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Analysis of stability against falling by movement simulation in human walking

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ABSTRACT

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Falls pose a threat to society through fatality, injuries, and related economic burden. Falls are mainly related to tripping or slipping while walking. Means to prevent walking-related falls are, thus, needed. This Thesis addresses fall prevention from the perspective of mechanics and neuromuscular control of walking. To this end, stability of walking against falling is addressed through movement simulations. Bulk of the Thesis reviews a novel simulation-based method of a tripping perturbation for assessing the stability of walking, and application of the method. In addition, the definition and neuromuscular control of stability are dealt with.

A small whole-body angular momentum has previously been envisaged to indicate a good balance and a low fall risk during locomotion. Experimental data on the whole-body angular momentum have been reported, but theoretical reasoning regarding the meaning and importance of the angular momentum has so far been limited. Therefore, relation of the angular momentum to balance and falling remains ambiguous. An analysis of a two link system provided in the Thesis disclosed that a human can fall even if its whole body angular momentum is zero. This analysis showed that minimizing the whole-body angular momentum is not a universal way to prevent a fall. Consequently, the central nervous system must use some other control criteria in addition to or instead of minimizing the whole-body angular momentum to prevent a fall. In this Thesis, control of stability against falling is expected to be conducted through control of the trunk.

In this Thesis, stability of walking studied over a range of gait speeds by means of a tripping perturbation is reviewed. Fast walking has previously been related to high likelihood of falling due to tripping. It is unclear whether the increase in likelihood of falling is explicable by an increase in instability. Stability with respect to a constant tripping perturbation was quantified as the immediate passive response of the trunk to the perturbation. Subject-specific muscle-driven simulations of young healthy subjects walking at four speeds were analyzed. In the simulations, short perturbations were performed several times throughout the swing phase by applying a constant backward force to the swing foot of the model. This tripping perturbation revealed that gait speed has a weak and varying effect on stability. Thereby, fast walking is not necessarily more unstable than slow walking.

Thesis also reviews studies that made use of a tripping perturbation to detect impaired stability of walking. Subject-specific muscle-driven simulations of walking of children with and without cerebral palsy (CP) were analyzed by the tripping perturbation. Results showed that unimpaired subjects were stable for a larger percentage of the swing phase than CP subjects. Thus, CP subjects are more sensitive to a tripping perturbation, and, consequently, are likely to be more susceptible to falling than unimpaired subjects.

Thesis provides an examination of an unimpaired control of the trunk during walking. Studying an unimpaired control of the trunk reveals characteristics of good control. These characteristics can be pursued in the rehabilitation of an impaired control. Impaired control of the trunk during walking is associated with aging and many movement disorders. This is a concern as it is considered to increase the fall risk. Muscles that contribute to the trunk control in normal walking may also contribute to it under perturbed circumstances, attempting to prevent an impending fall. Analyses of muscle-driven simulations demonstrated that the abdominal and back muscles exhibited large contributions throughout the gait cycle in both the sagittal and frontal planes. The proximal lower-limb muscles contributed more than the distal muscles in the sagittal plane, while the both proximal and distal muscles showed large contributions in the frontal plane.

Keywords: Biomechanics, Human walking, Stability, Falling, Musculoskeletal model, Simulation

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PI R. Klemetti, P. Moilanen, J. Avela, J. Timonen. Effects of gait speed on stability of walking revealed by simulated response to tripping perturbation. *Gait Posture*, 39, 2014.

PII R. Klemetti, K. M. Steele, P. Moilanen, J. Avela, J. Timonen. Contributions of individual muscles to the sagittal- and frontal-plane angular accelerations of the trunk in walking. *J. Biomech.*, 47, 2014.

PIII R. Klemetti, P. Moilanen, J. Avela, J. Timonen. Deteriorated stabilization of walking in individuals with spastic cerebral palsy revealed by a simulated tripping perturbation. *Research*, 1, 2014.

PIV R. Klemetti, P. Moilanen, J. Avela, J. Timonen. Zero angular momentum can result in falling: A theoretical study. *Submitted for publication*.

The author of this Thesis conducted all analyses in and drafted manuscripts of articles [PI], [PII], [PIII], and [PIV].

RELATED CONFERENCE ARTICLE

[Klemetti13] Detection of impaired stability of walking by simulated response to tripping perturbation. XXIV Congress of the International Society of Biomechanics.

1 INTRODUCTION

Falls are the leading cause of unintentional-injury deaths in Finland [Korhonen11]. In the elderly, a fall is involved in 96 % of the hip fractures [Norton97] which are a significant economical burden [Bessette12]. Falls are mostly related to tripping or slipping, and occur most commonly while walking [Bentley98, Li06]. Means for preventing walking-related falls are, thus, needed. An understanding of the mechanisms underlying such falls is needed as well.

Falls can be prevented, for example, by building safer walking environments, both indoors and outdoors. This Thesis addresses fall prevention from the perspective of mechanics and neuromuscular control of walking. Individuals with a high fall risk can be rehabilitated so as to enhance their functional capacity, which, in turn, can help prevent falls. To do this rehabilitation efficiently, risk populations must first be screened for individuals with a high fall risk. Effective screening methods are, however, lacking. Also, similar methods are required for monitoring the outcomes of rehabilitation. Before conducting rehabilitation, mechanisms leading to, and preventing, a fall must be understood so that dysfunction of the neuromuscular system can be treated. This Thesis deals with metrics for stability against falling, effects of gait speed on stability, detection of an impaired stability, and neuromuscular control of stability in walking.

There is agreement that stable walking lowers the risk of falling. What, then, is stable walking? Answering this question has proven a challenge (even though the first-known biomechanical analysis of human walking has been conducted already by the Ancient Greek philosopher Aristotle [Klette08]). Muscles have to be activated in a coordinated manner to maintain the stability during walking. Yet, neural signals that activate muscles are themselves unstable [Kang09]. Moreover, the body is far from mechanical equilibrium during walking. Astonishingly, the central nervous system usually manages to control the body such that a fall is averted. Such facts pose a challenge for defining and quantifying stable walking. There is no commonly accepted way to define or quantify the stability of walking.

Several measures for the stability of walking have been introduced. For example, concepts of orbital and local stability have been used: Orbital stability describes how purely a periodic system responds to perturbations discretely from one step to the next in the case of walking. Local stability describes how a system responds to very small perturbations continuously in time. [Dingwell07] In addition, the concept of margin of stability, based on the inverted-pendulum model, has been introduced. The margin of stability is a measure of distance from the extrapolated center of mass to the boundary of the base of support. [Hof05] Also, the foot placement estimator has been introduced, which measures how well the foot is placed to restore the balance in walking [Millard09]. Relationship of these or other measures to falling, however, is not evident.

In order to directly assess stability against falling, invasive natural-type perturbations such as tripping or slipping, preferably leading to a fall, should be conducted. Such experiments involve, however, many ethical, psychological, and technical concerns. First of all, safety of the subjects is of top priority, hence equipment like a safety harness must be used to ensure safe falls. Awareness of safety can affect the way the subjects respond to the perturbation; compared with real-life circumstances, where a fall can be fatal, there is no actual need to avert a fall. Moreover, good research ethics require that the subjects are informed that the perturbation can occur at some point of the experiment. Because of such information, subjects can anticipate the perturbation. Their responses may not, thus, reflect responses to a really unexpected perturbation. A response to a single perturbation may not be descriptive of the subject's behavior, and adaptation can occur if the perturbation is repeated. In addition, the magnitude and timing of the perturbation have to be the same for all conditions and subjects in order to make the measure of stability standardized. This may be difficult to achieve. Simulation can resolve the above concerns.

Recent development in simulation tools has enabled fast creation of subject-specific muscle-driven simulations [Delp07]. Such simulations are created by combining measured gait data (kinematics and kinetics) of a subject with a musculoskeletal model. Gait data can be easily obtained in a subject-friendly fashion. The resulting simulations allow advanced and in-depth analyses of human movement as a whole or at a level of an individual muscle. These analyses include, but are not limited to, studying cause-effect relationships of muscle coordination and predicting what would happen if a perturbation took place. The framework of the muscle-driven simulation has been validated [Delp07, Liu08, Steele10].

This Thesis introduces a simulation-based method of a tripping perturbation for assessing the walking stability as it relates to falling after tripping. This method works such that a backward force is applied to the swing foot of the model, and an immediate passive response is observed. The passive response quantifies the stability of walking dynamics with respect to tripping, *i.e.*, to the trip-related fall risk.

Since there is no consensus about the definition of stability, neither is there a solid understanding about how the central nervous system realizes fall prevention and control of stability. This Thesis provides a proof that the whole-body angular momentum is not a good quantity for defining the stability against falling. Further, a rationale for choosing the trunk kinematics instead is given. Consequently, in this Thesis, control of stability is expected to be conducted through control of the trunk. Thus, neuromuscular control of the trunk is examined. To understand how the central nervous system coordinates muscles to control the trunk, forces of all muscles need to be known. Unfortunately, measuring the muscle forces in human subjects is not feasible. Muscle forces must thus be obtained from a muscle-driven simulation. Thesis reviews a study that identified, by means of a muscle-driven simulation, muscles that contribute to the trunk control in walking.

Chapter 2 deals with the choice by which the stability is defined. First, an analysis of a two-link system is provided to show that a whole-body angular momentum is not an eligible quantity for defining the stability against falling, even though some researchers have envisaged so. Second, the importance of control of the trunk is discussed.

Chapter 3 is devoted to the methodology of muscle-driven simulations. First, a musculoskeletal model is described. Second, steps to create a subject-specific muscle-driven simulation from experimental gait data are described. Third, the tripping-perturbation simulation is introduced and discussed in detail. Fourth, an induced-acceleration analysis to study muscle contributions is reviewed.

Chapter 4 reviews results for application of the tripping perturbation to study how gait speed affects the stability of walking. This information helps find out whether slowing down would ameliorate the stability and lower the risk of falling. Chapter 5 shows that the tripping perturbation detects an impaired stability in individuals with cerebral palsy.

Chapter 6 reports what muscles young healthy subjects were found to recruit for controlling the trunk.

2 DEFINITION OF THE STABILITY AGAINST FALLING

This Chapter deals with the choice by which the stability against falling should be defined. There is no agreement on how and by which quantity the stability against falling should be defined. Some researchers have argued that minimizing the whole-body angular momentum is the key to prevent a fall. Section 2.1 provides a mathematical proof that a human can fall even if its whole-body angular momentum is zero. This proof was accomplished by using a two-link system [PIV], and it indicates that the whole-body angular momentum is not a proper quantity for this task. In Section 2.2, the importance of minimizing the trunk motion is established through literature review and reasoning, suggesting that the stability against falling should be defined by trunk kinematics.

2.1 On the control of angular momentum

2.1.1 Introduction

The whole-body angular momentum has been reported to be small in the steady-state walking [Herr08]. Yet, it has been emphasized that regulation of the whole-body angular momentum is not a general feature across all human movement tasks, and that this regulation is not a necessary condition of stability [Herr08]. Nevertheless, a small whole-body angular momentum has been envisaged to indicate a good stability/balance and a low fall risk: Increasing change in the angular momentum has shown a negative correlation with balance in clinical tests among post-stroke hemiparetic patients [Nott14]. On the other hand, elderly fallers and non-fallers have exhibited similar behaviors as for the whole-body angular momentum [Simoneau00]. Pijnappels et al. [Pijnappels05] have concluded that the essence of preventing a fall after tripping is to reduce the whole-body angular momentum, yet their data have shown that a high angular momentum does not always result in a fall. Overall, the aforementioned studies have reported valuable experimental data on that angular momentum, but theoretical discussion of the meaning and importance of angular momentum has been limited. Therefore, the relation of angular momentum to balance and falling remains ambiguous.

If reducing the whole-body angular momentum prevents falling, a zero angular momentum can be conjectured to ultimately prevent a fall. In the case of a single rigid body, a vanishing angular momentum implies that the body is not rotating. In multi-segment systems, such as a human body, however, implications of a zero angular momentum are equivocal. A vanishing angular momentum can result from either of the following two situations: there is no rotation at all, or segments are rotating such that their angular momenta cancel each other out. It was shown mathematically that a human can fall even if its whole-body angular momentum is zero [PIV]. This was accomplished by solving the equations of motion for a two-link system in the case that the system's angular momentum about its center of mass is zero. It was shown that among the solutions to these equations, there exist solutions that yield a decreasing elevation of the center of mass. In general, a system literally falls if a decrease of its center-of-mass elevation is unbounded. From a mathematical perspective, finding a single solution with a zero angular momentum and an unbounded decrease of the center-of-mass elevation suffices to prove that the notion of a zero angular momentum preventing a fall is not true in general.

2.1.2 Theory and methods

The sagittal-plane motion of a human body was analyzed by a planar two-link system. A schematic illustration of the system is shown in Fig. 1. Link 1 represents the shank and thigh rotating about the ankle joint. Link 2 represents the pelvis and trunk rotating about the hip joint. The inertial properties and anthropometry for the two-link system were obtained from [Delp90] and [Anderson99]. When the lower end of link 1 lies in the origin of a fixed frame of reference, its center-of-mass placement is given by

$$\vec{r_1} = R_1 \cos\theta_1(t)\hat{x} + R_1 \sin\theta_1(t)\hat{z}, \tag{1}$$

where $\theta_1(t)$ is the angle between link 1 and the x axis (*i.e.*, the horizontal coordinate axis) as a function of time t, R_1 is the center of mass of link 1 along the link measured from its lower end, and $\hat{x}$ and $\hat{z}$ are the horizontal and vertical unit vectors, respectively. The center-of-mass placement of link 2 is similarly given by

$$\vec{r_2} = [L_1 \cos\theta_1(t) + R_2 \cos\theta_2(t)]\,\hat{x} + [L_1 \sin\theta_1(t) + R_2 \sin\theta_2(t)]\,\hat{z}, \tag{2}$$

where $\theta_2(t)$ is the angle between link 1 and the x axis, R_2 is the center of mass of link 2 along the link measured from its lower end, and L_1 is the length of link 1. The placement of the two-link center of mass can be expressed in the form

$$\vec{r}_{CM} = \frac{m_1\vec{r_1} + m_2\vec{r_2}}{m_1 + m_2}, \tag{3}$$

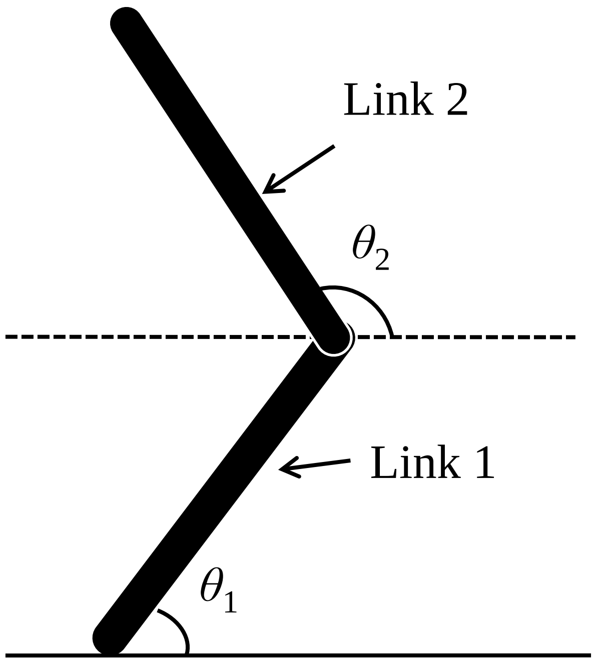

FIGURE 1 A schematic illustration of the two-link system. Link 1 represents the shank and thigh rotating about the ankle joint. Link 2 represents the pelvis and trunk rotating about the hip joint.

where m_1 and m_2 denote the masses of link 1 and link 2, respectively. Differentiation of Eqs. (1), (2) and (3) with respect to time, t, yields the respective velocities ($\vec{v}_1$, $\vec{v}_2$ and $\vec{v}_{CM}$) of the centers of mass. The angular momentum of the two-link system about its center of mass is given by

$$\vec{H} = \sum_{i=1}^{2} \left[I_i \dot{\theta}_i + m_i (\vec{r}_i - \vec{r}_{CM}) \times (\vec{v}_i - \vec{v}_{CM}) \right], \tag{4}$$

where the first term inside the sum corresponds to the angular momentum of link i about its center of mass, I_i and $\dot{\theta}_i$ the moment of inertia and angular velocity of link i, respectively, and the second term inside the sum corresponds to the angular momentum of m_i about the two-link center of mass. In the case that $\vec{H} = \vec{0}$, Eq. (4) is reduced to the form

$$\sum_{i=1}^{2} \left[I_i \dot{\theta}_i + m_i (\vec{r}_i - \vec{r}_{CM}) \times (\vec{v}_i - \vec{v}_{CM}) \right] = 0. \tag{5}$$

Substituting Eqs. (1), (2), (3), and their time-derivatives in Eq. (5) yields a first-order nonlinear differential equation for dependent variables θ_1 and θ_2. In order to solve this differential equation, one of the dependent variables needs to be predetermined. Therefore, θ_2 was determined to be

$$\theta_2 = \omega_2 t + \phi, \tag{6}$$

where ω_2 is constant (angular velocity) and ϕ is an initial angle. Subsequently, the differential equation for θ_1 was solved numerically by Matlab as an initial-value problem. The initial condition was an upright posture, *i.e.*, $\theta_1(0) = \theta_2(0) = \phi = \frac{\pi}{2}$. A solution was obtained for three different choices for ω_2 so as to demonstrate that existence of the solution does not rely on the choice of ω_2. The solutions for θ_1 were then substituted in Eqs. (1), (2) and (3) to see whether these solutions result in an unbounded decrease of the center-of-mass elevation, z_{CM}, of the two-link system.

A numerical solution, in general, is never exact, but it is an approximation. To verify the accuracy of the present numerical solutions of the differential equation, convergences of the solutions were tested by changing the length of the time step of the integration. The values of θ_1 converged with a precision of 1×10^{-13} rad. Thus, it is evident that these solutions are accurate.

2.1.3 Results and discussion

A set of solutions was found to the equations of motion of the two-link system with a zero angular momentum. These solutions are shown in Fig. 2. They yield an unbounded decrease of the two-link center-of-mass elevation, even in an accelerating manner, as illustrated in Fig. 3. These results indicate that if link 2 (*i.e.*, the upper extremity) rotates with a constant angular velocity, link 1 (*i.e.*, the lower extremity) rotates in the opposite direction with an increasing angular velocity so that the zero whole-body angular momentum is maintained. Of course, there are plenty of other solutions as well, but the objective was to show the existence of such a solution. In order to solve the equations of motion, the angular velocity of link 2 was predetermined to be constant, ω_2. Varying ω_2 over time would, naturally, affect the solution. The presented solutions, nonetheless, show that the notion that a zero whole-body angular momentum prevents a fall is not true in general. A human body was modeled as a two-link system. This is not a limitation: a human can act as a two-link system (*i.e.*, the knee joint is locked, and rotation occurs in the ankle and hip joints), so the solutions reported are relevant.

Emphasis was not to show or claim that a zero whole-body angular momentum always results in falling, but that it does so in some circumstances (as in the cases presented). Control of the whole-body angular momentum may play a role in the prevention of an impending fall, but minimizing the angular momentum does not alone suffice to prevent a fall in general. This suggests that the whole-body angular momentum should be used with caution when analyzing the balance and fall risk. More comprehensive control criteria for the prevention of a fall must be searched for.

2.2 On the control of the trunk

Control of the trunk is physiologically related to control of the overall balance as follows: minimization of the trunk motion reduces acceleration of the head, which allows stabilization of the optic flow, a more effective processing of the

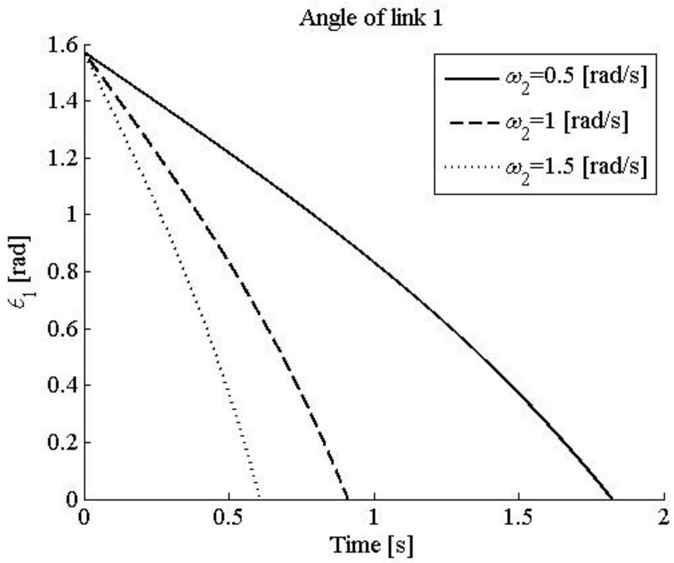

FIGURE 2 Solutions of the two-link system, which have a zero angular momentum. Angle of link 1, θ_1, as a function of time is shown for three values of the angular velocity of link 2, ω_2. These results indicate that if link 2 (*i.e.*, the upper body) rotates with a constant angular velocity, link 1 (*i.e.*, the lower body) rotates in the opposite direction with an increasing angular velocity so that the zero whole-body angular momentum is maintained.

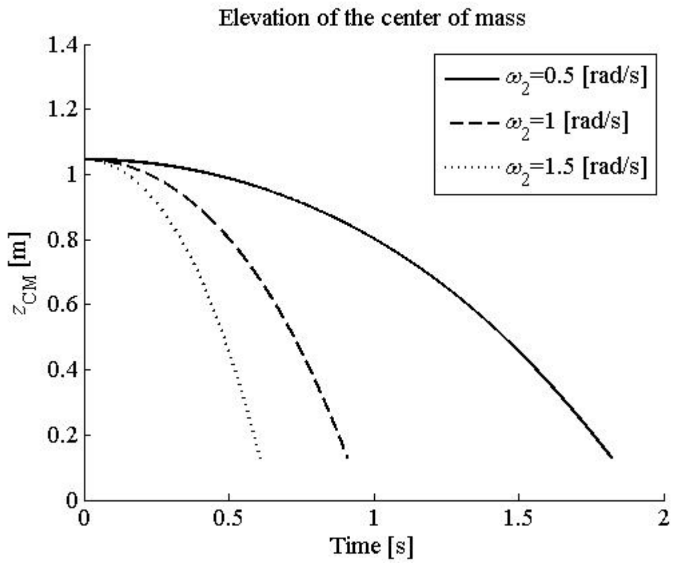

FIGURE 3 Elevation of the center of mass of the two-link system as a function of time for three values of the angular velocity of link 2, ω_2, with a zero angular momentum of the whole system.

vestibular system signals, and a consequent control of balance [Cappozzo81, Mazza08]. From a mechanical point of view, the trunk is massive and its elevation is high, so it stores a large amount of gravitational potential energy. Thus, the trunk must be controlled in order to avert transformation of the potential energy to kinetic energy, *i.e.*, falling. Control of the whole-body center of mass or angular momentum, for instance, does not guarantee such physiological and mechanical aspects of balance control.

Stability of the trunk during walking has been reported to be associated with fall history [Toebes12]. Moreover, ability to limit the trunk motion after a trip has been shown to distinguish non-fallers from fallers [Pavol01]. An impaired control of the trunk during walking is associated with aging and many movement disorders. This has been demonstrated as an increased motion of the trunk, for example, in the elderly [Goutier10], Parkinson's disease [Cole10, Adkin05], myotonia congenita [Horlings09], spinocerebellar ataxia [van de Warrenburg05], and multiple sclerosis [Spain12]. In the light of the above discussion, an impaired control of the trunk is a concern since it physiologically and mechanically increases the fall risk.

Angular motion of the trunk definitely contributes to the whole-body angular momentum, but the whole-body angular momentum is not fully or always related to the motion of the trunk. Therefore, the motion of the trunk is a more reliable and sensitive measure for stability. Stability against falling should thus be defined as minimization of the trunk motion.

3 MUSCLE-DRIVEN SIMULATIONS

This Chapter describes the framework of muscle-driven simulations, which is the core methodology in this Thesis. Muscle-driven simulations are the basis of the tripping perturbation, which was used to study effects of gait speed on the stability of walking [PI] and to detect an impaired stability of walking [PIII, Klemetti13]. In addition, neuromuscular control of the trunk was examined by means of muscle-driven simulations [PII]. Section 3.1 provides an introduction to muscle-driven simulations. Section 3.2 describes a generic musculoskeletal model that was used for the simulations reviewed below. Then, procedures for creating and testing a subject-specific muscle-driven simulation are described in Section 3.3. In Section 3.4, a novel method of a tripping perturbation is introduced. Section 3.5 explains use of an induced-acceleration analysis to study the neuromuscular control of the trunk.

3.1 Background

Components of the neuromusculoskeletal system interact with each other so as to generate a coordinated movement. Scientists have performed plenty of studies to describe these components. Resulting data characterize the muscle mechanics, the geometric relationships between bones and muscles, and the joint motions. Clinicians have studied the movement kinematics and neuromuscular excitations of thousands of patients, both before and after treatment interventions. Nevertheless, synthesizing descriptions of the components of the neuromusculoskeletal system, with measurements of movements for establishing an integrated understanding of the normal movement as a basis for restoring an abnormal movement, remains a challenge. [Delp07]

Using experiments alone for understanding the dynamics of movement has two fundamental limitations. First, variables such as muscle forces are not generally measurable. Second, establishing cause-effect relationships in complex systems from experimental data alone is difficult. Consequently, revealing functions

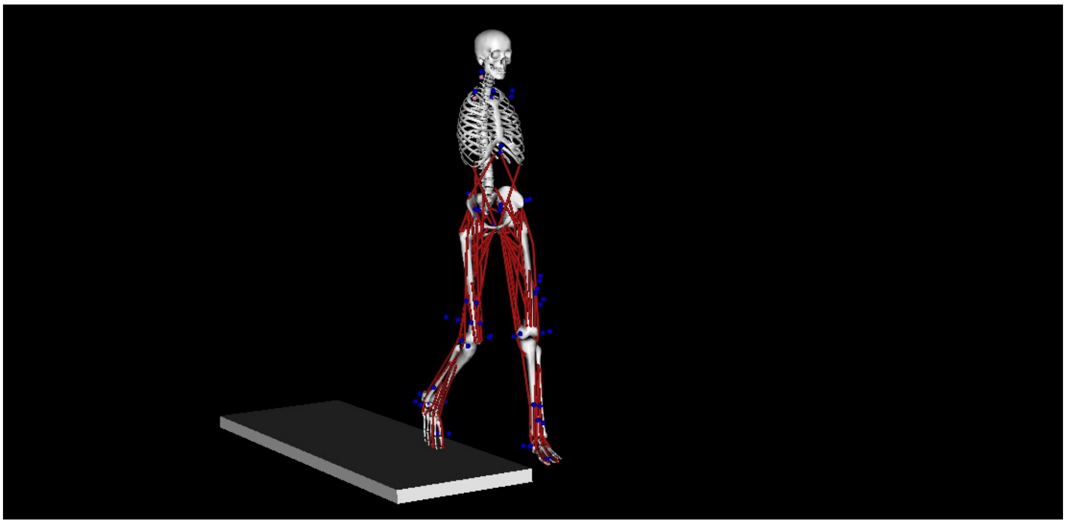

FIGURE 4 The musculoskeletal model used to create muscle-driven simulations. The model has 23 degrees of freedom and 92 muscle-tendon actuators. The image of the model was created with OpenSim [Delp07].

of muscles from experimental data is not straightforward. For example, electromyographic (EMG) recordings indicate when muscles are active, but these recordings do not enable to determine which motions of the body arise from the muscles' activity. Determination of how individual muscles contribute to motions of the body is difficult, since a muscle can accelerate body segments to which it does not attach and joints that it does not span. [Delp07]

In the light of the above, a theoretical framework is required to reveal the principles that govern the coordination of muscles during normal movement, to determine how neuromuscular impairments are related to abnormal movement, and to predict outcomes of treatments. To this end, the theoretical framework has to reveal cause-effect relationships between neuromuscular excitations, muscle forces, and the body motions. Such a framework is provided by dynamic simulation of a movement that integrates models which describe the anatomy and physiology of the neuromusculoskeletal system and the dynamics of multijoint movement. Muscle-driven simulations complement experiments by providing estimates for variables, such as muscle forces, which are difficult to measure. Such simulations also allow identification of cause-effect relationships and making predictions. [Delp07]

In order to create a muscle-driven simulation of walking, one must have a dynamic model of the musculoskeletal system. Such a model accounts for muscle-contraction dynamics, musculoskeletal geometry, and body-segmental dynamics. Once a dynamic musculoskeletal model is formulated, muscle excitations that produce a coordinated movement can be found. Excitations can be found by solving an optimization problem in which the objective of a movement is defined or in which the objective is to track experimental motion data. Simulations are evaluated by how well they match measured experimental data such as kinematics, kinetics, and EMG data. [Delp07] After testing a simulation's accuracy, it can be analyzed to study contributions of muscles to motion of the body

[Delp07] and its response to a perturbation.

Determination of muscle excitations that produce a coordinated movement is one of the biggest challenges in creating a muscle-driven simulation. Conventionally, the computational cost of creating a coordinated muscle-driven simulation of movement has been high, requiring computer time from days to months. Recent development in the application of robotic control techniques to biomechanical simulation has drastically reduced the computer time. For example, a computed muscle-control algorithm determines muscle excitations that track measured pedaling dynamics in ten minutes, which is two orders of magnitude faster than conventional dynamic optimization techniques. A simulation of a half-gait cycle has been created in thirty minutes. Speed of the computed muscle control makes it practical to create subject-specific muscle-driven simulations. [Delp07]

3.2 Musculoskeletal model

A generic musculoskeletal model (lower extremity from [Delp90] and the trunk from [Anderson99]) was used in the simulation analyses described below. This model has 23 degrees of freedom and 92 muscle-tendon actuators. The model is illustrated in Fig. 4. Procedures that have been used to create the model are described below.

To acquire the bone surface data, bone surfaces have been marked with a mesh of polygons, and the coordinates of the vertices have been determined by a three-dimensional digitizer. These coordinates have been used to display the pelvis, thigh, shank, and foot bones. Paths of musculotendon actuators (*i.e.*, the lines of action) have been defined based on the anatomical landmarks of the bone-surface models. Each musculotendon path has been represented as a series of line segments. In some cases, a muscle path has been described by the origin and insertion of the muscle. For example, soleus has been represented by a single line segment. Intermediate points have been introduced in other cases, where a muscle wraps over bone or is constrained by retinacula. For example, peroneus longus has been represented by a series of six line segments. [Delp90]

The lower extremity has been modeled as seven rigid-body segments: pelvis, femur, patella, tibia/fibula, talus, foot (consisting of the calcaneus, navicular, cuboid, cuneiforms, and metatarsals), and toes, with reference frames fixed in the segments. Relative motions of these segments have been defined by models of the hip, knee, ankle, subtalar, and metatarsophalangeal joints. The hip joint has been modeled as a ball-and-socket joint. Thus, transformation between the pelvic and femoral frames of reference has been determined by three successive rotations of the femoral frame about the femoral head. [Delp90]

A planar model of the knee has been modified so as to characterize the knee extensor mechanism. This model has one degree of freedom, and it accounts for kinematics of both the tibiofemoral and patellofemoral joints in the sagittal plane as well as the patellar levering mechanism. Transformations between the femoral, tibial, and patellar referance frames have been specified as functions of the knee angle. Tibiofemoral kinematics has been determined in the following way. The femoral condyles have been described as ellipses; the tibial plateau has been described as a line segment. Transformation from the femoral frame of reference to the tibial frame of reference has been determined such that the femoral condyles remain in contact with the tibial plateau during the entire range of knee motion. Tibiofemoral contact point depends on the knee angle and has been specified according to experimental data. Length of the patellar ligament has been assumed to be constant. Consequently, the angle between the patellar ligament and the tibia determines the translation vector from the tibial frame of reference to the patellar frame of reference. Rotation of the patella with respect to tibia has been specified according to experimental data on patellar rotation. [Delp90]

The ankle, subtalar, and metatarsophalangeal joints have been modeled as frictionless revolute joints [Delp90]. Degrees of freedom between the pelvis and trunk have been modeled as a ball-and-socket joint [Anderson99].

In order to compute the musculotendon force as a function of musculotendon length, a specific model for each musculotendon actuator has been formulated. Each model has been formed from a generic model that accounts for the static properties of a muscle and a tendon. Each musculotendon actuator's force-length relation has been computed by scaling the generic model by a peak isometric force, optimal muscle-fiber length, pennation angle, and tendon-slack length. Physiological cross-sectional areas, muscle-fiber lengths, and pennation angles have been taken from previous literature, while tendon-slack lengths have been estimated. [Delp90]

Combining the geometric musculoskeletal, joint, and musculotendon models described above has enabled computation of the force and the joint moment that each muscle can develop for any position of the body. For a given position of the body, musculotendon lengths and moment arms have been computed. Then, the maximum isometric muscle forces at the computed musculotendon lengths have been computed. The joint moments for muscles have been computed as products of the tendon forces and moment arms. It has been found that the active joint moments computed, exerted by all muscles, are in good agreement with the measured ones. [Delp90]

A sensitivity analysis has been performed to see how the musculotendon parameters and musculoskeletal geometry affect the muscle force. The joint angle at which a muscle produces peak force has been reported to depend on the slack length and optimal muscle-fiber length of the tendon. It has been found that the angle of peak force is more sensitive to change in the tendon length for actuators with a high ratio of tendon length to moment arm (e.g., gastrocnemius) than for those with a low ratio (e.g., gracilis). Similarly, the angle of peak force is more sensitive to change in the optimal muscle-fiber length for actuators with a high ratio (e.g., gracilis) of fiber-length to moment arm than for those with a low ratio (e.g., gastrocnemius). In general, the angle of peak force is more sensitive to the tendon-slack length than to muscle-fiber length for most actuators. It has been noted that estimates of the tendon slack length are correct for muscles with a high ratio of tendon slack length to moment arm. [Delp90]

It must be noted that the musculoskeletal geometry and musculotendon parameters have been specified only for one nominal subject [Delp90].

3.3 Creation of muscle-driven simulations

The basis of creating a subject-specific muscle-driven simulation is to combine a musculoskeletal model with experimental kinematics and kinetics of a subject's movement so as to find the muscle-excitation patterns that produced that movement. In addition, measured EMG patterns can be used to fine-tune simulated muscle activations. Subject-specific muscle-driven simulations studied in this Thesis have been created by the OpenSim software [Delp07]. In OpenSim, the process of creating a muscle-driven simulation involves four steps. These steps are: scaling, inverse kinematics, residual reduction algorithm, and computed muscle control. These steps are described in more detail below. A flow chart of creation and analyses of muscle-driven simulations in this Thesis is shown in Fig. 5.

In the scaling-step, a musculoskeletal model is scaled based on each subject's anthropometry. In this scaling, the segmental-mass properties, muscle-fiber lengths, and tendon-slack lengths are also scaled accordingly. In the inverse-kinematics step, kinematics of the scaled model is determined by tracking experimental marker data obtained from motion capture. Because of measurement errors in experiments and modeling-assumptions, model kinematics and measured ground reaction forces are often dynamically inconsistent. The residual reduction algorithm reduces this dynamic inconsistency by applying residual forces and torques to the pelvis of the model, and adjusting the model's mass properties and kinematics. [Delp07] Computed muscle control is used to determine the muscle excitations that, in concert with the ground reaction forces, generate a forward dynamic simulation of each subject's gait pattern [Thelen03, Thelen06]. Computed muscle control uses a static optimization criterion for distributing muscle forces across synergist muscles. Further, a proportional derivative control generates the forward dynamic simulation that tracks the kinematics derived by the residual reduction algorithm. This simulation includes first-order activation and contraction dynamics of the muscles. [Delp07]

3.4 Tripping perturbation

Balance can be controlled in a proactive and reactive manner [Wang12]. A reactive control means that the central nervous system reacts to a perturbation in

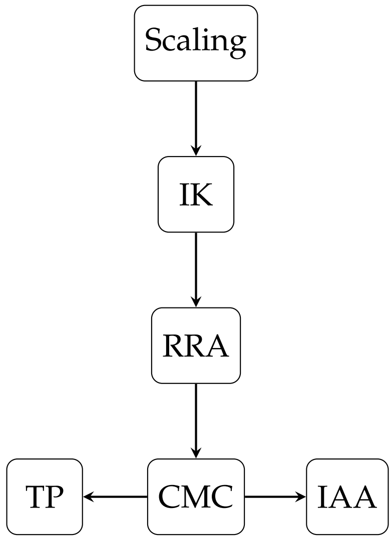

FIGURE 5 A flow chart of creation and analysis of muscle-driven simulations in this Thesis. Scaling, inverse kinematics (IK), residual reduction algorithm (RRA), computed muscle control (CMC), induced acceleration analysis (IAA), and tripping perturbation (TP) are shown.

order to minimize the impact of this perturbation. A proactive control means in turn that the central nervous system makes walking dynamics stable against a possible perturbation before and without being aware of the impending perturbation. One way to test such stability is to apply a short perturbation for probing the mechanical state of the system. In the studies reviewed in this Thesis, probing was carried out by the method of a simulated tripping perturbation [PI, PIII, Klemetti13]. This tripping perturbation is a forward-dynamics analysis, and it works so that in a subject-specific muscle-driven simulation of walking, created from experimental gait data, a force is applied to the swing foot, and resulting changes in the trunk kinematics are observed.

Advantages of the tripping perturbation are that it is standardized for all conditions and subjects, ensuring their fair comparison, and that it can be done and repeated free of any anticipation or adaptation by the subject. Tripping perturbation is also ideal for studying a proactive control because it avoids the influence of the reactive control. Moreover, tripping experiments are invasive, require special equipment, and are time consuming. Thus, they have a limited value in large-scale clinical applications. A tripping-perturbation simulation complements and provides to some extent an alternative for tripping experiments. Tripping perturbation is built on the OpenSim software [Delp07] which is already in clinical use. A tripping-perturbation simulation predicts a subject's passive response to tripping before the first spinal reflexes, thus testing the stability of walking dynamics with respect to a tripping force.

A tripping-perturbation simulation is generated by running a subject-specific muscle-driven forward dynamic simulation of an unperturbed walking, using the computed muscle excitations and the experimental ground-reaction-force data, while applying a time-invariant additional external force to the swing foot of the model. This tripping force is applied to the center of mass of the swing foot's cal-

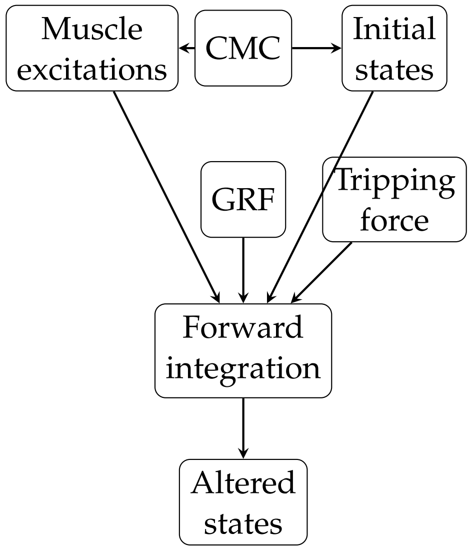

FIGURE 6 A flow chart of the tripping perturbation. States of the muscles and joint kinematics computed with computed muscle control (CMC) serve as initial states for a tripping-perturbation simulation. The equations of motion of the model are integrated forward having computed muscle excitations, experimental ground reaction forces (GRF), and a tripping force as input variables. This forward integration yields altered states.

caneus segment, and is directed horizontally backward. Magnitude of the force is a matter of choice. Magnitudes of 5-15 % of a subject's body weight were used. These magnitudes are similar, but are not meant to be identical to observed magnitudes of backward forces between a tripping foot and an obstacle in an actual tripping of a human [Pijnappels04]. Time evolution of the tripping force was not considered as the objective was to study stability against a perturbing force, similar in magnitude to a real tripping force, but not the actual tripping.

In a single tripping-perturbation simulation, the model's equations of motion are integrated forward for a short period t_t. Note that the tripping perturbation force is applied during the entire t_t. In order for the simulations to be physiologically relevant, t_t cannot be longer than about 20 ms, because then spinal reflexes take over the muscle excitations, and subsequently the force production. For instance, the latencies of deep tendon reflex of rectus femoris and soleus, at the age of 10 yrs, are 13 ms and 25 ms, respectively [Pereon04], and the electromechanical delay between the onset of muscle excitation and the force production is 13-15 ms [Nicol98].

The initial state for each tripping-perturbation simulation is the state of the unperturbed simulation at a given instant of time. This analysis is repeated at 1 ms intervals throughout the swing phase. A batch of the tripping-perturbation simulations was run by using a program written in Matlab (MathWorks, Inc.), which automatically executed OpenSim [Delp07]. Program for the batch simulations is freely available at www.simtk.org. A flow chart of the tripping perturbation is shown in Fig. 6.

3.5 Induced-acceleration analysis

Leonardo da Vinci has stated: "it is indispensable for a painter to become totally familiar with the anatomy of nerves, bones, muscles, and sinews, such that he understands their various motions and stresses, which sinews or which muscles cause a particular motion." [Klette08]

Contributions of individual muscles to the angular acceleration of the trunk in walking were addressed [PII]. It is not obvious what muscles contribute significantly to the motion of the trunk since every muscle affects more or less everything because of inter-segmental coupling. In order to determine the contributions of muscles, forces from all muscles must be known. Collecting such data experimentally is not feasible. Consequently, muscle forces must be obtained from a muscle-driven simulation.

To this end, the equations of motion of the model need to be solved. Solving these equations is not straightforward as, along with muscle forces, also ground-reaction forces, gravity, and Coriolis and centrifugal forces accelerate the trunk. Problem is that ground-reaction forces are not independent of the other aforementioned forces, but are literally reaction forces to them. One approach to resolve this problem is a perturbation analysis [Liu08]. In the perturbation analysis, the force of each muscle is augmented in turn, and resulting changes in the kinematics are observed. This procedure reveals the muscle contributions. Another approach is an induced-acceleration analysis [Riley99, Hamner10]. The results of this Thesis were obtained by the latter method.

In the induced-acceleration analysis, ground-reaction forces are replaced by kinematic constraints which model the foot-ground contact. A rolling-without-slipping constraint was used in the present analyses [Hamner10]. This means that foot is assumed to roll without slipping on the ground, and the ground-contact forces are calculated accordingly. Kinematic constraints enable solving the equations of motion of the model so as to find muscle contributions to contact forces, and, subsequently, to the angular acceleration of the trunk.

4 GAIT SPEED AND STABILITY

This Chapter deals with effects of gait speed on stability of walking as revealed by the tripping perturbation. This information is an addition to understanding the association of gait speed with falling, and can help prevent falls. Subject-specific muscle-driven simulations of young healthy subjects walking over a range of gait speeds were analyzed by means of the tripping perturbation [PI]. First, this Chapter provides an introduction to the enigma of association of walking speed with stability and falling after tripping. Second, specific details on the subjects and performed analyses are given. Third, results of the tripping perturbations are reviewed and discussed.

4.1 Introduction

Slow gait speeds are common in the elderly, and more pronounced in the elderly with a movement impairment [Alexander96]. It is unknown why the elderly select a slow walking speed. It may be that an age-related deterioration and other impairments of the function of their neuromusculoskeletal system prevents the elderly from walking faster. On the other hand, the neuromusculoskeletal system does not necessarily limit the walking speed, but the elderly may voluntarily slow down for some reason. Some studies have reported that, among the elderly, slow walkers tend to fall frequently [Bergland03]. It is unclear, however, whether the slow walking speed as such is a risk factor for falling or just results from the deteriorated function of the neuromusculoskeletal system which is in itself a risk factor for falling.

Walking fast has been shown to be associated with a high likelihood of falling due to tripping, and it may be the greatest cause of falling following a trip in healthy older adults [Pavol01]. One possible explanation for such causality is instability of fast walking. Inherent local dynamic instability, quantified by local divergence exponents and Floquet multipliers, has indeed been demonstrated to increase with walking speed in younger and older adults [Kang08]. Nonetheless, local divergence exponents have displayed different speed effects in different directions of motion, suggesting that slow walking is not necessarily more stable than fast walking [Bruijn09]. Such measures of stability, however, only quantify responses to local (*i.e.*, very small) intrinsic perturbations, and results may not extend to responses to large perturbations like tripping (*i.e.*, global stability) [Kang08]. Therefore, instability of fast walking remains equivocal.

4.2 Analyses

Previously created subject-specific muscle-driven simulations of walking were studied by the tripping perturbation [PI]. Eight healthy subjects walking at four speeds were examined. Their ages ranged from 7.0 to 18.0 yrs with a mean of 12.9 yrs. Each subject's walking trials were categorized post-hoc as follows: very slow, slow, free, and fast speed. Average walking speeds were 0.54, 0.72, 1.15, and 1.56 m/s for a very slow, slow, free, and fast speed, respectively. One step of each subject at each speed was analyzed.

Magnitude of the tripping force was 10 % of the subject's body weight. In addition, magnitudes of 5 and 15 % of the body weight were used for two subjects so as to find if results were force dependent. The period of the tripping perturbation, t_t, was 20 ms. To quantify responses to the tripping perturbation, changes in the velocity ($\vec{v}$) of the trunk center of mass and changes in the angular velocity ($\vec{\omega}$) of the trunk were computed:

$$\Delta\vec{v}(t) = \vec{v}_p(t + t_t) - \vec{v}_u(t + t_t) \tag{7}$$

and

$$\Delta\vec{\omega}(t) = \vec{\omega}_p(t + t_t) - \vec{\omega}_u(t + t_t), \tag{8}$$

where subscript p refers to a perturbation simulation, u refers to an unperturbed simulation, and t corresponds to time at which tripping was started. In other words, at a given instant of time, value of the unperturbed simulation was subtracted from the endpoint value of the perturbation simulation at that instant of time. $\vec{v}$ and $\vec{\omega}$ were represented in the coordinate system used in OpenSim: x was forward, y up, z right, and rotations were expressed about the respective axes according to the right-hand rule.

The absolute maximum of each component of $\Delta\vec{v}$ and $\Delta\vec{\omega}$ during the swing phase (*i.e.*, the extremity of each curve) for each trial was computed so as to find the most unstable point of each component. To see if the absolute maxima depended on the body size, a correlation analysis was performed: it was verified graphically that the both absolute maxima and leg lengths were approximately normally distributed, so Pearson's correlation analysis was used. Since there are natural correlations between these variables, absolute maxima were normalized to unity to allow comparison of subjects. A correlation analysis between these normalized values and walking speed was performed. In this analysis, walking speeds were expressed relative to each subject's free walking speed. Again, the

	Δv			$\Delta\omega$		
	x	y	z	x	y	z
r	0.3657*	0.4054*	-0.3999*	0.0069	0.5186**	-0.0580
Fast	0.95±0.06	0.93±0.08	0.71±0.09	0.76±0.24	0.92±0.06	0.92±0.09
Free	0.85±0.17	0.91±0.12	0.79±0.18	0.74±0.24	0.82±0.27	0.93±0.08
Slow	0.83±0.15	0.86±0.09	0.79±0.13	0.75±0.22	0.66±0.27	0.95±0.05
Very slow	0.80±0.23	0.83±0.10	0.92±0.19	0.77±0.20	0.63±0.23	0.90±0.06

TABLE 1 Effects of gait speed on normalized absolute maxima of changes in the components of linear and angular velocities of the trunk. Pearson's correlation coefficient is denoted by r. Statistical significance is indicated as follows: * refers to $p < 0.05$, ** refers to $p < 0.01$, and a result without an asterisk is not statistically significant. Mean values of the normalized absolute maxima, and their sample standard deviations, for each speed category are shown in the form mean ± standard deviation.

normalized maxima and the relative walking speeds were approximately normally distributed, so Pearson's correlation was justified. Also, for each speed category, subjects' group averages of the normalized absolute maxima, and their sample standard deviations were computed. Statistical analyses were run with the Matlab statistics toolbox.

4.3 Results

Figures 7 and 8 depict simulated responses to the tripping perturbation at different walking speeds for one subject, when the time of tripping perturbation was varied and the perturbing force was constant. All components of $\Delta\vec{v}$ and $\Delta\vec{\omega}$ for all subjects displayed both positive and negative values, with few exceptions. At fast speed, three subjects exhibited very high Δv_xs, when perturbation was performed at late swing, as exemplified by the subject presented in Fig. 7. Figures 9 and 10 illustrate for all subjects the absolute maxima of components of $\Delta\vec{v}$ and $\Delta\vec{\omega}$ for a constant perturbing force at different speeds, providing evidence that no clear effects of speed were identified. Correlation coefficients and group averages of the normalized absolute maxima are shown in Table 1. These results demonstrate that normalized maxima of components of $\Delta\vec{v}$ and $\Delta\vec{\omega}$ were only weakly correlated to walking speed. Table 2 shows that the absolute maxima of $\Delta\vec{v}$ and $\Delta\vec{\omega}$ were proportional at all speeds to the magnitude of the force applied, so all forces reveal a similar effect of speed on stability.

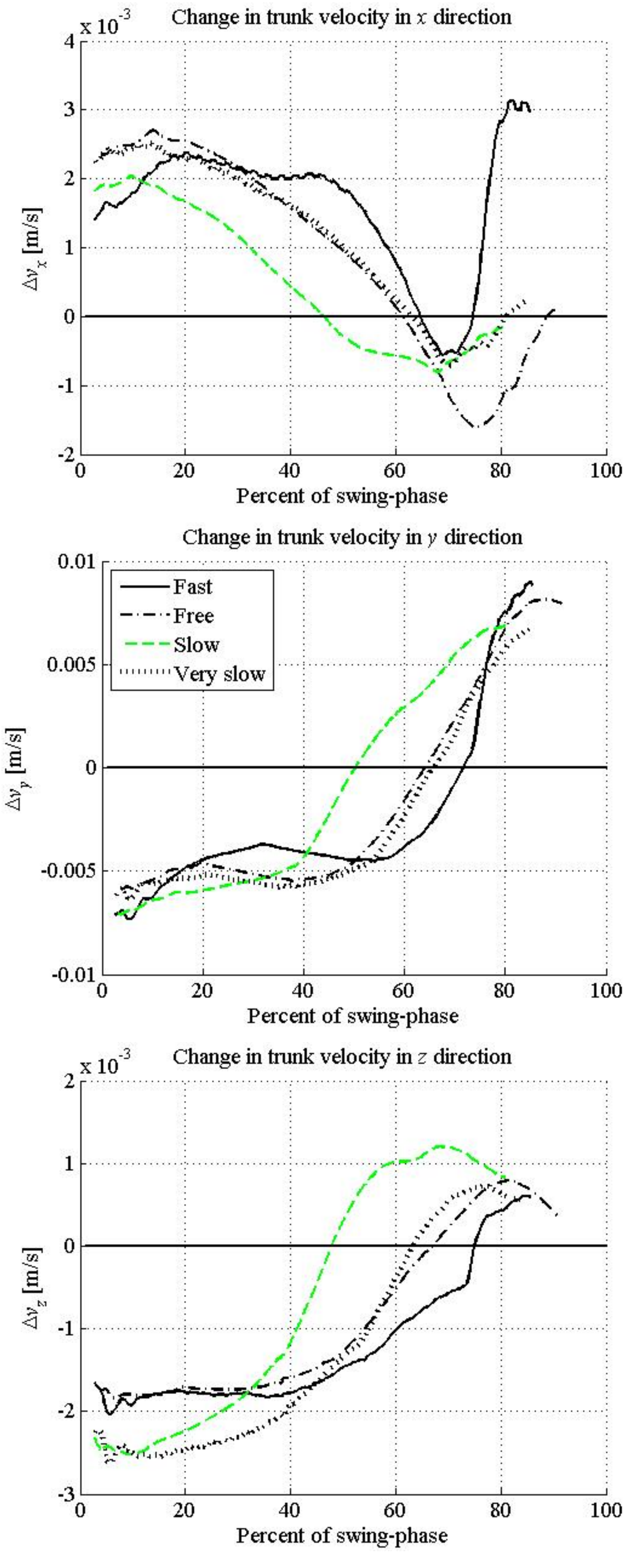

FIGURE 7 Changes in the linear velocity of the trunk as a function of timing of the tripping perturbation for one subject walking at a very slow, slow, free, and fast speed, when the left leg was tripped. Each data point represents a result of a single tripping simulation, and the entire curve the results of such simulations for varying the time of application of the tripping force during the swing phase.

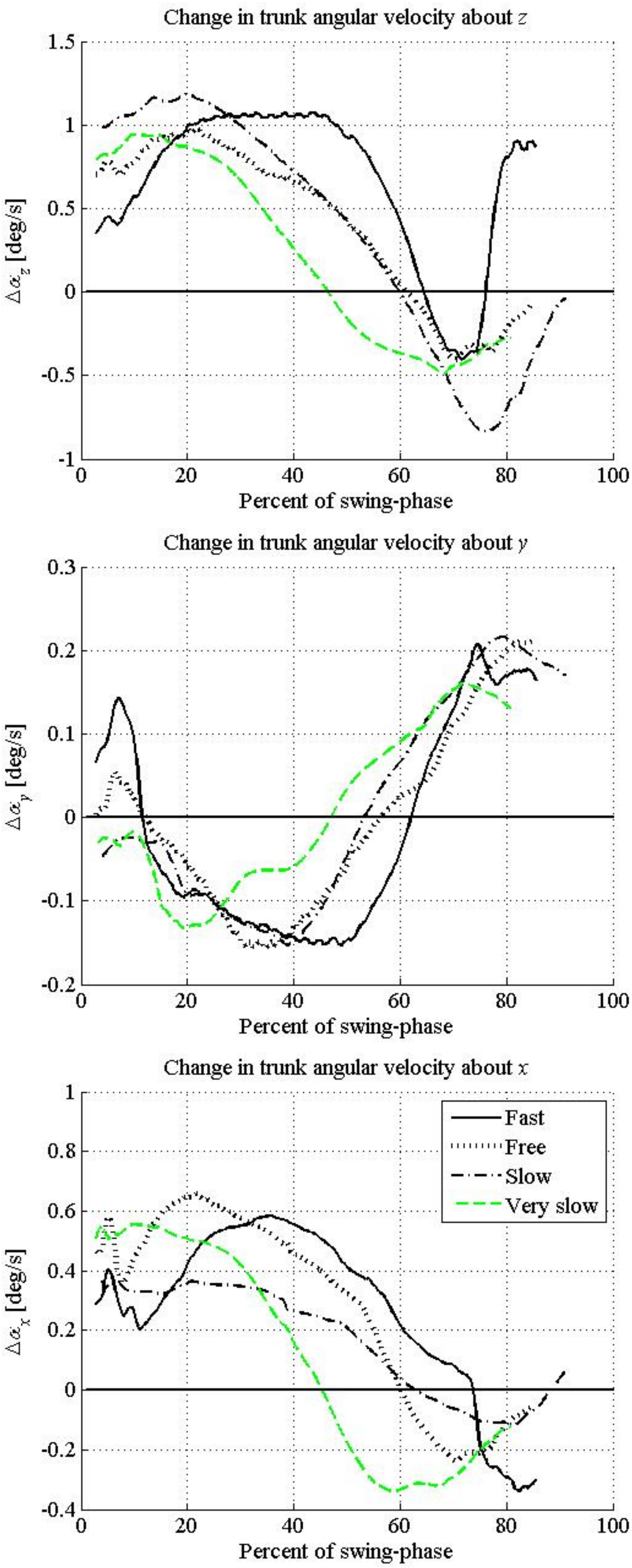

FIGURE 8 Changes in the angular velocity of the trunk as a function of timing of the tripping perturbation for one subject walking at a very slow, slow, free, and fast speed, when the left leg was tripped. Each data point represents a result of a single tripping simulation, and the entire curve the results of such simulations for varying the time of application of the tripping force during the swing phase.

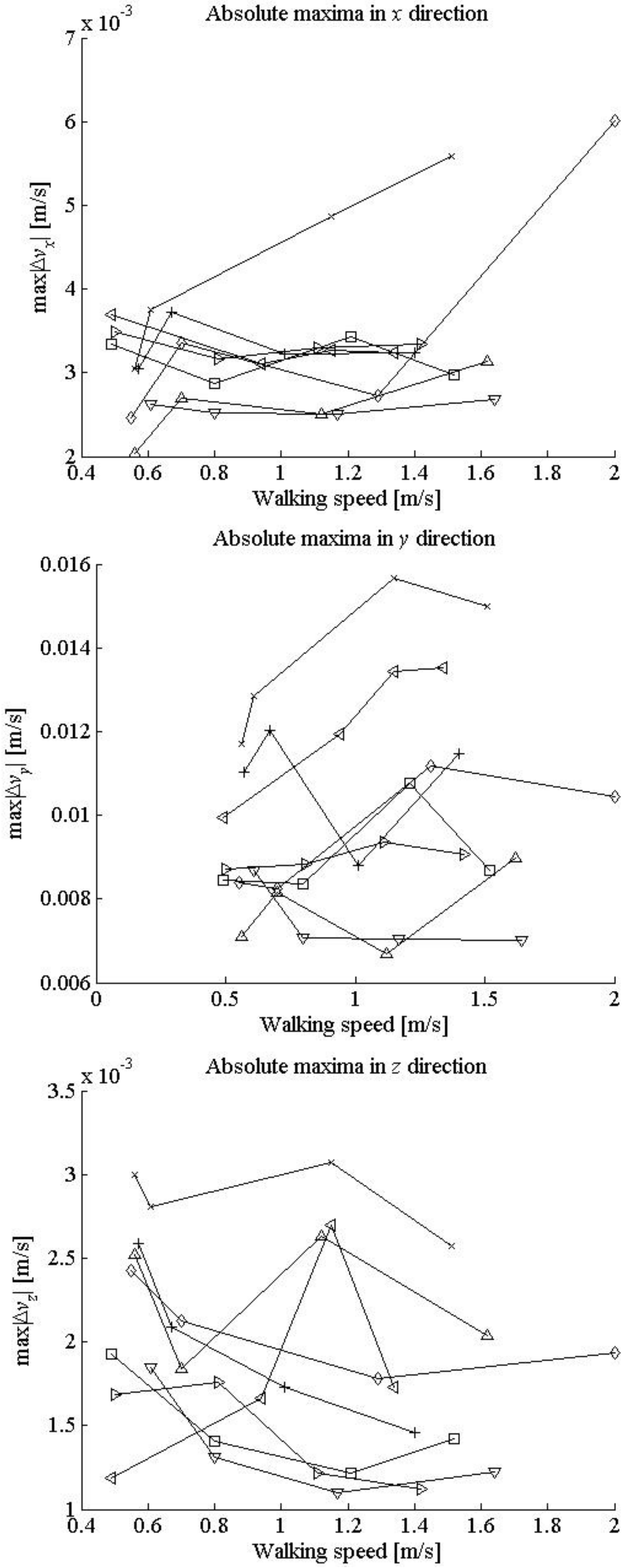

FIGURE 9 Absolute maxima of changes in the linear velocity of the trunk at different speeds for all subjects. Subjects are indicated by different marker types.

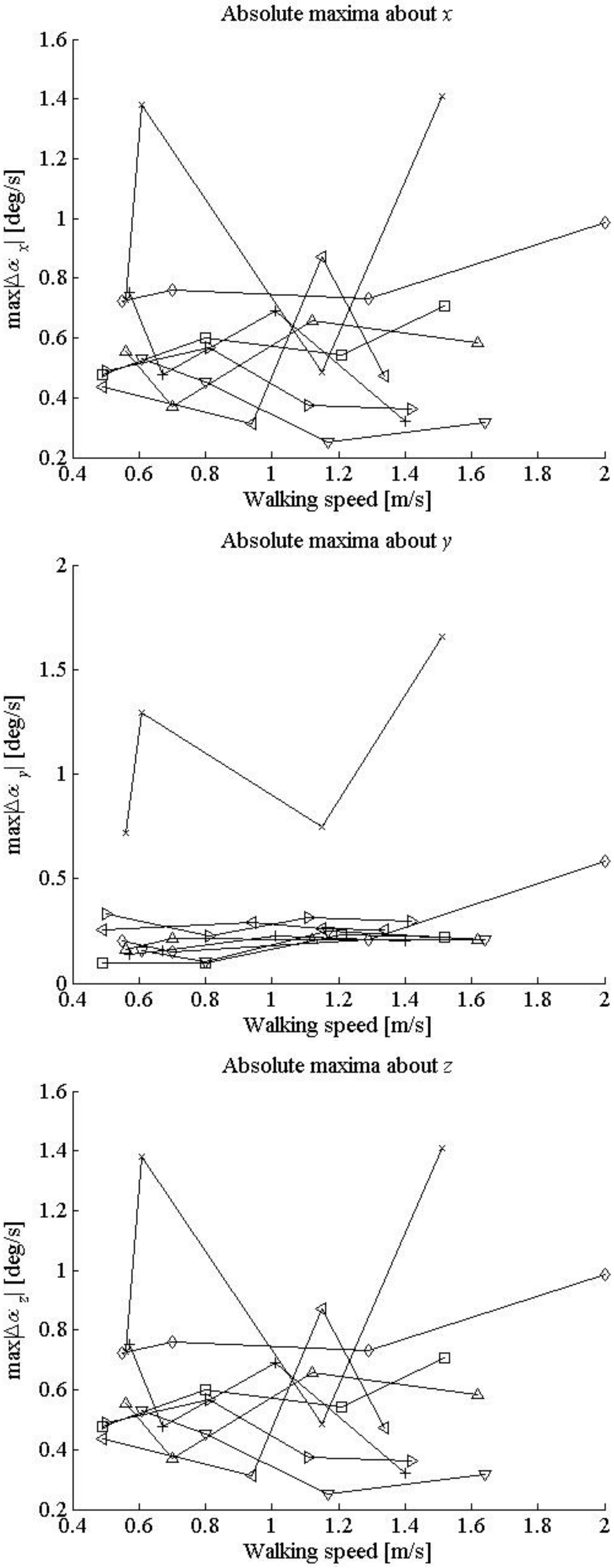

FIGURE 10 Absolute maxima of changes in the angular velocity of the trunk at different speeds for all subjects. Subjects are indicated by different marker types.

	Subject A							
	Fast		Free		Slow		Very slow	
	m $\times 10^{-4}$	b $\times 10^{-4}$	m $\times 10^{-4}$	b $\times 10^{-4}$	m $\times 10^{-4}$	b $\times 10^{-4}$	m $\times 10^{-4}$	b $\times 10^{-4}$
Δv_x	3.2920	0.1727	3.2680	0.3265	3.1968	-0.1445	3.5377	-0.2525
Δv_y	8.8420	1.2033	9.1941	0.8485	8.6607	0.7723	8.5982	0.5371
Δv_z	1.1258	-0.0179	1.2186	-0.0062	1.7804	-0.1117	1.7240	-0.2030
	m $\times 10^{-2}$	b $\times 10^{-2}$	m $\times 10^{-2}$	b $\times 10^{-2}$	m $\times 10^{-2}$	b $\times 10^{-2}$	m $\times 10^{-2}$	b $\times 10^{-2}$
$\Delta \omega_x$	3.5894	0.1274	3.7679	-0.0772	5.7111	-0.2171	4.9942	-0.4598
$\Delta \omega_y$	2.9921	-0.2679	3.0361	0.4703	2.2977	-0.1821	3.3984	-0.3872
$\Delta \omega_z$	10.4877	-0.3369	10.4169	1.6718	10.8945	-0.4736	10.4471	-0.4829
	Subject B							
	Fast		Free		Slow		Very slow	
	m $\times 10^{-4}$	b $\times 10^{-4}$	m $\times 10^{-4}$	b $\times 10^{-4}$	m $\times 10^{-4}$	b $\times 10^{-4}$	m $\times 10^{-4}$	b $\times 10^{-4}$
Δv_x	6.0871	-0.3439	2.7381	-0.0463	3.3882	-0.1266	2.4845	-0.0804
Δv_y	10.6645	-1.1378	11.0276	0.7221	8.2694	-0.1192	8.4706	-0.3555
Δv_z	1.9316	-0.0049	1.7805	0.0005	2.1238	0.0247	2.4334	-0.0460
	m $\times 10^{-2}$	b $\times 10^{-2}$	m $\times 10^{-2}$	b $\times 10^{-2}$	m $\times 10^{-2}$	b $\times 10^{-2}$	m $\times 10^{-2}$	b $\times 10^{-2}$
$\Delta \omega_x$	10.0328	-1.1008	7.2311	0.7288	7.5768	0.2257	7.2959	-0.3032
$\Delta \omega_y$	5.5417	1.4544	1.9026	0.8314	1.4815	-0.0037	2.0917	-0.3004
$\Delta \omega_z$	13.3967	-2.3498	13.0871	-0.3808	16.2108	-0.6383	14.2285	-0.1277

TABLE 2 A least-squares linear fit to the absolute maxima of changes in the components of linear and angular velocities of the trunk versus the magnitude of the applied force for magnitudes of 5, 10, and 15 % of each subject's body weight, for two subjects walking at a very slow, slow, free, and fast speed. Slope of the line is denoted by m, and the constant term by b.

4.4 Discussion

Stability of walking over a range of walking speeds was tested by simulated responses to the above tripping perturbation [PI]. Simulations of eight young healthy subjects walking at four speeds were examined. Responses to the tripping perturbation were quantified as changes in the (angular) velocity of the trunk. Specifically, the greatest change in each (angular) velocity component induced by the constant perturbing force during the swing phase was analyzed. Such changes in the (angular) velocity components were shown to be weakly correlated with the walking speed, but the effect of speed was shown to be small. Changes in the anterior-posterior and vertical components of the velocity displayed a positive and statistically significant correlation to the walking speed, whereas the medio-lateral component displayed a negative correlation. Among the angular velocity components, only the change in the component about the vertical axis displayed a statistically significant correlation which was positive.

It was speculated in Section 4.1 that fast walking may be unstable, which could explain the high likelihood of falling in fast walking [Pavol01]. This speculation was not supported by the present results: the above findings suggested that gait speed does not have a substantial effect on stability, in the sense of sensitivity to constant perturbation. Intuitively, in a real trip, impact forces may increase with walking speed. Such an increase alone would increase the instability. This is because changes in kinematics were shown to scale with the perturbing force, which likely increases the likelihood of falling. Assuming that the impact force in a trip scales with the walking speed, the effect of such an increase on the simulated response outweighs the observed small change in stability due to walking speed. Of course, the small increases in instability in the anterior-posterior and vertical directions exacerbate balance problems, but in the light of the present results, they do not seem to be dominant. Furthermore, stability was demonstrated to improve in the medio-lateral direction as the speed increases. What direction is critical in falling remains, however, unclear. Stability in the medio-lateral direction may be the most critical [Bruijn09].

It must be beared in mind that the association of fast walking with high likelihood of falling has been studied and demonstrated only in the case of old people [Pavol01], while the present results are for young subjects. These results may or may not extend to the elderly. It has been demonstrated previously that at least the direction of change in stability with gait speed in young people extends to old people [Kang08]. If the present results also extend so, slowing down is not a way for the elderly to improve their stability, but may be a preventive strategy to reduce impact forces of a potential tripping. On the other hand, one can walk fast and safely as far as one can avoid high impact forces. In order to achieve that, people could be trained to respond to tripping by moving the foot backward so as to reduce the impact force.

Tendencies of the effects of speed on changes in all components of the velocity revealed by the tripping perturbation were reminiscent of those of long-term divergence exponents, but compared with short-term divergence exponents, only the vertical velocity showed a reminiscent tendency [Bruijn09]. Such a behavior of divergences makes the effects of speed on local dynamic stability indefinite. Also, the origins and implications of divergences and the simulated responses to the tripping perturbation are very different as discussed in Section 4.1, so they do not have to be related to each other. Local dynamic stability may be one factor in the response of the trunk to a tripping perturbation.

Unfortunately, the data set is very small, which really do not allow statistical analysis of subject-specific speed effects. In three subjects, the anterior-posterior velocity displayed a drastic change, when the tripping perturbation was performed in late swing. This may indicate that these subjects have crossed a threshold after which walking really becomes unstable.

The reported responses are for a constant force. In real tripping, the corresponding tripping force is expected to vary based on the phase of the swing at which tripping occurs. This variation in the force alters the responses (*i.e.*, changes in the (angular) velocity components). The highest magnitudes of the tripping force are expected to occur in the mid-swing, when the momentum of the swing limb is highest. This kind of variation in the tripping force could alter the effects of gait speed on stability. On the other hand, toe clearance is highest in the mid-swing, so the likelihood of hitting an object is lowest during this phase. As can be seen, inclusion of the complexity of the real world in a tripping-perturbation simulation would complicate the analysis. The present implementation of the tripping perturbation in its simplicity provides an easy interpretation of results through which the real-world complexities can also be understood.

5 DETECTION OF AN IMPAIRED STABILITY

This Chapter demonstrates use of the tripping perturbation in detection of an impaired stability of walking. Subject-specific muscle-driven simulations of children with and without cerebral palsy were analyzed by means of the tripping perturbation to see if it can detect an impaired stability of walking [PIII, Klemetti13]. After the Introduction, specific details on the subjects and performed analyses are given. Then, results of the tripping perturbations are reviewed and discussed.

5.1 Introduction

To test whether the tripping perturbation can detect an impaired stability of a subject, it must be applied to a subject who is known to have an impaired stability. Cerebral palsy (CP) is a non-progressive disorder of movement and posture due to a defect or damage in the central nervous system. CP is a static encephalopathy [Nelson82]. Although CP appears to get worse, this is actually a result of deficits becoming more observable as a child grows over time. There is no cure for CP. Prevalence of CP is 2 cases per 1000 births [Gibson03]. One way to classify CP is to describe the predominant motor characteristics. These characteristics include spastic, hypotonic, athetotic, dystonic, and ataxic. In addition, topographical pattern of limb involvement, such as monoplegia, diplegia, triplegia, hemiplegia, or quadriplegia, is used. Spastic CP results from a defect or damage in the brain's corticospinal pathways, also known as the upper-motor-neuron damage. The prevalence of spastic CP is 70 to 80 % of the cases of CP. The major signs of CP include delayed motor milestones, abnormal neurologic examination, persistence of primitive reflexes, and abnormal postural reactions. [Taft95]

Children with and without spastic diplegic CP were examined by the tripping perturbation in two separate studies [PIII, Klemetti13]. CP results in balance impairment observable in walking [Hsue09a, Hsue09b, Liao97], hence a predictive method should be capable of detecting a difference between the subject groups. In light of this, it was hypothesized that, because of an impaired walking stability, tripping induces greater kinematic changes in subjects with CP than in unimpaired subjects. That is, since walking of the CP subjects is more unstable, it is easier to deflect their kinematics. Aim of the studies was not just to compare the groups, but also to provide means for an individualized analysis.

5.2 Stabilization of the trunk

5.2.1 Analyses

Previously created subject-specific muscle-driven simulations of walking were studied by the tripping perturbation [PIII]. Simulations were for children with spastic diplegic cerebral palsy and for unimpaired children walking at their self-selected speed. The age, body mass, and leg length of ten CP subjects were 8.1 $\pm$ 1.7 yrs, 27.1 $\pm$ 9.1 kg, and 0.65 $\pm$ 0.06 m, respectively. For comparison, eight subjects - age 12.9 $\pm$ 3.3 yrs, body mass 51.8 $\pm$ 19.2 kg, leg length 0.81 $\pm$ 0.09 m, respectively - were studied.

Magnitude of the tripping force was 10 % of the subject's body weight. Period of the tripping perturbation, t_t, was 20 ms. In each tripping perturbation, the difference between a perturbed position, x_p, and the corresponding unperturbed position, x_u, of the trunk center of mass, in the fore-aft direction, was computed as a function of time, t, as follows:

$$\Delta x(t) = x_p(t) - x_u(t). \tag{9}$$

The positive direction of x is forward.

Each tripping perturbation was used for seeing if walking was stable at the respective instant of time. It was shown previously that magnitudes of the responses to the tripping perturbation depend on the size of the subject [PI]. Consequently, these magnitudes are not useful when comparing subjects of different size. A more universal measure is, thus, needed. Walking was defined as stable at a given instant of time if the following condition was met [PIII]: the absolute difference between the perturbed and unperturbed positions of the trunk at the end of the perturbation was smaller than its absolute peak value during the perturbation. In other words, at a given instant of time, walking was defined as stable if deviation of the position of the trunk due to the perturbation was bounded. Such a stable state means that, after initially deviating from its unperturbed position, the trunk then approaches and stays close to that position. An unstable state means in turn that the trunk is moving away from its unperturbed position. Examples of a stable and an unstable state are illustrated in Fig. 11.

Percentage of the swing phase during which the subject was stable was determined from all the perturbations performed. The group average of that percentage was determined for each subject group. Student's t test was performed to test whether a difference between group averages was statistically significant.

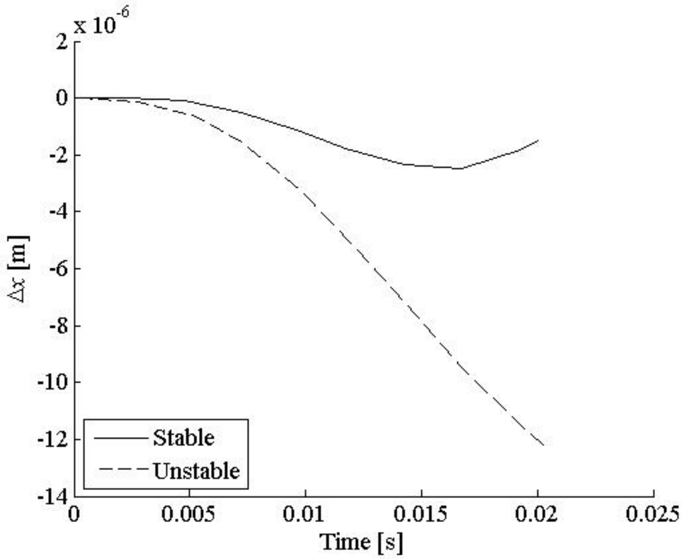

FIGURE 11 Examples of a stable and an unstable state. A difference between the perturbed and unperturbed positions of the trunk as a function of time for a stable and an unstable state.

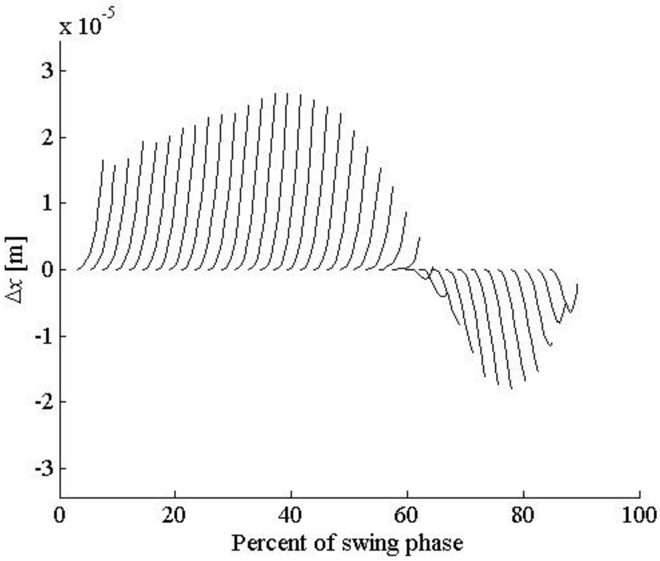

FIGURE 12 Responses to the tripping perturbations for one unimpaired subject. Each plotted curve represents a difference between the perturbed and unperturbed positions of the trunk as a function of the percentage of the swing phase. Every tenth perturbation performed during the swing phase is shown in the figure.

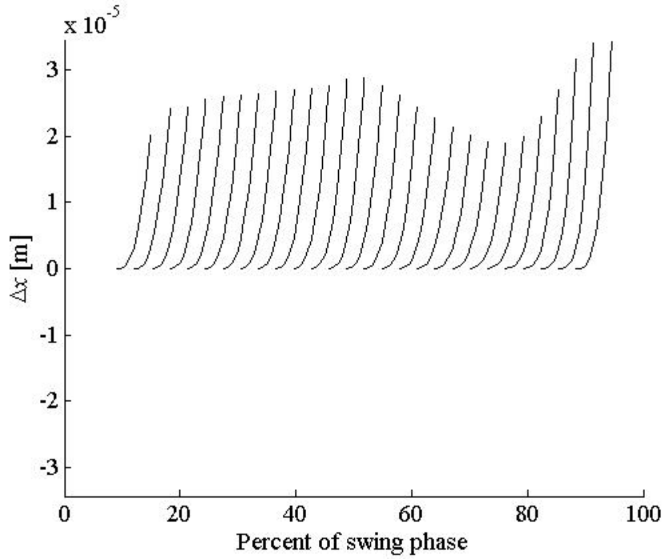

FIGURE 13 Responses to the tripping perturbations for one CP subject. Each plotted curve represents a difference between the perturbed and unperturbed positions of the trunk as a function of the percentage of the swing phase. Every tenth perturbation performed during the swing phase is shown in the figure.

5.2.2 Results

Responses to the tripping perturbation are depicted for one unimpaired subject and for one CP subject in Figs 12 and 13, respectively. All unimpaired subjects were stable at some point of their swing phases, whereas six out of the ten CP subjects were never stable. Unimpaired subjects were statistically significantly stable for a larger percentage of the swing phase than the CP subjects ($p < 0.05$). On the average, unimpaired subjects were stable for 8.5 % of the swing phase while CP subjects were stable for 1.5 % of the swing phase. These findings disclose that the bulk of the swing phase is unstable even in unimpaired individuals. However, CP seems to be related to a deteriorated proactive control of balance as CP subjects are less able to stabilize their walking.

5.3 Dynamic equilibrium of the trunk

5.3.1 Analyses

Previously created subject-specific muscle-driven simulations of walking were studied by the tripping perturbation [Klemetti13]. Simulations were for children with and without spastic diplegic cerebral palsy walking at a self-selected speed. The age, body mass, and leg length of the nine CP subjects were 8.2 ± 1.8 yrs, 24.4 ± 3.8 kg, and 0.64 ± 0.06 m, respectively, while they were 11.3 and 7.0 yrs, 32.4 and 26.1 kg, 0.72 and 0.66 m, respectively, for the two unimpaired subjects.

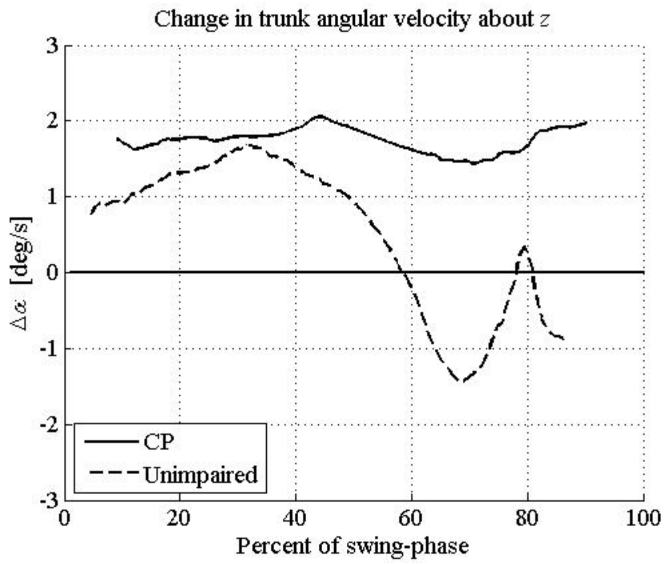

FIGURE 14 Changes in the angular velocity of the trunk in the sagittal plane for one CP subject and for one unimpaired subject, when varying the time of perturbation during the swing phase.

Two steps for each subject were analyzed. According to Eq. (8), $\Delta\vec{\omega}$ was plotted versus percentage of the swing phase for a tripping force with a magnitude of 10 % of the subject's body weight for a period of t_t=20 ms, so that each plotted data point represents a result of a single tripping perturbation, and the entire curve the results of such perturbations for varying the time of application of the force during the swing phase.

5.3.2 Results

Figure 14 depicts $\Delta\omega_z$ for one unimpaired subject and for one CP subject. The unimpaired subjects exhibited both positive and negative values of $\Delta\omega_z$ during each step, whereas the CP subjects exhibited only positive ones, with one exception; one CP subject exhibited positive values during one step and both positive and negative values during the other step. It is evident that if $\Delta\omega_z$ displays both positive and negative values, then there is a time during the swing phase such that, if perturbation is performed then, there is only a negligible $\Delta\omega_z$. This negligible $\Delta\omega_z$ indicates the existence of a dynamic equilibrium at that instant of time, since perturbation does not cause a clear motion. The CP subjects were thus concluded to have an impaired stability as they had difficulties in reaching a dynamic equilibrium.

5.4 Discussion

Stability of walking of children with and without CP, was studied by the tripping perturbation [PIII, Klemetti13]. CP subjects were shown to be more unstable, with respect to tripping, than unimpaired subjects. In other words, CP subjects are more sensitive to a tripping perturbation, and, consequently, likely to be more susceptible to falling than unimpaired subjects.

It should be noted that results are trip specific, other types of perturbation would likely produce very different characteristics. For example, the dynamic equilibrium is trip specific; that is, angular velocity is not affected much by the tripping force. In light of the above, results cannot be considered as a general or complete picture of walking stability. Regardless of this, trip-specific characteristics are important, when studying trip-related falls.

The EMG data have shown some disagreement with the simulated muscle activations [Liu08, Steele10]. Such disagreement may have been due, in part, to measurement errors in the experimental gait data. Also, the generic model does not account for subject-specific muscle-tendon properties.

Dynamic equilibrium - in the sense of a negligible change in the trunk kinematics - may result from stiffness of certain joints so that (angular) momentum is distributed over the body segments reducing the change in the trunk kinematics. Alternatively, it may result from weak coupling between the tripped foot and the trunk leaving the trunk unaffected. Further research is needed to find out how dynamic equilibrium is reached and why the subjects with CP tend to fail to reach it. Once the causes are known, they can be targeted in rehabilitation by designing special interventions to improve the balance.

6 NEUROMUSCULAR CONTROL OF THE TRUNK

This Chapter is devoted to the analysis of neuromuscular control of the trunk during walking. In this Thesis, the essence of controlling the stability against falling is considered to be control of the trunk. Control of the trunk was studied by identifying muscles that contribute to the angular acceleration of the trunk in walking of young healthy subjects [PII]. This task was carried out through the induced-acceleration analysis. Chapter begins by explaining the importance of knowledge about muscle contributions. Then, details of the subjects and analyses are given. Ultimately, muscle contributions in the sagittal and frontal planes in different parts of the gait cycle are reviewed and discussed.

6.1 Introduction

Simulation allows quantification of the effect of muscle force on movement generation. Analyzing such contributions of individual muscles to the angular acceleration of the trunk is the key to understanding how the trunk is controlled during walking. Studying the unimpaired control of the trunk reveals the characteristics of good control. These characteristics can be pursued in the rehabilitation of an impaired control. It has been demonstrated that muscles recruited in normal walking are also recruited in atypical phases of gait, accounting for both anticipatory gait modifications prior to perturbations and reactive feedback responses to perturbations [Chvatal12]. In the light of this, muscles that contribute to trunk control in normal walking may also contribute to trunk control under perturbed circumstances, attempting to prevent an impending fall.

In spite of its importance, muscle contributions to the angular acceleration of the trunk have not been addressed so far. Muscle contributions to trunk energetics have been studied [Allen12]. Such an analysis does not, however, reveal the direction in which muscles accelerate the trunk. Nott et al. [Nott10] have reported that all joint moments significantly contribute to the angular acceleration of the trunk. In addition to not including muscles, however, Nott et al. [Nott10] have modeled the pelvis and trunk as one segment, which may not accurately reflect the degrees of freedom between the pelvis and trunk. Compared to a model of separate pelvis and trunk segments, such a simplification has indeed previously been shown to have an impact on the role of the stance-limb hip moment, and on the magnitude of the joint moment contributions to trunk energetics [Patel07].

Individual lower-limb and trunk-muscle contributions to the angular acceleration of the trunk in normal walking of young healthy subjects, using muscle-driven simulations, were investigated [PII]. The musculoskeletal model with separate pelvis and trunk segments and 92 muscle-tendon actuators was used. Objective was to identify muscles that make the largest contributions to the angular acceleration of the trunk in the sagittal and frontal planes in different parts of the gait cycle. Since all joint moments have previously been reported to significantly contribute to the angular acceleration of the trunk [Nott10], it was hypothesized that muscles at all joints significantly contribute to this acceleration.

6.2 Analyses

Previously created subject-specific muscle-driven simulations of walking were studied by the induced-acceleration analysis [PII]. Seven healthy subjects walking at their free speed were examined. Their ages ranged from 7.0 to 18.0 yrs with a mean of 12.9 yrs. Walking speeds ranged from 1.0 to 1.3 m/s with a mean of 1.2 m/s.

The induced-acceleration analysis was employed to compute the contributions of individual muscles to the angular acceleration of the trunk in the sagittal and frontal planes. This analysis was done separately at each time point of the subjects' gait simulations. In this analysis, model's equations of motion are solved so as to determine the angular acceleration of the trunk induced by each force (muscle force, gravity, and velocity-related forces).

Since the data did not include the entire swing phases, assuming symmetry, the corresponding parts of previous swings were used so as to analyze an entire gait cycle. The induced-acceleration data of each force were divided into positive and negative parts. These parts were integrated with respect to percentage of the gait phase. The studied gait phases include the initial double support, and early and late single supports corresponding to the first and second halves of the single-support phase, *i.e.*, the single-limb stance, respectively. Mean and standard deviation of each force's contribution (*i.e.*, integral of the induced acceleration) over all subjects were computed for each phase. Forces that had induced angular accelerations larger than 15 percent of that of the largest contributor in each phase, were included for further analysis so as to only report forces that contributed more than the residuals. For comparison with previous studies, the induced-acceleration analysis was also conducted to identify contributions of the net joint moments to the angular acceleration of the trunk.

Abbreviation	Description
ERCSPN	Erector spinae
EXTOBL	External oblique
INTOBL	Internal oblique
GMAX(a, im, p)	Gluteus maximus (anterior, intermedius, posterior)
GMED(a, im, p)	Gluteus medius (anterior, intermedius, posterior)
ADDL	Adductor longus
ADDM(s, m, i)	Adductor magnus (superior, medius, inferior)
SMEM	Semimembraneous
STEN	Semitendineous
BFlh	Biceps femoris long head
RF	Rectus femoris
VASLAT	Vastus lateralis
SOL	Soleus
MEDGAS	Medial gastrocnemius
TIBANT	Tibialis anterior

TABLE 3 Descriptions of the abbreviations of muscles.

6.3 Results

Abbreviations for the reported muscles are described in Table 3. The induced angular-acceleration curves for one subject, illustrated in Fig. 15, exemplify the joint-moment and gravity contributions to the angular acceleration of the trunk. In the sagittal plane, the hip-flexion/extension and lumbar-flexion/extension moments had the largest contributions, whereas the knee-flexion/extension and ankle-plantar/dorsiflexon moments played a minor role. In the frontal plane, the majority of the joint moments made considerable contributions. For this same subject, induced angular-acceleration curves for a few selected muscles are shown in Fig. 16. Like the joint moments, proximal muscles were of the greatest importance in the sagittal plane, while both proximal and distal muscles contributed to the angular acceleration of the trunk in the frontal plane.

Figures 17 and 18 depict the group means and standard deviations of muscles' and gravity's contributions to the angular acceleration of the trunk in the sagittal and frontal planes, respectively. These results demonstrate that, during the initial double support in the sagittal plane, the back and hip-flexor muscles accelerated the trunk backward, while the hamstring, abdominal, and gluteus muscles accelerated it forward. In the frontal plane, the ipsilateral and contralateral abdominal and back muscles induced the largest ipsilateral and contralateral angular accelerations, respectively. In the sagittal plane during the single-support phase, the main contributors had the same role as during the double-support phase. Contributions of individual muscles, however, varied over time. For example, contributions of the iliacus and psoas of the swing limb were shown to decrease from the initial double-support phase to the swing phase. Further,

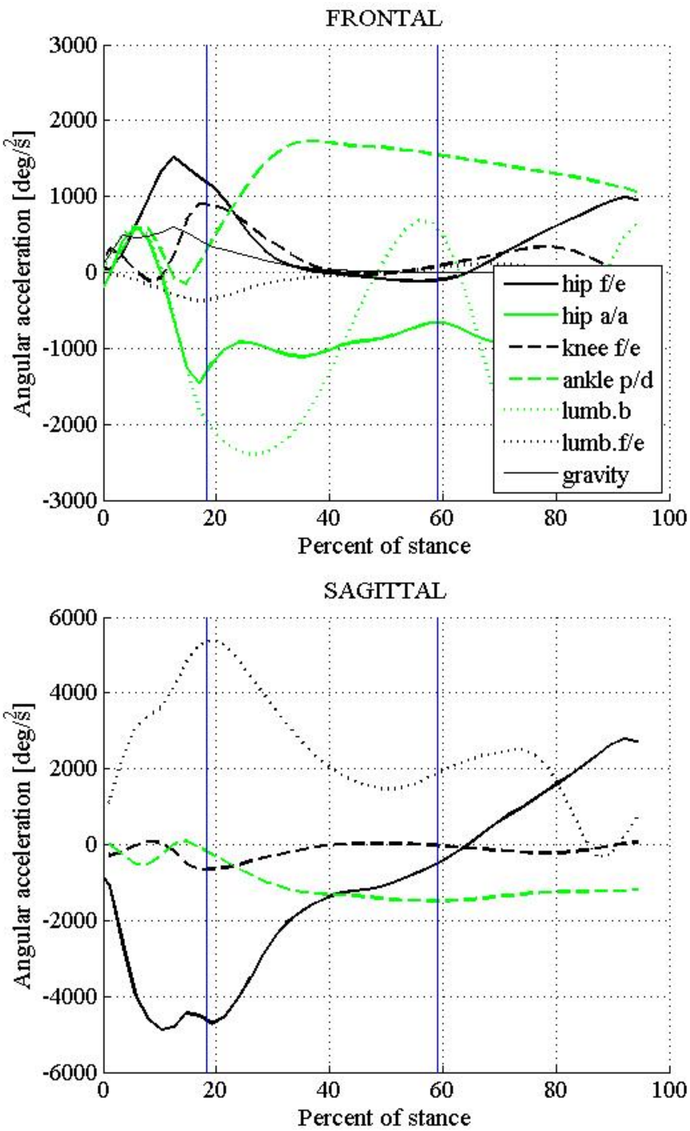

FIGURE 15 Contributions of the stance-limb joint moments and gravity to the angular acceleration of the trunk during a right-leg stance for one subject. In the sagittal plane, positive direction corresponds to rotation where the upper trunk moves backward. In the frontal plane, positive direction corresponds to rotation where the upper trunk moves rightward. The studied phases (initial double support, and early and late single supports) are separated by the vertical solid lines. In the sagittal plane, the hip-flexion/extension and lumbar-flexion/extension moments made the largest contributions, whereas the knee-flexion/extension and ankle-plantar/dorsiflexon moments played a minor role. In the frontal plane, the hip-flexion/extension-, hip-abduction/adduction-, ankle-plantar/dorsiflexion-, knee-flexion/extension- and lumbar-bending moments were all important.

the hamstring muscle group displayed an increasing contribution from the early-swing to the late-swing phase. There is time evolution in the muscle contributions in the frontal plane as well. For example, the gluteus medius of the support limb exhibited a gradually increasing contribution from the initial double-support to the late single-support phase. Also, the soleus and medial gastrocnemius of the support limb had a significant contribution only during the single-support phase. Gravity showed a considerable contribution only in the frontal plane during the double-support phase. The high error bars in Figs. 17 and 18 reflect a high inter-subject variability.

6.4 Discussion

Contributions of individual lower-limb, abdominal, and back muscles to the angular acceleration of the trunk during walking, in the sagittal and frontal planes at different parts of the gait cycle, were identified [PII]. Young healthy subjects walking at their free speed were studied by means of 3D muscle-driven simulations. Results show that, in the sagittal plane, roughly, the more distal the muscles are to the trunk, the less they contribute to its angular acceleration. In contrast to this, in the frontal plane both proximal and distal muscles exhibit large contributions. The abdominal and back muscles are of great importance throughout the gait cycle in both planes of rotation, while the lower-limb muscles display changes in their contributions depending on the phase of the gait cycle. In both planes, the swing-limb muscles also exhibit an appreciable contribution. There was a comparatively high inter-subject variability, which makes it impossible to assess the exact order of importance of the individual muscles in general. Nevertheless, a group of important muscles, such as the hip-flexors, hip-extensors and hip-abductors, abdominal, and back muscles, were identified.

It has been previously reported that, in the sagittal plane, all joint moments significantly contribute to the angular acceleration of the trunk [Nott10]. This result is not supported by the present findings. Such a difference between these two studies has likely resulted from the fact that, in the study by Nott et al. [Nott10], the model used has no degrees of freedom between the trunk and pelvis, *i.e.*, the trunk and pelvis are a combined segment. This has made the distal lower-limb segments more strongly coupled with the trunk, and, consequently, has led to larger contributions of the distal lower-limb joint moments to the angular acceleration of the trunk. Also, because of this simplification, the moment between the trunk and pelvis has been ignored, while the present results show that it is the largest contributor. In the frontal plane, the results of these two studies are similar in the sense that all the sagittal-plane lower-limb joint moments significantly contribute to the angular acceleration of the trunk. This can be explained as follows: the sagittal-plane joint moments cause reaction forces that have a moment about the trunk in the frontal plane, and these reaction forces are not dependent on the degrees of freedom of the model. Because of the difference between the

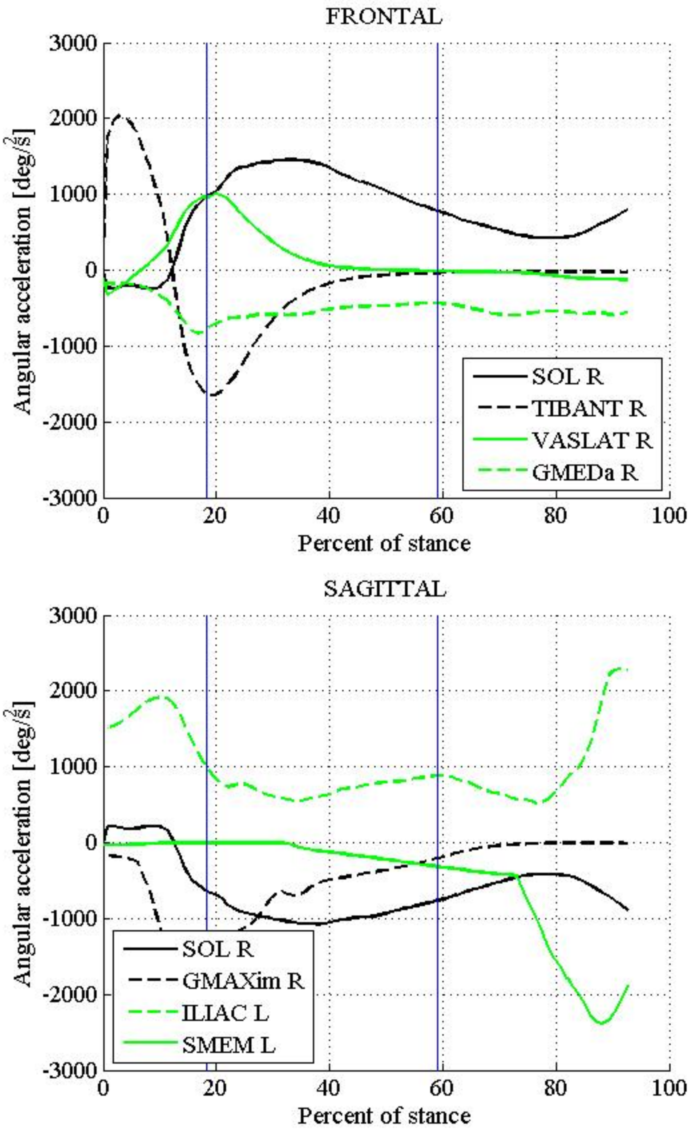

FIGURE 16 An example of contributions of individual muscles to the angular acceleration of the trunk during a right-leg stance for one subject. In the sagittal plane, positive direction corresponds to rotation where the upper trunk moves backward. In the frontal plane, positive direction corresponds to rotation where the upper trunk moves rightward. The studied phases (initial double support, and early and late single supports) are separated by the vertical solid lines. In the sagittal plane, proximal muscles were of the greatest importance, while distal muscles were of great importance, too, in the frontal plane.

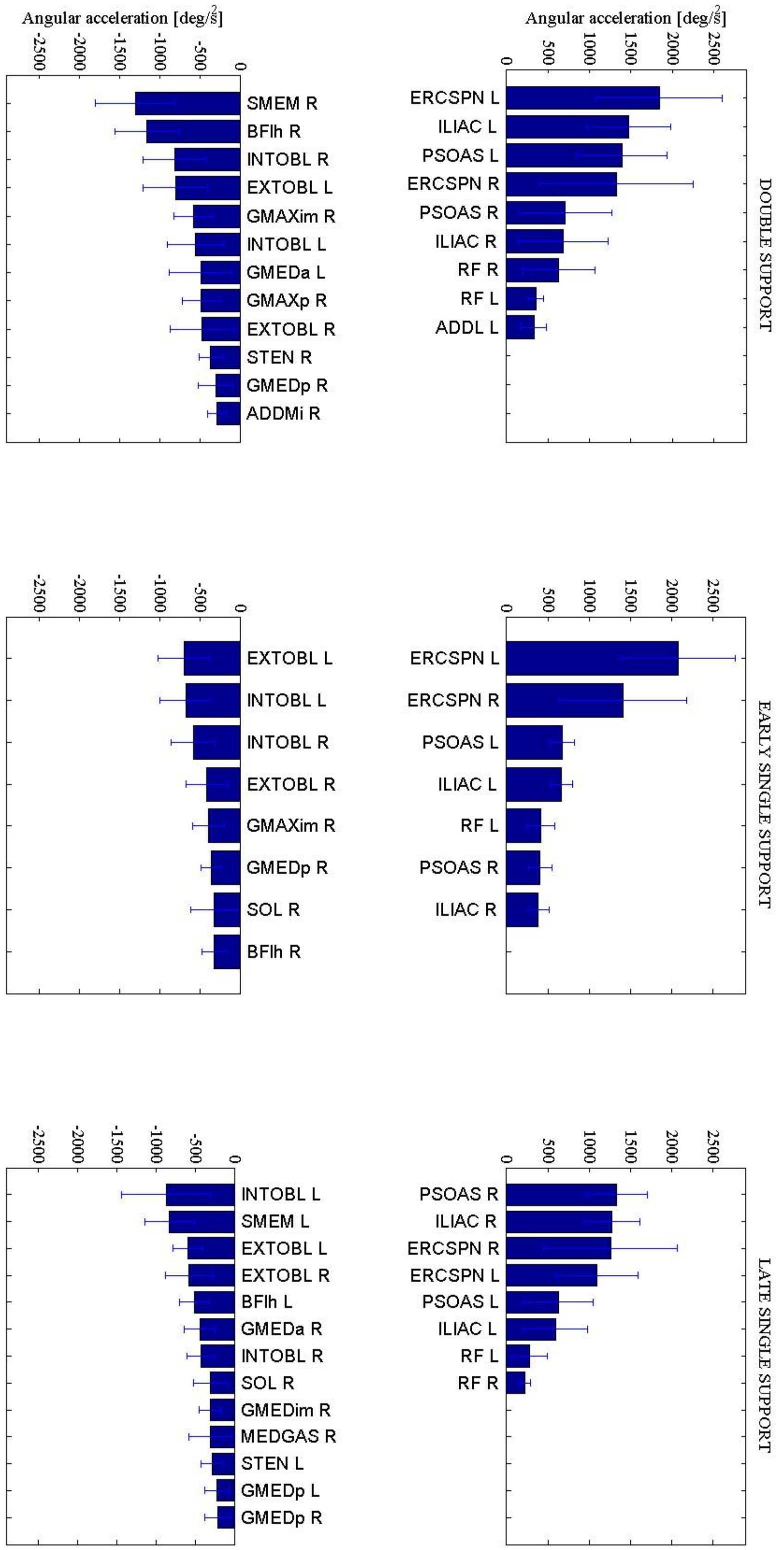

FIGURE 17 Results for the sagittal plane. The group means and standard deviations of individual muscles' and gravity's contributions to the angular acceleration of the trunk in different phases of the gait cycle. Means are indicated by bars and standard deviations by error bars. R refers to the leading leg (right leg) and L refers to the trailing leg (left leg). Positive direction corresponds to rotation, where the upper trunk moves backward.

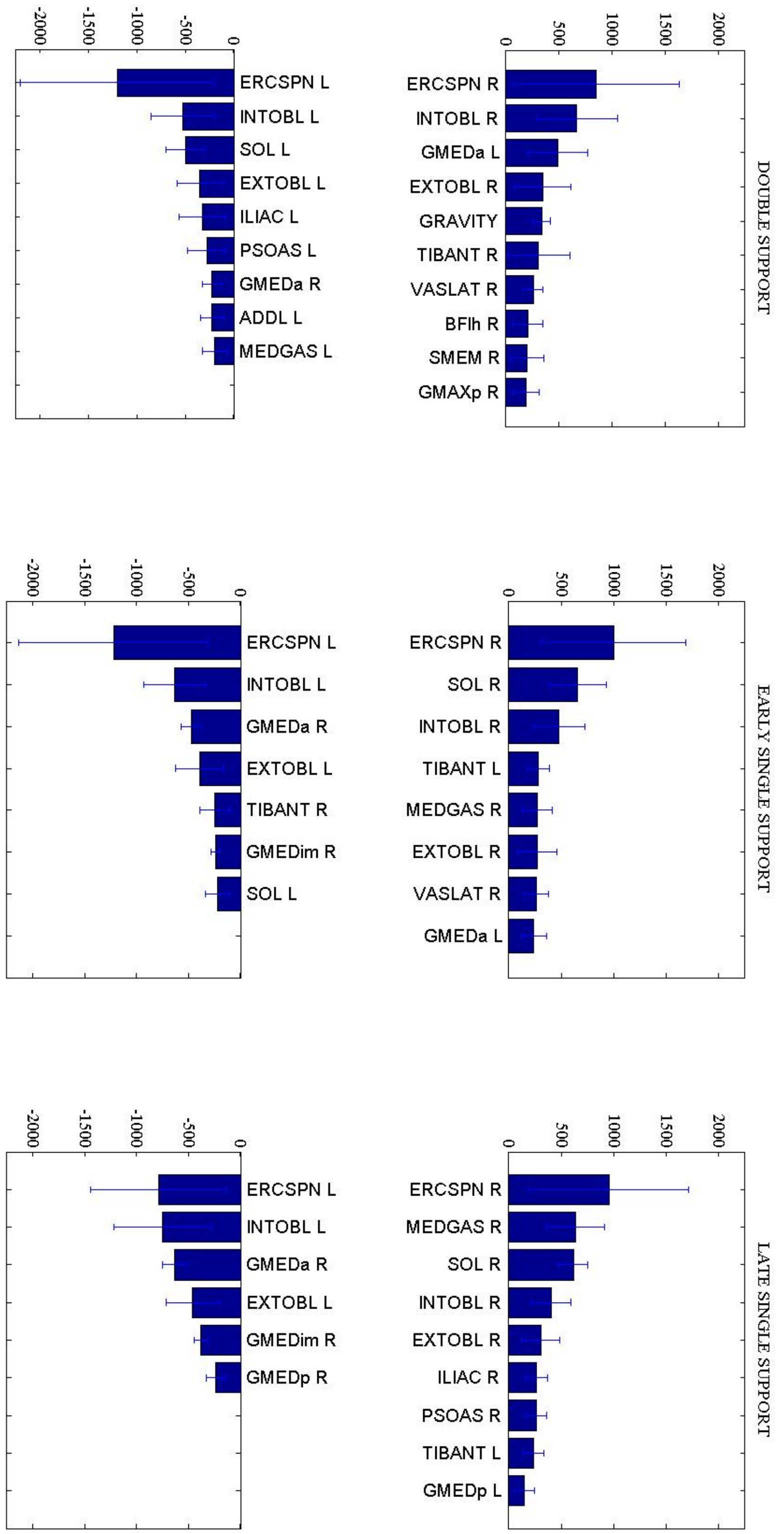

FIGURE 18 Results for the frontal plane. The group means and standard deviations of individual muscles' and gravity's contributions to the angular acceleration of the trunk in different phases of the gait cycle. Means are indicated by bars and standard deviations by error bars. R refers to the leading leg (right leg) and L refers to the trailing leg (left leg). Positive direction corresponds to rotation, where the upper trunk moves rightward.

models, the meaning of the trunk is not the same in these two studies, so the results cannot be compared in detail. There is currently no gold standard against which to evaluate the accuracy of the induced-acceleration-analysis predictions; however, modeling the pelvis and the trunk as separate segments is closer to the anatomical architecture of the body [Patel07].

Present results disclose that, along with the abdominal and back muscles, gluteus medius is an important muscle in opposing the trunk to sway in the frontal plane. Thus, enhancing the function of gluteus medius would likely ameliorate the stability of many pathologies by attenuating the trunk sway. In fact, weakness of the hip abductors is associated with the ipsiliateral trunk sway in patients with cerebral palsy [Krautwurst13], and in patients with lumbo-sacral myelomenginocele [Gutierrez03]. Present results also disclose that soleus and medial gastrocnemius accelerate the trunk ipsilaterally. This should be taken into consideration when designing rehabilitation management; increased ankle-plantarflexor forces are advantageous as they help propel forward, but are disadvantageous as they destabilize the trunk by causing trunk sway, so all movement tasks should be considered as a whole. Present findings show that gravity has a considerable contribution in the frontal plane. It may thus be that pathologies cannot resist such an effect of gravity, which may engender excessive trunk sway among pathologies.

The considerable contribution of the swing-limb muscles to the angular acceleration of the trunk is meaningful as it implies that the swing limb can stabilize or destabilize the trunk. For example, the swing limb can be moved quickly in response to a perturbation to stabilize the trunk, but the strong coupling between the swing limb and the trunk makes the trunk sensitive to external perturbations that are directed to the swing limb, e.g., tripping. In the light of the swing-limb contributions, responses to the tripping perturbations reviewed in this Thesis depend on the dynamics of the swing limb. This emphasizes the specificity of the responses to tripping. It is not likely that the responses to slipping, for example, would be strongly related to the dynamics of the swing limb.

It is worth mentioning that the muscles that accelerate the trunk, simultaneously accelerate other body segments as well, and are thus used to perform other movement tasks than to control the trunk alone. The central nervous system has to recognize such a complexity, when executing concurrent movement tasks. For example, the swing-limb muscles are primarily recruited to initiate and terminate a swing, but they also accelerate the trunk as discussed above.

It must be noted that the trunk model used is rather simple: the spine is modeled by a ball-and-socket joint, and only a few trunk muscles are included (*i.e.*, erectus spinae and internal and external oblique muscles). A more comprehensive spine model could have affected the results as reported because of a higher number of degrees of freedom. The additional degrees of freedom would have, in turn, complicated the analysis [Christophy12]. On the other hand, several studies have indicated that quite simple measures of the trunk motion can distinguish pathologies from normal individuals [Goutier10, Spain12]. This suggests that the simple model used is capable of capturing such differences. As for the trunk muscles, rectus abdominis, for example, would have likely made a large contribution, but it is not included in the model. Thus, the internal and external oblique muscles' contributions reported must be interpreted as the net effect of all the abdominal muscles. In other words, the present results demonstrate quite an intuitive fact that the abdominal and back muscles are important, but their exact individual contributions remain unclear. Also, the paraspinal muscles have been shown to play a key role in minimizing the head motion by controlling the trunk [Prince94]. Overall, the strength of the present results lies in identifying the lower-limb-muscle contributions in detail.

Although the musculoskeletal model used does not include the arms, the residual forces and moments applied to the model are intended, in part, to represent effects of the arms. Arm motion certainly contributes to the angular acceleration of the trunk. Nevertheless, contribution of the arms can be considered to be smaller than that of the muscles that were reported, since only muscles that contributed more than the residuals were reported. In general, as far as arms swing symmetrically, their angular momenta cancel each other in the sagittal and frontal planes, and, consequently, they do not affect the trunk motion. In the transverse plane, there is no such cancelation, but the transverse plane was not analyzed with this model.

Results should also be interpreted in the light of the fact that, in the induced-acceleration analysis, estimates of the muscle contributions depend on simulated muscle forces. A direct validation of the simulated muscle forces is not possible as the actual muscle forces of a human subject cannot usually be measured. Alteration of the muscle forces does not, however, affect the direction of the angular acceleration induced by each muscle, but it can alter their relative contributions.

7 SUMMARY AND OUTLOOK

This Thesis addressed stability against falling in walking with a special reference to perspective of movement simulations. Thesis began by discussing the definition of stability relevant for describing the fall risk. The bulk of the Thesis reviewed a novel simulation-based method to assess stability of walking, and applications of the method. In addition, neuromuscular control of stability as examined by muscle-driven simulations was reviewed.

A small whole-body angular momentum has previously been envisaged to indicate a good balance and a low fall risk during locomotion. In contrast to that, this Thesis disclosed results that a fall can result also with zero whole-body angular momentum. These results were obtained by solving the equations of motion for a two-link system in the case that the system's angular momentum about its center of mass is zero [PIV]. This analysis showed that among the solutions to these equations, there exist solutions that yield a decreasing elevation of the center of mass. The central nervous system could choose these undesirable solutions, if its strategy to prevent a fall was to minimize the angular momentum, which would be a failure to control the balance. This suggests the whole-body angular momentum is not a good measure of stability. Moreover, for the same reason, the central nervous system does not likely use, at least not solely, minimization of the whole-body angular momentum as the strategy to control stability and to prevent falling. In fact, because different states of the human body yield the same whole-body angular momentum, meaning of the whole-body angular momentum is equivocal. In this Thesis, control of stability against falling was expected to be conducted through control of the trunk. Importance of the control of the trunk was established through literature review and reasoning.

A simulation-based method of a tripping perturbation for quantifying stability of walking was introduced [PI, PIII, Klemetti13]. In this method, an immediate passive response to this tripping perturbation was used as the measure of stability. Specifically, response was quantified as changes in the trunk kinematics. Such response can be considered to reflect stability against falling due to tripping. Advantages of the tripping perturbation are that it is standardized for all conditions and subjects, ensuring fair comparison, and that the perturbation can be done and repeated free of any anticipation or adaptation by subjects. One disadvantage of the method is its inability to predict, from experimental data of unperturbed gait, subject's actual fall following the perturbation. This inability is due to lack of a model that would account for all neural responses. Consequently, the tripping perturbation cannot fully address the risk of falling. Another disadvantage of the method is the laboratory experiments and time required to create a simulation. For practicality, there may be a need to see if a simpler and faster method could provide reasonably similar information as revealed by the present tripping perturbation. This can be accomplished once more data have been obtained with the tripping perturbation.

The tripping perturbation was utilized to study association of gait speed with stability of walking in young healthy subjects [PI]. Results demonstrated that gait speed has just a small, but a statistically significant effect on stability as for the constant tripping perturbation. Increasing gait speed deteriorated the stability in the anterior-posterior direction, but ameliorated it in the medio-lateral direction. In other words, changes in kinematics due to the tripping perturbation increased and decreased in the anterior-posterior and medio-lateral directions, respectively. This shows that stability of walking does not in itself considerably affect the risk of falling in young healthy individuals. This is not necessarily the case in the elderly or in pathologies. However, gait speed can be expected to increase impact forces in tripping, and higher tripping forces were shown to amplify destabilization [PI]. Therefore, speeding up walking will increase the risk of falling.

It was demonstrated that the tripping perturbation applied can be used to detect an impaired stability of walking [PIII, Klemetti13]. This was manifested in individuals with spastic diplegic cerebral palsy. The tripping perturbation revealed that subjects with CP were stable for a smaller percentage of the swing phase than unimpaired subjects. Also, during the swing phase the CP subjects had difficulties in reaching a dynamic equilibrium, a state where kinematics of the trunk are not affected by the tripping force. In light of the above, this method was demonstrated to indicate that children with spastic diplegic cerebral palsy have balance problems when subjected to a generic tripping. Other patient groups may exhibit very different responses to the tripping perturbation compared to patients with CP. Therefore, the difficulty of reaching a dynamic equilibrium, for example, is not necessarily observable among other patient groups. In other words, other patient groups may have their own specific features that make their walking unstable. This must be kept in mind when using the tripping perturbation to examine other patient groups. Nevertheless, tripping perturbation is capable of detecting an anomaly of individuals with CP, so there is a promise that it is also capable of detecting anomalies in other groups. More research is, of course, needed to confirm that.

It should be noted that the results obtained by the tripping perturbation are trip-specific, other types of perturbation would likely produce very different characteristics. In light of this, they cannot be considered as a general or complete picture of walking stability. In spite of this, trip-specific characteristics are important, when studying trip-related falls, which was the purpose here. Also, the same simulation methodology can be applied to perform other types of perturbation, such as slipping.

Active responses play an important role in the prevention of falls. Tripping-perturbation simulations do not account for any active responses. Therefore, only the phase of passive responses, before the first spinal reflexes appear, is simulated. Nevertheless, the initial instability of walking and muscles' ability to passively resist destabilization determine the need for active responses, thus being factors of the fall risk. Importance of muscles' passive resistance in the stabilization of walking has recently been highlighted [John13]. Furthermore, stable passive responses enable more effective active responses to control the balance.

To understand how the central nervous system controls the stability of walking against falling, the muscles that contribute to the angular acceleration of the trunk were identified [PII]. Results indicate that abdominal and back muscles are of paramount importance. In the sagittal plane, proximal lower-limb muscles contribute more than distal muscles. In the frontal plane, both proximal and distal lower-limb muscles are important. Rehabilitation should thus be targeted especially to abdominal, back, and proximal lower-limb muscles, because tripping and slipping take place primarily in the sagittal plane. Along with the stance-limb muscles, the swing-limb muscles exhibit a considerable contribution. The contribution of the swing-limb muscles indicates a strong coupling between the swing limb and the trunk. This means that impact forces directed to the swing limb during tripping can easily destabilize the trunk. This should be considered in particular when rehabilitating individuals with the stiff-knee syndrome.

Muscle-driven simulations do not come without limitations. These simulations are as accurate as the model that they are based on. The generic musculoskeletal model used to create the simulations reviewed in this Thesis does not account for all features of the human body. Some of the missing features may even be important. Moreover, the model is based on limited experimental data on some parts of the neuromuscular system. Also, model was made subject-specific only by scaling it. Thus, even though many musculotendon properties scale with the size of the subject, many subject-specific properties were not included. This is a concern for the simulations of individuals with CP, whose musculotendon properties are known to differ from the nominal values. One could speculate about the accuracy of the model forever. Nevertheless, validity of the model depends on its application. The tripping perturbation is not expected to be sensitive to properties of, for instance, an individual muscle as it probes the overall state of the system dynamics. It is worth mentioning that this model is the most sophisticated out of all the models that have been used to examine the stability of walking. The computed muscle contributions are, of course, sensitive to properties of the muscles. However, the conclusions that were drawn are very broad in nature.

This Thesis did not yet show the full potential of muscle-driven simulations. Simulations created only by means of a tracking control were reviewed, *i.e.*, the subject's gait and muscle excitations were tracked. Development of predictive control will enable the prediction of muscle excitations and the resulting movement. While the present implementation of the tripping perturbation predicts only the passive responses, predictive control allows inclusion of the active responses. Moreover, predictive control makes it possible to conduct complex theoretical studies on falling and the prevention of falling. For example, the question of whether control of the whole body angular momentum or control of the trunk dominates the stability of walking, can then be addressed.

Muscle-driven simulations in general, and especially a tripping-perturbation simulation, show promise as a tool for rehabilitation. This tool can potentially be used for screening individuals with balance problems and monitoring progress in their physical therapy. More work must still be accomplished so as to streamline the process of creating and analyzing such simulations in order to make them more feasible in rehabilitation.

YHTEENVETO (FINNISH SUMMARY)

Kaatumiset ovat uhka yhteiskunnalle niiden aiheuttamien kuolemien, loukkaantumisten ja niistä seuraavien taloudellisten kustannusten myötä. Kaatumiset liittyvät pääasiassa kompastumiseen tai liukastumiseen kävellessä. Täten keinoja kävelyyn liittyvien kaatumisten ehkäisemiseksi kaivataan. Tämä väitöskirja tarkastelee kaatumisten ehkäisyä kävelyn mekaniikan ja hermo-lihashallinnan näkökulmista. Tätä varten kävelyn tasapainoa kaatumista vastaan tarkastellaan liikesimulaatioilla. Pääosa väitöskirjasta esittelee uutta simulointiin perustuvaa kompastushäiriö -menetelmää kävelyn tasapainon määrittämiseen ja tämän menetelmän soveltamista. Lisäksi tarkastellaan tasapainon määritelmää ja tasapainon hallinnan lihaskoordinaatiota.

Kehon pienen kokonaispyörimismäärän on aiemmin kuviteltu osoittavan hyvää tasapainoa ja pientä kaatumisriskiä liikkumisessa. Kokeellisia mittaustuloksia pyörimismäärästä on raportoitu, mutta pyörimismäärän merkityksen ja tärkeyden teoreettinen perustelu on ollut rajallista. Pyörimismäärän yhteys tasapainoon ja kaatumiseen on siksi tulkinnanvarainen. Väitöskirjassa esitetty kaksilenkkisen järjestelmän analyysi toi esiin, että ihminen voi kaatua, vaikka hänen kokonaispyörimismääränsä olisi nolla. Analyysi osoitti, että pyörimismäärän minimointi ei ole yleispätevä tapa ehkäistä kaatumista. Keskushermoston täytyy siten käyttää kaatumisen ehkäisemiseksi jotain muuta kontrollikriteeriä pyörimismäärän minimoimisen lisäksi tai sen sijasta. Tässä väitöskirjassa tasapainon hallinta kaatumista vastaan katsotaan tapahtuvan ylävartalon hallinnan kautta.

Tässä väitöskirjassa esitellään kävelyn tasapainoa eri kävelynopeuksilla kompastushäiriön avulla tutkittuna. Nopea kävely on aiemmin liitetty suureen kaatumisriskiin kompastumisen seurauksena. On epäselvää, onko suuri kaatumisriski selitettävissä epätasapainon lisääntymisellä. Tasapainoa suhteessa vakioituun kompastushäiriöön määritettiin ylävartalon välittömänä passiivisena vasteena häiriöön. Tätä varten analysoitiin nuorten, terveiden koehenkilöiden koehenkilökohtaisia, lihasohjattuja simulaatiota. Näissä simulaatioissa mallin heilahtavaan jalkaan suoritettiin lyhyitä häiriöitä useita kertoja heilahdusvaiheen aikana kohdistamalla siihen taaksepäin suuntautuva vakiovoima. Kyseinen kompastushäiriö paljasti, että kävelynopeudella on heikko ja vaihteleva vaikutus kävelyn tasapainoon. Nopea kävely ei siten välttämättä ole epätasapainoisempaa kuin hidas kävely.

Väitöskirjassa tarkastellaan myös tutkimuksia, jotka käyttivät kompastushäiriötä kävelyn heikentyneen tasapainon havaitsemiseen. Sitä varten analysoitiin kompastushäiriöllä CP-vammaisten ja vammattomien lasten koehenkilökohtaisia lihasohjattuja kävelyn simulaatioita. Tulokset osoittavat, että vammattomat lapset ovat tasapainossa suuremman osan jalan heilahdusvaiheesta kuin CP-vammaiset. Täten CP-vammaiset ovat herkempiä kompastushäiriölle, ja näin ollen todennäköisemmin herkempiä kaatumaan kuin vammattomat.

Väitöskirjassa tarkastellaan myös ylävartalon normaalia hallintaa kävelyn aikana. Normaalin hallinnan tutkimisella saadaan selville hyvän hallinnan tunnusmerkit. Näihin tunnusmerkkeihin voidaan pyrkiä heikentyneen hallinnan kuntouttamisessa. Heikentynyt ylävartalon hallinta kävelyn aikana liittyy ikääntymiseen ja moniin sairauksiin. Tämä on huolestuttavaa, koska sen voidaan katsoa lisäävän kaatumisriskiä. Lihakset, jotka osallistuvat ylävartalon hallintaan normaalin kävelyn aikana, voivat osallistua siihen myös ulkoisen häiriön kohdistuessa kävelyyn, pyrkien välttämään uhkaavan kaatumisen. Lihasohjattujen simulaatioiden analyysi osoitti, että vatsa- ja selkälihaksilla on suuri osuus ylävartalon hallinnassa koko askelsyklin ajan sekä sagittaali- että frontaalitasossa. Proximaalisilla jalkalihaksilla on suurempi osuus ylävartalon hallinnassa sagittaalitasossa kuin distaalisilla lihaksilla, mutta frontaalitasossa niin proximaalisilla kuin distaalisillakin jalkalihaksilla on suuri osuus.

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