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Postural Control and Consciousness and Their Applications to Lower-Limb Exoskeletons

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par

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To my parents, brother, grand-mother and friends, for their time, energy and love.

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Abstract

Human postural control requires high coordination and is distributed over the whole musculoskeletal and nervous systems. Hence, postural balance can be impaired due to various pathologies. Efficient rehabilitation programs and technological solutions are therefore needed to alleviate the burden due to the risk of falls and lack of mobility. In this thesis, we have sought solutions to improve postural balance with two different approaches. One approach is based on the application of neuroscience principles, while the other relies on engineering.

With the emergence of so-called exergames for balance training, it is essential to understand how the central nervous system integrates augmented visual feedback. Research in motor learning has demonstrated the need for experiencing errors to adapt to the environment and learn new skills. Hence, studies have proposed to visually amplify trajectory error to promote learning. As this strategy demonstrated promising results for upper limb rehabilitation, we investigated whether visual error augmentation promotes motor adaptation during a balance task. Based on the observer model, we also investigated whether there is a link between sensorimotor learning and a conscious experience of movements known as the sense of agency. Our results suggest that visual error augmentation may promote balance learning and that the optimal gain may be influenced by the sense of agency. Moreover, our results illustrate that humans monitor their actions in an effector-independent manner. Although motor learning is different from neurorehabilitation, these studies help to understand cognitive components of motor learning that are of scientific and clinical relevance.

Patients with severe balance deficits need engineered solutions to stand up and walk. We have developed a lower-limb exoskeleton, called TWIICE, with the goal to give back autonomy to people with a complete spinal cord injury. Most exoskeletons require that the user manages balance while walking or standing with the help of crutches. To facilitate the interaction with the environment while standing, we developed the first postural controllers enabling patients with complete paraplegia to stand in a low-actuator-count exoskeleton without the need for crutches. This could have important implications for the independence of individuals

with paraplegia, their inclusion in social activities and their potential inclination to use an exoskeleton on a daily basis for the associated health benefits.

Keywords: motor learning, postural control, motor awareness, visual feedback, exoskeleton, bodily self-consciousness, robotics

Résumé

Le contrôle postural humain exige une bonne coordination sur l'ensemble des systèmes musculosquelettique et nerveux. Par conséquent, l'équilibre postural peut être détérioré par diverses pathologies. Des programmes de rééducation efficaces et des solutions technologiques sont donc nécessaires pour alléger la charge due au risque de chute et au manque de mobilité. Dans cette thèse, j'ai cherché des solutions pour améliorer l'équilibre postural en utilisant deux approches différentes. L'une est basée sur l'application de principes neuroscientifiques, et l'autre sur l'ingénierie.

Avec l'émergence d'*exergames* pour l'entraînement à l'équilibre, il est essentiel de comprendre comment le système nerveux central intègre un feedback visuel augmenté. Les recherches sur l'apprentissage moteur ont démontré la nécessité des erreurs de mouvement pour s'adapter à l'environnement et acquérir de nouvelles compétences. Ainsi, des études ont proposé d'amplifier visuellement les erreurs de trajectoire pour favoriser l'apprentissage. Cette stratégie a donné des résultats prometteurs pour la rééducation des membres supérieurs, et nous avons donc étudié si l'augmentation de l'erreur visuelle favorise l'adaptation motrice lors d'une tâche d'équilibre. En nous basant sur le modèle de l'observateur, nous avons également étudié s'il existe un lien entre l'apprentissage sensorimoteur et l'expérience consciente du mouvement, appelée agentivité. Nos résultats suggèrent qu'augmenter l'erreur visuelle peut favoriser l'apprentissage de l'équilibre et que le gain optimal peut être influencé par l'agentivité. De plus, nos résultats montrent que les humains contrôlent leur action d'une manière indépendante des effecteurs. Bien que l'apprentissage moteur soit différent de la neuroréhabilitation, ces études aident à comprendre les composantes cognitives de l'apprentissage moteur et les applications cliniques.

Les patients souffrant de graves déficits d'équilibre ont besoin de solutions techniques pour se lever et marcher. Nous avons développé un exosquelette des membres inférieurs, appelé TWIICE, pour donner de l'autonomie aux personnes souffrant d'une lésion complète de la moelle épinière. La plupart des exosquelettes exigent que l'utilisateur gère son équilibre en marchant ou en se maintenant debout à l'aide de béquilles. Pour faciliter l'interaction avec l'environnement en position debout, nous avons développé les premiers contrôleurs posturaux permettant aux patients atteints de paraplégie complète de se tenir debout dans un exosquelette comportant peu d'actionneurs sans béquilles. Cela peut améliorer l'indépendance des personnes ayant une paraplégie, leur inclusion dans la société et leur propension à utiliser régulièrement un exosquelette, avec les bénéfices de santé associés.

Mots clés : apprentissage moteur, contrôle postural, conscience motrice, feedback visuel, exosquelette, conscience corporelle, robotique

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List of acronyms

AR augmented reality

BCI brain-computer interface

BoS base of support

BSC bodily self-conciousness

CNS central nervous system

CoM center of mass

CoP center of pressure

DoF degree of freedom

DoFs degrees of freedom

EA error augmentation

FPS frames per second

IMU inertial measurement unit

LED light-emitting diode

LCD liguid crystal display

Li-ion Lithium-ion

Li-Po Lithium-polymer

MA motor awareness

MAI motor awareness index

MP motor performance

PD proportional-derivative

PI proportional-integral

PID proportional-integral-derivative

PCB printed circuit board

PSE point of subjective equality

PTE peak trajectory error

RGB red-green-blue

RMS root mean square

SCI spinal cord injury

SNR signal-to-noise ratio

SoA sense of agency

UI user interface

UX user experience

VR virtual reality

2

1 Introduction

1.1 Motivations and goals

Humans demonstrate of outstanding coordination to maintain their balance under extreme conditions such as walking on a high-wire, in challenging postures such as during gymnastic or even when interacting with a dynamic object such when juggling with a football [1]. More commonly, having a good balance allows us, among other things, to walk without tripping, bend over to pick something up, stand up from a chair, or simply get dressed. Those basic functions are of tremendous importance for physical independence as they are integral to our daily activities. However, since postural control is distributed throughout the nervous and musculoskeletal systems, balance disorders are a common consequence of pathologies ranging from multiple sclerosis to stroke, spinal cord injury, Parkinson's disease, diabetes, and natural aging [2, 3, 4, 5]. Balance deficits are the most common causes of falls, and often lead to severe injuries, disability and limited quality of life [6], and thus are a major health issue.

In this thesis, we have researched the mechanisms of balance control by tackling optimization of balance motor learning through error augmentation (Chapter 3 and 5), by investigating its potential links with conscious motor control and awareness (Chapter 4 and 5), and by providing wearable robotic solutions based on a bioinspired approach for complete spinal cord injured patients (Chapter 6).

1.1.1 Optimizing balance motor learning

Improving motor learning is a common research topic since progressing as fast as possible is a desirable outcome in different fields. In sports, an ice skater wants to jump higher or rotate faster. Different means are used to help them learn these complex skills. A coach will demonstrate the correct movements (observational learning [7]) or give vocal instructions at the start of a jump to help with the correct timing (auditory or verbal feedback [8]). Sometimes technical displays will contribute to enhance learning. Video analysis of the athlete's performance (terminal visual feedback [9]) is a common method to highlight errors. Physical guidance (haptic feedback [10]) is also sometimes provided in the form of a harness, allowing to improve jumps and fall less often. All of these techniques may be combined, with the aim to increase the learning rate.

Similar methods are used to recover functional abilities, essential to everyday life activities, for patients with impaired balance. Therapists, to some extent, play the role of a sport trainer, and the patient likely wants to recover and learn as much and as fast as possible [11]. Therefore, *research aims to optimize instructions and feedback to improve motor learning.*

Technical progress allows the investigation of motor tasks in more ecological fashion through the implementation of visual, auditory, haptic or multimodal augmented feedback. The efficacy of unimodal and multimodal augmented feedback in the framework of motor learning has been thoroughly reviewed by Sigrist and colleagues [11]. They report that there is presently no consensus on the optimal way of providing efficient augmented feedback specifically for tasks and individuals. To pave the way towards personalized medicine, one needs to know which individual characteristics better predict the training outcomes.

The **first part** of the manuscript introduces the fundamentals of motor control, learning and awareness, as well as their related work. The **second part** of this thesis reports on *visual augmented feedback based on error augmentation for balance training.* Research has shown that error detection and correction are needed to enhance motor learning [12, 13, 14]. For this reason, several studies have used and tested arm reaching tasks that propose to amplify the magnitude of such artificially introduced errors in order to stimulate the error-driven adaptation and acquisition process in the brain. For example, it has been demonstrated that visual error augmentation can change movement patterns of healthy subjects [15, 16, 17] and trigger functional improvement in stroke patients [18]. Although balance training with visual feedback is getting popular with exergames on the Wii Balance Board [19], with the MindMotion Go [20] or on the Balance Master System (NeuroCom International Inc., Clackamas, OR, USA), most of the past research involving visual distortion such as error augmentation have been done in upper limbs. Only recently, studies started to transfer this paradigm to gait [21]. Despite the use of visual feedback for balance training, visual error augmentation dedicated for balance tasks remains little explored and, therefore, research on its effect may be of interest for the scientific and clinical communities.

In addition to transferring an error augmentation paradigm to balance studies, we were also driven by the idea of *predicting the optimal error augmentation gain* and personalization (i.e. to provide individual-tailored training). We hypothesize that EA gain, which depends on general factors such as task complexity and individual skill level [22, 23, 24, 25], may further be linked to conscious action monitoring. It has been demonstrated that humans perform unconscious motor corrections up to a certain level sensorimotor mismatch, called the motor awareness threshold [26, 27, 28, 29]. A lingering question in motor learning is whether those motor corrections have to remain unconscious (implicit learning) or consciously processed (explicit learning) to promote movement automaticity [30]. Error amplification therefore introduces an interesting conundrum: if the error augmentation (EA) gain is used to push the errors over the motor awareness (MA) threshold, will this be beneficial or detrimental for motor learning? Although the literature suggests that implicit learning favors automaticity [30] while explicit learning promotes skill transfer [31], the ideal balance between conscious adaptation and implicit learning so that it is automated and retained is still unclear. The EA gain might need to be adapted to skill or impairment to force a cognitive or automatic movement strategy.
Consequently, we were interested to *study full-body motor awareness* (i.e. explicit awareness about an artificially introduced error between the performed body tilting movement and the received visual feedback) during a goal-directed balance task [32], and investigate whether the level of motor awareness is a suitable predictor of the optimal error augmentation gain or reflects a difference in motor learning ability.
Furthermore, despite the fact that a fundamental aspect of bodily self-consciousness is that the self is defined as the single coherent representation of the entire body rather than separated body parts [33, 34], which has been experimentally verified in [35, 36], our task provided a new paradigm and unique insights into the mechanisms of full-body motor awareness, which have been less explored than distal body parts motor awareness such as fingers and arms.

1.1.2 Towards the use of lower-limb exoskeletons

The life experience of about 65 million persons worldwide is impacted by severe mobility impairment [37], which often leads to the necessary use of a wheelchair on a daily basis. It is usually associated with a major loss of independence [38], negatively affects social integration, leads to unemployment and loss of vocational motivation [39], and alters perception of self or induces a feeling of difference [40, 41]. The lack of mobility due to impaired postural control is also the cause of important but often disregarded secondary health issues such as loss of bone mineral density, bowel, bladder and sexual dysfunctions, musculoskeletal disorders, immunodeficiency, and pressure ulcers [42].

With the aim to provide solutions to the burden associated with loss of mobility, wearable assistive devices have been proposed among other technological solutions. They consist of mechanical components attached to patient's body parts that temporarily replace or assist the function of the affected limb. They can be actuated or passive, soft (fabric-based suit) or rigid. The rigid ones are referred to as exoskeletons. Despite the fact that their potential in walking assistance for motor complete spinal cord injury (SCI) patients and as a tool for neurorehabilitation has successfully been demonstrated [43], developing novel features to improve user safety, mobility and autonomy is still a constant research challenge. In 2015, we developed a full-mobilization exoskeleton called TWIICE [44] with the goal to give people with complete SCI some degree of autonomy. While the main targeted function for exoskeletons is walking, usually with the help of crutches, we were interested in the development of a postural controller that would enable the use of the hands while standing, and thus facilitate the user interaction with their environment.
Hence, the **third part** focuses on the *balance management in lower-limb exoskeletons* with two degrees of freedom per leg - actuation at the hip and knee joints. We adopted a bioinspired approach including identification of postural control strategies in young able-bodied adults constrained by a passive exoskeleton called INSPIIRE, development of two controllers, and performance characterization with a complete spinal cord injured pilot.

1.2 Thesis at a glance

Chapter 2 introduces the general framework of motor control, motor learning and motor awareness, and how augmented feedback based on error augmentation affects motor learning. Then, a summary of the existing balance rehabilitation therapy methods is provided, including the use of exoskeletons.

Chapter 3 presents a feasibility study on visual error augmentation for a full-body goal-directed balance task. The experiment was conducted using a balance training device called "Thera Trainer coro" which is also detailed in this chapter.

In **Chapter 4**, the effector-independent nature of motor awareness is discussed and tested. Motor awareness of the full-body is compared to the one of the hand through a matched, goal-directed task using the Thera-Trainer coro as a "full-body balance joystick" and a conventional joystick to control a cursor.

In **Chapter 5**, we assess full-body motor awareness and test four different error augmentation gain scenarios during a full-body goal-directed balance task. We studied the effect of the error

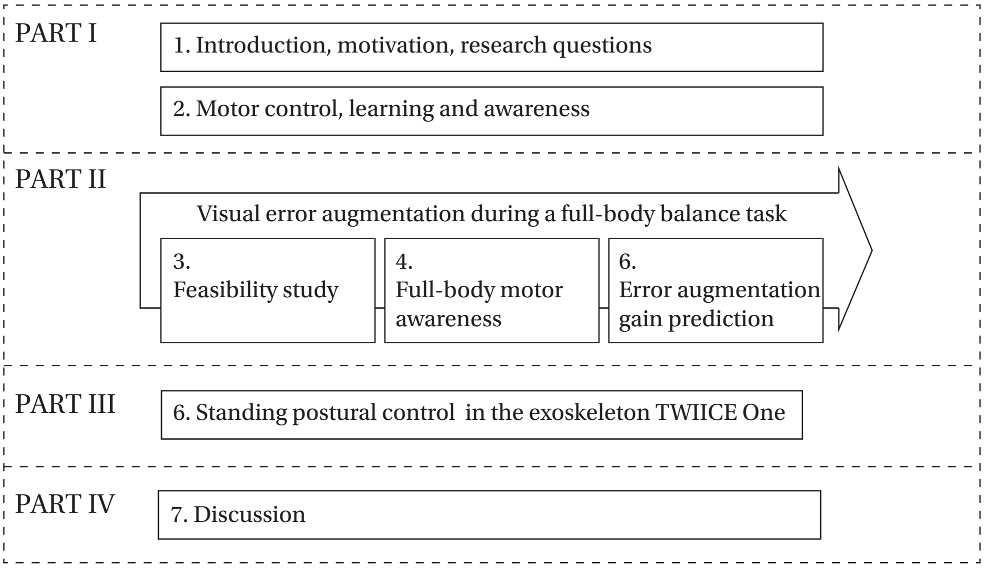

Figure 1.1 – Overview of the thesis with its seven chapters. In the first part, the motivations, the research questions and the state of the art of motor control, learning and awareness, as well as the related work are introduced. The second part investigates the effect of visual error augmentation during a balance task, and its link to motor awareness. In the third part, we investigate how balance management in the TWIICE exoskeleton can be improved while standing.

augmentation gains as function of motor awareness.

Finally, **chapter 6** presents the development of two postural controllers based on a bioinspired approach for the full-mobilization exoskeleton TWIICE, as well as their performance characterization with an individual with motor-complete SCI.

1.3 Personal contribution

1. **Chapter 3** is based on: **Fasola J**, Kannape O, Bouri M, Bleuler H, Blanke O. Error Augmentation Improves Visuomotor Adaptation during a Full-Body Balance Task, 2019 41st Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC).
   Personal contribution: technical and experimental design, implementation, recording, analysis and writing.

2. **Chapter 4** based on **Fasola J**, Kannape O, Blanke O. Motor Awareness is Effector Independent but Strongly Influenced by Kinematic Task Demand, *In preparation.*

3. **Chapter 5** based on **Fasola J**, Kannape O, and Blanke O, On the Relation between Error Augmentation Learning and conscious Motor Awareness in a Goal-Directed Balance Task, *In preparation.*
   Personal contribution: technical and experimental design, implementation, recording, analysis and writing.

4. **Chapter 6** based on :

   - **Fasola J**, Vouga T, Baud R, Bleuler H and Bouri M. Balance control strategies during standing in a locked-ankle passive exoskeleton. 2019 IEEE 16th International Conference on Rehabilitation Robotics (ICORR) Toronto, Canada, June 24-28, 2019.
     Personal contribution: Experimental design, recording, analysis and writing.
   - **Fasola J***, Baud R*, Vouga T, Bouri M and Ijspeert A. Bioinspired Postural Controllers for a Low-Actuator Count Exoskeleton Targeting Complete SCI Users, Frontiers in Robotics and AI, Biomedical Robotics, 2020, *Submitted.* *Equal contribution.
     Personal contribution: Experimental design, recording, analysis and writing.

2 Motor control, learning, and awareness

2.1 Principles of motor control

2.1.1 How do we move?

To understand motor learning, we may first start by examining the roots of motor control. Most our daily activities involve a high number of complex movements. However, humans are able to execute most of their actions with minimal cognitive effort. To control those actions, the central nervous system (CNS) is constantly integrating multimodal sensory information, from the environment and the body, and producing the necessary commands to recruit the adequate muscles via a planning process. *Motor control is thus known as a hierarchical top-down process, described in three phases: planning, programming and executing* (Logemann, 1985).
Years of studies involving brain imaging and electrophysiology permitted to establish the pathways of motor control (see Figure 2.2). Planning might happen in the prefrontal area of the brain while motor programs seem to be processed in the posterior parietal, premotor, and supplementary motor areas before being translated into a motor response in the primary motor area [45]. In this section, the objective is to provide an overview of these three phases and of widely used computational models of motor control.

Motor planning

Complex actions include a sequence of different steps that needs to be planned in advance in order to cope with the time needed to execute them, but also with the time delay of the sensory feedback. Planning specifies the direction, the distance and the speed of the movement to produce a new body state [46]. The *state* is defined by a set of time-varying parameters such position, speed or muscle synergies in a somatosensory (body-centered) reference frame. Studies have found evidence of the existence of a planning stage in which the selection of the movement profile depends on previous movements and is modulated by extrinsic and intrinsic constraints [47].

Motor program based on optimal control

Although a task can be executed differently depending on the chosen muscle synergies, it has been shown that the CNS presents stereotyped movement patterns across repetitions and individuals [48]. Indeed, in 1985, Flash & Hogan proposed that motor programs are based on maximizing movement smoothness and succeeded to reproduce empirical data by minimizing movement jerk. Nevertheless, it was well understood why the CNS would optimize smoothness and how it could cope with the calculation of this complex metric. Consequently, other optimization models have been proposed. Harris & Wolpert proposed a model that takes into consideration not only the motor variables, but also the end-goal of the performed task by suggesting that the CNS minimizes the final trajectory error, measured as variance about the target [49]. Using the movement error as a cost function is behaviorally pertinent and rather simple to compute [12].
In addition to error, humans motor control seems to minimize also the effort needed to perform a movement [50]. These two variables are defined as *natural cost function* for motor programs.

In the view of building any kind of motor program or plan, the CNS needs to know the current state of the body, which is estimated from the sensory signals. It is important to note that sensory afferents alone cannot estimate accurately the body state due to noise and delay. To cope with the inaccuracy of state estimates, it has been proposed and generally accepted that the CNS builds *internal forward models* of movements [51]. The name of the model reflects the fact that it *predicts* the behavior of the body. In the early fifties, Von Holst and Mittelstaedt already proposed that the forward model of the on-going movement uses as inputs the previous state and a copy of neural efference, called the *efference copy* [52], to simulate body dynamics and predict sensory feedback (see Figure 2.1, outer loop).
The CNS then compares and combines the forward model predictions with the sensory signals to estimate optimally the current state. The comparison between the two signals is known as the *observer framework* [51]. A Kalman filter [53] is often described as the optimal observer or comparator for linear systems since it generates the state estimate based on the least squared error [54]. This model was able to estimate correctly empirical data of the hand [51], head orientation [55] and posture [56].

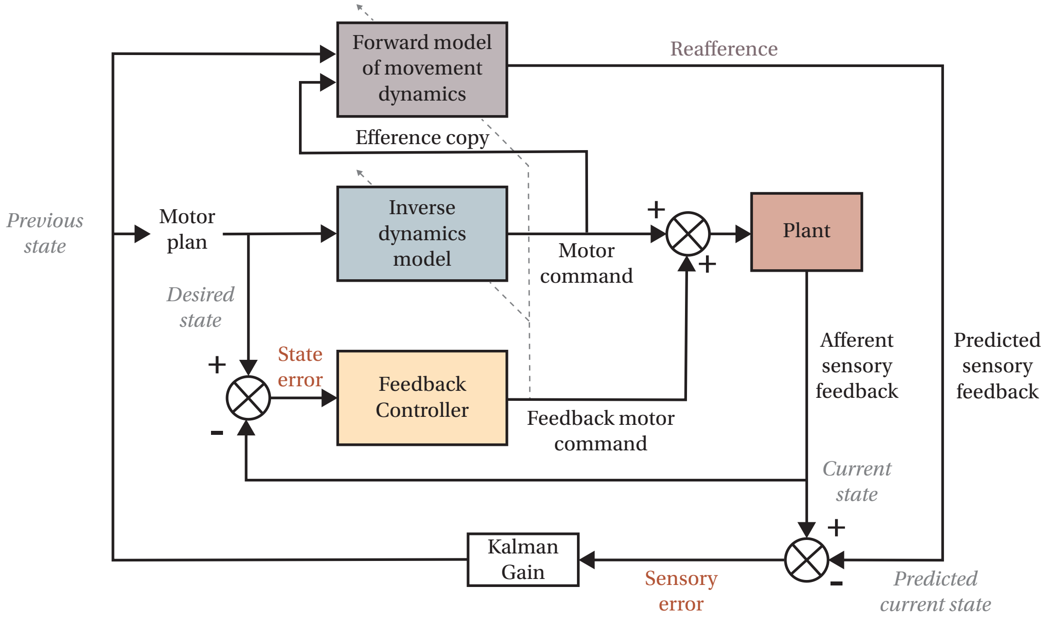

Figure 2.1 – Computational model of voluntary sensorimotor control adapted from [57]. Parallel forward and and feedback mechanisms are used to update motor commands generated by the inverse dynamic model (IDM). The forward dynamic model (FDM) generates a prediction of the sensory feedback. Predicted and afferent sensory feedback are compared resulting to a sensory error signal. This signal indicates how well the movement was executed according to the motor plan and can be used to update the motor plan and/or train the FDM. The feedback controller uses the error between the desired state and the current state estimated the perform online correction of the motor commands. Dashed lines crossing IDM and FDM means that both models can also be updated by state error during sensorimotor adaptation.

The discrepancy between predicted and actual sensory signals is used to control and improve sensory motor control without the need of additional conscious monitoring. The observer framework remains the main model for sensorimotor control, motor estimation, prediction, motor learning [12]. The use of this model will be discussed in section 2.1.3 for motor learning, but also in the case of the central monitoring framework in section 2.2.2.

Motor command generation

Once the motor program is created, the CNS needs to send the right instructions to effectors, i.e. the muscles, that match the desired behavior. One possibility is to activate the muscles needed to compensate for the dynamics of the body or for external perturbations [58, 59]. This kind of system is referred as an *inverse dynamics model.* The controller receives as an input the desired state and regulates the muscles activation (see Figure 2.1, middle loop). When the inverse model is refined, the motor commands generate the desired motor behavior, and the current state matches the desired state. Nevertheless, those inverse models need to

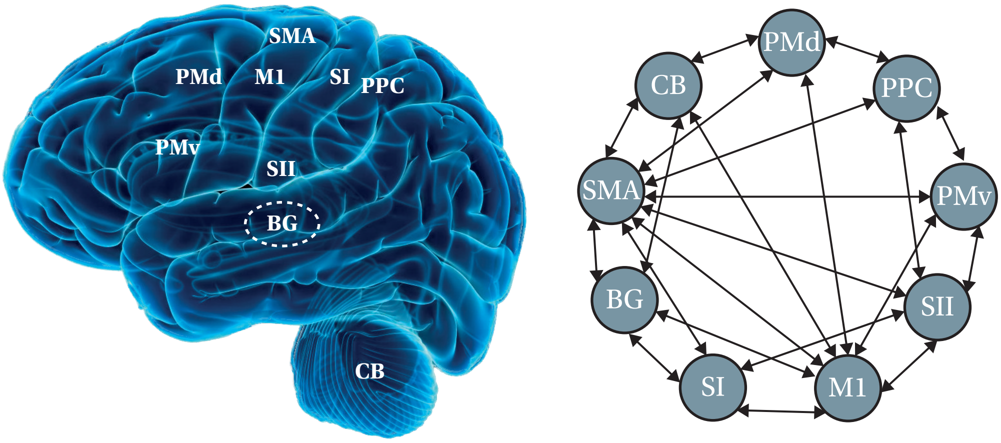

Figure 2.2 – Neural network connections involved in motor learning adapted from [64]. Dorsal premotor cortex (PMd), cerebellum (CB), supplementary motor area (SMA), basal ganglia (BG), primary somatosensory cortex (SI), primary moter cortex (M1), secondary somatosensory cortex (SII), ventral premotor cortex (PMv), and posterior parietal cortex (PPC).

be learnt and refined in order to generate optimal motor commands [46]. In fact, studies in arm reaching movements have demonstrated that the CNS learns to represent compensatory forces when the hand is exposed to a constant force-field generated by a robotic interface, and consequently updates the inverse internal model of the movement [60, 61]. For this reason, this kind of model is also called adaptive inverse model.

Nonetheless, it is not always possible to predict the forces the system will experience, and for this reason, a compensation strategy cannot not always be learnt. In order to produce the desired behavior in case of unforeseeable disturbances, a strategy is to increase the stiffness of the system using muscles co-contraction [62]. Moreover, it seems that the CNS is able to regulate stiffness geometry to stabilize unstable dynamics [63].

2.1.2 I'm still standing

Postural control describes the act of maintaining, achieving or restoring a state of balance, or equilibrium, during any posture or activity [65]. It is achieved by integrating sensory information coming from mainly visual, somatosensory and vestibular input to generate adequate motor outputs and maintain an upright posture [66].

Postural orientation and postural equilibrium are two main functions of postural control [67]. Postural orientation controls the body or body segments position with respect to some reference frames such as gravity, support surface, visual environment, and internal references.

Postural equilibrium refers to sensorimotor coordination strategies to keep the center of mass (CoM) over its base of support (BoS) during self-initiated and externally-triggered disturbances [67].

Postural control is required in most of our daily-living activities, which have been associated with three mains classes illustrated in Figure 2.3 [68]:

- Maintenance of a posture, such as sitting or standing
- Voluntary movement, such as leaning forward to grab a glass of water or dressing up (self-generated perturbations).
- Reaction to external perturbations, such as push, trip, slip

The term "postural equilibrium" includes management of stability during standing posture, but also during walking and voluntary tasks. The postural strategies that are adopted during walking or voluntary task are same than the ones used standing posture mentioned below; they are added to the locomotor pattern [69].

In this thesis, we focus only on stance posture. Indeed, in chapters 3 to 5, voluntary movements of the CoM over its base of support is addressed by a task requiring full-body weight shifts along anterior and mediolateral plane to selected targets displayed on a computer screen [32], see Figure 2.7.C. This task will be referred as *a goal-directed balance task* in the next chapters. In the frame of the development of postural controllers (chapter 6), the three classes mentioned above will be addressed.

Stance posture

Although bipeds have some passive stability, standing or stance posture requires active postural control to act against gravity. Standing humans are constantly making small postural corrections, called postural sway, to avoid losing balance. Postural sway can be measured by quantifying the displacement of the center of pressure (CoP), which describes the point of application of the ground reaction forces needed to adjust the CoM over the base of support.

To maintain standing stability, the CNS has two main postural control strategies: the fixed-support strategy and the change-in-support strategy [67]. The fixed-support strategy brings back the CoM over the actual base of support, while the change-in-support strategy involves a modification of the BoS by taking a step or reaching for a support. In this thesis, only fixed-support strategies will be studied in terms of motor learning and development of postural controller.

Automatic postural responses For the fixed-support strategy, studies have identified two main automatic postural responses, which often co-exist, in response to antero-posterior perturbations: the ankle and the hip strategies [70].
In the *ankle strategy*, the body is represented as a single-link inverted pendulum that rotates around the ankle joints thanks to active generation of ankle torque. In the *hip strategy*, the body is represented as a two-link inverted pendulum, and exerts a torque at the hip joints to quickly modify the CoM position. This strategy is characterized by the opposite rotation directions of the trunk and the shank segments. This ankle strategy is mainly used during small swaying motions, such during postural maintenance, while the hip strategy is involved for higher perturbations [71].
Finally, more rarely included in the automatic postural responses, the vertical strategy involves knees and trunk flexions, and is mainly used for backward falling motion [70, 72].

These automatic postural strategies can be described based either on body kinematics, which define the relationship between body segmental motion or body kinetics involving relationship of body segmental forces. In the chapter 6, we used body kinematics to define postural adaptation strategies adopted by healthy young adults in passive locked-ankle lower limb exoskeleton to maintain their balance during standing. Based on these observed strategies, we developed standing postural controllers.

Anticipatory postural adjustments In complement to automatic postural responses, postural control can be also achieved in a predictive manner for voluntary movements. Therefore, anticipatory postural adjustments are believed to be generated by a feedforward postural controller prior to any sensory feedback indicating postural instability [73]. The most striking examples are that the body moves forward and upon the stance leg prior to taking a step or leans forward for reaching an object [67].

2.1.3 Fundamentals of motor learning

Motor learning is not strictly defined, but rather consists of different categories such as skills acquisition, motor adaptation, and decision making [74]. It is characterized by an improvement of performance beyond baseline levels [75], and is often divided in three successive stages, see Figure 2.4, [76]. The first stage is called the cognitive stage where an initial representation of the motor command is created. This stage requires a high level of attention and is often accompanied by verbal self-talk. The movement is divided into sequences and large errors are usually observed. During the second stage (associative stage), movements are refined and movement variability is reduced thanks to errors detection. In the final stage (autonomous stage), precise movements can be performed in an automated manner, the

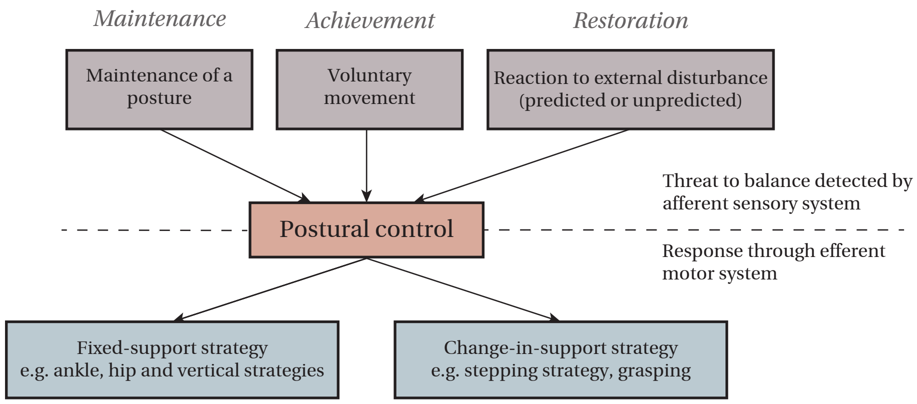

Figure 2.3 – Schema representing functions requiring postural control and types of postural control strategies adapted from [65].

learner being barely aware of the movement execution, which relieves the cognitive system for other task such as decision making or anticipation [76].
For a wide range of different tasks, the learning curve is as the shape of an exponential growth [76, 77]. The first stage presents a rapid improvement in performance, while the two other stages show a slower performance gain (see Figure 2.4.
This mode of learning is generally referred as *explicit learning* due to the generation of verbal knowledge at the cognitive stage [78]. On the other hand, *implicit learning* involves "an accumulation of procedural knowledge, which is inaccessible to consciousness" [30], and therefore, learners are usually unable to verbally explain the changes in technicalities that lead to an improvement [78, 79].
Literature suggests that implicit learning favors automaticity [30], while explicit learning promotes skill transfer [31]. However, the ideal balance between conscious adaptation and implicit learning so that the skill is automated and retained is still unclear [30].

In section 2.1, we discussed how motor commands are based on internal models. During practice, those internals models are constantly updated via feedback loops, leading to modifications of the motor output and movements' correction [75, 80].
In the next section, we will introduce two computational models describing how the CNS is able to learn from different error signals: *sensory prediction errors* and *state feedback-errors*. The first one is sensitive to prediction errors while the second one relies on goal-based performance errors [81]

Sensory prediction errors Sensory feedback allows us to estimate our body state and provides information about the environment. The combination of the observation and the prediction lead to a more accurate state representation (see section 2.1.1). However, the combination of each stream is only an advantage if the predictions are accurate, i.e. if the forward models perform well. If there is a persistent discrepancy between the two signals, the CNS needs to re-weight each stream and update the predictors [54]. For this purpose, the brain processes the sensory feedback from movements which can be used as training signals to learn forward models. Indeed, the discrepancy between the predicted and the actual sensory feedback can be used as a sensory error signal to train the predictors.

State feedback-errors Another possible way to generate motor adaptation is by updating or acquiring inverse internal models. However, the motor command error generated by the inverse model is not directly available. In fact, the brain receives a state error (discrepancy between the desired state and the state estimate) in sensory coordinates that need to be converted into motor errors to train the inverse model. One way to deal with this issue was proposed by Kawato et al. in 1990 [82]. They proposed the existence of a feedback controller generating a feedback motor command based on the state error (see Figure 2.1, inner loop). Consequently, the motor command sent to the plant is the sum of the output of the inverse model and the feedback motor command. If the state error is null meaning that the inverse model is perfectly trained, then the feedback controller is silent. Such a system, called feedback-error learning, allows on-line motor corrections and those motor corrections are directly used to train the inverse models.

> Studies in arm reaching have tried to dissociate feedback-error from sensory prediction error [83]. To do so, participants were asked to perform rapid movements preventing on-line corrections and "pointing" movements which are slow and thus enable on-line motor corrections. Results suggested that adaptation of motor commands is driven by sensory prediction errors rather than motor corrections. These results are supported by other studies in visuomotor adaptation [84] and in eye saccade adaptation [85, 86, 87].

Research in motor learning is most often based on adaptation paradigms since shorter time-scales are needed. There are two prominent adaptation paradigms that have been well studied for different type of effectors:

1. **Force-field adaptation** The experimenter introduces a physical perturbation that alters both visual and proprioceptive consequences of the motor commands. In reaching movements, this is implemented through the use of a robotic arm, called an actuated manipulandum [60], see Figure 2.5.A.
2. **Visuomotor adaptation** The experimenter introduces a perturbation that distorts the visual consequences of the motor commands, while the proprioceptive consequences

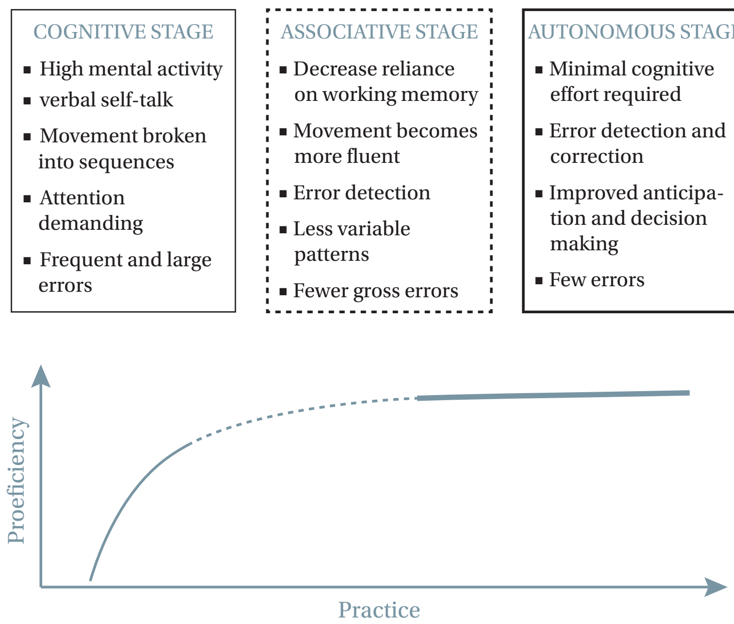

Figure 2.4 – Overview of the three main motor learning stages.

remain unchanged. This type of visual distortion is typically implemented by asking people to control a cursor on screen [88].

Both adaptation paradigms lead to *after-effects* once the perturbation is removed, meaning that participants make erroneous movements in the direction opposite to that seen during initial adaptation. The presence of after-effects is strong evidence that the CNS altered the motor command.

Augmented visual feedback

Research in motor learning aims to boost training efficiency by decreasing learning time and increasing retention. During motor learning, intrinsic feedback, which are related to bodily sensory afferences such as touch, vision, proprioception and vestibular stimuli, are always present and drive internal models updates and formation. Motor learning based on sensory prediction errors underlines the fact that extrinsic feedback, if provided the right way, may substitute or be additive to bodily sensory information in order to enhance motor learning. Extrinsic or augmented feedback are defined as signals that cannot be generated without a display (i.e. screen or headphone) or a trainer [89]. They can be visual, auditory, haptic or multimodal.

> In this thesis, visual feedback was chosen among the other modalities because it is the easiest and most effective way to transmit information and requires only little training before understanding the signal [90].

During the last decade, visual feedback has been extensively studied in the context of motor adaptation and motor learning [11]. However, studies reported opposing results when the methods slightly changed in terms of design, timing, focus of attention and learning phases. Indeed, numerous choices have to be carefully taken when designing a visual feedback. Sigrist et al. [11] reported some contextual insights gained from studies on visual feedback, starting with the moment at which the feedback is provided.

Feedback timing *Concurrent feeback* refers to feedback given during the task execution. There is evidence that concurrent feedback is beneficial for immediate performance during practice; however, the performance gain is often lost once the feedback is removed in the retention phases [91]. These findings are supported by the guidance hypothesis: permanent guidance can lead to dependency [92, 93]. In other words, dependency on the extrinsic feedback overcomes intrinsic afferent information, such as proprioception, essential for motor learning. As a consequence, the feedback itself becomes part of the task and is needed by the formed control strategies in the brain to execute the movement [94, 95]. However, Wulf et al. explained that congruent feedback could provide an external focus of attention, which promotes the automaticity of movements and decreases cognitive overload [96, 97].
Terminal feedback, also called post-response feedback, is given at the end of the task execution. This type of feedback seems to reduce accuracy and consistency during practice in comparison to concurrent feedback, but leads to better performance during the retention tests [98, 99]. However, the feedback frequency may not have the same effect depending on the complexity of the learned skills and may have to be adapted to the learning phases [100]. Indeed, studies found that learners of high complexity tasks, such as sport-related skills, might benefit of concurrent or very frequent feedback [101].
Bandwidth feedback adapts the frequency of the feedback to the skills of the individuals. The principle is to give a feedback when the error exceeds a certain threshold. This technique has shown to be effective in arm-hand skills training for stroke patients [102], as well as in learning of complex gymnastic skills [103].

Feedback design The design aspect of the visual feedback could range from abstract design, such as bar plots, gauges or stick figures, to more natural visualization such as 3-D virtual immersion and avatars [104, 105, 106]. In simple tasks, abstract design has proven to be more effective than virtual mirror or real picture. It allows to highlight most relevant information of the task, since the number of significant parameters is also reduced [107]. Similar results were observed for complex task, where augmented reality did not enhance learning of a complex hand movement [108]. Designs including 3-D perspective have shown good results when a virtual teacher was part of the training. In contrast, whole body superposition did not enhance performance, mostly because of optic flow interferences [109].

> To summarize, concurrent feedback seems to be adequate for early learning phase or for very complex tasks. Studies demonstrated that the frequency of the feedback should decrease in later learning phases to avoid guidance dependency. Terminal feedback could be provided as bandwidth or self-controlled feedback. An important aspect is to keep the feedback motivational and positive to encourage learning. Finally, simple visual design seems to work as well as more complex 3-D visualization as long as the practice stays entertaining.

Error augmentation strategies for neurorehabilitation

Physical guidance or assisting robotic therapy is often used to reduce movement errors while training. In other words, a haptic feedback helps the participants to perform the task better. However, there is little evidence that haptic guidance improves motor learning [110, 111]. It has even been suggested that it might impair motor learning based on the guidance hypothesis, which states that internal models become dependent of the feedback and therefore fail to learn the adequate motor command [93]. Furthermore, physical effort and attention during practice are believed to be essential factors to promote motor plasticity [112].
Several studies have found consistent results that error detection and correction are needed to enhance motor adaptation and learning [113, 84, 54, 83]. Since the neural system seems to rely on errors to update the internal models of movements, studies suggested increasing the magnitude of the errors rather than decrease them [114, 115, 116]. This type of learning strategy is called challenge-based adaptation. Patton et al. [117] demonstrated that error magnification by applying a force during training led to significant improvement in arm reaching movement in patients with stroke. Equivalently in lower limbs, increasing the phasing error in gait with a split-belt treadmill resulted in a longer after-effect in gait symmetry [116, 118]. In healthy participants, it has been shown that haptic error amplification is a more suitable strategy, especially for less skilled subjects, than error reduction in a simple locomotor task [24]. In contrary, Domingo and Ferris have demonstrated that short-term learning is not improved by increasing errors by destabilization or by beam width reduction during walking on a narrow balance beam [94].

It seems that the efficiency of the *error augmentation strategy is related to task complexity and motor skills.* In fact, training of a reaching task with error augmentation was more beneficial

for less impaired patients with stroke, whereas haptic guidance was more appropriated for patients with more severe impairments [25]. Consistent results were found in healthy individuals in a timing-based motor task such as playing pinball [119].
This can be explained by the *optimal challenge point theory* which states that learning is optimal when the difficulty of the task is adapted to the prior skills of the learner. In consequence, error augmentation might be detrimental for learning complex tasks because it might add instability to the sensory feedback and lead to frustration [120]. If applied to complex tasks, error augmentation should be used with proficient participants in order to refine their motor skills [121].

Visual error augmentation Studies investigated whether visual error augmentation led to similar results. Participants were asked to perform arm reaching movement under visuomotor rotation while holding the handle of a manipulandum (see Figure 2.5). Their arm was hidden below a screen displaying a dot representing their hand position. In the experimental groups, the hand position was distorted with an error augmentation factor. The groups that trained with the visual error augmentation had better learning outcomes both in healthy participants [16, 15] and stroke participants [18].
Most research on visual error augmentation has been done on upper limbs with promising results [18, 122, 16, 17, 15, 22].Only recently, studies have tried to apply this paradigm to gait rehabilitation [123, 21]. Marchal-Crespo compared the effect of visual and haptic error augmentation when learning a modified locomotor pattern in the exoskeletal robotic system Lokomat [21]. Haptic error augmentation did not impede motor adaptation and promoted the transfer of the learned asymmetric gait pattern to free walking. However, visual error augmentation hampered motor learning and reduced the perceived competence of participants.

The visual error augmentation strategy has been little explored in the field of balance learning. One recent study [124] applied time-dependent visual error augmentation on a weight-shifting task. The results showed that the error augmented feedback reduced subjects' time to reach their steady-state performance compared to unaltered visual feedback, although it did not change the final steady state performance. These findings suggest that visual error augmentation could improve balance rehabilitation in term of performance speed and retention.
Visual error augmentation does not alter the proprioceptive consequences by definition, and thus might be less efficient than haptic feedback. Nevertheless, it is an attractive strategy, especially for gait and balance rehabilitation because it does not apply forces, and therefore ensure training safety (i.e. risk of falls). Finally, visual error augmentation could easily be integrated in exergames and virtual reality (VR) trainings, which seems to increase motivation and active participation [125, 126].

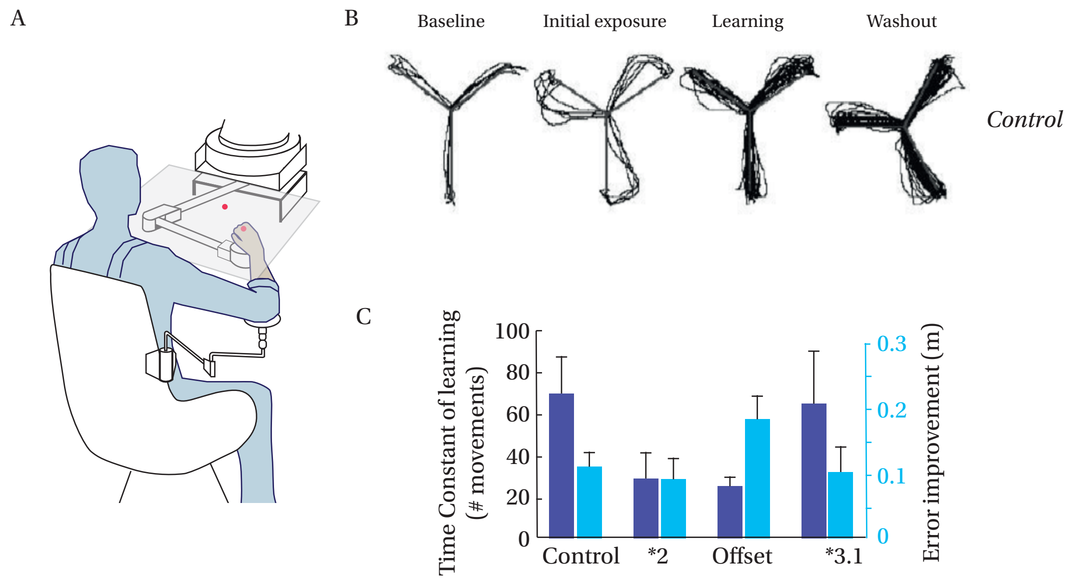

Figure 2.5 – Visual error augmentation during arm reaching movements with a manipulandum **(A)**. Healthy young participants were exposed to a 30° visual distortion and were asked to reach the targets in a straight line. Representative trajectories of the hand are shown in **(B)**. Results showed that participants that trained with an EA gain of 2 adapted faster than the control and improved as much **(C)**. The figures are adapted from [18].

2.2 Motor awareness and sense of agency

2.2.1 Introduction

The sense of agency (SoA) describes the feeling of being the one causing or generating an action and its sensory consequences[127]. Feeling in control and being aware of the consequences of our own actions allow to dissociate events that are self-generated from those caused by the environment [128] or another agent [129]. For this reason, the SoA has been linked to bodily self-consciousness, which is the sense of self (self-identification and self-location) stemmed from integration of bodily signals such as visual, tactile proprioceptive and vestibular signals [130]. The SoA has been divided in two distinct levels by Synofzik and colleagues in 2008 [131]. An implicit, low level of *feeling of agency* is described as on-going and pre-reflective sensorimotor processes. It is opposed to an explicit level of *judgement of agency* characterized by reflective and cognitive processes (i.e. awareness of who has caused the action), see also [132]. The latter seems to strongly depend on the action outcomes: whether it has been successfully achieved [133] or whether an external event coincides with the intention (i.e intentional binding, [134]. These distinct mechanisms behind the SoA have aroused interest since the early work of Nielsen in 1963 and 1978 [135, 136], and have more recently been intensely studied, especially by Chris Frith and colleagues and Marc Jeannerod

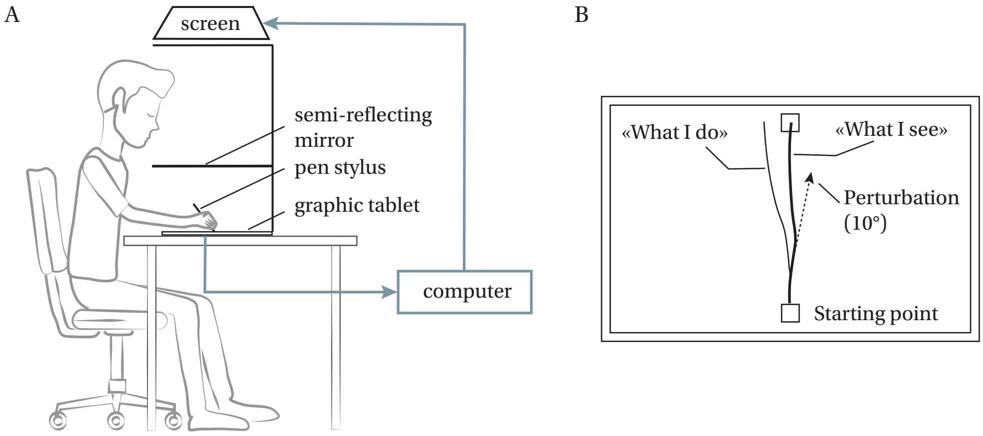

Figure 2.6 – Sense of agency. **(A)** Experimental setup used by Fourneret and Jeannerod in 1998 to investigate conscious action monitoring based on goal directed movements. **(B)** Participants were asked to draw a straight line but received a deviated visual feedback on the semi-reflecting mirror. Participants did not perceive the feedback to be deviated for perturbations up to 15°, even though they automatically compensated for the visuomotor mismatches.

and collaborators [137, 26].

Analog to research on motor adaptation, the aforementioned SoA research primarily explored *conscious action monitoring* of the upper limbs, as an approach to investigate the low-level of the sense of agency. Action monitoring therefore focuses on the awareness of a current sensorimotor experience and its conscious limitations. It was evaluated in response to sensorimotor incongruences between visual, proprioceptive and motor commands. Generally, participants were asked to perform repeated goal-directed movements during which their hand was hidden below a screen and replaced by a virtual hand that could be delayed in time or deviated in space, see Figure 2.6. Participants were asked to consciously monitor their action. At the end of each movement, participants were asked whether the movement they saw on the screen exactly corresponded to the movement they performed [138]. The awareness of "how" the action was performed is described as *motor awareness* and corresponds to participants' ratings of self-generated actions. Those research works demonstrated that participants self-attributed movement feedback that was deviated in space by up to 6°-15°, even though they unconsciously corrected their trajectory [26, 129, 139, 140]. Similarly, experiments with temporal delays between the participant's actual hand and the visual feedback showed that participants self-attributed delayed movements below a delay of 100-200 ms [138, 141].

2.2.2 Predictive processing in agency

In general, the SoA is reinforced when the action outcome corresponds to the expected immediate consequences. Thus, the concept of agency has often been associated with the computational model of sensorimotor control [51], the observer framework (see section 2.1.1). Indeed, the predictions of the sensory feedback can also be used to enhance perception [142]. Subtracting self-generated sensory feedback to actual sensory feedback allows the sensorimotor system to detect those that are externally generated. A striking example is that one cannot tickle oneself [143]. Therefore, sensory predictions not only allow to correct on-line movement errors and refine internal models, but also contribute to the sense of agency and conscious monitoring of your own actions.
The observer framework was thus adapted and extended to the concept of *central monitoring* to describe the neurocognitive model of agency and to account for abnormalities in motor awareness in patients with cognitive disorders or schizophrenic symptoms [137, 144].

> The fact that the sense of agency and sensorimotor control might share the same predictors, which are directly involved in motor learning, drove us to the idea that the motor awareness threshold might be related to motor learning and error perception. Several studies in motor learning have suggested that large, or even catastrophic errors need to be experienced in order to acquire or adapt motor skills [113, 84, 54, 83, 94]. A lingering question is whether those errors have to remain unconscious (implicit learning) or consciously processed (explicit learning) to promote movement automaticity [30]. Error amplification through visual feedback could push at times the errors over the MA threshold. Is it beneficial or detrimental for motor learning?
> A first step to answer this question is to study the relation between motor awareness and the effect of EA gain on motor learning. This will be explored in chapter 5.

2.2.3 Full-body agency and bodily self-consciousness

The sense of agency until recently, has been studied mostly in upper limbs. Nevertheless, research on bodily self-consciousness has demonstrated the importance of representing the self as a single spatially-situated entire body, rather than separated body parts [33].
Indeed, full-body representation of the self has been experimentally tested in healthy adults [130, 34, 145] following clinically induced "out-of-body" experience in neurological patients [146]. This line of work demonstrated the existence of a global body representation in the brain, which is dissociated from cortical representations of individual body parts (i.e. Blanke et al., 2015 [34]).
The research work of Menzer [27] and then Kannape in 2010 [28] extended the hand agency paradigms to the movement of the full-body by studying walking agency. Menzer et al. used

audio-motor conflicts of footsteps to assess full-body agency [27], while Kannape et al. measured spatial and temporal gait awareness using visuo-motor conflict by displaying participants virtual body on a large projection screen [147, 29, 28, 148, 149]. Interestingly, the temporal awareness threshold for full-body audio-motor cues were similar to the one of visuo-motor cues for body parts (i.e. temporal delays are not detected up to 100-200 ms) [27]. In line with these findings, the spatial and temporal thresholds of waking agency were comparable to the ones of upper limbs [28, 29].
These results suggest that the mechanism of agency is 1) independent of the effectors and 2) independent of the modality such as audition or vision. It has been thus proposed that the sense of agency may be based on an *effector-independent and supramodal mechanism* [27, 147].

> Nonetheless, no study up to now compared the motor awareness thresholds of two different effectors in a matched agency task and within participants. The framework of this thesis allows to directly compare full-body agency involved in balance control and hand agency. In addition of being two distinct effectors, studies in bodily self-consciousness have also demonstrated that they have separate body representations in the brain [130, 34]. Therefore, we assumed that they are two relevant contenders to test the effector-independent hypothesis [27, 147].

2.3 Neurorehabilitation of postural control

Balance control is related to different physiological mechanisms such as biomechanical constraints, cognitive processing, sensory modalities, sensory integration and reweighting, movement strategies and perception of verticality [150]. Therefore, any pathology in the nervous system or musculoskeletal system can hinder postural control. In case of spinal cord injury, the sensory pathways giving information about limb orientation, as well as the motor pathways activating muscle synergies are disrupted. Lesions in the brainstem, such in the vestibular nuclei, impairs sensory integration across modalities responsible for postural orientation. Damage to the cerebellum may lead to severe problems with postural stability and orientation due to difficulty to optimize postural strategies [151]. The important role of the basal ganglia in postural control can be deduced from the frequency of falls in patients with Parkinson's disease. The basal ganglia seems to play a central role for adaptation of postural strategies, regulating muscle tone, generating anticipatory and reactive postural responses, and perception of postural orientation [152]. The cerebral cortex in different stages of postural control such as generation of anticipatory postural adjustment and longer latency postural responses, formulation of the internal models of the body and its environment, and among others, regulation of postural strategies as function of cognitive states [153]. Consequently, any damage caused by a stroke can impair postural control.

One-third of the older population (+65 years old) reports balance difficulty and experiences a fall each year [154]. Common consequences of falls are fractures, especially hip fractures, fear of falling which restricts their daily living activities, and costly health problems [155, 156]. Therefore, improvement of balance function is a desirable outcome in rehabilitation programs.

There is evidence that injured sensorimotor systems can reorganize with intensive tast-oriented training due to neuroplasticity [157]. Indeed, imaging studies in stroke survivors [158, 159], in patients with SCI [160, 161], and with Parkinson's disease [162] showed that neural plasticity occurs at multiple levels including the brainstem, spinal cord, and peripheral nervous system. Based on motor learning principles, it is nowadays generally accepted that rehabilitation training has to be challenging, task-oriented, repetitive and motivating to promote plasticity changes [163].

In this section, the recent studies using augmented visual feedback for balance neurorehabilitation will be reviewed, as wells as the use of exoskeletons for assisting and managing standing postural control in various type of populations suffering from balance impairment.

2.3.1 Current balance rehabilitation techniques

In rehabilitation settings, specific metrics are used to evaluate standing postural control. Goldie et al. divided standing balance in three aspects: symmetry, steadiness and dynamic stability [164]. Symmetry refers to equal weight distribution between the feet. Steadiness describes the ability to maintain a posture with minimal corrective movements called sway. Finally, dynamic stability is the ability to react to changes while the body is in motion. These components are typically affected in elderly, patients with a stroke or with Parkinson's disease [164]. Therefore, commercial sensing platforms providing a visual feedback of the location of the CoP or projected CoM, as well as training protocols, have been developed. The type of exercises is often very similar (see Figure 2.7) and uses concurrent feedback: The users aim to keep their CoP or CoM inside a target, or shift their weight to follow a cursor or reach some targets [165, 166, 167, 168]. Some studies agreed that this kind of practices reduces immediate sway and increases symmetry [165, 169], but others did not find significant changes [168]. Mostly, studies also failed to detect retention once the feedback is removed [165, 170]. It is moreover not always recognized that trainings with visual feedback in combination to conventional therapy help to further improve functional balance. For example, a study with the SMART Balance Master showed that visual feedback did not offer additional benefit to conventional therapy in early stage of stroke rehabilitation [166, 171]. However, the same system has shown to improve significantly balance and functional outcomes for chronic stroke patients [172]. Wii Fit training alone did improve balance of healthy older adults,

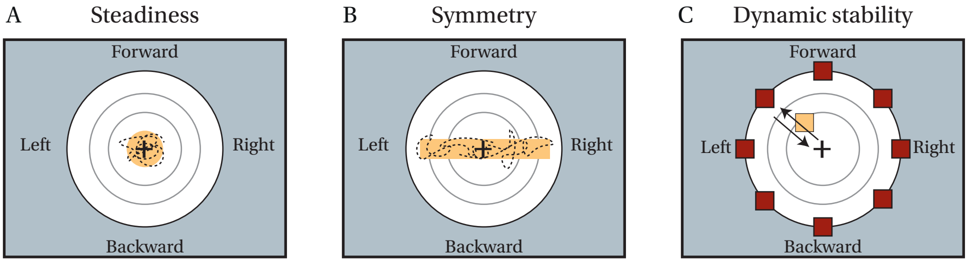

Figure 2.7 – Common balance exercises adapted from [32]. **(A)** Steadiness training with a central target. Participants are asked to keep the cursor (+) representing either the CoP or the CoM inside the orange circle (dotted line). **(B)** Symmetry training by performing repetitive lateral weight shift while keeping the cursor in the orange rectangle. Performance measures are typically the time spent inside the targets or the sway magnitude. **(C)** Dynamic stability is trained by asking the patients to move the cursor (orange rectangle) from the center to a lit target and then to come back to the center. The goal is to train weight shift in many directions. Performance metrics are typically trajectory errors and reaching time.

but physical therapy exercise or a combination of both had greater benefit [173]. It has been proven that visual feedback of the CoP is more effective than tactile and verbal cues in reducing asymmetrical position during standing for hemiplegic patients [165]. More recently, virtual reality-balance training has emerged and showed promising results [105, 106, 174, 175]. As summarize in a review of Mao et al. in 2014, VR-based training for balance seems to promote neural reorganization and improve spatial orientation in patients with neurological impairments. It may also facilitate transference to real-world tasks and reduce fear of falling [175].

VR-based balance training was essentially tested in neurological patients or elderly people. However, investigate the effects of VR on balance training in healthy adults through the applications of common motor learning principles, such as error augmentation, could provide some insight for the design of novel rehabilitation protocol.

2.3.2 Towards the use of exoskeletons

The rehabilitation methods mentioned in the previous section are not adequate for patients with severe postural control disorders, who cannot stand up and walk by their own. Lower-limb wearable exoskeletons have emerged as advantageous rehabilitation tools, allowing to stimulate neuroplasticity in the subacute phase, as soon as possible after patient's vital functions have been stabilized. Indeed, recent studies demonstrated that incorporating exoskeletons usage in intensive rehabilitation cares can promote partial restoration of sensorimotor functions in functionally complete SCI patients [176, 177]. For more details, the reader may refer

to previous excellent reviews highlighting the potential benefits of using an exoskeleton after SCI [178, 179, 180, 181].
Many exoskeletons have been developed in the aim to help paraplegics to recover (Lokomat[182], LOPES [183], Ekso Bionics [184], H2 [185] or Indego [186]), and to restore mobility in daily life, such as Rewalk [187], MINDWALKER [188] or leg braces [189]. Figure 2.8 presents three exoskeletons destinated to different usages. A systematic review of the current existing exoskeleton devices will not be provided here, however the reader may refer to recent review articles. Young and Ferris [43] compared the main mechanical aspects between different exoskeletons. In [190], the authors described and reviewed the most common control strategies for exoskeletons and prostheses, while Gorgey in 2018 [181] provided a critical view about exoskeletons and their acceptance in rehabilitation and daily life.

Among the existing lower-limb exoskeleton, most of them rely on the use of crutches to manage lateral and sagittal balance while standing and walking [184, 187, 186]. This has the inconvenience that the hands cannot be used for other activities.
Very few full-mobilization exoskeletons take care of balance management [191, 176, 192, 193]. Harib and colleagues [194] first demonstrated a controller enabling dynamic walking with the exoskeleton Atalante. SCI patients were able to walk without crutches on a 10-meter track, at speed up to $0.15\,\mathrm{m\,s^{-1}}$. Nonetheless, this comes at the cost of a low walking speed, and presumably, an important overall weight and size since six actuators per leg are needed. No details about mechanical implementation have been disclosed yet, including weight or materials.
Standing and walking balance are essential functions to promote exoskeleton usage during daily living activities and enhance neurorehabilitation, that should not be detrimental to other features such as climbing stairs and managing slopes. By design choice in favor of leanness, lightness and simplicity, exoskeletons are often actuated in the sagittal plane only. Nevertheless, lateral stability provided passively while standing if the distance between the feet is large enough [195], and thus assistance or mobilization in the sagittal plane can enhance standing stability. Hence, several research groups work nowadays on partial assistance during stance for incomplete SCI patients [196, 197, 198, 199]. These works mimic the most common postural strategies highlighted by Winter et al. [200]: the ankle, the hip and their combined strategies and are using torque control. Recently, Emmens et al. [197] proposed a balance controller to assist SCI-incomplete patients during standing, which regulates ankles torque based on proportional-derivative (PD) control law receiving body sway as input and keep the knees extended. They were able to reduce sway and enhance stability against perturbations. Farkhatdinov and colleagues proposed enhance standing balance by assisting the stepping strategy with an exoskeleton actuated at the hip and knee joints [198]. They demonstrated that exoskeleton assistance can decrease step duration and increase reaction speed, as well as increase hip flexion in healthy participants.

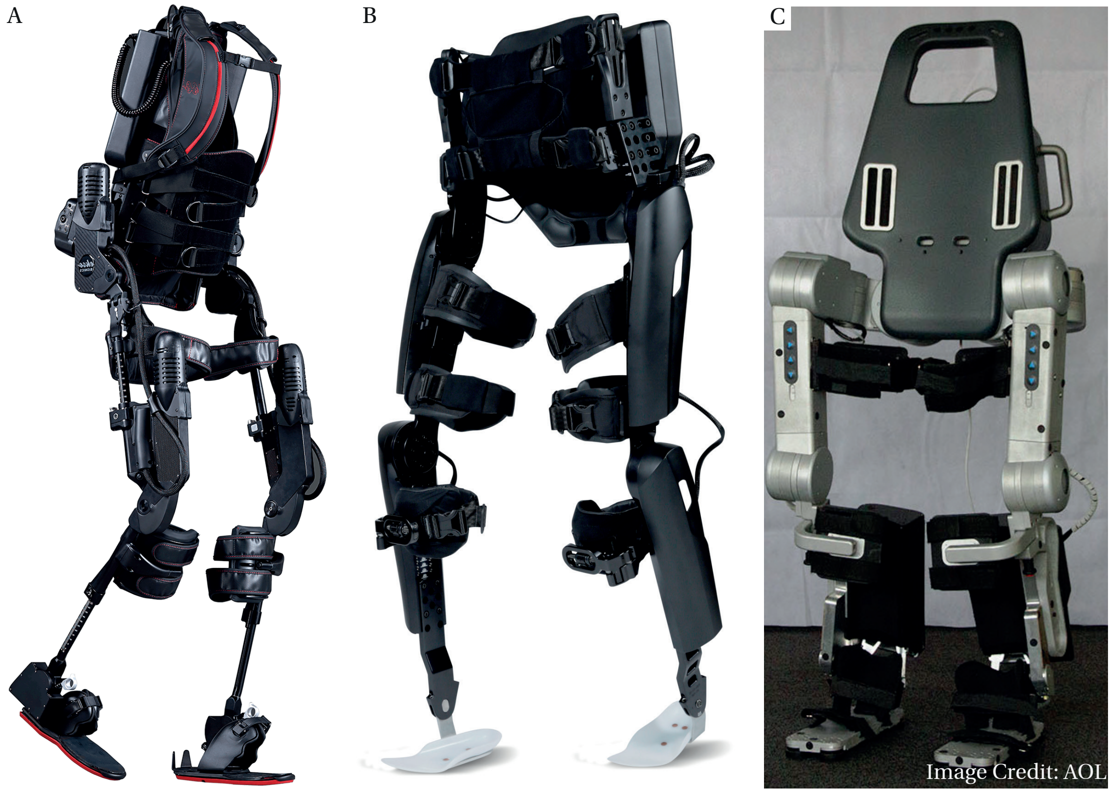

Figure 2.8 – Overview of three commercial exoskeletons: **(A)** Ekso Bionics which is used for neurorehabilitation, **(B)** Rewalk that is destinated to daily activities, and **(C)** Atalante, a self-balancing exoskeleton.

> These recent works on standing balance with lower-limb exoskeletons actuated only in the sagittal plane targeted incomplete SCI patients, and therefore provided assistance. However, one of the most widespread application of lower-limb exoskeletons is for the mobilization of motor-complete SCI patients. Standing balance is a key feature to promote exoskeleton usage on daily bases. Hence, in the second part of this thesis, we present two balance controllers for complete SCI users with TWIICE exoskeleton [44].

3 Visual error augmentation during a full-body balance task: feasibility study

Error Augmentation Improves Visuomotor Adaptation during a Full-Body Balance Task

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Abstract
Visual amplification of kinematic errors has successfully been applied to improve performance for upper limb movements. In this study, we investigated whether visual error augmentation can promote faster adaptation during a full-body balance task. Healthy volunteers controlled a cursor by shifting their weight on the THERA-Trainer coro platform. Two experimental groups and one control group were asked to reach visual targets. For the two experimental groups, the cursor's deviation from the ideal straight line trajectory was augmented by a gain of 1.5 and 2, respectively, while the control group did not experience visual error amplification (gain of 1). Error augmentation with a gain of 1.5 enhanced the speed and the amount of motor adaptation, while the highest gain might have decreased the stability of adaptation. As visual feedback is commonly used in balance training, our preliminary data suggest that integrating visual error augmentation in postural exercises may facilitate balance control.

3.1 Introduction

Since human balance is inherently unstable, postural control is achieved by constantly adjusting the relative motion of center of mass (CoM) of the whole body over the base of support [201]. Populations at high risks of falling, such as patients suffering from stroke and Parkinson's diseases, have shown impaired control of the CoM [202, 203]. This might be due to deficiencies in different sensorimotor mechanisms like sensory integration and reweighting, cognitive processing or kinesthetic body mapping.

Visual, somatosensory and vestibular afferents are the three main sensory modalities involved in balance control. When a mismatch appears between these modalities, vision often takes the lead over the different senses, i.e. visual dominance [204]. Therefore, visual feedback has been extensively used to improve static balance. However, functional balance, even with fixed based of support, often includes dynamic movements. For example, directional balance control is involved in many daily life activities such as leaning to reach an object, changing direction or anticipatory movements. A dynamic goal-directed standing task is a good way to assess directional balance control and is often used to restore balance in elderly, patients with Parkinson's disease and stroke. Although visual feedback seems to be the easiest way to augment sensory information, there is currently no consensus about the optimal strategy [11].

Literature on human sensorimotor control has proposed that the nervous system forms internal models of movement dynamics based on sensory inputs. These models are updated during learning to modify the motor outputs in order to produce accurate and automatic movements [75]. Moreover, pioneering research in motor learning suggested that kinematic errors drive the learning mechanisms [51]. Along those lines, several studies have demonstrated that error detection and correction are needed to enhance motor adaptation and learning [12, 13, 14].

Since the human nervous system seems to rely on errors to update internal movement models [12], studies suggested increasing the magnitude of the errors to improve training efficiency. Studies on upper [205, 117, 206, 207] and lower limbs [24, 208] demonstrated that error magnification with robot-generated forces during training significantly enhanced motor learning. Visual error augmentation for upper limb movements helped to speed up the adaptation process in healthy individuals [18, 16, 15, 17] and lead to functional improvement in a stroke population [18]. In addition to stimulate the error-driven learning mechanism, larger errors may augment the signal-to-noise ratio for sensory feedback [18, 209]. Error augmentation may shift task demands towards executive functioning and cognitive control. In this way, self-evaluation on the kinematic performance may be facilitated.

Nowadays, the literature on error augmentation for balance training is still sparse. Although balance training with visual feedback is increasing with systems like the Wii Balance Board or the Balance Master Systems, most of the research involving visual distortion such as error augmentation has been conducted on upper limbs. The technique itself however is well known by therapists who often exaggerate the errors of the patient in order to make them more noticeable. A recent study [124] applied time-dependent visual error augmentation on a weight-shifting task. They found that the error augmented feedback reduced subjects' time to reach their steady-state performance compared to unaltered visual feedback, although it did not change the final steady state performance. These findings suggest that visual error augmentation could improve balance rehabilitation in term of performance speed and retention.

This pilot study aims to investigate whether visual error augmentation applied to directional balance control can enhance visuomotor adaptation. To increase the difficulty of the task for healthy participants, a visual angular deviation of 30 deg was added to a goal-directed dynamic balance task [210]. Based on previous studies, the error was measured from a straight-line path between the center and the target and was multiplied by a linear gain. The magnitude of the error amplification consisted of gain of 1 (control), gain of 1.5 as in [124] and gain of 2 as in [18]. As in upper limb studies, we expect participants exposed to visual error augmentation to have a higher adaptation and learning rate, but reach the same steady-state performance. As we studied healthy participants, we do not expect an overall improvement in directional control after the angular deviation is removed.

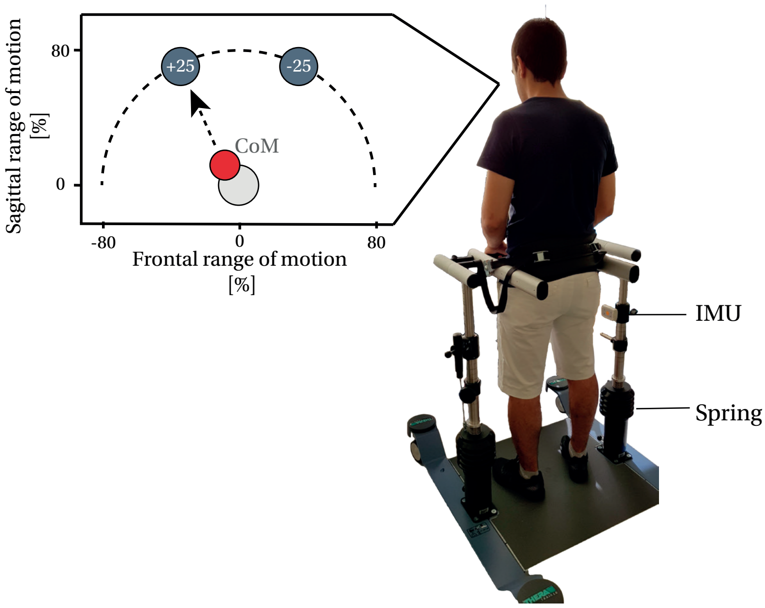

Figure 3.1 – : Experimental setup and visual feedback. Participants were secured at the pelvis level to the THERA-Trainer coro. The IMU on the frame of the device measured the tilting angles, from which the CoM displacement could be estimated and displayed on a screen. Participants were asked to lean towards and stop at the targets, which would appear at one of two predetermined positions in pseudo random manner.

3.2 Materials and Methods

3.2.1 Experimental setup: Thera Trainer Coro

One goal of the current study is to adapt the paradigm for clinical populations, e.g. with impaired balance; the experimental setup was thought to ensure the safety of participants at high risk of falls. The THERA-Trainer coro [25], a therapy device for safe dynamic balance exercises was therefore used Figure 3.1. Its frame is height-adjustable and equipped with two spring elements, whose resistance was set to minimal for this study. No degrees of freedom were locked during the experiment. It is equipped with an inertial measurement unit (IMU) which wirelessly transmits the inclination data of the frame at a sampling rate of 30 Hz. Participants were standing with the feet shoulder-width apart in front of a 15-in liguid crystal display (LCD) and were secured via straps at the pelvis level. They were instructed to maintain a static base of support and to keep their arms crossed over their chest. The frontal and sagittal displacement of the participant's center of mass was estimated by multiplying the pelvis height with the sinus of the roll and pitch angles given by the IMU, respectively.

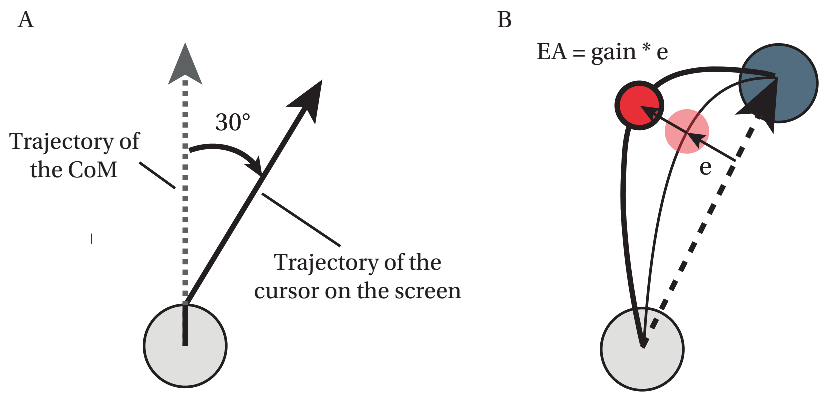

Figure 3.2 – Angular deviation and error augmentation schemas. A. The effect of the 30° angular deviation on a straight forward leaning movement (dashed arrow) as observed on the screen (solid arrow). Participants had to adapt to this constant clockwise visual rotation. B. Illustration of the error augmentation paradigm. The ideal straight-line trajectory is represented by the dotted black arrow, while the CoM trajectory between the resting center (grey dot) and the target (blue dot) is indicated by the thin line. With EA, the cursor position (red dot) is displayed as the online error (transparent red dot) multiplied by the augmentation gain (1.5 or 2).

3.2.2 Participants

We performed a single blind randomized study with nine naïve healthy participants from the research group (5 females), ages 25-33 years. The evaluation of visuomotor adaptation in a goal-direct dynamic balance task had a duration of 20 min without any resting time. To avoid fatigue, only healthy young participants with an intact musculoskeletal system and no history of orthopedic, neural or psychiatric disorders have been enrolled in the study. The experimental procedure described in this pilot study was approved by the local ethics committee in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki.

3.2.3 Protocol

A cursor representing the projection of participant's CoM in the transverse plane was displayed on the screen. A center target with radius of 250 mm (average gaze angle: 3.7 deg ± 0.1) indicated the up-right resting position. Once in the resting position, participants were asked to perform successive outward leaning-and-stop movement towards visually displayed targets as straight as possible. Two targets were placed at ± 25deg from medial axis along a circle with a radius of 80% participant's range of motion (Figure 3.1). After reaching the target, they had to return to the resting position to start the next trial. Return movements to the resting position were not analysed, since they were assisted by the spring system of the THERA-Trainer coro.

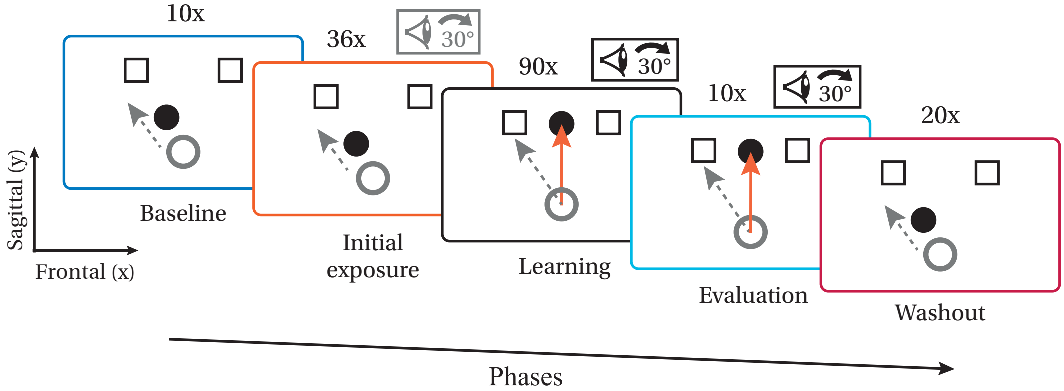

Figure 3.3 – Experimental timeline. Participants started with congruent visual feedback and then were exposed to visual angular deviation on every sixth trial in the initial exposure phase. The feedback had a constant clockwise visual deviation of 30° during training and evaluation. Visual deviation was removed for the washout phase.

Targets appeared randomly to the left or right of the medial axis, varying the shift direction to avoid anticipation effects.

All participants had to reach the targets in presence of a visual angular deviation of 30°. The onset of the deviation started beyond a distance of 250 mm from the resting position (Figure 3.2.A). As in [18], a perpendicular vector from the ideal straight-line trajectory was used to define the online error. The cursor location was moved by a gain of the current error vector (Figure 3.2.B). The control group had a gain of 1 and experienced the angular deviation without error augmentation. The two other groups had a gain of 1.5 and 2.

The protocol (Figure 3.3) was very similar, but shorter than the one from [18]. All groups had the same protocol consisting of five blocks. They started with 10 leaning-and-stop movements; 5 to each target to establish the baseline. During the second block (36 trials), they experienced initial exposure to the angular deviation every 6 trials. No error augmentation was displayed at this stage. The learning block was made of 90 trials with angular deviation for all groups and error augmentation, except for the control group. The two groups with error augmentation experienced catch-trials during training on every sixth trials. During these catch-trials, their respective error augmentation was removed. Training was followed by an evaluation block of 10 trials, where error augmentation was removed but the angular deviation was conserved. Finally, the last block is the washout phase of 20 trials, where the angular deviation and the error augmentation are removed. It allows to study how the human nervous system de-adapts.

3.2.4 Data analysis

Off-line data analysis was conducted in MATLAB. Estimated CoM positions were filter with a moving average filter. The main performance metric was the peak trajectory error (PTE). It was calculated as the maximum perpendicular distance between the actual CoM position and the straight line path between resting and target position [18]. To assess the change of trajectory error during training with or without error augmentation, an exponential curve was fitted to the PTE time series of the training phase. From this fit:

$$e_i = A\exp(-i/b) + C \tag{3.1}$$

Where i is the trial number, A represents the change of trajectory error (amount of learning), b is the time constant for the error to decrease 67% of the way to the asymptote (inverse of learning rate) and C f is the asymptote of the exponential curve (steady-state error value). Higher changes in trajectory error signify a larger amount of learning, while smaller time constants indicate a higher learning rate. Lower steady state values represent better adaptation performance at the end of the training phase. Note that the fit is applied to all trials of the training phase, not only on the catch trials. However similar results are obtained when only the catch trials are taken into account.

The average of PTE was calculated for the baseline and the evaluation phases to assess the baseline and adaptation performance, respectively. Finally, standard deviation of the peak trajectory error (PTE) for the aforementioned phases was computed in order to investigate the change in movement variability. Similar analysis was done for the washout phase to analyze the persistence of the after-effect for each group. Paired two tailed Student's t-tests on PTE evaluated significant differences between the phases. Unpaired two-tailed Student's ttests assessed differences between the groups. An alpha level of 0.05 was used on all statistics.

3.3 Results

All participants were capable of modifying the direction of their weight-shift to adapt to the 30° angular deviation. During the initial exposures to the visual deviation, participants changed the direction of their weight shift to reach the target towards the very end of the trajectory. Therefore, the first trajectories under exposure were hooked (Figure 3.4, Early learning). Participants were then able to adapt and reproduce a straight-line trajectory at the end of the training phase (Figure 3.4, Late learning). Once the error augmentation is removed in the evaluation phase, training with an amplification gain of 1.5 led to less variability in

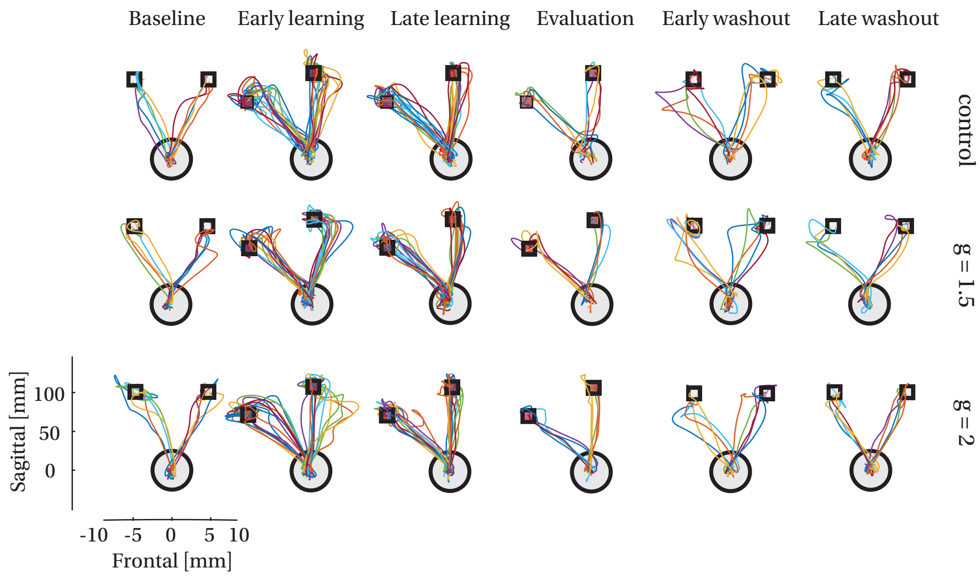

Figure 3.4 – CoM trajectories. Trajectories of representative participants of each group are shown for the main phases of the experiment.

trajectory patterns. In the washout phase, an after-effect of adaptation was observed that seemed to persist in the last washout trials although a bit of decay was observed. Learning curve and washout decay of a representative participant for each group are shown in Figure 3.5.

In-line with studies on arm reaching, an appropriate error augmentation gain has the potential to enhance visuomotor adaptation for directional balance control. The main finding of this pilot study is that the gain of 1.5 significantly reduced the learning time constant compared to the control condition ($p = 0.05$), while the gain of 2 tended to slow down adaptation. In fact, the group with the gain of 1.5 reached 67% of the way to the steady-state error value after an average of 5.5 ± 0.6 trials, while the control group and the group with the gain of 2 needed on average 13.7 ± 2.9 and 26.3 ± 13.9 trials, respectively (Figure 3.6.A).

While the steady state error values are similar for all groups at the end of the training phase, the two experimental groups learned more than the control group (Figure 3.6.B). Moreover, participants who trained with a gain of 1.5 tended to perform better, i.e. smaller PTE average (14.1 ± 1.1 mm), during the evaluation than the control group (16.5 ± 1.1 mm) and the group with the gain of 2 (16.5 ± 3.3 mm).

The same trend applies for the movement variability; PTE variability during the evaluation phase was reduced by training with a gain of 1.5 (4.7 ± 0.8 mm) and increased with the gain of

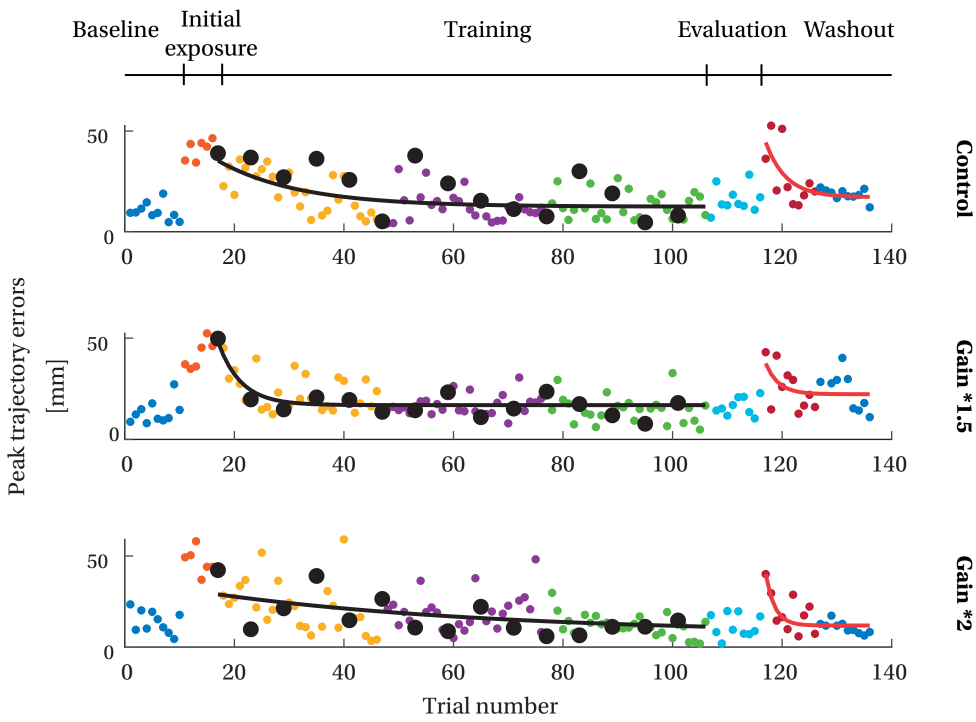

Figure 3.5 – Peak trajectories errors. PTE (dots) of representative participants for each group are shown for each phase of the experiment. Larger dots represent the catch trials. Black and red lines indicate the exponential fit for the training and washout phases, respectively

2 (10.1 ± 4.03 mm) in comparison to the control group (5.9 ± 0.6 mm).

Finally, data from the washout phase suggest that error augmentation with a gain of 1.5 lead to stronger adaptation, since the time constant of the after effect decay was higher (6.3 ± 3.7 trials) than the one of the control group (3.0 ± 1.04 trials) and the one of the group with the gain of 2 (1.8 ± 0.5 trials).

3.4 Discussion

This preliminary study applied visual error augmentation paradigm to a directional balance control task. The goal was to investigate whether error augmentation could be implemented in conventional balance exercises and to refine the optimal gain to boost adaptation to a visuomotor distortion. The error augmentation gain of 1.5 significantly accelerated learning. In contrast, the gain of 2 increased the learning time constant. Both gains seemed to increase the amount of learning, although this did not reach significance in this pilot sample. Our findings provide further evidence that visual error augmentation can have a direct positive effect on learning performance and supports the results of previous studies performed on upper limbs [18, 16][15, 17]. O'Brien tested visual error augmentation with a gain of 1.5 for a lateral weight shifting task and found similar results [124].

Participants were able to change the trajectory of their center of mass to adapt to the visual deviation. They all reached a steady-state in their performance before the end of the training phase. The experimental group with a gain of 1.5 adapted faster than the control, while the group with a gain of 2 took more time to reach the steady-state. Since the training phase was sufficiently long to reach the steady-state, the performances in the evaluation phase were equivalent for all groups. Nevertheless, participants were more consistent in their weight shift movement after training with a gain of 1.5, as they showed a smaller variability of trajectory errors. Finally, the gain of 1.5 might lead to a longer after-effect. While the gain of 1.5 promotes better learning performance, the gain of 2 may overstep the limits where error augmentation paradigm remains efficient for this task. Indeed, a larger gain might lead to over-corrections which reduce the stability of the adaptation process [18]. When incongruences occur between different sensory modalities, human central nervous system tends to rely more on vision, rather than proprioception [204]. Therefore, the magnitude of the gain has a strong effect on the learning performance. If the gain is too high, participants might notice the incongruence between their actual movement and the one displayed on the screen and might mistrust the visual feedback. Consequently, they would rely more on proprioception and decrease the weighting of the augmented feedback. However, when the optimal challenge point is reached, it might also be a source of motivation and self-evaluation. The data analysis focused on CoM movement accuracy as the main metric of task performance. Nevertheless, a deeper investigation of the CoM kinematics, such as reaching time or velocity profile, might be relevant to understand the learning strategies used by the central nervous system.

Our pilot study lends support to applying visual error augmentation in balance training for young, healthy participants. The study should be replicated in a larger cohort, for one to define an optimal error augmentation gain, for another to validate the paradigm for application in clinical populations suffering from balance impairment. Indeed, visual feedback is commonly used for balance training in rehabilitation therapy especially for elderly, and patients suffering from stroke, or Parkinson's disease [211, 212, 168, 173]. It is suggested that patients with Parkinson's disease take more time for re-weighting of somatosensory inputs and have impaired representation of their kinesthetic body map [4]. Therefore, augmenting the kinematic error during a postural control task might help refining the control of their body movements. Since the optimal error augmentation gain seems to be task-specific and even skill-specific [24], understanding the mechanisms correlated to error-detection and correction is essential to provide efficient visual feedback. As a consequence, a second research axis

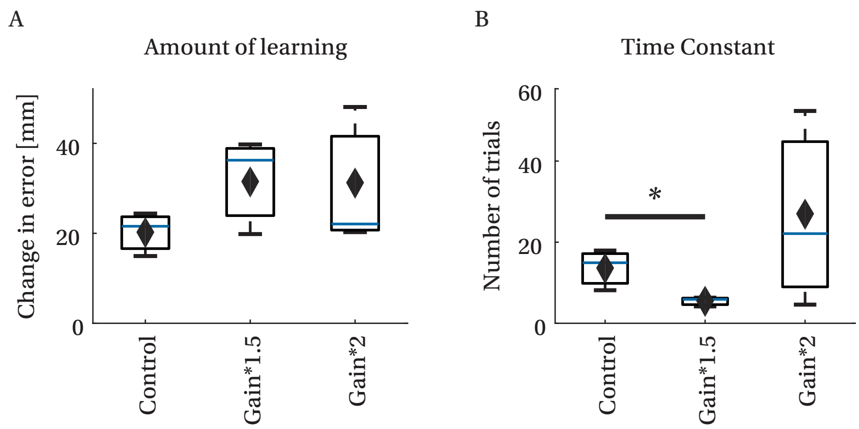

Figure 3.6 – Average effect of the error augmentation paradigm on the learning time constant **(A)** and amount of learning **(B)**. On each box plot, the diamond is the average, the central line indicates the median, the top and bottom of the box represent the 25th and 75th percentiles, respectively, and the whiskers extend to the most extreme data points.

would be to investigate whether a correlation exists between the optimal error augmentation gain and the ability to monitor one's actions.

4 Full-body motor awareness

Motor Awareness is Effector Independent but Strongly Influenced by Kinematic Task Demands

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In preparation.

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Abstract
Conscious action monitoring has been used to probe actual limits of agency by involving movements of different body parts such as the finger or the arm, and more recently of the full-body. As self-consciousness, it is believed that agency is unitary in experience, and the same as been argued for motor awareness in the sense that it is effector independent. However, the effector independent hypothesis has never been tested. The primary goal of this study is to investigate whether motor awareness differs between movements of trunk-centered (full-body) and distal (hand) effectors. Secondly, the effect of cognitive loading on motor awareness and performance was tested. Finally, the spatial accuracy of the effectors was tested in a blind reaching task to seek if there is a relationship between movement accuracy and motor awareness. The results support the effector independent hypothesis but also show a strong effect of kinematic task demands.

4.1 Introduction

The ability to monitor our own actions allows us to discriminate between events that are due to our own movements from those induced by other agents or by the environment. Therefore, motor awareness or action monitoring [128], is the subjective awareness that I am initiating, executing, and controlling my own movements. It is tightly linked to bodily self-consciousness, since action monitoring allows an implicit comprehension of the causal relation between one's self and the world [127]. In general, motor awareness is reinforced when the action outcome corresponds to the expected immediate consequence. Thus, the concept of motor awareness has often been associated with the computational model of sensorimotor control, called the observer framework. This model compares actual re-afferent feedback and the predictive future state to improve sensorimotor control without the need of additional conscious processing. Therefore, movement prediction not only allows to correct online movement error and refine movement internal model, but also to discriminate between our own actions from the ones generated by the environment [128] or other agents [129].
The mechanism behind motor awareness has roused research interest since the early work of Nielsen (1963, 1978), and has been intensely studied since then [26, 213, 138]. Most of the research has investigated action of fingers, hands, or arms [214]. In the aforementioned studies, motor awareness was evaluated in response to sensorimotor mismatches between visual and proprioceptive feedback and the motor commands. In such paradigms, participants are asked to perform repeated goal-directed movements during which their hand is hidden from view, e.g. below a screen and replaced by a virtual hand or marker that could be delayed in time or deviated in space. At the end of each movement, participants judge whether the movement of the virtual hand corresponds to the movement of their actual hand. A psychometric curve is then fitted to the responses as function of delay or deviation to extract

the motor awareness threshold at the point of subjective equality (PSE). These studies have consistently demonstrated that participants attributed feedback that was deviated in space by up to 6.5°-15° to their own movement, even though they corrected their trajectory to compensate for the deviation [129, 139, 140, 149].
Similarly, experiments with temporal delays between the participant's actual hand and the visual feedback illustrated that participants self-attributed movements with up to 150-200 ms of delay [141, 138]. These experiments also brought to light the fact that participants automatically compensated for a spatial or a temporal deviation to reach visual targets. These online-corrections allowing the compensation for the deviation referred as motor performance were generated unconsciously for small deviations or temporal delays.
Aside from reaching movements, motor awareness and the sense of agency have further been investigated for movements such as finger-tapping, or hand-opening and closing using auditory or visual feedback. Sensory attenuation, closely linked to the comparator framework, has been shown to be comparable for sounds generated by both the hand and the foot. Similarly, motor awareness for footsteps generated during walking, evoke similar psychometric responses as auditory feedback for upper-limb movement. More recently the research of Kannape et al. [28, 29] extended these full-body paradigms by studying walking agency using visual feedback. The thresholds in these are again comparable to those involving only upper-limb movements both with respect to temporal and spatial characteristics. These studies were further of relevance as they not only demonstrated that motor awareness thresholds are comparable between goal-directed hand movements and continuous actions involving the full-body, but that participants exhibited a trend to automatically attempt to synchronized with their delayed feedback. Taken together, these studies suggest that motor awareness is independent of the end-effector as well as supramodal, instead representing a central action monitoring mechanism [147, 27]. However, thus far no study has confirmed this effector-independence by studying full-body and hand-related awareness thresholds within individuals.
The **primary goal** of this study is to confirm whether motor awareness is effector-independent and therefore comparable between goal-direct movements of the hand and the full-body, in this case directional control of balance.
The **second objective** of this experiment was to assess whether motor awareness is modulated by kinematic demands and cognitive loading.
Finally, the **third goal** of this study is to capture participants representation of the effector-cursor dynamics in a blind reaching task. Their blind reaching accuracy and variability is determined in order to investigate whether there is a link between the spatial representation and action monitoring abilities.

We hypothesized a similar motor awareness threshold between the two effectors following the effector-independent theory. We expect that motor awareness will be higher when the kinematics demand is reduced, as the cognitive bias of completing the task will “override” the error signal. Concerning the relationship between motor awareness and blind reaching performance, it is also presumed that participants with more accurate representation of the effector-cursor dynamics will be better in action monitoring.

4.2 Materials and methods

4.2.1 Participants

22 healthy adults were enrolled for this study (15 females, mean age = 25 ± 3.9 years, height = 1.66 ± 0.07 m). Participants had intact or corrected to normal vision, intact musculoskeletal system and no history of orthopedic, neural or psychiatric. The experimental procedure described in this study was approved by the Cantonal Ethics Commitee Vaud, Switzerland (project number 2017-00765) in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki.
Two volunteers were discarded from the data analysis. The first one could not complete the balance task, since he felt dizzy. He was the only volunteer to report vertigo. The second exclusion was done post-analysis, since the motor awareness threshold of this participant was below the point of subjective equality for all the deviations, including the control trials with no deviation. It was assumed that he did not understand the task.

4.2.2 Experimental setup

Participants were secured via straps at the pelvis level inside the THERA-Trainer coro [215], a therapy device for safe dynamic balance exercises used to track the movement of the full-body (Figure 4.1). The main advantages of this device are the short setup time and the safety it provides, allowing to run the similar experiments with clinical populations in the future. Its frame is height-adjustable and composed with two spring elements, whose resistance was set to minimal for this study. The behavior of the spring system is comparable to the one of a joystick, meaning that it always pulls back towards the center. No degrees of freedom were locked during the experiment. It is equipped with an inertial measurement unit (IMU) which wirelessly transmits the tilt of the frame at a sampling rate of 30 Hz. The projection of the CoM in the frontal and sagittal direction was estimated by multiplying the pelvis height with the sinus of the roll and pitch angles given by the IMU, respectively.
The upper frame of the THERA-Trainer core was modified to host a joystick, whose position was also streamed at 30 Hz.

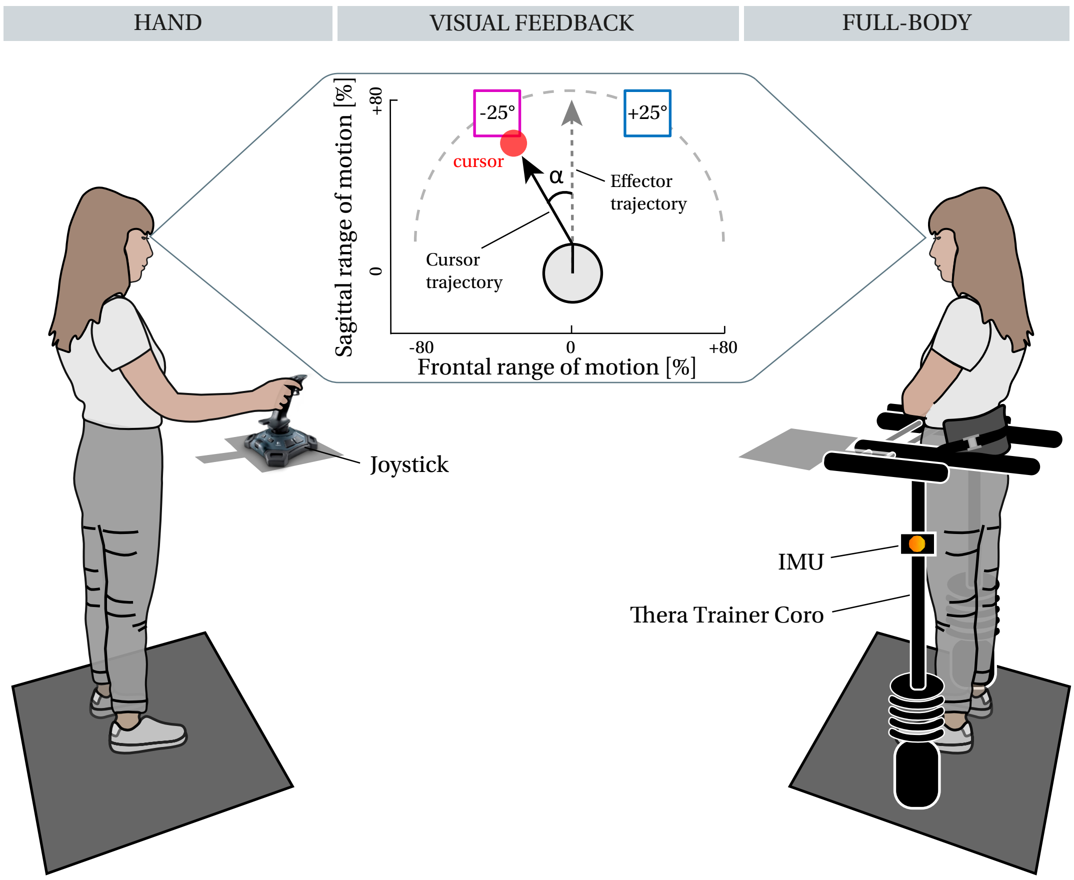

Figure 4.1 – Experimental setup and visual feedback. Participants either controlled the cursor by tilting their whole body inside the Thera-Trainer Coro or by moving the joystick with their dominant hand. In some trials, the cursor trajectory was deviated by an α-angle.

4.2.3 Protocol

This study was divided in two parts. It started with a blind reaching experiment, where participants were asked to reach a target without any visual feedback, and action monitoring study allowing the assessment of participants' MA threshold. The order of the studies was chosen as follows to not bias the state representation with the visuomotor conflicts present in the action monitoring task.

Study 1: Blind reaching

This study was composed of two experimental blocks, a full-body and a hand blind reaching block effectuated in a randomized order (Figure 4.2). A cursor representing either the projection of participant's CoM or the hand position in the transversal plane was displayed on LCD monitor. Each trial started from the center target representing the resting position (i.e.

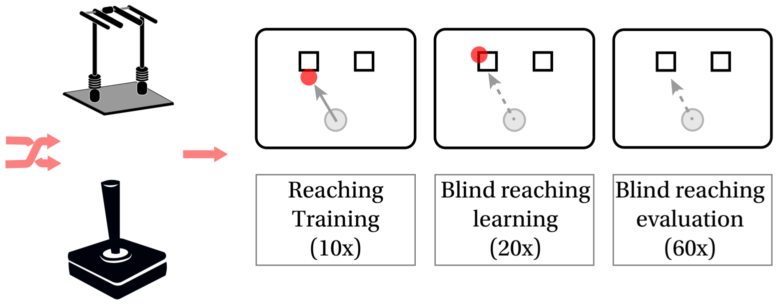

Figure 4.2 – Timeline of the blind reaching task. The effector order was randomized across participants. Participants started the experiment with a reaching training with a concurrent and congruent visual feedback to familiarize with the setup. They were then trained to reach the target with only a terminal feedback. During the evaluation, they had a terminal feedback every 6 trials.

standing straight or center of the joystick). A virtual target appeared on the top of the screen at one of two randomized locations. The two target locations are placed at ±25° from medial axis along a circle with a radius of 80% participant's range of motion. Participants are asked to perform a hand movement with a joystick or to lean with their full-body to reach the target. During the full-body block, they are instructed to keep the feet shoulder-width apart, maintain a static base of support. For the hand block, they used their dominant hand.

All trials begin from the resting position. The experiment started with a training phase where they were asked to reach one of the two virtual targets with a congruent visual feedback. In the second phase (blind reaching learning), the cursor disappeared once the virtual target was presented. Participants were asked to reach blindly the virtual target, stop and stay once they think they reached the virtual target. They received a terminal visual feedback of the cursor position to show them their reaching accuracy. They had 20 trials to improve their reaching accuracy. In the last phase, the terminal visual feedback appeared every 6-trial only.

Study 2: Action monitoring

Participants performed two experimental blocks, a full-body (FB) and a hand (H) action monitoring block, effectuated in a randomized order. These two blocks were divided in two sub-blocks, comprising a single task (ST) and a dual task (DT). The sub-blocks order were also randomized within blocks. The experimental procedure is illustrated in Figure 4.3.

Our action monitoring task is similar to the paradigm used in previous studies to test for motor awareness through perceptual visuomotor conflicts of voluntary movements [26]. As in study 1, they were able to control a cursor either with their hand or by tilting the body. In both cases, they were asked to produce a trajectory as straight as possible. After reaching the target, they returned to the resting position to start the next trials. Return movements to the resting position were not analyzed, since they are assisted by the spring system of the THERA-Trainer coro or of the joystick.

The experiment started with a training phase (20 trials) where they familiarized with the devices and the task. The cursor position is congruent with their movement. Participants then performed the action monitoring training (12 trials) and evaluation (88 trials). In some trials (75%) and beyond a distance of 250 mm from the resting position, the trajectory of the cursor was deviated either clockwise (+ sign) or counterclockwise (— sign) by an angular rotation of 7.5°, 15°, 22.5° or 30°. The amplitude and the side of the deviation was randomized and evenly distributed across targets (88 trials per sub-block, including 24 control trials, i.e. no deviations, and 16 trials per deviation) as in [28]. At the end of each trial, volunteers used the joystick for both blocks to answer a question displayed on the screen asking: “Did the movement you saw on the screen correspond to the movement of you hand/body?” [138].

In the dual task sub-blocks, participants performed the same action monitoring task, while executing a visual stroop task. A color name appeared inside the cursor along the trajectory in 8 different distances from the resting position. The color print was always mismatching from the color name. Participants were asked to say out loud the printed color as fast as possible and then reply to the question quoted above.

4.2.4 Data Analysis

Data are processed offline using Matlab (MathWorks, Natick, Massachusetts, USA) and R [216]. For both tasks, we first analyzed the reaching trajectories. Those are smoothed with a 5-window moving average filter, interpolated and finally averaged across targets and deviations for each participant. Trials where participants restarted their movement (i.e. came back to the center and reached again) were excluded.

4.2.5 Study 1: Blind reaching

The **reaching** error of each trial is defined by calculating distance between the trajectory end point and the target center. The error amplitude is a direct measure of the spatial accuracy. To quantify the overall accuracy of each effector, reaching errors are then averaged across participants. To evaluate the participants consistency in their performance, the **reaching variability** is calculated by taking the standard deviation of the reaching errors and averaged across participants to assess the precision of each effector.

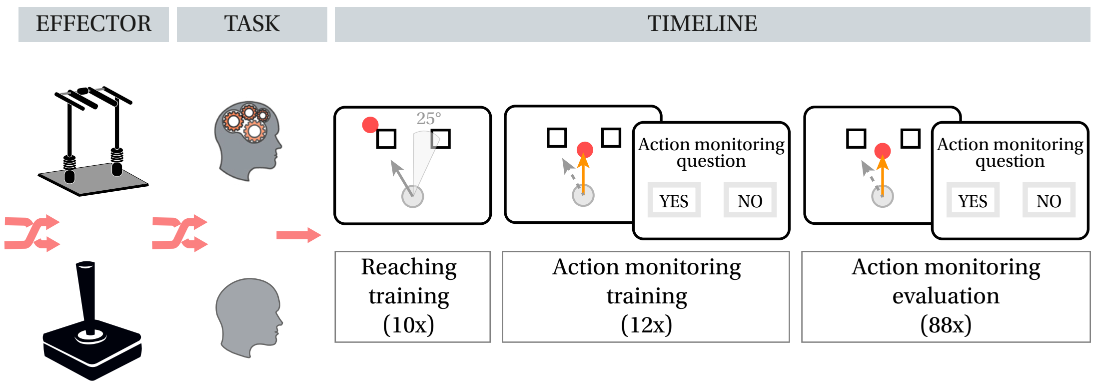

Figure 4.3 – Timeline of the action monitoring task. Participants started in a randomized fashion either by controlling the cursor with the full-body or the hand. The task (single vs dual task) with which they started was also randomized. The experiment started with a familiarization phase, then participants learnt to reply to the motor awareness question, and finally their MA threshold was evaluated. Once they completed both tasks with one effector, they repeated the experiment with the other effector.

4.2.6 Study 2: Action Monitoring

First, we analyzed the responses to the action monitoring question. **Motor awareness (MA)** is defined as the number of yes-responses out of all trials of the same deviation. Correct self-attribution or MA is a "yes" response for non-deviated trials and a "no" response for deviated ones (Equation 4.1). The MA threshold was determined by fitting a psychometric curve to the participants' responses and looking at the 50% point of subjective equality (PSE). PSE is then averaged across participants.

$$MA_\alpha = \frac{1}{N}\sum_{i=1}^{N}\left(f(r) = \begin{matrix} 1, r = Yes \\ 0, r = No \end{matrix}\right) \tag{4.1}$$

r = response, α = angular deviation [°], N= number of trials

Second, we measured the trajectory endpoint to assess **motor performance (MP)** (Equation 4.2). The trajectory endpoint corresponds to the effector position when reaching the target radial distance, which is not equivalent to the trial endpoint (i.e. center of the target). MP was defined as total angle compensated by the participants considering the endpoint of each movement trajectories and measured from the onset of the deviation, which was 25 mm from the resting position.

$$MP_{\alpha} = \frac{1}{N}\sum_{i=1}^{N}(c_i) \tag{4.2}$$

c = motor compensation [°]; α = angular deviation [°], N = number of trials

Finally, as additional variables, we analyzed the reaching and response time. The reaching time was calculated from movement onset to end of the trial. The movement onset was determined as the radial velocity being greater than 3% of the maximum trajectory velocity. The response time corresponds to the time from the appearance of the question to the joystick click.
The aforementioned dependent variables are analyzed as functions of the independent variables such as Deviation, Deviation Side, Target Side, Task, Mode and Effector. Deviation Side is defined as the direction of the deviations (i.e. clockwise or counterclockwise). The target side is the position of the target with respect to the midline (left or right). The mode defines if the deviation converges towards or diverges from the target position.

Statistical analysis

Statistical analysis were performed with R [216]. Mixed-effects models for repeated-measures ANOVA were conducted on MA, MP, and the two observatory variables with the factors Deviation, Deviation Side, Target Side, Task, Mode and Effector as within-subjects factors. The first model included the interaction Target Side × Deviation Side for the deviated trials. The second model tested the effect of Target Side for the control trials. The third model included the factor Deviation and the interaction Task x Effector. Finally, the interaction Mode × Effector was analyzed in the last conducted mixed-effects model.
Paired Student t-test were used to assess significant differences between full-body/hand reaching accuracy and variability.
For correlations analysis, we use a robust procedure called Pearson's skipped-correlations [217]. This method protects against bivariate outliers. We measure the strength of the association between average reaching error and PSE of each participant.

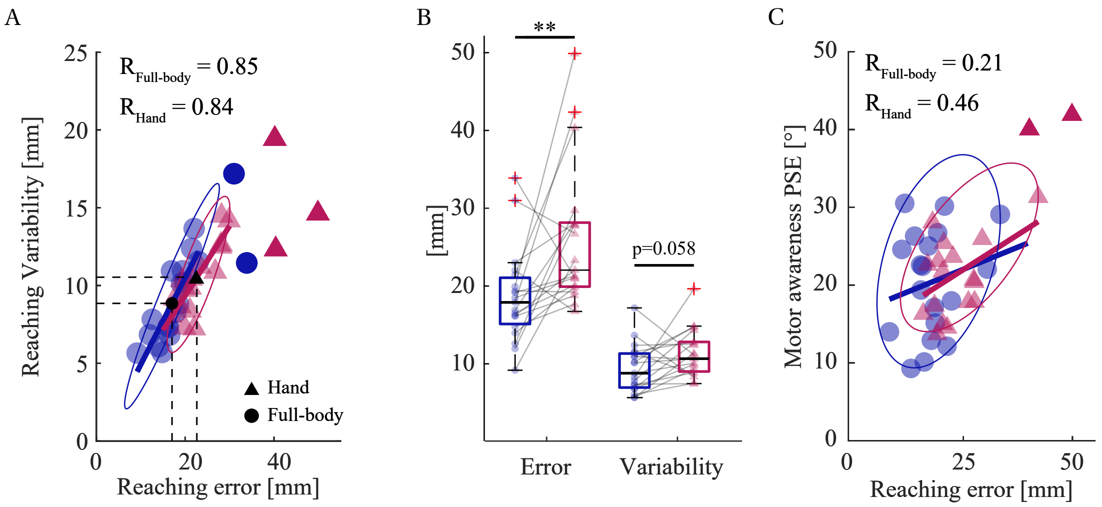

Figure 4.4 – **(A)** Blind reaching error and variability of each participants and for the full-body (blue dots) and the hand (pink triangles). Samples were obtained by averaging the 60 blind reaching trials of each participants. Opaque samples represent outliers, ellipses are the robust data centers and shaded areas are the 95% bootstrapped confidence interval. **(B)** In average, blind reaching with the full-body lead to more accuracy (smallest reaching error) and more precision (lowest variability), with the hand. **(C)**. Association between reaching error and motor awareness. Plotting symbols and shading conventions are the same than in A.

4.3 Results

Study 1: Blind reaching

Blind reaching error and variability of each participant and each effector are shown in Figure 4.4.A. This two metrics are highly correlated for the two effectors (Pearson's skipped $R > 0.84$). This means that participants with small reaching error present also a low reaching variability. Interestingly, the average reaching error was significantly lower for the full-body than for the hand ($p=0.007$). This suggests a better spatial accuracy for the full-body compared to the hand. Although the reaching variability was higher for the hand than for the full-body, the change was not significant ($p=0.058$), see Figure 4.4.B.

Study 2: Action monitoring

Motor awareness

The first step was to verify whether motor awareness during non-deviated was not significantly affected by independent variables such as Target Side ($p=0.7$). Then, the interaction between

Target Side and Deviation Side (i.e. sign of deviations) was tested in order to assess whether targets could be collapsed in a single location and absolute values of the deviation could be taken. A main interaction between Target Side and Deviation Side was found ($p<0.001$). Thus, the percentage of yes-responses was significantly higher when the deviation side converged towards the target side, i.e. when motor compensation was towards the center. Since, this interaction was very strong, it would not be wise to collapse the targets and take the absolute deviation. In order to keep the analysis consistent, we cancelled the Target Side effect by defining converging trials with negative deviation sign and diverging trials with positive deviation sign.
Participants correctly self-attributed 89.9 ± 0.3% and 91.3 ± 0.3 (mean ± SEM) of the non-deviated trials for the FB and H, respectively. This percentage significantly decreased as the angular deviations increased (main effect of Deviation, $p<0.001$, see Figure 4.5.A.
We then tested the effect of Task and Effector. For non-deviated trials, we failed to detect a main effect of Task or Effector, nor an interaction ($p=0.42$ and $p=0.65$, $p= 0.54$ respectively, see Figure 4.5.B). For deviated trials, an interaction was found ($p=0.02$). Indeed, the difference in motor awareness between the two effectors was stronger during the single task, where MA was significantly lower (better) for the full-body ($p<0.001$). MA for the hand significantly decreased when performing the dual task ($p=0.007$), see Figure 4.5.B.
We tested the effect of kinematic demands (Mode) and effector on the MA threshold for deviated trials (Figure 4.7.A). Deviations were split in two modes (converging and diverging) and their absolute value were taken. A significant effect of Mode ($p<0.001$), as well as an interaction Mode x Effector ($p=0.001$), was detected. MA was significantly better (lower) when deviations were diverging. The interaction was driven by a significant difference in diverging trials between the two effectors. While the MA thresholds were similar for converging trials, MA was significantly better for full-body (PSE of 16°, against 19.9° for the hand) during diverging trials ($p<0.001$).

Finally, we investigated whether there was an association between the participants' ability to monitor their action (motor awareness PSE) and their reaching accuracy (Figure 4.4.C). The Pearson's skipped correlation was 0.21 for the full-body and 0.46 for the hand, which represent low association factors.

Motor performance

Figure 4.6.A shows the trajectories of the full-body and the hand in the sagittal plan from the resting position to each of the two targets for non-deviated trials. The reaching performance for non-deviated trials regardless of the task was similar across participants and significantly better for the hand than the full-body (Main effect of Effector, $p<0.001$, $error_{hand}$=2.49°±0.06; $error_{full-body}$=4.01°±0.1, mean±sem). The dual task significantly reduced reaching error

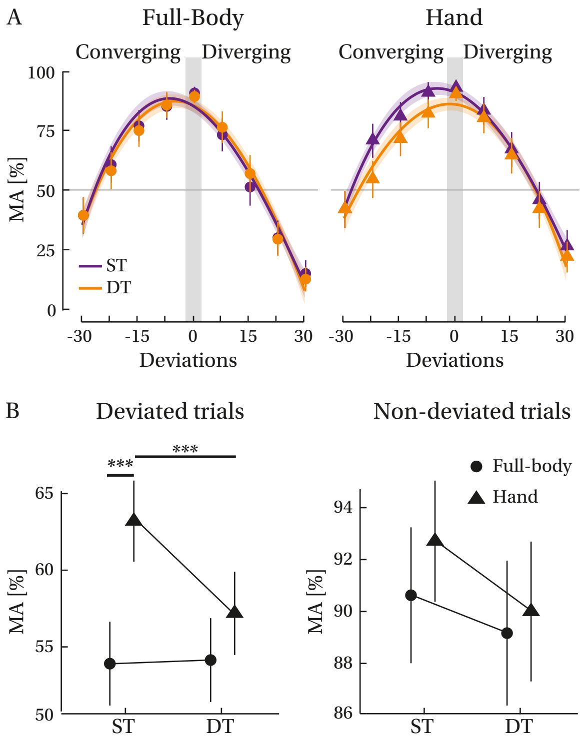

Figure 4.5 – **(A)** Motor awareness for the Full-Body (round markers) and the Hand (triangle markers) for each deviation. Grey shaded bands represent the control trials which don't belong to one of the two modes (e.g. converging and diverging). **(B)** Averaged MA across participants for deviated and non-deviated trials in single and dual tasks.

independently of the effector (main effect of Task, $p=0.01$). There was no interaction Task x Effector for non-deviated trials ($p=0.38$).
Trials with angular deviations (Figure 4.6.B) changed participant's reaching direction. Thus, trajectories were deviated in the opposite direction of the angular deviation, as reported for hand [26, 218, 219] and walking [28, 149] goal-directed tasks. Motor performance significantly increased as the angular deviation augmented (main effect of Deviation ($p < 0.001$), see 4.6.D. A main effect of Task and Effector was detected for the deviated trials ($p=0.02$ and $p<0.001$, respectively), but no interaction was found ($p=0.5$). Motor compensation was higher for the hand and during the dual task.
Target Side did not affect motor performance of control trials ($p=0.09$). However, we observed an interaction Target Side × Deviation Side ($p<0.001$) for deviated trials. By looking the effect of mode, we observed that MP was significantly higher when the compensation was towards the center, i.e. when the deviation side converged towards the target side (see Figure 4.7.B). Therefore, a main effect of Mode was detected ($p<0.001$).

Additional variables

Although we recorded response times, participants were not asked to reply to the motor awareness question as quickly as possible. They also could move to target at a speed that was comfortable for them. For the response time, a main effect of Effector was detected ($p<0.001$). Participants responded significantly faster when the effector was the hand (Figure 4.8.A). The response time was not affected by the deviations' amplitude ($p=0.34$).
For the reaching time, a main effect of Deviation, Task, Effector, and Mode was detected, as well as an interaction Task × Effector ($p<0.001$). The reaching time augmented significantly as the deviation increased ($p<0.001$), see Figure 4.8.B. In addition, the reaching time was significantly higher for the full-body than for the hand ($p<0.001$). The dual task significantly decreased the reaching duration. Finally, the reaching time was lower when the deviations were converging.

4.4 Discussion

In this study, we assessed and compared motor awareness for the full-body and the upper limbs, as well as their representation of the cursor dynamics, in a matched, goal-directed task. The primary objective was to test the effector-independent hypothesis by investigating, for the first time, comparable motor-tasks, both for the hand and the full body, within a single cohort of participants. Based on this hypothesis, a general control strategy, independent of the effector (arms, fingers, locomotion), is employed by the central nervous system to generate goal directed movements [27, 28]. Our results support the effector-independent hypothesis,

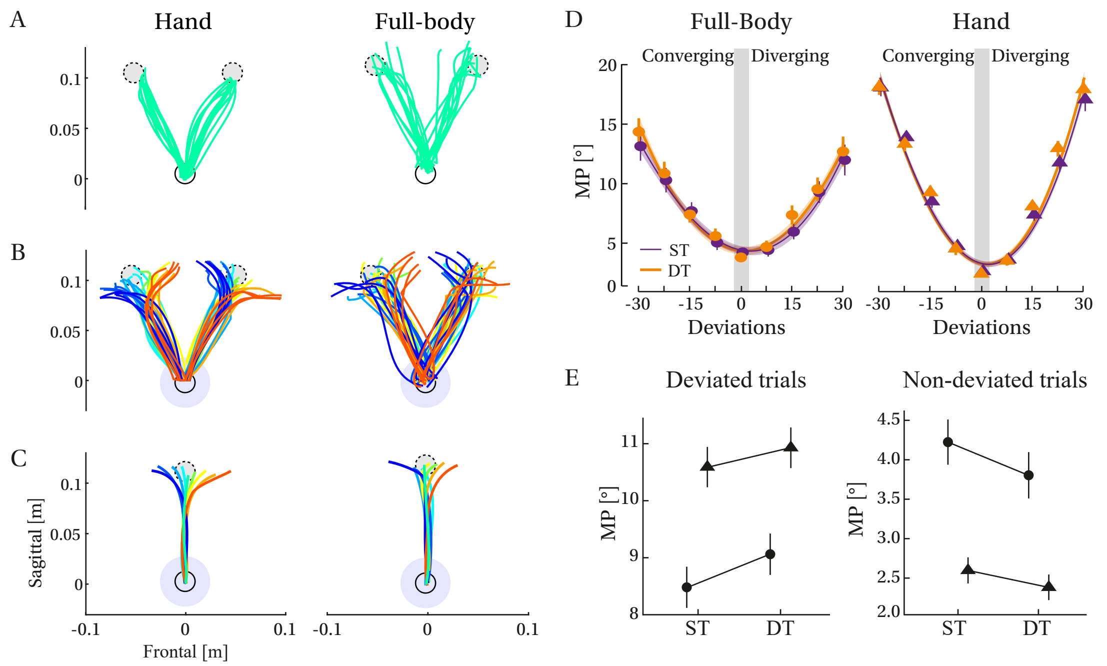

Figure 4.6 – Motor Performance. **(A)-(C)**. Effectors trajectories of a single representative participant during the single task. A. Effectors trajectories from the resting position (dot with solid line) towards the two target locations (dot with dashed line) in non-deviated trials. **(B)** Effectors trajectories towards the targets for trials with deviations to the left (red tone lines) and to the right (blue tone lines). The shaded blue dots represent the distance by which the deviations start. **(C)** Average trajectories of this participant after having centered the two targets. **(D)** Averaged MP across participants for full-Body (round markers) and hand (triangle markers) for each deviation. **(E)** Averaged MP across participants for deviated and non-deviated trials in single and dual tasks.

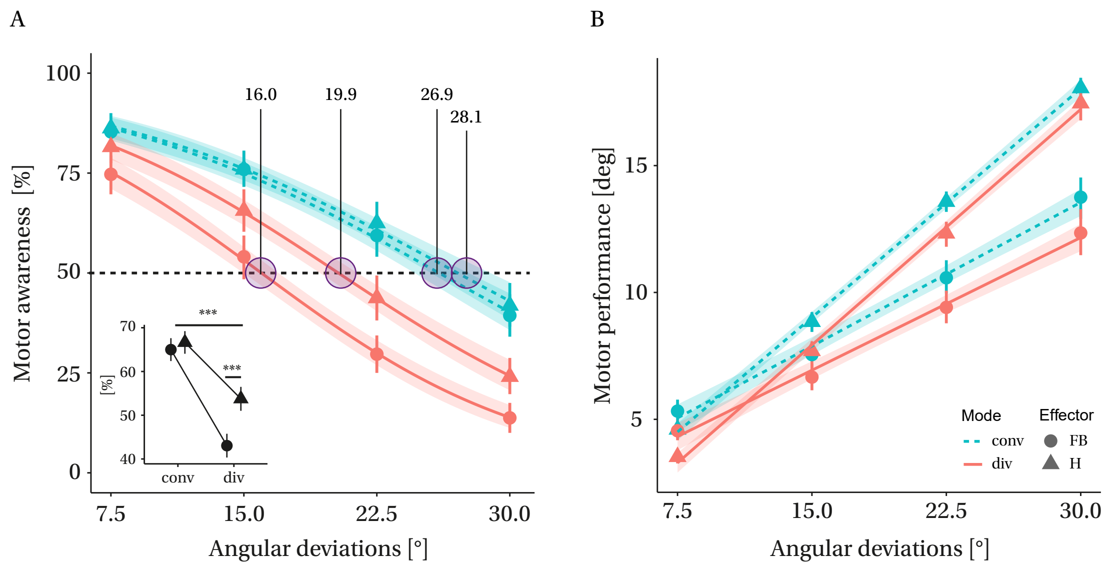

Figure 4.7 – Effect of kinematic demands on motor awareness **(A)** and motor performance **(B)**. The two effectors have similar motor awareness thresholds for converging trials, while MA threshold was better for full-body during diverging trials. Motor performance is better for the hand and during converging trials.

but showed that MA is strongly influenced by kinematic task demands. Finally, our results suggest a better reaching accuracy for the full-body in comparison to the hand.

Our findings demonstrate that MA for both full-body and hand movements was strongly modulated by kinematics demands. In fact, participants self-attributed deviated trials more often when deviations converged towards the targets, and participants thus needed to compensate towards the center leading to less a demanding task effort. In other words, their MA became worse due to the bias of converging on and reaching the target more easily. Motor performance, on the other hand improved when the kinematic demand was reduced. These results are supported by other studies showing that participants make more MA errors when MP is higher, minimizing the error of the visual feedback [149]. On the contrary, studies have found that in case of intentional effort, agency can be enhanced [220, 221].

Regarding the effector-independent hypothesis, we compared the MA threshold of the two effectors for converging and diverging modes. MA thresholds were similar for converging trials, but differed for diverging ones. Indeed, MA threshold for the hand was significantly higher than for full-body in diverging mode. Several limitations of the study suggest that it might not be due to the difference in effectors, but rather to differences in motor performance and reaching speed. MP was better for hand, and thus reduced the error of the visual distortion as discussed above. It might also be due to the fast reaching time observed in hand movements. Although we asked participants to perform straight-line and controlled movements for both effectors,

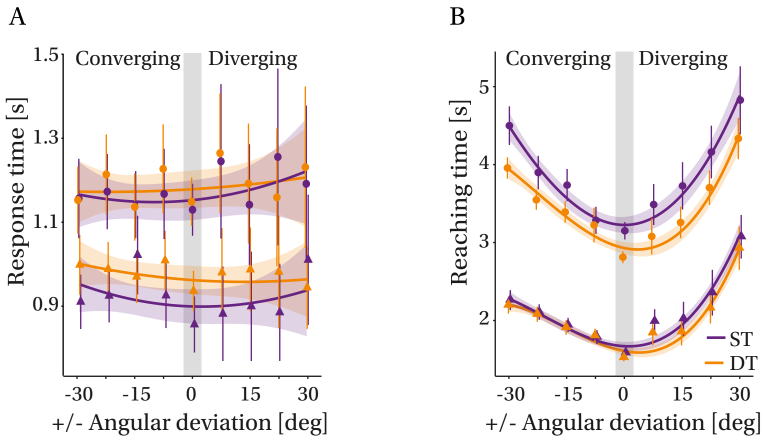

Figure 4.8 – Time needed to answer the action monitoring question **(A)** and to reach the targets **(B)**, from the moment participants start moving from the resting position, averaged across participants for each angular deviation.

the reaching speed was significantly higher for the hand, which could affect the sense of agency. It has been demonstrated the mindful movements described as continuous monitoring of present action are associated with slower speeds allowing to monitor moment-by-moment the quality of the performance [222]. While motor corrections were still possible, the fast reaching speed could have impaired the action monitoring the hand. Speed can therefore modulate perceptual motor awareness. Therefore, one main limitation of this current study was that the speed was not controlled. Hence, our current results don't argue against the effector-independent nature of motor awareness.

We found that cognitive loading significantly affected motor awareness, however post-hoc analysis does not reveal a clear nor a significant trend. Hon and colleagues [223] also observed lower MA ratings in high working memory load, suggesting preoccupied minds feel less in control. These results are contradictory to the study of Kannape 2014 and 2017, who found that walking motor awareness was impaired (increase of MA threshold) by cognitive loading [149, 148]. While there is no clear consensus on how taxing cognitive resources may affect the MA threshold , all studies agree that MA is modulated by cognitive loading. These results suggest that the sense of agency depends on the availability of conscious cognitive resources. The difference response times between both effectors may stem from the simple fact that the joystick was used to answer both for hand and full-body movements.

In previous action monitoring studies [26, 224, 213], the effectors are often hidden below a screen and the visual feedback is given in the same plane than the actual movement. In our study, participants had the monitor display in front of them at eye level. Although their hand or body were not hidden, they could not visually monitor their actual movements when

Table 4.1 – Results overview - Full-body and Hand motor awareness

			Condition				**Effects**					
mean±SEM			**ST**	**DT**	**Conv**	**Div**	**Deviation***	**Task****	**Effector****	**Task by Effector****	**Mode*****	**Mode by Effector**
MA [%YES]	Control	FB	90.6±0.3	89.2 ±0.3	NA	NA	NA	0.42	0.65	0.54	NA	NA
		H	92.7±0.3	90.0±0.3	NA	NA						
	Deviated	FB	53.9±0.5	54.1±0.5	65.0±0.5	43.0±0.5	**<.001**	0.9	0.1	**0.02**	**<.001**	**.001**
		H	63.2±0.5	57.2±0.5	66.6±0.5	53.7±0.5						
MP [°]	Control	FB	4.2±3.2	3.8±3.3	NA	NA	NA	**0.01**	**<.001**	0.38	NA	NA
		H	2.6±1.8	2.38±1.8	NA	NA						
	Deviated	FB	8.5±6.6	9.1±6.6	9.3±6.6	8.2±6.5	**<.001**	**0.02**	**<.001**	0.5	**<.001**	0.91
		H	10.6±6.4	10.9±6.5	11.3±6.2	10.2±6.7						
Reaching time [s]	Control	FB	3.1±1.2	2.8±0.8	NA	NA	NA	**<.001**	**<.001**	**.001**	NA	NA
		H	1.6±0.6	1.5±0.5	NA	NA						
	Deviated	FB	3.9±1.7	3.5±1.3	3.9±1.5	3.6±1.5	**<.001**	**<.001**	**<.001**	**<.001**	**<.001**	0.69
		H	2.1±0.9	2.0±0.8	2.3±0.9	1.9±0.8						
Response time [s]	Control	FB	1.13±0.7	1.14±0.6	NA	NA	NA	0.6	**<.001**	0.26	NA	NA
		H	0.85±0.7	0.94±0.5	NA	NA						
	Deviated	FB	1.17±0.7	1.18±0.7	1.17±0.7	1.18±0.7	0.34	0.4	**<.001**	0.3	0.7	0.96
		H	0.93±0.6	0.98±0.5	0.95±0.6	0.96±0.5						

Mean, standard error of the mean for the different conditions and results of separate mixed-effects models for repeated-measures ANOVA run for motor awareness, motor performance, reaching time and response time (p-value <0.05 are represented in bold)

looking at the screen. A limitation of this study was that the visual feedback was provided in the sagittal plane (screen in front on the participant) while it represented the projection of the hand and CoM in the transversal plane. Since individuals are constantly exposed to this kind of orientation mismatch, by using a mouse to control a cursor on a computer screen for example, cognitive burden related to this transformation should be minimal. Finally, the representation of effector-cursor dynamics was determined through a blind reaching task. Reaching error was used as a direct measure of the hand proprioceptive accuracy and the full-body proprioceptive-vestibular accuracy, while reaching variability assessed movement precision. Reaching error was significantly lower for the full-body suggesting better representation of the cursor dynamics. This might be explained by the additional contribution of the vestibular system. Reaching variability was also lower for the full-body. For both effectors, reaching error and variability were highly correlated supporting the control framework theory that the central nervous system minimizes trajectory error based on sensory feedback [113]. These findings corroborate the results of a more applied study suggesting that trunk-machine interface outperforms joystick to control both simulated and real drones [225].
We evaluated if MA was linked to the effector-cursor representation. Although it seems that better representation of cursor dynamics leads to better MA, these results remain inconclusive due to the confound of reaching speed. Indeed, the speed/accuracy trade-off might have affected both metrics.

This is the first study investigating effector-independence. Ours findings are in line with a general central monitoring strategy that is not specific to the effectors. However, we strongly believe that controlling for the speed will contribute to stronger evidences.

5 Relation between error augmentation learning and conscious motor awareness

On the Relation between Error Augmentation Learning and Conscious Motor awareness in a Goal-Directed Balance Task

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In preparation.

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Abstract
Despite the use of visual feedback for balance rehabilitation, visual error augmentation dedicated for balance tasks remains poorly explored. In this study, we investigate the effect of four error augmentation gains on the adaptation time in goal-directed balance task. In addition, we explored whether motor awareness could be a potential factor for predicting an optimal error augmentation gain. Therefore, we assessed the motor awareness threshold during a matched balance task. We found that an error augmentation gain of 1.5 significantly decrease the adaptation time. Moreover, the results suggest the optimal gain may be linked to motor awareness threshold. One could think to first assess motor awareness of the participants and then predict the optimal EA gain.

5.1 Introduction

Human balance requires constant postural adjustment of the whole body to keep the center of mass over its base of support. However, this basic function is often affected by natural ageing and various pathologies such as Parkinson's disease or stroke [202, 203]. In order to alleviate the burden linked to the risk of falls, research on motor learning seeks to understand the content and the structure of balance training that facilitate learning. With the emergence of in-home rehabilitation thanks to exergames, balance training could benefit of augmented visual feedback if provided the right way.
Research on motor learning suggests that the central nervous system relies on sensory error detection to adapt to the environment [113, 84] or to learn a new motor skills [83, 54]. Hence, error augmentation (EA), which consists of amplifying online trajectory errors with haptic feedback though the use of robotic devices [117] or with visual feedback [16], has been proposed as promising strategies to facilitate motor learning and enhance neurorehabilitation. However, research on error augmentation has essentially been limited to studies for upper limb motor-rehabilitation. The main reason might be that transferring this paradigm to locomotion or balance training requires additional safety efforts since larger movement errors could lead to falls. Only with the recent emergence of robotic devices assisting gait training [226], studies have started to investigate EA with haptic feedback for lower-limbs, for example by challenging controllers applying robotic forces to increase sensory errors during locomotor tasks [114, 121, 24, 21]. Findings with healthy individuals demonstrated that error amplification during walking accelerated adaptation rates [114]. Marchal-Crespo and colleagues pointed out that EA is advantageous for proficient individuals if the locomotor task is complex [121], while unskilled participants could benefit from it if the task is simple [24]. A task is defined as complex when the skill improvement requires long hours of practice, high attentional demands [100] and nested redundancy [227]. Along these lines, it has been shown that haptic EA impaired motor learning during a complex balance task [94] and negatively

affected participants motivation, and thus could be detrimental to motor learning [120, 94].

Although visual error augmentation has demonstrated promising results in upper limbs [16, 18, 17, 22, 228], it has been poorly explored in lower-limbs, especially for balance training. With the emerging exergames in balance rehabilitation, the use of visual feedback is predominant nowadays, and therefore a better understanding of how it should be provided to enhance learning is essential. Recently, [124] applied time-dependent visual error augmentation on a weight-shifting task. They found that the error augmented feedback reduced subjects' time to reach their steady-state performance compared to an unaltered visual feedback. Our pilot study aimed to investigate whether visual EA enhances visuomotor adaptation during a common rehabilitation exercise consisting of dynamic weight shifts to bring a cursor onto virtual targets [229]. Although the experiment was run in a small cohort, the results were promising, suggesting that visual error augmentation could potentially improve adaptation if the visual EA gain is adequate. The effect of two visual error augmentation gains were tested (1.5 and 2), in addition to the control group (gain of 1). Those gains were chosen based on previous studies [124, 16]. In line with previous studies [18, 16], it was found that visual EA can destabilize the adaptive process if the gain is too high. Indeed, large initial errors during practice may prevent the internal models used by the CNS from converging towards the correct motor command (Sanger 2004). It is yet unclear whether there might be an optimal EA gain, and whether it is task-dependent or related to other factors such as skills or error perception.

In this study, the **first aim** is to investigate whether visual EA can promote a faster adaptation rate during a balance task. Although the task is a goal-directed movement as in upper limbs studies where visual EA revealed to be efficient, it is slightly more complex since it involves multiple sensory inputs such as visual, vestibular, and proprioceptive. It also requires a high degree of coordination among multiple body segments in the upper and lower body. Since haptic EA showed mixed results for complex locomotor and balance tasks, it is relevant to investigate whether visual EA is an appropriate strategy for a simple balance task.
The **second objective** is to determine if an optimal learning gain exists for this task. Since previous studies suggested the learning rate was impaired by high EA gains, five EA gains including the control gain of 1, were tested in order to find at which gain is the learning time constant (i.e. the number of trials needed to reach the adaption plateau) is minimal.
Finally, our **third goal** is to explore which factors could predict the optimal EA gain. One possible contender, combining with task complexity or skill, may be conscious action monitoring. Fourneret and Jeannerod in 1998 have demonstrated that humans unconsciously perform motor corrections up to a certain amplitude, called the motor awareness threshold [26]. Visual EA could at times push the errors over the motor awareness threshold. This could help conscious detection of trajectory error and promote explicit learning. On the other hand, it may decrease the feeling of being in control of the visual feedback, which may lead

frustration and impact the adaptation process. The ideal EA gain might be just high enough to favor automatic sensory error detection (implicit learning). A first step to answers those questions is to assess the motor awareness threshold of each participants in a similar balance task and observe how it relates to motor learning, and the EA gain.

We hypothesize that visual error augmentation during a goal-directed balance task will decrease, i.e. improve the time needed to adapt to a visuomotor conflict, while reaching the same level of adaptation as the control. Additionally, we expect that the learning time constant as function of the EA gains will liken a u-shaped curve with a minimum around 1.5, based our pilot study [229]. Finally, an open question is whether there is a relationship, based on the observer model, between sensorimotor learning and our motor awareness thresholds.

5.2 Materials and methods

5.2.1 Participants

57 healthy adults were enrolled for this study (32 females, mean age = 26.3 ± 4.01 years, height = 1.69 ± 0.09 m). Participants had intact or corrected to normal vision, intact musculoskeletal system, and no history of orthopedic, neural or psychiatric conditions. The experimental procedure described in this study was approved by the Cantonal Ethics Committee Vaud, Switzerland (project number 2015-00092) in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki.
Overall 8 participants (15%) were excluded based on the following criteria: One participant was excluded from data analysis because he was not able to focus during the task (i.e. drowsiness). Three participants were excluded because they did not perform the action monitoring task correctly. They presented mean awareness ratings above the point of subjective equality even for the highest deviations (30°). One participant was excluded because they did not adapt to the visual distortion. Finally, three participants were discarded from the analysis, because the trajectory errors did not follow an exponential decay and had an adaptation rate higher than 100 trials (see details below). These exclusion criterion were not predetermined, but were included during data processing. Each group comprise therefore 10 volunteers, except the group with the gain of 1.5 which has 9 participants.

5.2.2 Experimental setup

The Thera-Trainer coro was used to track the tilting motion of the body. For details, see section 3.2.1.

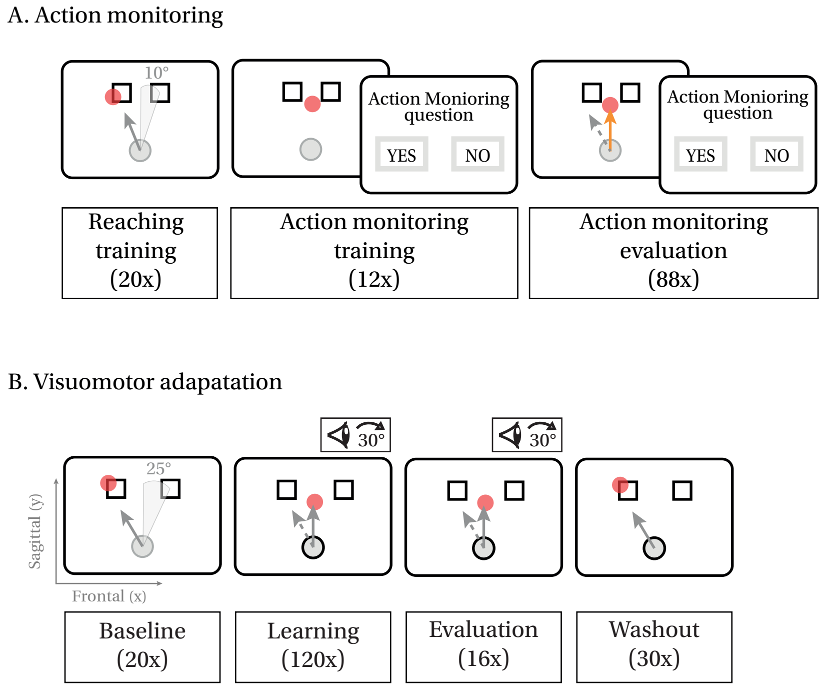

Figure 5.1 – Experimental timelines. A terminal feedback about reaching time was given during the reaching training of the action task and the baseline of the visuomotor adaptation task. For the other phases, participants received a terminal feedback only if they were too fast. Note that the targets position differs between the two tasks.

5.2.3 Protocol

All participants performed the action monitoring and the visuomotor task described below. For the visuomotor task, they were randomly assigned to one of 5 groups: a control group and 4 experimental groups which train with different error augmentation factors (1.25, 1.5, 1.75, and 2). A gain of 1 represents no error augmentation (control group) while a gain of 1.5 represent a 50% amplification for example.

All participants started with the action monitoring task and then performed the visuomotor task after a five-minutes break.

Task 1: Full-body action monitoring

The protocol was similar to the one described in section 4.2.3 for the full-body (FB). The targets were displaced towards the center at a location of ± 10° (instead of ±25°) from the midline. The reason was to try to minimize the effect of kinematic demands, and therefore have an unbiased measure of the motor awareness threshold. The experimental timeline was exactly the same, except that it was done only for the full-body (see Figure 5.1.A). They started with a

training with undistorted feedback, learned to answer the action monitoring question, and then performed action monitoring evaluation (88 trials). In some trials (75%) and beyond a distance of 250 mm from the resting position, the trajectory of the cursor is deviated either clockwise (+ sign) or counterclockwise (- sign) by an angular rotation of 7.5°, 15°, 22.5° or 30°. The amplitude and the side of the deviation is randomized and evenly distributed across targets (16 trial per deviation and 24 control trials, i.e. no deviations).
At the end of each trial of the AM training and evaluation phases, participants used a mouse to answer a question displayed on the screen asking: "Did the movement you saw on the screen correspond to the movement of you hand/body?" [138].

During the reaching training phase, participants received feedback on their reaching time to tell them if they moved too fast, too slowly or at the desired time (2.5 s < Reaching time < 4 s). The goal was to maintain comparable movement velocities across participants. During the other phases, participants received a feedback only if they were too fast.

Task 2: Visuomotor adaptation

The protocol was similar than the one described in section 3.2.3, except that the initial exposure phase was removed and that number of trials in each phase was increased (see figure 5.1.B). In contrary to the full-body action monitoring task, the targets position was kept at ±25° from the midline to promote movement control and prevent "forward falling" movements.
They started with a training phase of 20 trials, where they received a terminal feedback on their reaching time as for the AM task. During training and evaluation, all participants had to reach the targets in presence of a visual angular deviation of 30°. The onset of the deviation started beyond a distance of 250 mm from the resting position (see Figure 3.1.A). A perpendicular vector from the ideal straight-line trajectory was used to define the online error. During the training phase (120 trials), the cursor location was moved by a gain of the current error vector for the four experimental groups (see Figure 3.1.A). The control group had a gain of 1, and therefore experienced the angular deviation without visual error augmentation. During the training phase, every six trials were catch trials, meaning that EA was not present. This allows to have similar trials between groups through the training phase to see how fast they adapt to the visual distortion. Training was followed by an evaluation phase (16 trials) where EA was removed for all group. Finally, in the washout phase (30 trials), the visual distortion was removed and EA was not present. The goal of the washout phase is to evaluate the after-effect.

5.2.4 Data analysis

The data analysis for the action monitoring task was performed the same way than in section 4.2.6. Motor awareness (MA) is defined as the number of yes-responses out of all trials of the same deviation. Correct self-attribution or MA is a “yes” response for non-deviated trials and a “no” response for deviated ones. The MA threshold was determined by fitting a psychometric curve to the participants’ responses and looking at the 50% point of subjective equality (PSE). The PSE was assessed for converging and diverging trials and averaged across participants to observed the effect of kinematic demands. However, the general PSE (all trials combined) of each participant is used study the motor awareness and learning association.

For the visuomotor task the analysis is described in details in section 3.2.4. The evolution of peak trajectory errors (PTE) was used as the main metric of adaptation. An exponential decay was fitted to the training and washout phases:

$$e_i = A\exp(-i/b) + C \tag{5.1}$$

Where i is the trial number (1-120), a represents the change of trajectory error (amount of learning), b is the time constant (TC) for the error to decrease 67% of the way to the asymptote (inverse of learning rate) and C is the asymptote of the exponential curve (steady-state error value). Higher changes in trajectory error signify a larger amount of learning, while smaller time constants indicate a higher learning rate. Lower steady state values represent better adaptation performance at the end of the training phase. Note that the fit is applied only to the catch trials.

The mean peak trajectory error of the baseline and evaluation phase was computed, as well as the variability by taking the standard deviation of the PTE.

The statistical analysis was performed with r (R Core Team 2014). For the action monitoring task, mixed-effects statistical models were conducted on motor awareness and performance to assess the effect of Deviation, Deviation Side, Target Side, Task, Mode and Effector. The first model included the interaction Target Side × Deviation Side for the deviated trials. The second model tested the effect of Target Side for the control trials. The third model included the factor Deviation and the interaction Task x Effector. Finally, the interaction Mode × Effector was analyzed in the last conducted mixed-effects model.

Mixed-effects models were also conducted on the learning time constant, amount of learning, and steady state, as well on the mean PTE and the PTE variability for baseline and washout phases to evaluate the effect of EA and PSE. We also tested the interaction EA x PSE.

Table 5.1 – Results overview - Full-body motor awareness

	Angular Deviation					Condition		Effects		
Mean±SEM	**0°**	**7.5°**	**15°**	**22.5°**	**30°**	**Conv**	**Div**	**Mode**	**Deviation**	**Mode by Deviation**
MA [%YES]	94.1±0.7	79.1±1.5	55.2±1.8	28.9±4.5	8.6±1.0	52.7±1.3	33.1±1.2	**p<.001**	**p<.001**	0.62
MP [°]	3.4±0.1	4.7±0.1	8.9±0.1	14.2±0.1	20.9±0.1	12.2±0.17	12.1±0.18	p=0.3	**p<.001**	p=0.3
Reaching time [s]	2.7±0.03	2.9±0.03	3.1±0.04	3.4±0.04	3.9±0.05	3.3±0.03	3.4+0.03	**p=0.02**	**p=.002**	p=0.4
Response time [s]	1.15±0.01	1.38±0.02	1.32±.0.02	1.3±0.03	1.27±0.03	1.3±0.02	1.28±0.02	p=0.68	p=0.67	p=0.9

Mean, standard error of the mean for the different conditions and results of separate mixed-effects models for repeated-measures ANOVA run for motor awareness, motor performance, reaching time and response time (p-value <0.05 are represented in bold)

5.3 Results

5.3.1 Action Monitoring

An overview of the results is given in Table 5.1. Participants correctly self-attributed 94.1±0.7% (mean ± SEM) of the non-deviated trials. This percentage significantly decreased as the angular deviations increased (main effect of Deviation, p<.001). A main interaction between Target Side and Deviation Side was found for deviated trials (p<0.001). Thus, the percentage of yes-responses was significantly higher when the deviation side converged towards the target side, i.e. when motor compensation was towards the center. The data were consequently divided in two modes: converging and diverging trials. A significant effect of mode was detected (p<0.001, see Figure5.2.A). For converging trials, the point of subject equality is higher (19.8°±5.94, mean±std) than diverging (12.3°±10.2, mean±std). MA is therefore worse for converging trials. No interaction between Mode and Deviation was detected. By looking at MA independently to the modes, the general PSE is at 16.5°± 4.9 (mean±std) which is in line with other agency studies performed with different effectors (see Figure 5.2.B). Motor performance significantly increased as the angular deviations increased (main effect of Deviation, see Table 5.1). However, we did not observe an effect of Mode (p=0.3) nor an interaction Mode x Deviation (p=0.4). Reaching time increased also as the deviations increased (p=0.002), and was significantly lower for converging trials (main effect of Mode, p=0.02). The time to answer the action monitoring question was constant with an overall mean of 1.2±0.12 s.

5.3.2 Visuomotor adaptation task

The Figure 5.3 shows the main metrics of each phase. At baseline, the mean PTE as well as the error variability across groups were not significantly different (p=0.27 and p=0.74, respectively), see Figure 5.3.A-B. For the learning phase, statistics on the learning time constant show a significant effect of EA (p = 0.018, Figure 5.3.C). This effect is driven by a significant reduction of the learning TC for the gain of 1.5 with respect to the control group (gain of 1), p=0.026. In

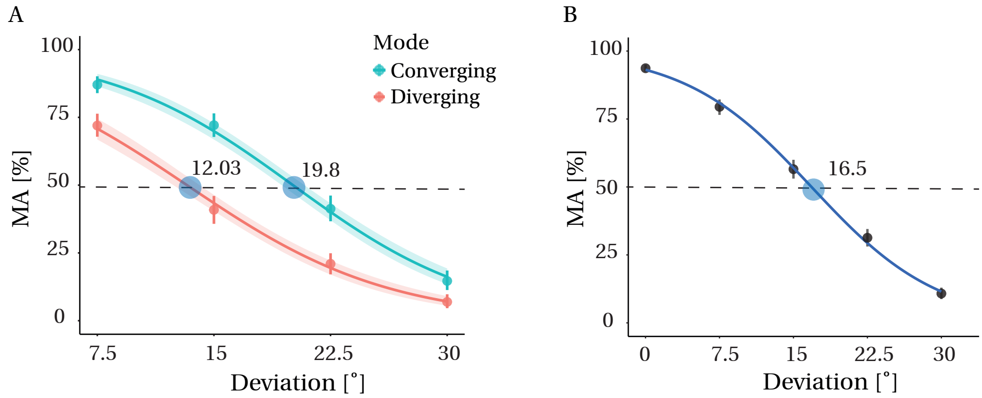

Figure 5.2 – Full-body motor awareness. **(A)** Motor awareness for diverging and converging trials. MA threshold is lower meaning better agency when the deviation side diverges from the target side. **(B)** The general motor awareness point of subjective equality (PSE) is at 16.5°, which is in line with previous studies based on upper limbs and walking agency.

other words, the group with the gain of 1.5 adapted significantly faster to the visual distortion than the control group. Although the learning TC seems to go up with the gain of 1.75, it does not liken to a u-shape, and therefore, the optimal gain cannot be extracted from a quadratic fit. No significant effect of EA was detected for the amount of learning nor for the steady-state error (p=0.1 and p=0.19, respectively, Figure 5.3.D-E). Training with different EA gains does not significantly change the mean error and the error variability of the evaluation phase (p=0.8 and p=0.76, Figure 5.3.F-G). However, a significant effect of baseline was detected all groups combined (p=0.035, see Figure 5.3.H). That is participants who start with a low average PTE during baseline present also a low average PTE in the evaluation phase, and conversely. Finally, washout time constant and steady-state error do not differ between groups (p=0.63 and p=0.49, respectively), see Figure 5.3.I-J. Nonetheless, the steady-state error augments significantly as the EA gain increases (main effect of EA, p=0.016). This effect is mainly driven by the gain of 2, which is significantly higher than the control group (gain of 1), p=0.036, see Figure 5.3.K.

5.3.3 Relation between motor awareness and learning

To study the relationship between motor learning and motor awareness, we first tested if the general PSE was distributed equally across the group, see Figure 5.4.A. Then, the learning time constant was correlated to the general PSE (Figure 5.4.B). All groups combined, the PSE does not seem to affect the learning TC (R = -0.09, p=0.51).
The influence of the PSE on the learning rate was then assessed for each group individually, see Figure 5.4.C. For the control group (gain of 1), participants with low PSE meaning good action monitoring ability adapted slower (high learning TC) than participants with high PSE. This is show by a negative correlation between TC and PSE (R = -0.44, p=0.21). A similar trend

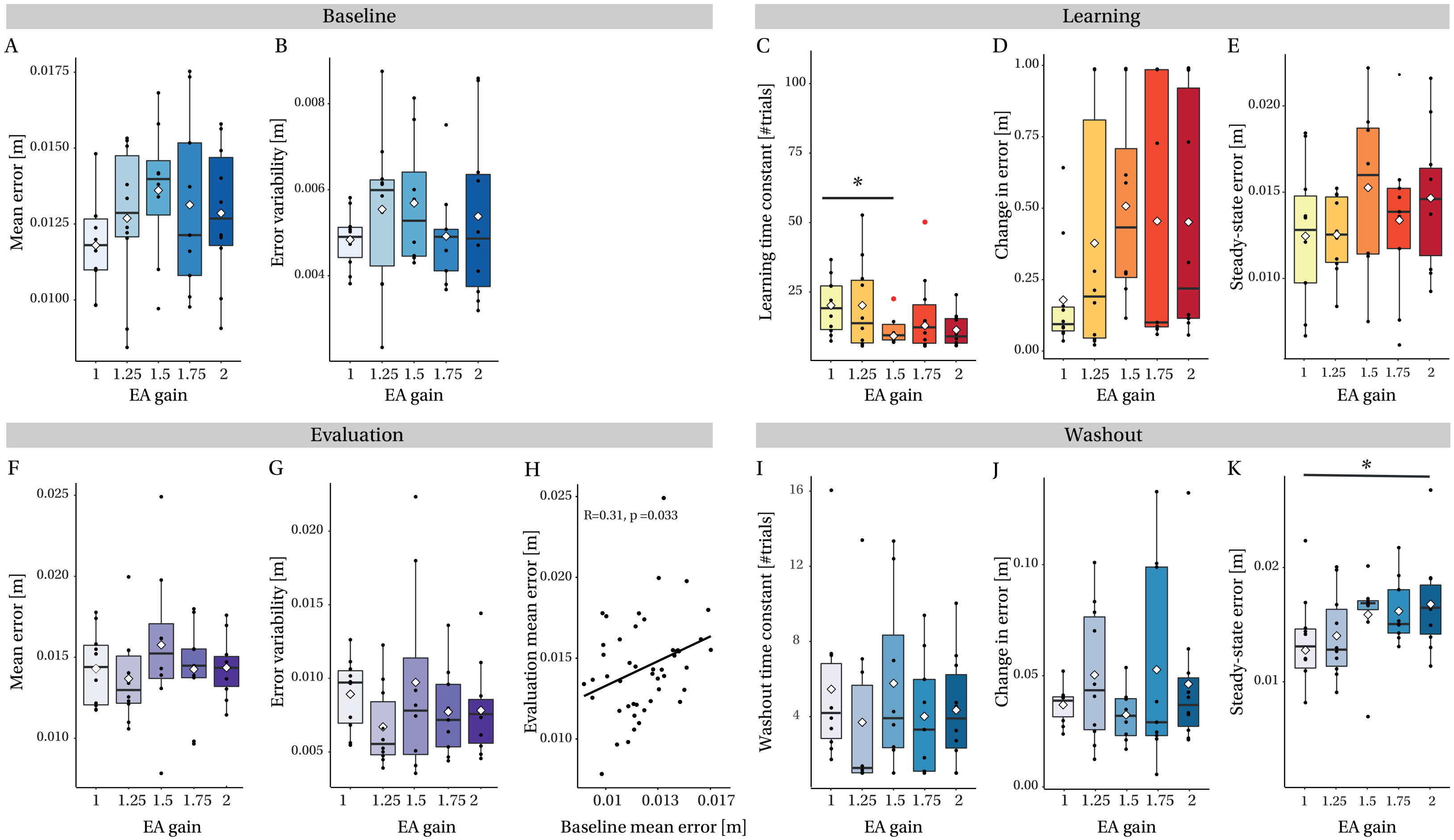

Figure 5.3 – Results overview of visuomotor adaptation during a balance task. **(A)-(B)** show the mean peak trajectory error and the error variability of the baseline phase, respectively. No difference was detected between groups at baseline. **(C), (D), and (E)** represent the main metrics of the learning phase. Training with EA reduced significantly the learning TC. This effect was mainly driven by the gain 1.5. **(F)-(G)** Similarly to baseline, no difference between groups was detected for the evaluation phase. **(H)** shows a moderate association between the mean error at baseline and evaluation. **(I),(J),(K)** represent the main metrics of the washout phases. The steady-state errors increased significantly with the increase of the EA gains.

is observed for the gain of 1.25, while learning TC is not influenced by PSE when training with gains of 1.5 and 1.75 (R=0.056 and R=0.028, respectively). The gain of 2 presents a positive correlation (R=0.57, p=0.09). In other words, participants with good motor awareness learned faster. These results describe a trend, since they are not significant. Indeed, we failed to detect an interaction between EA and PSE (p=0.11).

5.4 Discussion

In this study, we investigated the effect of visual error augmentation on a simple balance task and the association between motor learning and motor awareness. Our results demonstrate that the adaptation rate is significantly improved compared to control group with a gain of 1.5. Moreover, we find evidence that supports a relation between motor awareness threshold and the effect of EA gain on motor learning.

To investigate the effect of visual error augmentation for a standing balance task, participants were asked to perform dynamic weight transfers, while being exposed to a visuomotor mismatch. On top of the visuomotor distortion, the experimental groups had their online trajectory errors amplified by a constant gain. Our results show that visual error augmentation significantly modulated the learning time constant. Indeed, the group with a gain of 1.5 adapted significantly faster than the control group. The level of adaptation, meaning the mean peak trajectory error at the end of the training, did not differ between groups. Since it was proposed that large EA gain could destabilize the adaptation process [18, 16, 229, 22], we expected that the learning time constant would increase with the extreme gain of 2. Our data did not follow this behavior, and consequently, the optimal gain could not be determined with a quadratic function as expected. In contrast to our pilot study, but similarly to studies on simple arm reaching movement [18, 16, 15], EA may introduce destabilization in the learning process of this simple balance task when the gain is superior to one times the error (gain of 2). To understand these findings, we investigated the association between the motor adaptation and motor awareness threshold.

Firstly, the motor awareness threshold of each participant was assessed by implicitly asking them to monitor their movement during a goal-directed balance task. Results show that participants in average self-attributed feedback deviated in space by up to 16.5°, which is a similar mean subjective threshold found in other agency studies on upper limbs [26, 224, 139, 140] and locomotion [28, 149]. These results support the effector-independent hypothesis proposing that the central nervous system uses a general strategy to monitor one's own action during goal directed movements. In addition, the effect of kinematic demands highlighted in chapter 4 was replicated. Participants' motor awareness was impaired (i.e. more self-attribution of deviated trials) when the kinematic demand was reduced. While predictive

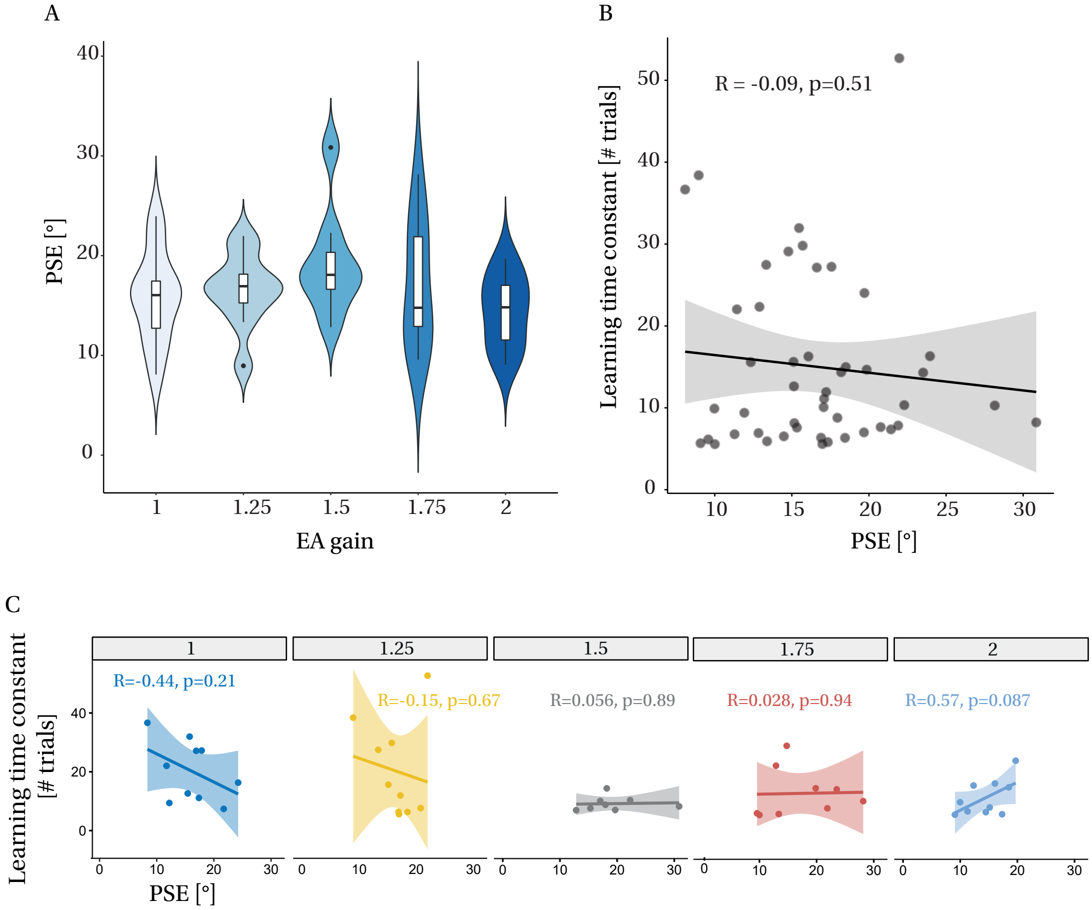

Figure 5.4 – Association between motor learning and awareness. **(A)** shows the similar distribution of the motor awareness threshold (PSE) across groups. **(B)** represents the association between the learning time constant and the PSE of each participant generalized to all groups. **C** shows the TC-PSE association for each group.

processes are often emphasized as central to the sense of agency [144], postdictive inferential processes play also a role [230]. Judgments of agency can be influenced by the cognitive bias due to goal-directedness (failure or success to accomplish the task). Kinematics demands may therefore be a form of goal-bias, where the ease to accomplish the task enhances the feeling of being in control.

Secondly, the relation between the learning time constant and motor awareness threshold was assessed. All groups combined, we failed to detect a relation between the learning TC and the MA threshold. We then examined the TC-PSE association of each groups. In the control group, participants with good motor awareness (low PSE) adapted more slowly than participants monitoring their actions less strictly (high PSE). The learning TC was not modulated by the PSE in participants who trained with the gain favoring the fastest adaptation (gain of 1.5). Interestingly and reversing the observations in the control group, participants with good motor awareness adapted faster than participants less efficient motor awareness when trained with the highest EA gain (2).
Although those tendencies should be confirmed by collecting more data, we may speculate on their meanings. Based on the observer framework presented by Wolpert in 1995 [51] and on the principles that catastrophic errors are needed to learn [94], one could argue that participants of the control group whom adapted faster have a less efficient comparator (as indicated by their high PSE), and therefore did not correct for the deviation causing to experience large errors in the first trials. Large errors may be used to update the internal models and promote implicit learning. However, participants with low PSE, stricter comparator, would automatically compensate for the deviation leading to smaller errors on first exposures. The sensory error being smaller, they may fail to consciously understand the correct movement pattern and needed more time to reach the learning plateau. In the same line of thought, a high EA gain makes participants with strict comparator (low PSE) experience drastic errors during the first exposures which may help to updates their internal model. In other words, high EA promotes explicit learning in participants with good motor awareness. The impaired learning performance in participants with high PSE and high EA gain might be explained by a “cumulation” of errors which prevents the internal model to converge towards the ideal motor commands [231]. This is in line with previous studies suggesting that lower gain may more suitable for complex task, and thus large errors, to avoid instability [22].
The gain of 1.5 promotes fast adaptation independent of other factors, and therefore could be close to the optimal gain for this task. This value is close to those proposed for learning simple task in upper limbs [18, 16, 17].

These rather speculative interpretation have to be confirmed by further data analysis and data collection. Further analysis should include the effect of motor performance on learning rate, as well as error variability during the learning phase. Moreover, similar analysis should be run on the washout phase to observe how motor awareness and performance influence the short-term retention. Although these results advocate that visual error augmentation could be beneficial for balance training, they should be taken cautiously, since long-term effect has not been studied.

This study showed that the learning time of a simple balance task may be reduced by applying the visual error augmentation paradigm. This could have implications in the way visual feedback is provided for balance rehabilitation. Moreover, these findings suggest the optimal EA gain may be related to the motor awareness threshold. Therefore, one could think to first assess the motor awareness threshold of the participants and then predict the optimal EA gain. This would be a first step towards personalized medicine for rehabilitation.

6 Standing postural control in the exoskeleton TWIICE One

6.1 Bio-inspired approach - from INSPIIRE to TWIICE One

To develop a bioinspired postural controller for the exoskeleton TWIICE One for complete SCI users, five healthy young adults were required to stand in a passive lower limb exoskeleton, called INSPIIRE, reproducing the same kinematic constrains. Both exoskeletons have only two degrees of freedom per leg (hip and knee) and round-shaped soles to compensate the lack of mobility of the ankle. We identified the changes in postural control and translated them into two position controllers on TWIICE One. The controllers have been preliminary tested with a SCI pilot, demonstrating ability to free the hands of exoskeleton users while standing.

The next sections include a description of INSPIIRE and TWIICE exoskeletons, the methods that allowed us to identify the adaptation in postural control, and finally a characterization of the developed postural controllers.

6.1.1 INSPIIRE - a locked-ankle passive exoskeleton

Humans exhibit impressive coordination and timing to keep balance in challenging situations. Inspiration should therefore be taken from human motor control and their capacity to adapt to their environment in order to the behavior, the reliability and the robustness of exoskeletons controllers. To improve the control of the exoskeleton TWIICE One (see description in section 6.1.2), a passive exoskeleton called INSPIIRE was developed at the Robotic Systems Laboratory (LSRO), see figure 6.1. This device was composed to study gait and balance of healthy individuals under constrained degree of freedoms. The understanding of sensorimotor adaptation has implications in many fields such as rehabilitation engineering, sport science and neuroscience.

Figure 6.1 – INSPIIRE outlook. Passive lower limb exoskeleton with locked-ankle.

INSPIIRE is therefore an ideal framework to observe sensorimotor adaptation and transfer them to legged robots.

INSPIIRE is composed of four parts: the feet, the shank and the thigh segments and the pelvic structure. The shank and thigh segment are adaptable in length to fit the user anatomy, while the pelvic structure is adjustable in width. Passive revolute joints allow the knees and hips to move freely in the sagittal plane, while the rigid pelvic structure prevents hip abduction and adduction. Absolute encoders (AMT20, CUI Inc., USA) measure the hip and knee joints angle with a resolution of 0.088°. The backpack unit is composed of a conventional mountaineering backpack with higher shoulder straps to provide a firm trunk maintenance during experiments. It embeds all the electronic. INSPIIRE has an adjustable locked ankles and round foot sole facilitating dynamic walking. The feet are the same modules as in the TWIICE exoskeleton [44], see Figure 6.2. They have four flat load cells measuring the vertical interaction forces of the foot with the ground. A custom PCB amplifies, samples, and sends the load cells signals to the on-board computer.

The next section presents how INSPIIRE was used to capture the postural strategies implemented to cope with the constrains of having locked ankles and round foot soles while standing.

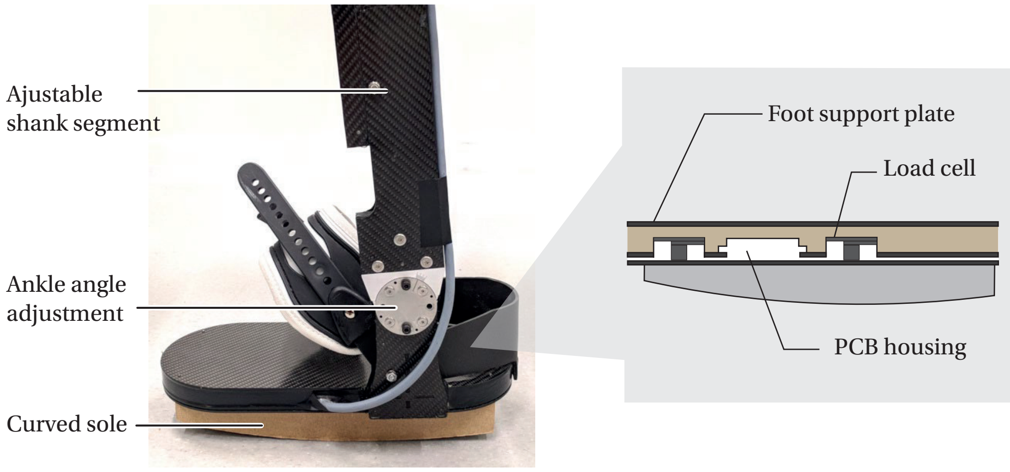

Figure 6.2 – INSPIIRE - Foot module details and load cells assembly (sagittal cut view, only two of the four load cells are visible).

6.1.2 TWIICE One - a full-mobilization lower limb exoskeleton

TWIICE One is the upgraded version of TWIICE introduced in [44], a lower-limb exoskeleton for complete SCI users (Figure 6.3). We use the terms user, pilot or test pilot to describe the person inside the exoskeleton, because these devices have not been tested for rehabilitation purposes. They were developed to participate to the Cybathlon and Cybathlon series [232], which aim to promote the development of assistive devices for daily activities.

TWIICE One has the same structure than INSPIIRE, except that each segment length is tailored to the user morphology. The ankles joints are locked at 90° and are not adjustable. As for INSPIIRE, the lack of mobility of the ankle is overcome thanks to round foot soles which allow to use the passive dynamics properties of a rocker-based inverted pendulum [233].
The exoskeleton firmly maintains the leg of the user with foot, shank and thigh cuffs, and moves the user's hip and knee joints in a position evolving in time (mobilization). These joints are mobilized solely in flexion and extension, while all other degrees of freedom are locked. The hip and knee joints are composed of compact and powerful actuators [234] maintaining the desired position stiffly in order to stand and walk. Successive joint positions are defined by a trajectory which corresponds the action the user intends to perform. The trajectory is played back at a predefined speed immediately after the user expressed their intention to move again.

The user needs crutches to manage the weight transfer from one leg to the other while walking and to keep sagittal balance when standing. The right crutch forearm holds a remote composed of three buttons and a trigger. The pilot chooses between actions types by selecting operating modes, each corresponding to a different activity such as slow/fast walking,

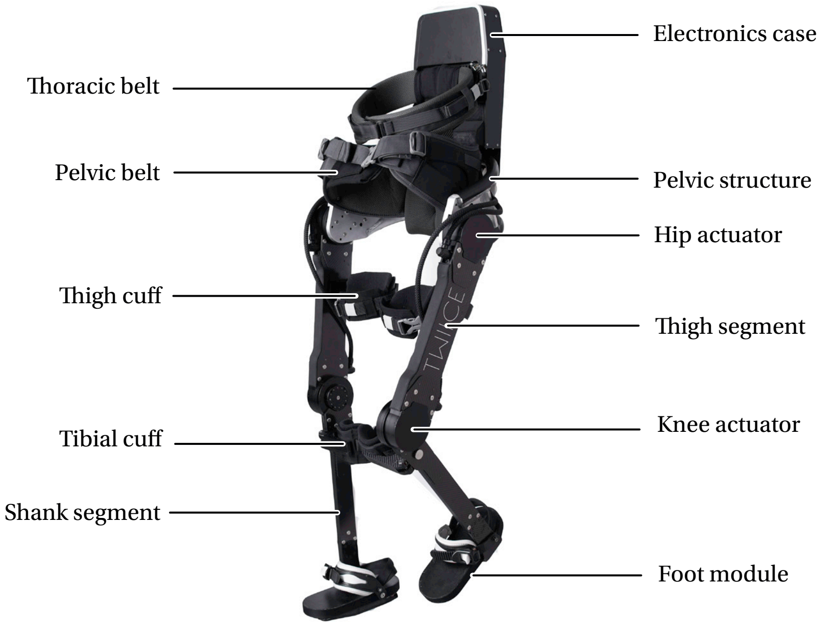

Figure 6.3 – TWIICE One - A lower-limb exoskeleton for individuals with complete paraplegia

ascending/descending stairs or walk up a slope (Figure 6.4.A). Visual feedback is given on a smartwatch (Figure 6.4.b). Each action (i.e. each step) are triggered using the point finger lever situated below the right crutch forearm. For more details on the control architecture, the reader is referred to [44].

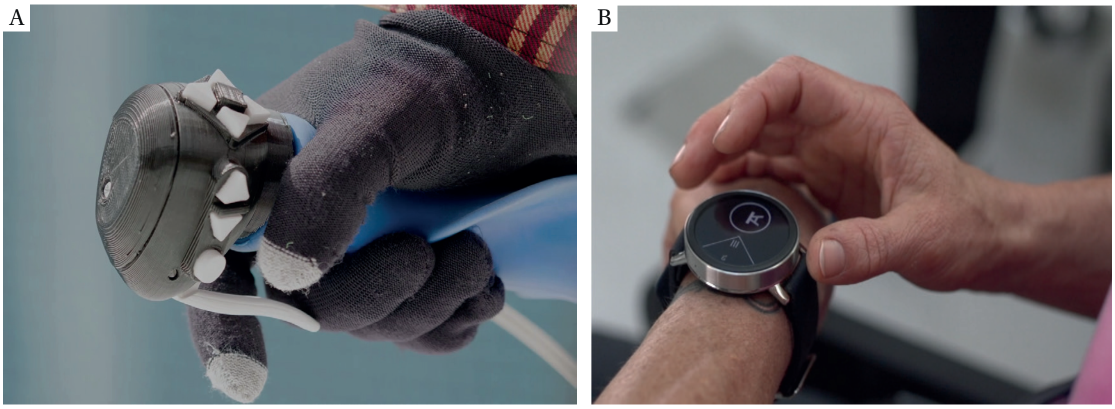

Figure 6.4 – Accessories for the exoskeleton control **(A)** Control remote allowing to navigate through the operating modes (up-down arrows) and to trigger each action with the point finger lever. **(B)** Smartwatch displaying the operating modes such as standing, sitting, walking at different speed or climbing stairs, among others.

6.2 Balance control strategies

Balance Control Strategies during Standing in a Locked-Ankle Passive Exoskeleton

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Abstract
This study investigates sensorimotor adaptation strategies of sagittal postural control in healthy subjects under kinematic constraints. A passive exoskeleton named CAPTUR, with locked ankle joints and legs motion restrained to the sagittal plane, is used to restrict and measure participant's movements. The aim is to assess the role of the orientation of the shank and the trunk segments in maintaining the body center of mass above its support base, while the ankle strategy is inhibited. Five young healthy participants were required to keep standing, while their balance was challenged by five experimental conditions. Participants mainly regulated quiet standing balance by flexing/extending the knees, in order to affect the shank and feet angles, and move the contact patch along the sagittal axis. In this case, the orientation of the trunk segment changes synchronously with the shank angle to keep an upright posture. Responses to more dramatic excursions of the center of pressure are ensured by changing the trunk tilt angle in opposition of phase with the shank angle. These observations could be used to implement a bioinspired balance controller for such constrained lower-limb exoskeletons.

6.2.1 Introduction

Humans are capable of highly dexterous coordination and balance even in challenging situations. Understanding how this motor coordination takes place has implications in many fields, including rehabilitation engineering, robotics, sports science and neurosciences. The underlying models over human balance and loCoMotion, if understood, could be implemented to legged robots, prosthetic and orthotic devices, assimilated to models of motor learning or to mechanisms of function restoration. More specifically, we are interested in how postural control in humans could be used as a bioinspired approach to control TWIICE, a lower-limb exoskeleton for complete SCI) patients [44]. The lightweight design of TWIICE has been achieved by selecting only key features of locomotion such as knee and hip flexion/extension. Thus, the ankle dorsi-flexion has been locked to favor lean manufacturing. The lack of mobility of this joint has been compensated by a curved foot sole allowing the foot to roll on the ground while walking [233]. The impact of ankle locking on gait kinematics has been observed in [235], however, its repercussions on standing balance have not been studied yet. Understanding how healthy individuals are able to keep their balance in similar kinematics conditions than those of the TWIICE exoskeleton could provide bioinspired control primitives, enabling SCI exoskeleton users to stand without the help of crutches.

To our knowledge, no other studies investigated sagittal postural control with locked degrees of freedom at the ankle joints in healthy individuals. Therefore, we aim to investigate how the coordination of the shank and the trunk segments is controlled during human standing under these kinematics constraints. For this purpose, a purely passive lower-limb exoskeleton, called INSPIIRE, was designed to mechanically lock the ankles at a fixed angle and prevent body movements in the frontal plane. It has the same mobile joints and the same instrumentation as the full-mobilization lower-limb exoskeleton TWIICE and fully curved foot soles.

A common approach for investigating human postural control is either to analyze spontaneous sway during quiet standing or to apply perturbations to the overall system. The type of perturbations such as visual [236][237], vestibular [238][239] or mechanical [240], [241] allows to assess the contributions of different physiological mechanisms to postural control. Thus, five experimental conditions such as eyes open and closed, visual feedback, cognitive load and mechanical perturbations were tested in this study to observe whether different postural strategies were implemented by the CNS.

In his work, Nashner [70] proposed to look at the body position in space, rather to the muscular synergies, to define which postural strategies are used. Three distinct control plans were identified: the vertical, the ankle and the hip strategies. The ankle strategy is often associated with quiet standing or low amplitudes of perturbations, while the hip strategy is required for larger disturbances. Since most of the postural models are represented as a two-links pendulum, the vertical strategy, involving knee flexion/extension, has often been neglected. However, with locked ankle joints, this approach might substitute the ankle strategy during quiet standing.

It was proposed to look at the inter-segmental coordination to identify which strategies were adopted by the CNS [242]. The sign of the covariance between the shank and the trunk segments pitch angle allows distinguishing the hip and the ankle strategies. In the absence of ankle movements, inter-segmental analysis can be used to determine whether the vertical or the hip strategy is favored to keep balance inside the passive exoskeleton. Therefore, the contributions of shank and the trunk segments to the sagittal displacement of center of mass (CoM) were analyzed, as well as the sign of their correlation coefficients over time-segmented trials [243]. In addition, the range of motions of the hip and knee joints and of the CoP were computed to observe if the different experimental conditions trigger distinct amplitudes of corrective movements and CoP excursions.

It is hypothesized that the role of the ankle joints will be substituted by a motion at the knee joints for short CoM excursions, i.e. quiet standing. Nevertheless, it is expected that the hip strategy, meaning opposite motion between the trunk and the shank segments, will intervene for bigger perturbations.

6.2.2 Methods

Participants: young healthy adults

Five healthy, young able-bodied participants took part to the study. They were recruited from a population of various ages (28 ± 3.8 years, only males) and similar physical shape (mass: 70.2 ± 3.63 kg; height: 179 ± 4.8 cm). The EPFL ethical committee approved this study for research within the LSRO at EPFL. All participants gave written informed consent in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki prior to study's start. Only participants with an intact musculoskeletal system and no history of orthopedic, neural or psychiatric disorders were enrolled in the study. Volunteers with foot length smaller than 240 mm (i.e. 8 US or 39 European shoes size) were excluded from the study due to the non-adjustable length of the foot sole.

INSPIIRE-a passive lower-limb exoskeleton

A non-actuated lightweight lower limb exoskeleton, called INSPIIRE, has been developed at EPFL by the Robotics Systems Laboratory for gait and balance analysis under kinematic constraints (Figure 6.1). It is dedicated to capture the change in locomotion patterns and balance control of healthy subjects. Its mobility, especially at the hip and ankle joints, reproduces the degrees of freedom of the most common full mobilization exoskeletons for paraplegic patients. The hip and the knee joints are limited to flexion and extension by revolute joints. All other movements are restricted. Its structure is made of stiff composite materials and aluminum, to prevent the wearer's movements in the frontal plane. Moreover, to enable the analysis to focus completely on the motion in the sagittal plane, the inter-foot distance was fixed to 100mm, since it is known that narrow stance widths (<80mm) augment lateral sway [195]. Some structural parts accommodate mechanical adjustment (shank length, thigh length, hip width and abduction angle), to adapt to the participants' morphology. The ankle is locked at a fixed dorsiflexion angle of 5° during the whole experiment, preventing the use of ankle strategies for balance control in stance. It is equipped of different types of sensors to measure the joints positions and the ground reaction forces while walking or standing. The absolute encoders (AMT20, CUI Inc., USA) measure the hip and knee joints angle, with a resolution of 0.088°. The embedded computer, located in the lower back structure, features an inertial measurement unit (IMU) (MPU-6050, InvenSense, USA) to measure the pitch and roll angles of the trunk. The sensory data is collected by the embedded computer of the CAPTUR (BeagleBone Black, Texas Instruments, USA). All the sensors are sampled synchronously at a frequency of 100 Hz and logged continuously to an SD card. The foot design is similar to the one of the TWIICE exoskeleton [44]. The four flat load cells are the only mechanical links between the two rigid layers, to measure accurately the contact forces. A custom printed circuit board (PCB) embedded in the sole amplifies, samples and sends the load cells signals to the on-board computer. The sole has a special curvature allowing the foot to roll on the ground while walking to compensate for the lack of mobility in the ankle. However, this sole curvature is also responsible for the dynamic instability while standing. Made of wood with an apex position towards the back of the foot, the contact surface with the floor is small and the friction is high.

Experimental procedure

The participants' segments lengths were measured and the exoskeleton segments adjusted accordingly. Once settled in the exoskeleton, they were asked to stand in front of a wall with a 24-inch computer monitor. They were first given a 1 min training session to get accustomed with the system. Then five conditions were assessed during the same session. Each condition had a maximum duration of 90 seconds and a 1 minute rest between them. For each condition, the participants were instructed to maintain a static base of support and to keep their aroot mean square (RMS) crossed against their chest. The order of the five conditions was randomized between subjects. They included:

- eyes opened (EO)
- visual feedback (VF) - A cursor showing the position of the participants' CoP was displayed on the monitor and they were instructed to keep it in the center of a fixed target
- cognitive load (CL)- The subjects had to perform serial-7 subtractions
- eyes closed (EC)
- random perturbation (RP) - Participants were pushed and pulled in the sagittal plane from the posterior part of the exoskeleton lower back with random perturbation forces ($P < 20N$).

Data processing

Off-line data analysis was conducted in MATLAB. The angular position of the trunk segment (θ_{Trunk}) was measured relative to the earth vertical (6.5). The angular motion of the shank (θ_{Shank}) was obtained by subtracting the angular position of the hip joint and adding the angular position of the trunk segment. The CoP was obtained by applying the barycentric formula over the eight load cells. The CoM was estimated to be at the height of the belly button. Its position in the sagittal plane was calculated from the direct kinematic of a human model consisting of three segments: shank, thigh and trunk. The sagittal displacement due to

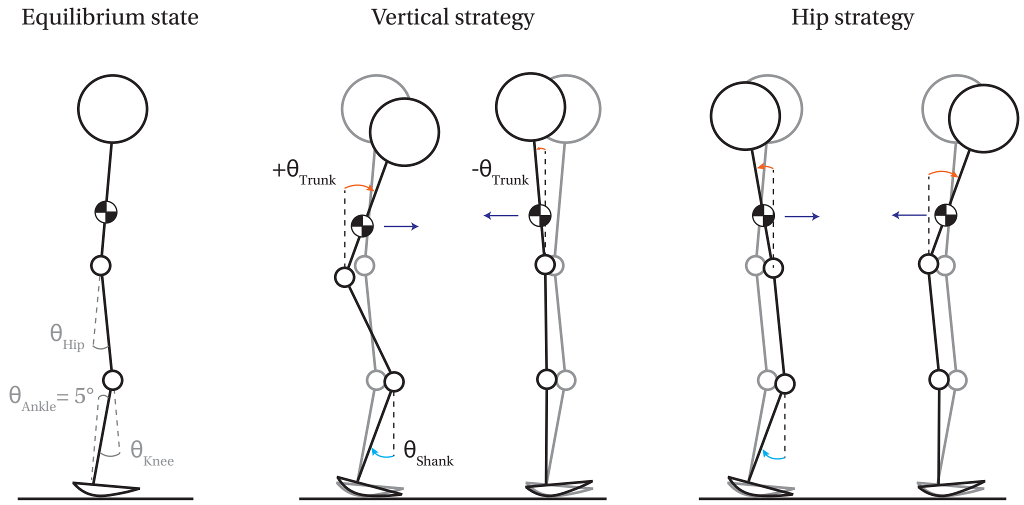

Figure 6.5 – Schematic diagram of the postural control strategies with the angles definition. The ankle joints are locked at 5°

the rolling foot soles was neglected since it was smaller than 20 mm. The first 30s of each trial were removed to discard the potential adaptation phases. The time series of the data were detrended linearly and filtered using a second order band-pass, zero phase lag Butterworth filter with a cut-off frequency of 0.5 Hz and 1Hz [243]. Note that increasing the low-pass filter value does not change the results. After having detrended the data, but prior to filtering, the RMS value was calculated for the knee (θ_{Knee}), hip (θ_{Hip}) and trunk (θ_{Trunk}) angular displacements, as well as for the CoP displacements, to evaluate their magnitude of motion and the amplitude of perturbation, respectively, for each condition. Their velocities and accelerations were obtained by differentiating the filtered angular positions.

Data analysis

In previous studies, postural coordination was assessed by looking at the inter-segmental covariance [242], [244] or correlation [243] of the shank and the trunk angular displacements. A hip or ankle strategy of postural control was identified with the sign of the correlation between these two segmental degrees of freedom. A positive correlation describes the ankle strategy, while a negative one represents the hip strategy. In our analysis, we propose an intermediary step in order to evaluate the contribution of the angular displacements of the shank and the trunk to the CoM sagittal motion. Instead of considering the direct correlation between the two segments angles, the Pearson correlations CoM-θ_{Shank} and CoM-θ_{Trunk} were computed for the entire and time-segmented trials. By computing the square of the correlation coefficient (R), the coefficient of determination (R^2) was obtained. It estimates

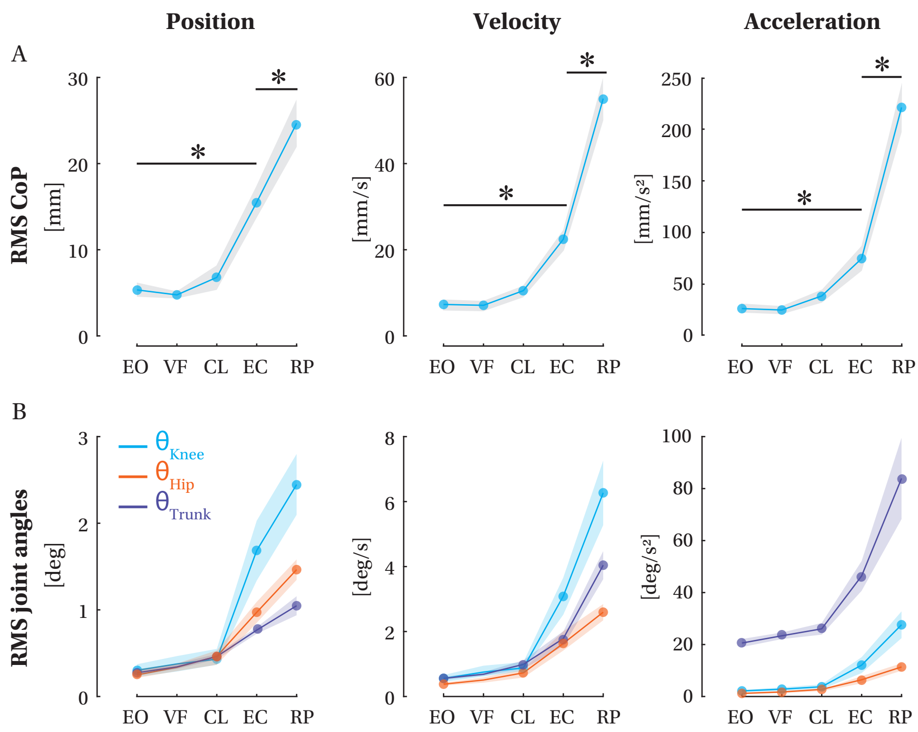

Figure 6.6 – Mean values of the RMS of the CoP displacement (a) and of the joints angular displacements, velocities and accelerations (b) averaged across all subjects (n=5).

the proportion of the CoM variance that is predictable by the motion of the shank and/or the trunk segments, and thus which segment orientation has a predominant role in postural maintenance. The sign of the correlations still allows to determine the postural strategy used to stabilize the CoM. If the two correlations are positive, then the shank and trunk segments rotate in the same direction. In this case, the vertical strategy, described first by Nashner [70], is used by the central nervous system to keep balance. Conversely, if the correlation coefficient between the CoM displacement and the rotation of the trunk segment is negative, the shank and the trunk segments are in opposition of phase and this state is assumed to be the hip strategy in our context (Figure 6.5).

A time-segmented analysis was performed to assess the variation of postural control strategies over time [243][245]. A 12s moving window marched along the 60s trials by step of 10 ms. At each step, the Pearson correlations were calculated with 1200 samples. To reach the statistical significant level of 0.05, the R value of the Pearson correlation should be out of the ± 0.7 interval. The average R^2 and the mean percentage of time spent using different control strategies for each condition were calculated across participants. Paired two-tailed Students t-tests evaluated significant differences between the conditions and variables. An alpha level of 0.05 was used on all statistics.

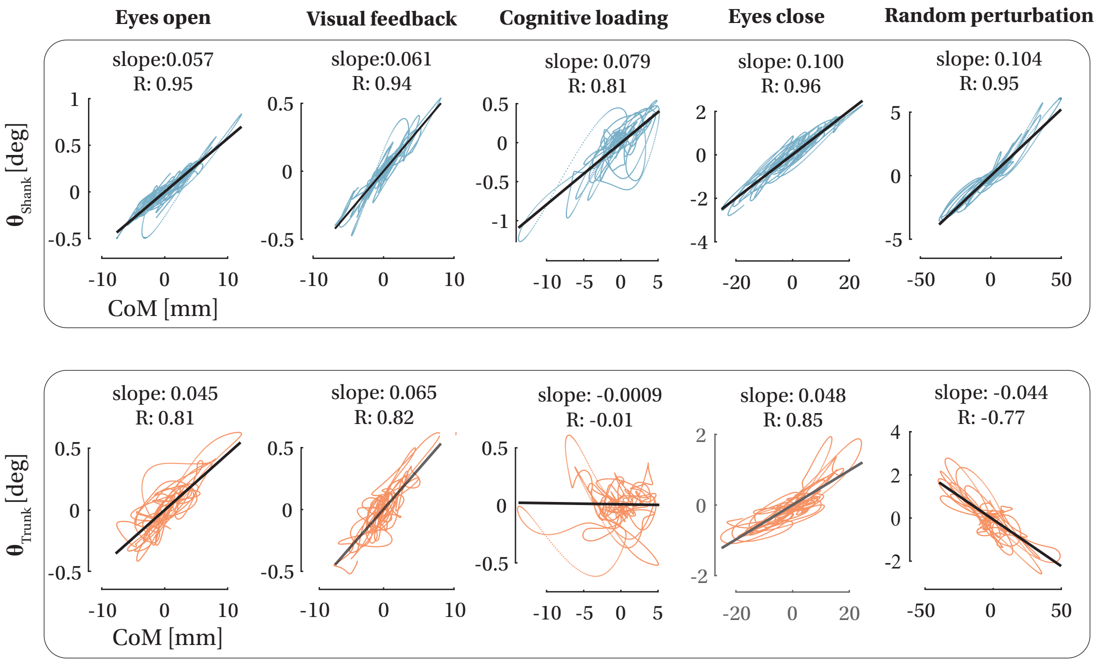

Figure 6.7 – Correlations of a representative participant for the entire trial of each condition

6.2.3 Results

Sway amplitude and magnitude of joints motion

The different experimental conditions was to obtain a range of perturbations. By looking at the RMS values of the CoP displacement (Figure 6.6.A), the eyes closed and the random perturbation conditions significantly increased the CoP excursion in comparison to the three other conditions ($p \leqslant 0.015$). The visual feedback did not significantly reduce the sway of the CoP with respect to the EO baseline condition (pEO-VF = 0.99). The cognitive load introduced more sway, but did not change significantly the amplitude of the CoP displacement with respect to the EO baseline (pEO-CL = 0.41). Similar results were obtained for the RMS values of the velocities and accelerations. Figure 6.6.B shows the mean RMS values of the joints angular displacements, velocities and accelerations averaged across all subjects. The range of motion of the knee is significantly larger than the one of the hip and the trunk for the EC and RP conditions ($p \leqslant 0.05$). Same observations can be done for the velocities, while the trunk accelerations are significantly larger than those of the hip and the knee joints for every conditions ($p \leqslant 0.008$).

Shank and trunk coordination

For the analysis over the entire trials, Figure 6.7 shows a representative sample of the behavior followed by the majority of the participants for the fives conditions. For the CoM-θ_{Shank} relationship, the sign of the correlation using the entire trial is positive for every subjects and each condition. The average correlation coefficients across subjects for each condition is higher than 0.93, except for the CL condition that has a mean R of 0.85 ± 0.05. Concerning the relationship between the trunk angular displacement and the CoM position, the sign of the correlation over the entire trials is positive for the EO (R = 0.53 ± 0.09), VF (R = 0.78 ± 0.02) and EC (R = 0.79 ± 0.06) conditions and negative in the RP condition (R = 0.53 ± 0.13) for every participant. The CL condition presents lower correlation coefficients and different signs in between subjects (R = 0.52 ± 0.15), due to mixed postural strategies as shown by the time-segmented analysis.

For the time-segmented analysis, the consistency of the postural behavior can be seen in Figure 6.8.A, where the correlation coefficients (R) are plotted as function of time for three conditions of a representative participant. When both correlation signs are positive, the vertical strategy is preferred, while opposite signs describe the hip strategies. The time-segmented correlations show very clearly that mixed strategies are used during the CL condition, while the vertical strategy is selected with EO and the hip strategy is used during RP. The time series of the CoM position, shank and trunk angular positions are plotted for selected time windows (Figure 6.8.B) to show the synchronization of the trunk and the shank segments during vertical strategy and the opposition of phase of these segments in the hip strategy. The amplitude of the CoM displacement differs in between conditions and corresponds to different postural strategies. The mean percentage of time spent in each strategy for each condition averaged across subjects are shown in Figure 6.9.B.

Main contributors of postural stabilization

The contribution of the shank and trunk angles in the variance of the CoM position are reported in Figure 6.9.A. The orientation of the shank segment can clearly be identified as the main contributor of the variance of the CoM motion. For every conditions, the shank displacement explains at least 87% of the CoM variance ($R^2 \geqslant 0.87$), except for the cognitive load where the displacement of the CoM is well shared between the two segments.

6.2.4 Discussion

This study investigated balance control strategies under sever kinematic constraints (locked ankle, motions limited to sagittal plane). Such constraints are typical for certain types of

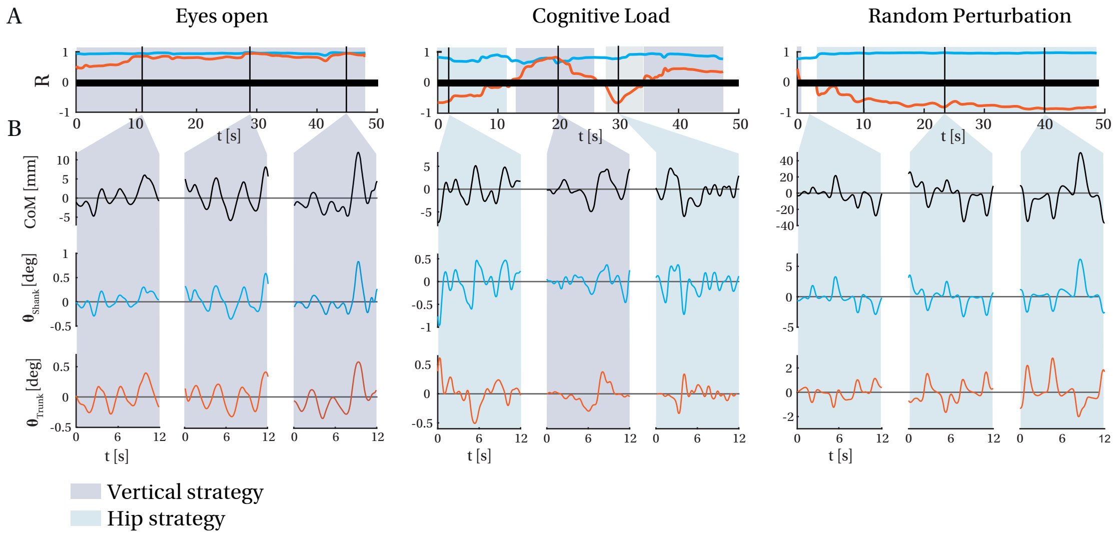

Figure 6.8 – Time-segmented correlation analysis of three representative conditions for a single participant.

exoskeletons. A first observation shows that a static base of support can be maintained without taking a step under perturbations up to 20N. Since the level of perturbation can be estimated with respect to the CoP excursion [246] and especially the CoP velocity [247], the analysis of the CoP motion shows that the participants are exposed to three levels of perturbation. Indeed, the EO, VF, CL conditions produce similar motion amplitudes of the CoP, while the EC and the RP present larger excursions. It was expected that the visual feedback would help stabilization and therefore reduce the sway motion of the CoP in comparison with EO. Although a trend in this direction is observed, this is not significant. The participants have very little sway motion with the eyes open and therefore not much to correct. Thus, the visual feedback does not contribute to lower this stability threshold. The CL condition does not directly change the sway, which is consistent with other studies with young participants [237][248][249] but has an impact on the motion synchronization of the shank and trunk segments. Indeed, the cognitive load introduced mixed postural strategies although the perturbation level is similar.

Participants spend most of their time in what we call the vertical strategy. They mainly regulate the position of their CoM during quiet standing balance by flexing/extending the knees, in order to affect the shank and feet angles, and move the contact patch along the sagittal axis. Minimal trunk movements in phase with the shank displacement are initiated to keep an upright posture. This result suggests that postural control in the TWIICE lower-limb exoskeleton for SCI users could be achieved by regulating the knee flexion/extension as function of the sagittal position of the CoM. However, when a dominant input perturbs the system (RP condition), young healthy subjects use another consistent strategy to counteract fast or large amplitude disturbances, as it is also the case when the ankle response is allowed

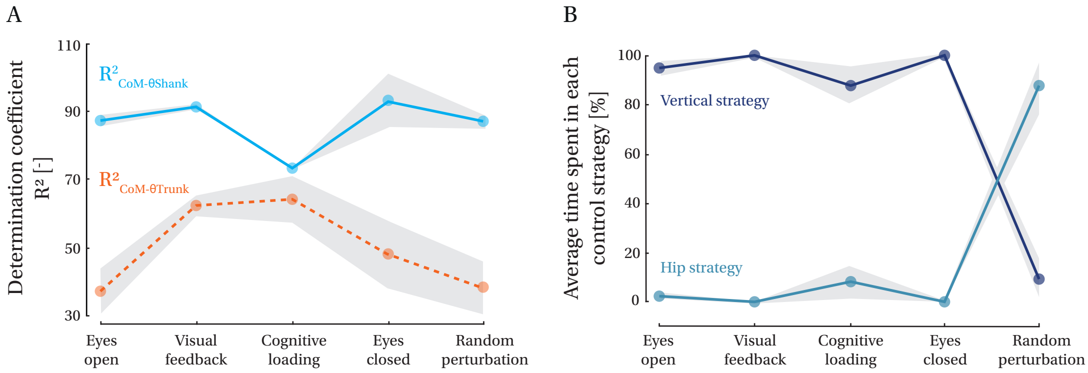

Figure 6.9 – **(A)** Average coefficient of determination across participants for the time-segmented analysis.**(B)**Average percentage of time spent in each postural control strategy.

[240][70]. Sagittal perturbations at the pelvis level trigger a counter reaction of the shank and the trunk segments to keep the CoM as static as possible.

For both strategies, the main contributor to the variance of the CoM position is the orientation of the shank with respect to the earth vertical. The flexion/extension of the knee joints is used either to re-position the CoM or to damp the perturbations. The main limitation of this study is that the arm movements, especially their accelerations, are neglected, since volunteers are asked to keep their arm crossed. Arm movements could contribute to restore postural stability and could lead to different limbs coordination.

The above results suggest that postural control in the TWIICE lower limb exoskeleton for SCI users could be achieved by regulating the knee flexion/extension as function of the sagittal position of the CoM. Control strategies for postural stability in lower-limb exoskeletons have been investigated, but with different goals and methods. Indeed, previous studies aimed to assist standing posture in in complete SCI patients with supportive torque at the ankle and knee joints [196][197]. In a different approach, our study provides bioinspired control primitives for a rigidly position-controlled exoskeleton, destined to complete SCI patients.

Moreover, observing how healthy subjects respond to a constrained environment and different balance conditions can contribute to the understanding of sensorimotor adaptation mechanisms taking place in the central nervous system for postural control. The time-segmented analysis reveals that the CNS maintains balance by using mainly the vertical control strategy when balance was unperturbed. These findings could be assimilated to models of motor learning or to mechanisms of function restoration. Future work will focus on transposing these results in a bioinspired controller for a lower-limb exoskeleton with the same degrees of freedom than the INSPIIRE passive exoskeleton. The goal is to enable standing posture without the need of an additional support point.

6.3 Postural controllers

Bioinspired Postural Controllers for a Low-Actuator Count Exoskeleton Targeting Complete SCI Users

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Abstract
Several lower-limb exoskeletons enable to overcome obstacles that would impair daily activities of wheelchair's users, such as going up stairs. Still, as most of the current commercialized exoskeletons require the use of crutches, they prevent the user from interacting efficiently with the environment. In a previous study, a bio-inspired controller was developed to allow dynamic standing balance for such exoskeletons. It was however only tested on the device without any user. This work describes and evaluates a new controller that extends this previous one with an online model compensation, and the contribution of the hip joint against strong pulling perturbations. Both controllers are tested with the exoskeleton TWIICE One, worn by a complete spinal cord injury pilot. Their performances are compared by the mean of three tasks: standing quietly, resisting external perturbations, and lifting barbells of increasing weight. The new controller exhibits a similar performance for quiet standing, longer recovery time for dynamic perturbations but better ability to sustain prolonged perturbations, and a significantly higher weightlifting capability.

6.3.1 Introduction

Lower-limb exoskeletons have gained much interest in the last decade. This growing interest is mainly driven by the aim of enhancing human performance and improving neuromotor rehabilitation. Therefore, developing novel features to improve user safety, mobility and autonomy is a constant research challenge. In the field of wearable robotic systems for complete spinal cord injured (SCI) patients, walking is the main function targeted by the majority of lower-limb exoskeletons. Balance management while walking and standing is generally performed by the user with the help of crutches, and thus impairing the use of their hands for other activities. Very few full-mobilization exoskeletons are able to self-stabilize [192, 193, 176], and thus, allow to free the user's hands. This comes at the cost of a low walking speed and an important overall weight. In addition, none of them are able to climb stairs. Standing and walking balance are essential functions to promote exoskeleton usage during daily activities, that however should not come to the detriment of other features.

In daily-life activities, manual tasks and environmental interactions happen mainly while standing (e.g. shaking hands, grabbing an object, drinking) rather than walking. Therefore, a valuable trade-off would be to enable the hands usage during standing for exoskeletons actuated only in the sagittal plane. While the fore-aft balance could be actively regulated, the lateral stability can be maintained passively if the space between the feet is large enough, thanks to the wider base of support (BoS).

Humans are constantly adjusting their posture to act against gravity and are capable to resist to moderate internally-generated or environmental perturbations using body coordination only (i.e. without stepping). To counteract these perturbations, proactive and reactive forms of postural movements are generated by the sensorimotor system to keep the center of mass (CoM) within the BoS [250]. Thus, coping with unexpected and self-generated perturbations requires a robust postural controller. As of today, there is no full-mobilization exoskeleton, position-controlled, actuated only in the sagittal plane capable of maintaining a standing posture with users that do not have any control of their lower limbs. Therefore, our goal is to develop a postural controller for TWIICE One, a lower-limb exoskeleton for complete SCI users. Several research groups work on partial assistance during stance with the goal to improve the balance of people with incomplete SCI. Most of these control strategies are using torque control [197, 196, 198, 199]. These studies mimic the most common postural strategies highlighted by Winter et al. [200]: the ankle, the hip and their combined strategies. However, the limited number of degrees of freedom of TWIICE One does not allow to directly adopt these control strategies. For that reason, in a previous study, we observed how young healthy participants adapted their postural control strategies when wearing a passive exoskeleton, called INSPIIRE, with the same kinematic constrains as TWIICE One with fully curved foot soles [251], see Figure 6.10. It has been found that healthy adults mainly manage postural balance by flexing and extending their knees in order to move the contact point along the anteroposterior axis while standing quietly inside a passive locked-ankle exoskeleton. This strategy is referred to a vertical strategy, meaning that the trunk and the shank orientations move in phase. In case of more consequent perturbations, the hip strategy was used to maintain balance. During the hip strategy, the shank rotation is not sufficient to keep the CoM of gravity over the base of support, then the trunk rotates in the opposite direction to compensate and reposition the CoM. Drawing inspiration from this human sensorimotor adaptation, a novel postural position controller has been implemented and tested on TWIICE with no user [252]. This controller regulated the balance with a proportional-derivative (PD) controller, setting the angle of the knee, and fed with the CoM position. In this article, an extended version of this knee controller will be described and evaluated with a complete SCI test-pilot. It is designed to resist to stronger long-term perturbations. These controllers are potentially useful for the current generation of full-mobilization exoskeletons, because they do not need torque control in the joints, or load cells in the feet. The hardware can then be kept minimal, so the device can be simpler, less expensive and more robust. The goal of this case study is to characterize the performance of the two postural controllers enabling a complete SCI user to stand without crutches.

6.3.2 Postural control framework

TWIICE One

Two postural controllers have been implemented on the lower limb exoskeleton TWIICE One 2018 (Figure 6.10.D). This exoskeleton is similar to the version of 2016 introduced in [44]. The

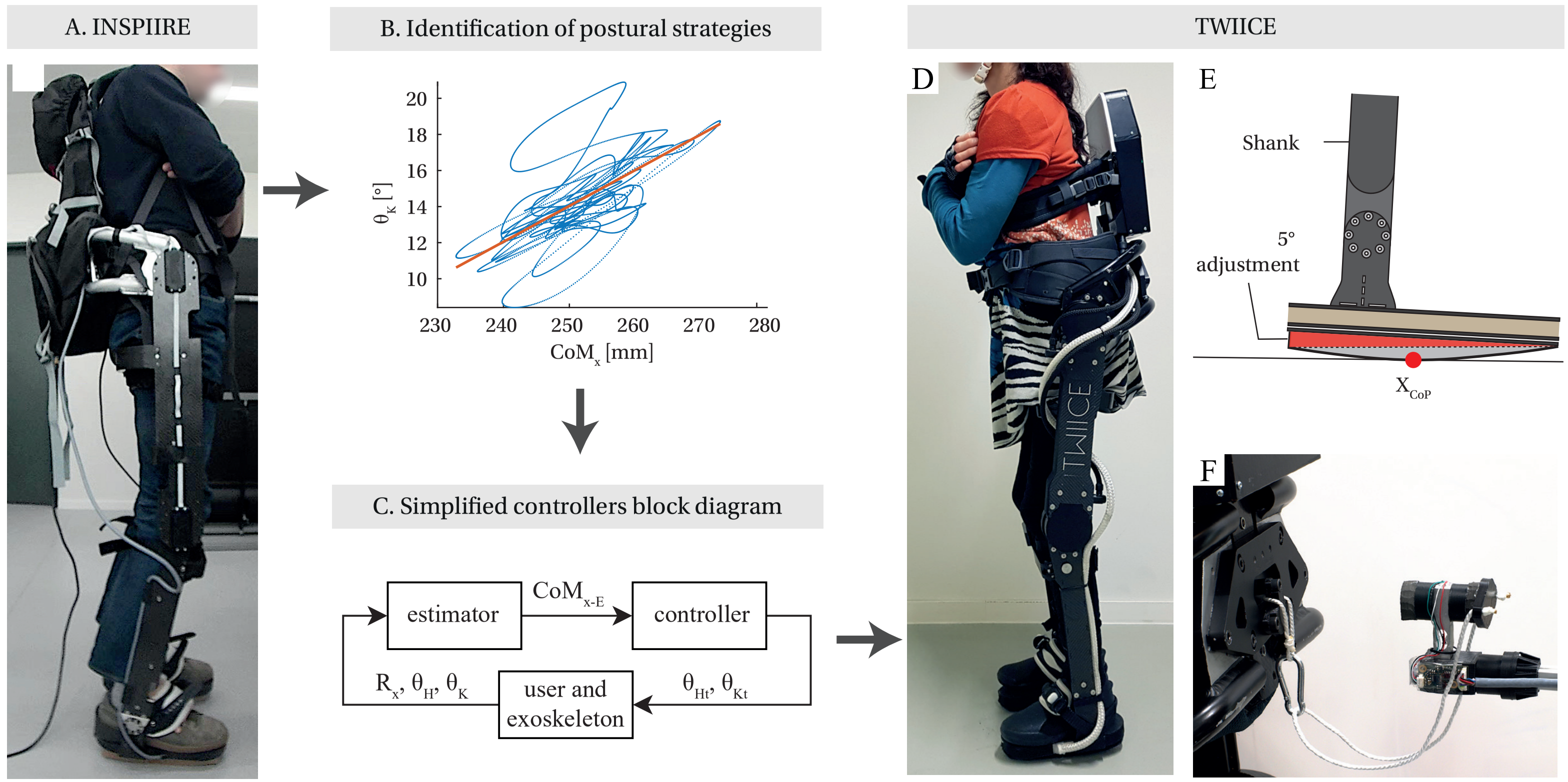

Figure 6.10 – **(A)** Healthy participant standing while being constrained by INSPIIRE, a passive exoskeleton. **(B)**. Identified relation between the knee angle and the CoMx for a typical young healthy participant in the eyes closed condition from [251] . **(C)** Overview of the controller block diagram. **(D)**. The TWIICE exoskeleton running with the knee controller (BKC) and a complete SCI user. **(E)** Close-up view on the TWIICE foot, with the rounded sole and the 5° wedge. **(F)** Instrumented push/pull stick.

mechanical design and the control framework are the same, while the actuators are more compact and more powerful [253]. TWIICE One provides two active DoFs per leg for the flexion/extension of the hip and knee joints in the sagittal plane. The ankle joints are locked at 90°. To match the experimental conditions of the passive exoskeleton [251], the soles have been modified. The foot soles are then fully curved (no flat part) to prevent passive postural stability. Their 0.65m radius is the same as the previous study, which is smaller than the height of the CoM of the test-pilot, so passive equilibrium is not possible. The sole is 227mm long, which corresponds to a maximum range of 242mm for the contact point, when the sole is rolling on the floor. The top part of the exoskeleton foot is tilted forward by 5°, so when the middle of the foot is in contact with the floor, the shank has a pitch angle of 5° with respect to the ground [252], Figure 6.10.E. The soles are covered with a rubber layer, in order to prevent slippage when standing. The width of the BoS is 244mm, measured between the two outer faces of the sole skates. The orientation (pitch and roll angles) is estimated from the inertial measurement unit (IMU) data with a simple complementary filter algorithm similar to [254]. Instead of using the trunk IMU as in [252], the IMU located in the left foot is used instead. It is expected to increase the performance for two reasons. First, when swinging fore-aft the whole body, the foot is the location with the lowest linear acceleration, which makes the state estimation more accurate. Second, there is more vibrations in the trunk, that is less rigid and in a cantilever configuration, which can generate closed-loop self-sustained uncontrolled oscillations. The embedded computer of TWIICE collects at 1 kHz the data from the inertial measurement unit and the joints encoders.

Proposed postural controllers

Baseline Knee Controller Drawing inspiration from the human sensorimotor adaptation investigated in [251], a novel postural position controller has been implemented and tested on TWIICE with no user [252]. This "Baseline Knee Controller" (BKC) regulates the balance with a proportional-derivative (PD) controller, setting the angle of the knee, and fed with the CoM position (Figure 6.11). The "Baseline Knee Controller" was described, simulated and experimentally tested. Basically, the knees are flexed proportionally to the estimated position of the CoM. This makes the foot sole rotate forward and backward, and move the point of contact with the floor. Since the sole is only in contact with the ground at one point, this point corresponds also to the center of pressure, xCoP, on the antero-posterior axis. CoM_x is the position of the projection on the ground of the CoM, in the antero-posterior axis. Its origin is defined at the middle of the foot when it is in contact with the ground. In this BKC controller, the CoM_x estimation is computed using a simple 2D model consisting of 3 segments (foot to knee, knee to hip, trunk including the head). The trunk length was measured on the user, while the shank and thigh lengths were obtained from the 3D model of the exoskeleton. The masses were obtained by summing the pilot's and exoskeleton's segments. The masses of

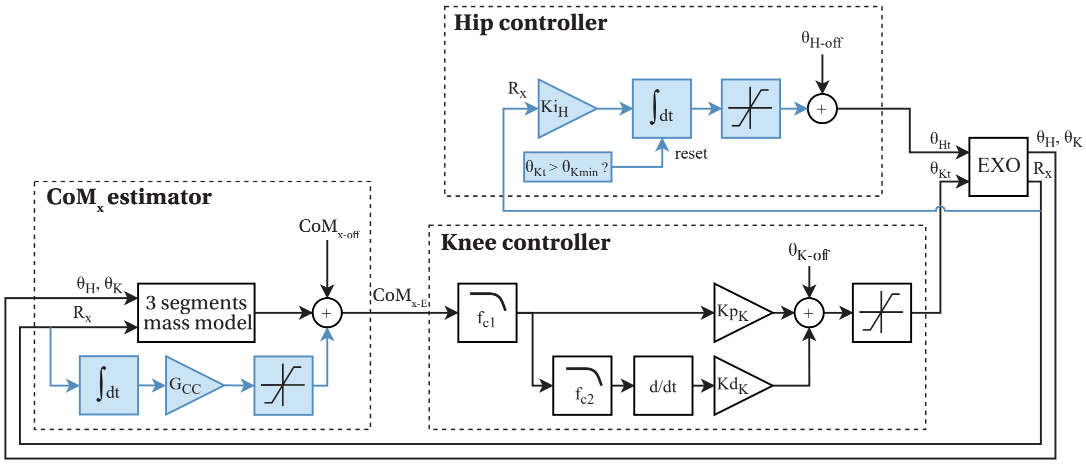

Figure 6.11 – Block diagram of the two controllers. The blue boxes represent the addition of the extended controller.

the user segments were estimated from the full bodyweight using the mass repartition from [255]. Finally, an offset CoM$_{x-off}$ is added to the estimation of CoM$_x$ to obtain CoMx-E, that is called CoM$_{x-E1}$ in the BKC case. This offset is necessary because the model is not accurate. CoMx-E is first filtered by a low-pass filter with a cut-off frequency f$_{c1}$, then fed into a proportional-derivative controller (PD) with the parameters KpK (proportional part gain) and KdK (derivative part gain). Before differentiation, the signal is filtered by a stronger low-pass filter with a cut-off frequency f$_{c2}$. This gives a knee flexion angle, which is offset by Θ_{K-off} to increase the flexion, and thus avoids hyperextension of the knee when the output of the BKC controller is negative. For safety, the value is finally clamped to the range [2° to 40°]. The hip joint is fixed at the angle Θ_{H-off}.

Extended Knee Controller A pilot study with the BKC controller has demonstrated its ability to make a complete SCI user stand dynamically with TWIICE. However, it was performing poorly for the task of grabbing heavy objects (several kilograms), unless they were close to the body. The first reason is that the CoM$_{x-E1}$ computation is not accurate since it does not consider the added mass. The other reason is that the controller is managing the balance by moving the position of the CoP along the foot length, but this does not work in the case the added weight shifts the CoM beyond the span of the feet.

The current section considers to improving the BKC controller to overcome the latter issues: 1) inaccuracy on the measurement of the CoM, and 2) inaccuracy on the position of the CoP. This Extended Knee Controller, which additionally takes into account a Hip strategy, is sketched out in Figure 6.11, blue boxes.

The first change concerns the extension of the CoM_x estimator to adapt the model online when a constant perturbation (added mass or horizontal force) arises. This is done by adding an integrator-like term to the calculation of the CoM. This term is obtained by integrating the angle of the foot pitch angle (Rx) and multipling by a gain (GCC) (Figure 6.11, CoM_x estimator, blue boxes). It is called CoM_{x-E2} for this controller. The idea is that in case of a permanent perturbation, the CoP will move durably, closer to the limits of the foot, which decreases the margin of stability in this direction. Continuously increasing the CoM_{x-E} offset will increase the correction of the PD controller, until the sole starts to roll in the other direction. This means that if a steady state exists, the center part of the foot will be in contact with the floor. A PID controller, which involves an integrator in the regulator instead, would not have the same effects. This will not be proved analytically here, but heuristically, in case of constant perturbation, the steady state will be reached when the CoM_{x-E} reaches zero, but the CoP will probably not be on the center of the foot, so the margin of stability is lower in one direction.

Second, EKC adds the hip contribution when the knee reaches the full extension. In the situation when the knee reaches the full extension, an integrator with a gain KiH will gradually increase the hip flexion angle, to bring the trunk forward, and thus shift the CoM toward the front (Figure 6.11, Hip controller, blue boxes). This flexion angle is limited to 60° for safety. If the knee is not saturated anymore, this integrator will go to zero smoothly at a 2°/s rate.

Simulations for a posterior perturbation have been performed with the same Simulink simulation environment described in (Baud et al. 2019). The results are on Figure 6.12. We notice that with BKC, the system is stable during the perturbation, but the equilibrium position is not on the center of the foot (foot pitch of 6°), which leaves little margin for further pushes. With EKC, the foot also reaches the same 6° of pitch angle when the perturbation is applied but returns slowly to horizontal, which allows further pushes. However, when the perturbation is removed, the CoM_x overshoots momentarily backward.

The overall behavior of the two controllers at steady state can be seen in Figure 6.13. In the BKC case, the system statically resists the perturbation by keeping the CoP more in front (B) or in the rear of the foot (C). In the EKC case, the system resists statically by keeping the CoM toward the back (D) or toward the front (E), such that the point of contact is at the center of the foot. In the last case (F), the pulling force is stronger, and the knee reaches the full extension and cannot extend more. The hip then flexes to move the CoM even more in front.

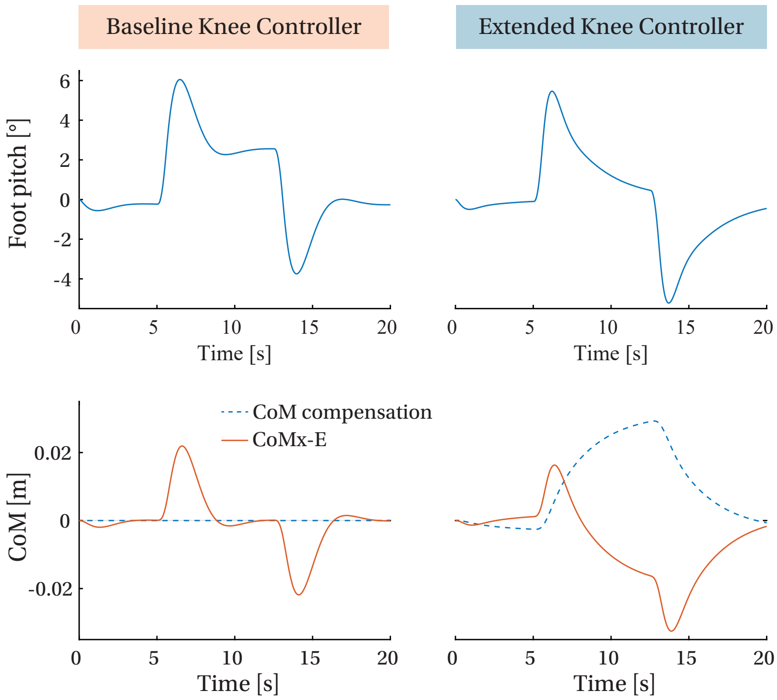

Figure 6.12 – Simulation results comparing BKC (with PID) and EKC when subject to a constant horizontal perturbation force. The horizontal pushing perturbation starts at t = 5 s, stops at t = 13 s, has an intensity of 20 N and a ramping time of 0.5 s.

Table 6.1 – TWIICE user description

User Profile	
Gender	Female
Years post injury	9
Lesion level	T10
AIS scale	A
Weight [kg]	46
Height [cm]	161
Knee ROM [deg]	0–160+
Hip ROM [deg]	-30–160+
Spasticity (Hips and knees)	MAS 0

6.3.3 Methods

Test-pilot: complete SCI user

The two postural controllers were tested with one chronic (10 years post-injury) and functionally complete SCI subject (ASIA A) with a lesion at the T10 level. Pilot's trunk control was limited as she had very little strength and control of the erector spinae. Her weight was 46kg and height was 1m61 (see Table 6.1. She had no spasticity most of the time (MAS 0) except for a slight ankle plantar flexion (MAS 1) in some instances. Because of her former profession (circus artist) her knee and hip ranges of motion were particularly large. No bone density measurements were carried out to screen for osteoporosis.
She provided informed consent after explanation of the possible risks associated with the use of exoskeletons as well as to those associated to the developing nature of the device [256].

Protocol

The experiment was composed of four tasks: 1) quiet standing, 2) pulse perturbations, 3) Static pull and push forces, and 4) object lifting perturbation. The test-pilot was instructed to keep her arms crossed and look straight at a cross on the wall in front of her during the whole experiment. The floor is made of hard linoleum floor, with virtually no rolling resistance.

At all times, there were one spotter in front and one behind the pilot to catch her in case of a fall, since the controllers will not trigger a step when the stability margin is exceeded. The spotters' hands are very close to the exoskeleton handles or the pilot's body to ensure quick grabbing in case of loss of balance. Contacts only occur in case of falling to prevent biasing the implemented postural controllers. The usual harness, cable and support frame cannot be used because the cable would disturb the balance, probably positively.

TUNING DAY The tuning of the controller parameters occurred during a preliminary session, 10 days before the actual experiment.

QUIET STANDING For both controllers, 1 min of quiet standing was performed in order to compare the sway amplitude without any perturbation.

PULSE PERTURBATIONS Then, the back part of the exoskeleton was pushed and pulled horizontally at the pelvic height (960mm above the ground level) with a stick to characterize accurately the responsiveness and stability of the two postural controllers. The experimenter applies the force from behind the pilot, so that she cannot expect the pulses. The stick is

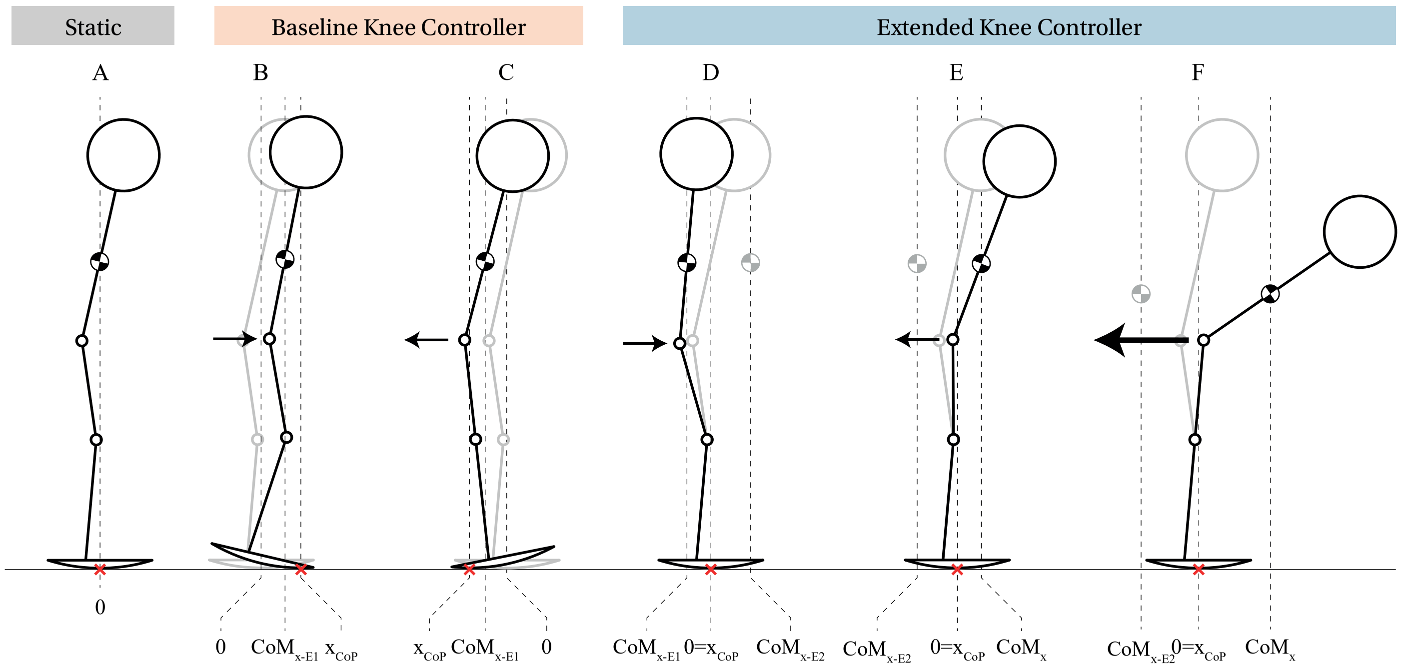

Figure 6.13 – Stick-figures showing the behavior of both controllers, at steady-state. The gray stick-figure is the initial rest position. The arrows indicate the direction of the perturbation while their size is proportional to the perturbation amplitude. The red crosses represent the position of the point of contact with the ground, which is equivalent to the CoP in the sagittal axis (x_{CoP}). The gray stick figures in background are the initial equilibrium position, same as the A

instrumented with a load cell, mounted with a stiff string such that it can push (posterior perturbation) and pull (anterior perturbation) the exoskeleton, or apply virtually no force when the pusher is not in contact and the string is loose (Figure 6.10.F). A custom amplifier and sampling board is also mounted on the stick, based on the ADS1146 chip (Texas Instruments, USA). It is wired to the exoskeleton embedded computer with four loose thin wires (0.129 mm^2) in order to apply only minimal parasitic force on the exoskeleton. This allows the exoskeleton to log the load cell signal with the same time base as the exoskeleton data, to avoid the manual synchronization step after the experiment. As in [197], the perturbation amplitude is defined by the push/pull force multiplied by the perturbation duration. The experimenter was keeping the pulses duration short and as constant as possible. As long as the user is swinging, the experimenter does not interfere with the movement. When the participant regains its steady state, meaning that the sway velocity is below 0.015 rad/s (0.86°/s) for more than 2 s, the LED on the backpack turns green and a new perturbation can be applied. The perturbations are applied randomly by the experimenter. The supervision laptop counts the perturbations and sorts them into the weak/medium/strong categories for both directions, to help the experimenter applying all types of perturbations.

STATIC PULL AND PUSH FORCES To assess the performance of both controllers during prolonged perturbations, the maximum applied force that is sustainable before losing balance was measured in both directions. The experimenter was instructed to push with the instrumented stick, increasing slowly and monotonically the force, until static equilibrium is lost. This procedure was repeated 3 times, and then was reiterated also 3 times by pulling the test-pilot backward. The user is caught and brought back to the vertical position by the experimenter at the end of each trial, so the recovery cannot be evaluated.

OBJECT LIFTING PERTURBATION Finally, to define the anterior static margin of stability in a situation close to an actual use case, the test-pilot was asked to lift a barbell in front of her and raise it gently at the shoulder height with the arms straight frontward, then lower it down. The mass of the barbell was changed from 0 kg (i.e. weight of the arms only) to 6 kg with increments of 2 kg. Each mass was lifted once. The task was failed if the spotters had to catch the test-pilot to prevent the fall, or if the test-pilot is unable to complete the task in less than 1 minute.

Data analysis

The analysis of the stability will be performed using the CoM_{x-E1} metric, because there was no extra instrumentation that could measure the actual CoM_x, and CoM_{x-E2} would be irrelevant when considering the static pull and push perturbations.

For quiet standing, a high-pass filter was applied to CoM$_{x-E1}$ to discard potential position shift due to the slow head movement of the test-pilot. Then, the root mean square (RMS) of the CoM$_{x-E1}$ was used to evaluate the amplitude of body sway for both controllers. For the pull and push task, perturbations with duration deviating more than 0.1s from the median duration were excluded. Thus, for each controller, only responses with similar perturbation duration were analyzed. Then, pull and push perturbations were sorted each in three categories based on the distribution of the perturbation magnitude. These categories were the same for the two controllers.
The main assessment metrics were the recovery time and the maximal perturbation magnitude that the controllers can handle in both directions. The recovery time was defined as the time needed after a perturbation for the CoM$_{x-E1}$ velocity to fall below a threshold set to 0.005 m/s. A moving average filter with a 36-sample window was applied to the CoM$_{x-E1}$ derivative. The maximal sustainable pulse perturbation amplitude in both directions was defined by the maximum perturbation amplitude that does not result in a loss of balance.
For the maximum sustainable pull and push force, the average of the 3 peak forces in each direction was computed.

We ran a repeated measures ANOVA to assess the effect of Controller and Category on the recovery time for dynamic antero-posterior perturbations. A repeated measure ANOVA was also used to evaluate the effect of Controller and Perturbation Direction on maximum pull and push forces. An alpha level of 0.05 was used on all statistics.

6.3.4 Results

Tuning session

The regulator gains were first set to zero to disable the closed-loop control. θ_{K-off} was fixed arbitrarily, then $x_{CoM-off}$, θ_{H-off} and were obtained by hand-tuning such that x_{CoM-E1} is zero when the exoskeleton stands still in the unstable equilibrium position, while the middle of the sole in contact with the ground. This procedure was repeated several times, to maximize θ_{K-off} under the condition that the posture is comfortable for the test-pilot.

Then, the low-pass filters and the PD parameters were tuned to maximize the disturbance rejection performance while no self-sustained oscillations or vibrations can be observed. Finally, Ki_H and GCC were tuned to the highest value that does not generate self-sustained oscillations. The results of the parameters tuning session are shown in Table 6.2.

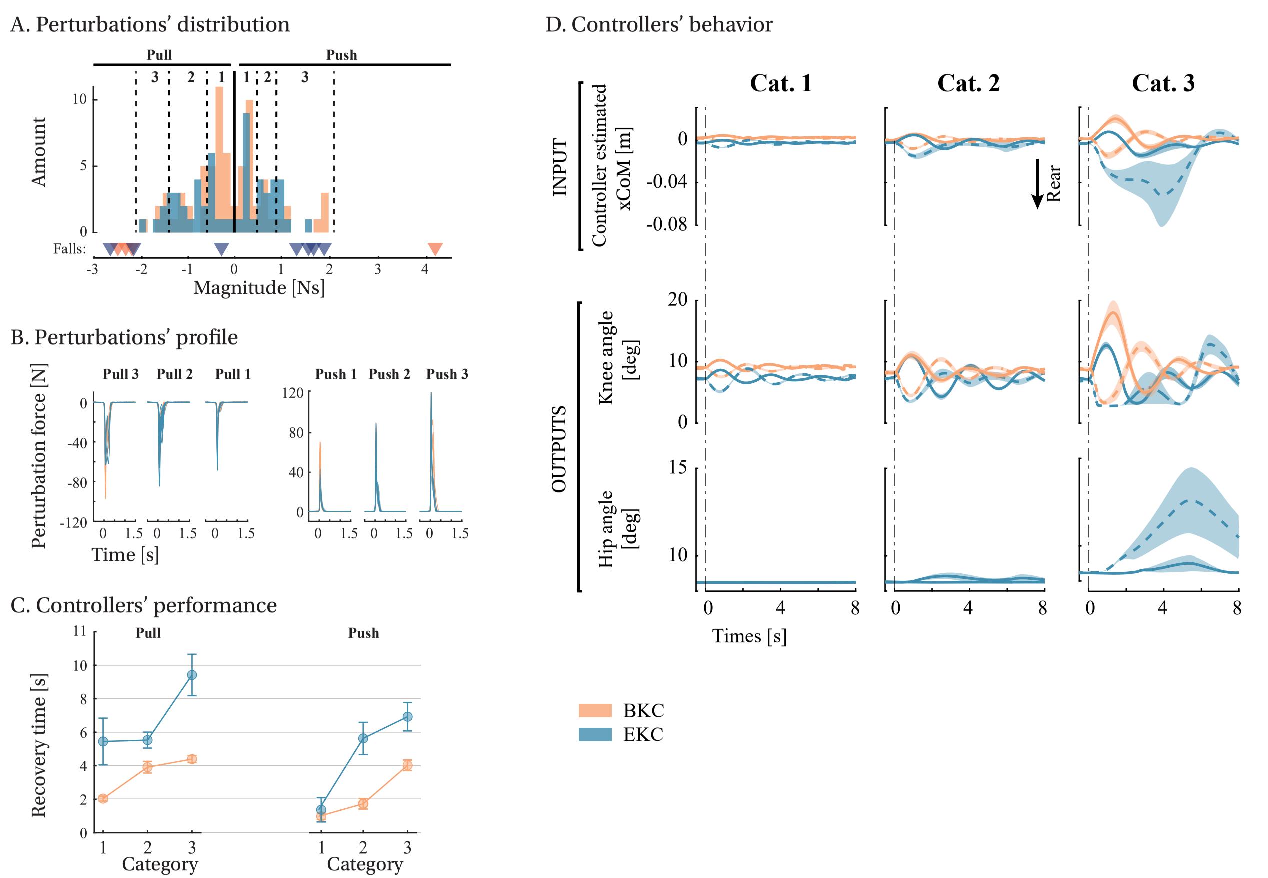

Figure 6.14 – **(A)** The perturbations of similar magnitude are grouped in 3 categories for each direction. Underneath the histograms, perturbations that led to a fall are presented by a Triangle. **(B)** Individual perturbation force profiles are shown by category. **(C)** Controllers' performance is evaluated with the recovery time. **(D)** average system response of the two controllers subject to the 3 categories of perturbation Magnitudes. Solid lines represent posterior perturbations, while dotted lines denote anterior ones. Overall, the colors correspond to the controllers (orange for KC and blue for EKC).

Parameters	BKC value	EKC value
$x_{CoM-off}$	0.04 m	
T1	0.05 s	
T2	0.2 s	
θ_{K-off}	8°	
θ_{H-off}	0°	
Kpk	420°/m	
KdK	110°/(m/s)	
KiH	0°/(°.s)	0.3°/(°.s)
Gcc	0 m (°.s)	0.000002 m/(°.s)

Table 6.2 – Parameters values of the two postural controllers

Quiet standing

The oscillation frequency is similar in both cases: 0.60 Hz for KC and 0.63 Hz for EKC. It was computed by finding the frequency of the highest peak in the Fourier transform of the x_{CoM-E1} signal. The RMS of the body sway was also similar (0.31 mm for KC and 0.38 mm for EKC).

Pulse perturbations

For BKC, 74 perturbations were applied, resulting in 4 fall initiations and 1 exclusion. For EKC, the test-pilot underwent 63 perturbations, including 7 fall initiations and 1 exclusion. The distribution of the perturbations magnitude and perturbations forces by category are shown in Figure 6.14.A and B. The average perturbation duration was 0.18 ± 0.006 s and 0.19 ± 0.006 s for BKC and EKC, respectively.

To characterize the robustness of the controllers, we determined the maximum anterior (pull) and posterior (push) perturbation amplitude the controllers can bear before a fall starts. For BKC, the maximal anterior perturbation magnitude that can be sustained is about 2 N.s. Beyond that threshold, 3 backward falls were recorded (see Figure 6.14.A, orange triangles). The threshold for posterior perturbations is between 2-4 N.s. A push with a magnitude of 4 N.s triggered a frontal fall. For EKC, the maximal anterior perturbation magnitude is also around 2 N.s. Two perturbations above this threshold triggered a backward fall. The maximal threshold for posterior perturbations is between 1.2 - 1.6 N.s. Indeed, 4 falls were observed when the perturbation magnitude was above this threshold (see Figure 6.14.A, blue triangles). It is important to note that the falls were in the backward direction although the perturbations were posterior (pushes). In summary, BKC was more robust than EKC for posterior perturbations, while they performed similarly for anterior perturbations.

To assess the performance of the controllers, the average duration of the measured recovery time has been extracted and plotted on Figure 6.14.C. Statistical analysis revealed a main effect of Controllers ($F=29.9$, $p<0.001$) and Category ($F=14.31$, $p<0.01$), but no interaction. Post-hoc analysis showed that BKC recovered significantly faster than EKC ($p<0.001$) and that recovery time significantly increased for each successive category ($p_{cat1-cat2}=0.033$, $p_{cat2-cat3}=0.025$).

The average system response is shown on Figure 6.14.D. For the perturbation categories 1 and 2, the response is similar, although the oscillations last longer with the EKC. There are more differences for the category 3. The pulling perturbations for EKC are producing a larger deviation of the CoM (4 cm instead of 2 cm for the other conditions), because the full extension of the knee was reached, and lowered the control capability. This is only the case for EKC, because initially, the knee was less flexed (the steady-state was not exactly the same), so there is less margin before the full extension is reached. It is also noticeable that even for the pushing perturbation, the hip contribution is used. This is because the oscillations have a high amplitude and a low damping, this is why the system also reaches the backward position and result in saturating the knee angle in full extension and starts using the hip contribution.

Maximum push and pull forces

The statistical analysis revealed a significant interaction between Controller and Perturbation Direction ($p=0.024$). The maximum pushing force for EKC was significantly higher (75.07 ± 3.55 N) than for BKC (13.69 ± 3.38 N, $p=0.009$). Although the maximum pulling force is also higher for EKC (27.91 ± 6.46 N) than for BKC (13.26 ± 1.15 N), the difference is not significant ($p=0.101$). EKC can sustain significantly higher static forces while the test-pilot is pushed forward than pull backward ($p=0.006$), while there is no significant effect of perturbation direction for BKC ($p=0.92$). Figure 6.15 shows an overview of the postural responses.

Object lifting perturbations

The results of this test are visible on Figure 6.16. With BKC, the test-pilot could lift his arms but failed to lift the 2 kg barbells because she fell forward before reaching the shoulder height, even though the ascent was slow. With EKC, the test-pilot could successfully lift the 2kg, 4kg and 6kg barbells.

6.3.5 Discussion

The goal of this study was to test and compare two postural controllers with a low-actuator count exoskeleton and a complete SCI pilot. Both controllers were able to manage quiet standing with almost no body sway and to cope with anterior-posterior perturbations. In that respect, postural adaptation strategies observed in healthy participants with a passive exoskeleton have been successfully transferred onto an active full-mobilization exoskeleton.

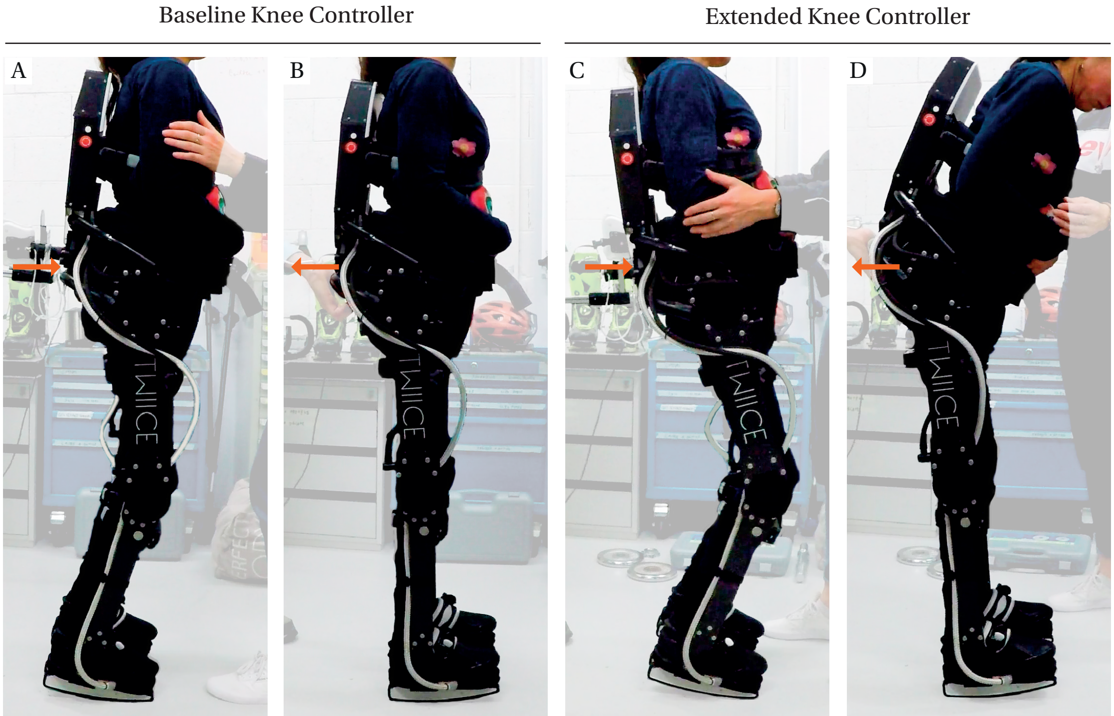

Figure 6.15 – Overview of postural responses for static perturbations

This results in the first exoskeleton with only two degrees of freedom per leg able to balance during standing with a complete SCI user. Overall, the EKC controller was more performant, although the recovery time is slightly slower. For pulse perturbations, EKC damps the oscillations more slowly due to its integrative behavior, and thus has significantly higher recovery time. Moreover, the falling direction was not always the same as the perturbation direction. Since the knees can flex more than they can extend in the actual configuration, it would possible to increase posterior margin of stability by increasing the knee offset angle (θ_{K-off}). However, it would imply that the hip flexion angle should also be increased to remain balanced, which results in an unnatural crouch standing. This also causes more load on the interfaces and in particular the trunk belt, which was reported to be an uncomfortable posture by the test-pilot during the tuning session. Overall, since the arms does not help to support the trunk through the crutches, the upper belt of the exoskeleton maintaining the torso should be sufficient and comfortable. For static perturbations, EKC could sustain significantly higher pulling forces (up to 7kg) thanks to the torso adjustment and repositioning of the CoM. It is important to note that EKC could resist even higher pulling forces, just by increasing the value of the maximum flexion of the hip, and if these forces change slowly. This would however be even more difficult to recover from. The main motivation for the curved sole is to walk by rolling the foot on the floor, to compensate for the lack of a mobile ankle joint. However, it also enables the use of these balance controllers, which could give more stability than passive balance with a flat sole

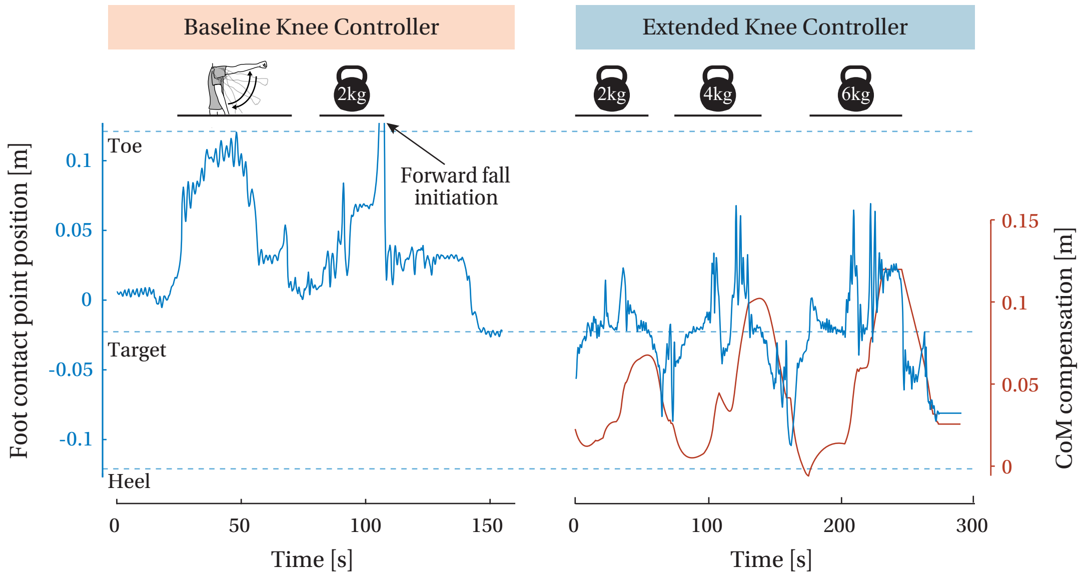

Figure 6.16 – Object lifting perturbations. Foot contact point position (x_{CoP}) along the curved soles during object lifting perturbations. For BKC, the x_{CoP} of one successful arm lift and one failed barbell lift is shown. When the toe tip has been reached, the pilot fell forward, was catched by the spotters and therefore the x_{CoP} quickly returned close to zero. The barbell was then removed, and the system stabilization is highlighted by the x_{CoP} returning to the target position. For EKC, the x_{CoP} of 3 successful barbell lifts with increasing weights is represented, as well as the CoM compensation.

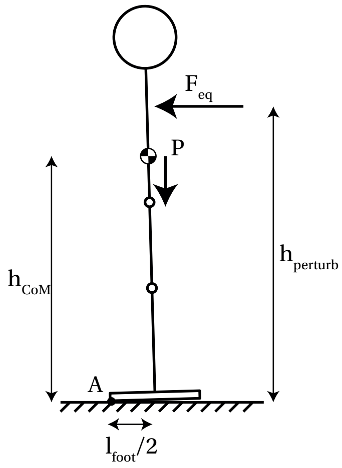

Figure 6.17 – Equivalent system for the static balance calculation with flat feet

of the same length. In practice, this ability was limited by the clamping on the knee and hip angles. The theoretical maximum force that the system could resist without falling in the same conditions can be computed by the simple static equilibrium model depicted in Figure 6.17. At equilibrium:

$$\sum M_A = 0 \Rightarrow \frac{l_{foot}}{2} P = F_{eq} h_{perturb} \Leftrightarrow F_{eq} = \frac{\frac{l_{foot}}{2} P}{h_{perturb}} = \frac{\frac{l_{foot}}{2} mg}{h_{perturb}} = 70.2N \tag{6.1}$$

So, the EKC can resist a higher pushing force (75.07 N) than the passive balance with a flat foot (70.2 N, see equation), but this is not the case with the pulling force (27.91 N). This limitation comes from the fact the knee cannot overextend, and that the hip joint was limited to 60° of flexion. The static perturbations assessment give us some functional insights on how much the pilot, while standing in the exoskeleton, could pull and push on object during daily activities such as opening a door, reaching for a pack of water on a supermarket shelf or closing a car trunk. The current EKC controllers could make this kind of activities possible without the need of crutches. Another functional assessment was the object lifting task. With the EKC controller, the user is able to manipulate a heavy object far from his body, but again, this is only possible if the movements are slow, otherwise the point of contact with ground may reach one end of the foot, and the user will start falling. Nevertheless, the EKC controller would for example enable to drink from a 1L bottle without worrying about balance management. To facilitate the user to apprehend the ability of the exoskeleton to manage balance, an acoustic or haptic feedback could be given when the CoP is close to the limits, so the user could for example decelerate the movement. Sensory feedback in addition to be warning signals could also promote embodiment, and thus facilitate acceptation of the device [257, 258]. More extensive training with the devices and the controller are necessary to further improve the performance and to apprehend the behavior the exoskeleton should follow in case of risk of fall.

Conclusion

A major result of this study was that postural adaptation strategies observed in healthy participants and elicited by standing in a passive exoskeleton could be ported onto an active exoskeleton with equivalent mobility. This conducted to the first full-mobilization exoskeleton able to balance during standing with only two degrees of freedom per leg. This could have important implications for the independence of individuals with paraplegia, their inclusion in social activities and their potential inclination to use an exoskeleton on a daily basis for the associated health benefits.

Conflict of Interest

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.

Author Contributions

JF conceived the idea and concept, designed the experiment, collected, analyzed and interpreted the data, and drafted the manuscript. RB and TV helped in conceiving the idea and concept. RB implemented the experiment, designed the controllers, participated in data acquisition, analysis and interpretation, and took part to the redaction. TV predominantly developed the two exoskeletons used in this study. AI and MB helped in drafting the manuscript and critically revising it.

6.4 Conclusion

Standing in a passive locked-ankle exoskeleton elicited an adaptation of their postural strategies. These changes were successfully measured by the passive exoskeleton and could be interpreted. We found that individuals during quiet standing flex and extend their knees to displace the center of pressure along the curved foot sole. The trunk moves only slightly to stay upright. However, we observed that a different strategy is adopted during strong external perturbations. Individuals moved their trunk in the opposite direction of the perturbation in order to reposition the CoM quickly. The observed trunk contribution can be associated to the hip strategy used in natural postural control.

The adapted postural strategies have then be implemented onto the active exoskeleton TWIICE One. This resulted in the first exoskeleton able to balance during standing with only two degree of freedom (DoF) per leg. It may have important implications in the propention of SCI user for adopting this kind of assistive devices in their daily activities.

7 Conclusion and outlook

In this final chapter, we summarize the key findings of this thesis and discuss the limitations and implications of our results. Subsequently, we relate the presented studies to previous research on error augmentation, pondering the importance of full-body agency in bodily ownership and self-consciousness. We also discuss how the sense of agency and ownership could contribute to enhance exoskeleton user experience.

7.1 Visual error augmentation for balance training

7.1.1 Main contributions and findings

The first goal of this thesis was to test the visual error augmentation paradigm on a balance task. The initial step was to find the most suitable balance exercise where visual acEA could be beneficial. We found in the literature that standing dynamic stability was trained in elderly and patients with stroke or Parkinson's disease with a goal-directed balance task, where patients were asked to control a cursor by tilting their body and reach virtual targets presented on a screen [32, 202]. In our study, the objective was to increase the stability margin, but also to promote directional control, i.e straight-line pass towards the targets. The design of the task was very similar to the one for arm reaching movement used in Patton et al. 2013 [16]. They were the first to demonstrate that visual EA can significantly increase the adaptation rate in healthy participants, as well as decrease the trajectory error in stroke survivors [18]. Therefore, this goal-directed balance task was suitable to assess the effect of visual error augmentation on balance control, in addition to be commonly used in rehabilitation programs.
Although we tested the paradigm only with young healthy participants, the experimental setup

was thought to be easily adopted in clinical settings. By using the balance rehabilitation device, called Thera-Trainer Coro, we ensured the feasibility of future studies with patients with balance disorder in a safe training environment. Furthermore, the setup used only one sensor, reducing experimental burden and set-up time. The transfer of motor learning techniques to neurorehabilitation is discussed in 7.1.3.
To increase the difficulty of the task for healthy young participants, a distortion of 30° was introduced, and we looked how fast they adapted.

We tested four visual error augmentation gains ranging from 1.25 to 2 times the error (25% to 100% of error amplification). The EA gains selection was based on previous studies on upper limbs which propose an optimal gain of 2 times the error for a simple reaching task [18, 16, 15, 17], but also on studies suggesting that the gain should be decreased if the task complexity increased [22, 21]. In addition, our pilot study (chapter 3) also indicated that the optimal gain for this balance task was smaller than 2. Indeed, there is evidence that high EA gains may destabilize the learning process [229, 22, 18, 16]. Therefore, we expected that the learning time as function of EA gains will decrease until it reaches a minimum located around 1.5, and then increase again due to the destabilization nature of high EA gains.

Our results demonstrate that the group with the gain of 1.5 (50% of amplification) adapted significantly faster to the visual distortion than the control group (no EA). Interestingly and in opposite to our pilot study, the highest gain tested (2) did not increase significantly the learning time with respect to the 1.5 gain. This indicates that the EA gains tested did not have a destabilizing action on the learning process. Consequently, EA gains up to 3 times the error, as in [18, 16] may be required to obtain a u-shaped curve for the learning time constant from which the optimal EA gain could be estimated. The small cohort (i.e. three participants per group) used in the the preliminary study may have misguided us in the selection range of the gain.

One main research question was whether visual error augmentation should push at times the errors over the motor awareness threshold, so that participants consciously correct for their kinematic errors. A first step to answer this question was to assess the relation between motor awareness and learning with error augmentation. Our results suggest that the optimal EA gain for each individual may be related to their motor awareness threshold. We found that an individual with good motor awareness may benefit from a higher EA gain in order to experience large and thus implicit errors in the early training stage. On the other hand, an individual with low motor awareness would be destabilized by high EA gains. In other words, visual error augmentation seems to promote explicit learning, which appears to be a good strategy for this adaptation task.

In summary, our findings suggest that the learning time of a simple balance task can be re-

duced thanks to visual error augmentation. Moreover, the optimal EA gain may be tailored to each individual as function of their motor awareness.
This could have implications in the way visual feedback is provided for balance rehabilitation. Indeed, visual error augmentation could be integrated in the emerging VR-based balance training [105, 106] for home-based and in-clinic rehabilitation. A main advantage of visual error augmentation, over haptic error augmentation for example, is the simplicity and affordability of the technical displays required. Finally, a motor awareness task could help tuning a rehabilitation protocol by predicting the optimal visual EA gain for each individual. This would be a first step towards personalized medicine for balance rehabilitation.

7.1.2 Limitations

"All too often we forget about the seductive and often misleading temporary changes in performance and take them to reflect learning when in fact, little persistence of that change is evident even after a short interval."

Winstein et al. 2003, [259]

A crucial component of motor learning is whether the performance, or in our case the adaptation, persists. In our study, there was no clear evidence that EA gains promote longer lasting after-effect. The group with the higher EA gain presents a significantly larger steady-state error than the control group during the washout phase. This may indicate that the washout phase was not long enough for this group to de-adapt completely, and therefore the retention was better for higher gains.
Moreover, we did not assess how the training with visual error augmentation generalize to similar tasks (i.e. transference) and only young healthy participants were tested. Therefore, the impact of balance training with visual error augmentation on patients with neurological diseases or elderly is not clear and should be assess in future experiments.

Since one of our interpretations is based on the amplitude of the errors experienced at the early stage of learning, further data analyses should be done to evaluate how motor performance (i.e compensation for the deviation) is related with the learning rate.

The relation between motor awareness and learning with visual error augmentation could not be directly investigated in the same task, since the adaptation task was also based on a visuomotor conflict. It would be interesting to apply this paradigm on a balance task that is *per se* challenging for healthy individuals or on patients with balance deficit, so the errors are not due to the visuomotor conflict, but to a poor balance control. That way, motor awareness could be tested with respect to the EA gain.

Finally, we did not test the effect of EA gains on motivation. Indeed, several studies reported that EA could lead to frustration and a decrease of motivation [120, 260, 21]. Our studies do not provide any insight on intrinsic motivation, which could be directly link to performance [261, 262].

7.1.3 Transfer of motor learning principles in neurorehabilitation

Rehabilitation can be defined as a process of relearning how to move [263, 264]. Consequently, it is generally accepted that recovery is a form of motor learning [265]. A better understanding of the mechanisms enhancing motor learning in healthy individuals can provide insights for the development of novel strategies for neurorehabilitation [266]. Since neurorehabilitation and motor learning are based on similar mechanisms of neural plasticity [157], it is believed that strategies that works on neurological intact individuals may work for neurorehabilitation [64]. Therefore, many rehabilitation approaches are based on theories of motor learning. The most fundamental principle of motor learning, applied to rehabilitation, states that skill acquisition is as function of the amount of practice [267].
Other rehabilitation principles stemmed from motor learning are task specificity, intensity, transference or intervention timing [163, 157].
Among others and as recent examples, two recent rehabilitation techniques based on motor learning principles are robot-assisted therapy [268, 226] and virtual reality-based rehabilitation [269, 270].

Although our findings need to be confirmed by more comprehensive data and longitudinal studies with healthy and neurological patients, they show promising effect of visual error augmentation for balance rehabilitation. A potential improvement could be the integration of visual error augmentation in a more immersive virtual environment. Indeed, the modularity of VR platforms offers an infinite possibility of controlled experimental designs for motor learning and for studying the sense of agency.

7.2 Full-body motor awareness

7.2.1 Main contributions and findings

Most of the studies that have addressed the sense of agency focused on how we consciously monitor our actions. These non-conceptual sensorimotor processes seem to be the only way to probe the actual limits of agency. Previous studies have shown that minor disruptions in the attended movement do not automatically enter awareness, although the motor corrections were performed [135, 26, 213, 28]. Similar motor awareness thresholds (6° 15°) were found for different body parts such as finger and arm, but also for tasks involving full-body motion such as walking. Following these results, it has been argued that the brain mechanism behind the sense of agency is effector-independent [27, 147]. However, this has never been tested in a matched task and within a single cohort of participants.

In this thesis, we explicitly tested the effector-independent hypothesis in a comparable task for hand and trunk. The task for the full-body was very similar to the Nielsen experiment (i.e. full-body tilt while standing over a mirror), while the hand task was similar to other upper limb agency study [271, 272]. The comparison of these two effectors was of particular interest due to the relevance of trunk-representation as central reference frame of bodily self-consciousness [34]. By investigating full-body agency during a balance task, we additionally looked at highly automated movements with important control centers in the brainstem [273], whereas upper limb control more strongly relies on cortical mechanisms [274]. Humans are normally able to correct for small balance perturbations without conscious control or without explicitly focusing on bodily information such as visual, vestibular and proprioceptive signals used to control balance [275].

Our results support the effector-independent hypothesis, but showed that motor awareness is strongly influenced by kinematic task demands.
In our first study on agency (chapter 4), we found that the motor awareness threshold is similar for hand and full-body in trials where the deviations were converging towards the targets location. Although we found a difference in the motor awareness thresholds for diverging trials between the two effectors, further data analysis revealed that this difference may results from a reaching velocity bias. Indeed, the reaching speed for the hand was significantly higher than for the full-body, which could have impaired the sense of agency. These results are in line with other studies supporting that an increased movement speed can hamper the sense of agency [222]. Therefore, the main limitation of our study comparing full-body and hand agency is that we did not control for reaching speed. In a second study (chapter 5), we also evaluated the full-body agency in a goal-directed balance task and we obtained a motor awareness threshold of 16°, which is in the same range as previous studies on body parts agency [26, 213, 139, 140] and full-body walking agency [28, 149].

Therefore, our findings corroborate the idea that effector-independent brain mechanisms are involved for agency in general, even when the control of actions relies on different mechanisms (i.e subcortical for gait and balance [276, 275] and cortical for upper-limbs [274].

Another important finding is the effect of kinematic demands on the sense of agency. Although this effect has been previously reported [28, 149], it has never been reproduced nor deeply studied. Our two studies on agency demonstrated that the motor awareness threshold was influenced by the deviation's side (i.e converging towards or diverging from the target side). The feeling of agency is increased when the kinematic demands is reduced, leading to more

self-attribution for converging trials. While predictive processes are often emphasized as central to the sense of agency [144], postdictive inferential processes also play a role in this kind of goal-directed task [230]. Kinematic demands may therefore be a form of goal-bias, where the ease to accomplish the task enhances the feeling of being in control. However, these findings are opposed to other studies suggesting that intentional effort enhance agency [220, 221].
Consequently, further work is needed to better understand the effect of kinematic demands on the sense of agency. Nevertheless, these opposing results support the hypothesis that the generation of goal-directed actions and the building of a conscious experience may be distinct brain processes [277, 128, 147].

7.2.2 Therapeutic relevance of the sense of agency

A better understanding of the fundamental aspects of the sense of agency may be of interest for improving therapeutic interventions. Experiencing the sense of agency is certainly a crucial aspect of sensorimotor rehabilitation, which is based on active patient contribution.
As previously discussed, motor awareness may be a parameter used to tune motor learning paradigms such as visual error augmentation. Furthermore, current neurorehabilitation trends rely on new technological developments such as assisting devices (e.g exoskeletons and prostheses) or virtual reality. The latter one often involves the visualization of a virtual body part such the arm or the hand or of a virtual avatar. Increasing embodiment over the virtual body through the sense of agency could contribute to the general motor performance, but also to motor learning.
In this direction, we plan to reproduce the Nielsen paradigm [136] in a virtual environment. Indeed, the Nielsen paradigm is very close the balance task we used for full-body agency. It is therefore very interesting to compare the motor awareness threshold obtained in an immersive virtual environment to the one obtained in our current studies, as well to the one of Nielsen.

The contribution of the sense of agency for the use of assisting devices, especially exoskeletons, was previously discussed in section 7.3.3.

Finally, the sense of agency can be disrupted under various pathological conditions such as schyzophrenia or anosognosia. While some of those pathologies are restricted to only body parts, others affect the global self [214, 278]. In addition, changes in the sense of agency is also a common feature of natural aging, and often linked to a physical impairment. [279].
Understanding the brain mechanisms behind the sense of agency may help developing interventions to remedy to these pathologies.

7.3 Balance management in a low-actuator count exoskeleton

7.3.1 Main contributions and findings

With the aim of developing a bioinspired postural controller for the full-mobilization exoskeleton TWIICE, five healthy young adults were asked to stand in a passive lower limb exoskeleton reproducing the same kinematic constrains. Locking of the ankle joint during standing elicited an adaptation of their postural strategies. These changes were successfully observed by the device and could be interpreted. We found that, during quiet standing, individuals flex and extend their knees to move the center of pressure along the curved foot soles, while the trunk moves slightly to stay straight. We observed for strong perturbations, that a different strategy is adopted. Individuals moves the trunk in the opposite direction of the perturbation in order to reposition the CoM quickly. The observed trunk contribution can be associated to the hip strategy used in natural postural control.

The adapted postural strategies have then be ported onto an active exoskeleton, the TWIICE. This resulted in the first exoskeleton able to balance during standing with only two DoF per leg which has important implications for the independence of patients, their inclusion in social activities and their potential inclination to use an exoskeleton on a daily basis for the associated health benefits.

Figure 7.1 shows a set of functional tasks that can be performed with the first balance controller we developed.
Those controllers have the advantage of requiring minimal sensing: An IMU to know the body tilt and encoders in the joints. Joints torque sensors or load cells below the feet are not required. This reduces the complexity and consequently improves the reliability of the device.

7.3.2 Limitations

A major challenge is to control the exoskeleton in an intuitive, reliable and safe manner. Unlike the majority of upper limbs movement, postural control errors may have severe consequences. An important question to raise is what are the adequate behaviors to adopt to avoid the risk of fall. So far, no security has been implemented on the TWIICE exoskeleton. The aim is to be able to use both hands in a standing position. However, where should the crutches be kept? And what happens if the user exceeds the balance capacity of the exoskeleton? Would it be wise to implement a stepping strategy in low-actuator-count exoskeleton? The behavior in case of risk of fall is very difficult to determine. Until today, in-home exoskeleton users need to have a companion to their side to be assisted and minimize risk of fall. Airbags have been developed to reduce the risk of injuries [280].
An efficient way to reduce the risk of fall might be the use of sensory feedback. Tactile feedback,

Figure 7.1 – Preliminary assessment of the balance controller with a SCI patient. The user could perform a set of tasks with hands free while remaining stable in standing. **A** Standing still. **B** Reaching for a 5-kg mass. **C** Depositing a 5-kg mass. **D** Lifting a ladder. **E-H** Playing with a ball, catching and tossing in various orientations.

being less obtrusive than auditory feedback, could be used to warn the user when close to the stability limits. Studies have demonstrated that tactile feedback can efficiently convey the states of an exoskeleton [281]. In addition to its practical use, haptic feedback may enhance ownership and embodiment [257, 258]. How these bodily components can enhance exoskeleton user experience will be discussed in the next section.

As other limitations, those controllers have only be tested in a controlled experimental environment. Their performance may be affected positively or negatively by the types of ground, whether they are soft or rigid, or even tilted. To assess the potential implications of such postural controllers for daily usage of lower-limb exoskeletons, they should be tested in less controlled scenarios. Moreover, only one case study is presented. Those controllers should be tested with different test-pilots to fully characterize their performance.

7.3.3 Applications of neuroscience principles

Rapid technological advances will contribute to enhance key features of wearable assistive devices such as light weight, speed and intelligent control. However, progress in this field should not be driven solely by engineering prowess. Pazzaglia and Molinari in 2016 [257] accurately enlightened that the development of wearable assistive devices should rely on the psychological and neuroscientific understanding of cortical body-representation and its modifiable aspect [130]. Walking again with an exoskeleton does not only imply the functional aspect, but also triggers emotional and psychological experiences. As an example, the Cybathlon competition which aims to promote the development of assisting devices sparked enthusiasm among the pilots who trained as real athletes.
A lingering question is how to enhance user experience. Studies have demonstrated that the human brain is capable to treat a tool as a part of their own body [282]. This special form of neuronal information processing is called embodiment [283]. Consequently, the sense of bodily borders may extend to mobility devices such as exoskeletons. The wheelchair is a striking example that tools can be represented in the body schema. Several testimonies reported that SCI patients no longer differentiate between their body and their wheelchair [284], and reference their body width to the one of the wheelchair [285]. Pazzaglia and colleagues suggested that body schema is modulated by a functional component which is the ability to move one's self, irrespective to a means [286]. Furthermore, it is the ability to control movements [287] and the feeling of being in control [288], which is defined as the sense of agency, that promote embodiment.
It has therefore been argued that enhancing the sense of agency could improve general user experience as well as sensorimotor functions [257, 289, 290].
Hence, to promote the acceptance of exoskeletons in daily activities and in clinical settings, an important effort has to be done to foster an active and predictive control of actions and awareness of them.

7.4 Conclusion

In conclusion, this thesis is an association of different research fields that could benefit from each other. We have shown that visual error augmentation is a promising strategy for balance training. Although it is generally accepted the sense of agency hinges on the forward model used for motor control and motor learning, few is known about the relation between the sense of agency and motor learning. In this thesis, we took a first step in this direction by demonstrating the optimal error augmentation gain may be related to motor awareness.
In addition, we tested whether the sense of agency is modulated by different effectors. Our results demonstrated that independently of the effector, conscious awareness is limited for spatial mismatches below 16°, even though analysis of motor performance illustrated au-

tomatic motor compensation for these sub-threshold visuomotor incongruencies. These findings are in line with previous studies on agency, and may explain why humans are able to maintain a unitary experience of self without cognitive effort.

Finally, the second part of this thesis is dedicated to an engineering approach which also aims to enhance postural control. Our research on human postural adaptation resulted in the first exoskeleton able to balance during standing with only two degrees of freedom per leg. We believe that bioinspired approaches, as well as embodied approaches, are required to increase the inclination of using exoskeleton in daily living and in clinical settings.

Bibliography

[1] Christina D. Davlin. "Dynamic Balance in High Level Athletes". en. In: *Perceptual and Motor Skills* 98.3_suppl (June 2004), pp. 1171–1176. ISSN: 0031-5125. DOI: 10.2466/pms.98.3c.1171-1176.

[2] Grace Vincent-Onabajo, Hadiza Y. Musa, and Emmanuel Joseph. "Prevalence of Balance Impairment Among Stroke Survivors Undergoing Neurorehabilitation in Nigeria". English. In: *Journal of Stroke and Cerebrovascular Diseases* 27.12 (Dec. 2018), pp. 3487–3492. ISSN: 1052-3057, 1532-8511. DOI: 10.1016/j.jstrokecerebrovasdis.2018.08.024.

[3] Ligie T. John, Binu Cherian, and Andrew Babu. "Postural Control and Fear of Falling in Persons with Low-Level Paraplegia". eng. In: *Journal of Rehabilitation Research and Development* 47.5 (2010), pp. 497–502. ISSN: 1938-1352. DOI: 10.1682/jrrd.2009.09.0150.

[4] Jeong-Ho Park, Yeo-Jeong Kang, and Fay Bahling Horak. "What Is Wrong with Balance in Parkinson's Disease?" In: *Journal of Movement Disorders* 8.3 (Sept. 2015), pp. 109–114. ISSN: 2005-940X. DOI: 10.14802/jmd.15018.

[5] Brooke Salzman. "Gait and Balance Disorders in Older Adults". en. In: *American Family Physician* 82.1 (July 2010), pp. 61–68. ISSN: 0002-838X, 1532-0650.

[6] Henning Stolze et al. "Falls in Frequent Neurological Diseases–Prevalence, Risk Factors and Aetiology". eng. In: *Journal of Neurology* 251.1 (Jan. 2004), pp. 79–84. ISSN: 0340-5354. DOI: 10.1007/s00415-004-0276-8.

[7] Natascha N Wesch, Barbi Law, and Craig R Hall. "The Use of Observational Learning by Athletes". en. In: (), p. 14.

[8] Souhail Hermassi et al. "Effect of Verbal Instruction on Motor Learning Ability of Anaerobic and Explosive Exercises in Physical Education University Students". English. In: *Frontiers in Psychology* 10 (2019). ISSN: 1664-1078. DOI: 10.3389/fpsyg.2019.02097.

[9] Roland Sigrist et al. "Terminal Feedback Outperforms Concurrent Visual, Auditory, and Haptic Feedback in Learning a Complex Rowing-Type Task". In: *Journal of Motor Behavior* 45.6 (Nov. 2013), pp. 455–472. ISSN: 0022-2895. DOI: 10.1080/00222895.2013.826169.

[10] Jaebong Lee and Seungmoon Choi. "Effects of Haptic Guidance and Disturbance on Motor Learning: Potential Advantage of Haptic Disturbance". In: *2010 IEEE Haptics Symposium*. Mar. 2010, pp. 335–342. DOI: 10.1109/HAPTIC.2010.5444635.

[11] Roland Sigrist et al. "Augmented Visual, Auditory, Haptic, and Multimodal Feedback in Motor Learning: A Review". eng. In: *Psychonomic Bulletin & Review* 20.1 (Feb. 2013), pp. 21–53. ISSN: 1531-5320. DOI: 10.3758/s13423-012-0333-8.

[12] D M Wolpert and Z Ghahramani. "Computational Principles of Movement Neuroscience". eng. In: *Nature neuroscience* 3 Suppl (Nov. 2000), pp. 1212–1217. ISSN: 1097-6256. DOI: 10.1038/81497.

[13] K A Thoroughman and R Shadmehr. "Learning of Action through Adaptive Combination of Motor Primitives". eng. In: *Nature* 407.6805 (Oct. 2000), pp. 742–747. ISSN: 0028-0836. DOI: 10.1038/35037588.

[14] R A Scheidt, J B Dingwell, and F A Mussa-Ivaldi. "Learning to Move amid Uncertainty". eng. In: *Journal of neurophysiology* 86.2 (Aug. 2001), pp. 971–985. ISSN: 0022-3077.

[15] Ian Sharp, Felix Huang, and James Patton. "Visual Error Augmentation Enhances Learning in Three Dimensions". In: *Journal of NeuroEngineering and Rehabilitation* 8 (Sept. 2011), p. 52. ISSN: 1743-0003. DOI: 10.1186/1743-0003-8-52.

[16] James L. Patton et al. "Visuomotor Learning Enhanced by Augmenting Instantaneous Trajectory Error Feedback during Reaching". In: *PLoS ONE* 8.1 (Jan. 2013). ISSN: 1932-6203. DOI: 10.1371/journal.pone.0046466.

[17] O. Celik, D. Powell, and M. K. O'Malley. "Impact of Visual Error Augmentation Methods on Task Performance and Motor Adaptation". In: *2009 IEEE International Conference on Rehabilitation Robotics*. June 2009, pp. 793–798. DOI: 10.1109/ICORR.2009.5209632.

[18] Yejun Wei et al. "Visual Error Augmentation for Enhancing Motor Learning and Rehabilitative Relearning". In: *9th International Conference on Rehabilitation Robotics, 2005. ICORR 2005*. June 2005, pp. 505–510. DOI: 10.1109/ICORR.2005.1501152.

[19] *Official Site – Wii Fit Plus*. http://wiifit.com/.

[20] *MindMotion GO – Gamified Neurorehabilitation*. en-US. https://www.mindmotionweb.com.

[21] Laura Marchal-Crespo et al. "Haptic Error Modulation Outperforms Visual Error Amplification When Learning a Modified Gait Pattern". English. In: *Frontiers in Neuroscience* 13 (2019). ISSN: 1662-453X. DOI: 10.3389/fnins.2019.00061.

[22] Ekin Basalp et al. "Visual Augmentation of Spatiotemporal Errors in a Rowing Task". en. In: *Human Movement and Technology : Book of Abstracts - 11th Joint Conference on Motor Control & Learning, Biomechanics & Training*. Shaker Verlag GmbH, 2016. ISBN: 978-3-8440-4707-3. DOI: 10.3929/ethz-a-010799461.

[23] Laura Marchal-Crespo, Nicole Rappo, and Robert Riener. "The Effectiveness of Robotic Training Depends on Motor Task Characteristics". eng. In: *Experimental Brain Research* 235.12 (Dec. 2017), pp. 3799–3816. ISSN: 1432-1106. DOI: 10.1007/s00221-017-5099-9.

[24] Laura Marchal-Crespo et al. "Learning a Locomotor Task: With or without Errors?" eng. In: *Journal of Neuroengineering and Rehabilitation* 11 (Mar. 2014), p. 25. ISSN: 1743-0003. DOI: 10.1186/1743-0003-11-25.

[25] B. Cesqui et al. "On the Use of Divergent Force Fields in Robot-Mediated Neurorehabilitation". In: *2008 2nd IEEE RAS EMBS International Conference on Biomedical Robotics and Biomechatronics*. Oct. 2008, pp. 854–861. DOI: 10.1109/BIOROB.2008.4762927.

[26] Pierre Fourneret and Marc Jeannerod. "Limited Conscious Monitoring of Motor Performance in Normal Subjects". en. In: *Neuropsychologia* 36.11 (Nov. 1998), pp. 1133–1140. ISSN: 00283932. DOI: 10.1016/S0028-3932(98)00006-2.

[27] Fritz Menzer et al. "Feeling in Control of Your Footsteps: Conscious Gait Monitoring and the Auditory Consequences of Footsteps". eng. In: *Cognitive Neuroscience* 1.3 (Sept. 2010), pp. 184–192. ISSN: 1758-8928. DOI: 10.1080/17588921003743581.

[28] O. A. Kannape et al. "The Limits of Agency in Walking Humans". en. In: *Neuropsychologia* 48.6 (May 2010), pp. 1628–1636. ISSN: 0028-3932. DOI: 10.1016/j.neuropsychologia.2010.02.005.

[29] O. A. Kannape and O. Blanke. "Self in Motion: Sensorimotor and Cognitive Mechanisms in Gait Agency". eng. In: *Journal of Neurophysiology* 110.8 (Oct. 2013), pp. 1837–1847. ISSN: 1522-1598. DOI: 10.1152/jn.01042.2012.

[30] Elmar Kal et al. "Does Implicit Motor Learning Lead to Greater Automatization of Motor Skills Compared to Explicit Motor Learning? A Systematic Review". en. In: *PLOS ONE* 13.9 (Sept. 2018), e0203591. ISSN: 1932-6203. DOI: 10.1371/journal.pone.0203591.

[31] Y. S. Lee and D. A. Vakoch. "Transfer and Retention of Implicit and Explicit Learning". eng. In: *British Journal of Psychology (London, England: 1953)* 87 (Pt 4) (Nov. 1996), pp. 637–651. ISSN: 0007-1269. DOI: 10.1111/j.2044-8295.1996.tb02613.x.

[32] Deborah S. Nichols. "Balance Retraining After Stroke Using Force Platform Biofeedback". en. In: *Physical Therapy* 77.5 (May 1997), pp. 553–558. ISSN: 0031-9023, 1538-6724.

[33] Olaf Blanke and Thomas Metzinger. "Full-Body Illusions and Minimal Phenomenal Selfhood". eng. In: *Trends in Cognitive Sciences* 13.1 (Jan. 2009), pp. 7–13. ISSN: 1364-6613. DOI: 10.1016/j.tics.2008.10.003.

[34] Olaf Blanke, Mel Slater, and Andrea Serino. "Behavioral, Neural, and Computational Principles of Bodily Self-Consciousness". eng. In: *Neuron* 88.1 (Oct. 2015), pp. 145–166. ISSN: 1097-4199. DOI: 10.1016/j.neuron.2015.09.029.

[35] Jane E. Aspell, Bigna Lenggenhager, and Olaf Blanke. "Keeping in Touch with One's Self: Multisensory Mechanisms of Self-Consciousness". en. In: *PLOS ONE* 4.8 (Aug. 2009), e6488. ISSN: 1932-6203. DOI: 10.1371/journal.pone.0006488.

[36] Bigna Lenggenhager et al. "Video Ergo Sum: Manipulating Bodily Self-Consciousness". en. In: *Science* 317.5841 (Aug. 2007), pp. 1096–1099. ISSN: 0036-8075, 1095-9203. DOI: 10.1126/science.1143439.

[37] *Spinal Cord Injury*. en. https://www.who.int/news-room/fact-sheets/detail/spinal-cord-injury.

[38] World Health Organization and Regional Office for South-East Asia. "Fact Sheet on Wheelchairs". en. In: (2010).

[39] Satoko Yasuda et al. "Return to Work after Spinal Cord Injury: A Review of Recent Research". eng. In: *NeuroRehabilitation* 17.3 (2002), pp. 177–186. ISSN: 1053-8135.

[40] A. Reitz et al. "Impact of Spinal Cord Injury on Sexual Health and Quality of Life". eng. In: *International Journal of Impotence Research* 16.2 (Apr. 2004), pp. 167–174. ISSN: 0955-9930. DOI: 10.1038/sj.ijir.3901193.

[41] P. Kennedy and L. Garmon-Jones. "Self-Harm and Suicide before and after Spinal Cord Injury: A Systematic Review". eng. In: *Spinal Cord* 55.1 (Jan. 2017), pp. 2–7. ISSN: 1476-5624. DOI: 10.1038/sc.2016.135.

[42] M. P. Jensen et al. "Frequency and Age Effects of Secondary Health Conditions in Individuals with Spinal Cord Injury: A Scoping Review". eng. In: *Spinal Cord* 51.12 (Dec. 2013), pp. 882–892. ISSN: 1476-5624. DOI: 10.1038/sc.2013.112.

[43] Aaron J. Young and Daniel P. Ferris. "State of the Art and Future Directions for Lower Limb Robotic Exoskeletons". en. In: *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 25.2 (Feb. 2017), pp. 171–182. ISSN: 1534-4320, 1558-0210. DOI: 10.1109/TNSRE.2016.2521160.

[44] Tristan Vouga et al. "TWIICE—A Lightweight Lower-Limb Exoskeleton for Complete Paraplegics". In: *Rehabilitation Robotics (ICORR), 2017 International Conference On*. IEEE, 2017, pp. 1639–1645.

[45] Shahabeddin Vahdat et al. "Functionally Specific Changes in Resting-State Sensorimotor Networks after Motor Learning". en. In: *Journal of Neuroscience* 31.47 (Nov. 2011), pp. 16907–16915. ISSN: 0270-6474, 1529-2401. DOI: 10.1523/JNEUROSCI.2737-11.2011.

[46] Daniel M. Wolpert and Zoubin Ghahramani. "Computational Motor Control". In: *Science* (2004).

[47] Jindrich Kodl, Gowrishankar Ganesh, and Etienne Burdet. "The CNS Stochastically Selects Motor Plan Utilizing Extrinsic and Intrinsic Representations". eng. In: *PloS One* 6.9 (2011), e24229. ISSN: 1932-6203. DOI: 10.1371/journal.pone.0024229.

[48] T Flash and N Hogan. "The Coordination of Arm Movements: An Experimentally Confirmed Mathematical Model". en. In: *The Journal of Neuroscience* 5.7 (July 1985), pp. 1688–1703. ISSN: 0270-6474, 1529-2401. DOI: 10.1523/JNEUROSCI.05-07-01688.1985.

[49] C. M. Harris and D. M. Wolpert. "Signal-Dependent Noise Determines Motor Planning". eng. In: *Nature* 394.6695 (Aug. 1998), pp. 780–784. ISSN: 0028-0836. DOI: 10.1038/29528.

[50] Ian O'Sullivan, Etienne Burdet, and Jörn Diedrichsen. "Dissociating Variability and Effort as Determinants of Coordination". en. In: *PLOS Computational Biology* 5.4 (Apr. 2009), e1000345. ISSN: 1553-7358. DOI: 10.1371/journal.pcbi.1000345.

[51] D. M. Wolpert, Z. Ghahramani, and M. I. Jordan. "An Internal Model for Sensorimotor Integration". eng. In: *Science (New York, N.Y.)* 269.5232 (Sept. 1995), pp. 1880–1882. ISSN: 0036-8075.

[52] Erich von Holst and Horst Mittelstaedt. "Das Reafferenzprinzip". de. In: *Naturwissenschaften* 37.20 (Jan. 1950), pp. 464–476. ISSN: 1432-1904. DOI: 10.1007/BF00622503.

[53] R. E. Kalman. "A New Approach to Linear Filtering and Prediction Problems" Transaction of the Asme~journal of Basic". In: 1960. DOI: 10.1115/1.3662552.

[54] Reza Shadmehr, Maurice A. Smith, and John W. Krakauer. "Error Correction, Sensory Prediction, and Adaptation in Motor Control". eng. In: *Annual Review of Neuroscience* 33 (2010), pp. 89–108. ISSN: 1545-4126. DOI: 10.1146/annurev-neuro-060909-153135.

[55] D. M. Merfeld, L. Zupan, and R. J. Peterka. "Humans Use Internal Models to Estimate Gravity and Linear Acceleration". eng. In: *Nature* 398.6728 (Apr. 1999), pp. 615–618. ISSN: 0028-0836. DOI: 10.1038/19303.

[56] A. D. Kuo. "An Optimal Control Model for Analyzing Human Postural Balance". eng. In: *IEEE transactions on bio-medical engineering* 42.1 (Jan. 1995), pp. 87–101. ISSN: 0018-9294. DOI: 10.1109/10.362914.

[57] Jürgen Konczak and Giovanni Abbruzzese. "Focal Dystonia in Musicians: Linking Motor Symptoms to Somatosensory Dysfunction". eng. In: *Frontiers in Human Neuroscience* 7 (2013), p. 297. ISSN: 1662-5161. DOI: 10.3389/fnhum.2013.00297.

[58] Justin Won and Neville Hogan. “Stability Properties of Human Reaching Movements”. en. In: *Experimental Brain Research* 107.1 (Nov. 1995), pp. 125–136. ISSN: 1432-1106. DOI: 10.1007/BF00228024.

[59] E. Burdet, M. Honegger, and A. Codourey. “Controllers with Desired Dynamic Compensation and Their Implementation on a 6 DOF Parallel Manipulator”. In: *Proceedings. 2000 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS 2000) (Cat. No.00CH37113)*. Vol. 1. Takamatsu, Japan: IEEE, 2000, pp. 39–45. ISBN: 978-0-7803-6348-9. DOI: 10.1109/IROS.2000.894579.

[60] R. Shadmehr and F. A. Mussa-Ivaldi. “Adaptive Representation of Dynamics during Learning of a Motor Task”. en. In: *Journal of Neuroscience* 14.5 (May 1994), pp. 3208–3224. ISSN: 0270-6474, 1529-2401. DOI: 10.1523/JNEUROSCI.14-05-03208.1994.

[61] Tie Wang, Goran S. Dordevic, and Reza Shadmehr. “Learning Dynamics of Reaching Movements Results in the Modification of Arm Impedance and Long-Latency Perturbation Responses”. In: *Biological cybernetics* 85.6 (Dec. 2001), pp. 437–448. ISSN: 0340-1200.

[62] N. Hogan. “Adaptive Control of Mechanical Impedance by Coactivation of Antagonist Muscles”. In: *IEEE Transactions on Automatic Control* 29.8 (Aug. 1984), pp. 681–690. ISSN: 2334-3303. DOI: 10.1109/TAC.1984.1103644.

[63] Etienne Burdet et al. “The Central Nervous System Stabilizes Unstable Dynamics by Learning Optimal Impedance”. en. In: *Nature* 414.6862 (Nov. 2001), pp. 446–449. ISSN: 1476-4687. DOI: 10.1038/35106566.

[64] Etienne Burdet, David W. Franklin, and Theodore E. Milner. *Human Robotics: Neuromechanics and Motor Control*. en. MIT Press, Sept. 2013. ISBN: 978-0-262-01953-8.

[65] Alexandra S. Pollock et al. “What Is Balance?:” en. In: *Clinical Rehabilitation* (July 2016). DOI: 10.1191/0269215500cr342oa.

[66] Yury Ivanenko and Victor S. Gurfinkel. “Human Postural Control”. In: *Frontiers in Neuroscience* 12 (Mar. 2018). ISSN: 1662-4548. DOI: 10.3389/fnins.2018.00171.

[67] Fay B. Horak. “Postural Control”. en. In: *Encyclopedia of Neuroscience*. Ed. by Marc D. Binder, Nobutaka Hirokawa, and Uwe Windhorst. Berlin, Heidelberg: Springer, 2009, pp. 3212–3219. ISBN: 978-3-540-29678-2. DOI: 10.1007/978-3-540-29678-2_4708.

[68] K. O. Berg et al. “Clinical and Laboratory Measures of Postural Balance in an Elderly Population”. ENG. In: *Archives of Physical Medicine and Rehabilitation* 73.11 (Nov. 1992), pp. 1073–1080. ISSN: 0003-9993.

[69] L. M. Nashner and H. Forssberg. “Phase-Dependent Organization of Postural Adjustments Associated with Arm Movements While Walking”. eng. In: *Journal of Neurophysiology* 55.6 (June 1986), pp. 1382–1394. ISSN: 0022-3077. DOI: 10.1152/jn.1986.55.6.1382.

[70] Lewis M. Nashner and Gin McCollum. "The Organization of Human Postural Movements: A Formal Basis and Experimental Synthesis". en. In: *Behavioral and Brain Sciences* 8.01 (Mar. 1985), p. 135. ISSN: 0140-525X, 1469-1825. DOI: 10.1017/S0140525X00020008.

[71] C. F Runge et al. "Ankle and Hip Postural Strategies Defined by Joint Torques". In: *Gait & Posture* 10.2 (Oct. 1999), pp. 161–170. ISSN: 0966-6362. DOI: 10.1016/S0966-6362(99)00032-6.

[72] John H. J. Allum, Mark G. Carpenter, and Flurin Honegger. "Directional Aspects of Balance Corrections in Man". eng. In: *IEEE engineering in medicine and biology magazine: the quarterly magazine of the Engineering in Medicine & Biology Society* 22.2 (2003 Mar-Apr), pp. 37–47. ISSN: 0739-5175. DOI: 10.1109/memb.2003.1195694.

[73] J. Massion. "Movement, Posture and Equilibrium: Interaction and Coordination". eng. In: *Progress in Neurobiology* 38.1 (1992), pp. 35–56. ISSN: 0301-0082. DOI: 10.1016/0301-0082(92)90034-c.

[74] Reza Shadmehr and Steven P. Wise. *The Computational Neurobiology of Reaching and Pointing*. en. MA: The MIT Press. Cambridge: The MIT Press, 2005.

[75] Lior Shmuelof, John W. Krakauer, and Pietro Mazzoni. "How Is a Motor Skill Learned? Change and Invariance at the Levels of Task Success and Trajectory Control". ENG. In: *Journal of Neurophysiology* 108.2 (July 2012), pp. 578–594. ISSN: 1522-1598. DOI: 10.1152/jn.00856.2011.

[76] Paul Morris Fitts and Michael I Posner. *Human Performance*. English. Belmont, Calif.: Brooks/Cole Pub. Co., 1967.

[77] John R. Anderson. "Acquisition of Cognitive Skill". In: *Psychological Review* 89.4 (1982), pp. 369–406. ISSN: 1939-1471(Electronic),0033-295X(Print). DOI: 10.1037/0033-295X.89.4.369.

[78] Melanie Kleynen et al. "Using a Delphi Technique to Seek Consensus Regarding Definitions, Descriptions and Classification of Terms Related to Implicit and Explicit Forms of Motor Learning". In: *PLoS ONE* 9.6 (June 2014). ISSN: 1932-6203. DOI: 10.1371/journal.pone.0100227.

[79] J. P. Maxwell, R. S. Masters, and F. F. Eves. "From Novice to No Know-How: A Longitudinal Study of Implicit Motor Learning". eng. In: *Journal of Sports Sciences* 18.2 (Feb. 2000), pp. 111–120. ISSN: 0264-0414. DOI: 10.1080/026404100365180.

[80] Mark J. Wagner and Maurice A. Smith. "Shared Internal Models for Feedforward and Feedback Control". en. In: *The Journal of Neuroscience* 28.42 (Oct. 2008), pp. 10663–10673. ISSN: 0270-6474, 1529-2401. DOI: 10.1523/JNEUROSCI.5479-07.2008.

[81] Jordan A. Taylor and Richard B. Ivry. "The Role of Strategies in Motor Learning". In: *Annals of the New York Academy of Sciences* 1251 (Mar. 2012), pp. 1–12. ISSN: 0077-8923. DOI: 10.1111/j.1749-6632.2011.06430.x.

[82] Mitsuo Kawato. "Feedback-Error-Learning Neural Network for Supervised Motor Learning". en. In: *Advanced Neural Computers.* Ed. by Rolf Eckmiller. Amsterdam: North-Holland, Jan. 1990, pp. 365–372. ISBN: 978-0-444-88400-8. DOI: 10.1016/B978-0-444-88400-8.50047-9.

[83] Ya-Weng Tseng et al. "Sensory Prediction Errors Drive Cerebellum-Dependent Adaptation of Reaching". eng. In: *Journal of Neurophysiology* 98.1 (July 2007), pp. 54–62. ISSN: 0022-3077. DOI: 10.1152/jn.00266.2007.

[84] Pietro Mazzoni and John W. Krakauer. "An Implicit Plan Overrides an Explicit Strategy during Visuomotor Adaptation". eng. In: *The Journal of Neuroscience: The Official Journal of the Society for Neuroscience* 26.14 (Apr. 2006), pp. 3642–3645. ISSN: 1529-2401. DOI: 10.1523/JNEUROSCI.5317-05.2006.

[85] Christopher T Noto and Farrel R Robinson. "Visual Error Is the Stimulus for Saccade Gain Adaptation". en. In: *Cognitive Brain Research* 12.2 (Oct. 2001), pp. 301–305. ISSN: 0926-6410. DOI: 10.1016/S0926-6410(01)00062-3.

[86] J. Johanna Hopp and Albert F. Fuchs. "The Characteristics and Neuronal Substrate of Saccadic Eye Movement Plasticity". eng. In: *Progress in Neurobiology* 72.1 (Jan. 2004), pp. 27–53. ISSN: 0301-0082. DOI: 10.1016/j.pneurobio.2003.12.002.

[87] Katharina Havermann and Markus Lappe. "The Influence of the Consistency of Post-saccadic Visual Errors on Saccadic Adaptation". In: *Journal of Neurophysiology* 103.6 (Apr. 2010), pp. 3302–3310. ISSN: 0022-3077. DOI: 10.1152/jn.00970.2009.

[88] John W. Krakauer et al. "Learning of Visuomotor Transformations for Vectorial Planning of Reaching Trajectories". en. In: *Journal of Neuroscience* 20.23 (Dec. 2000), pp. 8916–8924. ISSN: 0270-6474, 1529-2401. DOI: 10.1523/JNEUROSCI.20-23-08916.2000.

[89] Richard A. Schmidt and Craig A. Wrisberg. *Motor Learning and Performance: A Situation-Based Learning Approach.* en. Human Kinetics, 2008. ISBN: 978-0-7360-6964-9.

[90] Keith Nesbitt. "Designing Multi-Sensory Displays for Abstract Data". en. In: *PhD thesis, School of Information Technologies, University of Sydney, Australia.* (Jan. 2003).

[91] R. A. Schmidt and G. Wulf. "Continuous Concurrent Feedback Degrades Skill Learning: Implications for Training and Simulation". ENG. In: *Human Factors* 39.4 (Dec. 1997), pp. 509–525. ISSN: 0018-7208.

[92] C J Winstein, P S Pohl, and R Lewthwaite. "Effects of Physical Guidance and Knowledge of Results on Motor Learning: Support for the Guidance Hypothesis". eng. In: *Research quarterly for exercise and sport* 65.4 (Dec. 1994), pp. 316–323. ISSN: 0270-1367.

[93] Richard A. Schmidt. "Frequent Augmented Feedback Can Degrade Learning: Evidence and Interpretations". en. In: *Tutorials in Motor Neuroscience*. Ed. by Jean Requin and George E. Stelmach. NATO ASI Series 62. Springer Netherlands, 1991, pp. 59–75. ISBN: 978-94-010-5609-0 978-94-011-3626-6. DOI: 10.1007/978-94-011-3626-6_6.

[94] Antoinette Domingo and Daniel P Ferris. "The Effects of Error Augmentation on Learning to Walk on a Narrow Balance Beam". eng. In: *Experimental brain research. Experimentelle Hirnforschung. Expérimentation cérébrale* 206.4 (Oct. 2010), pp. 359–370. ISSN: 1432-1106. DOI: 10.1007/s00221-010-2409-x.

[95] Luc Proteau. "Visual Afferent Information Dominates Other Sources of Afferent Information during Mixed Practice of a Video-Aiming Task". en. In: *Experimental Brain Research* 161.4 (Mar. 2005), pp. 441–456. ISSN: 0014-4819, 1432-1106. DOI: 10.1007/s00221-004-2090-z.

[96] Wulf G. "Attentional Focus and Motor Learning: A Review of 10 Years of Research". In: (2007), pp. 4–14.

[97] John J. Buchanan and Chaoyi Wang. "Overcoming the Guidance Effect in Motor Skill Learning: Feedback All the Time Can Be Beneficial". eng. In: *Experimental Brain Research* 219.2 (June 2012), pp. 305–320. ISSN: 1432-1106. DOI: 10.1007/s00221-012-3092-x.

[98] C. J. Winstein et al. "Learning a Partial-Weight-Bearing Skill: Effectiveness of Two Forms of Feedback". ENG. In: *Physical Therapy* 76.9 (Sept. 1996), pp. 985–993. ISSN: 0031-9023.

[99] Catharine M. Walsh et al. "Concurrent Versus Terminal Feedback: It May Be Better to Wait". en-US. In: *Academic Medicine* 84.10 (Oct. 2009), S54. ISSN: 1040-2446. DOI: 10.1097/ACM.0b013e3181b38daf.

[100] Gabriele Wulf and Charles H. Shea. "Principles Derived from the Study of Simple Skills Do Not Generalize to Complex Skill Learning". en. In: *Psychonomic Bulletin & Review* 9.2 (June 2002), pp. 185–211. ISSN: 1069-9384, 1531-5320. DOI: 10.3758/BF03196276.

[101] Franz Marschall, Andreas Bund, and Josef Wiemeyer. "Does Frequent Augmented Feedback Really Degrade Learning? : A Meta Analysis". eng. In: *E-Journal Bewegung und Training* 1 (2007), S. 74–85. ISSN: 2123755-4.

[102] Annick A. A. Timmermans et al. "Technology-Assisted Training of Arm-Hand Skills in Stroke: Concepts on Reacquisition of Motor Control and Therapist Guidelines for Rehabilitation Technology Design". ENG. In: *Journal of Neuroengineering and Rehabilitation* 6 (Jan. 2009), p. 1. ISSN: 1743-0003. DOI: 10.1186/1743-0003-6-1.

[103] Jerzy Sadowski, Andrzej Mastalerz, and Tomasz Niznikowski. "Benefits of Bandwidth Feedback in Learning a Complex Gymnastic Skill". In: *Journal of Human Kinetics* 37 (2013 -7- 05), pp. 183–193. ISSN: 1640-5544. DOI: 10.2478/hukin-2013-0039.

[104] Yurong Mao et al. "Virtual Reality Training Improves Balance Function". In: *Neural Regeneration Research* 9.17 (Sept. 2014), pp. 1628–1634. ISSN: 1673-5374. DOI: 10.4103/1673-5374.141795.

[105] Thunyanoot Prasertsakul et al. "The Effect of Virtual Reality-Based Balance Training on Motor Learning and Postural Control in Healthy Adults: A Randomized Preliminary Study". In: *BioMedical Engineering OnLine* 17 (Sept. 2018). ISSN: 1475-925X. DOI: 10.1186/s12938-018-0550-0.

[106] Steven Phu et al. *Balance Training Using Virtual Reality Improves Balance and Physical Performance in Older Adults at High Risk of Falls*. English. https://www.dovepress.com/balance-training-using-virtual-reality-improves-balance-and-physical-p-peer-reviewed-article-CIA. Aug. 2019. DOI: 10.2147/CIA.S220890.

[107] Bernd Petzold et al. "A Study on Visual, Auditory, and Haptic Feedback for Assembly Tasks". In: *Presence: Teleoperators and Virtual Environments* 13.1 (Feb. 2004), pp. 16–21. ISSN: 1054-7460. DOI: 10.1162/105474604774048207.

[108] Ungyeon Yang and Gerard Jounghyun Kim. "Implementation and Evaluation of "Just Follow Me": An Immersive, VR-Based, Motion-Training System". In: *Presence* 11.3 (June 2002), pp. 304–323. ISSN: 1054-7460. DOI: 10.1162/105474602317473240.

[109] Michaël Huet et al. "Self-Controlled Concurrent Feedback and the Education of Attention towards Perceptual Invariants". In: *Human Movement Science* 28.4 (Aug. 2009), pp. 450–467. ISSN: 0167-9457. DOI: 10.1016/j.humov.2008.12.004.

[110] G. Wulf, M. Höß, and W. Prinz. "Instructions for Motor Learning: Differential Effects of Internal versus External Focus of Attention". ENG. In: *Journal of Motor Behavior* 30.2 (June 1998), pp. 169–179. ISSN: 0022-2895. DOI: 10.1080/00222899809601334.

[111] Laura Marchal-Crespo et al. "The Effect of Haptic Guidance, Aging, and Initial Skill Level on Motor Learning of a Steering Task". In: *Experimental Brain Research. Experimentelle Hirnforschung. Experimentation Cerebrale* 201.2 (Mar. 2010), pp. 209–220. ISSN: 0014-4819. DOI: 10.1007/s00221-009-2026-8.

[112] Jeffrey F. Israel et al. "Metabolic Costs and Muscle Activity Patterns during Robotic- and Therapist-Assisted Treadmill Walking in Individuals with Incomplete Spinal Cord Injury". eng. In: *Physical Therapy* 86.11 (Nov. 2006), pp. 1466–1478. ISSN: 0031-9023. DOI: 10.2522/ptj.20050266.

[113] D. M. Wolpert, R. C. Miall, and M. Kawato. "Internal Models in the Cerebellum". eng. In: *Trends in Cognitive Sciences* 2.9 (Sept. 1998), pp. 338–347. ISSN: 1364-6613. DOI: 10.1016/s1364-6613(98)01221-2.

[114] J.L. Emken and D.J. Reinkensmeyer. "Robot-Enhanced Motor Learning: Accelerating Internal Model Formation during Locomotion by Transient Dynamic Amplification". In: *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 13.1 (2005), pp. 33–39. ISSN: 1534-4320. DOI: 10.1109/TNSRE.2004.843173.

[115] James L Patton et al. "Evaluation of Robotic Training Forces That Either Enhance or Reduce Error in Chronic Hemiparetic Stroke Survivors". eng. In: *Experimental brain research. Experimentelle Hirnforschung. Expérimentation cérébrale* 168.3 (Jan. 2006), pp. 368–383. ISSN: 0014-4819. DOI: 10.1007/s00221-005-0097-8.

[116] Darcy S. Reisman et al. "Locomotor Adaptation on a Split-Belt Treadmill Can Improve Walking Symmetry Post-Stroke". eng. In: *Brain: A Journal of Neurology* 130.Pt 7 (July 2007), pp. 1861–1872. ISSN: 1460-2156. DOI: 10.1093/brain/awm035.

[117] James L. Patton et al. "Evaluation of Robotic Training Forces That Either Enhance or Reduce Error in Chronic Hemiparetic Stroke Survivors". ENG. In: *Experimental Brain Research* 168.3 (Oct. 2005), pp. 368–383. ISSN: 0014-4819. DOI: 10.1007/s00221-005-0097-8.

[118] Darcy S. Reisman et al. "Repeated Split-Belt Treadmill Training Improves Poststroke Step Length Asymmetry". en. In: *Neurorehabilitation and Neural Repair* 27.5 (June 2013), pp. 460–468. ISSN: 1545-9683. DOI: 10.1177/1545968312474118.

[119] Marie-Hélène Milot et al. "Comparison of Error-Amplification and Haptic-Guidance Training Techniques for Learning of a Timing-Based Motor Task by Healthy Individuals". eng. In: *Experimental Brain Research* 201.2 (Mar. 2010), pp. 119–131. ISSN: 1432-1106. DOI: 10.1007/s00221-009-2014-z.

[120] Jaime E. Duarte and David J. Reinkensmeyer. "Effects of Robotically Modulating Kinematic Variability on Motor Skill Learning and Motivation". eng. In: *Journal of Neurophysiology* 113.7 (Apr. 2015), pp. 2682–2691. ISSN: 1522-1598. DOI: 10.1152/jn.00163.2014.

[121] Laura Marchal-Crespo et al. "Effect of Error Augmentation on Brain Activation and Motor Learning of a Complex Locomotor Task". eng. In: *Frontiers in Neuroscience* 11 (2017), p. 526. ISSN: 1662-4548. DOI: 10.3389/fnins.2017.00526.

[122] Bambi Roberts Brewer, Roberta Klatzky, and Yoky Matsuoka. "Visual Feedback Distortion in a Robotic Environment for Hand Rehabilitation". eng. In: *Brain Research Bulletin* 75.6 (Apr. 2008), pp. 804–813. ISSN: 0361-9230. DOI: 10.1016/j.brainresbull.2008.01.006.

[123] Carlos Tobar et al. "The Effects of Visual Feedback Distortion with Unilateral Leg Loading on Gait Symmetry". eng. In: *Annals of Biomedical Engineering* 46.2 (Feb. 2018), pp. 324–333. ISSN: 1573-9686. DOI: 10.1007/s10439-017-1954-x.

[124] Kevin O'Brien, Charles R. Crowell, and James Schmiedeler. "Error Augmentation Feedback for Lateral Weight Shifting". eng. In: *Gait & Posture* 54 (May 2017), pp. 178–182. ISSN: 1879-2219. DOI: 10.1016/j.gaitpost.2017.03.003.

[125] Jeannine Bergmann et al. "Virtual Reality to Augment Robot-Assisted Gait Training in Non-Ambulatory Patients with a Subacute Stroke: A Pilot Randomized Controlled Trial". eng. In: *European Journal of Physical and Rehabilitation Medicine* 54.3 (June 2018), pp. 397–407. ISSN: 1973-9095. DOI: 10.23736/S1973-9087.17.04735-9.

[126] Lukas Zimmerli et al. "Increasing Patient Engagement during Virtual Reality-Based Motor Rehabilitation". eng. In: *Archives of Physical Medicine and Rehabilitation* 94.9 (Sept. 2013), pp. 1737–1746. ISSN: 1532-821X. DOI: 10.1016/j.apmr.2013.01.029.

[127] Shaun Gallagher. "Philosophical Conceptions of the Self: Implications for Cognitive Science". en. In: *Trends in Cognitive Sciences* 4.1 (Jan. 2000), pp. 14–21. ISSN: 1364-6613. DOI: 10.1016/S1364-6613(99)01417-5.

[128] Marc Jeannerod. "The Mechanism of Self-Recognition in Humans". eng. In: *Behavioural Brain Research* 142.1-2 (June 2003), pp. 1–15. ISSN: 0166-4328.

[129] Marc Jeannerod and Elisabeth Pacherie. "Agency, Simulation and Self-Identification". en. In: *Mind and Language* 19.2 (Apr. 2004), pp. 113–146. ISSN: 0268-1064, 1468-0017. DOI: 10.1111/j.1468-0017.2004.00251.x.

[130] Olaf Blanke. "Multisensory Brain Mechanisms of Bodily Self-Consciousness". eng. In: *Nature Reviews. Neuroscience* 13.8 (July 2012), pp. 556–571. ISSN: 1471-0048. DOI: 10.1038/nrn3292.

[131] Matthis Synofzik, Gottfried Vosgerau, and Albert Newen. "Beyond the Comparator Model: A Multifactorial Two-Step Account of Agency". eng. In: *Consciousness and Cognition* 17.1 (Mar. 2008), pp. 219–239. ISSN: 1090-2376. DOI: 10.1016/j.concog.2007.03.010.

[132] Nicole David, Albert Newen, and Kai Vogeley. "The "Sense of Agency" and Its Underlying Cognitive and Neural Mechanisms". en. In: *Consciousness and Cognition* 17.2 (June 2008), pp. 523–534. ISSN: 10538100. DOI: 10.1016/j.concog.2008.03.004.

[133] Sarah Jayne Blakemore and Chris Frith. "Self-Awareness and Action". eng. In: *Current Opinion in Neurobiology* 13.2 (Apr. 2003), pp. 219–224. ISSN: 0959-4388. DOI: 10.1016/s0959-4388(03)00043-6.

[134] Patrick Haggard, Sam Clark, and Jeri Kalogeras. "Voluntary Action and Conscious Awareness". en. In: *Nature Neuroscience* 5.4 (Apr. 2002), pp. 382–385. ISSN: 1546-1726. DOI: 10.1038/nn827.

[135] Torsten Ingemann Nielsen. "Volition: A New Experimental Approach". en. In: *Scandinavian Journal of Psychology* 4.1 (1963), pp. 225–230. ISSN: 1467-9450. DOI: 10.1111/j.1467-9450.1963.tb01326.x.

[136] Torsten Ingemann Nielsen. "Analysis and Synthesis of Human Acting, Concerning the Subject and from the Standpoint of the Subject". In: (1978).

[137] C. D. Frith, S. J. Blakemore, and D. M. Wolpert. "Abnormalities in the Awareness and Control of Action". eng. In: *Philosophical Transactions of the Royal Society of London. Series B, Biological Sciences* 355.1404 (Dec. 2000), pp. 1771–1788. ISSN: 0962-8436. DOI: 10.1098/rstb.2000.0734.

[138] N. Franck et al. "Defective Recognition of One's Own Actions in Patients with Schizophrenia". eng. In: *The American Journal of Psychiatry* 158.3 (Mar. 2001), pp. 454–459. ISSN: 0002-953X. DOI: 10.1176/appi.ajp.158.3.454.

[139] C. Farrer and C. D. Frith. "Experiencing Oneself vs Another Person as Being the Cause of an Action: The Neural Correlates of the Experience of Agency". eng. In: *NeuroImage* 15.3 (Mar. 2002), pp. 596–603. ISSN: 1053-8119. DOI: 10.1006/nimg.2001.1009.

[140] Matthis Synofzik, Peter Thier, and Axel Lindner. "Internalizing Agency of Self-Action: Perception of One's Own Hand Movements Depends on an Adaptable Prediction about the Sensory Action Outcome". eng. In: *Journal of Neurophysiology* 96.3 (Sept. 2006), pp. 1592–1601. ISSN: 0022-3077. DOI: 10.1152/jn.00104.2006.

[141] Sotaro Shimada, Yuan Qi, and Kazuo Hiraki. "Detection of Visual Feedback Delay in Active and Passive Self-Body Movements". en. In: *Experimental Brain Research* 201.2 (Mar. 2010), pp. 359–364. ISSN: 1432-1106. DOI: 10.1007/s00221-009-2028-6.

[142] D. M. Wolpert and J. R. Flanagan. "Motor Prediction". eng. In: *Current biology: CB* 11.18 (Sept. 2001), R729–732. ISSN: 0960-9822. DOI: 10.1016/s0960-9822(01)00432-8.

[143] S. J. Blakemore, C. D. Frith, and D. M. Wolpert. "Spatio-Temporal Prediction Modulates the Perception of Self-Produced Stimuli". eng. In: *Journal of Cognitive Neuroscience* 11.5 (Sept. 1999), pp. 551–559. ISSN: 0898-929X. DOI: 10.1162/089892999563607.

[144] Sarah Jayne Blakemore, Daniel M. Wolpert, and Christopher D. Frith. "Abnormalities in the Awareness of Action". eng. In: *Trends in Cognitive Sciences* 6.6 (June 2002), pp. 237–242. ISSN: 1879-307X. DOI: 10.1016/s1364-6613(02)01907-1.

[145] Hyeong-Dong Park and Olaf Blanke. “Heartbeat-Evoked Cortical Responses: Underlying Mechanisms, Functional Roles, and Methodological Considerations”. eng. In: *NeuroImage* 197 (Aug. 2019), pp. 502–511. ISSN: 1095-9572. DOI: 10.1016/j.neuroimage.2019.04.081.

[146] Olaf Blanke et al. “Stimulating Illusory Own-Body Perceptions”. eng. In: *Nature* 419.6904 (Sept. 2002), pp. 269–270. ISSN: 0028-0836. DOI: 10.1038/419269a.

[147] O. A. Kannape and O. Blanke. “Agency, Gait and Self-Consciousness”. eng. In: *International Journal of Psychophysiology: Official Journal of the International Organization of Psychophysiology* 83.2 (Feb. 2012), pp. 191–199. ISSN: 1872-7697. DOI: 10.1016/j.ijpsycho.2011.12.006.

[148] Oliver A. Kannape et al. “Distinct Locomotor Control and Awareness in Awake Sleepwalkers”. eng. In: *Current biology: CB* 27.20 (Oct. 2017), R1102–R1104. ISSN: 1879-0445. DOI: 10.1016/j.cub.2017.08.060.

[149] Oliver Alan Kannape et al. “Cognitive Loading Affects Motor Awareness and Movement Kinematics but Not Locomotor Trajectories during Goal-Directed Walking in a Virtual Reality Environment”. In: *PLoS ONE* 9.1 (Jan. 2014). ISSN: 1932-6203. DOI: 10.1371/journal.pone.0085560.

[150] Clarissa Barros de Oliveira et al. “Balance Control in Hemiparetic Stroke Patients: Main Tools for Evaluation”. eng. In: *Journal of Rehabilitation Research and Development* 45.8 (2008), pp. 1215–1226. ISSN: 1938-1352.

[151] F. B. Horak and H. C. Diener. “Cerebellar Control of Postural Scaling and Central Set in Stance”. eng. In: *Journal of Neurophysiology* 72.2 (Aug. 1994), pp. 479–493. ISSN: 0022-3077. DOI: 10.1152/jn.1994.72.2.479.

[152] F. B. Horak, J. Frank, and J. Nutt. “Effects of Dopamine on Postural Control in Parkinsonian Subjects: Scaling, Set, and Tone”. en. In: *Journal of Neurophysiology* 75.6 (June 1996), pp. 2380–2396. ISSN: 0022-3077, 1522-1598. DOI: 10.1152/jn.1996.75.6.2380.

[153] J. V. Jacobs and F. B. Horak. “Cortical Control of Postural Responses”. en. In: *Journal of Neural Transmission* 114.10 (Mar. 2007), p. 1339. ISSN: 1435-1463. DOI: 10.1007/s00702-007-0657-0.

[154] *Basic Facts about Balance Problems | Aging & Health A-Z | American Geriatrics Society | HealthInAging.Org.* https://www.healthinaging.org/a-z-topic/balance-problems/basic-facts.

[155] Jocelyn E. Harris et al. “Relationship of Balance and Mobility to Fall Incidence in People with Chronic Stroke”. eng. In: *Physical Therapy* 85.2 (Feb. 2005), pp. 150–158. ISSN: 0031-9023.

[156] Lone Jørgensen, Torgeir Engstad, and Bjarne K. Jacobsen. "Higher Incidence of Falls in Long-Term Stroke Survivors Than in Population Controls Depressive Symptoms Predict Falls After". en. In: *Stroke* 33.2 (Jan. 2002), pp. 542–547. ISSN: 0039-2499, 1524-4628. DOI: 10.1161/hs0202.102375.

[157] Jeffrey A. Kleim and Theresa A. Jones. "Principles of Experience-Dependent Neural Plasticity: Implications for Rehabilitation after Brain Damage". eng. In: *Journal of speech, language, and hearing research: JSLHR* 51.1 (Feb. 2008), S225–239. ISSN: 1092-4388. DOI: 10.1044/1092-4388(2008/018).

[158] Craig D. Takahashi et al. "Robot-Based Hand Motor Therapy after Stroke". eng. In: *Brain: A Journal of Neurology* 131.Pt 2 (Feb. 2008), pp. 425–437. ISSN: 1460-2156. DOI: 10.1093/brain/awm311.

[159] Rosemary A. Bosnell et al. "Motor Practice Promotes Increased Activity in Brain Regions Structurally Disconnected after Subcortical Stroke". eng. In: *Neurorehabilitation and Neural Repair* 25.7 (Sept. 2011), pp. 607–616. ISSN: 1552-6844. DOI: 10.1177/1545968311405675.

[160] Yuemin Ding, Abba J. Kastin, and Weihong Pan. "Neural Plasticity After Spinal Cord Injury". In: *Current pharmaceutical design* 11.11 (2005), pp. 1441–1450. ISSN: 1381-6128.

[161] James V. Lynskey, Adam Belanger, and Ranu Jung. "Activity-Dependent Plasticity in Spinal Cord Injury". In: *Journal of rehabilitation research and development* 45.2 (2008), pp. 229–240. ISSN: 0748-7711. DOI: 10.1682/JRRD.2007.03.0047.

[162] G. M. Petzinger et al. "Exercise-Enhanced Neuroplasticity Targeting Motor and Cognitive Circuitry in Parkinson's Disease". In: *Lancet neurology* 12.7 (July 2013), pp. 716–726. ISSN: 1474-4422. DOI: 10.1016/S1474-4422(13)70123-6.

[163] C. J. Winstein and S. L. Wolf. "Task-Oriented Training to Promote Upper Extremity Recovery". en. In: *Stroke Recovery and Rehabilitation.* Demos Medical Publishing. 17. New York, Nov. 2008, pp. 267–290.

[164] P. A. Goldie, T. M. Bach, and O. M. Evans. "Force Platform Measures for Evaluating Postural Control: Reliability and Validity". eng. In: *Archives of Physical Medicine and Rehabilitation* 70.7 (July 1989), pp. 510–517. ISSN: 0003-9993.

[165] A. Shumway-Cook, D. Anson, and S. Haller. "Postural Sway Biofeedback: Its Effect on Reestablishing Stance Stability in Hemiplegic Patients". ENG. In: *Archives of Physical Medicine and Rehabilitation* 69.6 (June 1988), pp. 395–400. ISSN: 0003-9993.

[166] C. Walker, B. J. Brouwer, and E. G. Culham. "Use of Visual Feedback in Retraining Balance Following Acute Stroke". ENG. In: *Physical Therapy* 80.9 (Sept. 2000), pp. 886–895. ISSN: 0031-9023.

[167] I.-Chun Chen et al. “Effects of Balance Training on Hemiplegic Stroke Patients”. ENG. In: *Chang Gung Medical Journal* 25.9 (Sept. 2002), pp. 583–590. ISSN: 2072-0939.

[168] E. Bisson et al. “Functional Balance and Dual-Task Reaction Times in Older Adults Are Improved by Virtual Reality and Biofeedback Training”. ENG. In: *Cyberpsychology & Behavior: The Impact of the Internet, Multimedia and Virtual Reality on Behavior and Society* 10.1 (Feb. 2007), pp. 16–23. ISSN: 1094-9313. DOI: 10.1089/cpb.2006.9997.

[169] C. M. Sackley and N. B. Lincoln. “Single Blind Randomized Controlled Trial of Visual Feedback after Stroke: Effects on Stance Symmetry and Function”. ENG. In: *Disability and Rehabilitation* 19.12 (Dec. 1997), pp. 536–546. ISSN: 0963-8288.

[170] C. J. Winstein et al. “Standing Balance Training: Effect on Balance and Locomotion in Hemiparetic Adults”. ENG. In: *Archives of Physical Medicine and Rehabilitation* 70.10 (Oct. 1989), pp. 755–762. ISSN: 0003-9993.

[171] Ruth Ann Geiger et al. “Balance and Mobility Following Stroke: Effects of Physical Therapy Interventions With and Without Biofeedback/Forceplate Training”. en. In: *Physical Therapy* 81.4 (Apr. 2001), pp. 995–1005. ISSN: 0031-9023, 1538-6724.

[172] Abhishek Srivastava et al. “Post-Stroke Balance Training: Role of Force Platform with Visual Feedback Technique”. In: *Journal of the Neurological Sciences* 287.1–2 (Dec. 2009), pp. 89–93. ISSN: 0022-510X. DOI: 10.1016/j.jns.2009.08.051.

[173] Hamid Bateni. “Changes in Balance in Older Adults Based on Use of Physical Therapy vs the Wii Fit Gaming System: A Preliminary Study”. In: *Physiotherapy*. Special Issue on Advancing Technology Including Papers from WCPT 98.3 (Sept. 2012), pp. 211–216. ISSN: 0031-9406. DOI: 10.1016/j.physio.2011.02.004.

[174] Wen-Chieh Yang et al. “Home-Based Virtual Reality Balance Training and Conventional Balance Training in Parkinson’s Disease: A Randomized Controlled Trial”. eng. In: *Journal of the Formosan Medical Association = Taiwan Yi Zhi* 115.9 (Sept. 2016), pp. 734–743. ISSN: 0929-6646. DOI: 10.1016/j.jfma.2015.07.012.

[175] Sevgi Sevi Yeşilyaprak et al. “Comparison of the Effects of Virtual Reality-Based Balance Exercises and Conventional Exercises on Balance and Fall Risk in Older Adults Living in Nursing Homes in Turkey”. eng. In: *Physiotherapy Theory and Practice* 32.3 (2016), pp. 191–201. ISSN: 1532-5040. DOI: 10.3109/09593985.2015.1138009.

[176] Ana R. C. Donati et al. “Long-Term Training with a Brain-Machine Interface-Based Gait Protocol Induces Partial Neurological Recovery in Paraplegic Patients”. en. In: *Scientific Reports* 6.1 (Sept. 2016). ISSN: 2045-2322. DOI: 10.1038/srep30383.

[177] Solaiman Shokur et al. "Training with Brain-Machine Interfaces, Visuo-Tactile Feedback and Assisted Locomotion Improves Sensorimotor, Visceral, and Psychological Signs in Chronic Paraplegic Patients". en. In: *PLOS ONE* 13.11 (Nov. 2018), e0206464. ISSN: 1932-6203. DOI: 10.1371/journal.pone.0206464.

[178] LE Miller, AK Zimmerman, and WG Herbert. "Clinical Effectiveness and Safety of Powered Exoskeleton-Assisted Walking in Patients with Spinal Cord Injury: Systematic Review with Meta-Analysis. - PubMed - NCBI". In: *Medical Devices* 9 (2016), pp. 455–466. DOI: 10.2147/MDER.S103102.

[179] S Federici et al. "The Effectiveness of Powered, Active Lower Limb Exoskeletons in Neurorehabilitation: A Systematic Review. - PubMed - NCBI". In: *NeuroRehabilitation* 37.3 (2015), pp. 321–340. DOI: 10.3233/NRE-151265.

[180] Dennis R. Louie et al. "Gait Speed Using Powered Robotic Exoskeletons after Spinal Cord Injury: A Systematic Review and Correlational Study". In: *Journal of NeuroEngineering and Rehabilitation* 12.82 (2015). DOI: 10.1186/s12984-015-0074-9.

[181] Ashraf S Gorgey. "Robotic Exoskeletons: The Current Pros and Cons. - PubMed - NCBI". In: *World Journal of Orthopedics* 9.9 (2018), pp. 112–119. ISSN: 2218-5836. DOI: 10.5312/wjo.v9.i9.112.

[182] Saso Jezernik et al. "Robotic Orthosis Lokomat: A Rehabilitation and Research Tool". In: *Neuromodulation* 6.2 (2003), pp. 108–115.

[183] Jan F. Veneman et al. "Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation - IEEE Journals & Magazine". In: *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15.3 (2007), pp. 379–386. DOI: 10.1109/TNSRE.2007.903919.

[184] Eliza Strickland. "Good-Bye, Wheelchair". en. In: *IEEE Spectrum* 49.1 (Jan. 2012), pp. 30–32. ISSN: 0018-9235. DOI: 10.1109/MSPEC.2012.6117830.

[185] Magdo Bortole et al. "The H2 Robotic Exoskeleton for Gait Rehabilitation after Stroke: Early Findings from a Clinical Study". In: *Journal of NeuroEngineering and Rehabilitation* 12.54 (2015). DOI: 10.1186/s12984-015-0048-y.

[186] Candy Tefertiller et al. "Initial Outcomes from a Multicenter Study Utilizing the Indego Powered Exoskeleton in Spinal Cord Injury". In: *Topics in Spinal Cord Injury Rehabilitation* 24.1 (2018), pp. 78–85. DOI: 10.1310/sci17-00014.

[187] A. Esquenazi et al. "The ReWalk Powered Exoskeleton to Restore Ambulatory Function to Individuals with Thoracic-Level Motor-Complete Spinal Cord Injury". In: *American Journal of Physical Medicine and Rehabilitation* 91.11 (2012), pp. 911–921.

[188] Shiqian Wang et al. “Design and Control of the MINDWALKER Exoskeleton”. en. In: *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 23.2 (Mar. 2015), pp. 277–286. ISSN: 1534-4320, 1558-0210. DOI: 10.1109/TNSRE.2014.2365697.

[189] Ryan J. Farris et al. “A Preliminary Assessment of Legged Mobility Provided by a Lower Limb Exoskeleton for Persons with Paraplegia”. eng. In: *IEEE transactions on neural systems and rehabilitation engineering: a publication of the IEEE Engineering in Medicine and Biology Society* 22.3 (May 2014), pp. 482–490. ISSN: 1558-0210. DOI: 10.1109/TNSRE.2013.2268320.

[190] Micheal Tucker et al. “Control Strategies for Active Lower Extremity Prosthetics and Orthotics: A Review | Journal of NeuroEngineering and Rehabilitation | Full Text”. In: *Journal of NeuroEngineering and Rehabilitation volume* 12.1 (2015), p. 1. DOI: 10.1186/1743-0003-12-1.

[191] Letian Wang et al. “Actively Controlled Lateral Gait Assistance in a Lower Limb Exoskeleton”. en. In: *2013 IEEE/RSJ International Conference on Intelligent Robots and Systems*. Tokyo: IEEE, Nov. 2013, pp. 965–970. ISBN: 978-1-4673-6358-7 978-1-4673-6357-0. DOI: 10.1109/IROS.2013.6696467.

[192] *Rex Bionics - Reimagining Rehabilitation.* https://www.rexbionics.com/.

[193] Thomas Gurriet et al. “Towards Restoring Locomotion for Paraplegics: Realizing Dynamically Stable Walking on Exoskeletons”. en. In: *2018 IEEE International Conference on Robotics and Automation (ICRA)*. Brisbane, QLD: IEEE, May 2018, pp. 2804–2811. ISBN: 978-1-5386-3081-5. DOI: 10.1109/ICRA.2018.8460647.

[194] Omar Harib et al. “Feedback Control of an Exoskeleton for Paraplegics: Toward Robustly Stable Hands-Free Dynamic Walking”. en. In: *arXiv:1802.08322 [cs]* (May 2018). arXiv: 1802.08322 [cs].

[195] B L Day et al. “Effect of Vision and Stance Width on Human Body Motion When Standing: Implications for Afferent Control of Lateral Sway.” en. In: *The Journal of Physiology* 469.1 (Sept. 1993), pp. 479–499. ISSN: 00223751. DOI: 10.1113/jphysiol.1993.sp019824.

[196] Vijaykumar Rajasekaran et al. “An Adaptive Control Strategy for Postural Stability Using a Wearable Robot”. en. In: *Robotics and Autonomous Systems*. Wearable Robotics 73 (Nov. 2015), pp. 16–23. ISSN: 0921-8890. DOI: 10.1016/j.robot.2014.11.014.

[197] Amber Emmens et al. “Improving the Standing Balance of Paraplegics through the Use of a Wearable Exoskeleton”. In: *2018 7th IEEE International Conference on Biomedical Robotics and Biomechatronics (Biorob)*. Aug. 2018, pp. 707–712. DOI: 10.1109/BIOROB.2018.8488066.

[198] Ildar Farkhatdinov et al. “Assisting Human Balance in Standing With a Robotic Exoskeleton”. In: *IEEE Robotics and Automation Letters* 4.2 (Apr. 2019), pp. 414–421. ISSN: 2377-3766, 2377-3774. DOI: 10.1109/LRA.2018.2890671.

[199] E. H. F. van Asseldonk et al. “Training Balance Recovery in People with Incomplete SCI Wearing a Wearable Exoskeleton”. en. In: *Wearable Robotics: Challenges and Trends*. Ed. by Maria Chiara Carrozza, Silvestro Micera, and José L. Pons. Biosystems & Biorobotics. Cham: Springer International Publishing, 2019, pp. 334–338. ISBN: 978-3-030-01887-0. DOI: 10.1007/978-3-030-01887-0_64.

[200] D. A. Winter. “Human Balance and Posture Control during Standing and Walking”. English. In: *Gait & Posture* 3.4 (Dec. 1995), pp. 193–214. ISSN: 0966-6362, 1879-2219. DOI: 10.1016/0966-6362(96)82849-9.

[201] Anne Shumway-Cook and Marjorie H. Woollacott. *Motor Control: Theory and Practical Applications*. en. Lippincott Williams & Wilkins, 2001. ISBN: 978-0-683-30643-9.

[202] Reuben T. Jessop, Christopher Horowicz, and Leland E. Dibble. “Motor Learning and Parkinson Disease: Refinement of Movement Velocity and Endpoint Excursion in a Limits of Stability Balance Task”. en. In: *Neurorehabilitation and Neural Repair* 20.4 (Dec. 2006), pp. 459–467. ISSN: 1545-9683, 1552-6844. DOI: 10.1177/1545968306287107.

[203] Aziz K. Alfeeli et al. “Postural Stability and Balance Training Program in Hemiparetic Stroke Patients”. en. In: *Macedonian Journal of Medical Sciences* 6.3 (Sept. 2013), pp. 251–254. ISSN: 1857-5773, 1857-5749. DOI: 10.3889/mjms.1857-5773.2013.0303.

[204] Marc O. Ernst and Martin S. Banks. “Humans Integrate Visual and Haptic Information in a Statistically Optimal Fashion”. en. In: *Nature* 415.6870 (Jan. 2002), pp. 429–433. ISSN: 1476-4687. DOI: 10.1038/415429a.

[205] J. L. Patton, F. A. Mussa-Ivaldi, and W. Z. Rymer. “Altering Movement Patterns in Healthy and Brain-Injured Subjects via Custom Designed Robotic Forces”. In: *2001 Conference Proceedings of the 23rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society*. Vol. 2. 2001, 1356–1359 vol.2. DOI: 10.1109/IEMBS.2001.1020448.

[206] Sharon Israely, Gerry Leisman, and Eli Carmeli. “Improvement in Hand Trajectory of Reaching Movements by Error-Augmentation”. eng. In: *Advances in Experimental Medicine and Biology* 1070 (2018), pp. 71–84. ISSN: 0065-2598. DOI: 10.1007/5584_2018_151.

[207] Sharon Israely and Eli Carmeli. “Error Augmentation as a Possible Technique for Improving Upper Extremity Motor Performance after a Stroke - a Systematic Review”. eng. In: *Topics in Stroke Rehabilitation* 23.2 (Apr. 2016), pp. 116–125. ISSN: 1074-9357. DOI: 10.1179/1945511915Y.0000000007.

[208] Laura Marchal-Crespo et al. “Optimizing Learning of a Locomotor Task: Amplifying Errors as Needed”. eng. In: *Conference proceedings: ... Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Annual Conference* 2014 (2014), pp. 5304–5307. ISSN: 1557-170X. DOI: 10.1109/EMBC.2014.6944823.

[209] Robert J. van Beers. “Motor Learning Is Optimally Tuned to the Properties of Motor Noise”. ENG. In: *Neuron* 63.3 (Aug. 2009), pp. 406–417. ISSN: 1097-4199. DOI: 10.1016/j.neuron.2009.06.025.

[210] John W. Krakauer. “Motor Learning and Consolidation: The Case of Visuomotor Rotation”. In: *Advances in experimental medicine and biology* 629 (2009), pp. 405–421. ISSN: 0065-2598. DOI: 10.1007/978-0-387-77064-2_21.

[211] Elaine Kearney et al. “Augmented Visual Feedback-Aided Interventions for Motor Rehabilitation in Parkinson’s Disease: A Systematic Review”. eng. In: *Disability and Rehabilitation* (Jan. 2018), pp. 1–17. ISSN: 1464-5165. DOI: 10.1080/09638288.2017.1419292.

[212] Moore S and Woollacott MN. “The Use of Biofeedback Devices to Improve Postural Stability”. EN. In: *Physical Therapy Practice* (1993), pp. 1–19.

[213] Günther Knoblich and Tilo T. J. Kircher. “Deceiving Oneself About Being in Control: Conscious Detection of Changes in Visuomotor Coupling”. In: *Journal of Experimental Psychology: Human Perception and Performance* 30.4 (2004), pp. 657–666. ISSN: 1939-1277(Electronic),0096-1523(Print). DOI: 10.1037/0096-1523.30.4.657.

[214] Niclas Braun et al. “The Senses of Agency and Ownership: A Review”. In: *Frontiers in Psychology* 9 (Apr. 2018). ISSN: 1664-1078. DOI: 10.3389/fpsyg.2018.00535.

[215] *THERA-Trainer: THERA-Trainer Coro.* EN. https://www.thera-trainer.de/en/thera-trainer-products/standing-balancing/thera-trainer-coro/. Feb. 2019.

[216] R Core Team. “R: A Language and Environment for Statistical Computing”. en. In: *R Foundation for Statistical Computing* (2014).

[217] Cyril R. Pernet, Rand R. Wilcox, and Guillaume A. Rousselet. “Robust Correlation Analyses: False Positive and Power Validation Using a New Open Source Matlab Toolbox”. English. In: *Frontiers in Psychology* 3 (2013). ISSN: 1664-1078. DOI: 10.3389/fpsyg.2012.00606.

[218] P. Fourneret et al. “Self-Monitoring in Schizophrenia Revisited”. eng. In: *Neuroreport* 12.6 (May 2001), pp. 1203–1208. ISSN: 0959-4965. DOI: 10.1097/00001756-200105080-00030.

[219] Indrit Sinanaj, Yann Cojan, and Patrik Vuilleumier. "Inter-Individual Variability in Metacognitive Ability for Visuomotor Performance and Underlying Brain Structures". In: *Consciousness and Cognition* 36 (Nov. 2015), pp. 327–337. ISSN: 1053-8100. DOI: 10.1016/j.concog.2015.07.012.

[220] Rin Minohara et al. "Strength of Intentional Effort Enhances the Sense of Agency". eng. In: *Frontiers in Psychology* 7 (2016), p. 1165. ISSN: 1664-1078. DOI: 10.3389/fpsyg.2016.01165.

[221] Jelle Demanet et al. "Power to the Will: How Exerting Physical Effort Boosts the Sense of Agency". eng. In: *Cognition* 129.3 (Dec. 2013), pp. 574–578. ISSN: 1873-7838. DOI: 10.1016/j.cognition.2013.08.020.

[222] José Raúl Naranjo and Stefan Schmidt. "Is It Me or Not Me? Modulation of Perceptual-Motor Awareness and Visuomotor Performance by Mindfulness Meditation". In: *BMC Neuroscience* 13 (July 2012), p. 88. ISSN: 1471-2202. DOI: 10.1186/1471-2202-13-88.

[223] Nicholas Hon, Jia-Hou Poh, and Chun-Siong Soon. "Preoccupied Minds Feel Less Control: Sense of Agency Is Modulated by Cognitive Load". In: *Consciousness and Cognition* 22.2 (June 2013), pp. 556–561. ISSN: 1053-8100. DOI: 10.1016/j.concog.2013.03.004.

[224] Günther Knoblich. "Self-Recognition: Body and Action". English. In: *Trends in Cognitive Sciences* 6.11 (Nov. 2002), pp. 447–449. ISSN: 1364-6613, 1879-307X. DOI: 10.1016/S1364-6613(02)01995-2.

[225] Jenifer Miehlbradt et al. "Data-Driven Body–Machine Interface for the Accurate Control of Drones". en. In: *Proceedings of the National Academy of Sciences* 115.31 (July 2018), pp. 7913–7918. ISSN: 0027-8424, 1091-6490. DOI: 10.1073/pnas.1718648115.

[226] Laura Marchal-Crespo and Robert Riener. "Chapter 16 - Robot-Assisted Gait Training". en. In: *Rehabilitation Robotics*. Ed. by Roberto Colombo and Vittorio Sanguineti. Academic Press, Jan. 2018, pp. 227–240. ISBN: 978-0-12-811995-2. DOI: 10.1016/B978-0-12-811995-2.00016-3.

[227] Danielle E. Levac, Meghan E. Huber, and Dagmar Sternad. "Learning and Transfer of Complex Motor Skills in Virtual Reality: A Perspective Review". en. In: *Journal of NeuroEngineering and Rehabilitation* 16.1 (Dec. 2019). ISSN: 1743-0003. DOI: 10.1186/s12984-019-0587-8.

[228] B. R. Brewer, R. Klatzky, and Y. Matsuoka. "Initial Therapeutic Results of Visual Feedback Manipulation in Robotic Rehabilitation". In: *2006 International Workshop on Virtual Rehabilitation*. 2006, pp. 160–166. DOI: 10.1109/IWVR.2006.1707546.

[229] Jemina Fasola et al. “Error Augmentation Improves Visuomotor Adaptation during a Full-Body Balance Task”. eng. In: *Conference proceedings: ... Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Annual Conference* 2019 (July 2019), pp. 1529–1533. ISSN: 1557-170X. DOI: 10.1109/EMBC.2019.8857523.

[230] Daniel M. Wegner. “The Mind’s Best Trick: How We Experience Conscious Will”. eng. In: *Trends in Cognitive Sciences* 7.2 (Feb. 2003), pp. 65–69. ISSN: 1879-307X.

[231] Terence D. Sanger. “Failure of Motor Learning for Large Initial Errors”. eng. In: *Neural Computation* 16.9 (Sept. 2004), pp. 1873–1886. ISSN: 0899-7667. DOI: 10.1162/0899766041336431.

[232] Robert Riener and Linda J. Seward. “Cybathlon 2016”. In: *2014 IEEE International Conference on Systems, Man, and Cybernetics (SMC)*. Oct. 2014, pp. 2792–2794. DOI: 10.1109/SMC.2014.6974351.

[233] Peter G. Adamczyk, Steven H. Collins, and Arthur D. Kuo. “The Advantages of a Rolling Foot in Human Walking”. eng. In: *The Journal of Experimental Biology* 209.Pt 20 (Oct. 2006), pp. 3953–3963. ISSN: 0022-0949. DOI: 10.1242/jeb.02455.

[234] Filiep Debaere et al. “Internal vs External Generation of Movements: Differential Neural Pathways Involved in Bimanual Coordination Performed in the Presence or Absence of Augmented Visual Feedback”. In: *NeuroImage* 19.3 (July 2003), pp. 764–776. ISSN: 1053-8119. DOI: 10.1016/S1053-8119(03)00148-4.

[235] J. Olivier et al. “Impact of Ankle Locking on Gait Implications for the Design of Hip and Knee Exoskeletons”. In: *2015 IEEE International Conference on Rehabilitation Robotics (ICORR)*. Aug. 2015, pp. 618–622. DOI: 10.1109/ICORR.2015.7281269.

[236] Johannes Dichgans and Thomas Brandt. “Visual-Vestibular Interaction: Effects on Self-Motion Perception and Postural Control”. en. In: *Perception*. Ed. by S. M. Anstis et al. Handbook of Sensory Physiology. Berlin, Heidelberg: Springer Berlin Heidelberg, 1978, pp. 755–804. ISBN: 978-3-642-46354-9. DOI: 10.1007/978-3-642-46354-9_25.

[237] M. C. Hunter and M. A. Hoffman. “Postural Control: Visual and Cognitive Manipulations”. eng. In: *Gait & Posture* 13.1 (Feb. 2001), pp. 41–48. ISSN: 0966-6362.

[238] R Fitzpatrick and D I McCloskey. “Proprioceptive, Visual and Vestibular Thresholds for the Perception of Sway during Standing in Humans.” en. In: *The Journal of Physiology* 478.1 (July 1994), pp. 173–186. ISSN: 00223751. DOI: 10.1113/jphysiol.1994.sp020240.

[239] R. Johansson, M. Magnusson, and P. A. Fransson. “Galvanic Vestibular Stimulation for Analysis of Postural Adaptation and Stability”. eng. In: *IEEE transactions on bio-medical engineering* 42.3 (Mar. 1995), pp. 282–292. ISSN: 0018-9294. DOI: 10.1109/10.364515.

[240] F. B. Horak and L. M. Nashner. “Central Programming of Postural Movements: Adaptation to Altered Support-Surface Configurations”. eng. In: *Journal of Neurophysiology* 55.6 (June 1986), pp. 1369–1381. ISSN: 0022-3077. DOI: 10.1152/jn.1986.55.6.1369.

[241] B. R. Bloem et al. “Is Lower Leg Proprioception Essential for Triggering Human Automatic Postural Responses?” eng. In: *Experimental Brain Research* 130.3 (Feb. 2000), pp. 375–391. ISSN: 0014-4819. DOI: 10.1007/s002219900259.

[242] A. D. Kuo et al. “Effect of Altered Sensory Conditions on Multivariate Descriptors of Human Postural Sway”. eng. In: *Experimental Brain Research* 122.2 (Sept. 1998), pp. 185–195. ISSN: 0014-4819.

[243] S. Gurses, R. V. Kenyon, and E. A. Keshner. “Examination of Time-Varying Kinematic Responses to Support Surface Disturbances”. In: *Biomedical Signal Processing and Control*. Biomedical Signal Processing(Extended Selected Papers from the 7th IFAC Symposium on Modelling and Control in Biomedical Systems(MCBMS’09)) 6.1 (Jan. 2011), pp. 85–93. ISSN: 1746-8094. DOI: 10.1016/j.bspc.2010.06.002.

[244] Arthur D. Kuo. “An Optimal State Estimation Model of Sensory Integration in Human Postural Balance”. eng. In: *Journal of Neural Engineering* 2.3 (Sept. 2005), S235–249. ISSN: 1741-2560. DOI: 10.1088/1741-2560/2/3/S07.

[245] Glen M. Blenkinsop, Matthew T. G. Pain, and Michael J. Hiley. “Balance Control Strategies during Perturbed and Unperturbed Balance in Standing and Handstand”. In: *Royal Society Open Science* 4.7 (July 2017). ISSN: 2054-5703. DOI: 10.1098/rsos.161018.

[246] Kathrin Freyler et al. “Reactive Balance Control in Response to Perturbation in Unilateral Stance: Interaction Effects of Direction, Displacement and Velocity on Compensatory Neuromuscular and Kinematic Responses”. In: *PLoS ONE* 10.12 (Dec. 2015). ISSN: 1932-6203. DOI: 10.1371/journal.pone.0144529.

[247] E. Zemková et al. “Postural and Trunk Responses to Unexpected Perturbations Depend on the Velocity and Direction of Platform Motion”. eng. In: *Physiological Research* 65.5 (Nov. 2016), pp. 769–776. ISSN: 1802-9973.

[248] L. Yardley et al. “Effect of Articulatory and Mental Tasks on Postural Control”. eng. In: *Neuroreport* 10.2 (Feb. 1999), pp. 215–219. ISSN: 0959-4965.

[249] Gerhard Andersson et al. “Effect of Cognitive Load on Postural Control”. In: *Brain Research Bulletin* 58.1 (May 2002), pp. 135–139. ISSN: 0361-9230. DOI: 10.1016/S0361-9230(02)00770-0.

[250] Mark W. Rogers and Marie-Laure Mille. “Balance Perturbations”. eng. In: *Handbook of Clinical Neurology* 159 (2018), pp. 85–105. ISSN: 0072-9752. DOI: 10.1016/B978-0-444-63916-5.00005-7.

[251] Jemina Fasola et al. “Balance Control Strategies during Standing in a Locked-Ankle Passive Exoskeleton”. In: *2019 IEEE 16th International Conference on Rehabilitation Robotics (ICORR)*. June 2019, pp. 593–598. DOI: 10.1109/ICORR.2019.8779500.

[252] Romain Baud et al. “Bio-Inspired Standing Balance Controller for a Full-Mobilization Exoskeleton”. In: *2019 IEEE 16th International Conference on Rehabilitation Robotics (ICORR)*. June 2019, pp. 849–854. DOI: 10.1109/ICORR.2019.8779440.

[253] Lionel Billet, Michaël Delbaere, and Tommy Pinel. “Compact Mechatronic Drive for Robotic Applications”. en-US. In: *Robotics Tomorrow* (May 2019).

[254] Pengfei Gui, Liqiong Tang, and Subhas Mukhopadhyay. “MEMS Based IMU for Tilting Measurement: Comparison of Complementary and Kalman Filter Based Data Fusion”. In: *2015 IEEE 10th Conference on Industrial Electronics and Applications (ICIEA)*. June 2015, pp. 2004–2009. DOI: 10.1109/ICIEA.2015.7334442.

[255] Y. Fang et al. “Anthropometric and Biomechanical Characteristics of Body Segments in Persons with Spinal Cord Injury”. eng. In: *Journal of Biomechanics* 55 (Nov. 2017), pp. 11–17. ISSN: 1873-2380. DOI: 10.1016/j.jbiomech.2017.01.036.

[256] Swissethics. *Lignes Directrices Dispositifs d’assistance Technique – Dispositifs Médicaux*.

[257] Mariella Pazzaglia and Marco Molinari. “The Embodiment of Assistive Devices-from Wheelchair to Exoskeleton”. eng. In: *Physics of Life Reviews* 16 (Mar. 2016), pp. 163–175. ISSN: 1873-1457. DOI: 10.1016/j.plrev.2015.11.006.

[258] Philipp Beckerle et al. “Feel-Good Robotics: Requirements on Touch for Embodiment in Assistive Robotics”. en. In: *Frontiers in Neurorobotics* 12 (Dec. 2018). ISSN: 1662-5218. DOI: 10.3389/fnbot.2018.00084.

[259] Carolee Winstein, Alan M. Wing, and Jill Whitall. “Motor Control and Learning Principles for Rehabilitation of Upper Limb Movements after Brain Injury”. In: *Motor Control and Learning Principles for Rehabilitation of Upper Limb Movements after Brain Injury*. 2nd Edition. Vol. 9. Handbook of Neuropsychology. J. Gaafman and L.H. Robertson, 2003.

[260] Domingo and Ferris. “Effects of Physical Guidance on Short-Term Learning of Walking on a Narrow Beam.” eng. In: *Gait & posture* 30 (2009), pp. 464–468.

[261] David J. Reinkensmeyer and Sarah J. Housman. “"If I Can’t Do It Once, Why Do It a Hundred Times?": Connecting Volition to Movement Success in a Virtual Environment Motivates People to Exercise the Arm after Stroke”. In: *2007 Virtual Rehabilitation*. Sept. 2007, pp. 44–48. DOI: 10.1109/ICVR.2007.4362128.

[262] Domen Novak et al. “Increasing Motivation in Robot-Aided Arm Rehabilitation with Competitive and Cooperative Gameplay”. In: *Journal of NeuroEngineering and Rehabilitation* 11.1 (Apr. 2014), p. 64. ISSN: 1743-0003. DOI: 10.1186/1743-0003-11-64.

[263] Janet H Carr and Roberta B Shepherd. “A Motor Learning Model for Stroke Rehabilitation”. en. In: *Physiotherapy* 75.7 (July 1989), pp. 372–380. ISSN: 0031-9406. DOI: 10.1016/S0031-9406(10)62588-6.

[264] F. B Horak. “Assumptions Underlying Motor Control for Neurologic Rehabilitation”. In: *Contemporary Management of Motor Control Problemsorary Management of Motor Control Problems*. Vol. VA: Foundations for Physical Therapy. Alexandria: Lister, M., editor, 1991, pp. 11–27.

[265] John W. Krakauer. “Motor Learning: Its Relevance to Stroke Recovery and Neurorehabilitation”. eng. In: *Current Opinion in Neurology* 19.1 (Feb. 2006), pp. 84–90. ISSN: 1350-7540. DOI: 10.1097/01.wco.0000200544.29915.cc.

[266] Volker Dietz and Nick Ward. *Oxford Textbook of Neurorehabilitation*. en. Oxford University Press, 2015. ISBN: 978-0-19-967371-1.

[267] Richard A. Schmidt and Timothy Donald Lee. *Motor Control and Learning: A Behavioral Emphasis*. en. Human Kinetics, 1999. ISBN: 978-0-88011-484-4.

[268] Gert Kwakkel, Boudewijn J. Kollen, and Hermano I. Krebs. “Effects of Robot-Assisted Therapy on Upper Limb Recovery After Stroke: A Systematic Review”. en. In: *Neurorehabilitation and Neural Repair* 22.2 (Mar. 2008), pp. 111–121. ISSN: 1545-9683. DOI: 10.1177/1545968307305457.

[269] Maureen K. Holden. “Virtual Environments for Motor Rehabilitation: Review”. eng. In: *Cyberpsychology & Behavior: The Impact of the Internet, Multimedia and Virtual Reality on Behavior and Society* 8.3 (June 2005), 187–211, discussion 212–219. ISSN: 1094-9313. DOI: 10.1089/cpb.2005.8.187.

[270] Heidi Sveistrup. “Motor Rehabilitation Using Virtual Reality”. In: *Journal of NeuroEngineering and Rehabilitation* 1.1 (Dec. 2004), p. 10. ISSN: 1743-0003. DOI: 10.1186/1743-0003-1-10.

[271] E Daprati et al. “Looking for the Agent: An Investigation into Consciousness of Action and Self-Consciousness in Schizophrenic Patients”. en. In: *Cognition* 65.1 (Dec. 1997), pp. 71–86. ISSN: 0010-0277. DOI: 10.1016/S0010-0277(97)00039-5.

[272] Esther van den Bos and Marc Jeannerod. “Sense of Body and Sense of Action Both Contribute to Self-Recognition”. eng. In: *Cognition* 85.2 (Sept. 2002), pp. 177–187. ISSN: 0010-0277. DOI: 10.1016/s0010-0277(02)00100-2.

[273] Dale Purves et al. “Motor Control Centers in the Brainstem: Upper Motor Neurons That Maintain Balance and Posture”. en. In: *Neuroscience. 2nd edition* (2001).

[274] Umberto Castiello. “The Neuroscience of Grasping”. eng. In: *Nature Reviews. Neuroscience* 6.9 (Sept. 2005), pp. 726–736. ISSN: 1471-003X. DOI: 10.1038/nrn1744.

[275] Marjorie Woollacott and Anne Shumway-Cook. “Attention and the Control of Posture and Gait: A Review of an Emerging Area of Research”. eng. In: *Gait & Posture* 16.1 (Aug. 2002), pp. 1–14. ISSN: 0966-6362. DOI: 10.1016/s0966-6362(01)00156-4.

[276] S. Grillner and P. Wallén. “Central Pattern Generators for Locomotion, with Special Reference to Vertebrates”. eng. In: *Annual Review of Neuroscience* 8 (1985), pp. 233–261. ISSN: 0147-006X. DOI: 10.1146/annurev.ne.08.030185.001313.

[277] U. Castiello, Y. Paulignan, and M. Jeannerod. “TEMPORAL DISSOCIATION OF MOTOR RESPONSES AND SUBJECTIVE AWARENESS: A STUDY IN NORMAL SUBJECTS”. en. In: *Brain* 114.6 (1991), pp. 2639–2655. ISSN: 0006-8950, 1460-2156. DOI: 10.1093/brain/114.6.2639.

[278] James W. Moore. “What Is the Sense of Agency and Why Does It Matter?” English. In: *Frontiers in Psychology* 7 (2016). ISSN: 1664-1078. DOI: 10.3389/fpsyg.2016.01272.

[279] John Mirowsky. “Age and the Sense of Control”. In: *Social Psychology Quarterly* 58.1 (1995), pp. 31–43. ISSN: 0190-2725. DOI: 10.2307/2787141.

[280] Amit Goffer. “Enhanced Safety of Gait in Powered Exoskeletons”. en. In: (), p. 2.

[281] Solaiman Shokur et al. “Assimilation of Virtual Legs and Perception of Floor Texture by Complete Paraplegic Patients Receiving Artificial Tactile Feedback”. en. In: *Scientific Reports* 6.1 (Sept. 2016), pp. 1–14. ISSN: 2045-2322. DOI: 10.1038/srep32293.

[282] Angelo Maravita and Atsushi Iriki. “Tools for the Body (Schema)”. en. In: *Trends in Cognitive Sciences* 8.2 (Feb. 2004), pp. 79–86. ISSN: 1364-6613. DOI: 10.1016/j.tics.2003.12.008.

[283] Frédérique de Vignemont. “Embodiment, Ownership and Disownership”. eng. In: *Consciousness and Cognition* 20.1 (Mar. 2011), pp. 82–93. ISSN: 1090-2376. DOI: 10.1016/j.concog.2010.09.004.

[284] Viviane de Souza Pinho Costa et al. “Social Representations of the Wheelchair for People with Spinal Cord Injury”. en. In: *Revista Latino-Americana de Enfermagem* 18.4 (Aug. 2010), pp. 755–762. ISSN: 0104-1169. DOI: 10.1590/S0104-11692010000400014.

[285] Franklyn N. Arnhoff and Marie C. Mehl. “BODY IMAGE DETERIORATION IN PARAPLEGIA”. en-US. In: *The Journal of Nervous and Mental Disease* 137.1 (July 1963), pp. 88–92. ISSN: 0022-3018.

[286] Mariella Pazzaglia et al. “A Functionally Relevant Tool for the Body Following Spinal Cord Injury”. en. In: *PLOS ONE* 8.3 (Mar. 2013), e58312. ISSN: 1932-6203. DOI: 10.1371/journal.pone.0058312.

[287] Roger Newport and Catherine Preston. “Pulling the Finger off Disrupts Agency, Embodiment and Peripersonal Space:” en. In: *Perception* (Jan. 2010). DOI: 10.1068/p6742.

[288] Manos Tsakiris, Gita Prabhu, and Patrick Haggard. “Having a Body versus Moving Your Body: How Agency Structures Body-Ownership”. en. In: *Consciousness and Cognition* 15.2 (June 2006), pp. 423–432. ISSN: 1053-8100. DOI: 10.1016/j.concog.2005.09.004.

[289] Tamar R. Makin, Frederique de Vignemont, and A. Aldo Faisal. “Neurocognitive Barriers to the Embodiment of Technology”. en. In: *Nature Biomedical Engineering* 1.1 (Jan. 2017). ISSN: 2157-846X. DOI: 10.1038/s41551-016-0014.

[290] Maria Niedernhuber, Damiano G. Barone, and Bigna Lenggenhager. “Prostheses as Extensions of the Body: Progress and Challenges”. eng. In: *Neuroscience and Biobehavioral Reviews* 92 (Sept. 2018), pp. 1–6. ISSN: 1873-7528. DOI: 10.1016/j.neubiorev.2018.04.020.

A Appendix

A.1 Learning to walk with the exoskeleton TWIICE One

A.1.1 Introduction

Walking with the exoskeleton TWIICE One requires the use of crutches to manage the orientation of the whole body in space, but also to perform weight transfer from one leg to the other during walking. Indeed, in order to start a step, most of the weight should be on the stance leg. Since TWIICE has no degree of freedom in the frontal plane, this weight transfer can be done solely by slightly tilting the upper body on the stance leg and mostly by applying a lateral force with the crutches on the ground. Therefore, learning to transfer properly the weight on the stance leg is crucial for walking with the exoskeleton TWIICE One. Since the gait trajectories are always following the same gait pattern, the sagittal orientation of the whole body matters for the quality of gait. In fact, if the body is tilted forward, the trajectories may not be able to pursue their full amplitude (i.e. the foot will hit the ground before the end of the trajectory), leading to shorter steps. On the opposite, if the body is tilted backward, triggering a step will give the sensation of falling backward.

Consequently, a novice exoskeleton user has to learn how to manage the posture of the whole system while walking. Since leg proprioception is impaired in SCI users, they mainly rely on visual and vestibular systems to orient their body in space, as well as arms proprioceptive feedback due to the use of the crutches. One of our pilots reported that standing in the exoskeleton procured the same balance sensations than sitting on an inflatable ball.
During training, the trainer can provide force feedback through the handle of the exoskeleton situated on the pelvis structure. A lingering question in motor learning with force feedback

Table A.1 – WIITE user description

User Profile	
Gender	Male
Years post injury	6
Lesion level	T7
AIS scale	A
Weight [kg]	70
Height [cm]	171
Knee ROM [deg]	10–?+
Hip ROM [deg]	-30–?+
Spasticity (Hips and knees)	MAS 3
Bone mineral density (Hips and knees) [gcm^{-2}]	0.65

is what type of training modalities such as guidance or error augmentation is more efficient. Guidance prevents the user from making large postural errors. On the other hand, error augmentation amplified the sensory feedback errors, forcing the user to overcorrect for the wrong postural orientation. Error augmentation thus allows experiencing larger errors.

In a case study, we investigated the effect of guidance and error augmentation on learning to walk with the exoskeleton TWIICE One and with a complete SCI, novice user. We expected that error augmentation would be more beneficial than guidance since it has been demonstrated that guidance can lead to feedback dependency [92, 93].

A.1.2 Test-pilot

The target user was a 26-year old male who was 6 years post-injury. His lesion was at level T7 and functionally complete (AIS A), see Table A.1. The screening was done in conjunction with the medical doctor specialized in physical medicine and rehabilitation who had followed the case since the injury.
The user exhibited important knee contractures that resorbed to 10 degrees in flexion after about 10 min of stretching and when applying 30Nm of torque. He provided informed consent after explanation of the possible risks associated with the use of exoskeletons as well as to those associated with the developing nature of the device. Tests were conducted pursuant to the Guidelines on assistive devices design by the Swiss Ethics Committees on research involving humans [256].

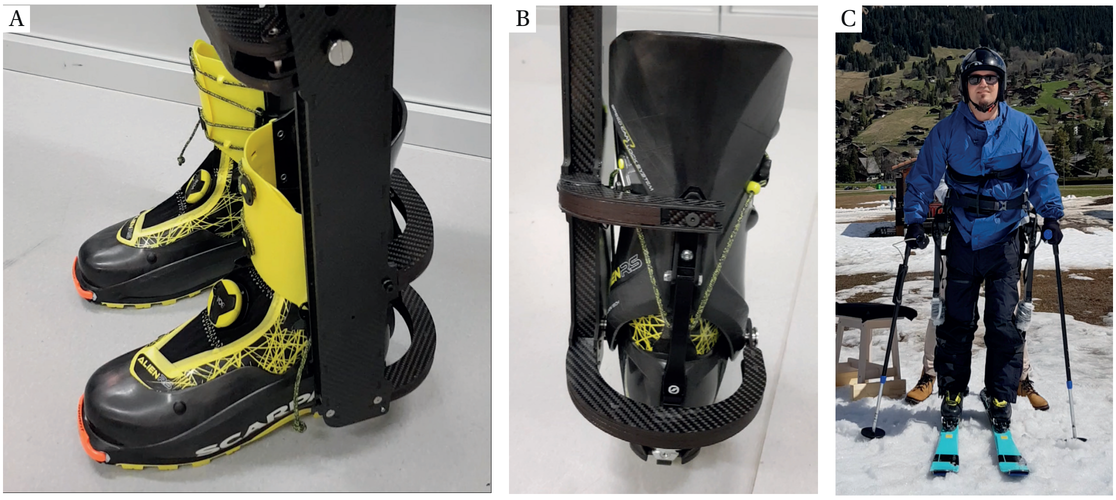

Figure A.1 – Overview of WIITE, an exoskeleton for ski-touring. **(A)-(B)** show the ski-touring boots and the shank segment. **(C)** presents an overview of the exoskeleton WIITE.

A.1.3 Materials

The ultimate goal of the pilot was to use the exoskeleton to perform ski-touring. In turn, the exoskeleton TWIICE One was modified: the foot was removed and shank modules were modified to be attached to ski-touring boots (Alien RS by Scarpa, Boulder, CO, USA). This new version of the exoskeleton is called WIITE, see Figure A.1. In order to practice overground walking, curved foot soles were added below the ski boots to cope with the lack of mobility of the ankle. This enabled the practice on flat ground and inside when weather conditions did not allow for ski-touring.

A.1.4 Experimental design

Protocol

We tested the effect of error augmentation (EA) and guidance (A) over two sessions of four training blocks (2xEA and 2xA). The four training blocks were randomized for each session and were followed by an evaluation phase. The two sessions started with a baseline representing a 10-meter walk. Each training block was a 20-meter walk, while the evaluation was a 10-meter walk. During the evaluation, the trainer did not provide any feedback and would intervene only in case of a risk of fall. After each evaluation and thus, before starting a new training block the pilot had a 5-minute break.

Figure A.2 – Error augmentation versus guidance. (A)Error augmentation. While the user is using its balance toward his right side, the trainer amplifies the postural error by applying a force to the right. (B) Guidance. The user is tilted too much to the left, the trainer corrects his postural orientation by applying a force to their right direction.

Learning strategies

The trainer behind the pilot was always monitoring the postural orientation of the pilot. He was evaluating if the postural orientation was adequate with respect to the gait phases. To promote learning of the correct posture, the trainer could provide force feedback through the pelvic structure of the exoskeleton in two different ways: postural error amplification or postural guidance (Figure A.2). For the error augmentation strategy, the trainer was amplifying the postural error by applying a force in the same direction as the incorrect postural orientation. For guidance, the trainer was minimizing the error by guiding the pilot to the correct posture. If the user was too much tilted forward, the trainer would pull him back to center him for example.

Performance metric

Since the walking trajectories and their velocity are predefined and constant during walking (see Figure A.6, the main component that can influence gait velocity is the duration of the double support phase (DSP), T_{DSP}, when the exoskeleton is not moving. During this phase, the pilot is in the double stance phase and has to position the crutches to be stable enough to

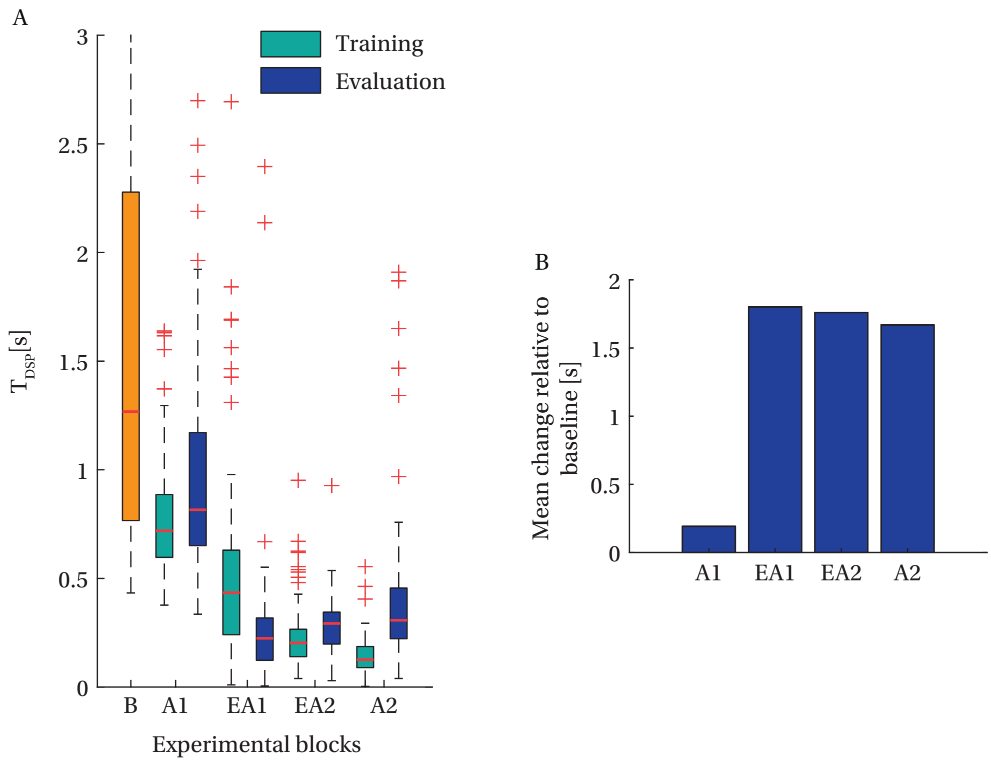

Figure A.3 – Results session 1. **(A)** DSP duration of each training and evaluation block. **(B)** Change in DSP duration with respect to baseline.

trigger the next steps. A reduction of the DSP duration reflects faster weight transfer from one to the other and a better grasp of the crutches placement to ensure his stability. Therefore, the average change in DSP duration in between block is the main metric to assess learning.

A.1.5 Results

The results of the first session are presented in Figure A.3. The pilot started with a mean DSP duration of 2.11s. The first training block was with the guidance strategy (A1). We can see that training with guidance reduced the DSP duration and inhibited long DSP associated with large postural errors. The average DSP duration increased slightly during the evaluation (A1.Evaluation) with a mean change relative to baseline was of 0.19s. The two next training blocks were with error augmentation. We observed that after the first training block with error augmentation (EA1.Training), the evaluation phase (EA1.Evaluation) presented a significantly lower mean DSP duration (0.3 s) and a mean change relative to the baseline of 1.8s. The next evaluation phases mostly showed a constant performance.

The second session happened 3 weeks later (Figure A.3). The mean DSP duration during

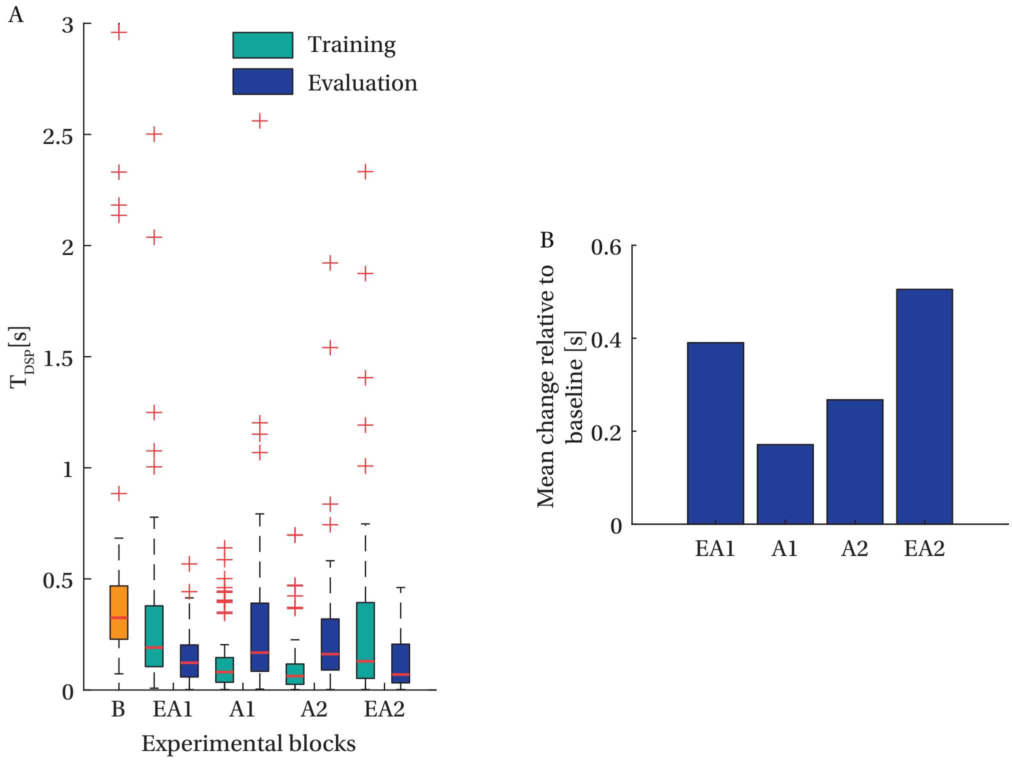

Figure A.4 – Results session 2.**(A)** DSP duration of each training and evaluation block. **(B)** Change in DSP duration with respect to baseline.

baseline was of 0.6 s. In this session, the pilot started to train with error augmentation. We observed a significant reduction of DSP duration (0.2 s) in the evaluation phase (EA1.Evaluation). The mean change relative to baseline was thus 0.4s. The next two blocks were with guidance. Guidance tends to shorten the DSP duration. However, the DSP duration tends to increase after training with guidance.

A.1.6 Discussion

The preliminary analysis of this pilot study suggests that amplifying postural errors could enhance learning to pilot the exoskeleton TWIICE one. Indeed the DSP duration was shorter after training with postural error augmentation. In opposite, training with guidance led to poorer performance during the evaluations. The DSP duration was low during guidance training suggesting a stable walk, but also no opportunity to experience catastrophic errors that are necessary to build the correct motor command [94]. This may also suggest that guidance training leads to a dependency of the feedback, and thus once the feedback was removed the performance decreased. This is in line with previous studies on motor learning with guidance.

The main limitation of this protocol is that error detection is based on the subjective evaluation of the trainer, as well as the produced forces. Although he tried to be rigorous, it necessarily introduces a bias.
This pilot study needs, however, a more strict experimental design and data analysis, as well as a larger cohort, but these preliminary data encourage us to pursue testing postural error augmentation on our future, novice exoskeleton pilots.

A.2 Walking performance evolution

This study uses data collected with all versions of the exoskeleton TWIICE, as it evolved throughout the years. The general architecture remained the same, with mobility at the hips and knees using electrical motors, and all other DoF locked. Several mechanical improvements were made, in particular during the first months. Those enhancements are presented and their effect is discussed in this study.

A.2.1 Materials and methods

Test-pilot: complete SCI user

The test-pilot for this study was as described in section 6.3.3. She provided informed consent after explanation of the possible risks associated with the use of exoskeletons as well as to those associated to the developing nature of the device. Tests were conducted pursuant to the Guidelines on assistive devices design by the Swiss Ethics Committees on research involving humans [256].

Protocol

This particular case study is a retrospective of all training sessions during which the pilot acquired skills on how to use the devices for different activities, as well as over the course of the improvements and development process of the exoskeleton. No strict protocol was followed as the purpose of the tests was to train the pilot and to briefly validate design decisions with short tests in an iterative manner. This provided insights on the design parameters and about where design effort was required.

A.2.2 Data collection and processing

All data were collected periodically by the exoskeleton's embedded system, measuring sensors' state at a 1 kHz sampling rate and stored on internal memory. There were retrieved and processed offline processing using Matlab (Mathworks, Natick, MA, USA). For each individual step, step duration T_S was measured between each press of the trigger. Double support phase duration T_{DSP} was measured as the time when the exoskeleton was not moving and T_{SSP} was the time during which the exoskeleton was producing movement with the legs. Step length A_S was measured based on a direct kinematic model of the device and using articular angular position measured just before movement trigger. The step velocity v_S was calculated using step length and duration. Training occurred during 121 sessions between July 2016 and July 2019. Each session was cut in intervals of the same activity called laps so that only active time was retained. For each walking lap, the average walking velocity was calculated. The maximum among all laps of the average walking velocity was kept and reported on Fig. A.5. The total distance was calculated as the sum of all step length, and the walking duration only considered active walking time. For both these parameters, a weekly rate was computed and averaged using a 1-week gaussian filter. The session maximum velocity was interpolated linearly between sessions at intervals of 5 hours. It was then derived using a gaussian differentiator with a cut-off frequency corresponding to a one week period. This yielded the walking velocity improvement rate, as reported on Fig. A.5.

A.2.3 Results

Over three years, a total of 30'000 a total of steps were taken for a total distance of 9271m and a total walking time of 13h 7min. The max velocity reached by our pilot is 0.44m/s. The Figure A.5 displays the evolution of the walking velocity based on the walking distance and duration over time. The major events in the developments are indicated with vertical lines, highlighting possible reasons for the evolution of velocity over time. We can observed that the walking velocity increased slowly over time. The walking improvement rate allows us to see which periods were particularly auspicious for gait improvement (positive rate). One should note that negative rate do not obligatory mean a loss of control from the user side, but could also reflects events where slower pace were required, such as demo.
We can observe the the walking velocity drastically increased during the first phase of development due probably to user learning how to position the crutches, but mainly to technical progresses such as new walking trajectories (Figure A.6 and the addition of a thoracic belt. Because of the level of the lesion, the trunk control of the pilot was limited. The pelvic belt was not sufficient leading to a slouching posture and high loads on the forearm. The thoracic belt greatly increased the trunk stability and alleviate the load on the arms. The trajectory modification (v2) happened after 6 training sessions. It decreased step duration thanks to smoother accelerations. This produces an increase of velocity without a major increase of the step length. The trajectory v3 and v4 increased the step length which also contributed to an important increase of velocity.

Another main event in the early training stage is the control of the exoskeleton via the crutch

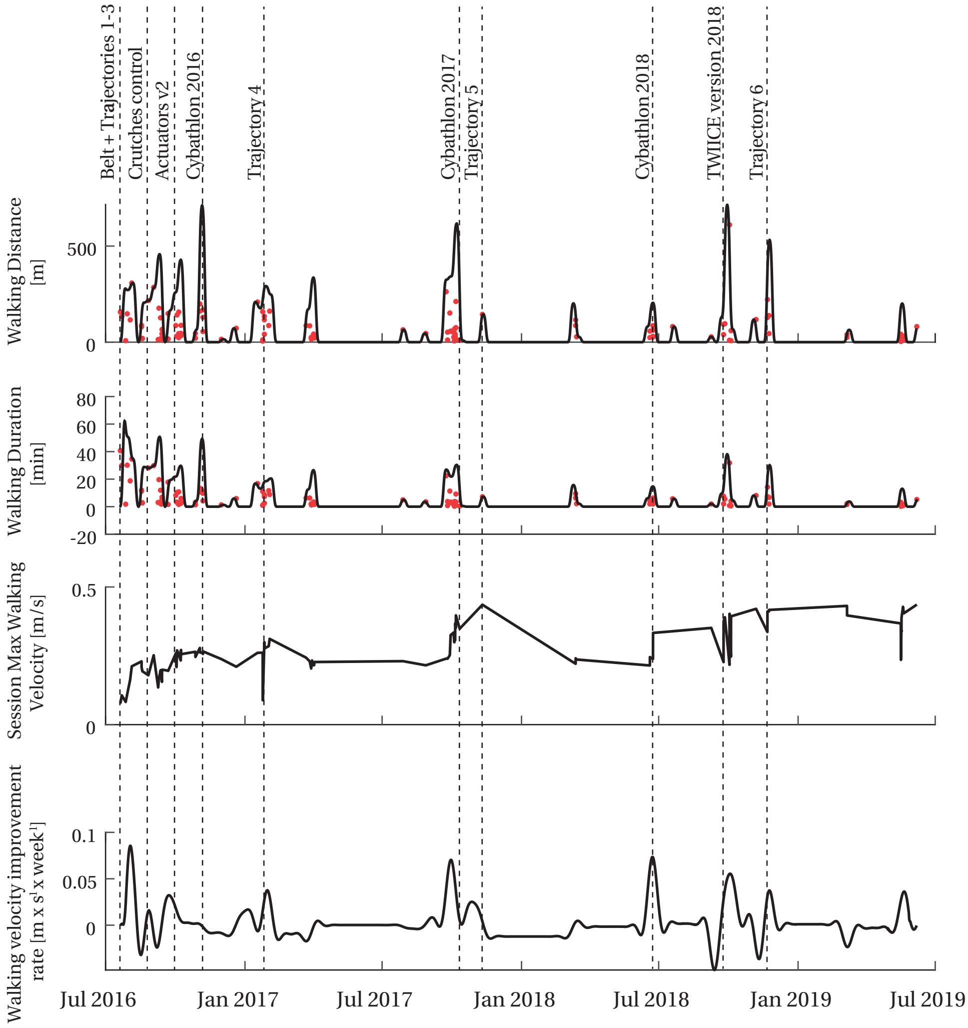

Figure A.5 – Evolution of walking velocity based on walking duration and distance

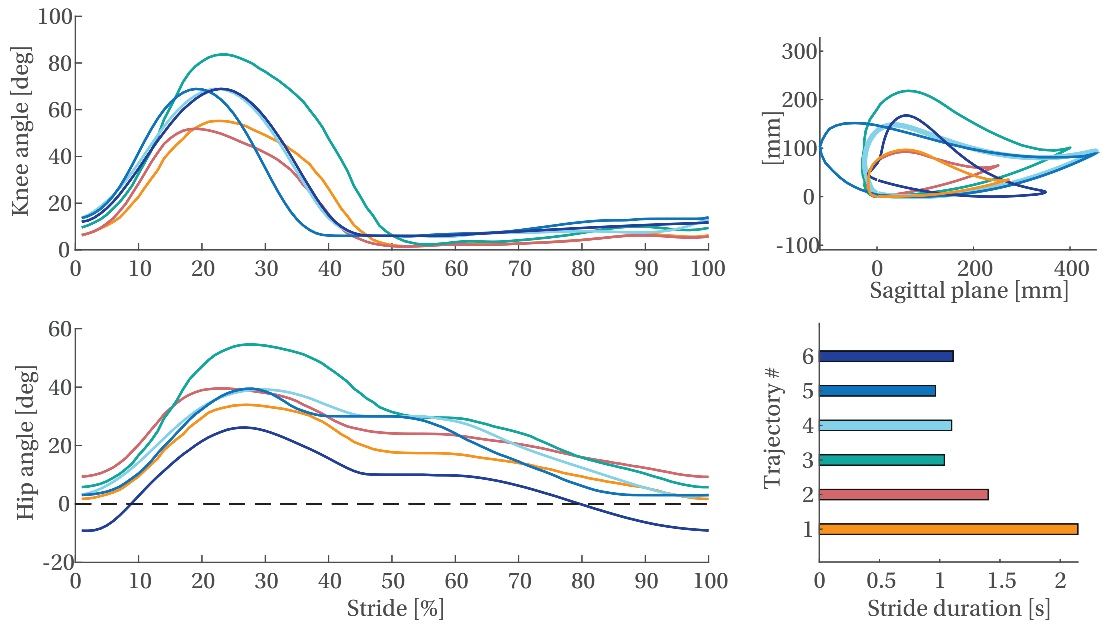

Figure A.6 – Evolution of walking trajectories

remote. During the initial training period, steps were triggered externally by the experimenter based on a subjective evaluation of the posture and weight shift. Active control was given to the user via a trigger placed below the right crutch forearm (Figure 6.4.A). This was first associated with a velocity decrease due to the cognitive load added to the user. After about one month of practice (2x times per week), the pilot learned to trigger the step in an automatized fashion which led to a long-term increase in velocity and higher self-reported confidence by the user.

We can also notice that training periods before the Cybathlon competitions (e.g 2017 and 2018) always led to a positive walking velocity improvement rate. This may be due to the fact that during these training the fastest trajectories are used, contrary to demos. Moreover, the pilot is fully focused on walking with few mental burdens around her.

After the 2017 edition of the Cybathlon, another training session was performed to focus exclusively on improving walking velocity. The trajectories were tuned (v5) and the pilot received feedback on about her posture and the crutches placement. This led to the fastest velocity ever achieved (0.44 m/s) but reached a level which was not felt particularly safe and stable for the pilot. This trajectory was kept as fast walking speed but was not used on all occasions. This again explains the coexistence of three velocity levels in the following time period.

Finally, it is relevant to notice that the new version of the TWIICE in 2018 involving new actuators and an overall ergonomic improvement did not have an important impact on the walking speed. The walking trajectories (v6) were again improved by shifting the hip joint further back to straighten the back. This progressively improved the walking velocity while yielding a more natural and safe feeling gait pattern as reported by the pilot.

A.3 Credits

There were many people involved in the developments of the TWIICE exoskeleton. Their work merits proper recognition. The present list is an attempt to highlight the amount of talent and dedication that was necessary for the achievements presented above.

- Mechanical design: Tristan Vouga, Paul Bertusi, Julien Pache
- Actuation design: Tristan Vouga, Paul Bertusi, Julien Pache
- Hardware fabrication: Tristan Vouga, Paul Bertusi, Julien Pache
- Electronics design & fabrication: Romain Baud
- Assistance strategies: Jemina Fasola, Romain Baud, Tristan Vouga
- Software and firmware implementation: Romain Baud
- Experiments and data collection: Jemina Fasola, Tristan Vouga, Romain Baud, Julien Pache
- Data analysis: Tristan Vouga, Jemina Fasola

Curriculum Vitae

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Swiss
Born September 8, 1989

EDUCATION

PhD Student at the Laboratory of Cognitive Neuroscience (LNCO) — since Oct. 2015
Swiss Federal Institute of Technology – EPFL, Switzerland

Summer School in Computational Sensory-Motor Neuroscience — August 2014
University of Minnesota, Minneapolis, MN, USA
Winner of the 2-week project competition
Award: poster presentation at the TCMC satellite meeting to the SfN annual meeting 2014

Master in Bioengineering — 2012-2014
Swiss Federal Institute of Technology – EPFL, Switzerland
Master project: *Robot-facilitated adaptive training for stepping in an unstable environment*
Robotics Lab, Rehabilitation Institute of Chicago and Northwestern University, USA
Under the supervision of Dr. James Patton and under the direction of Prof. Silvestro Micera, Translational Neural Engineering Lab, EPFL, Switzerland
Semester project: *Analysis of Distributed Spinal Network Activity during Locomotion and Electrical Stimulation*
Translational Neural Engineering Lab, EPFL, Switzerland
Under the direction of Prof. Silvestro Micera

Bachelor cycle in Life Sciences — 2008-2012
Swiss Federal Institute of Technology – EPFL, Switzerland
Bachelor project: *Assess the impact of social cues in bodily self-consciousness*
Laboratory of cognitive neuroscience (LNCO) and Virtual Reality Laboratory, EPFL, Switzerland
Under the supervision of Prof. Olaf Blanke

PROFESSIONAL EXPERIENCE

Research Assistant at the Robotic Systems Laboratory, LSRO, EPFL, Switzerland — 2015
Development of TWIICE, a lower limb exoskeleton for complete paraplegics

Neurotec Pharma, Barcelona, Spain — summer 2012
Contributed in the development of novel strategies for treatment of neurodegenerative diseases

Teaching Assistant, EPFL — 2010-2013
Internet and Robotics courses for children, Equal Opportunities Office

Ice Skating club, Moutier — 2004-2013
Teacher of figure skating

LANGUAGES

French: Mother tongue
German: Good – B1 level
Swiss-German understanding
English: Good – C2 level

ADDITIONAL INFORMATIONS

Training Certificate of Introduction to Research Ethics, Informed Consent and Research Ethics Evaluation of the TRREE training program in research ethics evaluation.

PROFILE

Vice Swiss champion of synchronized ice skating (junior and senior) — 2000-2008
International competitions

Swiss Olympic Talent — 2005-2008

Leisure: Kickboxing, traveling, soccer

PUBLICATIONS

- **Journal articles**
  - **Fasola J**, Baud R, Vouga T, Ijspeert A, and Bouri M, *Bioinspired Postural Controllers for a Low-Actuator Count Exoskeleton Targeting Complete SCI Users*, Frontiers in Robotics and AI, Biomedical Robotics, 2020, submitted.
  - **Fasola J**, Kannape O, Blanke O, *Motor Awareness is Effector Independent but Strongly Influenced by Kinematic Task Demand*, in preparation.
  - **Fasola J**, Kannape O, Blanke O, *On the Relation between Error Augmentation Learning and conscious Motor Awareness in a Goal-Directed Balance Task*, in preparation.

- **Conference articles**
  - Vouga T, Baud R, **Fasola J**, Bouri M, Bleuler H, *TWIICE – A Lightweight Lowerlimb Exoskeleton for Complete Paraplegics*, IEEE 15th International Conference on Rehabilitation Robotics (ICORR), London, 2017, published.
  - **Fasola J**, Kannape O, Bouri M, Bleuler H, Blanke O, *Error Augmentation Improves Visuomotor Adaptation during a Full-Body Balance Task*, 41st Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC), Berlin, 2019, published.
  - **Fasola J**, Vouga T, Baud R, Bleuler H, Bouri M, *Balance Control Strategies during Standing in a Locked-Ankle Passive Exoskeleton*, IEEE 16th International Conference on Rehabilitation Robotics (ICORR), Toronto, 2019, published.
  - Baud R, **Fasola J**, Vouga T, A. Ijspeert, M. Bouri, *Bio-inspired standing balance controller for a full-mobilization exoskeleton*, IEEE 16th International Conference on Rehabilitation Robotics (ICORR), Toronto, 2019, published.
  - Vouga T, Baud R, **Fasola J**, Ijspeert A, Bleuler H, Bouri M, *INSPIIRE – A Modular and Passive Exoskeleton to Investigate Human Walking and Balance*, 8th IEEE RAS/EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob), New-York, 2020, submitted.
- **Talks**
  - **Fasola J**, *Error Augmentation Improves Visuomotor Adaptation during a Full-Body Balance Task*, 41st Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)
  - **Fasola J**, *Balance Control Strategies during Standing in a Locked-Ankle Passive Exoskeleton*, IEEE 16th International Conference on Rehabilitation Robotics (ICORR), Toronto, 2019.
  - **Fasola J**, *Balance Control Strategies during Standing in a Locked-Ankle Passive Exoskeleton*, Cybathlon Symposium, 2020
- **Patents**
  - Vouga T, Baud R, **Fasola J**, Bouri M, BIO-INSPIRED STANDING BALANCE CONTROLLER FOR A FULL-MOBILIZATION EXOSKELETON, provisional US patent, No. 62/874,787