-to-step transition cost increases with increasing step length and width, mainly due to the increased changes in vertical COM velocity before and after foot contact (Donelan et al., 2002). This collision might be reduced by generating push-off prior to foot contact (Kuo, 2002; Kuo et al., 2005), which in turn also costs energy. A step-to-step transition cost can explain the increase in human metabolic expenditure with increasing step size (Donelan et al., 2001), but it does not explain a preferred foot placement location. This is because shorter steps have lower collision losses, such that steps of infinitesimal small length and duration would be the most efficient strategy.

Another cost can be associated with swinging the leg (Kuo, 2001; Doke et al., 2005; Doke and Kuo, 2007). When the desired movement is different from the swing that results from the leg's natural dynamics, the hip joint muscles must act on the leg to accelerate it in the desired direction, and decelerate it to position the foot. Unlike the step-to-step transition cost, the cost of leg swing increases with decreasing swing duration when swinging above the leg's natural dynamics (Doke et al., 2005). This could function as a trade-off against the very rapid stepping that would be preferred by the step-to-step transition cost. If leg swing would be the only cost, the optimal step would be governed by the leg's natural dynamics.

It is unclear to what extent energetic costs are considered in recovery steps in response to balance perturbations. There is presumably insufficient time to determine an optimal recovery motion while the recovery takes place. However, humans gain large amounts of experience with stepping and falling during early childhood (Adolph et al., 2012), which could lead to preferred and energetically favorable stepping responses. The purpose of this study is to investigate whether costs for leg swing and a cost for step-to-step transitions can explain the overall simultaneous modulation in foot placement location

and timing, similar to those observed in perturbed human subjects. We apply these costs to a simple linear inverted pendulum model (LIPM) of walking, under a constraint that the foot must be placed at a fixed distance from the extrapolated center of mass (XCOM). The latter is a COM velocity-dependent concept derived from the LIPM, which has shown to have predictive value in foot placement and center of pressure (COP) modulation in healthy subjects (Hof et al., 2010; Vlutters et al., 2016).

Methods

Linear inverted pendulum model

To mimic the single support phase of walking we use a ballistic LIPM such as in (Kajita and Tani, 1991), see Figure 1. The double support phase is ignored by instantaneously interchanging the model's telescopic legs at heel strike. One advantage of this model with respect to a non-linear inverted pendulum is that the foot placement location is not limited by the state of the COM with respect to the stance foot. If the legs would be of fixed equal length, the leading leg angle with the vertical must always be the same as that of the trailing leg angle with the vertical to realize foot contact. Furthermore, the relative simple equations of motion of the LIPM have a closed-form solution such that integration is not required for simulations. The equations of motion of the 3D LIPM are given by:

$$\omega_0^2 (\mathbf{r}_{COM} - \mathbf{r}_{COP}) = \ddot{\mathbf{r}}_{COM} \quad [1]$$

Figure 1. The 3D linear inverted pendulum model with swing leg. (A) Sagittal plane view, in which the leg swings from initial angle θ_{ix} to a final angle θ_{fx} in t_f seconds following a cosine function. The same applies to the frontal plane. Interaction dynamics between the pendulum and the swing leg are ignored for simplicity. (B) Representation of the center of mass ($\mathbf{r}_{COM}$), center of pressure ($\mathbf{r}_{COP}$) and XCOM ($\mathbf{r}_{XCOM}$) locations in the horizontal plane. The foot placement location is made state-dependent by setting the final swing leg angle corresponding to a point at a fixed distance $\mathbf{B} = [B_x, B_y]$ from $\mathbf{r}_{XCOM}$. The shortest angle between the legs is α .

in which $\mathbf{r}_{COM}$ and $\mathbf{r}_{COP}$ are each two-element column vectors containing the x and y coordinates of the COM and the COP in the horizontal plane, respectively, and the double-dot indicates the second derivative with respect to time. The term $\omega_0 = \sqrt{g/l}$ is a constant, with g the Earth's gravitational acceleration and l the fixed COM height as a result of the linearization. Note that this results in independent motions in the frontal (x) and sagittal (y) planes. For as long as $\mathbf{r}_{COP}$ is constant during the pendulum swing, the solutions to equation [1] are, cf. (Townsend, 1985):

$$\mathbf{r}_{COM f} = \mathbf{r}_{COP} + (\mathbf{r}_{COM i} - \mathbf{r}_{COP}) \cosh(\omega_0 t) + \frac{\dot{\mathbf{r}}_{COM i}}{\omega_0} \sinh(\omega_0 t) \quad [2]$$

$$\dot{\mathbf{r}}_{COM f} = (\mathbf{r}_{COM i} - \mathbf{r}_{COP}) \omega_0 \sinh(\omega_0 t) + \dot{\mathbf{r}}_{COM i} \cosh(\omega_0 t) \quad [3]$$

in which $\mathbf{r}_{COM i}$ and $\mathbf{r}_{COM f}$ correspond with the initial and final COM state at times $t = 0$ and $t = t_p$ respectively.

Perturbations

We will attempt to mimic the mediolateral (ML,x) and anteroposterior (AP,y) pelvis perturbations from (Vlutters et al., 2016), applied to walking human subjects. These perturbations were block pulses with a duration t_p and a constant force magnitude equal to a fraction p of the subject's body weight. All subjects therefore experienced approximately the same acceleration $p * g$, regardless of their mass. Though the perturbation device in (Vlutters et al., 2016) could not exactly track the input block pulses at every point in time, the time integral of the perturbations was close to the desired value. The net velocity change induced by an acceleration $p * g$ over t_p seconds is then modeled as:

$$\begin{aligned} \dot{x}_{COM}^+ &= \dot{x}_{COM}^- + t_p p g \\ \dot{y}_{COM}^+ &= \dot{y}_{COM}^- + t_p p g \end{aligned} \quad [4]$$

where the “-” and “+” indicate the COM velocity before and after the perturbation, respectively, t_p was the perturbation duration of 0.15 s, and $p = \{\pm 0.04, \pm 0.08, \pm 0.12, \pm 0.16\}$ is a fraction of g . For ML perturbations, negative p are referred to as "inward" (leftward for left stance) and positive p as "outward" (rightward for left stance) perturbations. For AP perturbations, positive corresponds to forward and negative to backward.

Extrapolated center of mass

A concept named the extrapolated center of mass (Hof et al., 2005) or capture point (Pratt et al., 2006) can be derived from equation [1], for which holds:

$$\mathbf{r}_{XCOM} - \mathbf{r}_{COM} = \frac{\dot{\mathbf{r}}_{COM}}{\omega_0} \quad [5]$$

in which $\mathbf{r}_{XCOM}$ contains the x and y coordinates of the XCOM in the horizontal plane. For the LIPM, if $\mathbf{r}_{COP} = \mathbf{r}_{XCOM}$, then $\mathbf{r}_{COM}$ converges toward $\mathbf{r}_{COP}$ in infinite time. As a result, the pendulum comes to an (approximate) upright stop. We introduce $\mathbf{B} = [B_x, B_y]$ as an offset from $\mathbf{r}_{XCOM}$, see Figure 1B. For normal walking it is preferred that the COM topples over the COP in the sagittal plane. This can be achieved by moving y_{COP} to a location for which $B_y < 0$. In the frontal plane it is desired that the COM returns toward the opposite direction, to facilitate lateral transfer to the other leg. This return movement can be achieved by moving x_{COP} to a location for which $B_x > 0$.

Swing leg model

To model a swing leg we use a method similar as in (Doke et al., 2005). The leg is represented as a single hanging pendulum that swings following a cosine waveform. The knee is assumed straight. The pendulum can make an angular displacement $\Theta = [\Theta_x, \Theta_y]$ with the vertical, rotating about the y and x axes respectively. Given telescopic legs, the leg swing can be treated decoupled between both axes. For simplicity floor scuffing is ignored, and the swing leg dynamics are treated independently of the LIPM dynamics such that they do not affect each others motion. The swing leg's equations of motion are given by:

$$\mathbf{M}_H = m_s l_s^2 \ddot{\Theta} + m_s g l_s \sin(\Theta) \quad [6]$$

In which m_s is the swing leg mass, and l_s is the distance of m_s from the hip joint along the leg. Any imposed angular displacement requires moments $\mathbf{M}_H = [M_x, M_y]$ about the x and y axes of the hip, respectively. We make the leg angle change from toe-off at $t = 0$ to heel strike at $t = t_f$ using a cosine wave, following:

$$\Theta = \Theta_i + A - A \cos(2\pi f t) \quad [7]$$

in which frequency f is given by:

$$f = \frac{1}{2t_f} \quad [8]$$

and amplitude A by:

$$A = \frac{\Theta_f - \Theta_i}{2} \quad [9]$$

in which Θ_i and Θ_f correspond to the initial (toe-off) and final (heel strike) leg angles, respectively, and t_f is the total swing time from toe-off to heel strike. By choosing f and A this way, the angular velocity at toe-off and heel strike is zero, while making the leg traverse from Θ_i to Θ_f in t_f seconds. When the LIPM model walks, Θ_i at toe-off is set to the last known angle between the previous stance leg and the vertical.

Cost of leg swing

By substituting equation [7] and its second derivative to time into equation [6], we get an expression for M_H as a function of Θ_p , Θ_j and t . Given these parameters, we define the cost of leg swing as the sum of the absolute required hip torque:

$$E_s = \int_{t=0}^{t_f} |M_H| dt \quad [10]$$

This assumes that any motion other than the natural dynamics of the legs takes effort. The cost is not normalized to the walking distance traveled, as this would emphasize making long and wide steps. We do not consider this to be an objective in recovery steps. For various other swing costs such as work or power the model converged to a gait that was not considered reasonable. Work tends to put a low penalty on small leg swing angles, making the model converge to a gait with very small, rapid steps. A similar effect was reported in (Kuo, 2001). Power puts a low penalty on slow movements, leading to long swing times, large steps, and issues converging toward a symmetric gait depending on the initial swing leg angles. Peak moment as a cost yields a symmetric gait similar as with equation [10], but makes the cost less sensitive to changes in the duration of the swing. This leads to less time modulation in response to disturbances.

Cost of step-to-step transition

Incorporating only cost [10] would make the model rely maximally on the swing leg's natural dynamics. This would not allow the model to make rapid compensatory steps above the natural swing frequency. To compensate, we incorporate a step-to-step transition cost similar as in (Donelan et al., 2002), where the energy lost in the transition from one leg to the next is dependent on the COM velocity before and after foot contact. This energy loss increases with step size, and has to be compensated to continue walking. However, it is difficult to determine whether or not a loss in horizontal velocity has unfavorable effects. Such transition losses might also aid in perturbation rejection, depending on the intended walking direction. Therefore only the vertical COM velocity is taken into account for this cost, as the COM must always be redirected back upward in the transition. As the LIPM has no vertical velocity component due to its linearization, we assume that the vertical velocity $\dot{Z}_{COM}$ is related to the stance leg orientation. We calculate $\dot{Z}_{COM}^+$ such that the resultant of $\dot{Z}_{COM}^+$ and the horizontal velocity components would be perpendicular to the stance leg, see Figure 2. For the transition of the COM to the leading leg after foot contact, we assume an instantaneous inelastic collision which reduces the velocity magnitude depending on the angle between the leading and trailing leg (Donelan et al., 2002):

$$|\dot{Z}_{COM}^+| = |\dot{Z}_{COM}^-| \cos(\alpha) \quad [11]$$

where $\dot{Z}_{COM}^-$ and $\dot{Z}_{COM}^+$ are the vertical COM velocity before and after impact, respectively, and α is the shortest angle between the legs. The energy that would be lost in the impact is then given by:

$$E_{sts} = 0.5 m (|\dot{Z}_{COM}^-| - |\dot{Z}_{COM}^+|)^2 \quad [12]$$

where m is the total mass of the pendulum plus swing leg. We will assume that the energy that would be lost in the step-to-step transition is instantly and fully compensated by some input that costs an equal amount of energy. Consequently, the velocity of the pendulum does not change in the transition to the next step. The model is furthermore not allowed to place a foot on the same side of the COM as the other foot, as this would lead to transition failure. The step-to-step transition cost is set to infinite for any potential step where the COM would not be between both feet.

Finding the swing time

Consider some desired final swing leg angles Θ_f . We look for the time t_f required to reach Θ_f from Θ_i such that the sum of the swing costs and the step-to-step transition cost is minimal over a finite discrete-time prediction horizon. As equations [2] and [3] allow easy calculation of the pendulum state at any future time, we simply compute the net cost for every possible time instance in the prediction horizon beyond $t = 0$, and find the time of heel strike corresponding to the minimum net cost:

$$t_f = \min_t [k_{sx} E_{sx} + k_{sy} E_{sy} + k_{sts} E_{sts}] \quad [13]$$

where the k represent gains of the individual costs. Note that the relative weighting of these gains determines the cost landscape. We then apply swing time t_f to the pendulum dynamics, place the foot at the location corresponding to Θ_f and repeat the procedure for the next step starting again at $t = 0$. For any successful transition to the leading leg, E_{sts} is minimal for as small as possible steps, preferably placing the foot continuously beneath the COM. Cost E_{sts} is therefore strictly increasing with increasing step length and width. Swing costs E_{sx} and E_{sy} are minimal when the swing leg can reach Θ_f through natural swing leg dynamics only. This is not always possible, however, depending on Θ_f . If the swing foot needs to "catch up" with some desired stepping location through a swing beyond the natural frequency, there is a trade-off between the angle over which the leg has to swing and the time within which this swing has to occur.

Figure 2. (A) Calculation of the vertical velocity component based on the horizontal velocity. (B) The vertical velocity components before (-) and after (+) foot contact are assumed to be related through leg angle α

Setting the final swing leg angle

In general, the model's COP modulation following foot placement is assumed to alter in line with the XCOM concept. That is, given the position $\mathbf{r}_{XCOM} + \mathbf{B}$ at heel strike, the COP is moved to that location at the subsequent toe-off (Vlutters et al., 2016). Since the model has a foot of infinitesimal size, and because the instances of heel strike and toe-off coincide due to a lack of a double support phase, the foot location always coincides with that of the COP. Hence, this COP modulation can be realized in the LIPM by simply setting $\boldsymbol{\Theta}_f$ to match a foot placement location at $\mathbf{r}_{XCOM} + \mathbf{B}$. This is similar to constant offset control proposed in (Hof, 2008). However a major difference is that we do not predefine any (fixed) single support duration, but instead determine the time of stepping based on equation [13].

It is important to note that the LIPM can only adjust the COP location and affect COM motion through foot placement adjustments. This is not the case in humans, who can also utilize other strategies. For example, we have previously shown that humans do not significantly modulate the location of the foot relative to the COM in response to AP perturbations, but do modulate the forthcoming COP location through other mechanisms (Vlutters et al., 2016), such as ankle moments. The model cannot replicate this. The model's foot placement is therefore expected to provide a better match to human data when foot placement modulation was the predominant strategy to realize COP modulation in the data.

Simulations

Simulations ran in discrete time with a fixed time step of 10^{-3} s. A prediction horizon of 1 s was used over which the net cost was computed. This horizon is assumed sufficiently long to cover one single support phase. An overview of default model parameter settings is given in Table 1. Parameters $\mathbf{B}$ were chosen according to the offsets of the COP from the XCOM following ML and AP perturbations found in (Vlutters et al., 2016), based on linear fits to the experimental data for slow walking (0.625 m s^{-1}). The model starts upright while standing on the left leg with $\mathbf{r}_{COM} = \mathbf{r}_{COP}$, $\dot{\mathbf{x}}_{COM} = 0.028 \text{ m s}^{-1}$, and $\dot{\mathbf{y}}_{COM} = 0.625 \text{ m s}^{-1}$. The initial swing leg angles $\boldsymbol{\Theta}_i$ are 0 at the start of the first step.

We let the model make 100 steps and assume that this is sufficient to make it converge toward a symmetric steady-state gait. Under this assumption, the steady-state gait parameters are independent of the initial conditions, and only depend on $\mathbf{B}$ and the cost that determines t_f . The ML and AP perturbations were applied at toe-off right, i.e. at the start of the left single support phase, following step 100. The resulting step locations and durations were analyzed and compared to human data from (Vlutters et al., 2016). In addition to the simulations with the default settings from Table 1, for ML perturbations gains k_{sx} and k_{sy} were numerically optimized through steepest descent to best match the model foot placement to the subject-average foot placement locations. Because the relative distribution of the gains affect how the model behaves, only k_{sx} and k_{sy} were optimized while the k_{sts} was left at 1. Furthermore, for ML perturbations the effect of the individual cost gains on the stepping responses was investigated by putting one or two of the three gains to 0.

Optimization was not done for the AP perturbations, because we already know that the human foot location does not change with the XCOM for these perturbations, while the model is constrained to do so. Therefore, to better match the foot placement locations for AP perturbations, the AP distance between the COM and the leading foot position was fixed to that of the steady state gait, ignoring the XCOM constraint after perturbation onset. The resulting model swing times were compared to the human data.

Results

Steady-state gait

The model requires only a few steps to converge toward a steady state gait. Typically, within 3-5 steps the changes in step length and width are less than 10^{-3} m, depending on the discrepancy between the initial and steady state COM velocity. The model's COM motion and foot locations for steady state gait and in response to the various ML perturbations are shown in Figure 3, using the model parameters from Table 1 for the simulations. For the 100-step steady state gait the step length was 0.485 m, step width 0.074 m, step time 0.485 s, and the average forward velocity 1.005 m s^{-1} . The resulting step size and forward speed are higher than in the human data from which $\mathbf{B}$ was derived. The sensitivity of the steady-state gait to changes in $\mathbf{B}$ and the k can be found in Figure S1. After optimization of the foot placement locations to those of the human data following ML perturbations, the 100-step steady state gait step length was 0.367 m, step width 0.062 m, step time 0.416 s, and the average forward velocity 0.880 m s^{-1} . A comparison with the human unperturbed walking data can be found in Figure 4. The gait velocity is still higher than in the human data, likely caused by differences in dynamics between the model and human gait. Humans have some control over the single support phase (Chapter 7) and have a foot roll-over, whereas the model does not.

Table 1: Default simulation parameter settings

Parameter	Value	Description
B_x	0.0132	Mediolateral COP offset from the XCOM, in m (Vlutters et al., 2016)
B_y	-0.1364	Anteroposterior COP offset from the XCOM, in m (Vlutters et al., 2016)
g	9.81	Earth gravitational acceleration, in m s^{-2}
k_{sx}	1	Gain for mediolateral leg swing cost
k_{sy}	1	Gain for anteroposterior leg swing cost
k_{sts}	1	Gain for step-to-step transition cost
l	1	COM height, in m
l_s	$0.447 * l$	Distance swing leg mass from hip joint assuming a straight knee, in m (Winter, 2009)
m	80	Total mass (pendulum + swing leg), in kg
m_i	$0.161 * m$	Swing leg mass, in kg (Winter, 2009)

Figure 3. Steady-state gait and perturbations. The model walked 100 steps to achieve a steady-state gait before it was perturbed. Model parameter settings were as in Table 1. During unperturbed gait, red and blue correspond to single support right and left, respectively. The colors during step 101 indicate the various perturbation magnitudes.

Contributions of individual costs

The contributions of the individual cost components to the steady-state gait and the perturbation recovery can be investigated by settings one or multiple k to zero, while leaving the other parameters at their default settings. Here the effects are shown for the recovery in response to ML perturbations.

First, the model cannot walk if stepping is fully governed by the step-to-step transition cost, while the swing cost gains are zero ($k_{sx}=0$, $k_{sy}=0$, $k_{sts}=1$). The swing cost gains were therefore set to a very small value ($k_{sx}=10^{-8}$, $k_{sy}=10^{-8}$, $k_{sts}=1$), resulting in very short steps of a short duration, see Figure 4AB. Some minor sensitivity to the ML perturbations remains, since k_{sx} and k_{sy} are not completely zero. Second, when stepping is only determined by the swing cost in the sagittal plane ($k_{sx}=0$, $k_{sy}=1$, $k_{sts}=0$), the model loses all coupling between the frontal and sagittal planes (Figure 4CD). The ML perturbations do not interfere with the sagittal plane swing cost, such that there are no time adjustment, nor any adjustment in the AP foot placement location. Note that the leg does adjust in the frontal plane because foot placement is constrained to a fixed distance from the XCOM, which is dependent on the ML COM velocity, and therefore affected by the ML perturbations. Third, when the frontal plane swing cost is the only cost ($k_{sx}=1$, $k_{sy}=0$, $k_{sts}=0$), the lateral movement fully determines the model behavior. Since the model is constrained to step at some distance B from the XCOM, it will always fall to the side and cannot walk in a perfect straight line. A longer single support duration therefore leads to a larger lateral velocity, which is associated with wider steps, and therefore a higher frontal plane swing cost. For steady-state gait the model therefore converges toward a gait with small steps of short duration, to minimize the step width and the lateral swing cost while sufficing the XCOM constraint. In the perturbed conditions, the -0.04 inward perturbation nearly compensates the steady-state lateral velocity, making it almost zero. This removes the need to step, leading to a very long and slow step (Figure 4EF). None of the individual costs can explain the human behavior, and a combination is required. For combinations with two of the three costs active see Figures S2 and S3.

Combined costs: recovery from ML perturbations

In response to perturbations the model generally places the foot in the direction of the fall under the default model parameter settings (Table 1). This might be expected given that the steps are based on the XCOM, which in turn is dependent on the horizontal COM velocity vector. However, even though foot placement is constrained to a fixed distance from the XCOM, the way both the location and timing of foot placement modulate with the various perturbations depends on the combined implemented costs.

For ML perturbations both model and human experimental data show comparable modulation patterns, see Figure 5AB. The model's step length tends to become shorter for wider steps, as is the case in the experimental data. Furthermore, as in the human data, the model's single support duration decreases with increasing outward perturbation magnitude, but first increases and then decreases with increasing inward perturbation magnitude.

In an attempt to better match the human data, k_{sx} (0.1174) and k_{sy} (0.2585) were decreased following the numerical optimization of the model foot placement location to that from the experimental data. This puts more relative emphasis on the step-to-step transition cost and ensures shorter and less wide steps, see Figure 5CD. The overall modulation remains intact for the steps in response to the perturbations. The main difficulty for the model is replicating the steep decrease in AP distance between the COM and the leading foot that accompanies wider steps.

Figure 4. Model responses to ML perturbations compared to human data using individual costs. (A) Foot locations relative to the COM using only a step-to-step cost (B) Single support duration using only a step-to-step cost. Other settings were as in Table 1. (C,D) As in (A,B), but for AP swing cost only. (E,F) As in (A,B), but for ML swing cost only. Triangles depict subject-average experimental data. Circles indicate the model output. Colors indicate the perturbation magnitudes, corresponding to the x-axis in (F).

Figure 5. Model responses to ML perturbations in comparison to human data. (A) Stepping locations relative to the COM. (B) Duration of the single support phase in which the perturbation is applied. Simulation settings were as in Table 1. (C,D) As in (A,B), except k_{sx} and k_{sy} were set by minimizing the model's leading foot position relative to the human experimental data. Triangles depict subject-average experimental data and indicate the perturbation direction. Circles indicate the model output. Colors indicate the perturbation magnitudes, corresponding to the x-axis in (D).

Figure 6. Model responses to AP perturbations in comparison to human data. (A) Stepping locations relative to the COM. (B) Duration of the single support phase in which the perturbation is applied. Simulation settings were as in Table 1. (C,D) As in (A,B), but for perturbation magnitudes that are three times lower. In (D), the experimental data is the same as in (B), but scaled on the x-axis to align it with the lowered model perturbation magnitudes. Furthermore, the AP distance between the COM and the leading foot was set to that of the steady-state gait response to the perturbations. Triangles depict subject-average experimental data and indicate the perturbation direction. Circles indicate the model output. Colors indicate the perturbation magnitudes.

Combined costs: recovery from AP perturbations

Under the default model parameter settings, the model also adjusts its foot placement according to the XCOM in response to AP perturbations. The AP distance between the leading foot and the COM increased for increasing forward velocity following forward perturbations. Backward perturbations resulted in opposite effects. This was to be expected given the XCOM constraint on the foot location. In contrast, humans keep the leading foot at a relatively constant AP distance from the COM. Given the mismatch in foot placement, the swing times do not match well either, see Figure 6AB.

When during the perturbation recovery the leading foot location relative to the COM is set to that of the steady state gait, the resulting stepping locations are similar to those in Figure 6C, but the resulting swing times remain comparable to those in Figure 6B. Remarkably, a better match in swing times could be obtained by additionally lowering the perturbation magnitudes with a factor 3, resulting in Figure 6CD. In such a case the model even predicts the slight increase in swing time for the lowest magnitude forward perturbation.

Discussion

We implemented costs for leg swing and step-to-step transition in a linear inverted pendulum model to investigate modulation in foot placement location and timing following various perturbations. The combined costs provide a way of discriminating between all possible times at which foot placement might occur at a given location. Because the location of foot placement was linked to the XCOM, which in turn is time dependent through the pendulum dynamics, both foot placement location and timing modulate simultaneously. The model can output human-like modulation in both the location and timing of foot placement, such as the wider steps of shorter length and duration in response to ML perturbations of increasing magnitude. However, the model also fails to replicate a steady-state gait with a human-like foot placement location, timing, and gait velocity all at the same time. The model furthermore has trouble replicating steps that do not correspond with the implemented XCOM constraint, such as the lack of foot placement adjustment following AP perturbations. It will require further model expansion to overcome these limitations, for example by allowing the dynamics to change through strategies other than foot placement.

Human-like simultaneous location and timing modulation

Balance controllers to realize perturbation rejection in walking models might struggle with the simultaneous selection of a location and timing for foot placement due to the abundance of available options. The XCOM-based constant offset control previously proposed in (Hof, 2008) did not adjust its fixed swing time in response to a perturbation. This causes foot placement in the sagittal plane to remain unaffected by ML perturbations, which is not the case in humans. In (Englsberger et al., 2013) an analytical foot placement planner was proposed based on the three-dimensional equivalent of the XCOM. It allowed an inverted pendulum to walk on uneven terrain, but the timing of foot placement had to be set in advance. A model predictive control scheme was used in (Diedam et al., 2008; Herdt et al., 2010) to realize an adaptive foot placement location in higher dimensional models. Here too a predetermined time of foot contact was used as a constraint on the foot placement location.

Simultaneous adjustment of foot placement location and timing was achieved in (Aftab et al., 2012), also using a model predictive control scheme. The optimization of the time was incorporated by

introducing a swing cost component, putting a penalty on the acceleration of the swing leg. The other costs that were minimized were the COM velocity and COM jerk (acceleration derivative), trunk angular velocity and angular jerk, and the COP distance from the center of the support foot. This made the model come to a stop, rather than have it walk. A comparison between this control scheme and human experimental data from literature was provided in (Aftab et al., 2016), for a forward fall from a standing position (tether-release). Absolute errors for the first recovery step were typically in the order of 10 cm for step length, and 0.02 s for step time throughout a range of initial body lean angles.

In walking, an LIPM model was also made to adjust the location and timing of foot placement (Khadiv et al., 2016). The swing time was obtained through an optimization procedure that attempted to realize steps with a length, width, and XCOM offset that were as close as possible to their predefined values. This assumes that humans try to enforce a certain step length, width, and XCOM offset, while using the swing time as a trade-off parameter. In the current study, we only assume that humans adhere to an XCOM constraint. A comparison with human data was not provided in (Khadiv et al., 2016).

In this work, the implemented costs provide a possible link to human behavior that is not strictly required for the LIPM to walk, or to have it make recovery steps. That is, the model can easily walk with other stepping strategies that are not necessarily human-like. For the LIPM specifically, a step location and time might be explicitly calculated from the solutions to the equations of motion, given some steady-state gait characteristics (De Boer, 2012). However there is no guarantee that the computed steps will compare well to human behavior. The similarities to the human modulation patterns realized in this study are therefore a direct result of the implemented costs and the XCOM constraint for foot placement.

The XCOM as constraint

The XCOM was used as a state-dependent constraint on the foot location, allowing the swing time to follow from the implemented costs. However the XCOM constraint does not hold on various occasions. It can be shown that humans do not adhere to this constraint in the AP direction for large magnitude ML perturbations (see Figure S4). Instead, the AP offset becomes more negative with increasing perturbation magnitude. This might explain why the model fails to replicate the steep decrease in AP distance between the COM and the leading foot following high magnitude ML perturbations. A similar issue occurs for AP perturbations, where the COP location modulates with the XCOM, but the average location of the foot does not (Vlutters et al., 2016).

For AP perturbations, a better match in location and timing was achieved by releasing the XCOM constraint and reducing the perturbation magnitude. The reduction in perturbation magnitude might be required because humans can counteract the perturbations during the single support phase, such as through ankle modulation, whereas the model cannot. Another explanation might be that humans move toward their leading foot during the double support phase. Because no double support phase occurs in the model, the COM has a relatively large distance from the COP when an AP perturbation occurs at toe-off. From a phase diagram (Figure S5) it can be shown that the model becomes increasingly sensitive to AP velocity perturbations with increasing distance between y_{COM} and y_{COP} . Together with the lack of control in the single support phase this might lead to a higher sensitivity to the perturbations than in humans.

Complex humans vs simple model

Our model can mimic the basic aspects of foot placement modulation, despite the absence of various factors that also contribute to human behavior. Inertial strategies such as upper body rotation, as well as center of pressure modulation through ankle moments in the stance leg play a role in balance during walking (Maus et al., 2010; Gruben and Boehm, 2014) (Chapter 7). For example, modulation of $\mathbf{r}_{\text{COP}}$ using the ankle of the stance leg might assist in perturbation rejection and reduce the need for foot placement adjustments. This could affect both the implemented swing cost and the step-to-step transition cost. In addition, the model does not take into account collision avoidance between the swing leg and the stance leg. This would be especially relevant for cross-steps following higher magnitude inward perturbations. Finally, other task-dependent costs such as those associated with the desired direction of walking and the global lateral position (i.e. “center of the road”) may also play a role.

We furthermore assumed that the implemented costs were representative of the metabolic costs required to swing the leg and transfer the COM to the leading leg. For leg swing specifically, a cost proportional to muscle force and inverse proportional to time has shown to provide the best match in predicting the speed-step length relationship in steady-state gait when combined with a cost for push-off (Kuo, 2001). Such a cost increases with increasing hip moment amplitude and with shorter moment duration, possibly due to the recruitment of less efficient muscle fibers. The swing cost in equation [10] appears to decrease with shorter durations t_f . However, it can be inversely proportional to time because $\mathbf{M}_H$ can rapidly decrease with increasing t_f . This occurs if a longer lasting swing is closer to the natural dynamics of the swing leg. Because the step-to-step transition cost tends to force the model into very rapid steps, the model generally walks at frequencies above the natural swing dynamics when both swing and step-to-step costs are active. As a result, equation [10] mainly operates in the domain where it is inversely proportional to t_f .

Conclusions

The current study results suggests that a swing leg cost and a step-to-step transition cost, together with a state-dependent foot placement indicator such as the XCOM, might provide the basic components for a human-like stepping strategy in response to ML disturbances. Both the XCOM concept and the implemented costs that contribute to the presented results provide a link to human behavior, with all concepts supported by experimental data (Donelan et al., 2002; Doke et al., 2005; Hof et al., 2010; Vlutters et al., 2016). Furthermore, only a combination of the implemented costs, and none of the individual costs, could approximate a human-like foot placement modulation pattern.

The study does not exclude other possible explanations for the observed foot placement modulation. Furthermore, the XCOM currently provides an overly strict constraint for a state-dependent foot location. Other constraints on - or predictors for - the foot location will have to be added to allow relaxation of the XCOM constraint. Conceptual models such as presented in this study provide an indication of whether the implemented concepts are feasible in predicting human-like balance control.

Supplementary material

On the sensitivity of the steady-state gait

The offsets $\mathbf{B}$ and the gains k affect how the model walks in steady state, see Figure S1. Recovery steps in response to perturbations alter in a similar way with changing $\mathbf{B}$ and k (not shown). With increasing positive B_x , steps become wider, while the walking speed, swing time, and step length all decrease. This might also be expected in humans given the limited leg length. Excessive wide steps would hamper forward progression. With a more negative B_y , the walking speed increases, while swing time and step width decrease. In humans, too, swing time and step width decrease with increasing walking speed, while step length increases (Grieve and Gear, 1966; Kuo, 2001). However in the model the step length saturates due to the (fixed) step-to-step transition cost gain.

For changes in the gains k , the walking velocity is affected in various ways, while the step length, width, and time are affected in a similar manner. Specifically, only for increasing k_{sy} the walking speed, step length, width, and time increase, while k_{sx} and k_{sts} have the opposite effect. For the swing cost gains, the sign of B_x , B_y , determines whether the steady state parameters increase or decrease with increasing gain k_{sx} , k_{sy} .

In humans the relative importance of the step-to-step transition cost likely lowers when a higher walking speed is desired. At higher speeds humans transfer to a more impulsive, bouncing gait with increased vertical COM motion (Orendurff et al., 2004; Srinivasan and Ruina, 2006), and eventually leave the floor completely when running. In the model, too, decreasing k_{sts} leads to an increase in step length and walking speed. When the step-to-step transition cost gain is fixed, the urgency to step becomes higher at higher walking velocities. Setting B_y to a more negative value generally leads to a higher walking velocity. This explains why the step length saturates with increasing walking speed under fixed k_{sts} and a more negative B_y . Consequently, both $\mathbf{B}$ and gains k will have to co-vary to mimic human-like steady-state gait over a range of walking velocities.

Figure S1. Steady-state gait sensitivity to model parameters. (A) Modulation of step length, width, time, and average forward gait speed with the choice of offset parameter $\mathbf{B}$. (B) Same as A, but for cost gains k_{sx} for the ML swing cost, k_{sy} for the AP swing cost, and k_{sts} for the step-to-step transition cost. The way the steady state gait modulates with $\mathbf{B}$ is dependent on the choice of the k and vice versa. When investigating the effect of a parameter, the others are set to their default value from Table 1, corresponding to the cross in the graphs. The recovery steps in response to perturbations would modulate in a similar way with changes in $\mathbf{B}$ and k .

Figure S2. Model responses to ML perturbations in comparison to human data. (A) Stepping locations relative to the COM for $k_{sx} = 0$ (B) Duration of the single support phase in which the perturbation is applied for $k_{sx} = 0$. Other simulation settings were as in Table 1. (C,D) As in (A,B), but for $k_{sy} = 0$. As in (A,B), but for $k_{sts} = 0$. Triangles depict subject-average experimental data and indicate the perturbation direction. Circles indicate the model output. Colors indicate the perturbation magnitudes, corresponding to the x-axis in (F).

Figure S3. Model responses to ML perturbations in comparison to human data. (A) Stepping locations relative to the COM for $k_{sts} = 0$, and k_{sx} twice as large as k_{sy} . (B) Duration of the single support phase in which the perturbation is applied for $k_{sts} = 0$, and k_{sx} twice as large as k_{sy} . Other simulation settings were as in Table 1. (C,D) As in (A,B), but for $k_{sts} = 1$ instead of 0. Triangles depict subject-average experimental data and indicate the perturbation direction. Circles indicate the model output. Colors indicate the perturbation magnitudes, corresponding to the x-axis in (F).

Figure S4. Relation between COM velocity at heel strike right (HSR) and the AP distance between the COM and the COP at toe-off left (TOL) following ML perturbations. For larger magnitude perturbations, the distance between the COP and the XCOM tends to increase. Triangles show subject means and indicate the perturbation direction. Ellipses represent the subject standard deviation. Colors indicate the various perturbation magnitudes. Pink solid and dashed lines indicate the subject mean and standard deviation, respectively, of the XCOM relative to the COM. Data is from (Vlutters et al., 2016) and is shown dimensionless.

Figure S5. Phase diagram for motion of the linear inverted pendulum in a single plane. Each curve corresponds with a ballistic COM trajectory, associated with a constant energy level per unit mass (E_{orb}) of the pendulum. Pink lines correspond with zero energy and with the XCOM. Blue vertical arrows indicate an instantaneous shift in COM velocity from the same COM trajectory. Both arrows result in the same velocity change, but one shifts the pendulum more energy levels than the other. Hence, if $E_{orb} > 0$ and the pendulum topples over the COP, the pendulum is increasingly sensitive to velocity changes with increasing distance between y_{COM} and y_{COP} .

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CHAPTER 10

General discussion

“Happy feet, wombo combo”

Introduction

The main aim of this thesis was to gain insight in balance strategies applied by humans following perturbations during walking. Specifically, the focus was on how humans change the location and timing of their foot placement to recover from the disturbances applied at the body's center of mass, located near the pelvis. Foot placement adjustments are of particular interest not only because these are an integral component walking, but also because it is not well understood how humans come to a consistent foot placement location and time given a disturbance. Because of this limited understanding it is challenging to ensure balance in bipedal robotic devices such as exoskeletons. A better understanding of human-like foot placement could therefore provide directions for the control of such devices. This might allow an exoskeleton to support its user in maintaining balance rather than the other way around. A series of perturbation experiments were therefore performed to collect and analyze data on human balance control during walking.

There are many location-time combinations to choose from when placing the foot. As shown in this thesis, this choice is affected by physical constraints on not only the foot placement strategy as in Chapter 4, but also on other strategies that complement foot placement as a balance strategy, such as the ankle strategy in Chapter 5. Fortunately, mostly consistent foot placement adjustments occur both within and across subjects given specific perturbation conditions, such that these might be predicted. The second aim of this thesis was to make predictions of human foot placement adjustments using low dimensional models and concepts derived from such models. In particular, a center of mass velocity dependent concept derived from a simple linear inverted pendulum model was compared to the collected data. Simulations of a linear inverted pendulum model with a swing leg were furthermore performed in an attempt to replicate the adjustments in foot location and time from in the experimental data. In the following sections the findings of the studies in this thesis will be considered together in a wider scope.

Findings and insights

Foot placement location should be assessed in a body-centered coordinate system

At least one foot is in contact with the environment during walking, often stationary on the floor until it is lifted and placed again to adjust the base of support. When an external perturbation occurs the body will move relative to the environment as a result of the perturbation, and therefore move relative to the foot or feet that are stationary on the floor. Meanwhile, the perturbed person might alter the base of support by adjusting the location of a foot in response to the disturbance. The distance between the feet is therefore affected by both the direct mechanical effects from a perturbation, as well as by foot adjustments induced by the subjects. Consequently, assessing a foot placement location relative to the other foot makes it more difficult to distinguish between direct mechanical effects and control exerted by the subject, and makes it impossible to see how the feet change relative to the body.

This becomes apparent in Chapters 3-5, where horizontal pelvis perturbations were applied during the single support phase of walking. The location of the trailing foot relative to the whole-body center of mass (COM) altered with the perturbation magnitude and direction. This is mostly because the body moves relative to the stationary trailing foot as a result of the perturbation. However, for anteroposterior (AP) perturbations the leading foot position relative to the COM was not significantly adjusted compared to the unperturbed condition. This leading foot position is mostly a result from controlled actions, as subjects had sufficient time to swing their leg in a desired direction (Chapter 3, 5). Consequently, if foot placement would have been evaluated relative to the other foot, the conclusion would have been that foot placement would adjust. In reality, only the body position relative to the trailing foot was altered, mainly through direct mechanical effects of the perturbation. The importance of expressing foot placement in a body-centered frame was also highlighted in stroke survivors in Chapter 6. When the feet locations are regarded relative to the whole-body COM, there exist differences between steps made with the paretic and the non-paretic legs. These differences become inconspicuous when the distance between the feet is used as an evaluation metric.

Foot placement adjustments are not always required to recover balance during walking

There is no straightforward answer to the question how humans adjust their foot placement in response to perturbations, due to the dependency on many factors. This thesis provides insight in foot placement adjustments, or lack thereof, specifically for horizontal perturbations applied at the pelvis during walking. Despite probing balance control with rather specific conditions, the findings contribute to a more general understanding of balance control during walking.

A previous study has used regression analysis in an attempt to make predictions of the leading foot position relative to the stance foot, based on the pelvis position relative to the stance foot (Wang and Srinivasan, 2014). It was shown that humans generally place their foot in the direction of the fall during unperturbed walking. However, this does not imply that a perturbation always requires foot placement adjustments compared to an unperturbed scenario to prevent a fall. As shown in Chapter 3, for mediolateral (ML) directed pelvis perturbations humans indeed place their leading foot more medially or laterally relative to the center of mass (COM) during the first recovery step, in the direction of the perturbation. The higher the perturbation magnitude, the larger the resulting change in COM velocity, and the further the foot is placed in the perturbation direction. Steps in which the leading foot is placed at a larger ML distance from the COM are generally also steps of shorter duration. In contrast,

for anteroposterior (AP) perturbations no significant adjustments in foot placement location relative to the COM occurred during the first recovery step, even when there was sufficient time to utilize a foot placement strategy. As shown in Chapter 7, other strategies such as ankle and hip moment modulation in the stance leg can contribute to the balance recovery. An ankle strategy is especially effective in the sagittal plane, which might reduce the need to make foot placement adjustments as compared to unperturbed walking. The lack of these leading foot adjustments also explains the poor predictive value of the pelvis state in predicting the anteroposterior (AP) location of the foot during normal walking (Wang and Srinivasan, 2014). A potential explanation for not adjusting the foot compared to unperturbed walking might be related to the energetic costs associated with the transition to the leading leg (Donelan et al., 2002). This cost rapidly increases with step size, following a loss in height of the COM that accompanies longer steps. Determining the costs that are associated with the relative use of the available strategies is one of the major challenges in understanding human balance control. Chapter 9 provides exploration of such costs.

Strategy availability affects the overall balance response during walking

Since multiple strategies can contribute to balance control, the need for foot placement adjustments is not only affected by the availability of the foot placement strategy itself, but also that of other strategies. This already becomes clear from perturbations applied during normal walking (Chapter 3). For example, an ankle strategy through modulation of ankle moments is especially effective in rejecting the AP perturbations, and less those of ML perturbations, given the dimensions of the human foot. This could explain the lack of foot placement adjustments in the first recovery step following AP perturbations, as observed in Chapter 3 and 4. In Chapter 4 and 5 the effect of various additional balance constraints was investigated.

Reducing the contributions of ankle joint moments and foot roll-over to the balance recovery can elicit foot placement adjustments. In Chapter 5 a pair of “pin-shoes” were introduced to exclude the ankle strategy. This set of modified ankle-foot orthoses physically blocked the ankle joints and strongly reduced the base of support area of each foot. In contrast to barefoot subjects, subjects that wore the pin-shoes showed significant foot placement modulation in response to AP perturbations as compared to the unperturbed case. The foot was moved further in the perturbation direction. This difference with barefoot subjects indicates that the use of an ankle strategy complements the foot placement strategy for these perturbations. Furthermore, the removal of the ankle strategy did not increase trunk motion in response to the perturbations, suggesting that inertial strategies by moving the trunk are not favored. Humans likely prefer to keep the upper body upright, perhaps to keep the eyes on the road ahead, or because it is simply an inefficient strategy.

In a similar way, limited availability of foot placement adjustments can elicit various other strategies. In Chapter 4 a time constraint was introduced on foot placement adjustments by applying perturbations briefly before foot contact. For example, ML perturbations applied short before heel strike can elicit swing leg abduction opposite of the perturbation direction in the subsequent single support phase, either as an inertial strategy or to provide a counter weight. Another example is pivoting of the stance foot to adjust the base of support without making an actual step. These responses demonstrate that healthy humans try to prevent a fall using the options they have. Finally, no actual falls occurred throughout the various constrained conditions in Chapter 4 and 5. This implies that there is redundancy, or better,

abundance in human balance control. The various balance strategies complement each other; even though one mechanism is deteriorated or temporarily less available, the healthy subjects that participated in the experiments were all able to stay upright and continue walking.

Foot placement is controlled, but not necessarily its location or timing

The range of possible combinations in the location and timing of foot placement to choose from is very large. Yet, the results in chapters 3-5 show that the modulation of these parameters is mostly consistent across healthy subjects within the given conditions. This suggests some form of preference. From Chapter 7 it appears that the swing leg is actively positioned prior to foot contact. The location and timing of foot placement during balance recovery therefore arise from control.

However, it remains unclear to what extent the location and timing of foot placement are explicitly controlled parameters, or emerge from other control mechanisms. This is dependent on the objectives we are trying to achieve during walking, which might quickly change depending on the situation. When asked to walk on stepping stones or at a metronome pace, the objective explicitly becomes to control the movement according to a certain location or timing of foot placement. When no such cues are provided, a more centralized level objective such as COM motion is likely to govern the observed responses (Safavynia and Ting, 2013), with the location and timing of foot placement as a possible by-product.

The differences in foot placement between Chapter 3 and Chapter 5 show that subjects do not forcefully adhere to a certain foot placement modulation pattern given the same perturbations, but different constraints on the ankle joint. Other carried-out perturbation experiments (not presented in this thesis) can show that the exact same perturbation conditions as in Chapter 3 can elicit different foot placement locations and timing when the subjects is given an additional high-level lane switch task. That is, a subject might be asked to shift the “center of the road” to the left or to the right whenever a ML perturbation is sensed. If the perturbation is in the same direction as the desired shift, it can help the subject move in the desired direction. If the perturbation is directed opposite of the desired shift, the perturbation interferes with the desired shift. This way, the exact same perturbation can be disturbing to various degrees, depending on the subject's intention. This is also reflected in the foot placement location and timing, which shift depending on whether the desired switch and the perturbation align or not, while the overall modulation pattern with the perturbations remains intact. These results clearly indicate that foot placement is a means of controlling the body, rather than the objective itself.

Furthermore, the location and timing of foot placement can be a by-product of another strategy. In Chapter 4, several perturbations lead to balance strategies during the single support phase that are not directly related to stepping. For example, swing leg abduction occurred in a direction opposite to that of the fall, possibly as inertial strategy or to provide a counterweight. This leg abduction resulted in a relatively long duration of the single support phase. The timing of foot placement is influenced by the use of another strategy before foot contact. It is therefore unlikely that subjects are trying to achieve a timing of foot placement associated with a specific swing duration, as measured from the instance of the preceding toe-off.

Different balance strategies can contribute to the same underlying relation

Previous work has provided the derivation of the extrapolated center of mass (XCOM) (Hof et al., 2005), also known as the capture point (Pratt et al., 2006), or its three-dimensional equivalent the divergent component of motion (Takenaka et al., 2009). The XCOM concept is directly related to the horizontal COM velocity through some factor $\omega_0^{-1} = \sqrt{l/g}$. A linear inverted pendulum model of walking comes to a full stop if its center of pressure (COP) is placed onto the XCOM. Hence the potential importance of this concept in balance control. Previous experimental findings suggest that humans laterally move their COP towards a (lateral) fixed distance from the XCOM following foot placement adjustments in response to ML perturbations (Hof et al., 2010). This is also known as constant offset control (Hof, 2008). If this were to hold, the COM velocity might be used as an instantaneous predictor for the location of the COP, and therefore for foot placement if it is the primary strategy to realize the COP adjustment.

The findings in Chapters 3-5 strongly support this predictive value of the COM velocity, but with several important differences. First, for higher walking speeds in Chapter 3 the COM velocity related linearly to the COP adjustments, but the similarities with the XCOM were lower compared to those in a lower walking speed. Second, in Chapter 3 it was shown that the predictive value of the COM velocity also holds in the AP direction following AP perturbations. However, the relation is realized through strategies other than foot placement adjustments. Chapters 3-5 show that this COM velocity – COP relation persists throughout various balance strategies, including those not involving foot placement adjustments. In Chapter 4, this relation persisted throughout inconsistent balance strategies across subjects, induced by a specific perturbation. In Chapter 5 the applied balance strategies changed as compared to those in Chapter 3 due to the introduced pin-shoes, but the relation between COM velocity and COP location persisted here as well. This relation is therefore actively established, and is not a result of passive dynamics only. This COM velocity dependent modulation of the COP that occurs in both ML and AP directions of movement might reflect an underlying objective towards which multiple balance strategies contribute.

The COM velocity – COP relation is actively established throughout multiple gait phases

Another difference between the findings in Hof's work and this thesis, is that in the thesis the best predictions of the COP location based on the COM velocity span two instances of the gait cycle. That is, given the COM velocity at heel strike, the proportionality constant ω_0^{-1} can be used to make a prediction of the COP location relative to the COM at the subsequent toe-off. However, the linear inverted pendulum model from which ω_0^{-1} is derived is generally used to model walking as a sequence of single support phases while ignoring the double support phase. It is unclear why a concept derived from such a model corresponds with human data that spans the double support phase.

The linear inverted pendulum model describes COM motion relative to a stationary COP. However, the equations can be rewritten for a stationary COM and a moving COP by simply shifting the origin of the model to the COM. During the double support phase, the COP moves from the trailing foot to the leading foot during the weight shift, at a much higher velocity than the COM. If the ground reaction force vector during the double support phase is treated as a leg attached to the COM (hence ignoring inertial effects), and all COP locations are shifted toward the same location, one ends up with a motion that resembles that of a linear inverted pendulum, see Figure 1. There might be more similarities

between the linear inverted pendulum model and the double support phase than initially appears. However, it remains unclear why the COP modulation occurs in accordance with the XCOM over multiple instances of the gait cycle. The relation is made up of data at the instance of heel strike and toe-off, such that the detection of both these instances can therefore affect the relation. In Chapter 5 gait phase detection was performed using the ground reaction forces, rather than the marker data such as in Chapters 3, 4, and 6. In the latter chapters, ML perturbations lead to cross-stepping, such that the subject occasionally stood with both feet on the same force plate of the treadmill. This makes force-based detection cumbersome. Still, the relations exist throughout the various detection methods in Chapter 3-5, making it unlikely that errors in gait phase detection are a major factor of influence. The COP shift during the double support phase is dependent on the forces delivered by the trailing and leading legs. These are in turn dependent on the perturbation characteristics, as well as on actions taken during the preceding single support phase. If the COP cannot shift because foot placement adjustments did not occur, the relation cannot arise. This is demonstrated in Chapter 4 for ML perturbations, which suggests that the COP modulation in accordance with the XCOM mainly occurs in the first recovery step if foot placement adjustments can play a role. If a perturbation is given early in the single support phase, the relation occurs in the first recovery step. It diminishes in the second step, possibly because other actions such as returning to the center of the treadmill begin to play a role. If a perturbation is given late in the single support phase and there is no time for foot adjustments, the relation disappears in the first step, and tends to re-appear in the second step. Hence, though the relation arises during the double support phase, it cannot be attributed to the double support only, and is also dependent on actions during the preceding single support phase.

Future experiments might point out whether these findings remain intact for variations in leg length l and gravitational acceleration g , which make up ω_0^{-1} . In Chapter 5 leg length l was already slightly altered by introducing the pin-shoes. A much larger effect might occur by artificially lowering gravity through a body weight support system, or by increasing it by pulling someone toward the floor. A lower gravity is expected to lead to foot placement further away from the COM, while a higher one should lead to foot placement closer to the COM. If under such conditions the relations would hold, it would further support that both leg length and gravity are crucial parameters in determining how we step.

Figure 1. The similarities between the weight shift that occurs between heel strike and toe-off during the double support phase, and the linear inverted pendulum model. The ground reaction force vectors might be treated as “legs”. Shifting the COP locations to a stationary point yields a pendulum-like motion.

The XCOM as constraint

The modulation of the COP with the COM velocity, in line with the XCOM, might be used as a constraint in mimicking human-like balance. Given a COM velocity, the challenge becomes employing the right strategies to realize the forthcoming COP shift. This is applied in Chapter 9 in a linear inverted pendulum model of walking, for which the COP and foot locations coincide. Foot placement is made state-dependent by limiting the foot location to a fixed distance from the XCOM. This leaves only the selection of a time for foot contact. The latter was achieved by implementing a combination of costs related to leg swing and the transition from one leg to the next. The resulting modulation in both the location and timing of foot placement were in line with those in human experimental data.

The linear inverted pendulum model can only to shift the COP and affect COM motion through foot placement. Similarities between model and human experimental data are therefore mainly expected for data in which the COP shift suffices the XCOM constraint, and is primarily realized through foot placement adjustments and not through some other strategy. Chapter 9 shows that the similarities between model and human foot placement strategies were indeed better for ML perturbations, for which humans mostly apply foot placement, than for AP perturbations, in which the foot hardly adjusts. Hence, the foot placement strategy can indeed be predicted – to some extent - using simple low dimensional models of walking, and concepts derived from these models. When strategies other than foot placement play a prominent role the problem becomes more complex, and a simple inverted pendulum model might no longer provide sufficient descriptive value. In such a case the degrees of freedom associated with other strategies need to be incorporated, and their interplay will have to be described with additional cost components.

Reactive balance control involves supra-spinal structures

A prominent professor in neuroscience once stated: "*we have a brain for one reason, and one reason only; that is to produce adaptable and complex movements*" (Wolpert, 2009). The reverse is not strictly true. Adaptable movements during walking can be realized through local sensory feedback loops without any input from the brain (Geyer and Herr, 2010). Passive dynamic walking does not even require any input at all, and is purely dependent on the dynamics of the body, using gravity to realize walking on a slope (McGeer, 1990). However, the results presented in Chapter 2, 7, and 8 show that human balance control requires active neural processes that lead to changes in muscle activation. Strange as it may sound, even not moving requires movement, due to the inherent instability of our upright body.

The movements we generate in response to external perturbations involve more than just local feedback loops. Other supra-spinal processes are involved that process information regarding the state of the entire body, and enable complex, adaptive responses. This is indicated by the results presented in Chapter 8, in which perturbation magnitude- and direction-dependent muscle responses occurred, even though those muscles were unlikely to be influenced by length and load changes through joint rotation, and even though those muscles had no functional contribution to balance. Given the 100 ms response latency and the unlikely contributions from (changes in) afferent feedback, it is likely that the observed responses involved supra-spinal pathways. This might also contribute to the ability of the stroke survivors in Chapter 6 to modulate the swing leg in response to perturbations, possibly bypassing brain areas affected by the stroke.

These responses are likely the “automated postural responses” described in (Nashner and Cordo, 1981), taking precedence over any voluntary induced motion. The occurrence of these responses motivates the inclusion of whole-body or task-level control loops in neuromuscular models of walking. To fully understand and replicate human balance control requires understanding and modeling the involved parts of the nervous system. This is a tremendous challenge, and indicates the complexity of human balance control.

Implications for fall prevention training and balance support

A previous study has shown that the angle of the stepping leg can be used as a predictor for backward falls following support surface translations in healthy young subjects (Weerdesteyn et al., 2012). As can be derived from Chapter 3, healthy subjects show a strong correlation between a foot placement location required to allow certain COP excursions, and the forces that are delivered after foot contact to redirect the body. If the same ground reaction force vectors would be delivered without displacing the COP, the body would rotate. This is because a lack in COP shift would create a large offset of these vectors from the COM. In this sense, foot placement adjustments could be a prerequisite for a successful recovery, as no other strategy might allow generation of the same combinations of COP excursions and ground reaction forces. This would motivate training adequate foot placement in impaired people to restore balance control and gait function.

However, foot placement does not guarantee anything. If no appropriate forces are delivered after foot contact, a fall will still occur. As shown in Chapter 7, each perturbation results in a unique profile of recovery moments distributed across multiple joints. If a person is capable of moving the foot to a certain location by swinging the leg, but cannot deliver the forces required to restore the desired body motion after foot contact, it becomes meaningless to realize such a step. Consequently, if an impaired person shows foot placement strategies that are different from those in a healthy subject, the question is whether the underlying cause is the inability to swing the leg, or the inability to control the body following foot contact of such a step. To determine which of these should be targeted in gait training or gait support of a specific subject, it is required to characterize the impairments that cause decreased balance performance. For instance, the results in Chapter 6 suggest that the investigated group of stroke survivors are capable of adjusting either leg in response to perturbations during walking. Enhancing the leg strength of these subjects might improve their foot placement strategies, even without directly assisting the leg swing. Experiments with (partially) impaired subjects supported by an exoskeleton might point out whether this is the case.

Limitations

The focus on this work was mostly on the first recovery step after a horizontal pelvis perturbation. Chapter 4 and 7 provide some insight into the second step, but subsequent steps have not been extensively analyzed. As shown in Chapter 4, the first recovery step influences the subsequent steps. Even though multiple steps often play a role in the perturbation recovery, it makes sense to start by characterizing the first. However, the subsequent steps are still a part of the recovery, during which actions such as the return to the center of the treadmill are more likely to occur. At this point it is unclear how many steps subjects need to recover from the disturbances (as this requires a definition of “recovered”), and what happens in each of these steps. Experimental data in (Matthis et al., 2017)

suggests that humans plan at least two steps ahead when walking on stepping stones, but it is unclear how this translates to walking without a foot placement objective. Another study used simulations in a 2D linear inverted pendulum model to argue that at most two steps is sufficient to achieve a target speed, without the need to plan far ahead (Zaytsev et al., 2015).

All experiments in this thesis were performed on a treadmill. Though the treadmill surface was sufficiently large to allow recovery steps in all directions, potential effects of the treadmill dimensions and the range of motion of the perturbation device cannot be excluded. Perhaps subjects would prefer to make a second lateral step following a ML perturbation during overground walking, which was limited possible in the used experimental setup. A study with an ambulatory perturbation device has suggested that stepping responses during treadmill walking and during overground walking are similar, and that a treadmill is even preferred due to the better access to force plates (Zadavec et al., 2016). However, a treadmill imposes a walking speed, forcing subjects to keep walking after a disturbance. Holding the feet in place after a recovery step for a prolonged time is not an option. Meanwhile, an overground environment encourages subjects to select their own preferred walking speed, which could introduce variability in the balance recovery responses *because* a specific walking speed is not enforced. Again, the imposed constraints on the subject might play an important role in how they perform a balance recovery.

The analysis in this thesis predominantly focused on high-level parameters such as the COM and COP and relations between them, analogous to those in simple low dimensional inverted pendulum model. More extensive analysis methods, perhaps data mining techniques, might be employed to find predictive value within larger parameter spaces (e.g. including combinations of joint angles and velocities) on future balance recovery actions. In addition, if common relations can be found throughout various perturbation conditions, not only for horizontal pelvis perturbations, it would be a step towards a more general understanding of balance responses. An extensive analysis over many variables and multiple perturbation conditions would require a large amount of experimental data, with comparable data available throughout all experimental conditions. This motivates a standardized and systematic approach.

Future considerations

Standardize

Balance control is dependent on the capabilities and constraints of the system at hand, while the observed balance responses are dependent on magnitude, direction, timing, and point of application of a perturbation. Meanwhile, researchers are approaching the problem of understanding balance in many different ways, with different perturbation characteristics and constraints, in different subject populations, and using different models to mimic and explain the observed behavior. Plowing through the existing literature in an attempt to unify various findings in a single framework is therefore a difficult task. It might be beneficial to standardize experimental protocols and perturbation characteristics, converge toward a generally accepted normalization of the results, and test balance control algorithms on various openly accessible models. A more unified and systematic approach could make it easier to compare experimental findings, to compare the performance of various balance controllers on the same model, and to complement and expand each others work.

Furthermore, existing frameworks of human movement control might provide a more general basis for

balance. For example, in optimal feedback control (Todorov and Jordan, 2002) humans regulate their movement with respect to task-level objectives, minimizing the movement variability that contributes to the task(s). These frameworks can include probabilistic information from sensory systems regarding the body state, as well as prior information regarding the objective (Körding and Wolpert, 2004). In the latter study, subjects performed a reaching task during which visual information of the hand position was manipulated. A visual shift was randomly drawn from some Gaussian distribution. Over many reaching trials, subjects could gain prior information on the average shift. When during a subsequent trial the visual information of the hand position was made less reliable, subjects relied more on the learned prior information, in accordance with a Bayesian model. In balance control, humans might develop prior information related to stepping and falling during early childhood, in which we gain major balance experience (Adolph et al., 2012). Perhaps an experiment analogous to that in (Körding and Wolpert, 2004) might be performed during walking, to investigate whether or not balance control is compatible with these frameworks of human movement control. For example, a pair of shoes with a sole thickness or sole orientation that can be adjusted during the swing phase could alter our prior knowledge of foot contact, or that of the relation between a foot position and the ground reaction forces associated with that position. Depending on whether or not feedback is provided about the sole, a subject might rely more or less on the learned prior information. This might be reflected in a specific foot placement location or timing.

Find the objectives of balance control

What exactly is it that we are trying to do when a disturbance occurs? On what level (e.g. joint, endpoint, whole-body) do such objectives have to be defined to be able to explain the observed behavior. What is the relative importance of - or the costs associated with - these objectives? If such questions can be answered, the objectives and the costs associated with them might be expressed mathematically and implemented in a human-like model.

Experiments have shown that variability in steady-state human walking can be divided in objective-relevant and objective-irrelevant variability (Dingwell et al., 2010). Humans appear to minimize movement variability that contributes to some imposed objective(s). Specific analysis methods, such as uncontrolled manifold analysis (Scholz and Schöner, 1999) or goal equivalent manifold analysis (Cusumano and Cesari, 2006) might aid in identifying such objectives. As a simple example, multiple combinations of step length and time can lead to the same walking speed. Both step length and step time might show considerable variation, while their combinations to obtain a certain speed might be consistent. If the variability in length and time is significantly larger than that in speed, then speed control might be considered as an objective that can be achieved through the step length and time. Specifically, the dimensionless ω_b^{-1} line presented in several chapters might be treated as a goal equivalent manifold: it provides those combinations of COM speed and COP distance that suffice an XCOM objective. However, the total objective space is undoubtedly much larger, non-linear, and with many more variables contributing to it.

Muscle responses such as those presented in Chapter 8 might also provide an indication of whether a specific disturbance interferes with an objective of balance during walking, and how. If a disturbance triggers an automated postural adjustment, it likely interfered with a major objective of balance. Repeating the experiment from Chapter 5 and 8 for perturbation instances across the gait cycle might

better reveal the functional contributions of the ankle muscles to balance, and what it is we are trying to control.

Investigate balance in human-exoskeleton coupled systems

A motivation for the presented work was to gain knowledge of human balance control during walking, as a basis for natural, human-like control of lower-extremity orthoses. Experiments were therefore conducted in healthy human subjects, but not in subjects coupled to any ambulatory device. The additional mass, inertia, capabilities, and control of such a device can affect how the user responds to a given disturbance, as well as how a perturbation affect the human-exoskeleton combined system. Where a human might need to adjust foot placement given some perturbation, the combined system might be able to maintain balance in a different way. A strong deviation between the human-exoskeleton combined system and the subjects investigated in this thesis might reduce the applicability of the findings in this thesis. Light and compact devices, perhaps exo-suits (Wehner et al., 2013), could ensure increased similarities with a healthy human in a mechanical way. In terms of control, only extensive testing with human subjects inside a device can point out how the user responds to the device, and whether the device is effective or not. The more the hardware and the control of that hardware advances, the more relevant the findings of balance control in healthy humans becomes.

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Summary

Foot placement in balance control – complex humans vs simple model

Maintaining balance in daily life is very common to us. For a healthy individual, a fall is simply not supposed to happen. Unfortunately, various conditions such as stroke, spinal cord injury, or aging can lead to balance problems and affect a person's mobility. Robotic devices such as powered orthoses, often referred to as exoskeletons, might provide an outcome in case of these balance problems.

In case of a lower-extremity exoskeleton, the user wears a construction around the legs that should provide support during standing and walking. This support can be to various purposes, such as to reduce the energetic costs of walking, to assist in gait rehabilitation, or to fully take over the walking motion. Although the purpose of the device might differ, most lower-extremity exoskeletons have one thing in common: they have no sense of balance. The exoskeleton cannot react to unexpected disturbances. Because of that, the user has to take the lead in making a balance recovery. This is especially troublesome when the user has balance impairments.

To tackle these issues, the control of exoskeletons needs to be improved. Specifically, if the device can assist in balance control in a way that feels natural and intuitive to the user, the device is less likely to conflict with the user's intention. To realize such human-like balance controllers, we must first understand what human balance is, and investigate the way healthy humans regain their balance when it is lost. This might be investigated by applying perturbations to experimental subjects. Disturbances will lead to a balance recovery response involving various balance strategies, such as adjustments in foot placement, or modulation of ankle and hip moments. Especially foot placement adjustments are considered a major balance strategy in human walking.

Though literature already provides a broad range of perturbation experiments and conditions, it is nearly impossible to investigate all possible forms of perturbations, applied to various body parts, during various circumstances. It is therefore crucial to converge towards models that can mimic and explain the observed human behavior when balance is disturbed. One of these models is the inverted pendulum, which describes the motion of a point mass on top of a single leg. This resembles the single support phase of human walking. Using experimental outcomes to validate simple models of human balance could provide a basis for the design of balance controllers for lower-extremity orthoses.

The focus of this thesis is on human balance recovery in response to external perturbations during walking. Because we mainly deal with walking, foot placement adjustments are expected to be a major, crucial strategy in balance control. This strategy might be replicated using simple inverted pendulum models of walking. To make comparison with such models more straightforward, perturbations are applied to the approximate location of the whole-body center of mass of the human, which is the pelvis. In particular, the thesis deals with the following questions:

I) How do humans adjust their foot placement in terms of location and time when subjected to horizontal pelvis perturbations during walking?

Ia) How do physical constraints affects the foot placement strategy for these perturbations?

Ib) How is the foot placement strategy complemented by other balance strategies for these perturbations?

II) Can the foot placement strategy be predicted using simple low dimensional models of walking, and concepts derived from these models?

The results presented in this work provide insight in how the location and timing of foot placement change following horizontal perturbations applied at the level of the center of mass. The most prominent foot placement modulation occurs for sideways perturbations, given that there is sufficient time to realize foot placement adjustments. The closer in time a disturbance occurs before foot contact, the less the step will be modulated. In those cases, adjustments partially transfer to the second step.

It also becomes clear that foot placement adjustments are not always the primary strategy to recover from perturbations during walking, even if it is an available option. Other strategies might be addressed instead. An example is the ankle strategy following fore-aft perturbations. Disabling the ankle strategy through a physical constraint elicits foot placement adjustments that were otherwise not present. This suggests that the ankle provides an alternative recovery mechanism.

The work furthermore reveals joint-level and muscle-level responses in reaction to disturbances. Such results can function as a benchmark for orthoses- and controller design in terms of actuator degrees-of-freedom and output. In addition, the muscle responses show that reactive balance control likely involves centralized, supra-spinal mechanisms, without the specific need for joint-level afferent feedback.

Finally, a concept derived from a simple inverted pendulum model is shown to have predictive value in human balance control. This concept provides a relation between the horizontal velocity of the body's center of mass, and the distance between the center of mass and the origin of the net ground reaction force following foot contact. When this concept is incorporated in a simple inverted pendulum model and is combined with two energetic costs, one for leg swing and one for leg transition, the model can yield human-like modulation of step location and step time in response to disturbances. These model results are in line with experimental results in perturbed human subjects, which brings us one step closer to replicating human-like balance in walking.

Samenvatting

Voetplaatsing tijdens balans controle – complexe mensen vs simpel model

In het dagelijks leven is balans de normaalste zaak van de wereld. Een gezond persoon hoort simpelweg niet om te vallen. Verschillende aandoeningen zoals een herseninfarct, een dwarslaesie, of gewoon ouderdom, kunnen echter leiden tot problemen met de balanshandhaving. Dit heeft een verminderde mobiliteit tot gevolg. In dit geval zou een robotisch hulpmiddel zoals een geactueerde orthese, ookwel exoskelet, een uitkomst kunnen bieden.

Bij gebruik van een exoskelet voor de onderste extremiteiten draagt de gebruiker een constructie rond de benen, welke ondersteuning kan bieden tijdens staan en lopen. Deze ondersteuning kan verschillende doelen hebben, zoals het verminderen van het energie verbruik tijdens de loopbeweging, het mogelijk maken van rehabilitatie en training van lopen, of zelfs het permanent overnemen van de gehele loopbeweging. Hoewel deze exoskeletten verschillende mogelijkheden kunnen bieden hebben ze vaak een ding in gemeen: ze hebben geen gevoel voor balans. Het exoskelet is vaak niet in staat om te reageren op onverwachte verstoringen. De gebruiker moet daarom het initiatief nemen om de balans te herstellen en een val te voorkomen. Dit is echter een probleem als de gebruiker zelf een verminderde balans heeft.

Om deze problemen aan te pakken moet de aansturing van exoskeletten verbeterd worden. Als het apparaat in staat zou zijn om de balans te ondersteunen op een menselijke manier, zal het dragen van een exoskelet natuurlijker aanvoelen, terwijl een val kan worden voorkomen. Om dit mogelijk te maken moet echter eerst worden onderzocht wat balans eigenlijk is, en wat mensen precies doen als ze hun balans verliezen. Dit kan duidelijk worden aan de hand van experimenten, waarbij proefpersonen worden verstoord doormiddel van een duw tegen het lichaam. Een verstoring zal leiden tot een balans herstel, waarbij verschillende strategieën kunnen worden toegepast. Vooral het aanpassen van de locatie en timing van de voetplaatsing wordt gezien als een belangrijke strategie voor balans herstel tijdens lopen.

Hoewel er in de wetenschappelijke literatuur een breed scala aan balans- en verstoringsexperimenten is te vinden, is het vrijwel onmogelijk om alle mogelijke condities in kaart te brengen. Er zijn vele soorten verstoringen (grootte, richting), die kunnen worden aangebracht op verschillende plekken van het lichaam, onder verschillende omstandigheden. Het is daarom belangrijk om modellen te vormen welke een verklaring kunnen geven voor wat mensen doen wanneer ze hun balans verliezen. Een voorbeeld van een zeer eenvoudig model is de omgekeerde slinger, welke de beweging beschrijft van een massa bovenop een been. Dit kan worden gebruikt als een benadering van de enkele standfase van de menselijke loopbeweging. Het gebruiken van experimentele uitkomsten voor het valideren van deze simpele modellen zou een basis kunnen leggen voor een natuurlijke aansturing van exoskeletten.

Dit proefschrift richt zich op balans herstel ten gevolge van verstoringen aangebracht tijdens lopen. Omdat de nadruk op lopen ligt, is de verwachting dat voetplaatsing een belangrijke rol speelt in de balans handhaving. Een voetplaatsing strategie zou kunnen worden gemodelleerd met behulp van een omgekeerd slinger model. Om de vergelijking tussen mensen en dit model eenvoudiger te maken kunnen experimenteel verstoringen worden aangebracht rond het massa-middelpunt van het menselijk lichaam, welke rond de heup ligt. Dit proefschrift behandelt de volgende specifieke vragen:

I) Hoe passen mensen hun voetplaatsing aan in termen van locatie en timing wanneer ze verstoort worden door een horizontale kracht op de heup tijdens het lopen?

Ia) Welk effect hebben fysieke begrenzingen op voetplaatsing voor deze verstoringen?

Ib) Hoe zijn andere balans strategieën complementair aan voetplaatsing voor deze verstoringen?

II) Kan een eenvoudig omgekeerd slinger model, en principes afgeleid van dit model, worden gebruikt om menselijke voetplaatsing te voorspellen?

De resultaten in dit proefschrift geven inzicht in hoe mensen hun voetplaatsing aanpassen in reactie op een horizontale verstoring aangebracht op de heup, ongeveer ten hoogte van het massa-middelpunt van het lichaam. De sterkste aanpassingen volgen in reactie op zijwaartse verstoringen, mits er voldoende tijd is om de aanpassing te doen. Als de verstoring wordt gegeven kort voor het moment van voetcontact zal er minder aanpassing zijn. Deze verschuift dan gedeeltelijk naar de volgende stap.

Tevens laten de resultaten zien dat voetplaatsing niet altijd de belangrijkste strategie is om balans te herstellen na een verstoring tijdens lopen, zelfs als de persoon in kwestie hier wel de mogelijkheid toe heeft. Andere strategieën kunnen ook worden angewend, zoals een enkel-strategie doormiddel van het beïnvloeden van de rotatie rond het enkelgewricht. Het fysiek blokkeren van de enkel strategie leidt tot aanpassingen in voetplaatsing welke anders niet aanwezig zijn. Dit wijst er op dat de enkel strategie als alternatief kan functioneren, en een voetplaatsing strategie dit overneemt in geval van disfunctioneren.

Verder wordt er inzicht gegeven in de reacties op verstoringen op het niveau van individuele gewrichten en spieren. Deze resultaten kunnen een richtlijn vormen voor de ontwikkeling van exoskeletten, in termen van actuator specificaties en vrijheidsgraden. De spier-responsies suggereren dat centrale, supra-spinale structuren een bijdrage leveren aan de balans aansturing, zonder dat daarbij feedback informatie nodig is van alle specifieke gewrichten.

Tot slot blijkt dat een concept afgeleid van een eenvoudig omgekeerd slinger model een voorspellende waarde heeft in balans controle. Het geeft een relatie tussen de horizontale snelheid van het massa middelpunt van het lichaam, en de afstand tussen dit massa middelpunt en de oorsprong van de grond reactie kracht volgend op voetplaatsing. Wanneer dit concept wordt gebruikt in combinatie met energetische kosten voor 1) het zwaaien van het been en 2) de transitie van het ene been naar het andere been, dan kan een omgekeerd slinger model een voetplaatsing strategie uitvoeren welke veel lijkt op die van de mens tijdens lopen. Dit brengt ons een stap dichtbij het nabootsen van menselijke balans.

Dissemination

Peer-reviewed publications

(Scan, or click for full Scopus search)

Awards

WeRob 2016 Best Student Contribution

Date: Oct 2016

Issuer: Prof. Dr. José Luis Pons Rovira & WeRob 2016 organizing committee

Description: Best student contribution for the WeRob 2016 conference on wearable exoskeletons.

Title: Joint-level responses to counteract perturbations during walking

Acknowledgments

The danger of naming is forgetting a name, and I do not intend to do that.

So thank *you*, dear reader, for your involvement in the realization or evaluation of this work and the studies presented in it, for being my supervisor, for having me as supervisor, for being my colleague, my roomie, my project partner, my experimental subject, or for simply reading this thesis out of interest or curiosity.

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This image shows a single sheet of white paper with horizontal ruling lines. The lines are evenly spaced and run across the width of the page. There are no margins, text, or other markings on the paper.

[illegible]

About the author

Mark Vlutters ('89, Enschede) obtained both his Bachelor ('10) and Master ('12) degrees in Biomedical Engineering with distinction (Cum Laude) at the University of Twente. For his Bachelor thesis he investigated ankle joint contributions to balance in standing humans. For his master thesis he investigated foot placement strategies to stabilize a linear inverted pendulum model of walking, and contributed to an exoskeleton simulation environment. After obtaining the MSc degree, he joined the European BALANCE project as a PhD researcher ('13). The aim of this project was to develop a lower extremity exoskeleton for the support of human balance during walking, in a natural and cooperative way. Mark his contributions are in gaining an understanding of how humans maintain their balance during walking when a disturbance occurs. A specific focus was on foot placement adjustments and the prediction thereof.