POSTURAL CONTROL IN RESPONSE TO EXTERNAL PERTURBATIONS: BIOMECHANICS OF EXTENDED SUPPORT

Jernej Čamernik

Doctoral Dissertation
Jožef Stefan International Postgraduate School
Ljubljana, Slovenia

Supervisor: Prof. Jan Babič, Jožef Stefan Institute, Ljubljana, Slovenia

Evaluation Board:
Prof. Igor B. Mekjavić, Chair, Jožef Stefan Institute, Ljubljana, Slovenia
Prof. Nejc Šarabon, Member, University of Primorska, Faculty of Health Sciences, Koper, Slovenia
Prof. Elmar Rueckert, Member, University of Luebeck, Luebeck, Germany

MEDNARODNA PODIPLOMSKA ŠOLA JOŽEFA STEFANA
JOŽEF STEFAN INTERNATIONAL POSTGRADUATE SCHOOL

Jernej Čamernik

POSTURAL CONTROL IN RESPONSE TO EXTERNAL PERTURBATIONS: BIOMECHANICS OF EXTENDED SUPPORT

Doctoral Dissertation

NADZOR ČLOVEKOVE DRŽE PRI ODZIVU NA ZUNANJE MOTNJE: BIOMEHANIKA RAZŠIRJENE OPORE

Doktorska disertacija

Supervisor: Prof. Jan Babič

Ljubljana, Slovenia, January 2020

To my family.

Acknowledgments

This dissertation would not have been possible without the help and support of a number of people over the last five years.

I would like to express my deepest appreciation to my supervisor, professor Jan Babič, for his continuous support of my PhD study and related research, for his patience, motivation, and immense knowledge. Thank you for giving me the unique opportunity to undertake these doctoral studies and work in your group. I deeply appreciate all the scientific ideas, concepts and solutions you have shared with me.

I thank my fellow labmates for the stimulating discussions, and for all the fun we have had in the last five years.

I am thankful to my parents (Brigita and Ciril) and in-laws (Marija and Stojan) for giving me the support to accomplish my ambitious goals.

Most of all, I would like to thank my family: my wife Katarina for her unwavering and selfless support, and my three shining stars who filled me with energy in times when I needed it the most.

Ljubljana, January 2020

Abstract

To reduce the risk of a fall while standing on a moving bus or climbing down the stairs we intuitively use handles and handrails to augment our balance control. However, a certain level of adaptation in posture and motor behavior needs to be developed to efficiently use these aids. This thesis characterizes the influence of holding a handle and making hand contacts with the environment on whole-body posture and balance recovery in young adults. A literature review of common strategies for balance recovery and using handles to augment balance control led to the design of perturbation-based experiments with the aim to explore balance recovery and adaptation in whole-body posture and to accomplish the following research goals:

1. To investigate the effects and the underlying mechanism of a supportive hand contact on postural control during a prolonged exposure of standing balance to perturbations.

2. To examine the mechanisms of assisted balance recovery and the associated loads on the body when holding a handle mounted at different heights or standing in different configurations.

3. To investigate the effects of active choice of supportive contacts on motor control.

During continuous perturbations to balance we compared whole-body adjustments and adaptation in postures with and without additional handle support. A handle was used to mainly reduce the movement of the body in the posterior direction and merely control the movement in the anterior direction. Furthermore, whole-body posture, measured by changes of joint angles, was adjusted to control balance and muscle activity of all postural muscles highly increased when no handle was used.

By applying an anterior force to the waist we investigated the effectiveness and mechanisms of balance recovery in five different postures: step stance and normal stance with or without holding to handles at different heights. Different posture configurations showed that both step stance and holding to handles at different vertical positions sufficiently assist balance recovery, compared to normal stance. While there was no significant effect of handle on displacement of the center of mass, the shoulder height handle required the lowest handle force, indicating a differential use of the handle. The total body effort was also reduced with the use of the handle, and the least effort to sustain balance was made by using the low and shoulder height positioned handle.

To study the effects of contacts an innovative full-body experimental paradigm was established. To examine adaptation during a reaching task, non-trivial postural perturbations of the subjects' support base were systematically introduced. Subjects adapted to the perturbations by establishing target dependent hand contacts. Moreover, the trunk motion adapted significantly faster than the motion of the arms. The most striking finding was that observations of the initial phase of the left arm or trunk motion (100–400 ms) were sufficient to faithfully predict the complete movement of the right arm. Overall, the results suggest that the goal-directed arm movements determine the supportive arm motions and that the motion of heavy body parts adapts faster than the light arms.

Povzetek

Na premikajočem se avtobusu ali pri hoji po stopnicah ljudje intuitivno uporabljamo razna oprijemala, ograje in ročaje. Z uporabo tovrstnih pripomočkov izboljšamo kontrolo ravnotežja in s tem zmanjšamo možnosti za morebiten padec. Za učinkovito uporabo teh pripomočkov pa mora človek ustrezno prilagoditi svojo držo in usvojiti določeno gibalno obnašanje. Pričujoča doktorska naloga se nanaša na raziskovanje človekovih gibalnih sposobnosti s poudarkom na proučevanju vpliva uporabe ročajev in vzpostavljanju kontaktov z objekti v njegovi bližini z namenom uravnavanja ravnotežja. Po pregledu obstoječe literature na temo gibalnih strategij za nadzor drže in ohranjanje ravnotežja so bile zastavljene tri študije, s katerimi sem želel doseči naslednje raziskovalne cilje:

1. Raziskati učinke in mehanizme nadzora drže in ravnotežja v stoji v primerih, ko je človek izpostavljen dlje časa trajajočim motnjam ravnotežja.

2. Ugotoviti, na kakšen način višina postavljenega ročaja in posledično različne drže človeka v primeru nenadnih, sunkovitih motenj ravnotežja vplivajo na nadzor ravnotežja in obremenitve na telo.

3. Ugotoviti učinke uporabe opornih kontaktov (naslanjanje na mizo) na motorično kontrolo in prilagajanje.

V prvi študiji smo neprekinjeno motili ravnotežje preiskovancev v dveh različnih pogojih – držanje za ročaj in prosta stoja. Ugotovili smo, da so preiskovanci aktivno uporabljali ročaj predvsem v smislu zmanjšanja gibanja telesa v smeri nazaj, medtem ko je bilo gibanje v smeri naprej podobno kot v primeru proste stoje. Razlike smo ugotovili tudi v drži preiskovancev, ki smo jo opredelili s pomočjo izmerjenih kotov v sklepih (posledica je bila znižanje težišča telesa v stoji prosto), ter v mišični aktivnosti (bistveno nižja pri pogoju, ko so se preiskovanci držali za ročaj).

V drugi študiji smo z nenadnimi sunki sile motili ravnotežje preiskovancev in preverjali vpliv višine postavljenega ročaja na nadzor ravnotežja. Poleg drže z uporabo ročaja, nameščenega na treh različnih višinah, so preiskovanci opravili poizkus tudi v dveh držah brez držanja za ročaj – v razkoraku in prosti stoji. Ugotovili smo, da sta stoja v razkoraku ter držanje za ročaj učinkoviti strategiji za ohranjanje ravnotežja, vendar z različnimi vplivi na obremenitev telesa. Le-ta je precej nižja predvsem v primeru uporabe ročajev, postavljenih v višini bokov, in ramen, kjer smo izmerili najnižjo silo, s katero so preiskovanci delovali na ročaj.

Za raziskovanje učinkov rabe opornih kontaktov, ki jih dnevno uporabljamo, ko se na primer naslanjamo na kuhinjski pult, medtem ko segamo po kozarcu na vrhnji polici, smo za tretjo študijo pripravili inovativno eksperimentalno okolje. Ob naslonu na mizo in seganju do oddaljene tarče, izrisane na zaslonu pred preiskovancem, smo hkrati sistematično motili njihovo ravnotežje s pomočjo premikanja površine, na kateri so stali. Ugotovili smo, da so preiskovanci prilagodili svoje gibanje na način, da so mesto naslona na mizo prilagodili glede na pozicijo izrisane tarče na zaslonu. Poleg tega smo ugotovili, da se je prilagoditev

drže najprej zgodila v trupu in šele nato v gibanju roke. S pomočjo novo razvite metode pa smo lahko na podlagi premika leve roke ali trupa v prvih 100–400 ms uspešno določili celotno gibanje desne roke.

Contents

List of Figures

List of Tables

Abbreviations

2D	...	Two-Dimensional
3D	...	Three-Dimensional
AP	...	Anterior-Posteriror
BCa	...	Bias-Corrected and Accelerated Bootstrap
BOS	...	Base of Support
CI	...	Confidence Interval
CNS	...	Central Nervous System
COM	...	Center of Mass
COG	...	Center of Gravity
COP	...	Center of Pressure
DOF	...	Degrees of Freedom
EMG	...	Electromyography
ES	...	Equilibrium Score
HF	...	High Frequency
LF	...	Low Frequency
MDF	...	Median Power Frequency
MF	...	Medium Frequency
MNF	...	Mean Power Frequency
NH	...	No-Handle Trial
PSD	...	Power Spectral Density
PSI	...	Postural Stability Index
UCM	...	Uncontrolled Manifold
WH	...	With-Handle Trial

Glossary

Adaptation: Any alteration in the structure, form or mechanisms, by which the organism becomes better suited to handle a new or changed environment.

Center of Mass: A point equivalent of the total body mass in the global reference system and the weighed average of the center of mass of each body segment in 3D space.

Center of Gravity: The vertical projection of the center of mass onto the ground.

Center of Pressure: The point location of the vertical ground reaction force vector. It represents a weighted average of all the pressures over the surface of the area in contact with the ground.

Feedback: A control system which senses the difference between actual and desired states, and produces counteractive effects to minimize the difference.

Feedforward: A control system involving an anticipatory output or response to an impending disturbance, generally with the aim to reduce the differences between the actual and desired states.

Perturbation: A disturbance causing the body to deviate from its regular movement.

Postural control: The act of maintaining, achieving or restoring a state of balance during any posture or activity.

Chapter 1

Introduction

1.1 Motivation

The ability to move voluntarily is one of the key differences between the living and non-living organisms. Human beings, as one of the many representatives of the living organisms, have evolved this characteristic to a remarkable state. Standing and walking on two feet, picking, holding, throwing, and manipulating with objects are just a few of the basic motor skills we learn in our early childhood. Through millions of years of evolution, the human ability to move evolved from quadrupedal to bipedal locomotion, which brought some important benefits. The legs took over the function of the primary way of moving in the environment and the arms took over the function of primary manipulators of the objects in the environment. However, the transition from quadrupedalism to bipedalism came at a cost of decreased postural stability. In order to be able to maintain stability, the central nervous system (CNS) developed such control mechanisms and strategies that can successfully cope with the unpredictable nature of human environment.

To study and understand the phenomena of the human control system that integrates sensory information and simultaneously coordinates the motion of the body requires an interdisciplinary approach. Motor control, as a field of science, aims at discovery of laws of nature describing the interactions between the central nervous system, the body, and the environment during the production of voluntary and involuntary movement (Latash & Zatsiorsky, 2015). Therefore, the scientific fields that contribute the most to the understanding of human motor control include physics, engineering, statistics, behavioral science, cognitive science, human factors, physiology, neuroscience and medicine.

Postural control, roughly described as the ability to control our body's position in space, is fundamental to everything we do. Hence, it is not surprising that this key area of human motor control interested many researchers. There have been numerous studies in the past decades that aimed to decipher the functioning and behavior of the human central nervous system with regard to postural control. The results of such studies can be used to improve human rehabilitation techniques or treatment of postural control disorders, which for example affect patients with osteoarthritis, patients with Parkinson's disease or children with developmental coordination disorder. Furthermore, the indisputable fact that the global population is increasingly aging (United Nations, 2017) goes in parallel with the increase of fatal falls rates (World Health Organization, 2007). Therefore, insight in balance control is vital in the development and production of more efficient orthoses and prostheses and support aids such as crutches and handrails.

In the following review of literature on postural responses to stabilize balance and prevent a fall, researchers underline the importance of supportive contacts and address many of the underlying mechanisms. However, there are still a number of important questions and topics that remain inconclusive. Specifically, (1) adaptation of postural control to continuous perturbations was only partially explained, (2) experimentally supported recommendations regarding optimal handles locations for balance recovery have not yet been given, (3) the workload on the human body has not yet been fully explained and (4) the question of how the coordination of supportive contacts and primary motor tasks develops or adapts remains only partly answered.

The main purpose of this dissertation is to form a higher level of understanding of how additional supportive contacts affect human motor control in balancing tasks. Specifically, we studied the mechanisms and strategies of postural control in response to external perturbations and how holding a handle or establishing hand contacts in such situations augments and/or facilitates balance control.

1.2 Thesis Structure

This thesis outlines a series of studies conducted to assess postural control mechanisms which act in situations when additional arm support is needed to prevent extensive movement of the body which could lead to a fall. In the following chapter, the thesis shortly reviews the past and recent findings regarding the control of stance in healthy adults, with an emphasis on studies that assessed postural adjustments (reactive and anticipatory) and adaptation in standing with additional hand support (i.e. holding a handle). The review covers the most commonly used methods for studying postural control and outlines the concepts of postural control strategies and synergies. In Chapter 3, the aims and objectives are presented. Chapters 5 to 7 include publications of the three relevant studies, including additional analyses and their conclusions. In Chapter 8, a summary of the conclusions from all three studies is given. Appendix A includes the medical ethics approval.

Chapter 2

Review of Literature

This review of literature begins by defining postural control and an overview of the main descriptive measures typically observed and analyzed in context of balance and postural control. It then continues with an overview of balance recovery strategies in standing and discusses the main factors which contribute to appropriate and effective responses to balance perturbations. Current research on the role of the arm support in standing (i.e. holding a handle) to prevent a fall or to recover from perturbations, which is also the topic of this thesis, is discussed in the context of a strategy of the extended base of support.

2.1 Postural Control

Human upright stance is inherently unstable because gravity continuously endangers body equilibrium. The task of maintaining equilibrium while standing is therefore very demanding and complex even without any external perturbations. There is no universal definition of postural control, however, maybe one of the most comprehensive is the one from Shumway-Cook and Woollacott (2012). In their book they define postural control as a complex motor skill which integrates neural control of postural equilibrium (stability) and postural orientation.

- **Postural stability** can be defined as a state in which the forces acting on the body are in balance, so that the body is at rest in a predetermined position. It involves coordination of sensory and motor strategies to maintain balance, that is, to stabilize the body's center of mass over its base of support. An important goal of postural stability control is to prevent falls during both self-initiated and externally triggered perturbations that affect balance. In the sense of control, postural stability can be further classified in-to static (stability) and dynamic (controlled mobility) postural control (O'Sullivan et al., 2014). Static postural control is defined as the ability to maintain the stability and position of the vertical projection of the COM within the BOS while the body is at rest, while the dynamic postural control has the same function, but in condition while parts of the body are in motion.

- **Postural orientation** relates to the active alignment of the trunk and head with respect to gravity, support surfaces, the visual surround, and internal references. Additionally, it integrates the sensory information from somatosensory, vestibular and visual systems.

In humans the goals of postural stability and postural orientation are independently controlled, however, interconnected through the ongoing task. Such control allows giving up one goal for another in order to be able to successfully complete a certain task. For example, a gymnast may give up the goal of orientation in order to achieve their goal of stable postural equilibrium at landing on a soft surface after a somersault. Given this example, postural control can be explained as an interaction of the individual (gymnast), the ongoing task (landing after a somersault) and the environmental constraints (soft landing surface).

Measures of Postural Stability.

The **center of mass (COM)** is a point that represents the average position of the body's total mass 2.1. The location of the COM in the human body is not fixed but depends on postural orientation. When standing with arms beside the body, it is located in the abdomen approximately 20 mm in front of the second lumbar vertebra. When flexed at the hips, however, the COM moves forward to a position outside the body. It is calculated by

$$\mathbf{r} = \sum_{i=1}^{N} \frac{m_i}{m} \mathbf{r}_i \tag{2.1}$$

where $\mathbf{r}$ is the vector of the COM of the body, N is the number of rigid body segments, m_i is the mass of the i-th segment, m is the total mass of the body and $\mathbf{r}_i$ is the vector of the COM of the i-th segment.

The central nervous system must control the position and motion of the body's COM as well as the body's rotation about its COM. Therefore, for a stable pose, the CNS must react in a way to control the gravity vector of the COM which must be pointing within the base of support and control the dynamic effects, i.e. COM velocity (Pai et al., 2000). It is the interaction between these two variables, rather than just the position of the COM alone, that determines whether a person will be able to remain stable within his or her current BOS or be required to take a step or reach for support in order to regain stability.

Based on these propositions some researchers of human postural control used COM position as a measure of postural stability (Maki et al., 2003). Moreover, it is hypothesized that the COM is the primary variable controlled by the nervous system during postural control (Scholz et al., 2007).

However, a mere observation of the COM position only gives information about body stability in static conditions. In case of dynamic stability one must also consider inertial forces and moments acting on the body. A constantly present force that pulls on all body segments, but acts through the COM, is the force of gravity. While standing or walking, the force of gravity is opposed by the ground reaction force, which pushes upward against each foot. The net ground reaction force occurs at an imaginary point on the ground called the **center of pressure (COP)**. It represents a weighted average of all the pressures over the surface of the area in contact with the ground and it is totally independent of the COM (Eq. 2.2) (Winter, 2009). If one foot is on the ground, the net COP lies within that foot. If both feet are in contact with the ground, the net COP lies somewhere between the two feet, depending on the relative weight taken by each foot. Thus, when both feet are in contact with the support surface, there are separate COPs under each foot. Because this can be easily measured by the force plate or with force sensors mounted or integrated in the shoes, it is the most commonly used stability measure in biomechanics for studying human walking and stability (Winter, 1995). On the flat surface, when all horizontal forces are compensated by the friction force, the COP point is calculated as:

$$\mathbf{p}_{COP} = \frac{\int_A \mathbf{p}\rho \mathrm{d}A}{\int_A \rho \mathrm{d}A} \tag{2.2}$$

where A is the surface area of the contact (support polygon), $\mathbf{p}$ is the location and ρ is the pressure of the infinitesimal calculus of the supportive surface area dA.

Aside from observing COM an COP characteristics, clinicians often use other measures for assessing stability like the Equilibrium Score (ES) or Posture Stability Index (PSI) Chaudhry et al. (2011). The ES is calculated by using specialized equipment for assessing stability like the Balance Master or Equitest, where the maximum anterior and posterior sway angle is measured. These parameters are measured in several conditions and assigned a score based on their values and the norm (i.e. 12.5° is assumed to be the limit of stability for a normal individual). The composite ES is then assumed from the weighted average of all scores. The main disadvantages of the ES are that the calculation does not include important biomechanical information of the subject (height, weight) and that the limits of stability may vary significantly from the norm used in the calculation. (Chaudhry et al., 2005) proposed an alternative measure, the PSI. This measure overcomes the limitations of the ES, as it depends on the entire sway history, and on the individual's body mass, height and the torque at the ankle. Nevertheless, this measure is limited for use only in conditions of a balancing strategy where the control of balance is maintained mostly by regulating the torque in the ankle joints.

Other Measures and Equipment for Analyzing Postural Stability

As already noted, both, the kinetic (or dynamic) quantities (which refer to the force and energy associated with movement) and kinematic ones (which refer to aspects of movement besides the consideration of mass and force), must be considered in the analysis of postural control. They are measured using different transducers (goniometers, accelerometers, dynamometers, etc.) and specific equipment, such as force plates and motion capture systems. Kinematic quantities (displacement, velocity and acceleration) are used to characterize local movements, i.e. segmental movements. Kinetic quantities (force, reaction force and torque) are used to measure the whole body movement. In particular, force plates, which measure support reaction forces, allow to determine the resultant forces and torques, as well as the COP. The resultant force expresses the center of gravity linear acceleration to within the body mass, and angular acceleration to within the body moment of inertia in relation to the center of gravity. Reaction forces (and torques) at joint level, however, can be noninvasively estimated only from a model (forward dynamics or inverse dynamics modeling). Besides kinetic and kinematic quantities, the measurement of activation of individual muscles is also a commonly used method. Surface electromyography is a unique method for specifying muscle activation since it measures the electrical impulses of muscles at rest and during contraction. Picking up the electrical signals by surface electrodes is a convenient way of determining muscle excitation and allows a description of muscular activation patterns.

Postural Sway

Muscle tone of the postural muscles has to adjust flexibly to changes in support, alignment and environmental conditions (Gurfinkel et al., 2006). The human musculoskeletal system therefore requires an active activation of postural muscles in order to maintain stance. This constant activation is needed to oppose the destabilizing effects of gravity and to orient the body segments appropriately to the environment. Therefore, while standing, humans make constant corrections by actively moving their body. These small movements are called postural sway. Maintaining balance while standing requires keeping the downward projection of the COM within the base of support (BOS), an imaginary area defined by those parts of the body in contact with the environment (2.1). Because the body is always in motion, even during a stable stance, the COM continually moves about with respect to the BOS. Postural instability is determined by how fast the COM is moving toward the boundary of its BOS and how close the downward projection of the body's COM is to the boundary.

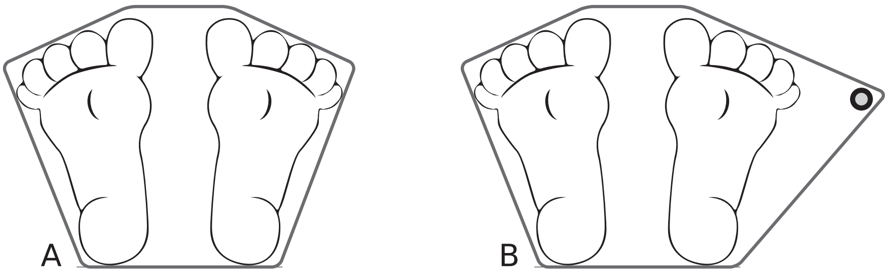

Figure 2.1: Base of Support. A schematic illustrating the area of base of support during standing (A) and how the use of a cane or handle can increase the stability by increasing the effective size of the base of support (B).

Integration of the Sensory Systems

Sensory integration refers to the process by which kinematic (orientation and motion) and kinetic (force-related) information from multiple sensory sources is combined in the nervous system for the purpose of generating motor action to compensate for the destabilizing effect of gravity and to resist external perturbations (Peterka, 2018). Sensory information from the somatosensory, visual and vestibular systems must be integrated in order to interpret complex sensory environments, because information from a single sensory channel can be ambiguous and misleading (Horak & Macpherson, 1996).

Somatosensory inputs are important for triggering the earliest automatic postural responses in response to external perturbations. They include pressure information from skin in contact with surfaces, limb segment orientation from muscle proprioceptors and joint receptors, as well as muscle length, velocity and force information. Somatosensory inputs are also important for providing information about the direction of perturbation and about the texture and stability of the support surface so that appropriate postural strategies can be selected. Somatosensory inputs can provide confusing, ambiguous information about the body center of mass motion because they cannot distinguish between body motion over a stable surface and surface motion under a stable body, such as when standing on a moving boat or pier (Horak & Macpherson, 1996).

Inputs from the **vestibular system** are important for orientation of the trunk and head to gravity, especially when the surface is unstable. However, the inputs from this system are not necessary to trigger automatic postural responses to external perturbations (Mergner et al., 2003). The sensory organs of the vestibular system – otolith organs (utricle and saccule) and semicircular canals – detect linear and angular motion of the head relative to space, as well as gravitational forces (Fox, 2011). Vestibular inputs can provide confusing, ambiguous information about the body center of mass motion because they cannot distinguish on their own between head motion over a stable body and head motion accompanying the body center of mass motion. Vestibular information is thought to help the somatosensory system distinguish a stable from an unstable surface and then become increasingly important for controlling postural orientation the more unstable is the surface.

Visual information provides knowledge of body sway and orientation in the environment and provides advanced information about potentially destabilizing situations. Vision can provide information about the direction and speed of body sway. It also allows perception and body orientation with respect to the vertical and horizontal visual environment. Thus, standing subjects exposed to slowly moving visual surrounds will sway with reference to the visual motion, even when unaware of it (Peterka, 2002).

The adjustment of the relative and absolute contribution of sensory inputs to control of posture by the central nervous system is called sensory reweighting. Researchers describe this term as an active feedback mechanism when the nervous system adjusts the relative and absolute contribution of sensory inputs depending on environmental conditions and subjects' perception (Assländer & Peterka, 2014; Peterka, 2002; Peterka & Loughlin, 2004).

2.2 Postural Control Responses to External Perturbations

Generally, researchers of postural control differentiate between two bodily responses to perturbations of balance: **reactive** and **anticipatory** responses.

2.2.1 Reactive Control of Balance

Postural control responses to unexpected perturbations of stance have been shown to be automatic, repeated and highly stereotyped in humans (Diener et al., 1988). Reactive, involuntary responses are triggered between 80-120 ms after the start of perturbation (Nashner & Cordo, 1981). This latency is faster than the fastest voluntary postural reactions, but slower than the fastest muscle stretch reflexes which are triggered by muscle spindles.

Automatic postural responses include responses in muscles that are shortened as well as stretched far from the site of perturbation that can exert torque against surfaces to correct posture (Ting & Macpherson, 2005). The recruitment of muscles in postural response depends on the goal of maintaining equilibrium and not on stereotyped reflexes. Automatic postural responses depend on the central set, so they are specific to the conditions of support and adapt to prior experience. The central set is the readiness of the central nervous system for an upcoming event based on initial conditions, prior experience and expectations. For example, leg muscles are activated in response to surface perturbations during free stance but arm muscles are activated and leg muscles suppressed in response to surface perturbations when holding onto a stable support (Cordo & Nashner, 1982).

In addition, muscles in the back of the legs are activated in response to forward body sway while standing but muscles on the front of the legs and in the arms are active when supported on the hands and feet (Macpherson et al., 1990). Postural responses change even in the first trial after a change in body configuration but continue to optimize the response for the particular conditions. For example, a gradual adaptation of the postural response can be observed during repeated trials of surface rotation. In response to the first rotation, a destabilizing response may be seen in the stretched ankle extensor muscle but with repeated rotations this activation of the extensor is suppressed and activation of the stabilizing ankle flexor gradually increases. Subjects can also influence which postural response is selected and the magnitude of their response based on experience, expectations and intention (Horak & Nashner, 1986).

Poor coordination of automatic postural responses can result in failure to return to equilibrium in response to external perturbations. Automatic postural responses can be defined by their **postural strategies** and **postural synergies**.

2.2.1.1 Postural Movement Strategies for Balance Recovery

The concept of postural strategies emerged as investigators struggled with a way to describe general sensorimotor solutions to the control of posture, including not only muscle synergies but also movement patterns, joint torques, and contact forces (Horak et al., 1997). Strategy selection and modulation are dependent on:

- the features of the perturbation (timing, direction, magnitude, predictability),
- the central set of the individual (affect, arousal, attention, expectations, prior experience),
- ongoing activity (cognitive or motor) and environmental constraints (on reaction force generation and limb movement).

Researchers identified two main types of movement strategies that can be used to return the body to equilibrium in a stance position: strategies that keep the feet in place and return the center of mass back over the base of foot support and the other strategy that changes the base of support through individual stepping or reaching (Horak & Nashner, 1986; Maki & McIlroy, 1997; McIlroy & Maki, 1996). These strategies can be characterized by their different muscle synergies, kinematics and joint torques.

Fixed-support Strategies (feet in place)

The **ankle strategy**, in which the body moves at the ankle as a flexible inverted pendulum, is appropriate to maintain balance in small amounts of sway when standing on a firm surface. To validate this assumption Horak and Nashner (1986) exposed standing subjects on a normal support surface to brief forward and backward horizontal surface perturbations. These perturbations elicited relatively stereotyped patterns of leg and trunk muscle activation with latencies from 73 to 100 ms. Activity began in the ankle joint muscles and then radiated in sequence to thigh and then trunk muscles on the same dorsal or ventral aspect of the body. This activation pattern exerted compensatory torques about the ankle joints, which resorted equilibrium by moving the center of mass forward or backward.

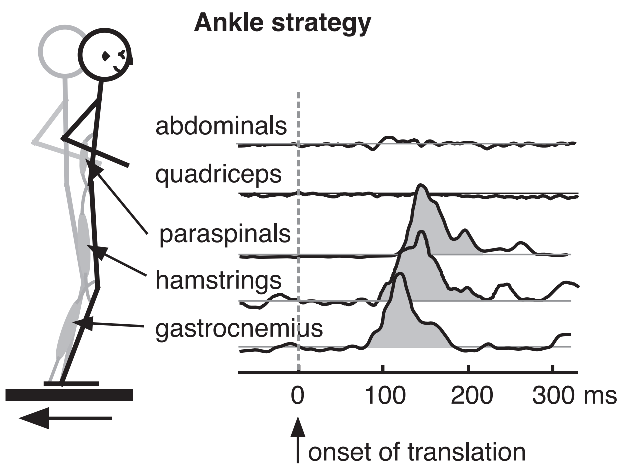

Figure 2.2: Ankle Strategy. When the subject stands on a wide platform that is moved backward, his posterior muscles – paraspinals, hamstrings, and gastrocnemius – are activated to bring the body back to the erect position by rotating at the ankles. The dashed vertical line indicates the onset of platform acceleration. Figure from Ting (2007).

The **hip strategy**, in which the body exerts torque at the hips to quickly move the body COM, is used when persons stand on narrow or soft surfaces that do not allow adequate ankle torque or when the COM must be moved quickly (e.g. as a response to larger perturbations) (Horak & Nashner, 1986).

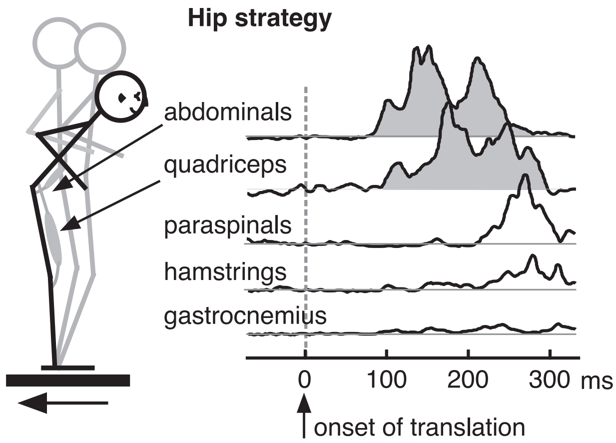

Figure 2.3: Hip Strategy. When a subject stands on a narrow beam that is abruptly moved backward, the anterior muscles – abdominals and quadriceps – are recruited to flex the trunk and extend the ankles, moving the hips backward. The dashed vertical line indicates the onset of beam acceleration. Figure from Ting (2007).

When subjects suddenly change from standing on a wide to a narrow surface, or vice versa, there is a gradual adaptation from an ankle to a hip strategy and vice versa with repeated perturbations. This gradual change in postural strategies suggests that they not only depend upon sensory feedback and current sensory conditions but also upon prior conditions based on the central set (Horak & Kuo, 2000). Furthermore, both strategies can also be concurrently activated, thus producing a continuum of possible postural corrections representing different mixtures of the two strategies (Creath et al., 2005; Runge et al., 1999).

Compensatory Stepping, Reaching, Touching and Holding

Typical human actions like using a handle while riding a bus, reaching for a glass on the top shelf using a supporting surface or making a step after a trip, involve coordination of multiple joints, significant shift of the center of mass, and control of equilibrium, either in static or dynamic conditions. These are skills learned early in human childhood but studied extensively in the context of motor disability, e.g. after neurological insults like stroke (Arienti et al., 2019), or in the elderly with reduced muscle power, joint flexibility and sensory loss (Correia et al., 2016; Dunsky et al., 2017; Osoba et al., 2019; Robinovitch et al., 2013).

Compensatory stepping. The occurrence of a step has typically been regarded as an inevitable consequence of the horizontal movement of the center of mass beyond the limits of the base of support. Taking a step to recover equilibrium is therefore common, especially during gait and when keeping the feet in place is not important. However, even when persons step in response to an external perturbation, they first attempt to return the COM to the initial position by exerting angle torque. To explore this strategy, forward stepping has been evoked by various sudden perturbations such as by pulling on a cable attached by the subject's waist (Mansfield & Maki, 2009; Mille et al., 2013; Pidcoe & Rogers, 1998) or by platform translations (Jensen et al., 2001; Maki et al., 2003). The studies show that the timing of the perturbation-evoked stepping response is faster in trials when the subjects were free to respond naturally, compared with trials where the subjects were instructed to step as quickly as possible. A compensatory stepping movement is typically completed in about 500 ms, which is approximately half the time required to step as fast as possible in reaction to visual cue.

Light touch. Making firm and supportive contacts with stationary objects while standing and walking (e.g. leaning on a cane or using a walker) increases the stability limits which are defined by the base of support (Bateni & Maki, 2005). However, researchers have shown that also a small force (< 1 N) applied by a stable support on different parts of the body (leg (Rogers et al., 2001), finger (Jeka, 1997; Jeka & Lackner, 1994; Kouzaki & Masani, 2008; Lackner et al., 2001), shoulder (Johannsen et al., 2007; Rogers et al., 2001), neck or head (Krishnamoorthy et al., 2002)) provides an additional sensory input, which helps with orientation of the body in space. The sensory information coming from this light touch has been proved to improve postural control in quiet standing by reducing the amplitude of center of pressure (COP) movement and muscle activity (Wing et al., 2011).

Krishnamoorthy et al. (2002) found that the light touch to the head or neck reduced body sway more effectively than a free finger touch to a pad. However, the effect of light finger touch improved significantly when the finger was fixed in a clip or when the subjects were holding a load suspended using a pulley system. These findings indicate that factors such as stability of the contact point and the transient force changes at the point of contact play a crucial role in the control of postural sway.

To determine whether the effect of the light touch is due to the tactile sensory input, as opposed to mechanical support, Kouzaki and Masani (2008) investigated the light touch effect on postural stability during quiet standing with and without somatosensory input from the fingertip. They modulated the somatosensory input by applying a tourniquet on the finger to induce ischemia and consequently a loss of tactile sensation in the finger. They found that the light touch effect (i.e. reduced mean velocity of COP) was absent due to the loss of finger tactile feedback, even though the mechanical support remained the same.

Rabin et al. (2008) concluded that the control of medial–lateral sway while lightly touching a stable surface is controlled based on information about the orientation of the arm. In their experiment they perturbed the arm proprioception by applying vibration to the upper arm flexor (biceps brachii) of the subjects and found that postural control degraded in a sense of increased postural sway.

Due to the plethora of findings coming only from a few studies of light touch mentioned here, it is clear that they substantially contribute to the understanding of postural control in general and especially of the importance of the somatosensory feedback.

Reach to grasp. A common change-in-support strategy is reaching toward a railing or other stable surface. The initiation of reaching reactions is even faster than stepping reactions. When a person reaches and grasps a stable nearby object, the additional support extends their base of support and consequently provides more options to stabilize the movement of the center of mass (McIlroy & Maki, 1995). The activation of the compensatory arm movement typically begins at a latency of 80 to 140 ms after the perturbation onset. These rapid responses occur even though there is no initial stretch or loading of the muscles by the postural perturbation and the muscle is not actively participating in the maintenance of stability. A study by Corbeil et al. (2013) suggests that the arm motion is modulated to serve stabilizing and/or has protective functions which depend also on perturbation direction. When the subjects experienced a perturbation with a forward translation of the support surface, the forward-directed arm movements were evoked. This was consistent with the counterweight strategy, since the mass of the arms was used to return the center of mass within the base of support. However, in leftward falling motion situations, the subjects used a combination of two different arm strategies. They moved their left arm to their left side which is consistent with the protective strategy (Chiu & Robinovitch, 1998; van den Kroonenberg et al., 1996) and their right arm rapidly rightward as a counterbalancing measure.

Holding a handle. A great deal of research has been conducted on the topic of reactive responses to postural perturbations, but few experiments have been carried out to discover the effects of supporting contacts established prior the onset of the perturbations. Especially how these contacts affect balance control and postural adaptations during longer periods of time. A proactive handle contact can be classified as an anticipatory as well as a reactive strategy since the person is holding a handle before the actual perturbation to balance occurred. This type of support to balance control is beneficial in several ways: 1) it provides the person with a tactile feedback, especially relevant when the perturbations are small; 2) it eliminates the need for a quick and accurate reach for the handrail; 3) it reduces the time lag of the reach-to-grasp movement and minimizes the momentum of the COM following a destabilization; 4) reduces the forces needed to apply to the handrail to counteract the perturbation (Maki et al., 1998).

Holding a fixed handle significantly increases the base of support of a standing person and enables generating necessary forces on the handle to minimize the effects of perturbations. A finding from a recent study by Babič et al. (2014) revealed that the location of the supporting hand contact is important to maximize its stabilizing potential and that the peak forces exerted at the handle during the support surface perturbations are related to the location of the handle with respect to the user.

Reach by using supports. Human reaching with arm support has not been extensively studied. There is almost no literature on the issue of how humans use one hand to extend their reach space. For example, to lean forwards requires a shift of the trunk and a shift of the center of mass (Huang & Ahmed, 2011). At some point it becomes advantageous to use a supporting surface, allowing a reduction in anticipatory postural adjustments and a simplified control strategy (Slijper & Latash, 2000). But the decisions about when to implement support using one arm, which will depend on the availability, reliability and compliance of a support surface, are almost unstudied (Krishnamoorthy et al., 2004).

2.2.1.2 Postural Synergies

Coordinated movements require precise control of the many joints and muscles in the body. In this section I briefly review the concept of postural synergies which is used to explain the complex organization of motor actions as a consequence of the response of the postural system. According to Torres-Oviedo and Ting, 2007, a muscle synergy can be defined as a group of muscles activated in synchrony with fixed relative gains. Therefore, postural synergy is a preferred pattern of muscle co-activation that is used by the nervous system to maintain standing balance. Postural synergies are thought to simplify the neural control task of selecting and coordinating multiple muscles across the body by eliminating the need to control each muscle independently. Since the postural strategies define the overall goals involved in the maintenance of balance, the postural synergies define the muscle activation patterns that are used by the nervous system to implement them.

Researchers Ting and Macpherson (2005) studied how synergy activation levels and biomechanical signals in cats varied during the automatic postural response to linear translation of the support surface in 16 directions in the horizontal plane. They analyzed EMG responses of hind limb muscles using a nonnegative optimization procedure. They assumed that 1) any given muscle can belong to more than one synergy and 2) the muscles within a given synergy have a fixed proportional activation. The results showed that only four or five synergies are needed to account for the activation patterns of 15 hind limb muscles of the cat during automatic postural responses to many directions of platform motion. Activation of each synergy produces a unique direction of force against the ground, suggesting that postural control is based on task-related variables such as the force between foot and ground rather than the contraction force of individual muscles. In another study by Torres-Oviedo and Ting (2007) - this time performed on human subjects not cats- the authors hypothesized that a few muscle synergies could account for within-the-trial variability in automatic postural responses from different perturbation directions, as well as different postural strategies. The composition and temporal activation of several muscle synergies identified across all subjects in the study were consistent with the previously identified ankle and hip strategies in human postural responses. The results from their study also suggest that muscle synergies represent a general neural strategy underlying muscle coordination in postural tasks.

To parse complex data of many muscles activation patterns into a small number of components, Danna-dos-Santos et al. (2008), Torres-Oviedo et al. (2006), Wojtara et al. (2014) used several mathematical techniques. The basic approach in these experiments was to: firstly, measure EMGs from a large number of muscles during a complex behavior (or more than one behavior); secondly, use a computational analysis such as non-negative matrix factorization or independent components analysis to identify a set of synergies from the recorded EMGs; thirdly, evaluate whether the observed EMGs can be well described as the combination of these synergies; and fourthly, relate the identified muscle synergies to task-relevant variables (Tresch & Jarc, 2009).

Another approach to proving the existence of synergies was developed by Scholz and Schöner (1999). This approach is called the uncontrolled manifold (UCM). The UCM hypothesis states that the controller acts in a high-dimensional space of elemental variables and organizes in that space a subspace (the UCM) corresponding to a desired value of a particular performance variable. Further, the controller organizes co-variation among the elemental variables (across trials or across time samples) in such a way that their variance is mostly constrained to the UCM. This may be interpreted as stabilizing the performance variable.

2.2.2 Anticipatory Control of Balance

To maintain a desirable posture under a wide range of individual, task and environmental constraints, the central nervous system minimizes deflections of the body from a desirable orientation through the use of postural adjustments that fall into two general categories, compensatory (described in the previous chapter) and anticipatory (Santos et al., 2010b).

Compensatory postural adjustments are feedback-based mechanisms that are initiated by sensory events following the loss of desirable posture (Alexandrov et al., 2005; Park et al., 2004). On the other hand, the anticipatory postural adjustments (APA) accompany the voluntary movements and act in a predictive manner. Their role is to help counteract the postural destabilization with a forthcoming movement. These adjustments are activated as feed-forward postural control, prior to any sensory feedback indicating postural instability (Aruin & Latash, 1995; Li & Aruin, 2009). The process of generation of APA is likely to be affected by expected magnitude (Bouisset et al., 2000) and direction (Santos & Aruin,

2008) of the perturbation, voluntary action associated to the perturbation (Shiratori & Aruin, 2007), postural task and body configuration. For example, when reaching for an object such as a book on a shelf, muscles in the trunk and legs activate in advance of muscle activity and movement in the shoulder and arm (Aruin & Latash, 1995). These contractions in trunk and leg muscles constitute APAs because they precede the principal (focal) movement of the arm. A study by Santos et al. (2010a) revealed that unpredictable perturbations, which were not associated with any anticipatory corrections, induced large compensatory changes in the angular positions of the ankle, knee, and hip joints, and larger displacements of the COM followed by displacements of the COP. In contrast, APAs seen in conditions with predictable perturbations initiated COP displacements, resulting in better arrangement of the body position prior to the impact; this led to smaller compensatory COP-COM excursions and smaller displacements in the lower extremities joint angles after the impact.

In conditions of high instability demands, the central nervous system may suppress APA as a protection against their possible destabilizing effects. In fact, a relation between APA and compensatory postural adjustments has been demonstrated in the first part of a study by Santos et al. (2010a). Based on the results of their study, they are suggesting the existence of an optimal utilization of APA in postural control.

2.3 Experimental Paradigms for Studying Postural Control in Standing

Studies of postural control of bipedal stance are predominantly conducted by observing the human motor behavior as a result of unpredictable external postural perturbations. Experimenters observe and record body segment motions, reaction forces and torques between the feet and the ground when forces, acting externally to the joint(s) of interest, perturb balance. The observed connection between motor reactions to the given perturbation and the nature of perturbation itself is then used to identify certain patterns of CNS functioning.

Quiet Standing

A very common paradigm, which appears deceptively simple, is maintaining standing posture in the context of the steady force due to gravity (Collins & De Luca, 1994). Although gravity is constant, equilibrium is unstable and small fluctuations are seen in balance measures that reflect continuous and intermittent muscle activity. The so-called quiet standing has been the subject of many studies and usually involves measuring the displacement of the COP in both anteroposterior and lateral directions. A pattern of small corrections of the upright body position which are made to oppose the destabilizing effect of gravity is known as postural sway. Recently Sarabon et al. (2010) investigated the sensitivity of body sway parameters during quiet standing to manipulation of support surface size. Subjects had to perform different balance tasks (parallel stance, semi-tandem stance, tandem stance, contra-tandem stance, single leg stance) for a period of 60 s on a force plate. During the trials the total COP distance, the oscillations amplitude and the frequency of oscillations were measured. Their study revealed that the monitored parameters of the COP were highly sensitive to the support surface size manipulations. The parameters which describe the total COP distance were the most sensitive to changes in the balance task, whereas the frequency of oscillation proved to be sensitive to a slightly lesser extent.

Self-imposed Perturbations

The second paradigm involves perturbations to posture that result from the test subject's own actions like displacing the limbs in a forward extended position (Slijper & Latash, 2000; Wing et al., 1997). For example a study by Hatzitaki et al. (2005) addressed the aging effects on the development of postural synergies in response to self-imposed perturbations. Two groups of older and younger participants performed repeatedly for 5 s hip flexion/extension movements using a full range of motion and maximum velocity. They concluded that performance of repeated single limb oscillations revealed altered postural responses in older adults characterized by a limited range of motion, weaker postural synergies and increased ankle instability.

Support-surface Translations and Rotations

To induce these postural perturbations, a use of a movable platform on which a human subject is standing is commonly used. These support surface perturbations replicate the conditions of slipping, tripping, stepping on an irregular surface, or accelerating or decelerating the support surface during vehicular motion. Platform motions can be either translational or rotational.

During platform translation, the body's base of support moves to a new position relative to space, while, during rotation, the tilted support base remains stationary relative to space. Balance disturbance induced by platform translation would be greater than that induced by platform rotation. The balance reaction to platform translation is to move the COM forward or backward to the new position within the displaced base of support. For platform rotation, the balance reaction is to minimize the displacement of the COM and to restore postural stability on the tilted support base (Szturm & Fallang, 1998). A recent study by Chen et al. (2014) confirmed that the rotational and translational perturbations induce different responses. They showed that the translational and forward/toe-up (rotation) perturbations induce larger upper body movements and faster and larger horizontal displacement of the COM than the toe-down rotation. Therefore, the two types of perturbations provide different insights into CNS functioning and have to be studied accordingly.

The motion of the platform is usually specified by the displacement from the original position or orientation, maximum velocity and maximum acceleration and deceleration. The displacement of the platform can also be defined by a fixed duration of its predefined motion.

Waist-pulls

The application of the perturbation at the waist is another type of postural perturbation that is used in several postural control studies (Hsiao-Wecksler et al., 2003; Mansfield & Maki, 2009; Mille et al., 2013; Pidcoe & Rogers, 1998). The study from Mansfield and Maki (2009) about age-related balance reaction impairments compared support surface translation perturbation and weight-drop waist-pull perturbation. Their findings showed that the surface translation is more destabilizing than waist-pull and could potentially be more effective in revealing age-related deficiencies of postural control. Other authors like Mille et al. (2013) or Peternel and Babič (2013) used a motor-driven waist-pull mechanism with cables attached to the human waist. In the first case, the waist-pull mechanism was used to perturb the subjects in both anteroposterior and lateral directions to investigate the stepping response in two different age groups. In the second case, the motor-driven pulling mechanism was used to transfer the human postural skill to the humanoid robot. The mechanism was used to allow the human to perceive the postural perturbation experienced by the robot and extract the appropriate reactions.

Chapter 3

Aims and Hypotheses

The preceding sections of the introduction highlighted the various biomechanical, and sensory mechanisms involved in postural control during standing with and without the additional hand support. The studies of this thesis aim to contribute to the understanding of how additional supportive hand contacts affect human motor control in disturbed pure-standing and combined standing-motor tasks. Specifically, I focused on the mechanisms and strategies of motor control in balancing tasks that included holding a handle or establishing hand contacts which in some manner augmented and/or facilitated balance control.

The aims of each study were to:

- Investigate the effects and the underlying mechanism of a supportive hand contact on postural control during a prolonged exposure of standing balance to perturbations **Study I**.
- Faithfully categorize handle positions in connection with posture configurations, by accounting for whole-body movement and forces exerted on the handle **Study II**.
- Examine the mechanisms of assisted balance recovery and the associated loads on the body **Study II**.
- Investigate the effects of active choice of supportive contacts on motor control **Study III**.
- Determine whether the supportive hand motions leading to contacts can predict the goal-directed reaching movements **Study III**.

3.1 Hypotheses

H1 A supportive hand contact will have a significant influence on postural balance, evaluated with the movement of the COM and COP, during a prolonged exposure to external perturbations.

H2 Utilization of the hand contact will be more prominent for postural perturbations in the backward direction, which are more threatening than the perturbations in the forward direction.

H3 Using an additional hand support will not only change the posture of the human body while the hand is in contact with the handle, but will also contribute to a change after the release of the handle.

H4 Holding a handle, mounted at different vertical positions, amid discrete perturbations of the COM while standing, will affect the mechanisms of balance control differently.

H5 Holding a handle, positioned at shoulder height, would have the greatest positive influence (lowest forces exerted on the handle and minimal recorded movement of the participant) on balance control.

H6 The motion of the arm reaching toward a target will determine the supportive hand contact strategies.

H7 Both motor tasks, reaching to a target and establishing a supportive contact, will adapt to the novel environment during consecutive repetitions of the task.

H8 There is a significant correlation between the tasks of reaching toward a target and establishing a supportive contact which would reflect in synchronized motor executions and a correlation between target locations and supportive hand contacts.

Overall, the results of this dissertation that arose from the experimental studies can be applied to improve human rehabilitation techniques, treatment of various postural control disorders or to produce more efficient support aids such as wearable robots (exoskeletons). Furthermore, since falls are becoming an increasingly large problem, understanding of the underlying motor processes is crucial for designing efficient fall prevention aids and exercise protocols. Moreover, the same studies can potentially provide solutions to robot control as well. The emergence of service robotics was aimed to build robots to assist humans in various tasks in the human environment. In attempt for the robots to best fit into such an environment, researchers emulated the human body structure to build humanoid robots. The similar structure of humanoid robots offers both practical and social benefits. Yet the bipedal nature brings the same drawbacks as with humans. The intrinsically unrobust balance of such robots requires proficient control mechanisms and strategies dedicated to postural control. Findings from human postural control studies could provide a shortcut for achieving adequate humanoid robot control algorithms.

Chapter 4

Thesis at a Glance

Table 4.1: Thesis at a glance. Outline of the studies along with short summaries of their results and conclusions.

Study	Question	Methods	Results	Conclusions
I	How humans utilize arms to augment balance control?	COP, forces exerted on the handle, EMG of postural muscles and full-body movement was recorded during standing with continuous perturbations under two conditions - holding a handle and standing freely	COP movement strongly correlated to the perturbation force and was more prominent in the anterior direction, hand forces were larger during posterior perturbations	Supportive hand contacts are efficiently used for balance control during continuous postural perturbations and such utilization of a handle has significant immediate and transitionary effects on the whole body posture.
II	How does the location of the handle alter balance control?	Measured whole-body movement (COM displacement), forces exerted on the handle and total body effort (produced torque in the joints) during 8 perturbation magnitudes applied to the participants' COM in 5 postural configurations (step stance, normal stance with or without holding handles at different heights)	Compared to normal stance, step stance and holding handles at different vertical positions sufficiently assisted balance recovery. There was no significant effect of handle in COM displacement. The shoulder height handle required the lowest handle force.	The use of different handles had a similar effect on balance control, the lowest forces were exerted on the shoulder height handle indicating a preferred handle position for balance recovery.
III	How humans coordinate supportive contacts and postural control?	COP, motions of both hands and contact locations of supportive and task performance contacts were measured during reaching towards a visual target.	Participants adapted to the perturbations by establishing target dependent hand contacts. Trunk motion adapted significantly faster than the motion of the arms. Observations of the initial phase of the left arm or trunk motion (100–400 ms) were sufficient to faithfully predict the complete movement of the right arm	Goal-directed arm movements determine the supportive arm motions and motion of heavy body parts adapt faster than the light arms.

Chapter 5

Study I - Utilization of Hand Contacts to Augment Balance Control

In this part, I present the results of the first study in which we focused on mechanisms of a supportive hand contact on postural control during continuous exposure to balance perturbations. People often experience such perturbations while standing on a moving bus or train and usually we hold on to a nearby handle or some other object to facilitate balance control and to prevent a fall. The main research question of this study was how does the control of balance change during the prolonged exposure to perturbations that vary in magnitude and direction and whether (and how) holding a handle affects this control. Changes in control of posture during continuous perturbations were also of special interest to us, since the majority of the perturbation-based balance control experiments involve only discrete perturbations. It is known that discrete perturbations predominantly evoke feedback motor responses (reactive responses) and our goal with this experiment was to explore the feed-forward component of the motor response (anticipatory postural adjustments).

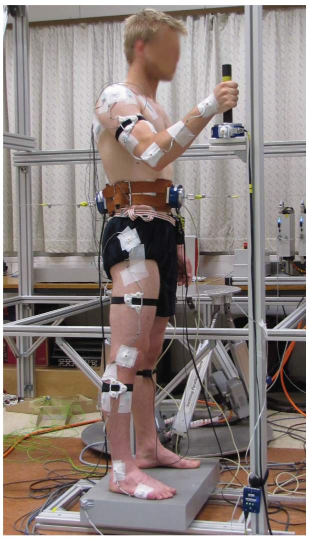

Figure 5.1: Study I. Utilization of hand contacts to augment balance control.

With this in mind we designed an experiment where we used a waist-pull mechanism to apply continuous perturbations to the COM of the participants. During the perturbations, we recorded their motor behavior in both conditions: when they were holding a handle or standing freely. This setup enabled us to effectively test the first three hypotheses of this thesis (H1-H3), where we speculate about the role and mechanisms of an additional support to balance.

The analysis of the experimental data was conducted in two parts. In the first part I describe the changes in posture due to the additional hand contact. The results are summed up in the paper "Holding a Handle for Balance during Continuous Postural Perturbations – Immediate and Transitionary Effects on Whole Body Posture" by Jernej Čamernik, Zrinka Potocanac, Luka Peternel and Jan Babič (Čamernik et al., 2016). The paper was published in a peer-reviewed journal *Frontiers in Human Neuroscience* in 2016. As a first author, I designed and performed the experiment, analyzed the data including programming of the data analysis procedures, and wrote the manuscript. The results of this analysis showed that the COP displacement was significantly affected by the additional support (holding a handle), resulting in smaller excursions in both anterior and posterior directions. Furthermore, we found that the fluctuations of the COP strongly correlated with the perturbation force, especially with the perturbations directed to the anterior direction. This implies that participants used the handle more for counteracting the backward-directed perturbations. When the participants released the handle, their posture changed as well. We found that participants flexed more in all lower body joints and had significantly higher values of muscle activation of the lower leg and trunk muscles when they were standing freely.

In the second part, I present additional results of the same experiment. I analyzed the postural control variable (COP displacement) in frequency domain by using the power spectrum density function. Based on the findings that the COP excursions and handle forces correlated with perturbation direction, the aim of this analysis was to explore how low and high perturbation frequencies are modulated by additional hand support. Results show that when the participants were holding a handle, the share of high frequencies (> 1 Hz) was lower, the share of low frequencies (< 0.25 Hz) was higher and the share of medium frequencies ($0.25 - 1$ Hz) remained the same. This result implies that higher perturbation frequencies are harder to control when standing freely even though the perturbation force is fairly low. An important aspect of effective balance control is also how much energy is needed to successfully accomplish this task. By using a simplified whole-body model and the inverse dynamics procedure I aimed to calculate the work done in each standing condition. Results show that the total body work was significantly higher when the participants were standing freely, in spite of inclusion of the arm joints in the calculation when they were holding a handle.

frontiers in Human Neuroscience

ORIGINAL RESEARCH
published: 26 September 2016
doi: 10.3389/fnhum.2016.00486

Holding a Handle for Balance during Continuous Postural Perturbations—Immediate and Transitionary Effects on Whole Body Posture

*Jernej Čamernik[1,2], Zrinka Potocanac[1], Luka Peternel[1,3] and Jan Babič[1]**

[1] Department for Automation, Biocybernetics and Robotics, Jožef Stefan Institute, Ljubljana, Slovenia, [2] Jožef Stefan International Postgraduate School, Ljubljana, Slovenia, [3] HRI2 Laboratory, Department of Advanced Robotics, Istituto Italiano di Tecnologia, Genoa, Italy

When balance is exposed to perturbations, hand contacts are often used to assist postural control. We investigated the immediate and the transitionary effects of supportive hand contacts during continuous anteroposterior perturbations of stance by automated waist-pulls. Ten young adults were perturbed for 5 min and required to maintain balance by holding to a stationary, shoulder-high handle and following its removal. Center of pressure (COP) displacement, hip, knee and ankle angles, leg and trunk muscle activity and handle contact forces were acquired. The analysis of results show that COP excursions are significantly smaller when the subjects utilize supportive hand contact and that the displacement of COP is strongly correlated to the perturbation force and significantly larger in the anterior than posterior direction. Regression analysis of hand forces revealed that subjects utilized the hand support significantly more during the posterior than anterior perturbations. Moreover, kinematical analysis showed that utilization of supportive hand contacts alter posture of the whole body and that postural readjustments after the release of the handle, occur at different time scales in the hip, knee and ankle joints. Overall, our findings show that supportive hand contacts are efficiently used for balance control during continuous postural perturbations and that utilization of a handle has significant immediate and transitionary effects on whole body posture.

Keywords: falls, handle, grasping, postural balance, balance recovery

Edited by:
Eric Yiou,
University of Paris-Sud, France

Reviewed by:
Jean-Louis Honeine,
University of Pavia, Italy
Teddy Caderby,
University of La Réunion, France

****Correspondence:***
Jan Babič
jan.babic@ijs.si

Received: *08 June 2016*
Accepted: *13 September 2016*
Published: *26 September 2016*

Citation:
Čamernik J, Potocanac Z, Peternel L and Babič J (2016) Holding a Handle for Balance during Continuous Postural Perturbations—Immediate and Transitionary Effects on Whole Body Posture.
Front. Hum. Neurosci. 10:486.
doi: 10.3389/fnhum.2016.00486

INTRODUCTION

With aging society, falls are becoming an increasingly large problem. A large proportion of falls occur due to the improper weight shifts (Robinovitch et al., 2013) and impaired postural control is a landmark of aging (Maki and McIlroy, 2006; Mansfield and Maki, 2009). When postural control is impaired, handrails, canes and handles are often used to assist maintaining balance by providing additional supportive contacts with the environment. This indicates that holding onto a physical aid is beneficial for postural control.

With respect to the use of hand contacts for postural control, one of the widely investigated phenomena is "light touch" (Jeka, 1997; Krishnamoorthy et al., 2002). These light, fingertip contacts with stationary objects can extend the base of support (Bateni and Maki, 2005) and provide an additional sensory input, which helps individuals to better position them in space (Jeka, 1997). Such sensory information improves postural control in quiet standing by reducing the amplitude of center of pressure (COP) movement (Jeka, 1997; Johannsen et al., 2007; Kouzaki and Masani, 2008; Wing et al., 2011).

On the other hand, in case of perturbed balance reaching arm movements with the aim to grasp for a nearby object is a widely utilized change-in-support strategy (Maki and McIlroy, 2006). Such hand contacts provide mechanical support in addition to the sensory augmentation of the light touch and thus offer a better stabilizing potential in the presence of perturbations (Maki and McIlroy, 1997). Specifically, holding onto a handle increases the base of support of a standing individual and enables a person to generate necessary hand forces to better counteract the perturbations (Babič et al., 2014; Sarraf et al., 2014). A recent study by Babič et al. (2014) showed that the location of the supporting hand contact is important to maximize its stabilizing potential and that the peak forces exerted at the handle during the support surface perturbations are related to the location of the handle.

Aforementioned studies were based on the discrete perturbations of balance which predominantly evoke feedback postural responses. A major component of such responses is comprised of motor actions that are related to various sensorimotor reflexes and to a lesser extent to the feed-forward components of the postural control (Mergner, 2010). Moreover, the discrete perturbations evoke reach-to-grasp arm movements even when the perturbations are so light that they do not physically disturb postural balance (McIlroy and Maki, 1995; Corbeil et al., 2013). In contrast to the discrete perturbations of balance, perturbations that continuously disturb postural balance evoke both feedback and feed-forward components of motor action and in this sense offer a complementary insight into the postural control (Dietz et al., 1993; Schmid et al., 2011).

The remaining question is what is the role of hand contact during continuous perturbations? Therefore, the aim of this article is to study situations where balance of an individual is challenged by continuous postural perturbations and to investigate the role of supportive hand contact in counteracting postural perturbations. Specifically, our goal was to investigate the immediate and the transitionary effects of a supportive hand contact on postural control of an individual whose balance is challenged by continuous anteroposterior perturbations of stance.

Our hypothesis is that a supportive hand contact has a significant influence on postural balance by reducing the COP excursion during the perturbation and that the utilization of the hand contact is more prominent for postural perturbations in the backward direction which are more threatening than the perturbations in the forward direction. Moreover, we hypothesize that utilization of the additional hand support not only alters posture of the human body while the hand is in contact with the environment but also after the release of the handle. To effectively address these hypotheses, we developed an experimental framework where we continuously perturbed postural balance and investigated the relationships between the perturbation force and the COP displacement, kinematical parameters of the human body, and the forces exerted by the supportive hand.

MATERIALS AND METHODS

Participants

Thirteen healthy right-handed young adults participated in this study after giving their written consent. Data of three subjects were excluded from the analyses due to technical problems during acquisition, therefore we used the data of ten subjects (average age = 22.3 years, SD = 2.2 years, average height 179.2 cm, SD = 5.9 cm and average weight = 76.9 kg, SD = 8.2 kg). The experimental procedures conformed to the latest revision of the Declaration of Helsinki and were approved by the Slovenian National Medical Ethics Committee (No. 112/06/13).

Measurement Protocol

Subjects were asked to step on a force plate, stand straight with the feet placed at hip width and look straight ahead. They were required to keep upright posture and maintain balance without making any unnecessary corrective steps while their balance was continuously perturbed in anteroposterior direction by a motorized waist-pull system (Peternel and Babič, 2013) as depicted in **Figure 1**. During the experiment, the subjects were not allowed to change their base of support. We marked their individual standing position on the force plate prior to the start of the experiment which was used as a reference for foot position during the experiment. This ensured that the different stance width would not affect subjects' balance since it has been shown by Bingham and Ting (2013) that active torque at ankle and hip joints scale with stance width. To emulate mild, daily life perturbations such as those during riding on buses, subways and trains (Graaf and van Weperen, 1997) the motorized waist-pull system perturbed the subjects using a band-pass filtered white noise signal (0.25–1.00 Hz) with the maximal perturbation force of 11% of the subjects' body weight (**Figure 2**).

The experiment consisted of two consecutive 5 min trials of different standing conditions: balancing while holding to a handle [*with-handle (WH)*] and without holding to a handle [*no-handle (NH)*]. First, the subjects were exposed to 5 min of perturbations in the *WH* trial. In the *WH* trial, subjects held onto a stationary handle (diameter = 3.2 cm, length = 12 cm) positioned at shoulder height with their right hand. After 5 min the perturbation stopped, subjects released the handle and folded their arms across their chest. Then, on average less than 60 s later, the second trial of 5 min of perturbation started (*NH* trial). In the *NH* trial, subjects were standing with their arms folded across their chest. In both trials, subjects were instructed to look straight ahead at all times at a fixed point positioned at the subject's eyelevel and 3 m in front of the experimental setup.

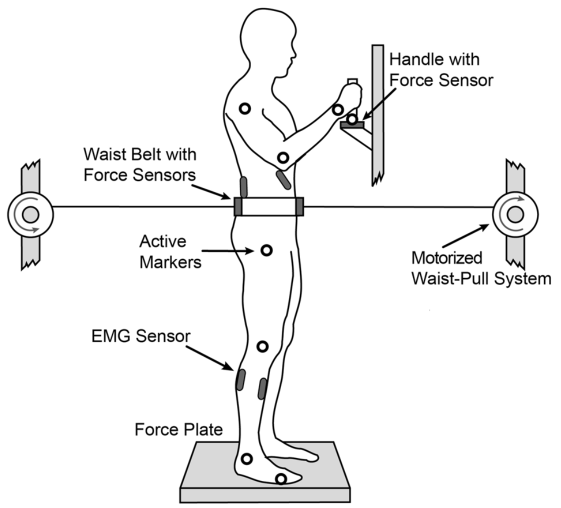

FIGURE 1 | Experimental setup. The subject is standing on a force plate, wearing a waist belt connected to the motorized waist-pull system which generated translational force perturbations in the anterior-posterior direction using a band-filtered white noise signal constructed to emulate mild, daily life perturbations.

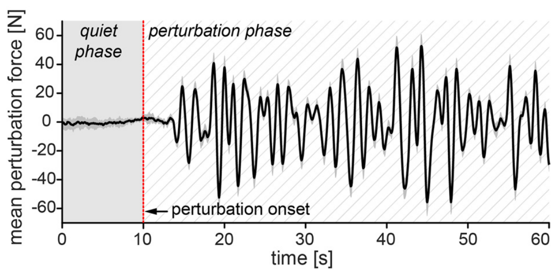

FIGURE 2 | Perturbation signal sample. Solid black curve represents first 60 s of mean perturbation signal from all 10 subjects with ±1 standard error of the mean (gray shade around the curve). In the first 10 s of each trial the perturbation value was at 0 N—quiet phase. After 10 s (perturbation onset), the perturbation gradually increased with alternated direction and amplitude—perturbation phase. Positive values of mean perturbation force represent forces in anterior direction and negative values represent forces in posterior direction. The perturbation signal was identical in both trials.

To induce response adaptations (Van Ooteghem et al., 2008; Schmid et al., 2011), the subjects were allowed to familiarize with the experimental procedures prior to the main experimental trials.

Kinetic data were collected using a force plate (9281CA, Kistler Instrumente AG, Winterthur, Switzerland) under the subjects' feet and a 3-axis force sensor (45E15A, JR3, Woodland, CA, USA) on the handle, both at 1000 samples/s.

Kinematic data were collected at a sampling rate of 100 samples/s using a contactless motion capture system (3D Investigator, Northern Digital Inc., Waterloo, ON, Canada) consisting of a 3 × 3 camera array. Nine active markers were placed at the apparent axis of rotation of the fifth metatarsophalangeal, ankle, knee, hip, shoulder, elbow and wrist joints on the subject's right side as well as at the base of the platform and the handle.

Anteroposterior displacement of the subject's COP was derived from the force plate data. In the first 10 s of each trial, the subjects stood quietly (holding a handle with right hand in the first trial, or both arms folded across their chest in the second trial) and no perturbation was applied at the waist (see **Figure 2** for reference regarding perturbation signal). The mean COP position from this time period (we refer to this as *quiet phase*) served as a baseline for calculation of COP excursions in the anterior and posterior direction in the following perturbations (*perturbation phase*).

Handle forces were calculated by considering the torques of the lever (distance from force sensor to the middle of the subject's hand on the handle). Kinematic data were low pass filtered (zero lag, 2nd order Butterworth filter with a cut-off frequency 20 Hz). Ankle angle was calculated as the angle between the foot (line connecting the fifth metatarsal and ankle) and the shank (line connecting the ankle and the knee), knee angle as the angle between the shank and thigh (line connecting the knee and the hip), and hip as the angle between the thigh and torso (line connecting the hip and the shoulder). To evaluate the adaptation effects of releasing the handle in the *NH* trial, an exponential fit of group average joint angles was calculated (Franklin et al., 2003). Adaptation was considered as final once the given joint angle reached the plateau defined by three time constants of the fitted exponential decay function, i.e., once the fitted exponential decay function fell to 5% of its starting value (Honeine et al., 2015; Assländer and Peterka, 2016).

Electromyographical (EMG) electrodes were placed on the right leg (TA, Tibialis Anterior; GA, Gastrocnemius Lateralis; and trunk (MF, Multifidus; OE, Obliques Externus) muscles and their activity was measured using Biometrics DataLOG MW8X at a sampling rate of 1000 samples/s. Preparation of the skin and positioning of the electrodes was performed according to the SENIAM protocol (Hermens et al., 2000). Before the start of the experiment, subjects performed three maximal voluntary contractions (MVC) against resistance of each of the measured muscles. MVC's were used in EMG post-processing for normalization, to establish a common ground when comparing data between subjects. All EMG signals were band-pass filtered (zero lag, 2nd order Butterworth filter with cut-off frequencies of 20 and 450 Hz), full-wave rectified and low pass filtered (zero lag, 2nd order Butterworth algorithm, 10 Hz cut-off frequency). Finally, EMG signals were normalized with respect to the MVCs and integrated over time (iEMG) to express the magnitude of muscle activity.

We divided COP excursions and contact forces exerted on the handle in two data sets based on their direction—anterior and posterior. For each data set, average values from all 10 subjects were calculated and used in the statistical analysis.

Average hip, knee and ankle angles over the 5 min for each subject were calculated and used for statistical analysis.

Differences between COP displacement in the anterior and posterior directions and subject average joint angles were analyzed using paired samples t-tests. Differences between the *WH* and *NH* trials in subject average COP displacements were analyzed for the anterior and posterior directions separately, using a paired samples t-test. The relationship between group average COP excursion and the magnitude of the perturbation and between group average perturbation magnitude and the exerted handle contact force was analyzed using separate linear correlations for anterior and posterior directions. All statistical analyses were performed using SPSS 21 Inc., Chicago, IL, USA at $\alpha = 0.05$. Effect size (d) was calculated using standard Cohen's equation (Cohen, 1988).

RESULTS

Supportive Hand Contact has Significant Influence on Postural Balance by Reducing COP Excursion during Perturbation

The diagram in **Figure 3** shows the comparison of mean COP excursion between the conditions when the subjects counteracted postural perturbations without using the additional hand contact (NH) and when they did use the handle (WH). Paired samples t-test showed significant effect in reducing the mean COP excursion when the subjects were holding to the handle compared to when they did not hold to the handle. The differences of COP excursion were significantly larger both in the anterior direction (difference of 20.3 mm, $t_{(9)} = 7.78$, $p = 0.001$, $d = -4.15$) and posterior direction (difference of 23.9 mm, $t_{(9)} = -11.09$, $p = 0.001$, $d = -3.8$).

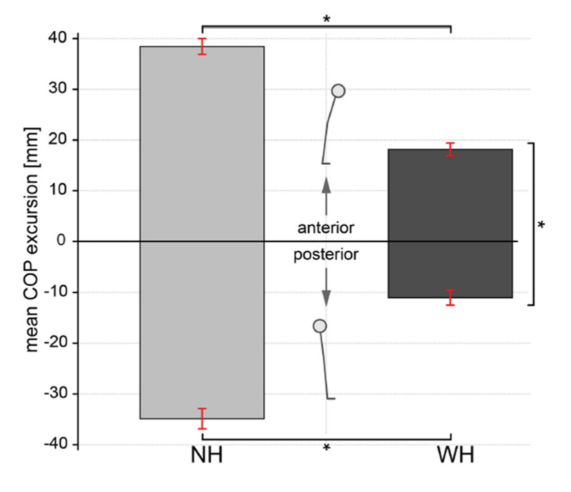

FIGURE 3 | Comparison of center of pressure (COP) excursion between trials when subjects either used additional hand contact or not. Bars represent mean COP excursion during no-handle (NH) and with-handle (WH) trials for the anterior (positive) and posterior (negative) directions. Error bars indicate ±1 standard error of the mean. Statistically significant differences are indicated (*$p \leq 0.02$).

Utilization of Hand Contact is more Prominent for Postural Perturbations in Backward Direction than for Perturbations in Forward Direction

In both, *NH* and *WH* trials, the COP excursion was larger in the anterior direction (mean ± SE: *NH* 38.5 ± 1.6 mm, *WH* 18.2 ± 1.2 mm) compared to the posterior (mean ± SE: *NH* −34.9 ± 2 mm, *WH* −11.0 ± 1.5 mm), but this difference was significant only for the *WH* trial ($t_{(9)} = 2.81$, $p = 0.02$, $d = 1.52$).

We further assessed the effects of utilizing the additional hand contact, and the direction and intensity of perturbation on the maximal COP displacement. The diagrams in **Figure 4** show correlations between the perturbation force and the group average COP excursion during *NH* and *WH* trials. Additionally, the correlation between the perturbation force and the handle force is shown for the *WH* trials. The group average COP excursion was strongly correlated with perturbation force in both posterior ($r_p = 0.77$ and $r_p = 0.67$) and anterior ($r_a = 0.82$ and $r_a = 0.89$) directions in the *NH* and *WH* trials, respectively (all $p < 0.001$). Similarly, the forces that

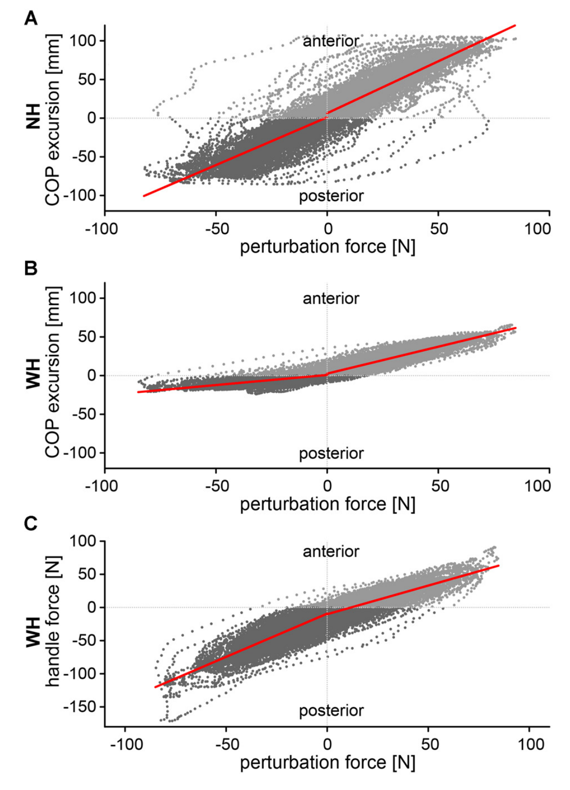

FIGURE 4 | Correlations between perturbation force and (A) COP excursion in NH trial, (B) COP excursion in WH trial and (C) handle force in the WH trial. Correlations were calculated separately for the anterior (positive) and posterior (negative) direction.

subjects applied on the handle was also strongly correlated with the perturbation force (**Figure 4C**) in both anterior ($r_p = 0.85$, $p < 0.001$) and posterior directions ($r_a = 0.81$, $p < 0.001$). Moreover, the slope of the regression line is significantly larger for perturbations in the posterior direction ($k_p = 1.3$), compared to the perturbations in the anterior direction ($k_a = 0.86$).

Supportive Hand Contact Affects Whole Body Posture

To investigate how the additional hand contact affects the body posture during the perturbations, we compared the mean values of ankle, knee and hip joint angles between the conditions when the subjects counteracted perturbations without the additional hand contact (NH) and when they did use the handle (WH). The comparison is shown in the diagram in **Figure 5**.

Multiple paired samples t-tests showed significant effect of the hand contact on all three observed joint angles. Specifically, mean joint angles were significantly lower during the *NH* trials compared to the *WH* trials. Differences were the largest in the knee (mean ± SE: 165.1 ± 1.9° for NH, 173.2 ± 1.5° for WH, $t_{(9)} = -6.70$, $p < 0.001$, $d = 1.4$), followed by the hip (mean ± SE: 171.4 ± 2.8° for NH, 181.5 ± 1.6° WH, $t_{(9)} = -6.68$, $p < 0.001$, $d = 1.1$) and the ankle (mean ± SE: 108 ± 1.2° for NH, 112 ± 1.1° WH, $t_{(9)} = -5.67$, $p < 0.001$, $d = 1.1$).

Utilization of Hand Contact has Non-Uniform Transitionary Effect on Whole Body Posture after Release of Handle

To investigate the effect of supportive hand contact on the body posture, an exponential curve was fitted to the group average ankle, knee and hip joint angles calculated during the *NH* trial that immediately followed the *WH* trial (**Figure 6**).

Exponential fits revealed that postural readjustments after the release of the handle did not occur simultaneously throughout the body. Instead, the readjustments occurred at different time scales in the hip, knee and ankle joints. Specifically, joint angles stabilized first in the ankle (mean ± SE: 133 ± 103.5 s after perturbation onset), followed by the hip (mean ± SE: 188 ± 90.8 s after the perturbation onset) and finally in the knee (mean ± SE: 195 ± 92.5 s after the perturbation onset). However, a paired-samples t-test did not show statistically significant difference between any of the compared pairs due to the high variability of data.

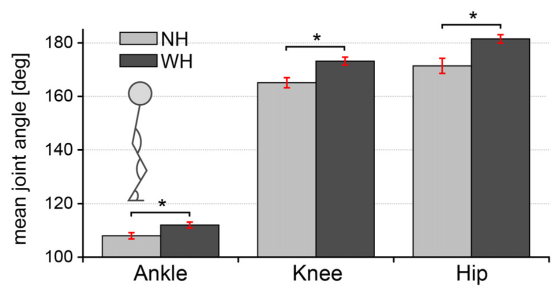

FIGURE 5 | Comparison of body posture between trials when subjects either used additional hand contact or not. Bars represent mean ankle, knee and hip joint angles during NH and WH trials. Error bars indicate ±1 standard error of the mean. Statistically significant differences are indicated (*$p \leq 0.02$).

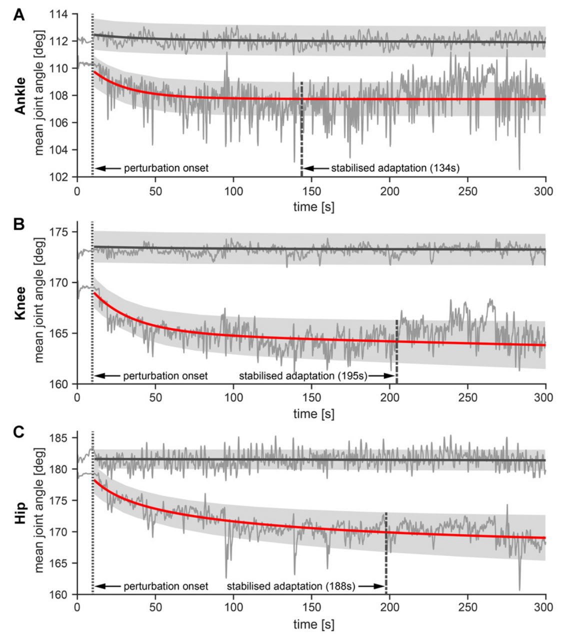

FIGURE 6 | Ankle (A) knee (B) and hip (C) angles over the time course of the perturbation. Thin solid lines represent mean joint angles from all 10 subjects during NH and WH trials. Thick solid lines represent exponential curve fit denoting adaptation of joint angles in the NH (red color) and WH trials (gray color), while shaded areas represent ±1 standard error of the mean of the exponential decay curve. Mean R^2 value for the exponential decay curves for ankle joint in the NH trial was mean ± SE: 0.27 ± 0.06, for the knee joint mean ± SE: 0.31 ± 0.08 and for the hip joint mean ± SE: 0.61 ± 0.08. The dotted vertical lines represent perturbation onset while the dashed vertical lines indicate the mean time of stabilized changes in the joint angles after perturbation onset.

Analysis of Muscle Activity

Muscle activity was significantly lower during the WH trial than during the NH condition both for the leg muscles (GA $t_{(9)} = 3.57$, $p = 0.04$, $d = -0.89$; TA $t_{(9)} = 6.41$, $p = 0.002$, $d = -23\ 1.85$) and one of the trunk muscles (MF $t_{(9)} = 6.5$, $p = 0.001$, $d = -1.01$), as can be seen in **Figure 7**. Leg muscle activity was 18.4 ± 4.9% lower in the GA (mean ± SE: NH: 28.9 ± 6.5% MVC, WH: 10.6 ± 2.3% MVC) and for 23.7 ± 3.5% in the TA (mean ± SE: NH: 27.2 ± 4% MVC, WH: 3.47 ± 1.88% MVC), while the trunk muscle activity was 14.3 ± 2.1% lower in the MF (mean ± SE: NH: 36.2 ± 4.5% MVC, WH: 21.8 ± 3.7% MVC), but no significant change was observed in the OE (mean ± SE: NH: 10.4 ± 7.4% MVC, WH: 8.97 ± 4.63% MVC).

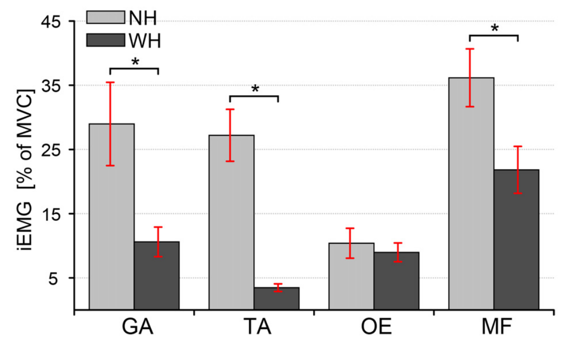

FIGURE 7 | Comparison of integrated Electromyographical (EMG) between trials when subjects either used additional hand contact or not. Bars represent iEMG of four muscles (GA, Gastrocnemius Lateralis; TA, Tibialis Anterior; OE, Obliques Externus; MF, Multifidus) during NH and WH trials. Error bars indicate ±1 standard error of the mean. Statistically significant differences are indicated (*$p \leq 0.02$).

DISCUSSION

We investigated how subjects used an additional hand support (handle) to maintain upright posture when exposed to mild and continuous perturbations elicited by waist-pull apparatus in the anteroposterior directions. The use of handle reduced the destabilizing effect of the applied perturbation by reducing the excursions of COP. These results are in line with previous studies that investigated light-touch contacts. Those studies found a reduction of COP excursions during bipedal stance with (Johannsen et al., 2007; Hausbeck et al., 2009) and without externally applied perturbations (Jeka, 1997; Krishnamoorthy et al., 2002; Kouzaki and Masani, 2008). However, in our case the handle compensated for a significant load and served more than just a light-touch contact. The mean force in the handle was over 24 N (mean ± SE: 24.5 ± 7.9 N) where in other light-touch studies, contact forces usually did not exceed 3 N (Krishnamoorthy et al., 2002; Johannsen et al., 2007; Kouzaki and Masani, 2008). Comparison between measured mean handle force and mean perturbation force, which was ~20 N (mean ± SE: 19.5 ± 1.7 N), indicates that a significant portion of perturbation on postural stability was counterbalanced by the hand.

COP excursions were strongly correlated with the perturbation force in both directions, indicating the perturbations were effective, albeit mild. When holding a handle, the excursions of the COP were larger in the anterior than in posterior direction and less correlated with the perturbation force. Due to the specifics of our design, i.e., the use of a continuous perturbation, it was impossible to investigate pure feedback postural responses to the specific direction of the perturbation. However, asymmetry in COP excursion might to be due to a differential use of the handle, indicated by a steeper slope of the regression line between the perturbation force and the forces exerted on the handle for the posterior direction. This indicates that subjects have utilized the handle more when they counteracted the posterior COP excursions. This may be related to a more threatening situation due to the inability of the subjects to see (look) behind them, as the directions to the subjects were to look straight ahead at all times, and due to smaller stability margin in the posterior direction (Pai and Patton, 1997; Hof et al., 2005). This finding is in line with our previous study using a similar handle location and discrete perturbations caused by support platform translations, in which COP excursions were larger in the anterior direction (Babič et al., 2014).

Using a handle for balance support was beneficial, since it resulted in less displacement of COP and a smaller deviation from the neutral posture, as evident by the average joint angles. Measured joint angles in *WH* condition stayed closer to the neutral anatomic position than joint angles in the *NH* condition. Additionally, leg and trunk muscle activity was also significantly lower during the *WH* trial compared to the *NH* trial. This is consistent with the decreased leg muscle activity reported previously, when subjects had to hold (Cordo and Nashner, 1982) or touch (Sozzi et al., 2012) a surrounding object. Unlike other muscles, we found no decrease of muscle activity in the OE, which controls the rotation of the torso (Ng et al., 2001). The unilateral hand support might have caused a rotation of the trunk, which the subjects had to counteract by the OE activity. Prior to experiments we instructed subjects to use the handle in any way they prefer. Overall reduced muscle activity in legs and trunk and increased activity in arm muscles in case of using handle indicate that a portion of significant perturbation load was shifted from legs/trunk to arms. The same can be confirmed by high measured handle forces and lesser COP excursions in case of *WH* trial. The use of hand contact to compensate a significant portion of perturbation, even though it could be counteracted solely by legs/trunk, might be preferred since legs in a stance already have to compensate the load of the body mass (Bateni and Maki, 2005; Mergner, 2010).

Finally, when subjects had to release the handle for balance control they prepared for the more difficult *NH* condition even before the perturbation onset, as evident from differences in the starting joint angles. When the perturbation began this preparation was even more pronounced and joint angles changed further. These postural readjustments appear to occur at different time scales in the hip, knee and ankle joints, however that was not statistically confirmed. The subjects bended their ankle, knee and hip joints which resulted in a more flexed leg and lower hip position. Hip position can serve as an indication of the COM position and lowering of the COM could facilitate balance control (Rosker et al., 2011). Hence, these changes indicate feed-forward preparation to ease control of balance when expecting more challenging conditions, i.e., in the absence of handle. Along the same line, anticipation of the upcoming perturbations was also reported to cause changes in kinematics during quiet stance (Santos et al., 2010), walking (Pijnappels et al., 2001), recovery stepping (Pater et al., 2015) and tripping (Potocanac et al., 2014).

AUTHOR CONTRIBUTIONS

JC, LP and JB designed the study. JC and LP performed the experiments. JC, LP, ZP and JB analyzed the data and wrote the manuscript.

FUNDING

The work presented in this article was supported by the European Community Framework Programme 7 through the CoDyCo project, contract no. 600716.

REFERENCES

Assländer, L., and Peterka, R. J. (2016). Sensory reweighting dynamics following removal and addition of visual and proprioceptive cues. *J. Neurophysiol.* 116, 272–285. doi: 10.1152/jn.01145.2015

Babič, J., Petrič, T., Peternel, L., and Šarabon, N. (2014). Effects of supportive hand contact on reactive postural control during support perturbations. *Gait Posture* 40, 441–446. doi: 10.1016/j.gaitpost.2014.05.012

Bateni, H., and Maki, B. E. (2005). Assistive devices for balance and mobility: benefits, demands and adverse consequences. *Arch. Phys. Med. Rehabil.* 86, 134–145. doi: 10.1016/j.apmr.2004.04.023

Bingham, J. T., and Ting, L. H. (2013). Stability radius as a method for comparing the dynamics of neuromechanical systems. *IEEE Trans. Neural Syst. Rehabil. Eng.* 21, 840–848. doi: 10.1109/TNSRE.2013.2264920

Cohen, J. (1988). *Statistical Power Analysis for the Behavioral Sciences.* 2nd Edn. Mahwah, NJ: Lawrence Erlbaum Associates.

Corbeil, P., Bloem, B. R., van Meel, M., and Maki, B. E. (2013). Arm reactions evoked by the initial exposure to a small balance perturbation: a pilot study. *Gait Posture* 37, 300–303. doi: 10.1016/j.gaitpost.2012.07.017

Cordo, P. J., and Nashner, L. M. (1982). Properties of postural adjustments associated with rapid arm movements. *J. Neurophysiol.* 47, 287–302.

Dietz, V., Trippel, M., Ibrahim, I. K., and Berger, W. (1993). Human stance on a sinusoidally translating platform: balance control by feedforward and feedback mechanisms. *Exp. Brain Res.* 93, 352–362. doi: 10.1007/bf00228405

Franklin, D. W., Osu, R., Burdet, E., Kawato, M., and Milner, T. E. (2003). Adaptation to stable and unstable dynamics achieved by combined impedance control and inverse dynamics model. *J. Neurophysiol.* 90, 3270–3282. doi: 10.1152/jn.01112.2002

Graaf, B. D., and van Weperen, W. (1997). The retention of balance: an exploratory study into the limits of acceleration the human body can withstand without losing equilibrium. *Hum. Factors* 39, 111–118. doi: 10.1518/001872097778940614

Hausbeck, C. J., Strong, M. J., Tamkei, L. S., Leonard, W. A., and Ustinova, K. I. (2009). The effect of additional hand contact on postural stability perturbed by a moving environment. *Gait Posture* 29, 509–513. doi: 10.1016/j.gaitpost.2008.11.014

Hermens, H. J., Freriks, B., Disselhorst-Klug, C., and Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *J. Electromyogr. Kinesiol.* 10, 361–374. doi: 10.1016/s1050-6411(00)00027-4

Hof, A. L., Gazendam, M. G. J., and Sinke, W. E. (2005). The condition for dynamic stability. *J. Biomech.* 38, 1–8. doi: 10.1016/j.jbiomech.2004.03.025

Honeine, J.-L., Crisafulli, O., Sozzi, S., and Schieppati, M. (2015). Processing time of addition or withdrawal of single or combined balance-stabilizing haptic and visual information. *J. Neurophysiol.* 114, 3097–3110. doi: 10.1152/jn.00618.2015

Jeka, J. J. (1997). Light touch contact as a balance aid. *Phys. Ther.* 77, 476–487.

Johannsen, L., Wing, A. M., and Hatzitaki, V. (2007). Effects of maintaining touch contact on predictive and reactive balance. *J. Neurophysiol.* 97, 2686–2695. doi: 10.1152/jn.00038.2007

Kouzaki, M., and Masani, K. (2008). Reduced postural sway during quiet standing by light touch is due to finger tactile feedback but not mechanical support. *Exp. Brain Res.* 188, 153–158. doi: 10.1007/s00221-008-1426-5

Krishnamoorthy, V., Slijper, H., and Latash, M. L. (2002). Effects of different types of light touch on postural sway. *Exp. Brain Res.* 147, 71–79. doi: 10.1007/s00221-002-1206-6

Maki, B. E., and McIlroy, W. E. (1997). The role of limb movements in maintaining upright stance: the "change-in-support" strategy. *Phys. Ther.* 77, 488–507.

Maki, B. E., and McIlroy, W. E. (2006). Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. *Age Ageing* 35, ii12–ii18. doi: 10.1093/ageing/afl078

Mansfield, A., and Maki, B. E. (2009). Are age-related impairments in change-in-support balance reactions dependent on the method of balance perturbation? *J. Biomech.* 42, 1023–1031. doi: 10.1016/j.jbiomech.2009.02.007

McIlroy, W. E., and Maki, B. E. (1995). Early activation of arm muscles follows external perturbation of upright stance. *Neurosci. Lett.* 184, 177–180. doi: 10.1016/0304-3940(94)11200-3

Mergner, T. (2010). A neurological view on reactive human stance control. *Annu. Rev. Control* 34, 177–198. doi: 10.1016/j.arcontrol.2010.08.001

Ng, J. K.-F., Parnianpour, M., Richardson, C. A., and Kippers, V. (2001). Functional roles of abdominal and back muscles during isometric axial rotation of the trunk. *J. Orthop. Res.* 19, 463–471. doi: 10.1016/s0736-0266(00)90027-5

Pai, Y. C., and Patton, J. (1997). Center of mass velocity-position predictions for balance control. *J. Biomech.* 30, 347–354. doi: 10.1016/s0021-9290(96)00165-0

Pater, M. L., Rosenblatt, N. J., and Grabiner, M. D. (2015). Expectation of an upcoming large postural perturbation influences the recovery stepping response and outcome. *Gait Posture* 41, 335–337. doi: 10.1016/j.gaitpost.2014.10.026

Peternel, L., and Babič, J. (2013). Learning of compliant human-robot interaction using full-body haptic interface. *Adv. Robot.* 27, 1003–1012. doi: 10.1080/01691864.2013.808305

Pijnappels, M., Bobbert, M. F., and van Dieën, J. H. (2001). Changes in walking pattern caused by the possibility of a tripping reaction. *Gait Posture* 14, 11–18. doi: 10.1016/s0966-6362(01)00110-2

Potocanac, Z., de Bruin, J., van der Veen, S., Verschueren, S., van Dieën, J., Duysens, J., et al. (2014). Fast online corrections of tripping responses. *Exp. Brain Res.* 232, 3579–3590. doi: 10.1007/s00221-014-4038-2

Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., et al. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. *Lancet* 381, 47–54. doi: 10.1016/S0140-6736(12)61263-X

Rosker, J., Markovic, G., and Sarabon, N. (2011). Effects of vertical center of mass redistribution on body sway parameters during quiet standing. *Gait Posture* 33, 452–456. doi: 10.1016/j.gaitpost.2010.12.023

Santos, M. J., Kanekar, N., and Aruin, A. S. (2010). The role of anticipatory postural adjustments in compensatory control of posture: 2. Biomechanical analysis. *J. Electromyogr. Kinesiol.* 20, 398–405. doi: 10.1016/j.jelekin.2010.01.002

Sarraf, T. A., Marigold, D. S., and Robinovitch, S. N. (2014). Maintaining standing balance by handrail grasping. *Gait Posture* 39, 258–264. doi: 10.1016/j.gaitpost.2013.07.117

Schmid, M., Bottaro, A., Sozzi, S., and Schieppati, M. (2011). Adaptation to continuous perturbation of balance: progressive reduction of postural muscle activity with invariant or increasing oscillations of the center of mass depending on perturbation frequency and vision conditions. *Hum. Mov. Sci.* 30, 262–278. doi: 10.1016/j.humov.2011.02.002

Sozzi, S., Do, M. C., Monti, A., and Schieppati, M. (2012). Sensorimotor integration during stance: processing time of active or passive addition or withdrawal of visual or haptic information. *Neuroscience* 212, 59–76. doi: 10.1016/j.neuroscience.2012.03.044

Van Ooteghem, K., Frank, J. S., Allard, F., Buchanan, J. J., Oates, A. R., and Horak, F. B. (2008). Compensatory postural adaptations during continuous, variable amplitude perturbations reveal generalized rather than sequence-specific learning. *Exp. Brain Res.* 187, 603–611. doi: 10.1007/s00221-008-1329-5

Wing, A. M., Johannsen, L., and Endo, S. (2011). Light touch for balance: influence of a time-varying external driving signal. *Philos. Trans. R. Soc. Lond. B Biol. Sci.* 366, 3133–3141. doi: 10.1098/rstb.2011.0169

Conflict of Interest Statement: The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

5.1 Additional Analyses and Conclusions

5.1.1 Frequency-domain representation of COP

In the above paper, the effects of additional hand support on the participants' balance were evaluated by analyzing the COP data from both trials in the time domain - quantitative analysis. I extended that analysis, by transforming the signals from time domain, to frequency domain, to search for the qualitative differences between both trials and estimate the changes in postural strategy.

Researchers have previously shown that balance tests (equilibrium score and sway path) after an exhausting exercise like walking (Lepers et al., 1997) or running (Nardone et al., 1998) reveal significant detrimental effects on the ability to maintain standing balance. The detrimental effects were shown in deteriorated ability of receiving sensory information and their integration, resulting in a decreased muscular efficiency. By inspecting the frequency profile of the COP, researchers (Dietz et al., 1980; Golomer et al., 1994) have shown that the so-called high frequencies of the COP signal (1 - 20 Hz) mostly account for a somatosensory feedback mechanism (i.e. information coming from the perception of touch or movement). The light touch of a finger with a stationary object while standing freely initiates changes in the COP position within the first 300 ms, however, a more prominent touch or holding a handle enables a much faster response, close to 30 ms (Jeka & Lackner, 1994).

These findings emphasize the importance of additional contact of the body with the environment to augment balance control, however, further qualitative investigations of balance during prolonged exposure to perturbations still remain to be done. In the following sections I first briefly explain the instrumentation and data acquisition which was used in the study, since the methodology of the experiment is thoroughly explained in the paper above. Second, I focus on the theory and methodology I used to transform the COP signals from time to frequency domain. Last, I explain the results and draw some conclusions.

Methods

During both trials of perturbed standing, the participants were standing on a Kistler force plate (9821CA, Kistler Instrumente AG, Winterthur, Switzerland) to record the ground reaction forces. The sampling rate was set at 1000 samples/s, and the anteroposterior displacement of the participants' COP was calculated from the collected forces and moments, Eq. (5.1).

$$\mathrm{COP} = \frac{M_x - (F_y \cdot dz)}{F_z}, \tag{5.1}$$

where M_x refers to the moments of forces around the frontal axis, F_y and F_z refer to horizontal and vertical components of the ground reaction force, respectively, and dz is the distance from the surface to the platform origin.

From the COP position signals, a power spectrum was obtained using the Fast Fourier Transformation with Hanning window using a procedure as described in Vieira et al. (2009). Then, I calculated the mean power frequency (MNF) and the median power frequency (MDF) (Phinyomark et al., 2012) of the COP signals to check if there was any change in these parameters which could indicate a direct effect of the handle on balance control.

The MNF is an average frequency value that is computed as a sum of product of the COP power spectrum and frequency, divided by a total sum of spectrum intensity

$$\mathrm{MNF} = \sum_{j=1}^{M} f_j P_j \Bigg/ \sum_{j=1}^{M} P_j \ , \tag{5.2}$$

where f_j is a frequency value at a frequency bin j.

The MDF is a frequency value at which the COP power spectrum is divided into two regions with an equal integrated power and can be expressed as

$$\sum_{j=1}^{MDF} P_j = \sum_{j=MDF}^{M} P_j = \frac{1}{2}\sum_{j=1}^{M} P_j \quad . \tag{5.3}$$

Next, I divided the power spectrum into three frequency intervals - low-frequency (LF) band (0 - 0.25 Hz), medium-frequency (MF) band (0.25 - 1 Hz) and high-frequency (HF) band (1 - 3 Hz). According to literature, the LF band is sensitive to vestibular stress and disturbances (Oppenheim et al., 1999) and is associated with visual regulation; the MF band is associated with the somatosensory activity and postural reflexes (Nagy et al., 2004; Wing et al., 2011); and the HF band reflects the dysfunctions in the central nervous system (Oppenheim et al., 1999) and is associated with the proprioceptive regulation (Golomer et al., 1994; Golomer et al., 1999). With regard to the experimental setup of our study, the MF band also corresponds to the frequency band of the perturbation signal (Figure 5.2, non-shaded area) and by checking the LF and HF bands I aim to investigate the aftereffects of the perturbation and the effect of holding a handle.

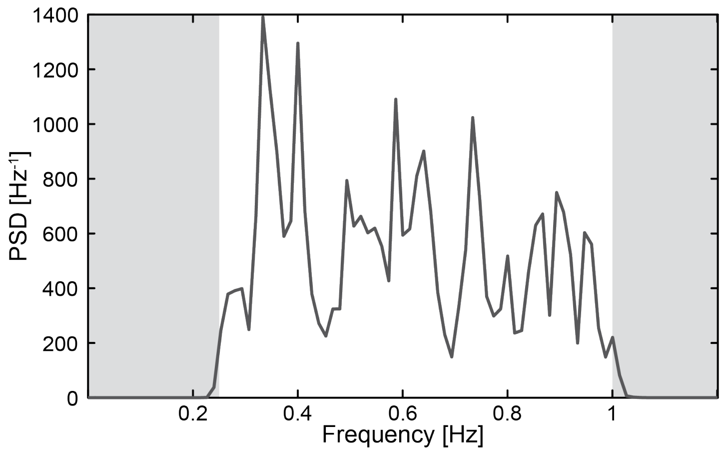

Figure 5.2: Power spectral density (PSD) of the perturbation signal. The diagram shows the perturbation signal's power versus frequency in a range between 0.25 and 1 Hz (non-shaded area). The PSD of the perturbation signal was identical for all participants in both trials.

Assisted standing decreases variability of COP excursions.

The analysis of mean (MNF) and median (MDF) frequency of the group average COP excursions showed a reduction when the participants were holding a handle. On average, the MNF of the participants' COP excursion frequency was lower in the *WH* condition ($M = 0.601, SE = 0.026$), compared to the *NH* condition ($M = 0.719, SE = 0.012$). This difference, 0.118, BCa 95 % CI $[0.076, 0.161]$, was found to be statistically significant $t(9) = 4.766$, $p = 0.001$ and represented a large-sized effect of $d = 1.51$. A similar result was shown for the MDF, there the values in *WH* condition ($M = 0.549, SE = 0.032$) were on average lower (0.097, BCa 95 % CI $[0.038, 0.168]$) than those in the *NH* ($M = 0.647, SE = 0.01$) condition. Like for MNF, the difference between *WH* MDF and *NH* MDF was statistically significant ($t(9) = 3.287$, $p = 0.009$, $d = 1.04$).

Holding a handle reduces the high frequency component of the COP excursions. Results of the frequency breakdown analysis of the COP signal are shown in Figure 5.4 where the individual frequency band is presented as a percentage of the sum of the three bands. The largest percentage of frequencies can be seen in the MF band, which is to be expected since this was the frequency range of the perturbation which significantly affected the movement and posture of the participants (Čamernik et al., 2016). However, a significant difference between the conditions was shown in the LF ($t(9) = 3.05$, $p = 0.014$, $d = 0.968$) and HF band ($t(9) = -7.1$, $p < 0.001$, $d = 2.245$). In LF band, the percentage of the total signal was 9.57 % higher in favor of the *WH* condition, and in HF band, the percentage was 5.85 % higher in favor of the *NH* condition.

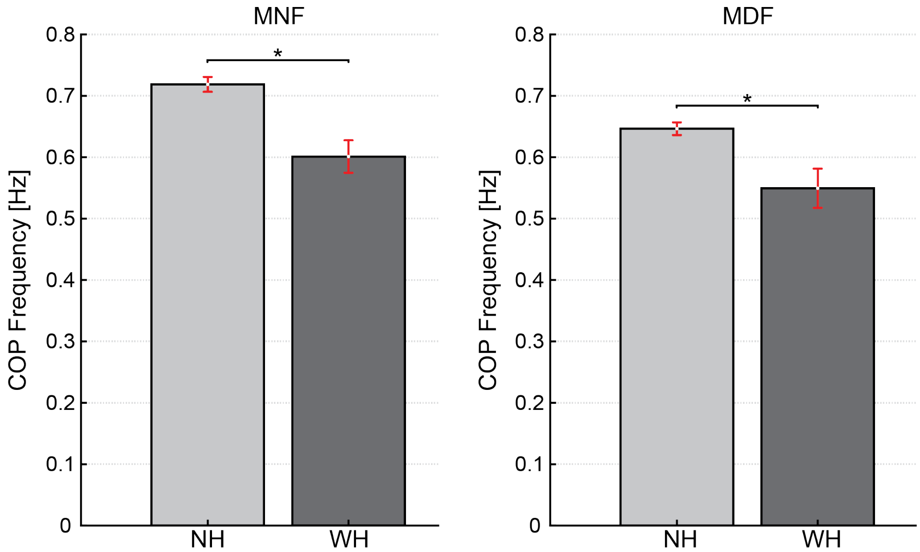

Figure 5.3: Mean and median frequency of COP excursions. Comparison of MNF and MDF of COP excursions between trials when the participants held the handle and stood freely. Both MNF and MDF show significantly lower values in *WH* condition, indicating a lower movement variability.

Conclusions

Frequency analysis of the COP signals yielded a few interesting finds. First, an increase in the mean and median frequencies in the *NH* trial indicates a significant decrease in postural stability. We have shown that the increased postural threat, induced by releasing the handle in the *NH* trial, was manifested in a lower position of the COM, an increase in muscle activity and in a higher frequency of whole-body movement.

Second, in a sense of mechanically compensating the effects of perturbation, holding a handle mostly affected the high-frequency (HF) component of the COP. Frequencies above 1 Hz on average occupied 5.8 % less share of all frequencies when comparing both conditions (Fig. 5.4). Lower share of frequencies in the HF band explains larger displacement of the COP as it was found in the initial investigation. Holding a handle caused the movement of the participants to be slower and more in sync with the perturbation.

Third, due to the lower contribution of the somatosensory system in the *NH* trial (i.e. no haptic information from the hand), the share of the HF band was significantly larger. Holding a handle caused a reallocation of the high frequency share to the medium and low frequency bands. As it was previously found in the light touch studies (Jeka & Lackner,

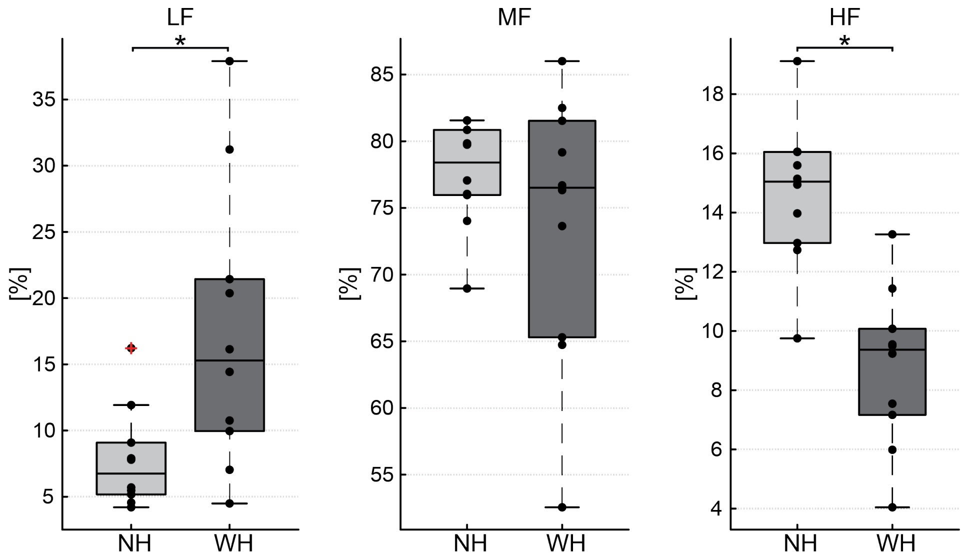

Figure 5.4: Low, medium and high-frequency bands of COP excursions. Comparison of LF, MF and HF bands of COP excursions between trials when participants either used additional hand contact or not. In both conditions, the largest share of frequencies was located in the MF band, followed by the HF and LF bands. Statistically significant differences between the *WH* and *NH* conditions are indicated for the LF and HF bands ($^*p \leq 0.02$).

1995; Jeka & Lackner, 1994), the haptic cues coming from the hand holding a handle which mechanically stabilized the body, also diminish the threshold of detection of body movement and thus reduce the HF component of the COP signal. The faster detection of body movement allowed corrective actions to begin earlier than in the *NH* condition and illustrates a feed-forward mechanism.

5.1.2 Estimation of whole-body effort

With regard to motor adaptation researchers found that people tend to move in an energetically optimal way (Huang et al., 2012; Selinger et al., 2015; Todorov & Jordan, 2002). Adaptation may occur due to the years of experience which allows people to easily anticipate and adjust their movements in familiar situations or as it was found more recently, such adaptation can also occur immediately, in only a few iterations of reaching and walking tasks (Huang et al., 2012; IJmker et al., 2015). Therefore, in addition to the time and frequency analysis of balance and posture during the two trials of continuous balance perturbations, I performed an estimation of whole-body effort to measure and evaluate the adaptation of the locomotor system.

To estimate the effect of holding a handle on whole-body effort during a prolonged exposure to balance perturbations, I calculated the cumulative joint torques with the inverse dynamics procedure. For this purpose, I used MuJoCo 2.0 (Fig. 5.5), a software developed by Emanuel Todorov for Roboti LLC (Todorov et al., 2012). MuJoCo stands for Multi-Joint dynamics with Contact and it is a physics engine aiming to facilitate research and development in robotics, biomechanics, graphics and animation. It is a commercial software, however it is also freely available to students for personal projects, upon request.

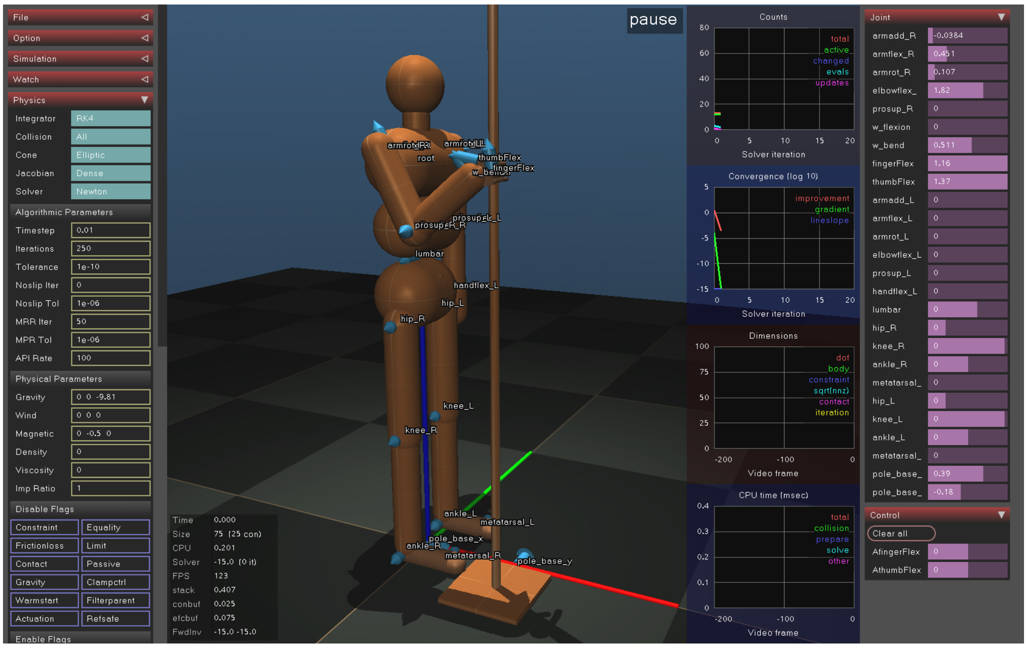

Figure 5.5: MuJoCo 2.0 graphical user interface.

The multi-joint humanoid model

To solve the inverse dynamics problem (i.e. calculate kinetics, based on experimental kinematics data) several steps had to be taken. First, I constructed a human-like model (humanoid) by using the MuJoCo specific modeling format called MJCF. This format has an XML-tree structure and consists of MuJoCo specific XML elements and their attributes to design a multi-joint body. Such model is built by different elements like bodies, geometries, joints, visual representation objects, etc., which have different attributes (position, orientation, inertial properties, etc.). The MJCF model can be constructed in a way to resemble real body dimensions and attributes, therefore, I designed a template model which can be used to fill in the body parameters specific to a real human subject.

The template of a humanoid model consists of 16 body segments and 24 joints (Fig. 5.7). A description of all body segments and joints is presented in Table 5.1.

Table 5.1: Description of a multi-joint model. A list and description of body segments and joints in the humanoid model used in the inverse dynamics procedure to calculate joint torques.

No.	Body Seg.	Description	Joint	DOF
1.	torso_U	thoracic region of the torso	*root*	6
2.	head	head and neck	*none*	
3.	uarm_R	right upper arm - shoulder to elbow	*armadd_R, armflex_R, armrot_R*	3
4.	larm_R	right lower arm - elbow to wrist	*elbowflex_R, prosup_R*	2
5.	palm	right palm	*w_flexion, w_bend*	2
6.	fingers	right index, middle, ring and little finger	*fingerFlex*	1
7.	thumb	right thumb	*thumbFlex*	1
8.	uarm_L	left upper arm - shoulder to elbow	*armadd_L, armflex_L, armrot_L*	3
9.	larm_L	left lower arm - elbow to wrist	*elbowflex_L, prosup_L*	2
10.	hand_L	combined left palm and fingers	*handflex_L*	1
11.	torso_M	lumbar region of the torso	*lumbar*	1
12.	torso_L	sacral region of the torso	*none*	
13.	thigh_R	right upper leg - hip to knee	*hip_R*	1
14.	shank_R	right lower leg - knee to ankle	*knee_R*	1
15.	foot_R	right foot - ankle to 5th metatarsal	*ankle_R*	1
16.	toes_R	right toes - 5th metatarsal to toe tip	*metatarsal_R*	1
13.	thigh_L	left upper leg - hip to knee	*hip_L*	1
14.	shank_L	left lower leg - knee to ankle	*knee_L*	1
15.	foot_L	left foot - ankle to 5th metatarsal	*ankle_L*	1
16.	toes_L	left toes - 5th metatarsal to toe tip	*metatarsal_L*	1

To calculate the human body parameters (body parts sizes, masses and inertial properties) I used a custom-made software BodyModel developed by Leon Žlajpah and Jan Babič - Fig. 5.6 (Žlajpah & Babič, 2014).

By inputting the subject's body weight and height, the algorithm calculates the individual body segment size, mass and inertial properties using coefficients based upon the regression analysis of de Leva (1996) performed for male adults. The model template is then populated with the subject-specific parameters so that the body dimensions and inertial parameters of the humanoid model approximately match the real human body.

The MuJoCo inverse dynamics procedure requires joint angles positions, velocities and accelerations as an input. Additionally, external forces acting on the participant during the task can be used as an input. In this case the external forces acting on the body would be the ground reaction forces (measured by the force plate), the perturbation (measured by the force sensors on the waist belt) and in case of the *WH* trial, forces acting on the hand that was holding the handle (measured by the force sensor on the handle). I calculated the joint angles positions by using the kinematics data, acquired during the experiment as described in Čamernik et al. (2019). The nature of the experiment yielded in predominantly planar movement of the participants, therefore, only the 2D position data of active markers placed on the participant's joint centers, bony landmarks and segments

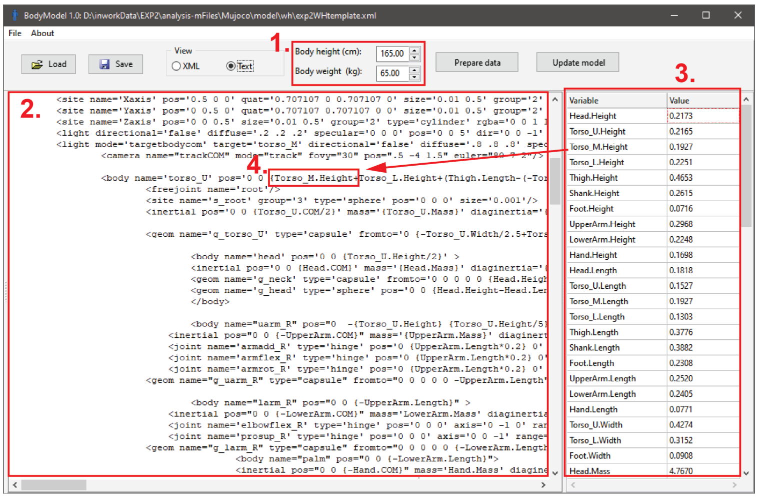

Figure 5.6: BodyModel 1.0. A custom-made application for calculating human body parameters. The input (1.) are the subject's body weight and height and (2.) the MJCF template. After the algorithm calculates the body parameters (3.), the MJCF template is populated with the appropriate values (4.)

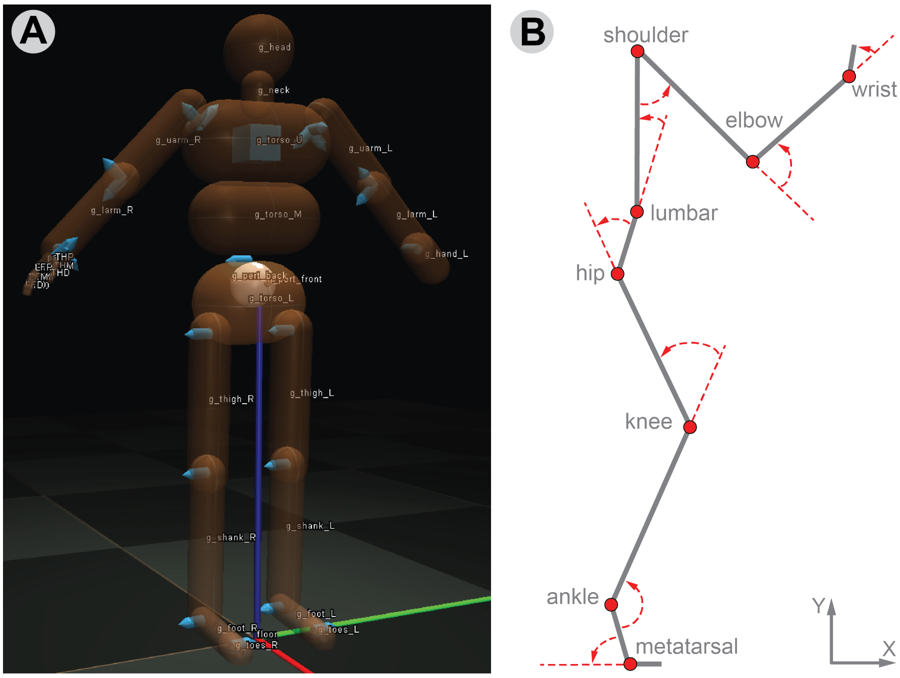

Figure 5.7: A multi-joint model of a human. **A:** A simplified model of a human body, consisting of 16 body segments and 24 joints. The joints between the body segments are shown as blue cones and the COM of the model is illustrated as a white sphere in the lower part of the torso. **B:** A stick-man model with joint positions and direction of rotation - only in sagittal plane.

were used to calculate the orientation of individual body segments (θ, Eq.5.4a). Then, the joint angles were obtained by calculating the angles between the adjacent body segments (q, Eq.5.4b). Joint velocities ($\dot{q}$, Eq.5.4c) and joint accelerations ($\ddot{q}$, Eq.5.4d) were calculated by using the first and second order derivative of the joint angles position, respectively.

$$\text{a) } \theta_{ij} = \arctan \frac{y_j - y_i}{x_j - x_i} \qquad \text{b) } q_i = \theta_j - \theta_i \qquad \text{c) } \dot{q}_i = \frac{\partial q_i}{\partial t} \qquad \text{d) } \ddot{q}_i = \frac{\partial \dot{q}_i}{\partial t} \tag{5.4}$$

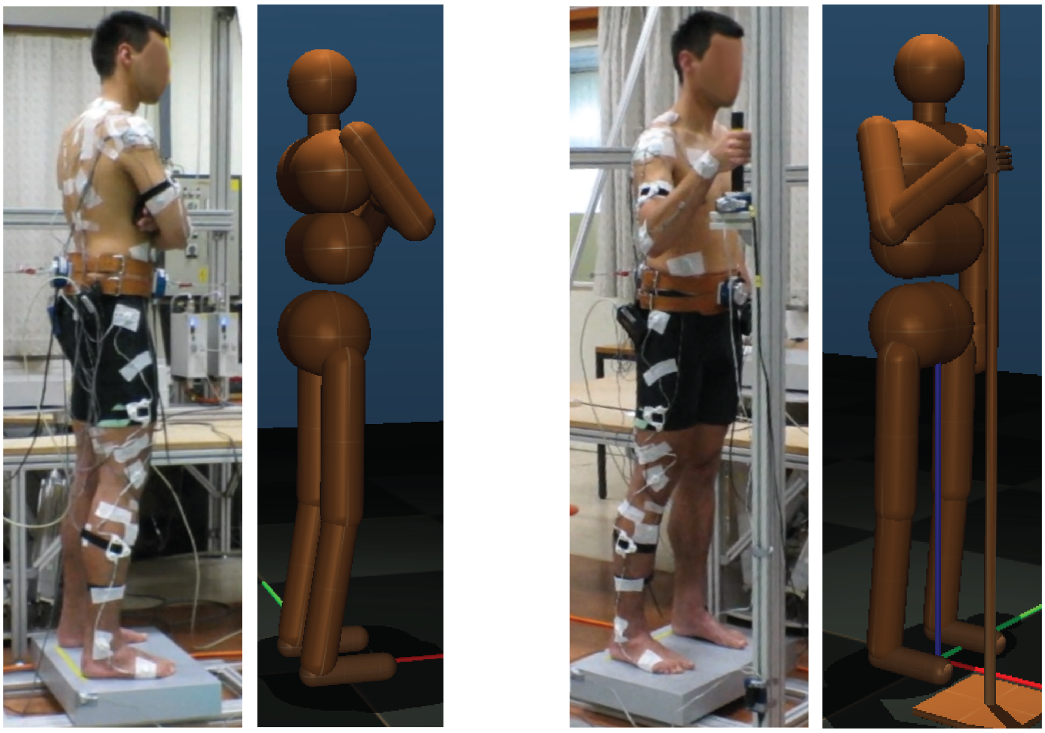

Figure 5.8: Whole-body model and real subject in *WH* and *NH* trial pose. A humanoid model was used in inverse dynamics procedure to calculate the joint torques. On the left panel of the figure a scaled model used in calculations for the *NH* trial is shown (participant is standing freely, both arms are folded together across the chest). On the right panel a scaled model for the *WH* trial is shown (participant is standing and grasping a stationary handle with his right hand).

To compensate for individual differences in body parameters of the participants, joint torque values of each joint were normalized by dividing them with a scaling factor specific to each participant (Eq. 5.5b). The scaling factor (s_{fac}) was calculated as a product of participant's weight (sub_w) times participant's height (sub_h) (Eq. 5.5a).

$$\text{a) } s_{fac} = sub_w \cdot sub_h \qquad \text{b) } \tau_{jnorm} = \frac{\tau_j}{s_{fac}} \tag{5.5}$$

where τ_{jnorm} is the normalized joint torque and τ_j is the calculated joint torque in each joint with a unit of Nm/kg·m.

The total body work, which is equivalent to the external work done by or on the body, was calculated as the summation of the work done by all the joint's torque. The measure of effort, i.e. the work done by a joint torque was calculated as the time integral of the product of joint torque times the angular velocity of the joint it crosses (Eq. 5.6) (Robertson et al., 2014),

$$W_{tb} = \int_{t=0}^{t=300} \sum_{joint=1}^{N_{joint}} P_j dt, \tag{5.6}$$

where W_{tb} is the total body work, P_j is the product of the joint torque (τ_j) and the angular velocity at j-th joint and N_{joint} is the number of joints. For NH condition the $N_{joint} = 7$ and includes the calculated work of the lumbar joint, both ankles, knees and hips. For the WH trial, the work of the shoulder and elbow joints of the active arm were added, hence $N_{joint} = 9$.

Results

The diagrams in Figure 5.9 show the comparison of mean normalized joint torques between the NH and WH conditions over the 5 minutes period of perturbed standing. Higher values, especially in the ankle and knee joints during the NH trial, indicate more movement and higher effort when participants were standing freely.

To estimate whole-body effort, the joint torques were transformed to work done in each joint. Group average results for each observed joint with confidence intervals are presented in Fig. 5.11. From visual inspection we can already observe the difference between the two conditions. However, to faithfully compare both conditions, all joints involved in balancing task need to be taken into consideration. The sum of joint work in WH trial thus includes the work in the shoulder and elbow joints of the arm holding a handle.

Due to the high variability of data, normality tests were performed on both trials. The Wilk-Shapiro test revealed the data calculated for cumulative joint work for each participant in NH trial were not normally distributed (Fig. 5.10, $p = 0.0046$), therefore, a Wilcoxon Signed-Ranks test was used to determine the difference between the two conditions.

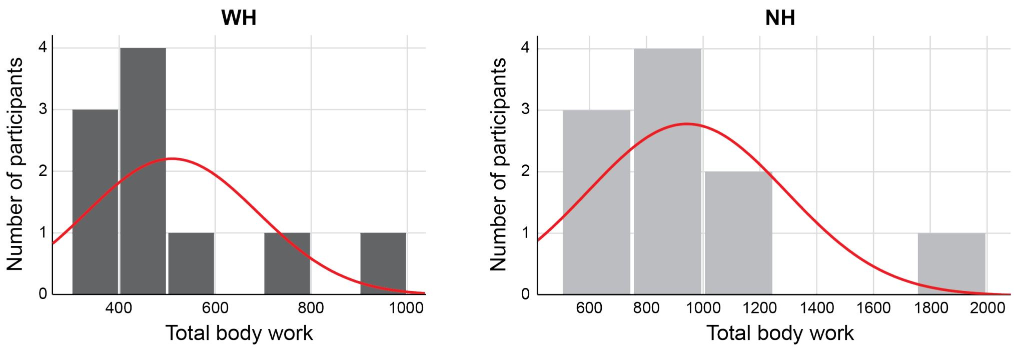

Figure 5.10: Normality test of cumulative joint work in WH and NH condition. The joint work scores in WH trial, $D(10) = 0.848$, $p = 0.055$, did not deviate significantly from normal; however, the NH trial $D(10) = 0.759$, $p = 0.005$, scores were significantly non-normal.

The test has confirmed that the cumulative joint work in the WH trial (mean rank = 5.78) was significantly lower than the cumulative joint work in the NH trial (mean rank = 3.0), Z = -2.5, $p = 0.01$ (Fig. 5.11, right panel).

Conclusions

Metabolic energy consumption in motion-rich environments has been shown to be an important variable in whole body movement tasks like walking (IJmker et al., 2014) and standing (Duncan et al., 2018). Authors used different approaches to evaluate the metabolic energy consumption, but came up with similar conclusions. IJmker et al. (2014) perturbed healthy adults while walking on a treadmill under four conditions of varying postural threat. By calculating the energy cost of walking from the data of oxygen uptake and respiratory exchange ratio, they found that the metabolic demand significantly increased with the largest postural threat. At the same time, the gait parameters have also changed (shorter and faster steps). Since the observed changes occurred independently of walking speed, they concluded that increasing postural threat is the more important factor that leads to a decrease in gait stability and economy. Similarly, Duncan et al. (2018) studied the effects of postural perturbations initiated by a moving environment on the metabolic energy consumption of healthy adults while standing. The participants were standing freely or performing a lifting task on movable platform and their cardiorespiratory and heart rate parameters ware recorded to estimate their metabolic cost. They found that a systematic increase in metabolic cost is associated with increased platform motions and the task.

In our study, we estimated the whole-body effort by using the inverse dynamics procedure. By designing a scalable humanoid model and applying the measured kinematics and dynamics properties, we aimed to faithfully calculate the generalized joint torques and estimate the work done at each joint. The analysis of the average cumulative joint work has shown a significant increase in case of the *NH* trial. This is interesting because even though the perturbations themselves were not large enough to force the participants to lose balance, they appeared to be threatening enough to induce a change in a postural strategy. This change was seen in more flexed joints, higher muscular activity and higher movement variability which all led to higher energy expenditure. Our findings here emphasize the importance of holding a handle to augment balance control. Not only that balance control is improved, also the total body energy expenditure is lowered. Furthermore, as in studies of IJmker et al. (2014) and Duncan et al. (2018), we have shown that the increased postural threat increases the metabolic cost and changes the postural strategy.

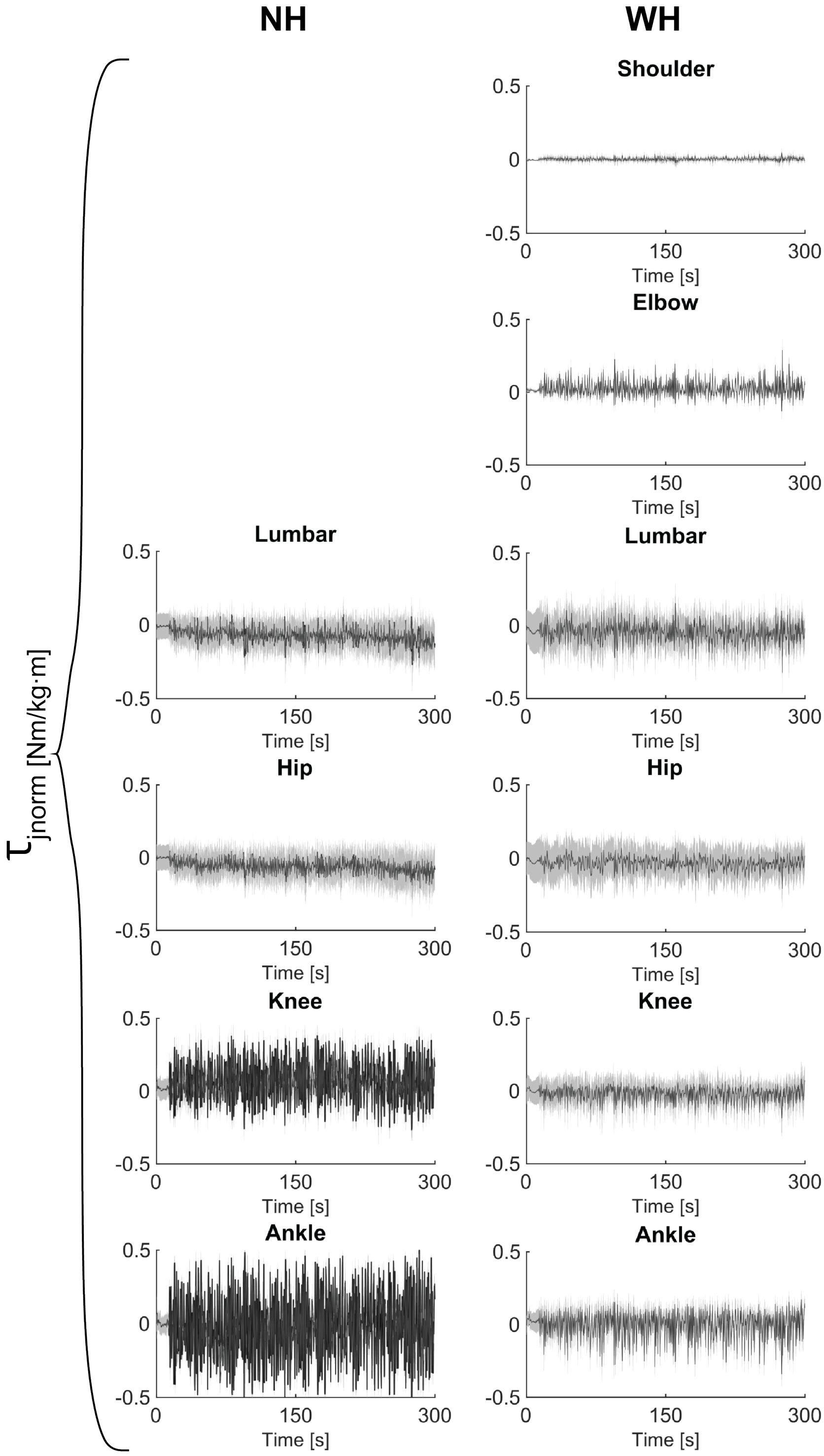

Figure 5.9: Generalized joint torques. Group mean values with 95 % CI of normalized joint torques during the 5-minute trials in *NH* and *WH* conditions. Torque values of shoulder and elbow joints of the arm holding the handle are included in the diagrams of the *WH* trial. The joint torques are expressed relative to the body mass times body height (Nm/kg·m).

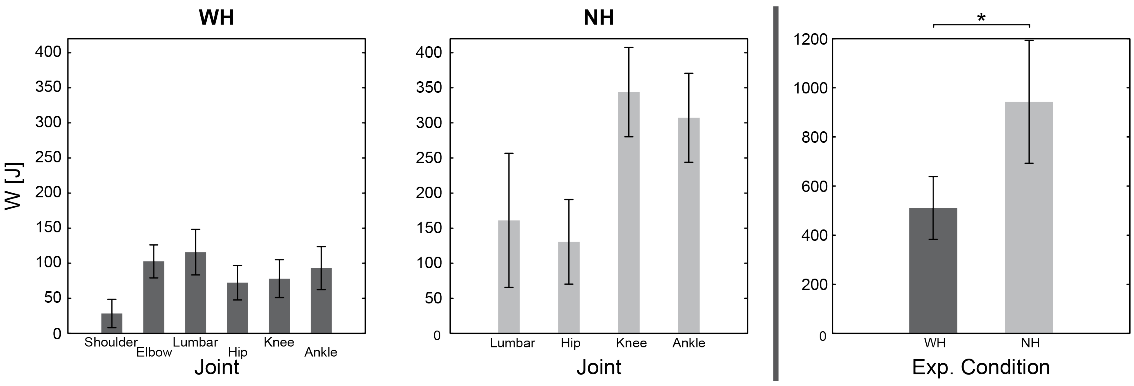

Figure 5.11: Individual and cumulative joint work (W_{tb}). In the left panel group mean values with 95 % CI of normalized joint work during the 5-minute trials in *NH* and *WH* trials are presented. On the right panel, the mean cumulative joint work with 95 % CI is shown. The values in the *NH* trial are significantly higher even though less joints are taken into consideration ($*p < 0.01$).

Chapter 6

Study II - Mechanisms of Postural Control Change with Location of Supportive Hand Contact

Holding a handle for balance is a common fall-prevention strategy when for example we are riding a bus. When the bus hits a hole or bump, suddenly breaks or is making a turn, different perturbations compromise the balance of standing passengers. Keeping balance in such situations can sometimes be really challenging and without holding a handle impossible. But how should one hold a handle to have the best control over his balance? Is it better to hold on to the handle at hip height or overhead?

In this part, I present the results of the second study with the focus on effects of posture and position of the handles on the ability to maintain balance after discrete perturbations. The goal of this second study was to identify the effects of the position of the handle with respect to the participant on postural control during discrete perturbations of varying magnitudes. We addressed the hypotheses H4 and H5 where we speculate that the mechanisms used to control balance are based on the position of the supportive hand contact and that holding a handle at shoulder height provides the best control of balance.

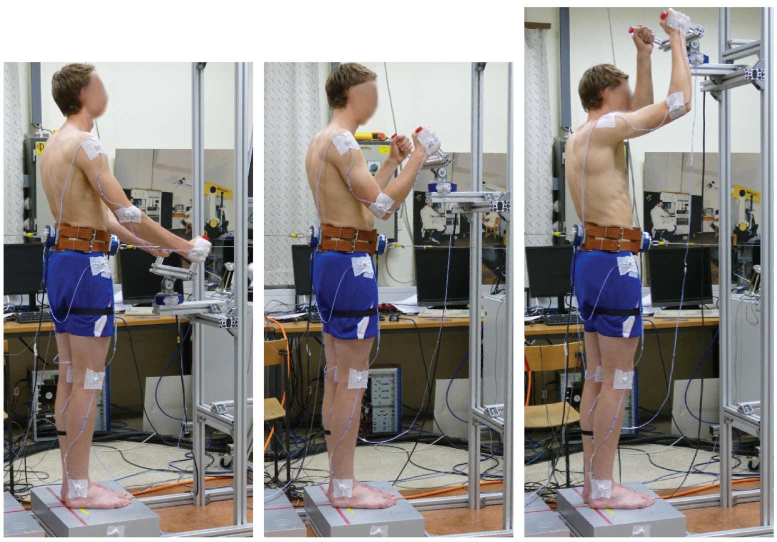

Figure 6.1: Study II. Mechanisms of postural control change with the location of supportive hand contact.

The experiment of the second study included five different postures: two without holding a handle and three with holding a handle, located at three different positions. We applied discrete perturbations of varying magnitudes directly to the COM of the participants, and observed their motor responses.

First, I present the summary of the main results and conclusions which are collected in a paper "Staying on your feet: the effectiveness of posture and handles in counteracting balance perturbation" by Jernej Čamernik, Morteza Azad, Luka Peternel, Zrinka Potocanac and Jan Babič (Čamernik et al., 2019). The paper was published in a peer-reviewed journal *Ergonomics* in 2019 and as a first author I participated in designing the experiment, conducted the measurements, analyzed the data including programming of the data analysis procedures, and wrote the manuscript.

Second, the results of additional analyses of whole-body effort are presented.

ERGONOMICS
https://doi.org/10.1080/00140139.2018.1559363

RESEARCH ARTICLE

Staying on your feet: the effectiveness of posture and handles in counteracting balance perturbation

Jernej Čamernik[a,b]*, Morteza Azad[d], Luka Peternel[a,c], Zrinka Potocanac[a] and Jan Babič[a]

[a]Laboratory for Neuromechanics and Biorobotics, Department of Automation, Biocybernetics and Robotics, Jožef Stefan Institute, Ljubljana 1000, Slovenia; [b]Jožef Stefan International Postgraduate School, Ljubljana, Slovenia; [c]HRI^2 Lab, Department of Advanced Robotics, Istituto Italiano di Tecnologia, Genoa, Italy; [d]School of Computer Science, University of Birmingham, Edgbaston, The United Kingdom

ABSTRACT

Stairways, public transport and inclined walkways are often considered as sites with higher likelihood of falls due to a sudden loss of balance. Such sites are usually marked with warning signs, equipped with non-slip surfaces and handles or handrails to avert or decrease this likelihood. Especially, handles are supposed to provide additional support in cases of a sudden loss of balance. However, the mechanisms of using handles for balance at different heights are not yet fully disclosed. We simulated full body perturbations by applying an anterior force to the waist and investigated effectiveness and mechanisms of balance recovery in five different postures: step stance and normal stance with or without holding handles at different heights. Results indicate that both step stance and holding handles at different vertical positions sufficiently assist balance recovery, compared to normal stance. While there was no significant effect of handle in CoM displacement, the shoulder height handle required the lowest handle force, indicating a difference in using the handle.

Practitioner summary: To investigate handle use for balance recovery, we perturbed healthy young adults in different standing positions. Even though the use of different handles had a similar effect, the lowest forces were exerted on the shoulder height handle indicating a preferred handle position for balance recovery.

Abbreviation: AP: antero-posterior; CNS: Central nervous system; CoM: Center of Mass; CoMmax: Maximal displacement of the center of mass; CoP: Center of pressure; FHmax: Maximal resultant force exerted on the handle; hFHmax: Maximal horizontal force exerted on the handle; vFHmax; Maximal vertical force exerted on the handle; M1-M8: Perturbation force magnitude

ARTICLE HISTORY
Received 23 May 2017
Revised 14 November 2018
Accepted 8 December 2018

KEYWORDS
Slips; trips and falls; biomechanics; hand forces; balance

Introduction

Unpredictable perturbations compromise our postural stability in many everyday situations. Falls, which are a direct consequence of balance loss, often lead to injuries or even fatalities. In EU, falls are reported to be one of the highest causes of injury and even of fatal injuries (17% of all fatal injuries) and remain a major cause of workplace injury (EuroSafe 2013). Falls and fall prevention have been extensively studied to understand the mechanisms involved, and research has led to various evidence-based prevention practices (Choi and Hector 2012; Cadore et al. 2013). Nevertheless, in a recent paper by Chang et al. (2016) which summarises the current state of science regarding occupational slips, trips and falls on the same level, the authors concluded that, in spite of the progress in recent decades, there is an urgent need to develop better fall prevention strategies. Furthermore, they emphasise that most investigators agree that reactive recovery characteristics following the perturbation are directly linked to how we maintain dynamic stability. Therefrom, they propose further efforts in improving reactive recovery scenarios, which include control over initiating circumstances and improvements in reactive recovery.

When standing, human posture is balanced if the ground projection of its centre of mass (CoM), falls within the convex hull of the foot support

CONTACT Jernej Čamernik jernej.camernik@ijs.si Laboratory for Neuromechanics and Biorobotics, Department of Automation, Biocybernetics and Robotics, Jožef Stefan Institute, Ljubljana 1000, Slovenia
Color versions of one or more figures in the article can be found online at http://www.tandfonline.com/terg.

area (base of support). However, this can be perturbed by movements of the COM or changes in the base of support (e.g. during walking) (Winter 1995). Hence, the central nervous system (CNS) continuously adjusts the strategies and mechanisms of postural control to prevent a potential loss of balance.

Postural reactions and adjustments following the loss of standing balance are based on the type, direction and magnitude of the perturbation (Diener, Horak, and Nashner 1988; Chen et al. 2014) and scale with biomechanical constraints (Park, Horak, and Kuo 2004). It has also been well established that additional haptic cues (e.g. touching a stable surface with a fingertip) play an important role in balance control while standing (Jeka and Lackner 1994; Albertsen, Temprado, and Berton 2010; Martinelli et al. 2015) and walking (Bateni and Maki 2005; Forero and Misiaszek 2013). Such tactile feedback provides the central nervous system with information of current position and orientation of the body and complement regulation of corrective reactions to balance disturbances.

Nevertheless, only lightly touching a stable surface is not sufficient to prevent balance loss following large perturbations (Kouzaki and Masani 2008). In such situations, one can achieve an additional mechanical support by extending the base of support by either making a step or grasping onto a stationary object. For example, a firm hand contact with a fixed handle on a moving bus or using a cane while walking contributes to a much better balance potential by increasing the size of the base of support and activating the arm muscles for balancing (Bateni and Maki 2005; IJmker et al. 2015; Čamernik et al. 2016). Apart from mechanical support, firmly holding onto handrails reduces energy cost of treadmill walking in stroke survivors, and alters their step parameters and muscle activity (IJmker et al. 2015). This indicates an important clinical role for handles, which should be shaped (Dusenberry, Simpson, and Dellorusso 2009) and positioned (Maki, McIlroy, and Fernie 2003) in an optimal way to facilitate balance control. Therefore, it is important to position handles appropriately with respect to a standing human in order to provide mechanically efficient, reliable and safe support.

In the context of working environment, one can make an analogy between handles used for balance and handles used in pushing and pulling activities. From biomechanical perspective, pushing objects produces forces in the hands from connection with the object (e.g. holding onto a handle on the cart). To overcome, sustain or resist these forces one must sufficiently activate appropriate muscles to produce enough torque at all the joints involved. Pushing and pulling was substantially studied to identify mechanical loading of the joints, factors affecting pushing and pulling forces and to estimate ergonomics consequences (for review, see Garg et al. [2014]). Conclusions from these studies vary quite substantially, however one of the common findings is that maximum initial and sustained forces for pushing and pulling depend on handle height, frequency of exertion and distance of pushing/pulling. As a risk factor Hoozemans et al. (2004) and Lett and McGill (2006) found the combination of force direction (pushing v. pulling), force magnitude, body posture and handle height to be responsible for detrimental movements in the shoulder and compressive and shear forces to low back. On the other hand, even though handles play an important role in fall prevention, current falls research literature is very limited when standing balance perturbations involve use of handles. To our knowledge, only a few studies investigated the role of handle location when healthy adults grasp the handle before the onset of perturbation (Maki, McIlroy, and Fernie 2003; Noé, Quaine, and Martin 2003; Babič et al. 2014; Sarraf, Marigold, and Robinovitch 2014). In a vehicle starts simulation study, Sarraf, Marigold, and Robinovitch (2014) concluded that overhead grasping minimises handle forces, while shoulder-height grasping minimises centre of pressure (CoP) displacement when perturbation occurs in the antero-posterior (AP) direction. Similarly, using AP perturbations and four handles, Babič et al. (2014) found that handles significantly reduced the maximal displacement of the subject's CoP regardless of the location, direction of the perturbation and its intensity. Importantly, handle location had a significant effect on forces exerted on the handle, so they speculated that a shoulder-high (or higher) handle enabled balancing with the least effort. Recently, two other studies were published, which investigated effects of handles on postural control from different perspectives. Beschorner et al. (2018), considered the downward-pull forces at different postures, but without including external perturbations, and Komisar, Nirmalanathan, and Novak (2018) focussed on CoM control and physical demands in perturbed reach-to-grasp scenarios. Beschorner et al. (2018) focussed on impacts of glove use and different handle heights on achievable downward pulling forces. They found that the achievable downward hand force increases with higher hand positioning and explained this result with the biomechanical mechanism of the length-tension relationship of the proximal muscles (biceps brachii and latissimus dorsi). On the

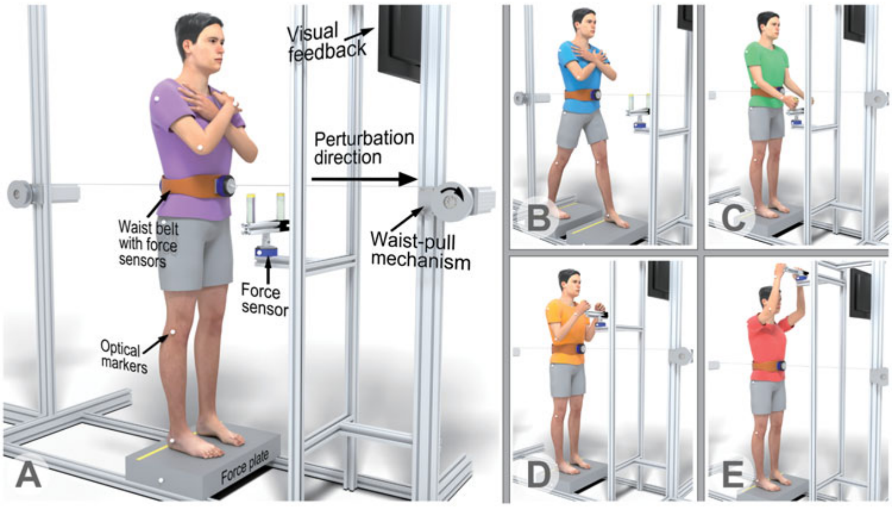

Figure 1. Schematic representation of the experimental setup. In the normal pose (A), subjects stood upright with their feet together and arms crossed over the chest. In the step stance pose (B), subjects stood with their arms crossed over the chest and their left foot 60 cm ahead of their right foot (ankle to ankle distance). In the remaining poses, the subjects stood upright with their feet together and held onto a handle located 30 cm in front of their bodies. The handle was vertically positioned at hip height in the low handle pose (C), at shoulder height in the middle handle pose (D), and above the head in the high handle pose (E).

other hand Komisar, Nirmalanathan, and Novak (2018), who positioned handrails at heights of 86.5, 96.5 and 106.5 cm, delivered balance perturbations to subjects and analysed their balance recovery reactions after reaching and grasping the handrail, found that higher handrails resulted in reduced forces exerted on the handrail and reduced peak CoM displacement.

The studies mentioned, mostly focussed on explaining the regularities in behaviour while using handles for balance (e.g. measured how underfoot centre of pressure position/movement changed with handle location) while the workload on human body was mainly neglected. Moreover, experimentally supported recommendations regarding optimal handles locations for balance recovery have not yet been given.

Here, we aimed to advance this work by investigating how changes in body pose (standing freely or holding a handle at different heights) affect the mechanical stabilisation of posture in response to discrete forward directed perturbations of varying magnitudes. Our goal was to categorise handle positions in connection with posture configurations, by accounting for whole body movement (CoM displacement) and forces exerted on the handle. Furthermore, we aimed to give a more profound insight into the mechanisms of assisted balance recovery and the associated loading. Based on the current literature on balance recovery and pushing/pulling activities, we hypothesised that the handles positioned at shoulder height would be the most efficient. The handle position of highest efficiency would be considered as the one with measured lowest forces exerted on the handle and the least displaced CoM during the recovery period occurring after the perturbation.

Methods

Subjects

Ten healthy male subjects (avg. ± SD: age 21.7 ± 2.2 years, height 183 ± 4.6 cm, body mass 76.8 ± 8.1 kg) participated in this study after giving informed consent. The study was approved by the Republic of Slovenia National Medical Ethics Committee (No. 112/06/13).

Measurement protocol

The experiment (Figure 1) consisted of five different standing poses; narrow (*normal*) and step stance (*step*) and holding onto a handle at three different heights (*low, middle and high*). We adjusted the horizontal distance and heights of the handles according to each subject's anthropometry. In the low handle pose, the handle was at subject's hip height (10° flexion in the elbow joints), in the middle handle pose at shoulder height (90° flexion in the elbow joints), and in the high handle pose above the head (10° flexion in the elbow joints). Our in-house made force-controlled pulling mechanism (Peternel and Babič 2013) perturbed the subjects in forward direction by producing a horizontal external force at the approximate position of

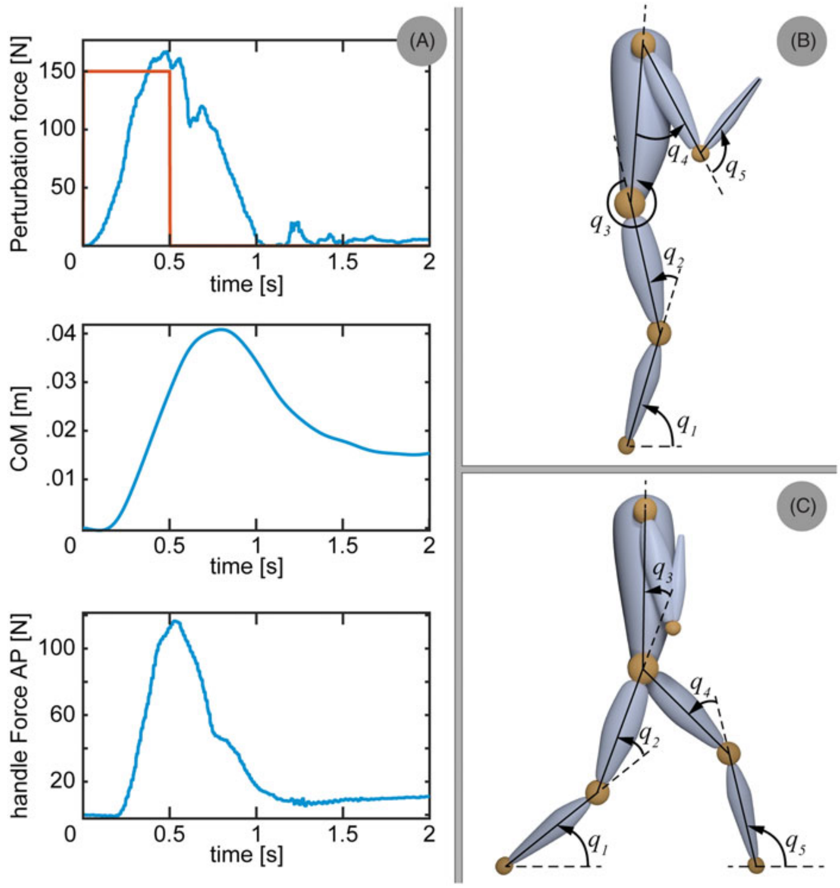

Figure 2. Details of the experimental methods. (A) An example of trajectories from a single trial corresponding to the first 2 s of perturbation and recovery period. In this example, perturbation magnitude of M8 was applied to a subject whose body weight was 76.5 kg while holding a handle at shoulder-height. In the top diagram, we show individualised perturbation force signal (red line) and actual force as measured by the force sensors on the waist belt (blue line) which was applied to the subject's approximate location of CoM. Trajectory of CoM displacement, is shown on the middle diagram and in the bottom diagram, we show force trajectory in antero-posterior direction as measured by the force sensor at the handle. (B and C) Diagrams of inverse dynamic model. (B) The model consists of five links connected by actuated revolute joints (q_1–q_5). Note that the first link (lower leg) was connected to the ground, as the subject's feet did not move during the experiments. To model the *normal* and *step* poses, we locked the degrees of freedom (DoF) of the arms, effectively creating a 3 DoF model (q_1, q_2 and q_3). (C) For the other poses, the model was constrained at the hand to model the handle contact.

their CoM. The command signal was a step with 0.5 s width (see Figure 2A) which was sent to the pulling mechanism after a random pause to avoid habituation. The actual perturbation force was controlled through a combination of system's dynamics model (including stiction and friction compensation) and a force-feedback PID controller. The feedback force was measured by a force sensor attached to the human body by a belt. The speed of pulling was not directly controlled; however, it was limited by an upper bound for safety reasons. For more details, including the design, parameter setting and accuracy analysis, please refer to (Peternel and Babič 2013; Peternel 2015).

We used eight perturbation force magnitudes (M1–M8), linearly increasing from 2.75% (M1) to 22% (M8) of the subject's body weight. In spite of the linear increase of the perturbation magnitude, the duration of the perturbation impulse remained the same - 0.5 s. Ten trials of each perturbation magnitude were performed for each pose, blocked by difficulty (80 trials per subject, per pose). Within a pose, blocks were presented in an increasing difficulty. A random pause varying between 0.5 and 5 s was given between trials to minimise the effects of anticipatory postural adjustments (Santos, Kanekar, and Aruin 2010). Between poses, the subjects rested for 10 min. In the first pose (*normal*) all subjects had to take a recovery step before the maximum perturbation magnitude was reached. When a step was made, the experimenter stopped the procedure and continued the experiment in the next pose. Compensatory stepping did not occur in other poses and the subjects returned to their initial position after each perturbation.

Kinematics were measured using the 3 D InvestigatorTM system (NDI, Waterloo, Ontario) consisting of 2 × 3 camera array at a sampling rate of 500 Hz. Active markers were firmly attached to the skin by means of double-sided adhesive tape on the apparent axis of rotation of the right ankle, knee, hip, shoulder, elbow and wrist joints. In post-processing, kinematic data were low pass filtered with a zero lag, 4th order Butterworth filter with a cut-off frequency 5 Hz (Bartlett 2007). Ground reaction forces and centre of pressure (COP) positions were measured using two force plates (9281CA, Kistler Instrument AG, Winterthur, Switzerland) at a sampling rate of 500 Hz (Winter 2009). There was no filtering applied to the CoP data. We placed a monitor in front of the subjects to provide visual feedback of starting position based on CoP position recorded prior to the start of each pose. After every trial, subjects were required to assume the same pose, by aligning their current CoP to the recorded CoP position, both displayed on screen. Custom made, vertically symmetrical handles (diameter = 3.2 cm, length = 12 cm, with adjustable positioning) were held with both hands in order to prevent rotations of the upper body and mounted on a 3-axis force sensor (45E15A, JR3, Woodland, USA) to measure the resultant forces between the handles and the subject. Force sensor data were not filtered and contribution of the torque due to the handle lever has been accounted for. The data from the force plate, force sensor in the handle and kinematics were collected synchronously at 500 Hz by using a custom-made application in MATLAB Release 2015b (The MathWorks, Inc., Natick, Massachusetts, United States).

We considered the recovery response as a 2 s period after the start of perturbation. During this recovery period, we calculated the maximal CoM displacement (CoM_{max}) and maximal resultant force exerted on the handle (F_{Hmax}). Additionally, we separately analysed horizontal (forward/backward, hF_{Hmax}) and vertical (vF_{Hmax}) components of the F_{Hmax}. We obtained the CoM_{max} by using a planar inverse dynamics model, built with a segmental method (Hamill and Knutzen 2008) and by using the segment inertial parameters from Leva (1996). To model the handle poses, we included five joints in the model (ankle, knee, hip, shoulder and elbow - Figure 2B). For *normal* pose model, we locked the joints of the arms, hence the model consists of three joints (ankle, knee and hip) and to model the *step* pose, we included the front leg joints and locked the joints of the arms (right ankle, right knee, right hip, left knee and left ankle - Figure 2C). The first segment of the model (lower leg) in all poses is connected to the ground. This is because we assume that the feet of the subjects do not move during the experiments. In no handle poses, the model is otherwise unconstrained and in handle poses, the model is constrained at the hand to model the handle contact. The total body CoM was then calculated as a weighted sum of the masses of the model's segments.

$$\vec{r} = \sum_{i=1}^{N} \frac{m_i}{m} \vec{r}_i, \tag{1}$$

where $\vec{\mathbf{r}}$ is the vector of the CoM of the body, N is the number of rigid body segments, m_i is the mass of the i-th segment, m is the total mass of the body and $\vec{\mathbf{r}}_i$ is the vector of the CoM of the i-th segment.

Maximal CoM_{max}, F_{Hmax}, hF_{Hmax} and vF_{Hmax} were averaged over pose and perturbation magnitude for each subject and used in statistical analysis. Statistical analyses were performed using SPSS 21 (SPSS Inc., Chicago, USA) and the level of significance for all statistical analyses was set at $p < .05$.

The effect of pose on CoM_{max}, F_{Hmax}, hF_{Hmax} and vF_{Hmax} was investigated using a one-way repeated measures ANOVA for each perturbation magnitude separately due to the missing data from the *normal* pose in higher perturbation magnitudes. The Greenhouse–Geisser correction was applied when Mauchly's test indicated violation of sphericity and paired t-tests were used for post-hoc testing. To account for multiple comparisons in paired t-tests, we corrected the calculated p values by applying the Bonferroni correction.

Results

CoM_{max} increased with an increase of applied perturbation magnitude as shown in Figure 3A. We did not find an effect of the pose for the weakest (M1) perturbation magnitude. However, pose had a significant effect on CoM_{max} in the subsequent three perturbation magnitudes (M2: $F(1.61, 14.51) = 14.08$, $p = .001$, $\eta_p^2 = .61$; M3: $F(1.58, 14.19) = 35.87$, $p < .001$, $\eta_p^2 = .8$; M4: $F(1.11, 7.79) = 25.64$, $p < .001$, $\eta_p^2 = .79$).

At the M2 perturbation magnitude the CoM_{max} was significantly larger in the *normal* pose compared to *step* and *middle handle* ($t(9) = 3.72$–5.43, $p < .048$) poses and significantly smaller in the *step* pose compared to all other poses ($t(9) = -5.38$–5.43, $p < .047$) (Figure 3B). As we increased the perturbation magnitude to M3, the

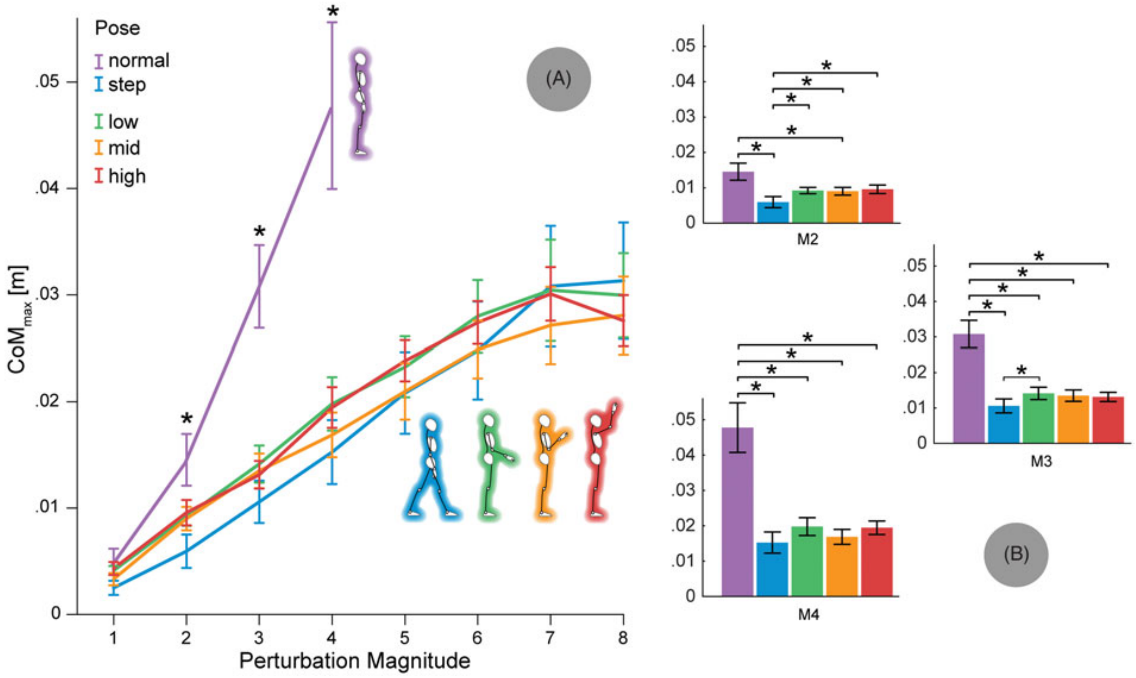

Figure 3. Maximal CoM displacements (CoM$_{max}$) across poses and perturbation magnitudes (A). At perturbation magnitudes of M2, M3 and M4, pose had a statistically significant effect on CoM$_{max}$ (B). Error bars indicate standard errors of the mean. Statistically significant effects of pose are marked with an asterisk.

CoM$_{max}$ in the *step* pose remained smaller only from the *normal* and *low handle* poses ($t(9) = -3.75–7.74$, $p < .046$), while the CoM$_{max}$ in the *normal* pose remained larger than all other poses ($t(9) = 5.79–7.74$, $p < .003$). A further increase in perturbation magnitude to M4 resulted in significantly larger CoM$_{max}$ in the *normal* compared to all other poses ($t(7) = 4.64 - 6.48$, $p < .024$). After the M4 perturbation magnitude, most of the subjects could not withstand the perturbation without making a step when positioned in *normal* pose. Therefore, the *normal* pose was excluded from the analysis of the remaining perturbation magnitudes. We found no statistically significant effects of pose at perturbation magnitudes above M4.

F$_{Hmax}$ is shown in Figure 4A. We found statistically significant differences between F$_{Hmax}$ in perturbation magnitude M2 ($F(2, 18) = 4.45$, $p = .027$, $\eta_p^2 = .33$) and the two highest perturbation magnitudes (M7 and M8, $F(2, 18) = 4.19–5.42$, $p = .014–.032$, $\eta_p^2 = .32–.38$), in which forces measured were higher for the *high handle* pose, compared to the *middle handle* pose (M2: $t(9) = 3.14$, $p = .036$, M7: $t(9) = 3.87$, $p = .011$, M8: $t(9) = 3.89$, $p = .011$).

Finally, by decomposing the resultant F$_{Hmax}$ to horizontal (forward/backward = hF$_{Hmax)}$ and vertical (vF$_{Hmax}$) components, we analysed how each component contributed to balance (Figure 4B and C).

Main effect of pose on hF$_{Hmax}$ (Figure 4B) was found in M1 ($F(2, 18) = 3.82$, $p = .042$, $\eta_p^2 = .3$) and in M5–M8 ($F(2, 18) = 3.88–7.01$, $p = .006–.04$, $\eta_p^2 = .3–.44$). Post hoc t-test showed that in M1, M6 and M8, values of hF$_{Hmax}$ were significantly higher in the *middle handle* pose, when compared to the *high handle* pose ($t(9) = 3.31–4.16$, $p = .007–.026$).

For vF$_{Hmax}$ (Figure 4C), this effect was found to be statistically significant in all perturbation magnitudes ($F(2, 18) = 11.86–32.53$, all $p < .001$, $\eta_p^2 = .3–.44$). The post-hoc t-test showed that in all perturbation magnitudes, values in *high handle* pose were statistically significantly higher than values in *middle handle* pose (mean difference: 10.37–74.21N, $t(9) = 5.78–8.73$, all $p < .001$). Moreover in perturbation magnitudes M1, M2, M5, M7 and M8, values of vF$_{Hmax}$ in *middle handle* were significantly lower than values in the *low handle* pose (mean difference: 6.26–43.88N, $t(9) = 3.09–4.32$, $p = .006–.039$)

Discussion

The main aim of this study was to further previous research investigating the effect of holding a handle on the mechanical stabilisation of posture by investigating whole body movement and forces exerted on the handle during recovery from discrete, forward

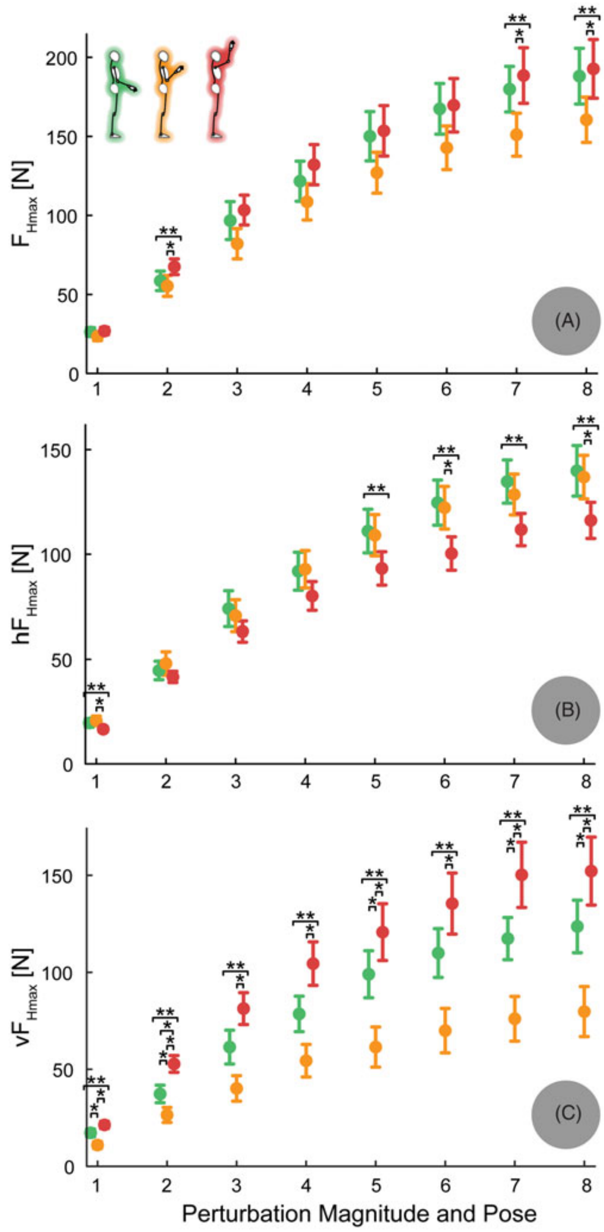

Figure 4. Maximal resultant force exerted on the handle (F_{Hmax}) across poses and perturbation magnitudes (A). At perturbation magnitudes of M2, M7 and M8, pose had a statistically significant effect on F_{Hmax}. Maximal forward/backward force exerted on the handle (hF_{Hmax}) across poses and perturbation magnitudes (B). At perturbation magnitudes of M1 and M5–M8, pose had a statistically significant effect on hF_{Hmax}. Maximal vertical force exerted on the handle (vF_{Hmax}) across poses and perturbation magnitudes (C). At all perturbation magnitudes, the effect of pose was significant (double asterisk). Post-hoc t-test showed that values of vF_{Hmax} from *high* and *middle handle* poses were significantly different in all perturbation magnitudes. Error bars indicate standard errors of the mean. Statistically significant effects of pose are marked with a double asterisk and statistically significant post-hoc results with a single asterisk.

directed perturbations. Our results show that actively holding a handle during perturbations significantly improved balance control compared to normal stance, but that the underlying mechanisms of handle utilisation appear to differ with the location of the handle.

Together with increased perturbation magnitudes in all poses (see Figure 3A), the average CoM_{max} displacement also increased. Displacement of the CoM_{max} was the highest in *normal* pose where, after perturbation magnitude M4, subjects could no longer effectively balance and had to make a step for recovery.

On the other hand, when subjects were holding a handle or were standing in the *step* pose, they easily controlled the CoM_{max} displacement and thus balanced effectively irrespective of the perturbation magnitude. We found no change in the CoM_{max} displacements with location of the handle, indicating that all of the handles used sufficiently assisted balance. This finding is in line with Komisar, Nirmalanathan, and Novak (2018) who found no effects of handrail height on COM control in anterio–posterior direction and with the findings of Babič et al. (2014), who analysed CoP displacements while subjects were holding a handle at four different location. In contrast, Sarraf, Marigold, and Robinovitch (2014) who emulated moving environments found that peak CoP displacement was greater when subjects were grasping a handrail overhead when compared to a shoulder-height position. A possible explanation for this discrepancy might lie in the methodological differences. Namely, Sarraf, Marigold, and Robinovitch (2014) used an experimental setup where the handle was moving together with the movable platform which they used for perturbations, while Babič et al. (2014) and we were using a setup where the handles were stationary. Furthermore, in their study Sarraf, Marigold, and Robinovitch applied perturbations to the platform and analysed the CoP, whereas in our study we applied them to the CoM and focussed on CoM analyses.

Interestingly our results show a significantly lower CoM_{max} displacement in the *step* pose, compared to all other poses at a low perturbation magnitude (M2) while at higher perturbation magnitudes (M4−M8), the CoM_{max} displacement was similar between the *step* pose and the three 'handle' poses. This was a surprising result since one would expect the handles to always have a greater influence on balance control. This counterintuitive finding could be due to the increased size of the base of support which has been reported to reduce the distance and maximal displacement of CoP in quiet standing (Sarabon et al. 2010) and thus reducing the velocity of the CoM which directly affects balance (Hof, Gazendam, and Sinke 2005). Apparently, in both the *step* and 'handle' poses the increased base of support was sufficient for effective balancing. Additionally, small differences in CoM_{max} displacements could be due to the CoM trajectory being controlled as the task-level variable in postural control (Scholz and Schöner 1999; Ting 2007). On the other hand, the base of support during step pose (which is similar to double support phase in gait) was between the subject's feet and during the 'handle' poses it was between the subject's feet and the pole to which the handles were attached. This implies a difference in the angle between the subject's CoM and CoP, therefore, the analysis of the angular motion of CoM around the base of support could provide with additional insights into differential effects of posture on standing stability. Namely, Lee and Chou (2006) found that the CoM–CoP inclination angle identified elderly people with imbalance during gait. In either case, the surprising finding of no additional benefit from using the handles warrants further investigation into the underlying mechanisms.

We investigated forces exerted onto the handles, to check whether all handles are used in a similar manner. Resultant forces exerted onto the handles were affected by the handle location in M2, M7, and M8. In all cases, the force exerted onto the middle handle was significantly lower than the force exerted onto the high handle. This finding is in a contrast to the one of Komisar, Nirmalanathan, and Novak (2018) who reported a decrease of handrail forces while increasing the height of the handrails. However, this might be due to the differences in the experimental setup since their highest handrail was mounted at 106.5 cm, while our high handle was mounted above the subject's head and the fact that our subjects were already holding to the handle before the start of perturbation. Interestingly, when we analysed the vertical and horizontal components of this force, we found this was mediated by a difference in handle use. Namely, across all perturbation magnitudes, subjects exerted much higher vertical component of the force when using the high handle, compared to the middle handle. This finding is in line with the ones by Beschorner et al. (2018), who reported the highest achievable down pulling forces, when the handle was positioned above the subject's head. In contrast, the horizontal component of the force measured in this study was lowest for the high handle. This probably reflects a different use of the handle, where subjects mostly 'hang' onto the high handle, while they 'push/pull' onto the middle and low handle. These results are in agreement with findings from Kumar (1995) who compared handle forces at different heights during pushing and pulling activities. They found that during standing, male subjects were able to generate on average 22% more pushing force on the handrail which was closer to their centre of mass (1 m from the floor - comparable to our *low handle* pose) when compared with forces on handles located at a high height (1.5 m from the floor - comparable to our *middle handle* pose). Similar results were reported in cart pushing/pulling studies by De Looze et al. (2000) and Al-Eisawi et al. (1999). In dynamic pushing and pulling of a movable cart and stationary bar, De Looze et al. (2000) found that in pushing, the direction of force changed from downward to near-horizontal pushing as the handle height increased. Their highest handle was positioned at 80% of the shoulder height, which corresponds to our middle handle. At that handle height they recorded the highest horizontal forces, which the subjects were able to exert during the pushing experiment. On the other hand, Al-Eisawi et al. (1999) investigated effects of handle height on initial hand forces in cart pushing and pulling and reported the smallest vertical forces recorded at elbow height, which is the height that corresponds to somewhere in between our low and middle handle. Given these findings and that our perturbation was also in horizontal direction (as in cart pushing studies), the horizontal forces measured at the middle handle might counteract the perturbation more effectively.

Limitations

We investigated balance recovery in a convenience sample of healthy, young adults. This group is typically able to recover balance well, hence this might be considered the main limitation of our study. However, our intention was to provide fundamental insights into the use of handles and a baseline for future comparisons. Future work should address handle in fall prone populations, such as older adults and various patient groups. Although healthy young adults are typically able to recover balance, it was recently shown that they fall more often than expected (Heijnen and Rietdyk 2016). Given that, healthy young males represent a high proportion of the workforce at risk of falling, our data might also have applicative value in informing workplace interventions (i.e. positioning the handles for safety at shoulder-height). Another limitation of this study was the choice of perturbation. We used a unidirectional perturbation (pulling subjects forward), to avoid non-planar movements of the subjects. However, in case of multidirectional perturbations, the mechanisms could be described to a greater extent, and this is something to consider in future studies. Furthermore, we controlled the CoP position to minimise feed-forward postural adjustments associated with preparation to predictable perturbations. Even so, monitoring muscle activations is another well adopted measure to monitor anticipatory postural adjustments and should be considered in the future. Lastly, further research should be addressing musculoskeletal injuries related to different poses (handle heights), as this is an important factor to consider when designing and positioning handles to assist balance control.

To conclude, our data confirm that both step stance and handles of any height are equally beneficial for balance recovery in young adults. Additionally, we found that the height of the handle influences the force needed to recover and the way handles are used. Young adults predominantly ‘hang’ onto handles positioned over their heads, while they ‘push/pull’ onto lower positioned handles. Given the ageing society, fall prevention is becoming increasingly important. One of the ways to prevent falling is by the use of handles to assist balance. Our data clearly show that handles are beneficial but also indicate that not all handles are equal. This provides the basis for optimal positioning of assistive handles, after future research investigates whether our findings hold in populations needing balance assistance the most, such as elderly and patients. Their higher reliance on balance support, coupled with limited capacity, might result in exaggeration of the effects we observed in healthy young. If elderly and patients also use handles differently depending on their position, this can inform optimal design of handrails, canes, etc. Additionally, we show that changing handle height is a simple and efficient way to individualise training difficulty during physical therapy.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

The work presented in this paper is supported by the European Community Framework Programme 7 through the CoDyCo project, contract no. 600716.

ORCID

Jernej Čamernik http://orcid.org/0000-0002-8434-098X

References

Al-Eisawi, K. W., C. J. Kerk, J. J. Congleton, A. A. Amendola, O. C. Jenkins, and W. G. Gaines. 1999. “The Effect of Handle Height and Cart Load on the Initial Hand Forces in Cart Pushing and Pulling.” *Ergonomics* 42 (8): 1099–1113. doi:10.1080/001401399185162.

Albertsen, I. M., J. J. Temprado, and E. Berton. 2010. “Effect of Haptic Supplementation on Postural Stabilization: A

Comparison of Fixed and Mobile Support Conditions." *Human Movement Science* 29 (6): 999–1010. doi:10.1016/j.humov.2010.07.013.

Babič, J., T. Petrič, L. Peternel, and N. Šarabon. 2014. "Effects of Supportive Hand Contact on Reactive Postural Control during Support Perturbations." *Gait and Posture* 40 (3): 441–446. doi:10.1016/j.gaitpost.2014.05.012.

Bartlett, R. 2007. *Introduction to Sports Biomechanics: Analysing Human Movement Patterns*. 2nd ed. Abingdon, UK: Routledge.

Bateni, H., and B. E. Maki. 2005. "Assistive Devices for Balance and Mobility: Benefits, Demands, and Adverse Consequences." *Archives of Physical Medicine and Rehabilitation* 86 (1): 134–145. doi:10.1016/j.apmr.2004.04.023.

Beschorner, K. E., G. P. Slota, E. M. Pliner, E. Spaho, and N. J. Seo. 2018. "Effects of Gloves and Pulling Task on Achievable Downward Pull Forces on a Rung." *Human Factors: The Journal of the Human Factors and Ergonomics Society* 60 (2): 191–200. doi:10.1177/0018720817742515.

Cadore, E. L., L. Rodríguez-Mañas, A. Sinclair, and M. Izquierdo. 2013. "Effects of Different Exercise Interventions on Risk of Falls, Gait Ability, and Balance in Physically Frail Older Adults: A Systematic Review." *Rejuvenation Research* 16 (2): 105–114. doi:10.1089/rej.2012.1397.

Čamernik, J., Z. Potocanac, L. Peternel, and J. Babič. 2016. "Holding a Handle for Balance during Continuous Postural Perturbations—Immediate and Transitionary Effects on Whole Body Posture." *Frontiers in Human Neuroscience* 10: 1–8. doi:10.3389/fnhum.2016.00486.

Chang, W.-R., S. Leclercq, T. E. Lockhart, and R. Haslam. 2016. "State of Science: Occupational Slips, Trips and Falls on the Same Level." *Ergonomics* 0139: 1–23. doi:10.1080/00140139.2016.1157214.

Chen, C.-L., S.-Z. Lou, H.-W. Wu, S.-K. Wu, K.-T. Yeung, and F.-C. Su. 2014. "Effects of the Type and Direction of Support Surface Perturbation on Postural Responses." *Journal of Neuroengineering and Rehabilitation* 11 (1): 50. doi:10.1186/1743-0003-11-50.

Choi, M., and M. Hector. 2012. "Effectiveness of Intervention Programs in Preventing Falls: A Systematic Review of Recent 10 Years and Meta-Analysis." *Journal of the American Medical Directors Association* 13 (2): 188.e13–188.e21. doi:10.1016/j.jamda.2011.04.022.

Diener, H. C., F. B. Horak, and L. M. Nashner. 1988. "Influence of Stimulus Parameters on Human Postural Responses." *Journal of Neurophysiology* 59 (6): 1888–1905.

Dusenberry, D. O., H. Simpson, and S. J. Dellorusso. 2009. "Effect of Handrail Shape on Graspability." *Applied Ergonomics* 40 (4): 657–669. doi:10.1016/j.apergo.2008.05.006.

EuroSafe 2013. *"Injuries in the European Union, Report on Injury Statistics 2008–2010."* Amsterdam: European Association for Injury Prevention and Safety Promotion. www.eurosafe.eu.com

Forero, J., and J. E. Misiaszek. 2013. "The Contribution of Light Touch Sensory Cues to Corrective Reactions during Treadmill Locomotion." *Experimental Brain Research* 226 (4): 575–584. doi:10.1007/s00221-013-3470-z.

Garg, A., T. Waters, J. Kapellusch, and W. Karwowski. 2014. "Psychophysical Basis for Maximum Pushing and Pulling Forces: A Review and Recommendations." *International Journal of Industrial Ergonomics* 44 (2): 281–291. doi:10.1016/j.ergon.2012.09.005.

Hamill, J., and K. Knutzen. 2008. *Biomechanical Basis of Human Movement*. Lippincott: Williams and Wilkins.

Heijnen, M. J. H., and S. Rietdyk. 2016. "Falls in Young Adults: Perceived Causes and Environmental Factors Assessed with a Daily Online Survey." *Human Movement Science* 46: 86–95. doi:10.1016/j.humov.2015.12.007.

Hof, A. L., M. G. J. Gazendam, and W. E. Sinke. 2005. "The Condition for Dynamic Stability." *Journal of Biomechanics* 38 (1): 1–8. doi:10.1016/j.jbiomech.2004.03.025.

Hoozemans, M. J. M., P. P. F. M. Kuijer, I. Kingma, J. H. van Dieën, W. H. K. de Vries, L. H. V. van der Woude, D. J. Veeger, (H.E.J.), A. J. van der Beek, and M. H. W. Frings-Dresen. 2004. "Mechanical Loading of the Low Back and Shoulders during Pushing and Pulling Activities." *Ergonomics* 47 (1): 1–18. doi:10.1080/00140130310001593577.

IJmker, T., C. J. Lamoth, H. Houdijk, M. Tolsma, L. H. V. van der Woude, A. Daffertshofer, and P. J. Beek. 2015. "Effects of Handrail Hold and Light Touch on Energetics, Step Parameters, and Neuromuscular Activity during Walking after Stroke." *Journal of NeuroEngineering and Rehabilitation* 12 (1): 70. doi:10.1186/s12984-015-0051-3.

Jeka, J. J., and J. R. Lackner. 1994. "Fingertip Contact Influences Human Postural Control." *Experimental Brain Research* 100 (3): 495–502. doi:10.1007/BF00229188.

Komisar, V., K. Nirmalanathan, and A. C. Novak. 2018. "Influence of Handrail Height and Fall Direction on Center of Mass Control and the Physical Demands of Reach-to-Grasp Balance Recovery Reactions." *Gait and Posture* 60: 209–216. doi:10.1016/j.gaitpost.2017.12.009.

Kouzaki, M., and K. Masani. 2008. "Reduced Postural Sway during Quiet Standing by Light Touch Is Due to Finger Tactile Feedback but Not Mechanical Support." *Experimental Brain Research* 188 (1): 153–158. doi:10.1007/s00221-008-1426-5.

Kumar, S. 1995. "Upper Body Push-Pull Strength of Normal Young Adults in Sagittal Plane at Three Heights." *International Journal of Industrial Ergonomics* 15 (6): 427–436. doi:10.1016/0169-8141(94)00062-8.

Lee, H.-J., and L.-S. Chou. 2006. "Detection of Gait Instability Using the Center of Mass and Center of Pressure Inclination Angles." *Archives of Physical Medicine and Rehabilitation* 87 (4): 569–575. doi:10.1016/j.apmr.2005.11.033.

Lett, K. K., and S. M. McGill. 2006. "Pushing and Pulling: Personal Mechanics Influence Spine Loads." *Ergonomics* 49 (9): 895–908. doi:10.1080/00140130600665869.

Leva, P. D. 1996. "Adjustments to Zatsiorsky-Seluyanov's Segment Inertia Parameters." *Journal of Biomechanics* 29 (9):1223–30. doi:10.1016/0021-9290(95)00178-6.

Looze, M. P. D., K. Van Greuningen, J. Rebel, I. Kingma, and P. P. F. M. Kuijer. 2000. "Force Direction and Physical Load in Dynamic Pushing and Pulling." *Ergonomics* 43 (3): 377–390. doi:10.1080/001401300184477.

Maki, B. E., W. E. McIlroy, and G. R. Fernie. 2003. "Change-in-Support Reactions for Balance Recovery." *IEEE Engineering in Medicine and Biology Magazine* 22 (2): 20–26. doi: 10.1109/MEMB.2003.1195691.

Martinelli, A. R., D. B. Coelho, F. H. Magalhães, A. F. Kohn, and L. A. Teixeira. 2015. "Light Touch Modulates Balance Recovery following Perturbation: From Fast Response to Stance Restabilization." *Experimental Brain Research* 233 (5): 1399–1408. doi:10.1007/s00221-015-4214-z.

Noé, F., F. Quaine, and L. Martin. 2003. "Mechanical Effect of Additional Supports in a Rocking on Heels Movement." *Gait and Posture* 18 (1): 78–84. doi:10.1016/S0966-6362(02)00163-7.

Park, S., F. B. Horak, and A. D. Kuo. 2004. "Postural Feedback Responses Scale with Biomechanical Constraints in Human Standing." *Experimental Brain Research* 154 (4): 417–427. doi:10.1007/s00221-003-1674-3.

Peternel, L. 2015. *Accelerated Robotic Learning for Interaction With Environment and Human Based on Sensory-Motor Learning*. Ljubljana: University of Ljubljana.

Peternel, L., and J. Babič. 2013. "Learning of Compliant Human–Robot Interaction Using Full-Body Haptic Interface." *Advanced Robotics* 27 (13): 1003–1012. doi: 10.1080/01691864.2013.808305.

Santos, M. J., N. Kanekar, and A. S. Aruin. 2010. "The Role of Anticipatory Postural Adjustments in Compensatory Control of Posture: 2. Biomechanical Analysis." *Journal of Electromyography and Kinesiology : Official Journal of the International Society of Electrophysiological Kinesiology* 20 (3): 398–405. doi:10.1016/j.jelekin.2010.01.002.

Sarabon, N., J. Rosker, S. Loefler, and H. Kern. 2010. "Sensitivity of Body Sway Parameters during Quiet Standing to Manipulation of Support Surface Size." *Journal of Sports Science and Medicine* 9 (3): 431–438. http://www.ncbi.nlm.nih.gov/pubmed/24149637.

Sarraf, T. A., D. S. Marigold, and S. N. Robinovitch. 2014. "Maintaining Standing Balance by Handrail Grasping." *Gait and Posture* 39 (1): 258–264. doi:10.1016/j.gaitpost.2013.07.117.

Scholz, J. P., and G. Schöner. 1999. "The Uncontrolled Manifold Concept: Identifying Control Variables for a Functional Task." *Experimental Brain Research* 126 (3): 289–306. doi:10.1007/s002210050738.

Ting, L. H. 2007. "Dimensional Reduction in Sensorimotor Systems: A Framework for Understanding Muscle Coordination of Posture." *Progress in Brain Research* 165: 299–321. doi:10.1016/S0079-6123(06)65019-X.

Winter, D. A. 1995. "Human Balance and Posture Control during Standing and Walking." *Gait and Posture* 3: 193–214. doi:10.1016/0966-6362(96)82849-9.

Winter, D. A. 2009. *Biomechanics and Motor Control of Human Movement*. Hoboken, NJ, USA: John Wiley and Sons, Inc. doi:10.1002/9780470549148.

6.1 Additional Analyses and Conclusions

In the second step of the investigation of the effect of posture and handles, we performed the inverse dynamics and analyzed the joint torques during the first 2 seconds after the start of perturbations - the recovery response as defined in the above paper.

6.1.1 Total effort Analysis - Total Joint Torque Analysis

As a first step, we derive the equations for the total joint torque ($\boldsymbol{\tau}_{total}$, Eq. 6.3) as a sum of average joint torques averaged over the recovery response. Let $\mathbf{q}$ denote the vector of generalized coordinates of the model which is described in detail in the paper above. Note that depending on the posture, the model has either 3 or 5 DOF. To calculate the torque of individual joints we used the following equation:

$$\boldsymbol{\tau} = \mathbf{H}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{C}(\mathbf{q}, \dot{\mathbf{q}}) + \mathbf{G}(\mathbf{q}) - \mathbf{J}_c^T(\mathbf{q})\mathbf{f}_c, \tag{6.1}$$

where $\boldsymbol{\tau}$ is the vector of joint torques, $\mathbf{H}$ is the joint-space inertia matrix, $\ddot{\mathbf{q}}$ represents the joint accelerations obtained from marker positions, $\mathbf{C}$ is the vector of Coriolis and centrifugal forces, $\mathbf{G}$ is the vector of gravity force, $\mathbf{J}_c$ is the Jacobian of the constraints and $\mathbf{f}_c$ are the contact forces measured by the force/torque sensors on the handle.

Next, we calculated the average of torque produced at each joint across 10 trials by

$$\bar{\boldsymbol{\tau}} = \frac{1}{2}\int_0^2 \Big(\frac{1}{10}\sum_{i=1}^{10} \boldsymbol{\tau}_i(t)\Big)dt, \tag{6.2}$$

where $\bar{\boldsymbol{\tau}}$ is the average joint torque of the recovery response, and $\boldsymbol{\tau}_i(t)$ is the torque over time of the i-th trial.

Last, we calculated the total joint torque ($\boldsymbol{\tau}_{total}$) by summing the values that we obtained in the previous step (Eq. 6.3). Since maximal torques of the arm joints vary with arm configuration, we normalized shoulder and elbow torques with respect to the maximal torque at the given joint. The normalization was performed by multiplying the calculated torques with a factor determined from the regression curves (Anderson et al., 2007; Bober et al., 2002; Gandevia et al., 1998; Moraux et al., 2013). For illustration, if the maximal elbow torque is 60 Nm at 90° flexion and the subject exerted 40 Nm at 90° elbow flexion, then the multiplication factor was 1.0 and the normalized torque remained 40 Nm. Moreover, if the subject exerted 40 Nm at 30° elbow flexion where the maximal elbow torque is also 40 Nm, then the multiplication factor was 1.5 which gives the normalized torque of 60 Nm. In effect, the normalization linearizes the non-linear relationship between the maximal achievable joint torque and the joint angle at which the torque is generated and at the same time preserves the units of Nm.

$$\boldsymbol{\tau}_{total} = \sum_{j=1}^{n_j} w_i \bar{\boldsymbol{\tau}}_j, \tag{6.3}$$

where w represents the weights to normalize the joint torques of the arm joints i, $\bar{\boldsymbol{\tau}}_j$ is the average torque produced at joint j, and n_j is the number of joints in the model.

Results

The results of the total joint torque are shown in Figure 6.2. We found a main effect of pose at perturbation magnitudes M4 ($F(1.32, 9.24) = 6.18$, $p = .028$, $\eta_p^2 = .46$), M6 ($F(3, 27) = 3.88$, $p = .02$, $\eta_p^2 = .30$), and M7 ($F(3, 27) = 8.61\, p < .001$, $\eta_p^2 = .49$). At M4 and M6 none of the post hoc t-tests were statistically significant (Figure 6.2B), but at M7 $\boldsymbol{\tau}_{total}$ was significantly larger in the step pose compared to low ($t(9) = 3.79$, $p = .025$) and middle handle ($t(9) = 4.29$, $p = .012$) pose.

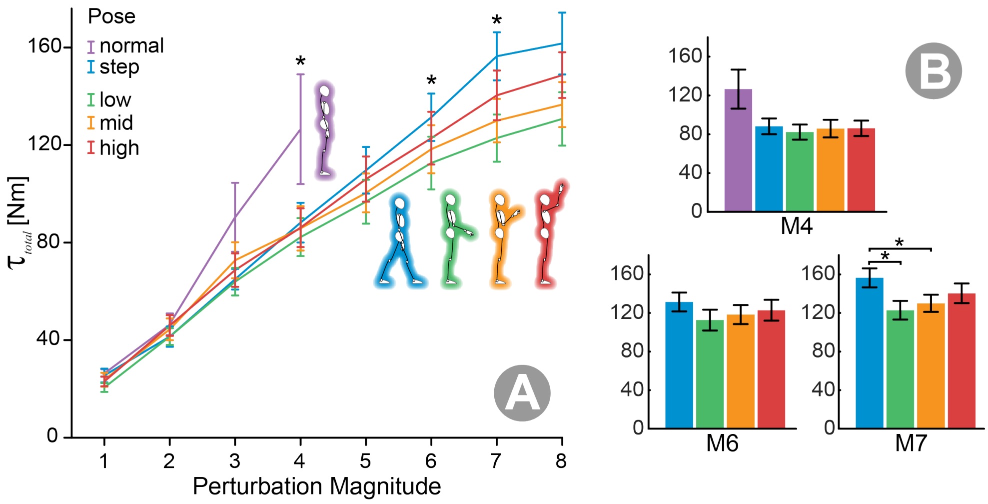

Figure 6.2: Sum of torques produced in the joints of the body during perturbations. **A)** Total joint torques ($\boldsymbol{\tau}_{total}$) across poses and perturbation magnitudes. **B)** At perturbation magnitudes of M4, M6, and M7, pose had a statistically significant effect on $\boldsymbol{\tau}_{total}$, but no post-hoc results were significant for M4 and M6. Error bars indicate standard errors of the mean. Statistically significant effects of pose are marked with an asterisk.

Discussion

The results of the total joint torque calculation show a significant effect of the handle at higher perturbation magnitudes. In those positions, the total joint torques were significantly higher for the step stance pose, indicating an implicit benefit of using the handle. Namely, while both step and handle poses sufficiently counteracted the perturbations, a higher torque, as evident in the step stance pose, could be detrimental. Lower loading of the joints can be linked with lower effort needed to counteract perturbations. For example, when stroke patients use a handrail or a cane during treadmill walking, the energy cost of walking is reduced (IJmker et al., 2015). Additionally, high torques could be problematic for populations at risk of falling, who presumably need support to balance the most (Pijnappels et al., 2008). Therefore, the patients with stroke and with higher risk of falling can benefit from using a handrail or cane for support as their metabolic effort for balance control is reduced.

Chapter 7

Study III - Learning Hand Contacts in Unpredictable Environment

In this part, I present the paper "Probabilistic Movement Models Show that Postural Control Precedes and Predicts Volitional Motor Control" by Elmar Rueckert, Jernej Čamernik, Jan Peters, and Jan Babič (Rueckert et al., 2016). The paper was published in a peer-reviewed journal *Scientific Reports* in 2016.

This study was conducted together with our colleagues from Technische Universität Darmstadt who performed the computational part of the analysis and led the writing of the paper. I participated in designing the experiment, performed the recordings of the experimental task, processed the data statistically, and wrote parts of the methods, results and discussions of the manuscript. In particular, I focused on analyzing and describing the task-dependent movement of the participants. The presented manuscript addresses hypotheses H6, H7, and H8.

Figure 7.1: Study III. Learning hand contacts in an unpredictable environment.

In this study we were questioning whether certain motor actions depend on the task, or some other higher motor priority. We put the participants on top of a movable platform and asked them to reach toward a target which appeared on the screen in front of them. During their reaching action we would perturb their balance by quickly moving the platform laterally. The movement of the platform depended on the movement of the participants' non-reaching arm, which they had to use to lean on the table in front of them. The screen with the targets was positioned in front of the participants at such a distance that it was impossible for them to touch the screen without leaning on the table. Participants received a monetary award for each successful reach, to stimulate them to complete the task as fast and as accurate as possible. With this study we show that in cases with perturbation to standing balance while completing a reaching task, the participants chose the supportive contact location based on the position of the target and not only based on the direction and the magnitude of the perturbation.

www.nature.com/scientificreports

SCIENTIFIC REPORTS

OPEN

Probabilistic Movement Models Show that Postural Control Precedes and Predicts Volitional Motor Control

Received: 22 February 2016
Accepted: 02 June 2016
Published: 22 June 2016

Elmar Rueckert[1], Jernej Čamernik[2], Jan Peters[1,3] & Jan Babič[2]

Human motor skill learning is driven by the necessity to adapt to new situations. While supportive contacts are essential for many tasks, little is known about their impact on motor learning. To study the effect of contacts an innovative full-body experimental paradigm was established. The task of the subjects was to reach for a distant target while postural stability could only be maintained by establishing an additional supportive hand contact. To examine adaptation, non-trivial postural perturbations of the subjects' support base were systematically introduced. A novel probabilistic trajectory model approach was employed to analyze the correlation between the motions of both arms and the trunk. We found that subjects adapted to the perturbations by establishing target dependent hand contacts. Moreover, we found that the trunk motion adapted significantly faster than the motion of the arms. However, the most striking finding was that observations of the initial phase of the left arm or trunk motion (100–400 ms) were sufficient to faithfully predict the complete movement of the right arm. Overall, our results suggest that the goal-directed arm movements determine the supportive arm motions and that the motion of heavy body parts adapts faster than the light arms.

Most of our every day motor skills involve strict control of postural stability in parallel to the execution of the primary motor task. A great deal of these tasks also require additional supportive hand contacts beside the feet that are in contact with the ground. An example task is the reaching for a glass on the highest kitchen shelf when we typically have to use the other hand to support the body by leaning on the kitchen counter. To successfully perform such a reaching motion and to simultaneously ensure the postural stability, the motion of the body and both arms have to be perfectly synchronized and coordinated. Till now it is unclear how this coordination develops or adapts during skill learning.

In general, supportive hand contacts increase the stability during balancing and complement humans' sophisticated motor abilities. Already *light touch* fingertip contacts provide sufficient feedback to enhance postural stability[1–4]. Notable mechanical support is gained by holding onto a handrail, which enables humans to generate contact forces to better counteract perturbations[5,6]. A notable exception to these static contacts studies are fall prevention experiments. It was shown that humans prefer reach-to-grasp strategies to rapid stepping with age[7,8]. All these works underline the importance of contacts but cannot answer the posed question of how the coordination of supportive contacts and primary tasks develops or adapts.

Like the ability to balance, the utilization of supportive hand contacts has to be learned[9,10]. In the early childhood, learning of postural balance is tightly intertwined with supportive hand motion. At eight months infants already learn how to balance on hands and knees during crawling; at ten months they start walking with support from hanging on a table or a couch; and at about one year infants already know how to balance on two feet and walk independently across the room[11,12]. While it is known that people develop optimal motor control strategies both for postural stability and manipulation[13–16], it is unclear how these two distinctive motor tasks relate to each other and how their relationship affect the *learning* of novel motor skills or when re-establishing motor abilities in novel environments. A thorough understanding of these issues is of particular interest for the field of

[1]Intelligent Autonomous Systems Lab, Technische Universität Darmstadt, 64289, Germany. [2]Department for Automation, Biocybernetics and Robotics, Jožef Stefan Institute, Ljubljana, SI-1000, Slovenia. [3]Robot Learning Group, Max-Planck Institute for Intelligent Systems, Tuebingen, 72076, Germany. Correspondence and requests for materials should be addressed to E.R. (email: rueckert@ias.tu-darmstadt.de)

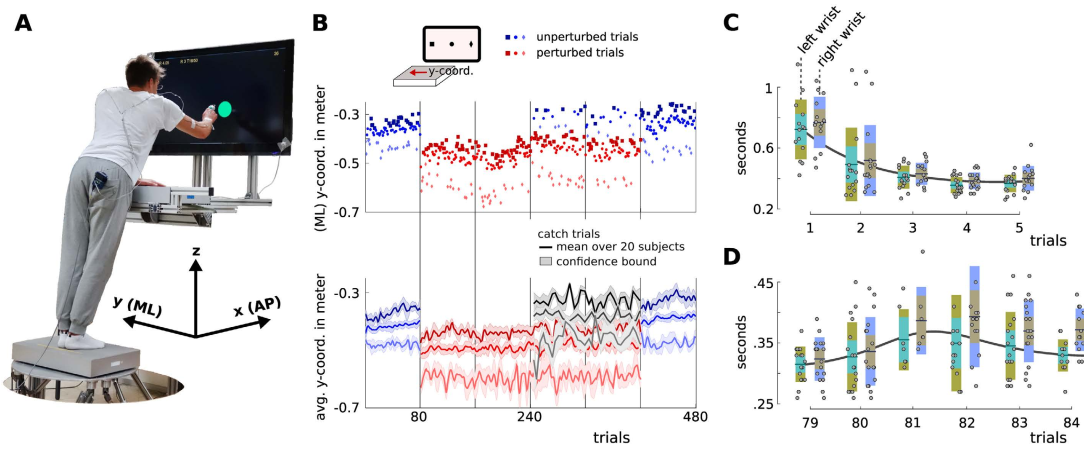

Figure 1. Experiment, target dependent contacts and synchronized arm motions. (**A**) Experimental setting. (**B**) The top row shows contact locations for a single representative subject and the bottom row shows the mean and the confidence bound over all 20 participants. The first 80 trials and the last 80 trials are unperturbed sessions. Catch trials were initiated during trials 240 to 400 and are denoted by the black lines in (**B**). (**C**) Illustration of the movement onsets of the wrists for the first five trials. (**D**) Movement onsets when transitioning to the perturbed session (trials 79 and 80 are still unperturbed).

rehabilitation, where correlations between the primary motor tasks and the underlying supportive motor actions could be exploited in progress monitoring and novel pre-tests of motor dysfunctions[17]. Such pre-tests promise to be less sensitive to factors like stress or sleep deprivation. However, to approach that goal we need to identify such correlations if existing.

Many of previous studies either focus on control of postural stability[18–21], study adaptation of arm reaching in confined lab environments[22–29], or study situations when postural stability and arm reaching are only indirectly related[2,6,30–32]. In our study we propose an innovative full-body experimental paradigm (Fig. 1A) that extends current experimental methods to a more ecological setting where postural stability and manipulation skills are tightly interrelated and interdependent to each other.

We hypothesize that the arm reaching motions determine the supportive hand contact strategies and that both motor tasks adapt during training. In particular, we assume that these two motor tasks are correlated which is reflected in synchronized motor executions and a significant correlation between target locations and supportive hand contacts. To effectively elaborate on these hypotheses we designed an experimental paradigm where we asked 20 healthy subjects to reach with their right arm for a target displayed on a screen while using their left arm to maintain postural stability by leaning on a table in front of them. Non-trivial postural perturbations of the subjects' support base were systematically introduced to examine adaptation and a novel probabilistic trajectory model approach was employed to analyze the correlations between the supportive arm motion and the motion of the arm to reach for the distant target.

Results

The experimental setup involved a 6 degrees-of-freedom Stewart platform on which human subjects stood and performed the arm-reaching movements. The sum of the anteroposterior components of the left and right hand displacements during the reaching was used to generate (or not, depending on the condition) perturbations in the mediolateral direction; the forward motion of hands caused a leftward displacement of the platform as depicted in Supplementary Figure S5. The subjects performed arm-reaching movements (called trials) in three types of conditions: perturbed (P), unperturbed (UP) and catch (C). In the unperturbed condition, the platform remained still while in the perturbed condition the platform induced perturbations. The catch condition was almost the same as the perturbed case. However, for some randomly selected squat-to-stand trials, the perturbation was turned off without the knowledge of the subjects.

To initiate the reaching movements of the subject, a target was presented on the screen at one out of three possible locations. The target appeared randomly at the left, the center or the right side on a horizontal bar at a random interval of one to three seconds after the previous trial. We motivated the subjects through a monetary reward that was proportional to the time needed to reach for the target. This reward was displayed after the target was reached. The number zero was displayed if the trial time exceeded two seconds. In our results, we focused on the y-coordinate of the contact locations as it corresponded to the direction of the translational perturbation. This is indicated by the arrow labeled by *y* in Fig. 1.

Subjects chose task dependent contact locations. Analysis of variance (ANOVA) showed a significant effect in the left hand contact location between sessions when the subjects were perturbed and unperturbed during reaching toward targets displayed on screen, $F(1,19) = 113.632$, $p < 0.001$, $\eta_p^2 = 0.857$.

The post-hoc analysis showed that the left hand contact location was significantly different between reaching toward all three distinctly positioned targets in perturbed, $t(19) = 6.85$to11.58, $p < 0.001$, and in unperturbed trials, $t(19) = 6.78–12.21$, $p < 0.001$.

There were also significant differences between means of contact locations for each target between perturbed and unperturbed trials (target one, $t(19) = 10.07$, $p < 0.001$, $d = 2.01$; target two, $t(19) = 10.22$, $p < 0.001$, $d = 2.38$; target three, $t(19) = 11.51$, $p < 0.001$, $d = 3.25$), see Supplementary Figure S1.

Further, we analyzed individual sessions of unperturbed, perturbed and catch trials. We compared them with each other to determine whether there are any differences between contact locations and for which target positions they apply.

The first 80 trials out of 480 training trials were unperturbed. We found that there was a significant effect of target locations on the supportive contact locations in these first 80 trials, $F(1.11,21.14) = 104.61$, $p < 0.001$, $\eta_p^2 = 0.846$, see Fig. 1B. Post-hoc tests using the Bonferroni correction[33] revealed that there was significant difference in chosen supportive contact location between all three targets ($t(19) = 6.92$ to11.66, $p < 0.001$).

After the initial training phase, all subjects had to adapt to translational perturbations of the support base. Note that to compare the contact locations and the marker trajectories of perturbed and unperturbed trials, we corrected all sensor readings by the motion of the moving base. We refer to this correction as the *subject frame*. Later we will introduce *shoulder frames* to separate adaptation in the trunk and in the arms. Again, we found that there was a significant effect of the target location on the contact location over all perturbed trials, $F(1.33,25.27) = 116.39$, $p < 0.001$, $\eta_p^2 = 0.86$, see trials 81to240 in Fig. 1B. Post-hoc tests showed a significant difference in the chosen supportive contact location among all three targets ($t(19) = 6.85$to11.58, $p < 0.001$).

Consolidation of the postural control strategies was tested through randomly initiating unperturbed trials (catch trials). In the last phase of perturbed trials (trials 320 to 400), where adaptation can be assumed to be converged, the perturbation was deactivated for 30 random catch trials. When comparing initial training trials with catch trials post-hoc tests showed no significant difference for all three targets, target one ($t(18) = -1.99$, $p = 0.062$, $d = -0.74$), target two ($t(18) = -3.28$, $p = 0.004$, $d = -1$) and target three ($t(18) = -1.03$, $p = 0.315$, $d = -0.39$). Comparison of perturbed and catch trials showed significant differences in contact locations for all three target locations ($t(18) = -11.24$ to -12.97, $p < 0.001$, $d = -2.44$ to -1.93). For the washout, no significant difference was observed between contact locations for individual targets when comparing catch and final trials 401 to 480 ($t(18) = -1.59$ to 1.28, $p = 0.13$ to 0.22, $d = -1$ to -0.39).

In summary, we found that subjects chose distinct contact locations dependent on the target location on the screen. Catch trial tests revealed that the participants learned specialized control strategies for unperturbed and perturbed conditions. However, whether the distinct contacts result from trunk or left arm adaptations can not be answered from looking at contact locations only. That requires a more detailed analysis of the temporal profiles for which we will introduce a probabilistic trajectory model.

Supportive and goal-directed movements synchronize. Decreasing reaction times are an indicator for adaption of a pre-processing phase[34]. In line with this, we found that the movement onsets of the left and the right wrist synchronize within 3–5 trials in the unperturbed training phase (left wrist: $323 \pm 44(SD)$ ms, right wrist: 335 ± 46 ms) and when the perturbations are experienced for the first time (left wrist: 333 ± 48 ms, right wrist: 342 ± 49 ms). The movement onsets of all subjects in the first five unperturbed trials are shown in Fig. 1C. In Fig. 1D onsets of the last two unperturbed trials (79 and 80) and of the first four perturbed trials are shown.

No significant lead or lag of the supportive left arm motion over the right arm reaching movements was found in the data. This is shown in an illustration of movement onsets of all subjects in Supplementary Figure S2.

Modeling joint distributions over limb trajectories. Reaction times are sensitive to signal noise, need to be verified through visual inspection and allow only for a limited view on the functional mechanisms during skill learning. As an alternative to reaction time studies, we propose to analyze adaptation on a trajectory basis. For that we developed a probabilistic trajectory model (PTM) that encodes a joint distribution over multiple limb trajectories and over multiple coordinates like x, y, z components of three-dimensional marker data. Here, we give a brief summary of PTMs and for a precise mathematical definition we refer the reader to the Methods section.

The main feature of the model is that it captures the correlations between individual input dimensions. The model builds on a linear function approximator using radial basis functions with fixed means and variances. The amplitudes of the basis functions are scaled by learnable feature weights. Now both, the basis functions and the weights, form a linear generative model of a trajectory, i.e., $\boldsymbol{\tau} = \boldsymbol{\Phi w}$ (with the trajectory $\boldsymbol{\tau} = [y_1, y_2, ..., y_t]$, time-varying observations y_t, basis functions in $\boldsymbol{\Phi}$ and feature weights $\boldsymbol{w}$). Figure 2A shows the encoding of a single trajectory using classical radial basis functions. Note that the model is linear in the feature space, however it can capture non-linear dependencies in the trajectory space (through non-linear basis functions). The model's complexity is controlled through the number of basis functions. For the reaching experiments ten Gaussian distributions per dimension were found to be sufficient, see Supplementary Figure S3.

The feature vector $\boldsymbol{w}$ can be learned in the most simple case through standard linear regression or in more sophisticated models through one of the many existing variational inference methods[35].

A PTM encodes multiple input dimensions through a concatenated feature vector (a two-dimensional example is illustrated in Fig. 2B). This concatenated feature vector scales the contribution of an *extended* basis function matrix. Thus, the only difference to standard radial basis functions is a clever arrangement of basis functions and feature vectors to represent multiple input dimensions in one model.

To model a distribution over trajectories the mean and the covariance over multiple trials are computed from the inferred feature vectors (e.g., through linear regression). An example is shown in the center panel in Fig. 2B. The mean and the covariance encode a distribution over trajectories in the feature space and through

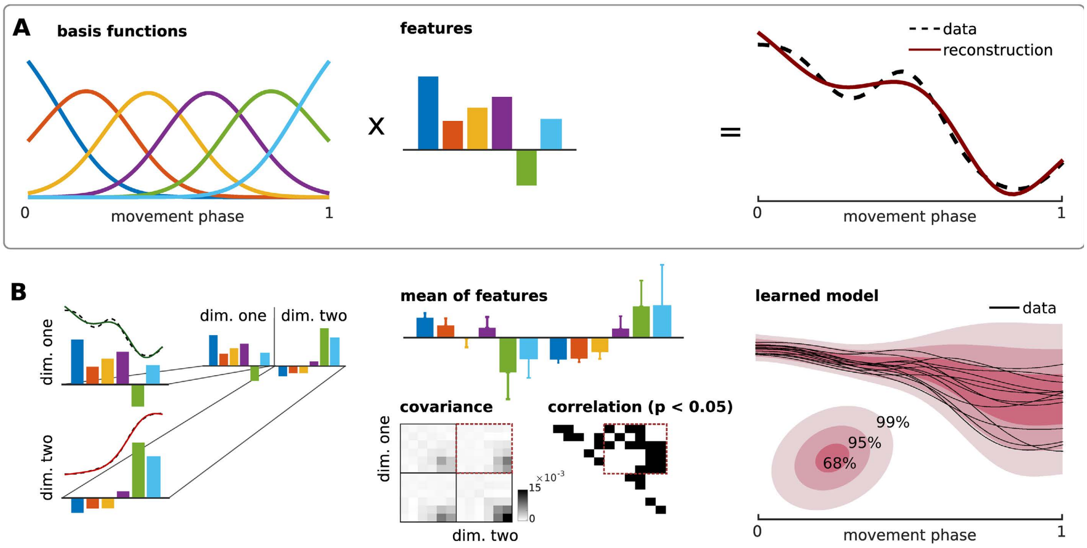

Figure 2. Probabilistic model of trajectories, from a feature space (left column) to trajectories (right column). (**A**) Generative model. Equally spaced (radial) basis functions are amplitude scaled by a feature vector to approximate a one-dimensional trajectory. A movement phase substitutes time to model trajectories of different lengths. In the inverse direction, feature vectors are computed given the input trajectories using, e.g., standard linear regression techniques or iterative variational approaches like expectation-maximization. (**B**) Learning the correlations between multi-dimensional input trajectories. Feature vectors computed from multiple input trajectories are concatenated and the mean and the covariance are computed from multiple trials (or subjects). The learned correlation in the center panel is the key features for computing predictions from partial observations. The distribution over trajectories is shown in the right panel.

the relationship $\boldsymbol{\tau} = \boldsymbol{\Phi w}$, the distribution can be mapped (back) to the time domain. It is worth mentioning that a PTM can represent trajectories of varying lengths by substituting time by a movement phase. The last panel in Fig. 2B shows an illustrative example of a trajectory distribution and the corresponding input trajectories using a movement phase.

To conclude, the advantage of the presented time-series model is that it can be formulated as a generative probabilistic model for which many learning algorithms and similarity measures exist. The model can encode the signal variation and can be used to compute operations like predictions, model comparisons or trial likelihoods. These operations are explored in the present study on postural control with supportive contacts.

Supportive contacts predict goal-directed movements. The learned and represented correlation of multiple limb trajectories can be exploited in computing predictions. This feature is used here to investigate if left wrist trajectories can predict the right wrist reaching trajectories. To separate the contributions of the trunk and the left arm, we transformed the wrist marker trajectories to a *shoulder frame*. For that the wrist positions in the subject's coordinate frame were additionally corrected by the time-varying shoulder marker positions.

For increasing observation horizons up to 400 ms, exemplary predictions of the y-coordinate of the arm trajectories are shown in Fig. 3. For these examples the target prediction error was below 5 cm when observing only the first 300 ms of the left wrist motion. Note that the distance between the two exterior targets on the screen was 80 cm which defines the maximum error. Averaged over all subjects, the prediction error was 12.3 ± 9.8 cm for an observation horizon of 300 ms. When observing the complete trial the error was 6.4 ± 5.5 cm. Per subject errors are shown in Fig. 4. Note that to obtain these results we used 18 trials for training (trials 170–320) and 18 trials for testing (trials 321–400).

In summary, the PTM was used to train models of trajectory distributions that could predict the goal-directed reaching motions from observing solely the supportive left wrist movements. Accurate predictions at an early execution phase (i.e., the first 300 ms), where the effect of visual feedback is negligible, indicate that supportive contacts are at least partially pre-processed.

Trunk adaptation precedes arm adaptation. To investigate adaptation of the supportive contacts we fitted Gaussian distributions to the contact locations recorded in the three phases of learning—the unperturbed training phase, the phase of perturbed trials and the final washout phase. Histograms over the contact locations in the shoulder frame are shown in Fig. 5A, where *UP1* denotes the initial training phase, *P* the perturbed trials and *UP6* the sixth session that was the washout phase. During the perturbed trials of the experiment, we found a shift of contacts to the right, or in other words, the left hand is moving closer to the center of the table in Fig. 1A. Note that the particular values in meters differ in Fig. 1A and in Fig. 5A. In the later the contacts are plotted in the shoulder frame.

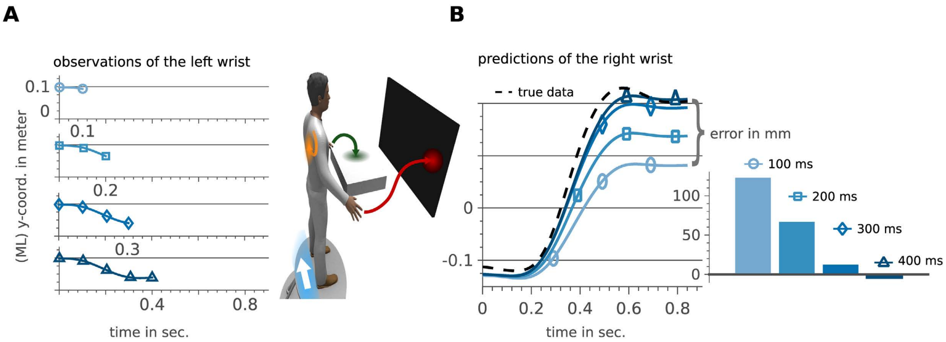

Figure 3. Supportive contacts predict goal-directed movements. (**A**) Partial observations of the left wrist predict right wrist future states in (**B**). For increasing observation horizons of the y-coordinate of the left wrist, the predicted right wrist trajectories converge to the true trajectory (illustrated as dashed line in (**B**). The final Euclidean error (to the true reached target on the screen) of less than 2 cm after 300 ms is 40 times smaller than the distance between the two outer targets (that is 80 cm).

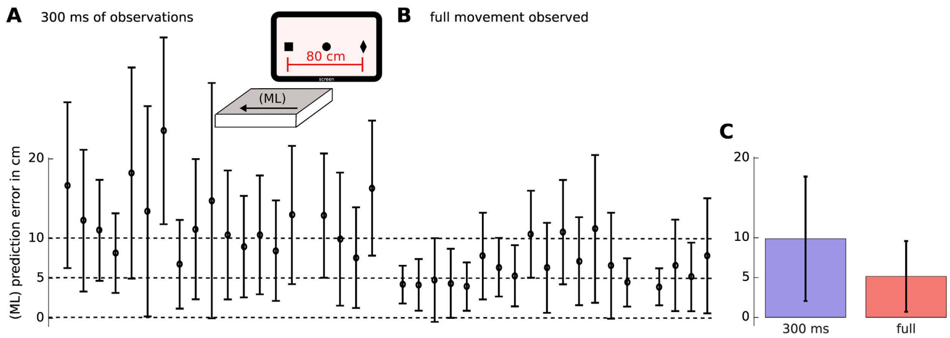

Figure 4. Target prediction error over all subjects. (**A**) Averaged prediction error for 19 subjects when observing 300 ms of the left wrist marker trajectories in perturbed trials. (**B**) The prediction error when observing the whole left wrist motion. Illustrated are the mean and the standard deviation over 18 test trials (six per target). One subject of 20 was excluded as less than six trials per target were recorded. Note that the maximum error or the distance between the two outer targets in the screen is 80 cm. (**C**) Mean and standard deviation over all subjects.

The fitted Gaussian distributions in Fig. 5A indicate a symmetric shift of *UP6* with respect to *P*, which implies a possible after effect of perturbations. On the other hand, the reduced variance of UP6 with respect to UP1 and the overlap of both distributions suggest the increased precision of supportive hand contact at UP6 compared to UP1. To better understand this, we performed additional statistical analysis where we compared contact locations of UP1 and UP6. We found out that for two out of three targets, the contact location at UP1 was statistically different to the locations at UP6 ($t(19) = -4.29$ to -4.16, $p < 0.001$, $d = -1.01$ to -0.75) while for one target (right most) there was no difference between the contact locations ($t(19) = 0.07$, $p = 0.948$, $d = 0.01$).

To compare the similarities between the fitted Gaussian distributions for the three phases of learning, we used a Kullback-Leibler divergence (KL) distance measure. In line with the observation of the shift of contacts we found strong similarities between the unperturbed phases, i.e., KL(UP1||UP6) = 0.019. In contrast for the transition to the perturbed trials we computed a distance of KL(UP1||P) = 0.20. For the transition to the washout phase, we found KL(P||UP6) = 0.126.

A more detailed investigation was conducted by analyzing trajectory similarities using the PTM. As reference model, trajectories of the last 20 trials in P or UP6 were used. We compared the reference model to a trajectory model trained from 20 trials in a moving window. The moving window traverses from trial 10 to trial 460 which is denoted by the shaded boxes in Fig. 5B,C. Individual models were trained for the left wrist, the right wrist and the trunk. We found that the trunk model converged about five times faster than the left wrist model. For that we used one-term exponential models of the form $y = c\exp(\lambda x)$, where for the trunk $\lambda_t = 0.03$ and for the left wrist $\lambda_l = 0.006$. This result is highlighted in the inlay in Fig. 5B.

We found that the distribution over model parameters for generating trunk kinematics converges faster than the model for generating the arm kinematics. This is an potential indicator that postural control precedes volitional motor control. Our finding can be explained by the importance of correct trunk motions to prevent

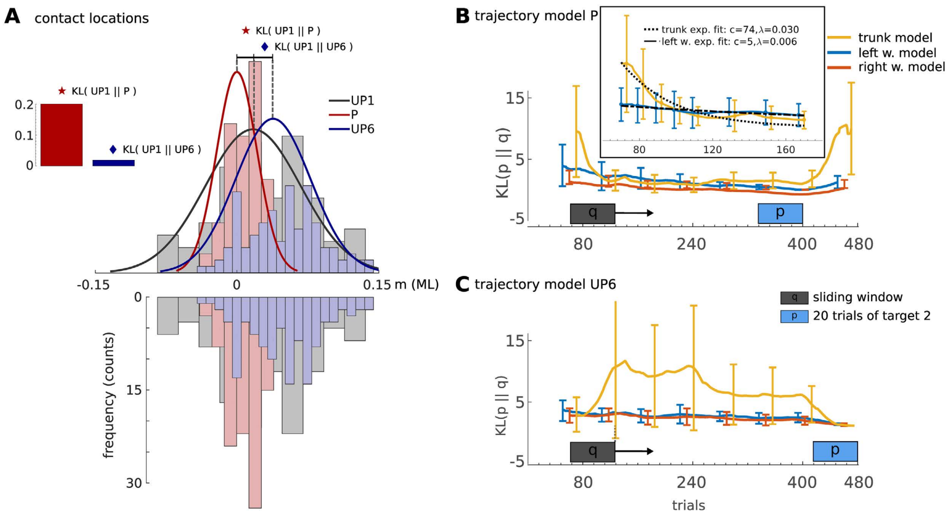

Figure 5. Trunk adaptation precedes arm adaptation. (**A**) The adaptation process of the contact location is illustrated by computing the KL-divergence between the first unperturbed session (UP1), the perturbed trials (P), and the last unperturbed session (UP6). The underlying data is shown as histogram with the Gaussian model fits as overlay. (**B**,**C**) For a detailed temporal analysis, the KL-divergence between a set of training trials (a sliding window of 20 trajectories) and a set of test trials (**B**) last 20 trials in P, (**C**) last 20 trials in UP6) is investigated. An exponential model fit is presented in the inset in (**B**).

falling as observed in related work[31,36]. When comparing the prediction accuracy however, we made an interesting observation. During early phases of the motor execution the left wrist results in more accurate predictions However, when observing the complete motion, the trunk was more informative for predicting the right arm movements, see Supplementary Figure S4. This result suggests that the supportive motion has a stronger effect on the task performance during early phases compared to the massive trunk.

Discussion

It is known that supportive contacts aid postural control in humans[37]. Contacts increase the subjects' belief about their poses in board balancing tasks[2], in tasks without visual cues[32], and have an effect on learning balancing strategies during task adaptation[6]. While these studies demonstrated the importance of contacts for human motor control, the contacts where *static* and pre-defined based on the experimental setting. In this study, we investigated the effect of the *active choice* of supportive contacts on motor control. Under natural conditions subjects were able to choose supportive contact locations to aid target reaching. Utilizing a developed probabilistic trajectory model (PTM), we found that the supportive motions leading to contacts could predict goal-directed reaching movements. A comparison of the rates of convergences of the model parameters in our PTM indicated that learning of postural control and arm reaching strategies proceeds on multiple time scales. These findings have important implications on medical care in diseases related to central nervous system disorders (such as dementia Alzheimer's, Parkinson's disease or stroke), computational neuroscience and robotics.

A pre-test for central nervous system disorders affecting postural control. Many central nervous system disorders affect not only cognitive abilities related to memory consolidation but also postural control. For example, an underdevelopment of postural control is a well known symptom in autism[38,39]. Early detection of these diseases is of utter importance for medical care. However, most pre-tests focus on cognitive functions that are influenced by many factors such as stress, sleep deprivation and age. Classical tests targeting motor coordination abilities are unnatural like for example the grooved pegboard test[40], where the goal is to fit pegs of various shapes into punched holes. The presented experimental setting has the potential to become an alternative pre-test that investigates postural control and motor learning under natural conditions. Deficits in motor control can be quantified in terms of the model prediction performance as shown in Fig. 3. In the tested subjects, the average prediction error of the goal-directed target reaching motion was $6.4 \pm 5.5(SD)$ cm out of a total range of 80 cm Note that for these predictions the model input was solely the x,y,z coordinate of the supportive motion, see Fig. 4B. While the predictions were accurate for the tested healthy subjects, we speculate that participants with motor dysfunctions would show inferior prediction performance scores.

However, a limitation of the presented experimental setting is its simplicity. In our experiments the participants reached for a static target at one out of three possible locations on a horizontal line. Synchronization of movement onsets within 3–5 trials (see Fig. 1C,D) indicates that the task might be too simple to show significant effects in subjects with central nervous system disorders. More complex tasks could consider for example moving targets at arbitrary locations on the screen.

Evidence for learning on multiple time scales. For many tasks motor memory consolidation proceeds on different time scales[25,41,42]. In particular, postural control is adapted on a faster rate in contrast to goal-directed movements. For example, Huys *et al.* showed that postural sway precedes eye and head movements (3:2 or also 3:1) in subjects learning to juggle[43]. In line with this finding, we found that trunk adaptation to maintain balance seems to precede the learning of optimal (here task correlated) supportive contact motions. Concretely, the learned generative model of trunk kinematics converged about 5 times faster than the model of the left wrist kinematics, see Fig. 5B. This result is another indicator for a hierarchical organization of motor control with the difference that we analyzed adaptation on a trajectory level in contrast to single time step models[19,25,41]. The developed PTM may extend future computational models for optimal feedback control by combining sequential predictions of feed forward commands and the integration of perceptual feedback.

A potential deficit of such an optimal feedback controller is that we analyzed adaptation on a kinematic level where limb inertia has a delayed effect on the recorded marker trajectories. For example, while expressive motion vectors can be observed for the light weight arms, the massive trunk may have moved just for few millimeters which could result in inaccurate predictions. This hypothesis is confirmed by our results, where at early movement phases (up to 300 ms) the left wrist leads to more accurate predictions than the trunk, see Supplementary Figure S4. For observations of the complete trial however, the trunk motion is more informative. To avert the effect of limb inertia the PTM could be trained from Electromyography (EMG) patterns. Such models could be used to predict motor commands on a muscle level and could help to better understand the underlying motor control mechanisms.

A robot controller that actively seeks for contacts. In this research we studied whole-body coordination mechanisms of arm reaching and postural control with additional hand contact. We investigated how humans perform reaching movements with the right arm in postural challenged conditions. Postural balance was maintained by establishing a supportive hand contact with the other arm. In effect, both arms had to perform reaching movements where one arm reaches for a distant target and the other arm reaches for a supportive hand contact. We found that humans preferred distinct contact locations for each of the three targets that were displayed. Such a context dependent controller could advance current abilities of humanoid robots as it was envisioned in the European project CoDyCo. In particular, our results suggest that different supportive contact strategies should be initiated based on future intentions. This has not been done so far as research on anticipatory controller focused largely on balancing[44–46] or on compensating for external forces[47–50].

In our experiments the modulating factors (the context) were the distance to the target, its location on the screen and the postural perturbation. The distance was chosen such that subjects *always* made a contact to avoid falling when reaching. We only investigated the effect of the contact location and used pre-defined translateral support base perturbations. The perturbations had no effect on the results. In unperturbed and perturbed conditions the subjects chose target dependent contact locations, see Fig. 1B. Future research may explore the effect of distance to complement a robot controller that autonomously decides in which situations to make supportive contacts.

Methods

Participants. Twenty right-handed male subjects (*age*: $20.8 \pm 1.8(SD)$) participated in the reaching experiments. None reported any neurological or musculoskeletal disorders (self-reported). Prior to their participation, the subjects were informed about the course of the study and were required to sign an informed consent approved by the National Medical Ethics Committee Slovenia (NO. 112/06/13). The experiments were carried out in accordance with the approved guidelines of the National Medical Ethics Committee Slovenia (NO. 112/06/13). All experimental protocol were approved by the National Medical Ethics Committee Slovenia (NO. 112/06/13).

Apparatus. Participants stood on a force plate (9281CA, Kistler Instrumente AG, Winterthur, Switzerland) mounted on top of a Stewart platform[51]. The force plate was used to measure the six components of the ground reaction forces and torques to determine the center-of-pressure (CoP). The CoP was used to improve the quality of the automatically computed movement onsets of the left and the right wrist. In addition to a commonly used peak velocity criteria (two percent was used as threshold here) we ensured that the CoP velocity profiles exceeded a threshold of 0.001 meter per second. By that we could eliminate most trials with incorrectly computed movement onsets due to small movements of the wrists prior to the actual reaching motion. The correctness of the automatically computed movement onsets was verified through a visual inspection of the velocity profiles of all 9600 trials (480 trials $\times$ 20 subjects). Trials with incorrect movement onsets were either manually labeled or rejected.

A second force plate of the same type was mounted anteromedial to the subject's hip position. In concrete this force plate had an offset of $\Delta x = 0.42$ m, $\Delta y = 0$ m, and $\Delta z = 0.92$ m with respect to the base. This force plate was used like a table providing additional support during reaching. The contact locations shown in Fig. 1 were defined as the left wrist position at the peak supportive contact force. In front to the contact force plate we mounted a screen. Both, the force plate and the screen were adjusted in height based on the subject's trunk height (56.5 ± 2.4 cm).

The Stewart platform was used to apply translational perturbations in the mediolateral direction (ML). Specifically, the displacement of the platform was proportional to the sum of the anteroposterior (AP) components of the left and right hand displacements (denoted by Δx_{LW} and Δx_{RW}). The maximal displacement of the Stewart platform was set to 0.2 m and corresponded to the situation when the subject's left hand was at the far edge of the table and the finger on the right hand touched the screen, see Supplementary Figure S5. For this scenario the platform position in the (ML) direction was computed as $pos = 0.1[\Delta x_{\mathrm{LW}}/d_{\mathrm{LW}} + \Delta x_{\mathrm{RW}}/d_{\mathrm{RW}}]$, where $d_{\mathrm{LW}} = 0.68$ m and $d_{\mathrm{RW}} = 1.12$ m. For tracking the desired platform position a P-controller with a gain of $K_p = 9$ was used.

A motion capture system (NDI 3D Investigator) was used to track the participants wrists and trunk movements. Three motion tracking markers were attached to each wrist to compensate for occlusions during the reaching motions. At least one marker needed to be visible and for more than one visible marker the position was computed through averaging. Two markers were attached to the back at the scapulae to transform wrist trajectories to shoulder frames and to estimate the trunk motions (i.e., the center location of the two shoulder markers). Additional markers were placed at the screen, at a wearable thimble to estimate the right index fingertip position and at the force plate mounted on top of the Stewart platform.

Coordinate frames. If not stated different, the results are defined in the subject's coordinate frame. To compensate for the translational perturbations of the Stewart platform a marker was attached to the support force plate. The subjects coordinate frame was defined as the force plate marker position minus the initial left wrist location (to correct for small deviations of the subject's feet placement throughout the experiment).

Wrist trajectories (see Fig. 3) were computed in the shoulder frames to isolate trunk and arm adaptations. In particular, the wrist positions in the subject's coordinate frame were additionally corrected by the time-varying shoulder marker positions.

Procedure. Participants had to perform 480 trials of reaching in blocks of 80 trials. After each block the subjects had a five minutes break. Only the first and the last 80 trials were unperturbed (see Fig. 1B). To investigate negative after effects, the perturbation was deactivated for 30 randomly selected reachings during trials 240 to 400. The participants were not informed about these catch trials.

Prior to each trial, subjects were required to stand upright moving as little as possible. A bar on the screen indicated anteroposterior fluctuations of the CoP in real time with respect to the initial CoP value. This initial CoP was measured at the beginning of the experiment. If the mean of the deviations averaged over 500 ms was less than 1 cm the start of the trial was initiated. After one second the target was presented at one out of three locations (target onset). The targets were displayed at a height of 1.5 m (z-coord. in Fig. 1A) and at a distance of 1.12 m (x-coord. in Fig. 1A). The y-coordinates of the three targets were −0.4, 0 and 0.4 meter. Note that three targets were chosen to define a non-trivial prediction problem that can be solved using the probabilistic trajectory model. In this case the trivial prediction in form of the mean over all kinematic trajectories would result in bad prediction performances for target one and target three.

After a random period (1–3 sec) the target's color switched and the subjects were allowed to move. Movements prior to that visual cue terminated the trial. The target was reached if the distance between the fingertip and the target on the screen was less than 1 cm.

Participants received a monetary reward after each trial based on the time needed for reaching, i.e. EUR = 0.1–0.05t where t denotes the reaching time in seconds. The reaching time was defined as the difference between the target onset and the time until the target was reached. On average the subjects received five cents per trial which corresponds to a reaching duration of one second, see Supplementary Figure S6. If the target was missed, or if the target was not reached within two seconds, no reward was given.

Data processing. All measurements of the force plates and the motion capture system were recorded at a rate of 100 Hz. Redundant marker settings on the wrists and on the thimble were used to compensate for occlusions. For that three markers were used for both wrists and the thimble. If all markers were occluded the trial was excluded from the analysis. For two or three correctly observed markers we computed the average for each coordinate (x,y,z). All marker trajectories were low-pass filtered at 5 Hz using 2nd-order Butterworth filter.

Statistical tests. Statistical analyses were performed using SPSS 21 (SPSS Inc., Chicago, USA). For each subject, an average of left hand contact location in mediolateral direction was calculated for every combination of balance perturbation (unperturbed, perturbed and unperturbed catch trials) and target locations. The average values of the individual subjects were then used for statistical analysis. Effect of perturbation was investigated using one-way repeated measures ANOVA. Differences between perturbed and unperturbed sessions for each combination of independent variables were tested with post hoc t-tests with Bonferroni correction. The level of statistical significance was set to 0.05.

Computing predictions through conditioning. Conditional probabilities and predictions are related concepts in statistical modeling. A prediction problem can modeled as computing the conditional probability denoted by

$$p(B|A) = p(A, B)/p(A), \tag{1}$$

where A and B represent two random variables (RVs). In our experiment, the variable A could encode the probability of a certain contact location on the table and the target location is represented by the RV B. The joint distribution $p(A, B)$ is the learned model and $p(A)$ is a prior over all possible contact locations. Given a particular contact location $p(A = a)$, we can compute the most likely target the human tries to reach. Throughout the paper we will use the shorthand $p(a)$ to denote $p(A = a)$, which is the probability that the RV A takes the value a to keep the notation uncluttered.

A particular interesting class of distributions are Gaussian distributions

$$N(\boldsymbol{x}|\boldsymbol{\mu}, \boldsymbol{\Sigma}) := \frac{1}{(2\pi)^{n/2}\sqrt{\det\boldsymbol{\Sigma}}} \exp\left(-\frac{1}{2}(\boldsymbol{x} - \boldsymbol{\mu})^T \boldsymbol{\Sigma}^{-1}(\boldsymbol{x} - \boldsymbol{\mu})\right), \tag{2}$$

with the n-dimensional RV $\boldsymbol{x} \in \mathbb{R}^n$, the mean $\boldsymbol{\mu}$ and the variance $\boldsymbol{\Sigma}$. In Gaussian distributions, conditional distributions can be computed in closed form. To see this, we define a multivariate normal distribution by concatenating two RVs. The two RVs denote the contact locations and the targets, i.e., $\boldsymbol{x} = \begin{pmatrix} a \\ b \end{pmatrix}$, $\boldsymbol{\mu} = \begin{pmatrix} \mu_a \\ \mu_b \end{pmatrix}$ and $\boldsymbol{\Sigma} = \begin{pmatrix} \Sigma_{aa} & \Sigma_{ab} \\ \Sigma_{ba} & \Sigma bb \end{pmatrix}$. Using these definitions in (2) and after rearranging the terms, the parameters $\mu_{b|a}$ and $\Sigma_{b|a}$ of the conditional distribution can be computed as

$$p(b\,|\,a) = \mathcal{N}(b|\mu_{b|a}, \Sigma_{b|a}), \tag{3}$$

$$\text{with} \quad \mu_{b|a} = \mu_b + \Sigma_{ba}\Sigma_{bb}{}^{-1}(a - \mu_a),$$

$$\text{and} \quad \Sigma_{b|a} = \Sigma_{bb} - \Sigma_{ba}\Sigma_{bb}{}^{-1}\Sigma_{ab}.$$

Given multiple measurements of a and b we can compute the statistics $\boldsymbol{\mu}$ and $\boldsymbol{\Sigma}$. For some during training unseen contact location a^* we can now predict the most likely target location (parametrized through $\mu_{b|a^*}$ and $\Sigma_{b|a^*}$). In the following we will extend this powerful feature of computing predictions to time-series data or trajectories.

Phase modulated probabilistic models of trajectories. The same conditioning operation can be used for computing predictions in trajectories if the RVs encode feature vectors in function approximation, see Fig. 2 for a sketch. Let $\boldsymbol{y}_t = \begin{pmatrix} a_t \\ b_t \end{pmatrix}$ denote a concatenated state at time t representing the contact location and the target. Only for the sake of clarity we will again assume that the contact location and the target are scalars. As we will see later the results generalize to multi-dimensional states. In addition, we will assume for simplicity that each dimension in $\boldsymbol{y}_t$ is approximated through a J-dimensional feature vector $\boldsymbol{w} \in \mathbb{R}^{J\times 1}$ and basis function vectors $\boldsymbol{\phi}_t \in \mathbb{R}^{1\times J}$ e.g. $a_t = \boldsymbol{\phi}_t \boldsymbol{w}_a$ with $\boldsymbol{\phi}_t = [\phi_{t,1}, ..., \phi_{t,J}]^T$. A large variety of possible basis functions can be used for time series approximation. A popular choice for rhythmic movements are Von-Mises basis functions, whereas Gaussian basis functions are widely used for point to point movements[52,53]. In this study we used Gaussian basis functions with

$$\phi_{t,i} = \frac{1}{\mathcal{Z}} \exp\left(-\frac{1}{2h}(z(t) - c_i)^2\right),$$

where $\mathcal{Z} = \sum_{l=1}^{J} \exp(-1/2h(z(t) - c_l)^2)$ denotes a normalization term. The function $z(t)$ implements a mapping from discrete time steps to a movement phase, i.e., $z(t) = (t-1)/(T-1)$ with z: $t \in [1, T] \mapsto [0, 1]$. In this notation state sequences of different length can be modeled and are aligned through the movement phase. Note that in our reaching experiments $z(t) = 0$ denotes the movement onset and $z(t) = 1$ the event when the target was reached. The scalar $c_i \in [0, 1]$ denotes the center of the i-th basis function and h is the bandwidth parameter.

In general, D-dimensional states can be approximated in $\boldsymbol{y}_t = \boldsymbol{\Phi}_t\, \boldsymbol{w}$ using block diagonal matrices $\boldsymbol{\Phi}_t \in \mathbb{R}^{D\times JD}$ and concatenated feature vectors $\boldsymbol{w} = [w_1^T, \ldots, w_D^T]^T \in \mathbb{R}^{JD\times 1}$. For example, when modeling contacts and targets $\boldsymbol{w} = [w_a^T, w_b^T]^T$ and a block diagonal matrix of the form $\boldsymbol{\Phi}_t = \begin{pmatrix} \phi_t & \boldsymbol{0} \\ \boldsymbol{0} & \phi_t \end{pmatrix}$ is used.

Sequences of T states, denoted by $\boldsymbol{\tau} = \boldsymbol{y}_{1:T}$, can be compactly represented by

$$\boldsymbol{\tau} = \boldsymbol{\Phi}_{1:T}\boldsymbol{w} \quad \text{with} \quad \boldsymbol{\Phi}_{1:T} = [\boldsymbol{\Phi}_1^T, \ldots, \boldsymbol{\Phi}_T^T]^T \in \mathbb{R}^{TD\times JD}, \tag{4}$$

where $\boldsymbol{\Phi}_{1:T}$ denotes an extended basis function matrix. Using the function approximation in (4) we can define the generative probabilistic model for trajectories

$$p(\boldsymbol{\tau}|\boldsymbol{w}) = \prod_{t=1}^{T} \mathcal{N}\left(\boldsymbol{y}_t \middle| \boldsymbol{\Phi}_t \boldsymbol{w}, \boldsymbol{\Sigma}_y\right) = \mathcal{N}\left(\boldsymbol{y}_{1:T} \middle| \boldsymbol{\Phi}_{1:T}\boldsymbol{w}, \boldsymbol{\Sigma}_y\right),$$

where $\boldsymbol{\Sigma}_y$ models zero mean independent and identically distributed (i.i.d.) Gaussian noise in $\boldsymbol{y}_t = \boldsymbol{\Phi}_t\, \boldsymbol{w} + \varepsilon_y$ with $\varepsilon_y \sim \mathcal{N}(\varepsilon_y|\boldsymbol{0}, \boldsymbol{\Sigma}_y)$. In order to represent a distribution over trajectories $p(\boldsymbol{\tau})$, we can apply the product rule shown in (1) and compute the marginal. For Gaussian distributions, in particular for $p(\boldsymbol{w}) = \mathcal{N}(\boldsymbol{w}|\boldsymbol{\mu}_w, \boldsymbol{\Sigma}_w)$, the marginal can be computed in closed form

$$\begin{aligned} p(\boldsymbol{\tau}) &= \int p(\boldsymbol{\tau}|\boldsymbol{w})p(\boldsymbol{w})d\boldsymbol{w} \\ &= \int \mathcal{N}(\boldsymbol{y}_{1:T}|\boldsymbol{\Phi}_{1:T}\boldsymbol{w}, \boldsymbol{\Sigma}_y)\mathcal{N}(\boldsymbol{w}|\boldsymbol{\mu}_w, \boldsymbol{\Sigma}_w)d\boldsymbol{w} \\ &= \mathcal{N}(\boldsymbol{y}_{1:T}|\boldsymbol{\Phi}_{1:T}\mathbf{w}, \boldsymbol{\Phi}_{1:T}\boldsymbol{\Sigma}_w\boldsymbol{\Phi}_{1:T}^T + \boldsymbol{\Sigma}_y)\ . \end{aligned} \tag{5}$$

The mean $\boldsymbol{\mu}_w$ and the covariance matrix $\boldsymbol{\Sigma}_w$ can be learned from data by maximum likelihood using the Expectation Maximization (EM) algorithm[54]. A simpler solution that works well in practice is to compute first the most likely estim ate of $\boldsymbol{w}^{[i]}$ for each trajectory $\boldsymbol{\tau}^{[i]}$ independently, where the index i denotes the i-th demonstration. In particular, given the trajectory $\boldsymbol{\tau}^{[i]}$, the corresponding weight vector $\boldsymbol{w}^{[i]}$ can be estimated by a straightforward least squares estimate

$$\boldsymbol{w}^{[i]} = (\boldsymbol{\Phi}_{1:T}^T \boldsymbol{\Phi}_{1:T} + \lambda \boldsymbol{I})^{-1} \boldsymbol{\Phi}_{1:T}^T \boldsymbol{\tau}^{[i]}. \tag{6}$$

Note that the least squares estimate is also known as regularized pseudo-inverse solution to the generative model $\boldsymbol{\tau}^{[i]} = \boldsymbol{\Phi}_{1:T} \boldsymbol{w}^{[i]}$. The regularization parameter λ is used to avoid numerical singularities. In our recordings we used $\lambda = 1e-6$. After computing the vectors for each trajectory, the mean and the covariance of $p(\boldsymbol{w})$ can be estimated by the sample mean and sample covariance of the $\boldsymbol{w}^{[i]}$ vectors.

Computing predictions in trajectories. Let us assume that we have observed a sequence of states $\boldsymbol{y}_{t_1}$ to $\boldsymbol{y}_{t_M}$ at $m = 1, 2, ..., M$ different time points, which do not need to be sampled at uniform intervals. In addition, not all of the dimensions in the vector $\boldsymbol{y}_{t_m}$ might be observed (e.g. due to occlusions, where irrelevant dimensions are assumed to be set to some real number). We introduce the homogeneous covariance matrix $\boldsymbol{\Sigma}_{\boldsymbol{o}}$ to control the importance of the observations, where small diagonal variances denote important dimensions and large scalars irrelevant dimensions. For example, we would use $\boldsymbol{\Sigma}_{\boldsymbol{o}} = \begin{pmatrix} 1e-5 & 0 \\ 0 & 1e5 \end{pmatrix}$ if only the first out of two dimensions is observed (as in the example with contact locations and targets). Let us further denote $\boldsymbol{\Phi}_{\boldsymbol{o}}$ as the concatenation of the basis function matrices for these time points and $\boldsymbol{o}$ as concatenation of the $\boldsymbol{y}_{t_m}$ vectors. Given these observations, we can obtain a conditioned distribution $p(\boldsymbol{w}|\boldsymbol{o})$ over the weight vectors

$$\begin{aligned} p(\boldsymbol{w}_o|\boldsymbol{o}) &\propto \mathcal{N}(\boldsymbol{o}|\boldsymbol{\Phi}_{\boldsymbol{o}}\boldsymbol{w}_o, \boldsymbol{\Sigma}_{\boldsymbol{o}})p(\boldsymbol{w}) \\ &:= \mathcal{N}\left(\boldsymbol{w}_o|\boldsymbol{\mu}_{w|\boldsymbol{o}}, \boldsymbol{\Sigma}_{w|\boldsymbol{o}}\right), \\ &\text{with} \quad \boldsymbol{\mu}_{\boldsymbol{w}|\boldsymbol{o}} = \boldsymbol{\mu}_{\boldsymbol{w}} + \boldsymbol{\Sigma}_{\boldsymbol{w}}\boldsymbol{\Phi}_{\boldsymbol{o}}^T(\boldsymbol{\Sigma}_o + \boldsymbol{\Phi}_{\boldsymbol{o}}\boldsymbol{\Sigma}_{\boldsymbol{w}}\boldsymbol{\Phi}_{\boldsymbol{o}}^T)^{-1}(\boldsymbol{o} - \boldsymbol{\Phi}_o\boldsymbol{\mu}_{\boldsymbol{w}}), \\ &\text{and} \quad \boldsymbol{\Sigma}_{\boldsymbol{w}|\boldsymbol{o}} = \boldsymbol{\Sigma}_{\boldsymbol{w}} - \boldsymbol{\Sigma}_{\boldsymbol{w}}\boldsymbol{\Phi}_{\boldsymbol{o}}^T(\boldsymbol{\Sigma}_o + \boldsymbol{\Phi}_{\boldsymbol{o}}\boldsymbol{\Sigma}_{\boldsymbol{w}}\boldsymbol{\Phi}_{\boldsymbol{o}}^T)^{-1}\boldsymbol{\Phi}_{\boldsymbol{o}}\boldsymbol{\Sigma}_{\boldsymbol{w}}, \end{aligned}$$

which recovers the conditioning result in (3) in the feature space.

To predict the state sequence $\tilde{\mathbf{y}}_{1:\mathbf{T}}$, the conditional distribution is projected back into the trajectory space using (5). The result is a distribution over trajectories

$$p(\tilde{\tau}) = \mathcal{N}\left(\tilde{\boldsymbol{y}}_{1:T}|\boldsymbol{\Phi}_{1:T}\boldsymbol{\mu}_{\boldsymbol{w}|\boldsymbol{o}}, \boldsymbol{\Phi}_{1:T}\boldsymbol{\Sigma}_{\boldsymbol{w}|\boldsymbol{o}}\boldsymbol{\Phi}_{1:T}^T + \boldsymbol{\Sigma}_y\right), \tag{7}$$

where for $t_M < T$ in the observations $\boldsymbol{o} = [\boldsymbol{y}_{t_1}, \boldsymbol{y}_{t_2}, \ldots, \boldsymbol{y}_{t_M}]$ future states can be predicted. For details on the implementation and usage of the model we refer the reader to the Supplementary Code to this manuscript[55].

Computing model comparisons in trajectories. Using the proposed probabilistic trajectory model (PTM), two trajectory models can be compared through computing the Kullback-Leibler (KL) divergence. For Gaussian distributions as in (5) the KL divergence can be computed in closed form

$$\mathrm{KL}(\mathcal{N}_1 \| \mathcal{N}_2) = \frac{1}{2} \log \frac{|\boldsymbol{\Sigma}_2|}{|\boldsymbol{\Sigma}_1|} - n + \mathrm{tr}(\boldsymbol{\Sigma}_2^{-1}\boldsymbol{\Sigma}_1) + (\boldsymbol{\mu}_2 - \boldsymbol{\mu}_1)^T \boldsymbol{\Sigma}_2^{-1} (\boldsymbol{\mu}_2 - \boldsymbol{\mu}_1),$$

where $\mathcal{N}_1$ denotes a Gaussian with the n-dimensional mean $\boldsymbol{\mu}_1$ and the covariance $\boldsymbol{\Sigma}_1$. The symbol tr denotes the matrix trace.

Computing task likelihoods in trajectories. Another important probabilistic operation is the computation of the task likelihood. This operation can be used to test which candidate model out of several best explains the data. Let $\mathcal{N}_p(.|\boldsymbol{\mu}, \boldsymbol{\Sigma})$ denote the model under test. Given a single trajectory sample $\boldsymbol{\tau}^*$ and its feature space representation $\boldsymbol{w}^*$ we can compute

$$\begin{aligned} L\left(\boldsymbol{w}^* \middle| \mathcal{N}_p\right) &= \log \mathcal{N}_p(\boldsymbol{w}^*|\boldsymbol{\mu}, \boldsymbol{\Sigma}) \\ &= -\frac{1}{2}(\boldsymbol{\mu} - \boldsymbol{w}^*)^T \boldsymbol{\Sigma}^{-1} (\boldsymbol{\mu} - \boldsymbol{w}^*) + \text{a constant}, \end{aligned}$$

which quantifies how likely $\boldsymbol{\tau}^* = \boldsymbol{\Phi}_{1:T}\boldsymbol{w}^*$ was generated by model $\mathcal{N}_p(.|\boldsymbol{\mu}, \boldsymbol{\Sigma})$.

References

1. Jeka, J. J. & Lackner, J. R. Fingertip contact influences human postural control. *Experimental Brain Research* **79(2),** 495–502, doi: 10.1007/BF00229188 (1994).
2. Slijper, H. & Latash, M. The effects of instability and additional hand support on anticipatory postural adjustments in leg, trunk, and arm muscles during standing. *Experimental Brain Research* **135(1),** 81–93, doi: 10.1007/s002210000492 (2000).
3. Dickstein, R., Peterka, R. & Horak, F. Effects of light fingertip touch on postural responses in subjects with diabetic neuropathy. *Journal of Neurology, Neurosurgery & Psychiatry* **74(5),** 620–626, doi: 10.1136/jnnp.74.5.620 (2003).

4. Kouzaki, M. & Masani, K. Reduced postural sway during quiet standing by light touch is due to finger tactile feedback but no mechanical support. *Experimental brain research* **188(1),** 153–158, doi: 10.1007/s00221-008-1426-5 (2008).
5. Sarraf, T. A., Marigold, D. S. & Robinovitch, S. N. Maintaining standing balance by handrail grasping. *Gait & posture* **39(1),** 258–264 doi: 10.1016/j.gaitpost.2013.07.117 (2014).
6. Babič, J., Petrič, T., Peternel, L. & Šarabon, N. Effects of supportive hand contact on reactive postural control during suppor perturbations. *Gait & posture* **40(3),** 441–446, doi: 10.1016/j.gaitpost.2014.05.012 (2014).
7. Maki, B. E. & McIlroy, W. E. The role of limb movements in maintaining upright stance: the "change-in-support" strategy. *Physica therapy* **77(5),** 488–507 (1997).
8. Maki, B. E. & McIlroy, W. E. Control of rapid limb movements for balance recovery: age-related changes and implications for fal prevention. *Age and ageing* **35** (suppl 2), ii12–ii18, doi: 10.1093/ageing/afl078 (2006).
9. Reed, E. S. An outline of a theory of action systems. *Journal of motor behavior* **14(2),** 98–134 (1982).
10. Von Hofsten, C. Prospective control: A basic aspect of action development. *Human development* **36(5),** 253–270 doi: 10.1159/000278212 (1993).
11. Adolph, K. Learning to keep balance. *Advances in child development and behavior* **30,** 1–40, doi: 10.1016/S0065-2407(02)80038-((2002).
12. Adolph, K. E. & Berger, S. E. Motor development. *Handbook of child psychology* **2,** 161–213 (2006).
13. Pai, Y.-C. & Patton, J. Center of mass velocity-position predictions for balance control. *Journal of biomechanics* **30(4),** 347–354 doi: 10.1016/S0021-9290(96)00165-0 (1997).
14. Scott, S. H. Optimal feedback control and the neural basis of volitional motor control. *Nature Reviews Neuroscience* **5(7),** 532–546 doi: 10.1038/nrn1427 (2004).
15. Todorov, E. Optimality principles in sensorimotor control. *Nature neuroscience* **7(9),** 907–915, doi: 10.1038/nn1309 (2004).
16. Scott, S. H., Cluff, T., Lowrey, C. R. & Takei, T. Feedback control during voluntary motor actions. *Current opinion in neurobiology* **33** 85–94, doi: 10.1016/j.conb.2015.03.006 (2015).
17. Duncan, P. W., Weiner, D. K., Chandler, J. & Studenski, S. Functional reach: a new clinical measure of balance. *Journal of gerontolog* **45(6),** M192–M197, doi: 10.1093/geronj/45.6.M192 (1990).
18. Horak, F. B. & Nashner, L. M. Central programming of postural movements: adaptation to altered support-surface configurations *Journal of neurophysiology* **55(6),** 1369–1381 (1986).
19. Kuo, A. D. An optimal control model for analyzing human postural balance. *IEEE Transactions on Biomedical Engineering* **42(1)** 87–101, doi: 10.1109/10.362914 (1995).
20. Winter, D. A. Human balance and posture control during standing and walking. *Gait & posture* **3(4),** 193–214, doi: 10.1016/0966- 6362(96)82849-9 (1995).
21. Lockhart, D. B. & Ting, L. H. Optimal sensorimotor transformations for balance. *Nature Neuroscience* **10(10),** 1329–1336 doi: 10.1038/nn1986 (2007).
22. Shadmehr, R., Mussa-Ivaldi, F. A. & Bizzi, E. Postural force fields of the human arm and their role in generating multijoin movements. *The Journal of neuroscience* **13(1),** 45–62 (1993).
23. Wolpert, D. M., Ghahramani, Z. & Jordan, M. I. Are arm trajectories planned in kinematic or dynamic coordinates? an adaptatior study. *Experimental brain research* **103(3),** 460–470, doi: 10.1007/BF00241505 (1995).
24. d'Avella, A., Portone, A., Fernandez, L. & Lacquaniti, F. Control of fast-reaching movements by muscle synergy combinations. *Th Journal of neuroscience* **26(30),** 7791–7810, doi: 10.3389/fncom.2013.00042 (2006).
25. Diedrichsen, J., Shadmehr, R. & Ivry, R. B. The coordination of movement: optimal feedback control and beyond. *Trends in cognitiv sciences* **14(1),** 31–39, doi: 10.1016/j.tics.2009.11.004 (2010).
26. Burdet, E., Osu, R., Franklin, D. W., Milner, T. E. & Kawato, M. The central nervous system stabilizes unstable dynamics by learning optimal impedance. *Nature* **414(6862),** 446–449, doi: 10.1038/35106566 (2001).
27. Franklin, D. W. *et al.* Endpoint stiffness of the arm is directionally tuned to instability in the environment. *The Journal of neuroscienc* **27(29),** 7705–7716, doi: 10.1523/JNEUROSCI.0968-07.2007 (2007).
28. Franklin, D. W. *et al.* CNS learns stable, accurate, and efficient movements using a simple algorithm. *The Journal of neuroscienc* **28(44),** 11165–11173, doi: 10.1523/JNEUROSCI.3099-08.2008 (2008).
29. Berger, D. J., Gentner, R., Edmunds, T., Pai, D. K. & d'Avella, A. Differences in adaptation rates after virtual surgeries provide direc evidence for modularity. *The Journal of Neuroscience* **33(30),** 12384–12394, doi: 10.1523/JNEUROSCI.0122-13.2013 (2013).
30. Flash, T. & Mussa-Ivaldi, F. Human arm stiffness characteristics during the maintenance of posture. *Experimental brain researcl* **82(2),** 315–326 (1990).
31. Stapley, P. J., Pozzo, T., Cheron, G. & Grishin, A. Does the coordination between posture and movement during human whole-body reaching ensure center of mass stabilization? *Experimental Brain Research* **129(1),** 134–146, doi: 10.1007/s002210050944 (1999).
32. Johannsen, L., Wing, A. M. & Hatzitaki, V. Effects of maintaining touch contact on predictive and reactive balance. *Journal o neurophysiology* **97(4),** 2686–2695, doi: 10.1152/jn.00038.2007 (2007).
33. Bland, M. *et al. An introduction to medical statistics.* Ed. 3, doi: 10.1002/sim.4780071212 (Oxford University Press, 2000).
34. Klatzky, R. L., Fikes, T. G. & Pellegrino, J. W. Planning for hand shape and arm transport when reaching for objects. *Acta Psychologic* **88(3),** 209–232, doi: 10.1016/0001-6918(93)E0068-D (1995).
35. Rueckert, E., Mundo, J., Paraschos, A., Peters, J. & Neumann, G. Extracting low-dimensional control variables for movemen primitives. In *IEEE International Conference on Robotics and Automation (ICRA), Seattle, WA, USA,* 1511–1518 (IEEE, 2015).
36. Bouisset, S. & Zattara, M. A sequence of postural movements precedes voluntary movement. *Neuroscience letters* **22(3),** 263–270 doi: 10.1016/0304-3940(81)90117-8 (1981).
37. Balasubramaniam, R. & Wing, A. M. The dynamics of standing balance. *Trends in cognitive sciences* **6(12),** 531–536, doi: 10.1016/ S1364-6613(02)02021-1 (2002).
38. Schmitz, C., Martineau, J., Barthélémy, C. & Assaiante, C. Motor control and children with autism: deficit of anticipatory function *Neuroscience letters* **348(1),** 17–20, doi: 10.1016/S0304-3940(03)00644-X (2003).
39. Minshew, N. J., Sung, K., Jones, B. L. & Furman, J. M. Underdevelopment of the postural control system in autism. *Neurology* **63(11)** 2056–2061, doi: 10.1212/01.WNL.0000145771.98657.62 (2004).
40. Klove, H. Clinical neuropsychology. *The medical clinics of North America* **47,** 1647–1658 (1963).
41. Wolpert, D. M. & Ghahramani, Z. Computational principles of movement neuroscience. *Nature Neuroscience* **3,** 1212–1217 doi: 10.1038/81497 (2000).
42. Smith, M. A., Ghazizadeh, A. & Shadmehr, R. Interacting adaptive processes with different timescales underlie short-term motor learning. *PLoS Biol* **4(6),** e179, doi: 10.1371/journal.pbio.0040179 (2006).
43. Huys, R., Daffertshofer, A. & Beek, P. Multiple time scales and multiform dynamics in learning to juggle. *Motor Control* **8,** 188–212 doi: 10.1016/j.humov.2004.08.009 (2004).
44. Sugihara, T. & Nakamura, Y. Whole-body cooperative balancing of humanoid robot using cog jacobian. In *IEEE/RSJ Internationa Conference on Intelligent Robots and Systems (IROS), Lausanne, Switzerland,* vol. 3, 2575-2580 (IEEE, 2002).
45. Rueckert, E., Mindt, M., Peters, J. & Neumann, G. Robust policy updates for stochastic optimal control. In *International Conferenc on Humanoid Robots (Humanoids), Madrid, Spain,* 388-393 (IEEE, 2014).
46. Tassa, Y., Erez, T. & Todorov, E. Synthesis and stabilization of complex behaviors through online trajectory optimization. Ir *International Conference on Intelligent Robots and Systems (IROS), Vilamoura, Algarve, Portugal,* 4906-4913 (IEEE, 2012).

47. Atkeson, C. G. & Stephens, B. Multiple balance strategies from one optimization criterion. In *IEEE-RAS International Conference on Humanoid Robots, Pittsburgh, PA, USA,* 57–64 (IEEE, 2007).
48. Hyon, S.-H., Hale, J. G. & Cheng, G. Full-body compliant human-humanoid interaction: balancing in the presence of unknown external forces. *IEEE Transactions on Robotics* **23(5),** 884–898, doi: 10.1109/TRO.2007.904896 (2007).
49. Rueckert, E. & Neumann, G. A study of morphological computation by using probabilistic inference for motor planning. In *International Conference on Morphological Computation, Venice, Italy,* 51–53 (2011).
50. Toussaint, M. *et al.* Dual execution of optimized contact interaction trajectories. In *International Conference on Intelligent Robots and Systems (IROS), Chicago, Illinois, USA,* 47–54 (IEEE, 2014).
51. Stewart, D. A platform with six degrees of freedom. *Proceedings of the institution of mechanical engineers* **180(1),** 371–386, doi: 10.1108/eb034141 (1965).
52. Rueckert, E. & d'Avella, A. Learned parametrized dynamic movement primitives with shared synergies for controlling robotic and musculoskeletal systems. *Frontiers in Computational Neuroscience* **7(138),** doi: 10.3389/fncom.2013.00138 (2013).
53. Ijspeert, A. J., Nakanishi, J., Hoffmann, H., Pastor, P. & Schaal, S. Dynamical movement primitives: learning attractor models for motor behaviors. *Neural computation* **25(2),** 328–373, doi: 10.1162/NECO_a_00393 (2013).
54. Dempster, A., Laird, N. & Rubin, D. Maximum likelihood from incomplete data via the EM algorithm. *Journal of the Royal Statistical Society, Series B* **39(1),** 1–38, doi: 10.1.1.133.4884 (1977).
55. Rueckert, E. & Paraschos, A. Supplementary matlab demo code to probabilistic trajectory models. http://www.ausy.tu-darmstadt.de/uploads/Main/HomePage/PTMDemo.zip (2016). Supplementary Matlab Demo Code to Probabilistic Trajectory Models.

Acknowledgements

The research leading to these results has received funding from the European Community's Seventh Framework Programme (FP7/2007-2013) under grant agreements No. 600716 (CoDyCo) and No. 610967 (TACMAN). The authors would like to thank Oriane Dermy, Rudolf Lioutikov, Serena Ivaldi and Alexandros Paraschos for their comments that greatly improved the manuscript.

Author Contributions

E.R., J.P. and J.B. conceived the story and designed the experiment. J.C. and J.B. executed the experiments and performed the statistical tests. E.R. developed the probabilistic model and conducted the model-based analyses. E.R., J.C., J.P, and J.B. wrote the manuscript.

Additional Information

Supplementary information accompanies this paper at http://www.nature.com/srep

Competing financial interests: The authors declare no competing financial interests.

How to cite this article: Rueckert, E. *et al.* Probabilistic Movement Models Show that Postural Control Precedes and Predicts Volitional Motor Control. *Sci. Rep.* **6**, 28455; doi: 10.1038/srep28455 (2016).

Supplement to *Probabilistic Movement Models Show that Postural Control Precedes and Predicts Volitional Motor Control*

Elmar Rueckert[1,*], Jernej Čamernik[2], Jan Peters[1,3], and Jan Babič[2]

[1]Intelligent Autonomous Systems Lab, Technische Universität Darmstadt, 64289, Germany
[2]Department for Automation, Biocybernetics and Robotics, Jožef Stefan Institute, Ljubljana, SI-1000, Slovenia
[3]Robot Learning Group, Max-Planck Institute for Intelligent Systems, Tuebingen, 72076, Germany
[*]rueckert@ias.tu-darmstadt.de

ABSTRACT

This supplement provides additional figures to the human recordings (Fig. S1, S2), the probabilistic trajectory model (Fig. S3, S4) and the experimental setting (Fig. S5, S6).

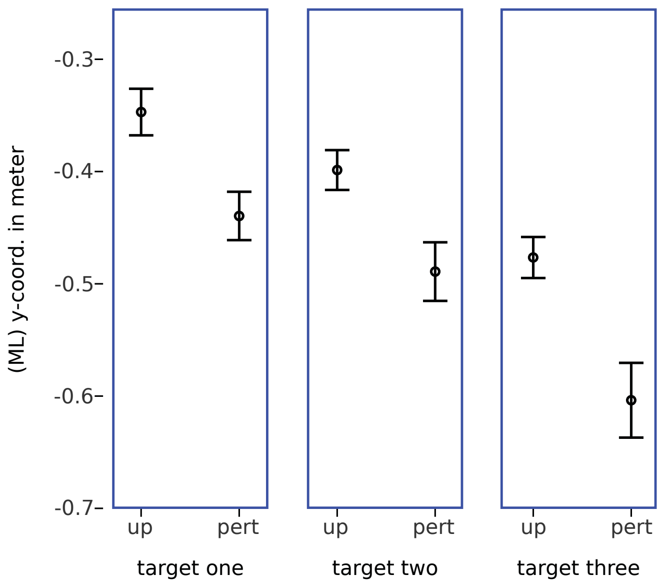

Figure S1. Supplementary figure, Means of the contact locations: From left to right the means and the standard deviations over all 20 subjects are shown for the three target locations. In each panel we contrast the unperturbed and the perturbed session.

Details to Supplementary Figure S7 The primary role of the monetary reward was to motivate the subjects to perform fast reaching movements. By that we could study skill adaptation within the first few hundred milliseconds where feedback had no effect. Here we provide additional results that show that the target location had an effect on the reward but not the experimental condition (unperturbed or perturbed).

Analysis of variance (ANOVA) showed significant effect in the amount of monetary reward between sessions when the subjects were perturbed and unperturbed during reaching toward the targets, $F(1,19) = 8.581$, $p = .009$, $\eta_p^2 = .311$.

Post hoc analyses showed that the amount of received reward was significantly different between reaching toward all three distinctly positioned targets in unperturbed trials,$t(19) = -15.44$ to 11.77, $p \leq .038$, and between targets 1 and 2 and targets 2 and 3, $t(19) = -15.51$ to -12.51, $p < .001$ in perturbed trials.

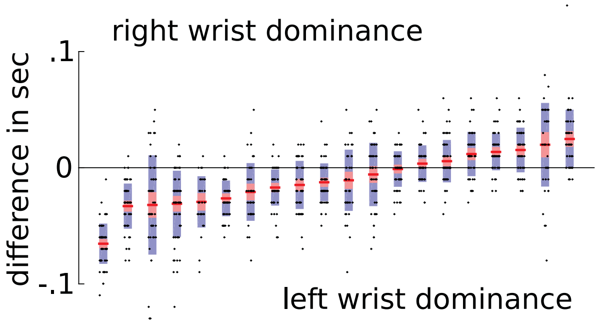

Figure S2. **Supplementary figure, Lead of the left wrist in 12 out of 20 subjects:** We show the difference of the movement onsets (left wrist minus right wrist). The underlying data are representative trials of the 2nd perturbed session (170 to 240), where learning converged.

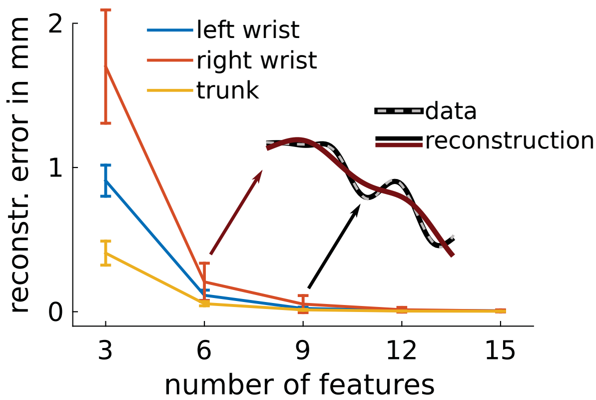

Figure S3. **Supplementary figure. Model complexity (number of basis functions) determines the reconstruction error:** With an increasing number of basis functions the reconstruction error (shown in millimeter) decreases. The reconstruction error is defined as the Euclidean distance of the generated trajectory to its observed counterpart. For the three limbs, left wrist, right wrist and trunk, the error converges to zero with more than nine basis functions. Ten basis functions are used in the experiments in the manuscript.

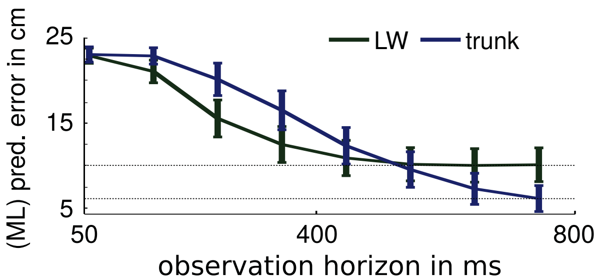

Figure S4. **Supplementary figure, Trunk vs. Left Wrist prediction error:** Average prediction error in cm over all subjects given the left wrist (LW) or the trunk motion as observation.

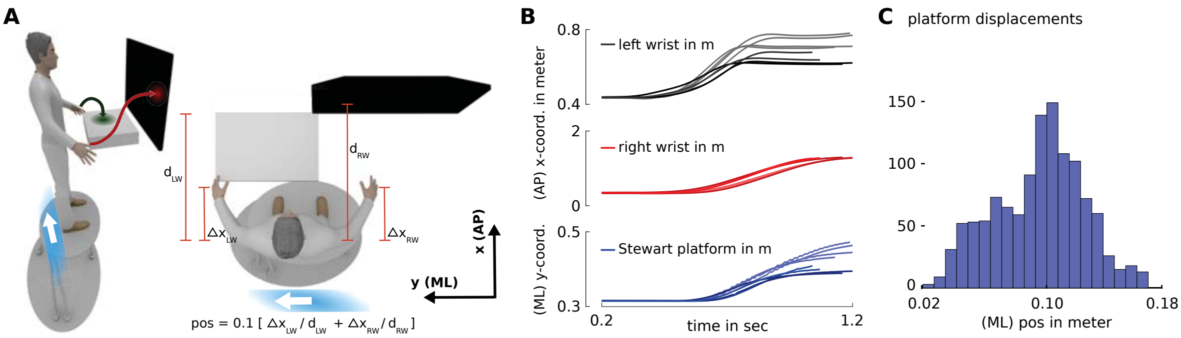

Figure S5. **Supplementary figure, Translateral perturbations: A)** The Stewart platform was used to apply translational perturbations in the mediolateral direction. Specifically, the displacement of the platform was proportional to the sum of the anteroposterior components of the left and right hand displacements. The maximal displacement of the Stewart platform was 20 cm and corresponded to the situation when the subject's left hand was at the far edge of the table and the finger on the right hand touched the screen. **B)** Sample trajectories of the left wrist, the right wrist and the displacement of the platform.

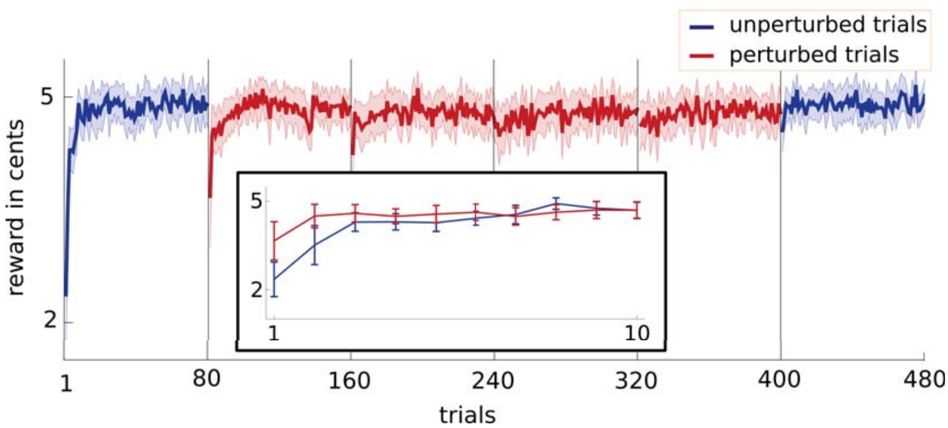

Figure S6. **Supplementary figure, Monetary rewards received:** Illustration of the received rewards averaged over all 20 subjects. The shaded region denotes the confidence bound.

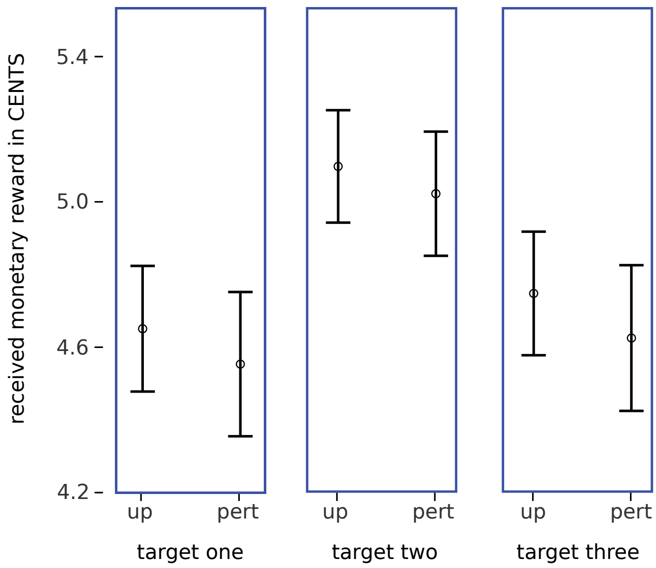

Figure S7. **Supplementary figure, Means of monetary rewards received:** From left to right the means and the standard deviations over all 20 subjects are shown for the three target locations. In each panel we contrast the unperturbed and the perturbed session.

However, the post hoc analysis showed that there were no significant differences between means of received reward for each target between perturbed and unperturbed trials (target one, $t(19) = 3.42$, $p = .057$; target two, $t(19) = 2.9$, $p = .171$; target three, $t(19) = 3.47$, $p = .057$).

Chapter 8

Conclusions

The aim of this thesis was to answer a few specific questions regarding postural control. Specifically, the focus was on control and adjustments of posture in situations where whole-body balance is compromised while standing and holding a stationary handle. The effects of this active support were evaluated in a series of experiments. We used the perturbation-based experimental paradigms where balance of the participants was challenged by either moving their base of support or applying forces directly to their COM. In this way we established a controlled experimental environment where we could faithfully measure the effects of the extended base of support on standing balance. Furthermore, it enabled us to observe and measure the mechanisms of postural control with and without additional hand support.

The first goal of this thesis was to explore the mechanisms of postural control which act during mild, continuous and varying perturbations. The second goal was to characterize the different standing postures, with and without holding a handle during discrete perturbations and to find out which of these postures is least affected in terms of loss of balance and energy expenditure. Finally, the third goal was to explore whether the positioning of the supportive contact (i.e. leaning on a table) is determined by an ongoing task, an unexpected perturbation or both. I successfully answered all the questions which we set at the beginning of the thesis. Therefore, the main findings are as follows:

Study I. The extended base of support, achieved by firmly holding a stationary handle, plays a significant role in continuous balance-perturbing environments. Furthermore, besides the reduced movement of the COP, the results also show a significant change in the overall posture. Namely, we found that when the participants did not hold the handle during the perturbations, their posture changed from complete extension in their knees and hips to slightly flexed configurations of these joints. As a consequence, the position of their COM moved closer to the base of support and the range of motion in these joints increased. A lower vertical position of the COM allows more control over balance, and a larger range of movement in the joints facilitates more possibilities for movement and adjustments of posture. Beside the changes in posture, we found an interesting change in the variability of movement of the COP. It appears that the COP movement was larger when the participants were holding a handle compared to when they were standing freely. This finding implies that because of the extended base of support, the range of whole-body movement was automatically 'allowed' to be larger. By examining the activity of postural muscles, we found an overall reduction in trunk and leg muscle activity when the participants held to the handle. The reduced muscle activity due to the different configuration of posture (holding a handle, engaging/including the arm muscles and extending base of support) was evident right from the start of the perturbation period. And even though the additional support actually allowed for more movement of the body, it seems that controlling this movement did not require more effort - quite the opposite. It appears that using muscles that are closer to the source of perturbation actually facilitated balance control. In the additional analyses we have also shown that the frequency component of the COP movement changed when the participants released the handle and were standing freely. A significant share of frequencies shifted to a higher frequency band (1 - 3 Hz) which indicates a loss of somatosensory cues which were present when the participants were holding a handle. Finally, by calculating the cumulative joint work, we estimated the total body energy expenditure which increased with the increased postural threat (i.e. when the participants were not holding a handle).

Study II. Handles are meant to provide additional support and to prevent falls due to a sudden loss of balance. However, the mechanisms of using handles mounted at different heights are not yet fully disclosed. We simulated full body perturbations by applying an anterior force to the waist of the participants to induce backward motion and investigated effectiveness and mechanisms of balance recovery in five different postures: step stance and normal stance with or without holding handles mounted at different heights. Results indicate that both step stance and holding handles at different vertical positions sufficiently assist balance recovery, compared to normal stance. While there was no significant effect of handle in COM displacement, we found that at higher perturbation magnitudes the calculated total joint torque started to differentiate between the handle positions. Furthermore, using the shoulder height handle for balance required the lowest force exerted on the handle and the direction of the applied force changed significantly which indicates a difference in using this aid. A finding that young adults predominantly 'hang' onto handles positioned over their heads, while they 'push/pull' onto lower positioned handles provides the basis for optimal positioning of assistive handles. Additional analysis of calculated joint torques showed that the loading of the joints was significantly lower when the participants were using the handles. However, the difference between handles mounted at different heights was not established.

Study III. While supportive contacts are essential for many tasks people do every day, little is known about their impact on motor learning. In the last study we were questioning whether certain motor actions depend on the task, or some other higher motor priority. We put the participants on top of a movable platform and asked them to reach forward and hit with their finger a target which appeared on the screen in front of them. During their reaching action we perturbed their balance by quickly moving the platform laterally. The movement of the platform depended on the movement of the participants' other arm, which they had to use to lean on the table in front of them. We found that participants adapted their reaches and supportive contacts to randomly appearing targets in front of them while experiencing a whole-body perturbation. Specifically, we found that they chose a supportive contact location based on the position of the target and not based on the direction of the perturbation as was observed in previous research from other authors. We attribute this to the case of the dual task present in our experiment - reaching toward a target and keeping balance - and confirms our assumptions about the task-dependent adaptation. Moreover, we found that adaptation of the trunk motion was significantly faster compared to the adaptation of the motion of the arms. This indicates to a movement priority given to the movement of the COM. Lastly, with a novel probabilistic trajectory model approach, we found the observations of the initial phase of the left arm or trunk motion (100 - 400 ms) to be sufficient to faithfully predict the complete movement of the right arm.

Appendix A

Ethical Approval

KOMISIJA REPUBLIKE SLOVENIJE ZA MEDICINSKO ETIKO

Luka Peternel, univ. dipl. inž. el.
Odsek za avtomatizacijo, biokibernetiko in robotiko
Institut Jožef Stefan
Jamova 39, 1000 Ljubljana

Štev.: 112/06/13
Datum: 17. 7. 2013

Spoštovani gospod inž. Peternel,

Komisiji za medicinsko etiko (KME) ste 27. 5. 2013 poslali v oceno predlog raziskave z naslovom:

"Razumevanje in modeliranje človekovega gibanja v fizični relaciji z okoljem."

Prostovoljca boste postavili na gibajočo se »robotsko ploščad«, kjer bo lovil ravnotežje. Podatke boste uporabli za razvoj algoritmov za vodenje humanoidnih robotov.

Svetujemo, da uporabljate standardno izrazje: "Navodila za kandidate" → pojasnila za osebe, povabljene v raziskavo. "Vprašalnik za kandidate" → privolitveni obrazec.

KME je na seji 11. junija 2013 ocenila, da je raziskava etično sprejemljiva, in Vam s tem izdaja svoje soglasje.

Lep pozdrav,

prof. dr. Jože Trontelj
predsednik Komisije RS za medicinsko etiko

Naslov:
Prof. dr. Jože Trontelj, Inštitut za klinično nevrofiziologijo, Univerzitetni klinični center Ljubljana, Zaloška 7, 1525 Ljubljana, telefon 01/ 522 1525.
Tone Žakelj; Univerzitetni klinični center Ljubljana, Zaloška 7, 1525 Ljubljana, telefon 01/ 522 1509.
Telefaks 01/ 522 1533; naslova za elektronsko pošto: joze.trontelj@kclj.si, tone.zakelj@kclj.si
Spletni naslov: http://www.kme-nmec.si/

References

Alexandrov, A. V., Frolov, A. A., Horak, F. B., Carlson-Kuhta, P., & Park, S. (2005). Feedback equilibrium control during human standing. *Biological Cybernetics*, *93*(5), 309–322. https://doi.org/10.1007/s00422-005-0004-1

Anderson, D. E., Madigan, M. L., & Nussbaum, M. A. (2007). Maximum voluntary joint torque as a function of joint angle and angular velocity: Model development and application to the lower limb. *Journal of Biomechanics*, *40*(14), 3105–3113. https://doi.org/10.1016/j.jbiomech.2007.03.022

Arienti, C., Lazzarini, S. G., Pollock, A., & Negrini, S. (2019). Rehabilitation interventions for improving balance following stroke: An overview of systematic reviews (S. Kumar, Ed.). *PLOS ONE*, *14*(7), e0219781. https://doi.org/10.1371/journal.pone.0219781

Aruin, A. S., & Latash, M. (1995). The role of motor action in anticipatory postural adjustments studied with self-induced and externally triggered perturbations. *Experimental Brain Research*, *106*(2), 291–300. https://doi.org/10.1007/BF00241125

Assländer, L., & Peterka, R. J. (2014). Sensory reweighting dynamics in human postural control. *Journal of Neurophysiology*, *111*(9), 1852–1864. https://doi.org/10.1152/jn.00669.2013

Babič, J., Petrič, T., Peternel, L., & Šarabon, N. (2014). Effects of supportive hand contact on reactive postural control during support perturbations. *Gait & Posture*, *40*(3), 441–446. https://doi.org/10.1016/j.gaitpost.2014.05.012

Bateni, H., & Maki, B. E. (2005). Assistive devices for balance and mobility: Benefits, demands, and adverse consequences. *Archives of Physical Medicine and Rehabilitation*, *86*(1), 134–145. https://doi.org/10.1016/j.apmr.2004.04.023

Bober, T., Kulig, K., Burnfield, J. M., & Pietraszewski, B. (2002). Predictive torque equations for joints of the extremities. *Acta of Bioengineering and Biomechanics*, *4*(2).

Bouisset, S., Richardson, J., & Zattara, M. (2000). Are amplitude and duration of anticipatory postural adjustments identically scaled to focal movement parameters in humans? *Neuroscience Letters*, *278*(3), 153–156. https://doi.org/10.1016/S0304-3940(99)00912-X

Čamernik, J., Azad, M., Peternel, L., Potocanac, Z., & Babič, J. (2019). Staying on your feet: The effectiveness of posture and handles in counteracting balance perturbation. *Ergonomics*, 1–11. https://doi.org/10.1080/00140139.2018.1559363

Čamernik, J., Potocanac, Z., Peternel, L., & Babič, J. (2016). Holding a handle for balance during continuous postural perturbations—immediate and transitionary effects on whole body posture. *Frontiers in Human Neuroscience*, *10*(September), 1–8. https://doi.org/10.3389/fnhum.2016.00486

Chaudhry, H., Bukiet, B., Ji, Z., & Findley, T. (2011). Measurement of balance in computer posturography: Comparison of methods-a brief review. *Journal of Bodywork and Movement Therapies*, *15*(1), 82–91. https://doi.org/10.1016/j.jbmt.2008.03.003

Chaudhry, H., Findley, T., Quigley, K. S., Ji, Z., Maney, M., Sims, T., Bukiet, B., & Foulds, R. (2005). Postural stability index is a more valid measure of stability than equilibrium score. *The Journal of Rehabilitation Research and Development*, *42*(4), 547. https://doi.org/10.1682/JRRD.2004.08.0097

Chen, C.-L., Lou, S.-Z., Wu, H.-W., Wu, S.-K., Yeung, K.-T., & Su, F.-C. (2014). Effects of the type and direction of support surface perturbation on postural responses. *Journal of NeuroEngineering and Rehabilitation*, *11*(1), 50. https://doi.org/10.1186/1743-0003-11-50

Chiu, J., & Robinovitch, S. N. (1998). Prediction of upper extremity impact forces during falls on the outstretched hand. *Journal of Biomechanics*, *31*(12), 1169–1176. https://doi.org/10.1016/S0021-9290(98)00137-7

Collins, J. J., & De Luca, C. J. (1994). Random walking during quiet standing. *Physical Review Letters*, *73*(5), 764–767. https://doi.org/10.1103/PhysRevLett.73.764

Corbeil, P., Bloem, B. R., van Meel, M., & Maki, B. E. (2013). Arm reactions evoked by the initial exposure to a small balance perturbation: A pilot study. *Gait & Posture*, *37*(2), 300–303. https://doi.org/10.1016/j.gaitpost.2012.07.017

Cordo, P. J., & Nashner, L. M. (1982). Properties of postural adjustments associated with rapid arm movements. *Journal of Neurophysiology*, *47*(2), 287–302. https://doi.org/10.1152/jn.1982.47.2.287

Correia, C., Lopez, K. J., Wroblewski, K. E., Huisingh-Scheetz, M., Kern, D. W., Chen, R. C., Schumm, L. P., Dale, W., McClintock, M. K., & Pinto, J. M. (2016). Global sensory impairment in older adults in the united states. *Journal of the American Geriatrics Society*, *64*(2), 306–313. https://doi.org/10.1111/jgs.13955

Creath, R., Kiemel, T., Horak, F., Peterka, R., & Jeka, J. (2005). A unified view of quiet and perturbed stance: Simultaneous co-existing excitable modes. *Neuroscience Letters*, *377*(2), 75–80. https://doi.org/10.1016/j.neulet.2004.11.071

Danna-dos-Santos, A., Degani, A. M., & Latash, M. L. (2008). Flexible muscle modes and synergies in challenging whole-body tasks. *Experimental Brain Research*, *189*(2), 171–187. https://doi.org/10.1007/s00221-008-1413-x

de Leva, P. (1996). Adjustments to zatsiorsky-seluyanov's segment inertia parameters. *Journal of Biomechanics*, *29*(9), 1223–1230. https://doi.org/10.1016/0021-9290(95)00178-6

Diener, H. C., Horak, F. B., & Nashner, L. M. (1988). Influence of stimulus parameters on human postural responses. *Journal of Neurophysiology*, *59*(6), 1888–1905. https://doi.org/10.1152/jn.1988.59.6.1888

Dietz, V., Mauritz, K.-H., & Dichgans, J. (1980). Body oscillations in balancing due to segmental stretch reflex activity. *Experimental Brain Research*, *40*(1), 37–46. https://doi.org/10.1007/BF00236666

Duncan, C. A., MacKinnon, S. N., Marais, J. F., & Basset, F. A. (2018). Energy cost associated with moving platforms. *PeerJ*, *6*(8), e5439. https://doi.org/10.7717/peerj.5439

Dunsky, A., Zeev, A., & Netz, Y. (2017). Balance performance is task specific in older adults. *BioMed Research International*, *2017*, 1–7. https://doi.org/10.1155/2017/6987017

Fox, S. (Ed.). (2011). Vestibular apparatus and equilibrium. In *Human physiology* (12th ed., pp. 275–279). McGraw-Hill Publishing Company.

Gandevia, S. C., Herbert, R. D., & Leeper, J. B. (1998). Voluntary activation of human elbow flexor muscles during maximal concentric contractions. *The Journal of Physiology*, *512*(2), 595–602. https://doi.org/10.1111/j.1469-7793.1998.595be.x

Golomer, E., Dupui, P., & Bessou, P. (1994). Spectral frequency analysis of dynamic balance in healthy and injured athletes. *Archives Internationales de Physiologie, de Biochimie et de Biophysique*, *102*(3), 225–229. https://doi.org/10.3109/13813459409007543

Golomer, E., Crémieux, J., Dupui, P., Isableu, B., & Ohlmann, T. (1999). Visual contribution to self-induced body sway frequencies and visual perception of male professional dancers. *Neuroscience Letters*, *267*(3), 189–192. https://doi.org/10.1016/S0304-3940(99)00356-0

Gurfinkel, V., Cacciatore, T. W., Cordo, P., Horak, F., Nutt, J., & Skoss, R. (2006). Postural muscle tone in the body axis of healthy humans. *Journal of Neurophysiology*, *96*(5), 2678–2687. https://doi.org/10.1152/jn.00406.2006

Hatzitaki, V., Amiridis, I. G., & Arabatzi, F. (2005). Aging effects on postural responses to self-imposed balance perturbations. *Gait & Posture*, *22*(3), 250–257. https://doi.org/10.1016/j.gaitpost.2004.09.010

Horak, F. B., & Nashner, L. M. (1986). Central programming of postural movements: Adaptation to altered support-surface configurations. *Journal of Neurophysiology*, *55*(6), arXiv arXiv:1011.1669v3, 1369–1381. https://doi.org/10.1152/jn.1986.55.6.1369

Horak, F., & Kuo, A. (2000). Postural adaptation for altered environments, tasks, and intentions. In *Biomechanics and neural control of posture and movement* (pp. 267–281). New York, NY, Springer New York. https://doi.org/10.1007/978-1-4612-2104-3_19

Horak, F. B., Henry, S. M., & Shumway-Cook, A. (1997). Postural perturbations: New insights for treatment of balance disorders. *Physical therapy*, *77*(5), 517–33.

Horak, F. B., & Macpherson, J. M. (1996). Postural orientation and equilibrium. In *Handbook of physiology. exercise: Regulation and integration of multiple systems* (pp. 255–291). Hoboken, NJ, USA, Oxford University Press.

Hsiao-Wecksler, E. T., Katdare, K., Matson, J., Liu, W., Lipsitz, L. A., & Collins, J. J. (2003). Predicting the dynamic postural control response from quiet-stance behavior in elderly adults. *Journal of Biomechanics*, *36*(9), 1327–1333. https://doi.org/10.1016/S0021-9290(03)00153-2

Huang, H. J., Kram, R., & Ahmed, A. A. (2012). Reduction of metabolic cost during motor learning of arm reaching dynamics. *Journal of Neuroscience*, *32*(6), 2182–2190. https://doi.org/10.1523/JNEUROSCI.4003-11.2012

Huang, H. J., & Ahmed, A. A. (2011). Tradeoff between stability and maneuverability during whole-body movements (A. Lucia, Ed.). *PLoS ONE*, *6*(7), e21815. https://doi.org/10.1371/journal.pone.0021815

IJmker, T., Lamoth, C. J., Houdijk, H., Tolsma, M., van der Woude, L. H. V., Daffertshofer, A., & Beek, P. J. (2015). Effects of handrail hold and light touch on energetics, step parameters, and neuromuscular activity during walking after stroke. *Journal of NeuroEngineering and Rehabilitation*, *12*(1), 70. https://doi.org/10.1186/s12984-015-0051-3

IJmker, T., Lamoth, C. J., Houdijk, H., van der Woude, L. H., & Beek, P. J. (2014). Postural threat during walking: Effects on energy cost and accompanying gait changes. *Journal of NeuroEngineering and Rehabilitation*, *11*(1), 71. https://doi.org/10.1186/1743-0003-11-71

Jeka, J. J., & Lackner, J. R. (1995). The role of haptic cues from rough and slippery surfaces in human postural control. *Experimental brain research*, *103*(2), 267–76.

Jeka, J. J. (1997). Light touch contact as a balance aid. *Physical Therapy*, *77*(5), 476–487. https://doi.org/10.1093/ptj/77.5.476

Jeka, J., & Lackner, J. (1994). Fingertip contact influences human postural control. *Experimental Brain Research*, *79*(2), 495–502. https://doi.org/10.1007/BF00229188

Jensen, J. L., Brown, L. A., & Woollacott, M. H. (2001). Compensatory stepping: The biomechanics of a preferred response among older adults. *Experimental Aging Research*, *27*(4), 361–376. https://doi.org/10.1080/03610730109342354

Johannsen, L., Wing, A. M., & Hatzitaki, V. (2007). Effects of maintaining touch contact on predictive and reactive balance. *Journal of Neurophysiology*, *97*(4), 2686–2695. https://doi.org/10.1152/jn.00038.2007

Kouzaki, M., & Masani, K. (2008). Reduced postural sway during quiet standing by light touch is due to finger tactile feedback but not mechanical support. *Experimental Brain Research*, *188*(1), 153–158. https://doi.org/10.1007/s00221-008-1426-5

Krishnamoorthy, V., Latash, M. L., Scholz, J. P., & Zatsiorsky, V. M. (2004). Muscle modes during shifts of the center of pressure by standing persons: Effect of instability and additional support. *Experimental Brain Research*, *157*(1), 18–31. https://doi.org/10.1007/s00221-003-1812-y

Krishnamoorthy, V., Slijper, H., & Latash, M. L. (2002). Effects of different types of light touch on postural sway. *Experimental Brain Research*, *147*(1), 71–79. https://doi.org/10.1007/s00221-002-1206-6

Lackner, J., Rabin, E., & DiZio, P. (2001). Stabilization of posture by precision touch of the index finger with rigid and flexible filaments. *Experimental Brain Research*, *139*(4), 454–464. https://doi.org/10.1007/s002210100775

Latash, M. L., & Zatsiorsky, V. M. (2015). *Biomechanics and motor control: Defining central concepts*. Elsevier Inc. https://doi.org/10.1016/C2013-0-18342-0

Lepers, R., Bigard, A. X., Diard, J.-P., Gouteyron, J.-F., & Guezennec, C. Y. (1997). Posture control after prolonged exercise. *European Journal of Applied Physiology*, *76*(1), 55–61. https://doi.org/10.1007/s004210050212

Li, X., & Aruin, A. S. (2009). The effect of short-term changes in body mass distribution on feed-forward postural control. *Journal of Electromyography and Kinesiology*, *19*(5), 931–941. https://doi.org/10.1016/j.jelekin.2008.05.003

Macpherson, J., Horak, F., Dunbar, D., & Dow, R. (1990). Stance dependence of automatic postural adjustments in humans. *Experimental Brain Research*, *80*(2), 291–299. https://doi.org/10.1007/BF00228173

Maki, B. E., & McIlroy, W. E. (1997). The role of limb movements in maintaining upright stance: The "change-in-support"strategy. *Physical therapy*, *77*(5), 488–507.

Maki, B. E., Perry, S. D., & McIlroy, W. E. (1998). Efficacy of handrails in preventing stairway falls: A new experimental approach. *Safety Science*, *28*(3), 189–206. https://doi.org/10.1016/S0925-7535(98)80008-8

Maki, B. E., McIlroy, W. E., & Fernie, G. R. (2003). Change-in-support reactions for balance recovery. *IEEE Engineering in Medicine and Biology Magazine*, *22*(2), 20–26. https://doi.org/10.1109/MEMB.2003.1195691

Mansfield, A., & Maki, B. E. (2009). Are age-related impairments in change-in-support balance reactions dependent on the method of balance perturbation? *Journal of Biomechanics*, *42*(8), 1023–1031. https://doi.org/10.1016/j.jbiomech.2009.02.007

McIlroy, W. E., & Maki, B. E. (1995). Early activation of arm muscles follows external perturbation of upright stance. *Neuroscience Letters*, *184*(3), 177–180. https://doi.org/10.1016/0304-3940(94)11200-3

McIlroy, W. E., & Maki, B. E. (1996). Age-related changes in compensatory stepping in response to unpredictable perturbations. *Journals of Gerontology - Series A Biological Sciences and Medical Sciences*, *51*(6), 289–296. https://doi.org/10.1093/gerona/51A.6.M289

Mergner, T., Maurer, C., & Peterka, R. J. (2003). A multisensory posture control model of human upright stance. *Progress in Brain Research*, *142*(1), 189–201. https://doi.org/10.1016/S0079-6123(03)42014-1

Mille, M. L., Johnson-Hilliard, M., Martinez, K. M., Zhang, Y., Edwards, B. J., & Rogers, M. W. (2013). One step, two steps, three steps more... directional vulnerability to falls in community-dwelling older people. *Journals of Gerontology - Series A Biological Sciences and Medical Sciences*, *68*(12 A), 1540–1548. https://doi.org/10.1093/gerona/glt062

Moraux, A., Canal, A., Ollivier, G., Ledoux, I., Doppler, V., Payan, C., & Hogrel, J.-Y. (2013). Ankle dorsi- and plantar-flexion torques measured by dynamometry in healthy subjects from 5 to 80 years. *BMC Musculoskeletal Disorders*, *14*(1), 104. https://doi.org/10.1186/1471-2474-14-104

Nagy, E., Toth, K., Janositz, G., Kovacs, G., Feher-Kiss, A., Angyan, L., & Horvath, G. (2004). Postural control in athletes participating in an ironman triathlon. *European Journal of Applied Physiology*, *92*(4-5), 407–413. https://doi.org/10.1007/s00421-004-1157-7

Nardone, A., Tarantola, J., Galante, M., & Schieppati, M. (1998). Time course of stabilometric changes after a strenuous treadmill exercise. *Archives of Physical Medicine and Rehabilitation*, *79*(8), 920–924. https://doi.org/10.1016/S0003-9993(98)90088-0

Nashner, L., & Cordo, P. (1981). Relation of automatic postural responses and reaction-time voluntary movements of human leg muscles. *Experimental Brain Research*, *43-43*(3-4), 395–405. https://doi.org/10.1007/BF00238382

Oppenheim, U., Kohen-Raz, R., Alex, D., Kohen-Raz, A., & Azarya, M. (1999). Postural characteristics of diabetic neuropathy. *Diabetes Care*, *22*(2), 328–332. https://doi.org/10.2337/diacare.22.2.328

Osoba, M. Y., Rao, A. K., Agrawal, S. K., & Lalwani, A. K. (2019). Balance and gait in the elderly: A contemporary review. *Laryngoscope Investigative Otolaryngology*, *4*(1), 143–153. https://doi.org/10.1002/lio2.252

O'Sullivan, S. B., Schmitz, T. J., & Fulk, G. (2014). *Physical rehabilitation* (6th ed.). F.A. Davis Company.

Pai, Y.-C., Maki, B., Iqbal, K., McIlroy, W., & Perry, S. (2000). Thresholds for step initiation induced by support-surface translation: A dynamic center-of-mass model provides much better prediction than a static model. *Journal of Biomechanics*, *33*(3), 387–392. https://doi.org/10.1016/S0021-9290(99)00199-2

Park, S., Horak, F. B., & Kuo, A. D. (2004). Postural feedback responses scale with biomechanical constraints in human standing. *Experimental Brain Research*, *154*(4), 417–427. https://doi.org/10.1007/s00221-003-1674-3

Peterka, R. J. (2002). Sensorimotor integration in human postural control. *Journal of Neurophysiology*, *88*(3), 1097–1118. https://doi.org/10.1152/jn.2002.88.3.1097

Peterka, R. J. (2018). Sensory integration for human balance control. In *Handbook of clinical neurology* (1st ed., pp. 27–42). Elsevier B.V. https://doi.org/10.1016/B978-0-444-63916-5.00002-1

Peterka, R. J., & Loughlin, P. J. (2004). Dynamic regulation of sensorimotor integration in human postural control. *Journal of Neurophysiology*, *91*(1), 410–423. https://doi.org/10.1152/jn.00516.2003

Peternel, L., & Babič, J. (2013). Learning of compliant human–robot interaction using full-body haptic interface. *Advanced Robotics*, *27*(13), 1003–1012. https://doi.org/10.1080/01691864.2013.808305

Phinyomark, A., Thongpanja, S., Hu, H., Phukpattaranont, P., & Limsakul, C. (2012). The usefulness of mean and median frequencies in electromyography analysis. In *Computational intelligence in electromyography analysis - a perspective on current applications and future challenges*. InTech. https://doi.org/10.5772/50639

Pidcoe, P. E., & Rogers, M. W. (1998). A closed-loop stepper motor waist-pull system for inducing protective stepping in humans. *Journal of Biomechanics*, *31*(4), 377–381. https://doi.org/10.1016/S0021-9290(98)00017-7

Pijnappels, M., Reeves, N. D., Maganaris, C. N., & van Dieën, J. H. (2008). Tripping without falling; lower limb strength, a limitation for balance recovery and a target for training in the elderly. *Journal of Electromyography and Kinesiology*, *18*(2), 188–196. https://doi.org/10.1016/j.jelekin.2007.06.004

Rabin, E., DiZio, P., Ventura, J., & Lackner, J. R. (2008). Influences of arm proprioception and degrees of freedom on postural control with light touch feedback. *Journal of Neurophysiology*, *99*(2), 595–604. https://doi.org/10.1152/jn.00504.2007

Robertson, D. G. E., Caldwell, G. E., Hamill, J., Kamen, G., & Whittlesey, S. N. (2014). *Research methods in biomechanics* (2nd ed.). Human Kinetics.

Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., Sims-Gould, J., & Loughin, M. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: An observational study. *The Lancet*, *381*(9860), 47–54. https://doi.org/10.1016/S0140-6736(12)61263-X

Rogers, M. W., Wardman, D. L., Lord, S. R., & Fitzpatrick, R. C. (2001). Passive tactile sensory input improves stability during standing. *Experimental Brain Research*, *136*(4), 514–522. https://doi.org/10.1007/s002210000615

Rueckert, E., Čamernik, J., Peters, J., & Babič, J. (2016). Probabilistic movement models show that postural control precedes and predicts volitional motor control. *Scientific Reports*, *6*(1), 28455. https://doi.org/10.1038/srep28455

Runge, C., Shupert, C., Horak, F., & Zajac, F. (1999). Ankle and hip postural strategies defined by joint torques. *Gait & Posture*, *10*(2), 161–170. https://doi.org/10.1016/S0966-6362(99)00032-6

Santos, M. J., & Aruin, A. S. (2008). Role of lateral muscles and body orientation in feedforward postural control. *Experimental Brain Research*, *184*(4), 547–559. https://doi.org/10.1007/s00221-007-1123-9

Santos, M. J., Kanekar, N., & Aruin, A. S. (2010a). The role of anticipatory postural adjustments in compensatory control of posture: 1. electromyographic analysis. *Journal of Electromyography and Kinesiology*, *20*(3), arXiv NIHMS150003, 388–397. https://doi.org/10.1016/j.jelekin.2009.06.006

Santos, M. J., Kanekar, N., & Aruin, A. S. (2010b). The role of anticipatory postural adjustments in compensatory control of posture: 2. biomechanical analysis. *Journal of Electromyography and Kinesiology*, *20*(3), 398–405. https://doi.org/10.1016/j.jelekin.2010.01.002

Sarabon, N., Rosker, J., Loefler, S., & Kern, H. (2010). Sensitivity of body sway parameters during quiet standing to manipulation of support surface size. *Journal of sports science & medicine*, *9*(3), 431–8.

Scholz, J. P., Schöner, G., Hsu, W. L., Jeka, J. J., Horak, F., & Martin, V. (2007). Motor equivalent control of the center of mass in response to support surface perturbations. *Experimental Brain Research*, *180*(1), 163–179. https://doi.org/10.1007/s00221-006-0848-1

Scholz, J. P., & Schöner, G. (1999). The uncontrolled manifold concept: Identifying control variables for a functional task. *Experimental Brain Research*, *126*(3), 289–306. https://doi.org/10.1007/s002210050738

Selinger, J. C., O'Connor, S. M., Wong, J. D., & Donelan, J. M. (2015). Humans can continuously optimize energetic cost during walking. *Current Biology*, *25*(18), 2452–2456. https://doi.org/10.1016/j.cub.2015.08.016

Shiratori, T., & Aruin, A. (2007). Modulation of anticipatory postural adjustments associated with unloading perturbation: Effect of characteristics of a motor action. *Experimental Brain Research*, *178*(2), 206–215. https://doi.org/10.1007/s00221-006-0725-y

Shumway-Cook, A., & Woollacott, M. (2012). *Motor control : Translating research into clinical practice* (4th ed.). Wolters Kluwer Health : Lippincott Williams & Wilkins.

Slijper, H., & Latash, M. (2000). The effects of instability and additional hand support on anticipatory postural adjustments in leg, trunk, and arm muscles during standing. *Experimental Brain Research*, *135*(1), 81–93. https://doi.org/10.1007/s002210000492

Szturm, T., & Fallang, B. (1998). Effects of varying acceleration of platform translation and toes-up rotations on the pattern and magnitude of balance reactions in humans. *Journal of vestibular research : equilibrium & orientation*, *8*(5), 381–97.

Ting, L. H. (2007). Dimensional reduction in sensorimotor systems: A framework for understanding muscle coordination of posture. In *Progress in brain research* (pp. 299–321). https://doi.org/10.1016/S0079-6123(06)65019-X

Ting, L. H., & Macpherson, J. M. (2005). A limited set of muscle synergies for force control during a postural task. *Journal of Neurophysiology*, *93*(1), 609–613. https://doi.org/10.1152/jn.00681.2004

Todorov, E., Erez, T., & Tassa, Y. (2012). Mujoco: A physics engine for model-based control, In *2012 ieee/rsj international conference on intelligent robots and systems*, IEEE. https://doi.org/10.1109/IROS.2012.6386109

Todorov, E., & Jordan, M. I. (2002). Optimal feedback control as a theory of motor coordination. *Nature Neuroscience*, *5*(11), 1226–1235. https://doi.org/10.1038/nn963

Torres-Oviedo, G., Macpherson, J. M., & Ting, L. H. (2006). Muscle synergy organization is robust across a variety of postural perturbations. *Journal of Neurophysiology*, *96*(3), 1530–1546. https://doi.org/10.1152/jn.00810.2005

Torres-Oviedo, G., & Ting, L. H. (2007). Muscle synergies characterizing human postural responses. *Journal of Neurophysiology*, *98*(4), arXiv 1612.05467, 2144–2156. https://doi.org/10.1152/jn.01360.2006

Tresch, M. C., & Jarc, A. (2009). The case for and against muscle synergies. *Current Opinion in Neurobiology*, *19*(6), arXiv NIHMS150003, 601–607. https://doi.org/10.1016/j.conb.2009.09.002

United Nations. (2017). *World population ageing 2017 - highlights*. New York, NY, United Nations, Department of Economic; Social Affairs, Population Division.

van den Kroonenberg, A. J., Hayes, W. C., & McMahon, T. A. (1996). Hip impact velocities and body configurations for voluntary falls from standing height. *Journal of Biomechanics*, *29*(6), 807–811. https://doi.org/10.1016/0021-9290(95)00134-4

Vieira, T., Oliveira, L., & Nadal, J. (2009). Estimation procedures affect the center of pressure frequency analysis. *Brazilian Journal of Medical and Biological Research*, *42*(7), 665–673. https://doi.org/10.1590/S0100-879X2009000700012

Wing, A. M., Flanagan, J. R., & Richardson, J. (1997). Anticipatory postural adjustments in stance and grip. *Experimental Brain Research*, *116*(1), 122–130. https://doi.org/10.1007/PL00005732

Wing, A. M., Johannsen, L., & Endo, S. (2011). Light touch for balance: Influence of a time-varying external driving signal. *Philosophical Transactions of the Royal Society B: Biological Sciences*, *366*(1581), 3133–3141. https://doi.org/10.1098/rstb.2011.0169

Winter, D. A. (1995). Human balance and posture standing and walking. *Gait and Posture*, *3*, 193–214.

Winter, D. A. (2009). *Biomechanics and motor control of human movement*. Hoboken, NJ, USA, John Wiley & Sons, Inc. https://doi.org/10.1002/9780470549148

Wojtara, T., Alnajjar, F., Shimoda, S., & Kimura, H. (2014). Muscle synergy stability and human balance maintenance. *Journal of NeuroEngineering and Rehabilitation*, *11*(1), 1–9. https://doi.org/10.1186/1743-0003-11-129

World Health Organization. (2007). Magnitide of falls - a worldwide overview. *WHO Global Report on Falls Prevention in Older Age.*, 49.

Žlajpah, L., & Babič, J. (2014). *Bodymodel 1.0 quick user manual* (tech. rep.).

Bibliography

Publications related to the Thesis

Journal Articles

Čamernik, J., Azad, M., Peternel, L., Potocanac, Z., & Babič, J. (2019). Staying on your feet: The effectiveness of posture and handles in counteracting balance perturbation. *Ergonomics*, 1–11. https://doi.org/10.1080/00140139.2018.1559363

Čamernik, J., Potocanac, Z., Peternel, L., & Babič, J. (2016). Holding a handle for balance during continuous postural perturbations—immediate and transitionary effects on whole body posture. *Frontiers in Human Neuroscience*, *10*(September), 1–8. https://doi.org/10.3389/fnhum.2016.00486

Rueckert, E., Čamernik, J., Peters, J., & Babič, J. (2016). Probabilistic movement models show that postural control precedes and predicts volitional motor control. *Scientific Reports*, *6*(1), 28455. https://doi.org/10.1038/srep28455

Published scientific conference contribution

Čamernik, J., Peternel, L., & Babič, J. (2014). Role of hand contact in continually challenged postural equilibrium (B. Zajc & A. Trost, Eds.). In B. Zajc & A. Trost (Eds.), *Zbornik triindvajsete mednarodne elektrotehniške in računalniške konference erk 2014, 22. - 24. september 2014, portorož, slovenija*, Ljubljana, IEEE Region 8, Slovenska sekcija IEEE.

Published scientific conference contribution abstract

Čamernik, J., Peternel, L., & Babič, J. (2015a). The role of additional arm support when used in an active balansing task, In *Progress in motor control x : 22-25 july, 2015, budapest, hungary : Program book*, Budapest, Hungarian Society of Sport Science.

Čamernik, J., Peternel, L., & Babič, J. (2015b). Role of hand contact in continually challenged postural equilibrium, In *Ispgr world congress, june 28 - july 2, 2015, seville, spain : Program.* Seville.

Other Publications

Journal Articles

Maurice, P., Čamernik, J., Gorjan, D., Schirrmeister, B., Bornmann, J., Tagliapietra, L., Latella, C., Pucci, D., Fritzsche, L., Ivaldi, S., & Babič, J. (2019). Objective and subjective effects of a passive exoskeleton on overhead work. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 1–1. https://doi.org/10.1109/TNSRE.2019.2945368

Romano, F., Nava, G., Azad, M., Čamernik, J., Dafarra, S., Dermy, O., Latella, C., Lazzaroni, M., Lober, R., Lorenzini, M., Pucci, D., Sigaud, O., Traversaro, S., Babic, J., Ivaldi, S., Mistry, M., Padois, V., & Nori, F. (2018). The codyco project achievements and beyond: Toward human aware whole-body controllers for physical human robot interaction. *IEEE Robotics and Automation Letters*, *3*(1), 516–523. https://doi.org/10.1109/LRA.2017.2768126

Published scientific conference contribution

Čamernik, J., Oztop, E., & Babič, J. (2018). Understanding human adaptation in squat-to-stand movements., In *2018 türkiye robotbilim konferansi, tork 2018 boğaziçi üniversitesi, istanbul, 12-14 nisan, 2018.*

Maurice, P., Čamernik, J., Gorjan, D., Schirrmeister, B., Bornmann, J., Tagliapietra, L., Pucci, D., Ivaldi, S., & Babič, J. (2019). Objective and subjective effects of passive exoskeleton on overhead work, In *Proceedings, 9th international conference on safety of industrial automated systems, sias 2018, 10-12 october, nancy, france*, Nancy.

Published scientific conference contribution abstract

Čamernik, J., Kezić, S., & Babič, J. (2019). Threat-related changes in postural control in virtual environments, In *2019 isprg world congress, june 30 to july 4, edinburgh, scotland : Abstract book*, Edinburgh.

Čamernik, J., Rueckert, E., Oztop, E., & Babič, J. (2018). Configurable dynamical environment simulation platform for studies of whole- body motor control and learning, In *Congress programme, 8th world congress of biomechanics, 8-12 july 2018, dublin, ireland*, Dublin.

Gorjan, D., Bellicha, A., Čamernik, J., Bachta, W., & Babič, J. (2019). Postural control during induced stabilization of the center of mass and light touch, In *2019 isprg world congress, june 30 to july 4, edinburgh, scotland : Abstract book*, Edinburgh.

Unpublished scientific conference contribution abstract

Arditi, E., Čamernik, J., Babič, J., Ugur, E., Nagai, Y., & Oztop, E. (2019). Explorations on inverse reinforcement learning for the analysis of sensorimotor data, In *Winter workshop on mechanism of brain and mind 2019, rusutsu, 9.-11.1. 2019*, Rusutsu.

Kunavar, T., Čamernik, J., Babič, J., Oztop, E., & Kawato, M. (2019). Does danger of injury influence human motor adaptation? In *Winter workshop on mechanism of brain and mind 2019, rusutsu, 9.-11.1. 2019*, Rusutsu.

Maurice, P., Čamernik, J., Gorjan, D., Schirrmeister, B., Bornmann, J., Tagliapietra, L., Pucci, D., Ivaldi, S., & Babič, J. (2018). Objective and subjective effects of passive exoskeleton on overhead work, In *Jnrh 2018, journées nationales de la robotique humanoide 2018, 4,15 juin 2018, nancy, france*, Nancy.

Biography

Jernej Čamernik was born on July 8, 1979, in Celje, Slovenia. He finished his primary education in Radeče, and secondary education at the Srednja ekonomska šola Celje. In 2007, he graduated at the University of Ljubljana, Faculty of Sport, with a thesis entitled "Comparison of kinematic parameters of the sprint start and block acceleration". In the first years after graduation he worked at the Biomechanics lab at the Institute of Sport - Faculty of Sport in Ljubljana. As a lab assistant, he was conducting and analyzing results of biomechanical and physiological studies mostly carried out on top Slovenian athletes.

In January 2014, he joined the team of Prof. Dr. Jan Babič at the Jožef Stefan Institute - Department of Automation, Biocybernetics and Robotics, Ljubljana, Slovenia, and in October, same year, he started his Ph.D. studies at the Jožef Stefan International Postgraduate School, Ljubljana, Slovenia. During his PhD studies, he was mainly carrying out research activities within the scope of the EU project CoDyCo (Whole-body Compliant Dynamical Contacts in Cognitive Humanoids) and since 2017 within the EU projects AnDy (Advancing Anticipatory Behaviors in Dyadic Human-Robot Collaboration) and Spexor (Spinal exoskeletal robot for low back pain prevention and vocational reintegration).

Within the scope of the CoDyCo project, he was conducting human behavior experiments, with the focus on human balance in balance challenging conditions, which were a baseline of his PhD work. During the PhD studies, he attended several conferences on the topics of human motor control, biomechanics and ergonomics and attended a Motor Control Summer School. As a result of his work at JSI, he wrote/co-wrote five peer-reviewed papers and presented his work at several international conferences. A common topic of most of his papers and the doctoral thesis is human postural control in balance challenging situations. Specifically, he focused on changes in full-body postural adaptations, motor learning and mechanisms of postural control augmented with hand contacts made with the surrounding environment. Recently he also successfully transferred his research expertise and skills to applied research studies. Namely, he has been working on biomechanical assessment and validation of lightweight upper-body exoskeletons used in car manufacturing.