Haptic interactions that alter the control of body balance
- from single and paired individuals to human-robot interactions

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1. Abstract

More than any other sensory modality, touch requires active motor participation on the part of the individual, which means that processes of action control and perception are closely interwoven in order to enable fine, instantaneous control of the contact forces during tactile exploration. In this respect, sensory and contextual predictions play a central role in haptic interactions. A special context, which is the central topic of this habilitation thesis, is the optimized sensorimotor control of body balance during light contact with an external reference point. However, light touch with an environmental reference enforces context-specific, including social, constraints that limit an individual's action degrees of freedom. However, these restrictions can be used therapeutically. For example, when light touch between a therapist and her patient are used in a targeted manner during balance and gait exercises in order to optimize movement control. Ultimately, mobile robotic systems that apply light touch can contribute to betterment of the quality of care for future generations of older individuals in our aging society.

2. Introduction

Our skin is a truly impressive but versatile sensory organ as evidenced by its responsiveness to a diverse range of physical stimulations. Specialized mechanoreceptors, embedded in glabrous as well as in hairy skin, are sensitive to static pressure and changes in pressure, skin indentation and stretch, movements across the skin, vibration, temperature, as well as touch applied in a social context. The perception and interpretation of a tactile afferent signal, however, is not a straight forward affair but requires a representation of the context such as the touch location on the body, the body's current posture, the perception of environmental factors, the demands of the current activity or postural task, any planned actions as well as the actors psychological state and intentions. This means that disambiguation of a tactile signal necessarily involves high-level, supraspinal, subcortical and cortical neural circuits culminating in the experience of qualia, for example the pleasantness or unpleasantness of tactile stimulation in a social setting. In the recent past, the social implications of interpersonal touch were acknowledged by reports of sexual harassment or the discussion of the mental and emotional consequences of social distancing rules in the still ongoing SARS-CoV-2 pandemic.

As the current pandemic-related social restrictions fittingly express, our living environment is affected by constant change either due to natural causes or human interventions. Being able to adapt to changing environmental requirements and engaging successfully in the planning of appropriate actions and the anticipation of their consequences in terms of intended and unintended effects on the environment or the own body, we need to internally represent these environments and any possible interactions therein. This requires learning and skill acquisition in the sense of generating and updating our internal representations of the dynamics of these environments based on the experience we gather while being immersed in them. It is not only our living environment that changes with time but our body as well. For example, as an individual is getting older, the properties of his or her body changes in terms of its physical anthropometry, strength and flexibility. Consequently, an individual's brain is constantly altered in terms of its structure and functionality due to development and learning of motor function but may also become subject to degradation caused by ageing-related biological factors.

One of the most remarkable achievements in the evolution of motor functions of the human organism is the ability of keeping a stable vertical posture against the pull of gravity. The components of the neuromusculoskeletal system and the interaction between these components is a great marvel which's emergence can be observed in the motor development of every single human individual from infant to adult. While in the animal domain quadrupedalism can be achieved literally within hours after birth, unsupported bipedalism in the human takes a minimum of nine months to one year to achieve and several years to optimise with respect to an integrated feed-forward and feedback control of body balance and posture. Learning to stand and walk unsupported usually means that during motor development each individual pass through a period in which support is provided by another human individual. In this period of learning to walk, social interactions between infant and parent show a qualitative change towards increased bidirectional modes of interaction. During this time, tactile feedback may be involved in the generation of internal models and body schema representations for the control of limb movements and body balance. Learning to integrate multisensory information may be directly related to the development of coherent body representations.

It is unsurprising, therefore, that almost any kind of neural deficit is likely to result in some reduction of the efficiency of the postural control system and an increased falls risk. Due to the necessity to integrate a multitude of sensory channels with each other but also with the motor system, postural control is very susceptible to any disruption of the sensorimotor components of central nervous system. Our capability to keep a stable posture is easily compromised by a physical impairment or neurological pathology such as Stroke or Parkinson's disease. The importance of the ability to control body balance for an independent and satisfactory lifestyle is expressed in the loss of the capability to stand without external support demonstrated in frail older adults. When a human individual reaches a life's phase of old age, it is seen frequently that individuals use either an external balance aid or receive manual balance support by another person. Taking the developmental analogy, it may be transferred into the context of movement education and rehabilitation in older individuals who experienced a lesion of the central nervous system. Adapting to compensating for physical impairments and updating internal body schema representations require experience-dependent relearning of multisensory congruencies involving the visual, vestibular, somatosensory and haptic channels. Haptic interactions between a therapist and a patient after stroke for example play an important role when retraining body balance during standing and walking following the brain lesion. It becomes clear that the skin as a sensory organ for the perception of physical interactions with the environment, the ability to control body balance and the demands of social interaction and interpersonal postural coordination converge in situations when either a parent helps an infant to become able to walk or when a therapist assists a patient in a training session.

The complexity of these social postural activities become apparent when human ingenuity tries to replicate them in artificial systems. Progress expressed by humanity's ambition to create autonomous, biologically inspired mechatronic devices, such as humanoid robots, imitating humans' movements and postures has been impressive but gradual. Although these robotic devices improve, it is still apparent that they are still a long way off in terms of the functional versatility, flexibility and adaptability observed in human postural activities.

Neural control of body balance and locomotion, neural degeneration with aging and following brain injury, haptic social interactions for balance support are the main themes that converged in my research in the past years. This present Habilitation treatise will summarize the essence of 17 studies conducted in about fifteen years of post-doctoral research into the confluence of haptic processing for the control of body balance. Beginning with eight studies into the benefits of light touch augmentation for the stabilization of an individual's externally perturbed body balance, this synopsis will cover the expansion of the light touch paradigm into the social domain in four studies as well as in four additional studies its application to haptic interactions with individuals with motor disorders and balance impairments following congenital or later in life acquired neural deficits. Finally, the translation of the deliberately light interpersonal touch approach into the domain of human-robot interactions for balance support will be discussed by one study.

My main intention is to convince the reader that balance control in general, but also balance control with light touch augmentation in particular, is much more than just an issue of the quality of sensory feedback but entails a close coupling between perception and action. In other words, I believe at this point of writing that controlling body balance and posture during standing or walking while maintaining light haptic contact with the external world, be it an earth-fixed reference or another person, imposes a very specific postural context. Light touch contact provides feedback cues for the stabilization of equilibrium but at the same time it also imposes task-specific constraints on postural control by the requirement to keep touch light. In addition, the current stance posture and any cognitive or perceptual tasks additional to the postural task need to be considered when deciding whether or not to utilize any tactile information available for the control of body balance.

The basic understanding is summarized in the following. As touch is an active sense, which involves some degree of motion, either by moving my finger across a surface texture or by having the surface move beneath my finger, actively suppressing body sway, for example by increasing postural stiffness, results in an improved tactile signal-to-noise ratio at the contacting skin section. In this way, the cutaneous feedback becomes informative about the own body sway relative to the external reference. In an analogous way, we hear better, when we are silent. Nevertheless, perception and disambiguation of a tactile afferent signal requires experience and access to previously acquired world knowledge, such as expectations about how I am affected by and how I am affecting myself the object I am in contact with. The high-level representations and beliefs about my own body and its dynamics as well as the environment determine how I perceive my current and future postural state and whether light touch can be considered to be hindering or assisting when keeping body balance within stable limits. This reasoning applies the more to situations, in which contact is kept with another individual, who could be more or less stable than oneself. In this sense, I have arrived at an understanding that ecological principles ought to be taken into account when considering the mechanisms of sensorimotor control of body balance but also movements in general.

I will use terms such as body balance and body equilibrium interchangeably. Measures of body sway were the main dependent variables in most of the research studies conducted. Nevertheless, body sway is generally ambiguous in nature as reduced sway may be linked to improved balance control by means of augmented self-motion feedback or it may also be caused by increased postural stiffness due to increased agonist-antagonist muscle contractions at a specific joint. While the former means improved postural stability in the face of postural destabilization, the latter frequently means less efficient balance control with reduced flexibility and stability and increased energetic expenditure.

2. Control of body balance

Regulation of postural stiffness

Quiet upright standing, as an example of a simple, apparently static posture, is still a highly dynamic activity caused by the interaction between body displacement through gravity pull and opposing muscle-produced torque. This interaction causes the body to show swaying motion in the horizontal plane (Winter, 1995). As long as the body's Centre-of-Mass (CoM) ground projection remains within the limits of the support base, stance can be considered stable. Successfully keeping body equilibrium when reacting to an externally imposed perturbation or when performing a voluntary action means that the duration of unstable equilibrium states is reduced to a minimum. Stiffness of the body or its individual limbs refers to the resistance against a force acting upon it. With respect to postural equilibrium, this can be an external force, such as the pull of gravity or an additional pull or push by environmental objects or agents, or a self-imposed force. The opposite of stiffness is compliance and means the amount of elasticity an object shows when affected by an external force. The importance of the active control of body stiffness for keeping the body in an equilibrium state during upright standing has been debated for a while.

The dynamics of body sway in upright stance can be well approximated by the application of a single inverted pendulum model (IPM) with a single joint at the ankles, simulating an "ankle strategy" (Winter et al., 2003). Winter and colleagues (Winter et al., 1998) also proposed an IPM for the mechanics of body sway in the frontal plane. The IPM is appealing as it simplifies the mechanical system to a single degree-of-freedom problem of balancing the body's CoM against gravity's pull above the ankle joint. The inverted pendulum is considered stable when the stiffness around the ankle joint is equal to or exceeds the pull of gravity. The total muscular activity generating torque around the ankle joint is modelled as a mass-spring-damper (MSD) with specific stiffness and damping characteristics, which are thought to arise partly due to passive viscoelasticity of the muscle-tendon system but also partly due to active CNS control (Morasso & Schieppati, 1999). Using mechanical or sensorial perturbations, the stiffness and damping parameters can be determined experimentally.

IPMs have been used to simulate the dynamics of body and limbs during standing and walking. Often the intention was to demonstrate the efficiency of MSD-like sensorimotor control strategies in terms of energy-return in steady postural states. The expression of MSD-like properties, however, implies regulation of stiffness, perhaps in a context-specific manner. In many studies, the IPM has been considered a useful simplification of the dynamics of human body sway in the sagittal plane during upright standing (Loram & Lakie, 2002b; Morasso & Schieppati, 1999; Peterka, 2002; Winter et al., 2001). Depending on the postural context, however, a double inverted pendulum model may be more appropriate for describing postural adjustment strategy, for example by including a second joint at hip level (Horak & Nashner, 1986). Thus, the single IPM as well as the MSD approach have been criticized as too simplistic and inappropriate for capturing the biological complexity of the human postural control system during perturbed or unperturbed standing (Creath et al., 2005; Kistemaker & Rozendaal, 2011; Pinter et al., 2008). For example, Pinter and colleagues (2008) used Principle-Component-Analysis on the angular movements of the ankle, the knee and the hip to conclude that postural control during free upright standing is better analysed with a double or triple IPM. Even if one adopts a body model consisting of three segments (lower leg, upper leg and trunk), however, the single trunk segment above the hips is criticized as not considering the flexibility of the human spine. For example, anti-phasic movements of the upper trunk relative to the pelvis have been observed following either backward rotations of the support surface or sideways roll (Gruneberg et al., 2004). On the other hand, Schweigart and Mergner (2008) re-evaluated the validity of the inverted pendulum model for dynamic postural control following a tilt of the support surface in the anterior-posterior direction and concluded that, despite specific nonlinearities in the observed dynamics, the IPM can still be considered a legitimate simplification.

As the concept of motor equivalence has been extended to human control of postural sway (Scholz et al., 2007), one would expect from the postural control system's ability to flexibly form muscle synergies adapted to the current postural context, that specific restrictions of the body's degrees-of-freedom would actually force the human postural control system to employ an inverted pendulum control strategy. Thus, any conclusions about the systems sensory processing, reweighting and conflict resolution strategies based on the IPM would remain valid still even when the postural control system would adopt different muscle synergies in a postural context with more body degrees-of-freedom at its disposal.

Brief increases in body and limb stiffness have been observed in abundance as part of functional compensatory postural adjustments following external perturbations to body balance in sitting, standing, and walking, for example in the form of translations of the support base, forces acting onto the body directly as well as slips and trips. The origins of these responses are considered to reside in supraspinal, long-latency reflexes, that demonstrate a considerable context-sensitivity (Diener et al., 1983; Kurtzer, 2014). Thus, higher-order predictive processes representing the situational demands are quite likely involved in the phasic control of compensatory postural responses. This may extend to high-level mental representations of the body and its integrity and the motivation to prevent physical harm, for example following the experience of a fall (Do & Chong, 2008). Dysfunctional increases in body stiffness as a consequence of psychological stress, for example fear of falling and fear of heights, have also been reported as a consequence of muscular co-contractions in the legs and trunk (Brandt & Huppert, 2014; Wuehr et al., 2014).

Multisensory integration

The human postural control system possesses a multitude of sensory channels to distinguish between self-motion and motion of the environment, including vision, vestibular sensation, proprioception in the leg muscles and cutaneous sensations in the soles of the feet. In a complex, dynamic environment, the human central nervous system (CNS) has developed context-dependent strategies for how to resolve sensory ambiguities and conflict between these sensory modalities as well as how to cope with sensory deprivation through intersensory substitution and reweighting (Maurer et al., 2006).

Vision is a sensory modality which is especially important for the control of body balance during standing as well as walking. Loss of vision invariably causes postural instability and increased body sway (Diener & Dichgans, 1988). Retinal stimulation by environmental lamellar optic flow has been shown to alter postural sway probably by disguising the swaying motion of the body resulting in insufficient and less carefully timed muscle activation (Stoffregen, 1985). Involuntary postural sway entrainment to optic flow (Lee & Lishman, 1975) can be reduced by the presence of a second afferent channel, for example vestibular sensation. For example, MacNeilage et al. (2012) demonstrated that vestibular input facilitates discrimination between self-motion induced optic flow and externally caused optic flow, which should lead to more adequately tuned postural adjustments and thus less sway. In terms of this visual-vestibular interaction, it has been shown that it is also effective in the opposite direction. For example, it has been shown that the visual channel is important for the disambiguation of vestibular otolith information invoked by either static tilt of linear acceleration (MacNeilage et al., 2007).

Predictive mechanisms in the control of balance and posture

In order to achieve postural equilibrium, the postural system needs to consider the physical properties and the dynamics of the segments of the body for the preparation of postural adjustments through the coordinated production of joint torques and forces. The system requires the capability to predict the consequences of its actions but also the capability to predict any change in its afferent input resulting from either a specific voluntary action or an external perturbation. Merfeld and colleagues (1999) proposed that neural processes representing internal models of physical dynamics are involved in the fusion of multimodal sensory input and the disambiguation of sensory conflict in order to predict the consequences of voluntary actions as well as external perturbations on body balance and current sensory afferences. Kuo (2005) extended this assumption by suggesting that internal models predict the stochastic behaviour of body sway and estimate the current equilibrium state based on the available noisy sensory input. Kiemel and colleagues (2002; 2011) proposed a direct link between state estimation and body sway and attributed observed body sway to computational noise in the optimal state estimator. Thus, human postural control is considered a mixture of feed-forward and feedback control processes that regulate the body's segments in the gravity field by minimizing of a cost function, which could be implemented as a continuous control process (van der Kooij & de Vlugt, 2007).

An overview on the current computational models of the postural control system has been provided by Kuo (2005). The majority of these include an error signal, often the deviation of the angle of the ankles from a set target value, as critical variable to be minimized during successive postural adjustments. For example, a linear feedback control system approach, which is frequently adopted in the field of robotics, regulates vertical orientation of an inverted pendulum system by continuously driving the system's motor plant with a signal composed of three error-related modes resembling a Proportional-Integral-Derivative (PID or a PD controller alternatively) controller. This feedback controller is paired with a “Kalman filter”, which is effective in the minimization of the variable error of the system's state estimation by consideration of the system's processing and measurement noise (Peterka, 2002, 2003).

The implicit assumption that a continuous control process regulates body sway has been criticized more recently by arguing that intermittent feedback control seems to be more plausible from a biological point of view. Although Winter et al. (1998) identified control of postural stiffness as the major component of balance control in quiet standing, he discounted feedback control mechanisms.

As intrinsic joint stiffness is insufficient to passively stabilize body sway and as feedback control is affected by time delays, it has been convincingly demonstrated (Lakie et al., 2003; Loram & Lakie, 2002a, 2002b) that anticipatory central mechanisms are involved in the actively control of ankle stiffness to modulate body sway and to keep body balance stable. Loram and colleagues assumed that feedback control of body balance is a serial, ballistic process in which the postural state is observed continuously but adjustments occur in an intermittent, predictive open-loop fashion (Gawthrop et al., 2011; Loram et al., 2011). While peripheral mechanisms for balance control are supposed to have a high processing bandwidth, the bandwidth of central balance control mechanisms is considered relatively low. For example, relatively long feedback time delays of latencies longer than 150 ms indicate low bandwidth but a more flexible control of the direction and amplitude of body sway by the involvement of intentional control mechanisms (Loram et al., 2009). In contrast, shorter response latencies are more likely to be controlled by automatic mechanisms which influence the amplitude of a response but not its direction. Loram et al. (2005) reported that muscle adjustments to stabilize an external load occur with an approximate rate of about two to three adjustments per second based on high-level processes such as multisensory integration, anticipatory planning and internal model representations.

An important distinction between theoretical approaches that assume an intermittent feedback control scheme is the question whether any stability interventions are performed at regular temporal intervals or in an event-dependent fashion, by which a postural adjustment is triggered when the estimated postural state, either measured or predicted, transgresses a threshold criterion. For example, the work of several work groups suggested that body sway is controlled in an event-related intermittent manner. Bottaro et al. (2005) reviewed the literature on prominent postural control schemes and concluded that stiffness control and continuous error-based feedback control schemes make unrealistic assumption such as a high level of intrinsic stiffness and a high level of noise in the background signal. As an alternative, they provide a theoretical framework ("sliding mode control theory"), which assumes that stabilization of posture alternates between phases of passive falling and ballistic stabilization attempts (Bottaro et al., 2005). Further, Asai et al. (2009) argued that an intermittent feedback control scheme represents a more physiological model of human body sway control than continuous control models. A similar conclusion was reached by Delignieres et al. (2011), who analysed the fractal dynamics of Centre-of-Pressure (CoP) motion during quiet stance and argued that intermittent velocity-based control of posture is a better descriptor of body sway than position-based control. Likewise, Tanabe et al. (2016) proposed that passive tendon and muscle viscoelasticity in combination with intermittent feedback control governs postural stabilization during upright standing. Tanabe et al. (2017) used a three-segment inverted pendulum model to model electrical myographic (EMG) activations of proximal and distal leg muscles during upright standing. They assumed that intermittent on and off switching of muscle activations is triggered by the ratio of stability/instability manifolds in the phase plane. The found, however, that the change of rate of the stability ratio preceded the on- and off-switching of muscle activation by around 160ms to 200 ms. The event-driven intermittent feedback control model by Tanabe et al. (2017) assumes muscle-specific switching between on- and off-states to generate corrective joint torques while being affected by a feedback delay. This mechanism is complemented by passive joint torque through muscular stiffness and damping, which partly stabilizes body sway without delay. The decision at which point to activate and deactivate specific muscles could be made by a high-level monitoring process, that considers the rate of change in the stability component in a phase space representation (Tanabe et al., 2017). Morasso et al. (2019) proposed that body sway is controlled by a hybrid scheme in which intermittent feedback control adjusts ankle torque, while the hip joint is stabilized by its passive stiffness properties only.

On the other hand, the assumption that intermittent feedback control of sway is exercised by higher-level processes has also been questioned. Elias et al. (2014) simulated leg muscle activation for sway stabilization and proposed that some muscles such as the medial and lateral gastrocnemius are controlled in an intermittent fashion while the soleus is controlled continuously. Further, they argued against cortical involvement in the control of quiet standing and suggested that reciprocal inhibition on a spinal level contributes to the regulation of body sway instead (Elias et al., 2014). Nevertheless, it seems justified to conclude that more automatic processes may be regulating the maintenance of postural stiffness and damping, while high-level processes monitoring and predicting states of postural stability will provide additional corrective joint torques as required according to the representation of the own stability state and the current postural context.

The role of predictive, feed-forward control of balance and posture is also illustrated by the occurrence of anticipatory postural adjustments prior to any voluntary postural activity. It is well established that ground reaction forces and torques precede distinct movements, such as taking a step or performing an upper limb reaching action, with the aim to reduce self-imposed disturbance of body balance (Benvenuti et al., 1997; Bouisset & Zattara, 1987). Massion (1994) suggested that internal forward models of the body and its dynamics are involved in the prediction of the postural consequences of voluntary movements. As my research, that is fed into this treatise, did not focus in anticipatory postural adjustments, with one exception in Johannsen et al. (2007), I will not discuss this topic and the associated paradigms involving discrete, voluntary limb action in more detail here.

Involvement of central structures of the nervous system in the control of body balance

The understanding of the structures of the central nervous system (CNS) contributing to the control of body balance and posture relies mostly on research into postural and balance impairments seen in individuals after brain injury. However, in the last two decades neurophysiological and neuroimaging techniques, such as near-infrared spectroscopy (NIRS), electroencephalography (EEG), functional magnetic resonance imaging (fMRI) and positron emission tomography (PET), have contributed additional insights into the role of supraspinal structures of the brain in balance control. A discrepancy in terms of the dependent variables associated with balance control exists between studies, that used patients' lesion information to infer important supraspinal structures for balance control, and studies in the normal population, such as younger and older adults, that investigated the sensorimotor control of body sway. The former studies mainly relied on behavioural criteria, such as the inability to keep balance stable in standing or walking or the demonstration of deficits in spatial orientation perception of the environment or the own body or severe postural lean in a specific direction. In contrast, the latter employed quantitative methods to describe the variability or complexity of body sway directly.

The processing of visual, vestibular and somatosensory afferents for the control of posture and balance is assigned to certain regions of the brain stem, the cerebellum, the thalamus and the cerebral cortex (Brandt & Dieterich, 2000; Dichgans & Diener, 1989; Dieterich & Brandt, 1992, 1993, 2019; Fukushima, 1997). Impairments in standing and walking have been reported as a result of lesions of the brain stem, cerebellum, thalamus and cerebral cortex (Brandt & Dieterich, 2000; Dieterich & Brandt, 1992). Patients with "Wallenberg Syndrome" after a brainstem lesion usually show a tendency to lose their balance to the side while standing, but are able to sit upright without support (Dieterich & Brandt, 1992; Kim et al., 2004). In the case of "thalamic astasia", however, a sitting posture as well as standing and walking can lead to a loss of balance (Brandt & Dieterich, 1993; Masdeu & Gorelick, 1988). Remarkably, these patients show no additional deficits that indicate a disturbance of the vestibular system, such as a deviation of the subjective visual vertical (SVV), but tend to have rather mild sensorimotor impairments of the contralateral half of the body (Masdeu & Gorelick, 1988). Cortical projections of the ventral posterolateral and posteromedial nuclei of the thalamus terminate in the postcentral gyrus, the secondary somatosensory cortex in the parietal operculum, and the island region (Engelborghs et al., 1998; Jones, 1985). Damage to the posterolateral thalamus and the posterior island region results in a misperception of the orientation of the visual environment and is thus understood as an integral component of the vestibular system (Brandt et al., 1994; Dieterich & Brandt, 1993).

In the brain of the macaque monkey, Grüsser et al. (Grusser et al., 1990a, 1990b; Guldin & Grusser, 1998) delineated a network of cortical regions which integrates vestibular and visual information with proprioception of the neck muscles. At the centre of this network is the “vestibular cortical system”, which is also known as the “parieto-insular vestibular” cortex (PIVC). The PIVC lies deep in the Sylvian groove, posteriorly in the island region (retroinsular and granular insular cortex) and is closely linked to the homologous region in the other hemisphere and all other regions in the network (Guldin & Grusser, 1998). The importance of the posterior island region and the parietal operculum as well as the transition region from the temporal to the parietal cortex in the processing of vestibular stimuli and the perception of body orientation has also been validated in the human subject by means of electrical stimulation and methods of functional imaging (Kahane et al., 2003; Petit & Beauchamp, 2003). Lacquaniti and colleagues (Indovina & Sanes, 2001; Zago & Lacquaniti, 2005) suggest that the posterior-insular vestibular cortex is not only a region that integrates multimodal sensory afferents for perception of spatial orientation, but also contains a representation of gravity in an allocentric reference system that is used for both motor as well as for cognitive tasks.

From a theoretical point of view, a consistent distinction between the two hemispheres regarding their role in the control of body balance and body sway has not been proposed. For example, posture and balance is quite often impaired in stroke patients. While hemiparesis is the most prominent cause of postural instability in stroke patients, the integration of afferent sensory input also seems to be affected (Marigold et al., 2004). Their deficit might be caused either by sensory loss or by affected internal processes of sensory integration and postural control. Hemiparetic stroke patients show abnormal anticipatory muscle activity in response to oscillations of the support base (Hocherman et al., 1988), impaired postural adjustments during voluntary limb movements (Horak et al., 1984; Palmer et al., 1996), increased sway and prolonged stabilization following external disturbances to the hip (Wing et al., 1993), disordered temporal sequences of muscle activity when going onto tiptoes (Diener et al., 1993), delayed muscle response onset latencies and inefficient postural control strategies in more dynamic situations such as during locomotion or following external perturbations to body balance (Badke et al., 1987; Dietz & Berger, 1984; Holt et al., 2000; Kirker, Jenner, et al., 2000; Kirker, Simpson, et al., 2000) as well as body weight bearing asymmetry during upright standing and walking (Bohannon & Larkin, 1985; Dettmann et al., 1987; Sackley, 1991).

Much more severe than a weight-bearing asymmetry is the inability to keep a vertical posture during sitting or standing as observed in patients with symptoms of contraversive pushing. The first study, which systematically examined the lesions of patients with severe pusher symptoms, following conservative criteria of diagnosis, comes from Karnath et al. (2000). The lesion data from 23 patients were merged, compared with control patients without pusher symptoms and resulted in the posterolateral thalamus as the neuroanatomical structure that caused the symptoms with the highest frequency after damage. A prospective follow-up study confirmed this finding insofar as haemorrhages in the posterior thalamus in both hemispheres are more likely to lead to pusher symptoms, whereas this was never the case for infarcts of the anterior thalamus (Karnath et al., 2005). With regard to the brain lesions that spare the thalamus but include the cerebral cortex, Johannsen et al. (2006) that compared with patients without pusher symptoms, the left hemispheric posterior island region, the superior temporal gyrus, the inferior parietal lobe and the right hemispheric postcentral gyrus are more frequently damaged in patients with pusher symptoms. Ticini et al. (2009) examined the cerebral blood flow in pusher patients with thalamic and extrathalamic lesions and were able to show that undersupply of cortical or thalamic regions in the respective group is not related to the pusher symptoms. Extra-thalamic lesions with pusher symptoms showed an undersupply in structurally intact areas of the inferior frontal gyrus, the middle temporal gyrus and the inferior parietal lobe. It can be concluded from this that the pusher symptoms can be caused independently of one another by lesions in both areas.

In an extensive study, Babyar et al. (2019) investigated lesion imaging and functional impairment in patients with and without symptoms of lateropulsion after stroke and reported that lesions in the inferior parietal lobe in the junction between Brodmann areas 2 and 40 correlates most frequently with lateropulsion. The volume of the lesions was also larger in these patients and motor functions and functional independence were more impaired. Baier and colleagues (2012) reported that pusher patients with right-hemispheric lesions tend to show damage to the posterior insula region, the operculum and the superior temporal gryus, while left-hemispherically lesioned patients tend to show damage to the anterior insula region, the operculum, the internal capsule as well as the lateral thalamus. They interpret their findings as an indication of a possible functional connection between posture control and the processing of vestibular information for body orientation in space.

Frank and Greenlee (2018) postulated a division of the traditionally called posterior insular vestibular cortex into two functionally distinct areas: an area in the parietal operculum that is inhibited by visual information processing and that uses vestibular afferents to process head and body movements while appreciating the spatial alignment, as well as an area located in the retroinsular cortex, which combines visual and vestibular afferents in order to determine the spatial orientation and to differentiate between proper movement and movement of the environment on the basis of visual information. From there, projections could be made to the temporoparietal transition in order to map signals of proper movement and alignment in the spatial self in an egocentric frame of reference (Frank & Greenlee, 2018).

Active stability state estimation

Sensorimotor control of movements, posture and balance is strongly dependent on predictive processes for accurate and optimal estimation of the postural state of the body. Predictions are by nature sensitive to the current action context. Instead of relying exclusively on passive sensory stimulation to cross-validate multiple sensory systems and update internal models of the current postural state, active exploration, in other words deliberate probing of sensory feedback, of the body's stability state based on anticipated sensorimotor effects of self-imposed balance perturbations is a much more promising strategy for tuning the components of the sensorimotor system.

A central assumption from an ecological point of view is that the postural control system purposefully generates sensory feedback for estimating the equilibrium state of the body. In the domain of postural control, Riccio and Stoffregen (1988) demonstrated that active perception of body orientation is grounded on the perception of corrective actions required to keep a chosen tilted or listing posture. They suggested that a trade-off between less effortful postural corrections when nearer the equilibrium point and more reliable interpretation of the forces acting on the body with greater deviation from the equilibrium point governs postural control. Based on this interpretation, Ehrenfried et al. (2003) speculated that sway reduction under increased cognitive load could express reduced effectiveness of active stability state exploration and suggested that the dynamics of body sway involve active, attention-demanding probing of self-motion and the state of balance stability in the context of noisy and potentially disruptive sensory stimulation.

3. Human tactile sensation and peripersonal space

According to Bajcsy (1988), "active" sensation and perception is an intelligent sensory data acquisition process. Thus, touch as a sense that affords active perception and exploration requires a minimum of body motion dynamics, also to counter potential effects of habituation to any tactile stimulation (Prescott et al., 2011). For example, only by moving against and across a surface will one perceive the properties of the surface, such as smoothness, compliance and rigidity.

The human skin contains a multitude of mechanoreceptors sensitive to different forms of physical stimulation. Glabrous skin contains the majority of mechanoreceptors responsible for the discrimination of tactile stimulation in the hands and feet such as texture or shape of an object or roughness of the ground during gripping or locomotion (Zimmerman et al., 2014). Four types of fast conducting Aß-mechanoreceptors detect static skin stimulation (slowly-adapting receptors innervating Merkel cells), skin stretch (slowly-adapting receptors ending in Ruffini corpuscles), movement across the skin (rapidly-adapting receptors innervating Meissner corpuscles) and high frequency vibration (up to 200 Hz; rapidly-adapting receptors terminating in Pacinian corpuscles). Hairy skin, which covers up to 90% of the body surface, also contains several low threshold mechanoreceptors associated with hair follicles to detect skin indentation (slowly-adapting receptors innervating Merkel cells) and hair deflection (Zigzag, Guard, Awl/auchene receptors innervating hair follicles; Zimmerman et al., 2014).

Tactile acuity varies considerably between regions of the skin surface. The two-point discrimination (2PD) thresholds as a popular measure of tactile sensitivity depends on the size, density and overlap of tactile receptive fields distributed over the surface of the skin. Thresholds for 2PD are lowest in the fingertips and the palms of the hands indicating finest tactile sensitivity and highest on the lower back and the upper thighs (Stevens & Choo, 1996). Nevertheless, even comparable mechanoreceptor densities in different regions of the skin can result in differences in tactile acuity possibly mediated via central, high-level perceptual processes (Mancini et al., 2014).

The skin is an important also channel for social information exchange via slow conducting C-type tactile (CT) mechanoreceptors. Hairy skin especially is associated with the emotional and social quality of touch (Morrison et al., 2010). McGlone et al. (2014) proposed a distinction between sensory discriminative touch and social affective touch. The separation of affective and discriminative touch begins with the type of mechanoreceptors involved. Depending on the speed of tactile skin stimulation, affective CT mechanoreceptors show maximum discharge frequency at intermediate stimulation speeds. Loeken et al. (2009) demonstrated that intermediate stroking velocities maximizing CT receptor discharge frequencies resulted in the highest pleasantness ratings of the stroking. In contrast, discriminative mechanoreceptor discharge frequency rises in a linearly increasing fashion with increasing stimulation speed (McGlone et al., 2014). It seems that stimulation of CT mechanoreceptors at their optimal stimulation rate results in correlated brain activations in the right-hemisphere dorsolateral prefrontal cortex as well as the anterior and posterior insula regions (Perini et al., 2015).

The separation between affective and discriminative touch is continued in dorsal spinal cord projections via different thalamic nuclei to separate cortical regions such as the basal medial and posterior thalamus and the insula region for affective touch and the ventral posterior lateral thalamus and the primary and secondary somatosensory cortex for discriminative processing (Morrison et al., 2010). Social touch also interacts with the topography of the body's skin surface dependent on the interpersonal and emotional relationship between touch provider and receivers ("Correction for Suvilehto et al., Topography of social touching depends on emotional bonds between humans," 2015; Suvilehto et al., 2015). While individuals in a partnered relationship are allowed to touch each other all over the entire body, for strangers most of the body is a taboo zone with the exception of the hand and the arms. In their review, Gallace and Spence (2010) discussed the relative importance of interpersonal touch for the creation of interpersonal relationships, emotional communication and general wellbeing. The region of insula cortex seems to be engaged in the processing of social touch (Suvilehto et al., 2020).

When we keep physical contact with an environmental reference, we normally have a good perception of the contact's relative position and motion. It is therefore conceivable that a body schema representation is involved in the localization of the contact and its relative motion in an egocentric frame of reference taking into account a specific limb orientation and body posture. The localization of touch in terms of the location on the body and the location in egocentric space requires high-level representations of the body and its surface. Holmes and Spence (2004) proposed the concept of a body schema representation to subsume an integrated multimodal representation of the body and the near environmental space (peripersonal and interpersonal space) based on a network of interacting cortical and subcortical centres processing multisensory information, such as vision and touch, in task-context-specific frames of reference. According to Cardinali et al. (2009) are body schema and peripersonal space two expressions of the same underlying body representation. Longo et al. (2010) suggested that the remapping of tactile afferent information requires a schema of the body's skin surface to result in the somatic localisation of touch. In combination with the localisation of the body and its segments in space, based on the integration of proprioceptive afferents and efferent motor commands into a postural body schema as well as anthropometric representations of one's body's size and shape, the egocentric localisation of tactile stimulation can be derived (Longo et al., 2010). Within the somatosensory system, parallel processing of somatosensory information may take place along two separate streams (Dijkerman & de Haan, 2007). One stream for the integration of tactile and proprioceptive information into a body schema would enable target-related action, while a second stream would serve perceptual performance such as tactile recognition of objects and the own body (body image). Both posterior parietal cortex and the insula region are considered to be involved in both somatosensory processing streams (Dijkerman & de Haan, 2007).

The flexibility of peripersonal space representations in the context of the subjectively perceived bodily self has been demonstrated by the visual-tactile Full Body Illusion (FBI; Noel et al., 2015; Serino et al., 2018). Individuals experience the FBI, a transfer in their subjective body-ownership and subjective localization within the environmental space, during upright standing when presented with a third-person perspective of their own person, while receiving simultaneous and visually congruent (feel what they see) haptic stimulation of their body. The human does not only possess one single representation of peripersonal space but multiple body segment-centred and a trunk-centred full body peripersonal space representation (Cléry & Hamed, 2018; Serino et al., 2015). A similar illusion can also be achieved for individual limbs of the observer's body, for example by the Rubber Hand Illusion (RHI; Ehrsson et al., 2005). Nevertheless, qualitative neuroanatomical dissociations have been reported between distinct co-activation networks overlapping in the left-hemisphere parietal cortex of the human brain concerned with the multisensory representations of either (i) peripersonal space for body-object interactions or (ii) body ownership, which may be important for the distinction between self and other during haptic interactions (Grivaz et al., 2017). De Vignemont and Iannetti (2015) proposed a generalized functional dissociation between two main purposes of sensorimotor processing concerning the space surrounding the body. One purpose would be the protection from bodily harm and the other would be the guidance of goal-directed action. Bufacchi and Iannetti (2018) reconceptualized the concept of peripersonal spaces as specific action-relevance fields. If touching or being touched are considered actions then it may follow that applying touch for stabilization of body balance both requires and is subject to representations of the peripersonal space and the body.

4. Sensory augmentation of body balance control by light touch

In the previous section, the theoretical assumptions and associated empirical evidence were presented that serve as the background to the discussion of how the provision of additional tactile afferent information through skin contact with environmental references alters the control of body balance and postural stability.

Touch-based interactions with the environment

The versatility of the human postural control system is expressed by the adaptation of non-plantar cutaneous afferences for the regulation of body sway. Haptic interactions with the environment provide powerful cues for inferring the current stability state. For example, upper limb tactile feedback can be recruited as an additional source of information about body sway. Actively produced or passively received haptic interactions with the environment provide powerful cues for inferring the current state of balance and postural stability. Holden and colleagues (1994) showed in healthy participants that mechanically non-supportive fingertip contact (<1 N) with an earth-fixed reference results in sway reductions, which may be even more efficient than sway reduction caused by visual feedback (Jeka & Lackner, 1994). Body stiffness is directly affected by the postural context of keeping light touch. Franzen and colleagues (2011) demonstrated that postural tone of hip muscles increased by 44% with light touch, which correlated with the observed sway reductions.

Even in unstable sitting without feet, arm or back support, fingertip light touch improves body balance (Albertsen et al., 2014). It has been shown that both cutaneous receptors detecting shear force variation through skin stretch and muscle proprioception of the contacting body segment provide sensory feedback about relative body sway (Kouzaki & Masani, 2008). Withdrawal of the tactile differential by simultaneously referencing the contact location to body sway removes the benefit of light touch feedback (Reginella et al., 1999). Skin feedback can also be utilized for sway control when the tactile contact is received passively (Krishnamoorthy et al., 2002; Rogers et al., 2001). In passive light touch, the receiver is generally not directly involved in the application of the contact and therefore not able to control aspects such pressure of the contact. A necessary qualification is, however, that the participant could opt for either a touch-compliant or touch-avoiding postural strategy. It has been shown that cutaneous receptors sensitive to skin stretch detect small differences in shear forces at the contact location but also muscle proprioception of the contacting upper limb may provide sensory feedback about the direction and amplitude of body sway relative to the contact location (Rabin et al., 1999). While the majority of studies demonstrated the benefits of haptic feedback exclusively in quiet standing, a few have reported increases in postural stability in a more dynamic context such as treadmill walking (Dickstein & Laufer, 2004; Forero & Misiaszek, 2013; Fung & Perez, 2011) and staircase negotiation (Reeves et al., 2008; Reid et al., 2011). In older adults, light contact with handrails during stair negotiation changes ankle and knee coordination during ascent and improves postural stability in terms of the separation distance between CoM and Centre-of-Pressure (Reeves et al., 2008), although Reid and colleagues (Reid et al., 2011) did not find indications for improved stability when using a handrail. During treadmill walking, handrail use induces specific changes in reflex patterns all over the body that might facilitate postural stability during gait (Lamont & Zehr, 2007). The dependency of postural control with light touch on contextual expectations and previous experience has been demonstrated by Bryanton et al. (2019).

Balance perturbations

Traditionally, improved body balance control in terms of reduced body sway with light touch has been demonstrated in quiet stance. As we have seen above, however, reduced body sway is inconclusive in the sense that it might be caused by improved self-motion feedback and more efficient postural control. On the other hand, light touch being an implicit supra-postural precision task might also result in reduced body as would a postural stiffening strategy, which might result in sway reduction but would not necessarily lead to improved postural stability. The exposure to a mechanical perturbation and the observation of subsequent postural compensation might provide a more direct measure of postural stability or stabilization with light touch. Dickstein et al. (2003) demonstrated that the postural response to a horizontal, backward support surface translation was altered by the presence of light touch at the fingertip in terms of increased sensitivity to the initial scaling of the postural response. Martinelli et al. (2015) demonstrated that following an abrupt backward pull at the hip, fingertip light touch reduced the amplitude and peak velocity of CoP motion with greater effects without visual feedback and when standing on an unstable surface.

In Johannsen et al. (2007), we designed an unimanual pulling paradigm to investigate the time course of light touch contact effects on standing balance after predictable self-imposed (with voluntary pull) or unpredictable externally imposed (with reactive pull) perturbation. In both cases, perturbing forces at the right hand in the region of 20 N produced forward directed movement of the CoP in the anteroposterior direction. We predicted light touch conditions, involving shoulder contact with a fixed reference, would improve the efficiency of reactive components of the response in reflex pull, facilitating earlier reduction in sway compared with no touch conditions. In contrast, we did not expect any effects of light touch on stabilization after voluntary pull under the assumption that feed forward anticipatory postural adjustments would already optimally stabilize body balance after the perturbation. We concluded that the immediate response to a voluntary perturbation and the stabilization after both a voluntary and a reflex perturbation are altered by light touch. To maintain light touch, CoP fluctuations are differentially modulated in the anteroposterior and mediolateral directions after a voluntary perturbation. Thus, light touch influences CoP fluctuations not only by providing a sensory spatial reference but also by constraining the movements of the body after a perturbation, which may be an indication that the requirement to keep light touch imposes specific task-related constraints that shape the postural responses as well after a perturbation. We demonstrated that the effect of light contact with an external reference is equally effective irrespective of a self-imposed or externally imposed perturbation of balance. This was surprising as one could expect that anticipatory postural adjustments preceding a self-imposed perturbation would result in lesser postural perturbation and more optimal sway compensation and therefore would not benefit from augmented tactile sway-related feedback.

Using a similar approach as in Johannsen et al. (2007) in a follow-up study, Johannsen et al. (2017) modified the availability of and dynamics the in the visual feedback exposed participants to vertical perturbations of the support, in order to investigate any interactions between support compliance, complexity of the visual surrounding and the availability of light touch contact to a compliant support. Light touch offered a substantial benefit to participants ability to balance in dynamic visual environments and on compliant supports. A visual environment of moving forest canopy did significantly destabilise humans' balance. The impact on postural stability of the dynamic forest environment combined with standing on the compliant support was as severe as when the participants were blindfolded. A similarity in the sway response for being blindfold and when viewing the dynamic visual environment suggests that vision was down-weighted in the latter to reduce its destabilising impact, amounting to a deprivation of visual feedback. Due to the visual-vestibular discrepancy in the forest canopy condition, body sway was significantly higher on the compliant compared to the stiff support. It also had a substantial effect on thigh muscle activity since rectus femoris was 40% more active and vastus lateralis 29% more active after the perturbation. Thus, the central nervous system initiated inappropriate muscle activation when it was harder to distinguish between movement of the body and movement of the naturalistic environment. We expected that light touch on a compliant hand support would provide sufficient additional proprioceptive and cutaneous feedback from the fingers to reduce the destabilising effect of the dynamic physical and visual environment without destabilising the body by displacing the hand support. We showed that light touch with a compliant reference improves postural compensation of external vertical perturbations of a compliant support and reduces muscular activations in the lower extremity. Even though the hand support was highly flexible, we found that light touch reduced body sway by 24% in quiet standing, independent of the visual and support conditions. After the perturbation it significantly reduced sway for all visual conditions, but particularly for the dynamic visual environment and when blindfolded (22% and 29% respectively). Further, the relative impact of light touch was most powerful for activity in the upper leg where it reduced rectus femoris and vastus lateralis muscle activity by more than 30% after the perturbation.

Haptic sensory transitions

The (re-)organization of the sensorimotor control loop for balance is a time-consuming process. Even when anticipating an upcoming change between states of available sensory feedback. Interestingly, the integration of touch information into the postural control loop seems to happen faster than the time required for the integration of visual information (Lackner & DiZio, 2005; Rabin et al., 2006). On the other hand, it has been argued that the processing of touch may be more complex than the processing of visual feedback and therefore result in longer integration latencies in specific postural contexts (Sozzi et al., 2012). Reported delays of more than 300 ms between light fingertip haptic feedback and subsequent postural adjustments, nevertheless, imply high complexity of internal processing of the light touch signal, for example to disambiguate the feedback signal in the current postural and environmental context (Rabin et al., 2006).

In Johannsen et al. (2014), we aimed to investigate the processes involved in reconfiguring the postural set following either the addition or withdrawal of active fingertip light contact to or from the sensorimotor feedback loop governing body sway during upright standing. Using an intermittent touching paradigm, we examined how a train of short duration light touch periods alters body sway. We assumed that sway would be reduced by touch contact within 2 seconds after onset. In addition, our expectation was that a time-consuming, asymptotic integration process would result in incomplete integration and weaker sway after-effects during touch exposure durations shorter than 1.5 seconds. With our intermittent light touch paradigm, we not only confirmed the postural stabilization effects of light touch at the fingertips, but also revealed after-effects after withdrawal of touching in dependency on the duration that light touch was available for. Our findings that intermittent light touch of longer than 2.5 seconds resulted in delayed return to baseline levels of body sway may be an indication that switching between postural control sets with and without light touch inclusion is a process that is not immediate but may involve phases of top-down reduced body sway, for example by means of increased postural stiffness.

Our follow-up study by Kaulmann et al. (2017) was intended to replicate the previously observed after-effects on body sway following removal of intermittent light contact at the fingertip. In addition, based on the assumptions by Sainburg (2014) that both hands are associated with distinct functional roles in bimanual activities and that this distinction is based on characteristics of functional brain lateralization in humans, we expected that effects of light touch on body sway might be mediated by the specific hand used for the sampling of light touch for balance control. Sainburg (2014) proposed that the left hemisphere in right hand-dominant individuals is more likely to control the processes that predict the effects of body and environmental dynamics. In contrast, the right hemisphere was considered to be involved in impedance control, meaning limb stiffness, to minimize movement error when arm movements are unexpectedly perturbed by mechanical forces as well as to keep accurate position of the limb in a steady-state posture. In Kaulmann et al. (2017), We observed that actively removed intermittent light touch at the fingertip led to a rapid increase in sway within 500 ms after contact removal for short contact durations but a more persistent contact aftereffect for longer durations when kept with the dominant hand. Intermittent light touch at the fingertip of the non-dominant hand showed rapid increase for short and long contact durations. This difference could not be explained by the tactile sensitivity of index fingers of the two hands, which was not different. The general progression of sway during a contact removal transition was in line with the previous study of Johannsen et al. (2014). In contrast to this previous study, Kaulmann et al. (2017) tested contact durations longer than 5 seconds, which are more likely to result in steady-state sway with light contact. Indeed, also for steady-state sway with light touch at the fingertips, we found that the sway progression increased at a lower rate after touch removal but only when the touch was established with the dominant hand. We proposed that consolidation of the central postural set for light touch balance control with the non-dominant hand might possess a longer time constant due to lack of experience, for example compared to exploring the environment with the dominant hand, and possibly due to differences in functional hemispheric laterality. The observations might indicate that the left-hemisphere might take longer to readapt to a voluntary switch in the postural control set, especially when haptic information is to be removed from the control loop.

Suprapostural task requirements

The requirement of lightly keeping contact affords precise control of the contacting forces in terms of the perceived tactile variability. It has been argued that the demands of the sensorimotor task of keeping touch light reduces body sway. In addition to augmented own body sway-related feedback, keeping light contact represents the goal of a suprapostural task with an external attentional focus (McNevin et al., 2013; McNevin & Wulf, 2002; McNevin et al., 2000; Riley et al., 1999). Riley and co-workers (1999) demonstrated that the light touch effect on sway is dependent on the salience of the contact within the current postural context. In other words, participants, for whom finger contact occurred only coincidentally, did not show any reductions in sway. Thus, any light touch effects on body sway are not only a consequence of tactile feedback processing but may also indicate proactive sway control when light touch contact is available. In this context, it is remarkable that merely the intention to establish light contact with an earth-fixed reference in the immediate future (less than 5 seconds of the present moment) can result in effects on body sway similar to actual contact (Bove et al., 2006). Traditionally, the interpretation of body sway serving performance in a supra-postural task was evidenced by the use of visual cognitive tasks requiring oculomotor precision, for example a visual search paradigm (Stoffregen et al., 2000; Stoffregen et al., 1999), which assumes a direct coupling between retinal slip induced by oculomotor activity and body sway. Direction-specificity of sway control was reported in visuomanual aiming tasks (Balasubramaniam et al., 2000) but also in visual search involving saccadic eye movements instead of smooth pursuit. Damping body sway by increasing postural stiffness should enable greater oculomotor precision and visual sensitivity. Chen et al. (2015) showed that light touch improved search performance. Demands of the visual search task, however, reduced sway independent of light touch availability so that two processes seemed to act in parallel (Chen et al., 2015).

In the study published by Kaulmann et al. (2018), we intended to contrast two functional couplings between noise in a visual signal detection task and either the variability of body or the variability of fingertip light touch. We expected that a coupling of visual jitter in the signal detection task to variability of light touch at the fingertip would create a hierarchy between sway control and light touch. Our results also demonstrated a direction-specific reduction in mediolateral body sway below a level achieved by light touch sway-related feedback augmentation alone if an implicit feedback coupling was present between visual jitter and either body sway or light touch variability. Both direct and indirect involvement of fingertip contact in the specific implicit feedback coupling condition minimized sway. This observation implies either that no control hierarchy existed for whole body sway and fingertip contact, for example in terms of an integration of both control processes, or that any hierarchy can be reversed flexibly, for example so that one control process facilitates the other, if it serves the implicit goal of reduced perceptual noise and enhanced performance within the context of a supra-postural visual signal detection task. This study demonstrated the flexibility of any task- and context-specific recruitment of body sway control.

Cortical involvement in the control of body balance with tactile feedback

Transcranial magnetic stimulation (TMS) over the motor cortex has been used to investigate corticospinal projections to trunk muscles. Chiou et al. (2018) showed that corticospinal excitability of the erector spinae is modulated by the context of specific postural tasks. Only a few studies investigated cortical processes involved in the control of body balance with light touch. Bolton et al. (2012) reported that dorsolateral prefrontal disruption by TMS alters the processing of a body sway-related tactile signal. On the other hand, Ishigaki and coworkers (2016) showed that EEG activity in the left-hemisphere posterior parietal cortex (lPPC) is related to body sway with light touch and that sway is disrupted during cathodal transcranial direct current stimulation over this location.

Using repetitive TMS (rTMS) in healthy participants, we demonstrated that the left-hemisphere inferior parietal gyrus plays a role in transient sensory reorganisation and conflict resolution following sudden, externally controlled removal of a light touch reference (Johannsen et al., 2015). Our central observation consisted of an increase and overshoot in body sway upon unpredictable and abrupt onset as well as removal of fingertip contact. After onset of contact, sway increased relative to sway without contact and only began to settle towards a less variable steady state after 2 s from onset. It then took another 2.5 to 3 s until steady-state sway with fingertip contact was achieved. After contact removal, a rapid increase in sway was observed, which culminated in an overshoot about two to three times the magnitude of the overshoot at onset. Our participants were not able to anticipate the time point at which contact was established. Thus, the initial overshoot might have been the consequence of an involuntary startle response to touch plate contact. Alternatively, abrupt contact with the touch plate, which was driven by a linear motor, could have resulted in a slight perturbation to the fingers of the right hand. However, as the respective peak overshoot did not follow immediately after the event, the possibility of a mechanical perturbation affecting body sway seems unlikely. Instead, we observed a gradual build-up in sway, which reached maximum within 1 to 2.5 s following the contact event. Unintended overshoot in body sway variability following unpredictable removal of sway-specific fingertip feedback is altered after disruption of left-hemisphere IPG by rTMS. Reduction in transient postural disorientation on haptic deprivation after left inferior parietal gyrus disruption could be attributed to degradation of perceived intermodal conflict or efficiency increases in the detection of intersensory discrepancy and intermodal conflict resolution due to lowered intra- or interhemispheric inhibition. We concluded that the left-hemisphere inferior parietal gyrus plays a role in transient sensory reorganisation and conflict resolution in the stabilization phase following sudden passive removal of a light touch reference at the dominant, contralateral hand.

As left-hemisphere inferior parietal gyrus disruption by rTMS did not directly affect the integration of light touch into the postural control loop, we assumed that the light touch contact position and body sway relative to this position would be processed within an egocentric frame of reference (Kaulmann, Hermsdorfer, et al., 2017). Some studies have demonstrated that the utilization of light touch for controlling body sway does enforce a local reference frame linked to the relative movement of the contact (Assländer et al., 2018; Franzen et al., 2011). Asslander and colleagues (2018) proposed a postural control model with light touch, which contains two integrated and nested feedback control loops. The first control loop aimed to minimize change in the relative distance between the body's Centre-of-Mass and fingertip position, which maintains the current postural configuration, while the second control loop adjusts body sway when the light touch reference is moving and comprises intersensory reweighting. Azanon et al. (2010) demonstrated that right-hemisphere posterior parietal cortex (rPPC) disruption impairs the localization of a tactile cue in a head-centred reference frame. Therefore, we aimed to evaluate the effects of disruption by continuous theta burst stimulation (cTBS) of the PPC in both hemispheres on the processing of fingertip light touch for body sway control in Tandem Romberg stance (Kaulmann, Hermsdorfer, et al., 2017). Disappointingly, neither left nor right-hemisphere PPC disruption affected transient states of light touch integration or removal or steady-state body sway with light touch. Surprisingly, however, after stimulation of the rPPC, the general level of sway variability was decreased. This encompassed all trial phases including those in which light fingertip contact was applied and body sway reduced by the augmented sensory feedback. Light touch changed the sway dynamics in a direction-specific manner in favour of the mediolateral direction. In the mediolateral direction, however, a second effect of rPPC disruption became visible. After the stimulation, the sway dynamics degraded in those phases in which light contact was kept with the non-dominant, contralateral hand. we replicated the traditional effect of light touch on body with decreased sway variability but showed direction-specific changes in its complexity. Moreover, we showed that as overall sway variability decreased, in addition to the light touch effect, sway complexity decreased after rPPC disruption when utilizing haptic information from the non-dominant, contralateral hand. We speculated that an increase in postural stiffness could result from lowered reciprocal inhibition of stiffness regulation by a disrupted process, which is engaged in actively exploring the body's stability state. We proposed a simple functional model of interhemispheric interactions, which could explain our results pattern by the assumption of an asymmetry between the rPPC and lPPC regarding bilateral utilization of haptic information for the control of body sway.

From the previously reported findings, the following picture seems to emerge. Two processes that require higher-order, attentional resources, for example activation of specific neural circuits, to a different amount are involved in the control of body sway: stiffness control, which is susceptible to attentional distraction to a lower degree, and active exploration, which poses greater demands on attention. In addition, these two processes are coupled in a reciprocal inhibitory fashion. Small amounts of attentional distraction disrupt active exploration, which loses inhibitory influence on stiffness control, so that body stiffness is increased. In addition, it seems that the left hemisphere inferior parietal gyrus is more involved in the processing of tactile information for the control of body sway than the right hemisphere, at least when the haptic signal is sampled using the dominant hand. In contrast, the right-hemisphere posterior parietal cortex seems to counteract postural stiffness regulation, possibly for the purpose of active stability state exploration, irrespective of the availability of light touch.

The previous study by Kaulmann et al. (2017) did not provide any evidence for the assumption that either the lPPC or rPPC is directly involved in the processing and integration of light touch for the control of body sway. In that study, however, we used a quiet stance paradigm and it has been reported that rPPC is involved in postural responses to expected mechanical balance perturbations (Mihara et al., 2008). Therefore, Kaulmann et al. (2020) pursued two main objectives. The first was to investigate whether light fingertip contact improves balance compensation following a perturbation unpredictable in its relative force so that generation of a context-specific central postural set would be hindered. The second was to assess the role of the right posterior parietal cortex for the control of postural stiffness by disrupting the rPPC using cTBS. We expected strong effects of light fingertip contact on body sway and muscle activations before, at and after a perturbation indicative of light touch feedback resulting in improved postural stability. Disruption of rPPC, on the other hand, was expected to hinder facilitation of sway stabilization with light touch but also affect the immediate response to a perturbation and sway stabilization by induced greater postural stiffness. We found a strong effect of Light Touch, which resulted in improved stability following an unpredictable perturbation. Light Touch decreased the immediate sway response, as well as the steady state sway following re-stabilization. Decreased sway is accompanied by reduced muscle activity of the ankle muscles. We assumed that improved sway response would lead to increased stability, which required less torque production around the ankles in order to stabilize the body. However, we did not find an improvement of the time constant in response to the perturbation with Light Touch. The lack of improvement might be a result of a different postural context or the unpredictability of the force of the perturbations. We observed a gradual decrease of muscle activity, which is indicative of an adaptive process in terms of lower leg muscle activity, following exposure to repetitive trials of perturbations. This supports the idea that exposing people repetitively to a perturbation leads to an optimization of the postural response. Given the range of the perturbations we suspect that the postural control system settled for a compromise across the three different perturbation forces and prepared for a medium configuration. This is supported by the notion that we see greater decrease of muscle activity in the medium force push condition. Regarding the effects of the disruption of the rPPC we were not able to confirm our hypothesis that disruption of the rPPC leads to increased postural stiffness. On the other hand, we found an unexpected effect of cTBS stimulation in terms of improvements of the aforementioned adaptive process. After disruption of the rPPC muscle activity of the Tibialis Anterior was decreased even more compared to sham. We can conclude that rPPC disruption enhanced the intra-session adaptation to the disturbing effects of the perturbation.

In conclusion, the assumption that an egocentric frame of reference, and possibly the implication of action within peripersonal space, as represented in the right-hemisphere PPC determines the processing of own body sway relative to a light touch contact location was refuted in both of the reported cTBS studies. The evidence that we accumulated points in the direction indicated by the Sainburg model (2014), which assumes that the right hemisphere is involved in limb impedance control. In contrast, the left hemisphere inferior parietal gyrus seems to be associated to a certain degree but not primarily with the processing of light haptic signals for balance control involving the dominant hand.

Interim conclusions

Availability of light touch feedback during a mechanical perturbation optimizes the postural response, in terms of a shortened stabilization time constant and reduced lower limb muscle activity. Whether the perturbation is self-imposed, and therefore predictable, or externally imposed does not seem to make a difference to response with light touch. The advantage of light touch feedback becomes more apparent in conditions where the other sensory channels are challenged by information reduced in its reliability. Transitions between states of tactile sway-related feedback utilization require reorganization of the central postural set. This process seems to be altered by the hand involved in the contact, such that contact with the dominant hand in right-handers controlled by the left hemisphere seems to lead to more inert representations of a central postural set with tactile inclusion. Although the processing of a light touch reference implies the processing of spatial information in an egocentric frame of reference, disruption of the brain areas, that are traditionally attributed with representing these functions, such as the posterior parietal lobes, has no effect on state-transitions and steady-steady body sway with light touch. Instead, it seems that the inferior parietal gyrus of the left hemisphere is a better candidate region

5 Social postural coordination with light interpersonal touch

According to the evolutionary interpretations of Shepard (1984), an organism's internal representations of significant environmental patterns will cause behavioural resonance when the actual patterns are encountered. In the human context, interpersonal mimicry may occur due to psychological and social constraints, for example the latent intention to build good rapport with another person (van Baaren et al., 2009). Smith and Mackie (2016) defined psychological closeness as the overlap between self and other based on mental representations of the other's observed actions and experiences, which could be unconsciously influence the observer's own actions. Chartrand and Bargh (1999) used the term "Chameleon Effect" to describe the unconscious and unintentional imitation of the postures and behaviours of an interaction partner and suggested that this phenomenon might arise from direct perception-action couplings. For example, visual observation of other moving human bodies has an influence on self-motion perception based on vestibular stimulation (Lopez, 2015). Lopez et al. (2015) proposed that a mechanism of interpersonal sensorimotor resonance between the self and other humans caused this influence. Tia and colleagues (Tia et al., 2012; Tia et al., 2011) used the term "postural contagion" to describe an involuntary destabilizing effect when visually observing the motion dynamics of another individual in a challenging balancing task. They interpreted postural contagion as an expression of insufficient inhibition of automatic imitative tendencies when viewing the balancing efforts of another human individual. A correspondence between the balancing constraints of the observer's current stance posture and the observed individual's balancing efforts enhanced postural contagion. Thus, it is reasonable to conclude that a distinction between self and other is necessary to inhibit imitative behaviour when trying to achieve one's own goals (Brass & Heyes, 2005).

Pezzulo et al. (2013) suggested that when the context makes actions and intentions during social interactions ambiguous, then prediction and imitation, intentional non-verbal, sensorimotor forms of interpersonal signalling facilitate interpersonal coordination in concert with automatic mechanisms of interpersonal resonance. Neri et al. (2006) provided evidence that two individuals interacting in a meaningful synchronized fashion, for example when dancing or martial arts fighting, utilize interpersonal predictive coding for the processing of visual information about the partner's action.

Spontaneous postural coordination between two individuals down to the level of body sway has been reported in several joint activities. For example, Shockley and colleagues (2007; 2003) demonstrated the effect in cooperative conversation. When two people conversed with each other in a joint problem-solving activity without mutual visual feedback, their sway patterns became more similar, possibly due to a task-specific convergence in their verbal stress patterns (Shockley et al., 2007). Richardson et al. (2005) argued that the observed interpersonal postural coordination in this context may result from implicit mimicry between the paired individuals to serve the primary conversational task. Tolston et al. (2014) generalized these observations by arguing that interpersonal coordination is sensitive to the constraints imposed by any single and jointly partnered actions. In addition, Varlet et al. (2011) showed spontaneous postural coordination in visually coupled individuals performing a voluntary swaying task. In another recent study, Athreya et al. (2014) demonstrated that interpersonal postural coordination with mutual visual feedback between the partners seems to rely more on interpersonal visual entrainment and not on the constraints of a manual precision task such as keeping two laser points aligned.

Deliberately light interpersonal touch

Studying interpersonal haptic interactions is the ideal field to observe context-dependent associations and dissociations between stiffness control, respectively adjustment of compliance, and active state perception and exploration. While the paradigms mentioned above mostly relied on visual interactions between two individuals, interpersonal haptic interactions provide a unique window into the perception of the movement intentions and dynamics of the partner by the exchange of forces. Interpersonal haptic interactions provide a unique window into the perception of the movement intentions and dynamics of the partner by the exchange of forces. Therefore, in paired dancing, and also specific martial arts forms, interpersonal touch is a major channel for interpersonal postural coordination.

When considering visual "postural contagion", it is not surprising that light interpersonal touch with a dynamic reference such as another human leads to interpersonal sway synchronization potentially, but not necessarily, due to similar perception-action coupling with the intention to minimize perceived contact force fluctuations. Increased interpersonal sway coordination during quiet standing and voluntary swaying was observed with a haptic coupling as well (Sofianidis et al., 2012).

Sofianidis et al. (2012) reported that haptic contact stabilizes spontaneous coordination dynamics of two partners performing paired periodic voluntary swaying. The strength of the emerged synchrony depended on the individuals' expertise to integrate tactile and auditory information about sway, such dance experience.

Light touch contact with a non-biological environmental reference that demonstrates own oscillatory motion causes involuntary postural sway entrainment as well as increases in body sway compared to a static contact for oscillation frequencies less than 0.8 Hz (Jeka et al., 1997). The entrainment of body sway to the contact's motion depends on the contact's oscillation frequency and complexity (Wing et al., 2011). This effect may be an expression of a default misattribution of any tactile sensation during upright stance to own body motion and automatic postural adjustments coordinated with the velocity as well as the position of the contact point. Saini et al. (2019) reported recently, that the information gained about one own's trunk velocity may be central to the postural adjustments during passive light interpersonal touch received at the trunk.

In Johannsen et al. (2009), we investigated, if light contact with another person through the fingertips improves body sway in a sample of five pairs of 10 older adults with an average age of 65 years. Standing next to each other, body sway of each individual was recorded as a function of the light touch condition at the fingertips. No touch feedback was contrasted with contacting an earth-fixed standing or the other partner in a pair. We found that interpersonal light touch had a beneficial effect on the variability of postural adjustments and body sway during quiet standing. Variability of postural adjustments and sway were both significantly reduced in paired individuals compared to standing without additional tactile input. The reduction in sway variability during interpersonal light touch was smaller than in the light touch condition involving an earth-fixed reference. This reduction in the effect of light touch in the interpersonal light touch condition most likely reflects the variability of the tactile input caused by the partner's sway compared to the constancy of the tactile signal received through the fingertips during earth-fixed reference light touch. Indeed, we noted a small but statistically reliable cross-correlation between members of each pair in the interpersonal light touch condition. This suggests that there is transfer of sensory information between the individuals which affects their own body sway dynamics. In contrast to previous studies that investigated the cross-correlations between postural adjustments while standing upright and fingertip shear forces during light touch with an earth-fixed reference, the interpersonal coordination during interpersonal light touch in terms of cross-correlations appear small. Moreover, postural sway can be quite strongly driven by a spatially oscillating referent, if it is the target for light touch (Wing et al., 2011). Therefore, we speculate that the small correlations between individuals may affected by an inhibitory mechanism that prevents the members of the pair from following the other person too exactly as this might result in amplification, rather than reduction, of body sway in the interpersonal light touch condition.

In the follow-up study by Johannsen et al. (2012), we intended to replicate and extend the findings reported in Johannsen et al. (2009) by examining how light interpersonal touch influences body sway depending on contact location and the stance asymmetry between two partners. We predicted differences in the degree to which an individual would utilize the haptic signal for the fine tuning of postural adjustments and this would depend on the variability of the signal as well as the demands for precise direction-specific modulation of own body sway. Thus, we expected that sway would be reduced less if the partner stood in tandem stance, compared to a partner in bipedal stance, due to the generally increased variability of the touch signal from the partner. This should apply to situations, in which both individuals adopted differing stance postures (asymmetrical interpersonal stance posture: one person in normal bipedal, one in tandem Romberg). Our expectation for this situation was that the more stable individual would contribute less to the variability of the touch signal and therefore would receive less specific feedback about own body sway and as a consequence would show smaller sway reductions. Finally, given the requirement of precision control of body sway, we assumed that shoulder-to-shoulder contact would force participants to constrain their sway more actively in order to compensate for the lower number of postural degrees of freedom of this more proximal contact. The study investigated in young adults the effect of light touch contact between two individuals on each individual's control of body sway as a function of the skin contact site, each individual's stance posture and the interpersonal stance symmetry. Reliable reductions in sway were found during both forms of interpersonal touch. Distal interpersonal touch at the fingertip differed, however, from more proximal interpersonal touch at the shoulder with respect to the proportional reduction in sway depending on an individual's and the partner's stance posture. In-phase interpersonal postural coordination with near zero lag became evident during shoulder-to-shoulder contact on the mediolateral axis. We proposed two different mechanisms for maintaining interpersonal touch during shoulder and finger contact. During shoulder contact, the implicit requirement to minimize variability of the haptic signal might have cued participants to increase their postural stiffness proactively thereby reducing the reciprocal signal noise. While shoulder contact afforded only one control loop to adjust body sway and to minimize variability of the contact, finger interpersonal touch might comprise a control loop for the upper extremity to minimize contact variability, which provides an efference signal that is feed into the body sway control loop. A more recent study by Ishigaki and coworkers (2017) confirmed our observations and demonstrated that interpersonal postural entrainment with social light touch is mediated by the social relationship of both interaction partners.

In the publication by Wing et al. (2011), we aimed to assess to what degree participants' body sway would entrain to a more complex external driving signal. We contrasted entrainment of finger contact by a superimposed signal of two sinusoidal oscillations against a "natural" body sway signal from another individual. We reported that passive exposure to pre-recorded body sway of other individuals did not result in similar reductions in body sway variability as real-time contact with an actual human partner. Like the superimposed signal, passive exposure to a person's body sway increased body sway beyond sway without any contact. This implies that sway reductions with interpersonal touch emerge as a mutually adaptive process between two contacting individuals, perhaps as a direct consequence of minimising the interaction torques at the contact location. The haptic interaction between two individuals does not seem to be the result of a mechanical coupling between both individuals but instead represents the effect of mutually shared sensory information (Reynolds & Osler, 2014). Reynolds and Ostler (2014) proposed a model of the direct exchange of body sway-related information, either by light touch or peripheral vision, which results in spontaneous interpersonal sway entrainment entirely due to increased sensory weighting.

While previous studies focussed on spontaneous postural coordination during quiet standing, which did not elicit clear leader-follower relationships, Steinl and Johannsen (2017) designed a study to assess interpersonal postural coordination with light interpersonal touch in a more dynamic context of a discrete forward reaching task. We intended to modulate the leader-follower relationship by the creation of asymmetric interpersonal dependencies. While the contact receiving and forward reaching individual was deprived of any visual feedback in the less stable postural state, he or she was supposed to rely more strongly on the contact provider when no alternative source of haptic information was available. On the other hand, the contact provider's responsiveness to the contact receiver's motion was expected to vary with the visuotactile interpersonal context in terms of the available visual feedback and any specific instruction regarding the interpersonal touch provision. We observed temporal movement coordination between the contact-provider and contact-receiver to depend on the presence of an external object and the visuotactile interpersonal context.
An object at the reaching person's fingertips influenced the reaching performance with respect to the precision demands, such as speed and accuracy, as expressed by less variable reaching velocity. In the reach end-state, increased amplitude with an additional object coincided with reduced body sway variability. Despite low friction of the fibreglass surface, the interaction with the object could have resulted in haptic feedback at the fingertips facilitating control of balance and resembling a non-rigid, haptic 'anchor' as conceptualized by Mauerberg-de Castro and colleagues (2004).
Interpersonal postural coordination was strongest when deliberately light interpersonal touch was

provided without the presence of an additional object at the contact-receiver's fingertips. As the leader-follower relationship between both partners was also modified by the visuotactile interpersonal context of the contact-provider, the sensorimotor states of both partners have to be considered of equal importance with respect to the observed interpersonal postural coordination. With the exception of one condition, the contact provider as the less constraint individual always took the follower role. Only when no external object was available to the contact receiver and the contact provider closed the eyes as well while interpersonal contact was kept, then the contact provider tended to lead the motion of the contact receiver.

Interim conclusions

In the studies summarized above we demonstrated that light interpersonal contact leads to a stabilization of body sway and to a certain degree of spontaneous interpersonal postural coordination between both contacting partners. These interactions are influenced by the intrinsic postural stability of each partner, which the less stable individual benefitting to much greater proportion, and the non-tactile sensory information available such as vision. It seems also, that these factors may interaction in a fashion, where the more stable partner takes a follower role, if they are able to coordinate their movements with the interaction partner visually. These findings indicate that the interpersonal postural coordination observed in pairs with light interpersonal touch is determined by higher-level representations of the joint action context and may not resemble a spontaneous low-level postural entrainment phenomenon.

6 Clinical applications of deliberately light interpersonal tactile balance support

Cunha et al. (2012) demonstrated that individuals who suffered a stroke are able to use fingertip light contact to reduce body sway. Further, Baldan et al. (2014) reviewed the literature on the effect of light touch on postural sway in individuals with balance problems due to aging, brain lesions or other motor or sensory deficits. They suggested that the maintenance of the fingertip lightly touching an external surface provides additional somatosensory information for individuals with poor balance and that it could be used as a strategy to improve the control of upright standing during intervention programs (Baldan et al., 2014). Indirect light contact with the environment mediated by a balancing aid such as a cane improves postural stability, too. Jeka (1997) suggested that augmented haptic feedback supplied by canes and sticks might lead to the development of mobility aids for balance-impaired populations. In subacute stroke patients, grounded light touch contact through a cane during overground walking facilitates activity of weight-bearing muscles of the paretic leg which results in improved lateral pelvic stability compared to grounded contact with greater force (Boonsinsukh et al., 2009). Balancing ability and lower extremity function determine ability to use grounded cane light touch strategy beneficially (Boonsinsukh et al., 2011).

A specific context where the control of stiffness becomes central to the task goal is the coordination of movements between two individuals. Each individual needs to show some degree of responsiveness to the movements of the partner. If at least one individual lacks responsiveness completely, both will find it impossible to collaborate with each other and joint action will be unsuccessful. Adjusting interpersonal stiffness or compliance therefore is a key aspect of successful interpersonal coordination. When both partners are in physical contact, verbal communication and visual feedback are not particularly well suited to control interpersonal compliance. For example, visual feedback has two obvious disadvantages. First, it requires a certain distance of spatial separation to perceive the parts of or the entire body posture of the partner. Second, a visual, attentional focus on the partner will make it impossible to perform visually directed movements towards an external task goal. A basic assumption in this context is, that a perceptual symmetry persists between both partners. In real-life situations, however, this is often not the case. For example, when moving a heavy object, one partner might need to walk backwards thus relying on verbal prompts from the partner. Alternatively, one partner might be genuinely blind or show other forms of sensorimotor impairments that affect balance and posture, which needs to be taken into consideration of the less impaired partner.

In specific circumstances, light interpersonal touch may be a promising strategy for patient guidance in clinical settings. Manual support to stroke patients provided by healthcare professionals during training of walking are routine activities in physical therapy. The specific features of these manual support techniques, however, are usually not elaborated in much detail. For example, Plummer et al. (2007) reported that supportive therapist-patient handholds were used to facilitate balance during walking. A critical aspect with respect to motor learning in stroke patients is to what degree any supportive handholds constrain patients' movement degrees of freedom and impair or even prevent their active participation in the generation of a locomotor pattern. We argue that the application of deliberately light interpersonal touch for balance support is much less likely to lower the voluntary drive and effort for self-control in patients with balance impairments. Not only does interpersonal touch augment sensory feedback about own body sway, carefully placed interpersonal touch on a patient's body might resemble a soft cue that may prompt the patient to actively search for a behavioural strategy, which optimises the patient’s control of postural degrees of freedom by an affording an external focus of attention.

A central characteristic of interpersonal interactions with light haptic contact is the amount of which both partners are able to contribute to the haptic exchange. For example, in most settings described above, both partners are equally capable of actively contributing to and controlling the interaction, such as when both partners have the same number of movement degrees of freedom available. In an alternative setting, only one partner is actively contributing to the haptic interaction, while the other is more constrained and thus more passively engage in the interaction by performing a different postural task simultaneously, such as the forward reaching task employed in Steinl and Johannsen (2017), or by having fewer postural degrees of freedom available, as we will see below. In Steinl et al. (2018), we contrasted the effects of active and passive modes of participation in the provision of light interpersonal touch balance support by a therapist to balance-challenged older adults. Our ambition was to determine which mode would lead to what amount of engagement in the therapists' client. The participants were not patients per se, but individuals, whose balance control appeared less optimal within their age-group. In the passive mode, the participant faced away from the therapist while receiving haptic support to the back and therefore was not able to contribute significantly to the interaction in terms of controlling contact force precision, presumably. In contrast, shared grip between the therapist and the participant on a manipulandum, while facing each other, allowed the participant to directly influence the precision of the interaction force by the utilization of the extended arm's full movement degrees of freedom. We hypothesized that a mode of passive interpersonal touch reception would result in more stable body sway with greater stabilization under progressively more challenging sensory conditions. In addition to changes in body sway, we characterized spontaneous interpersonal postural coordination as expressed by the interaction forces and correlations between both individual's spatiotemporal sway dynamics. Interestingly, our expectations were not confirmed as the active mode led to lesser sway variability compared to the passive mode. Our findings imply that in balance training, both support modes are able to augment control of body balance in a participant. Whereas the passive support mode demonstrated its advantages in increased strength of the interpersonal coordination the active mode decreased the postural sway of the participant to a greater extent. With regards to future balance rehabilitation, our study showed first indications that training could be more effective when partners face to each other and adopt a more collaborative, partnership-based training approach.

In a study by Johannsen et al. (2017), we evaluated the benefits of passive deliberately light interpersonal touch on the variability of standing balance both in patients with Parkinson's disease and in patients with chronic hemiparetic stroke. We expected that a superior contact location at shoulder level would result in greater reductions in postural sway compared to a contact location at waist level. We reasoned that the inverted pendulum-like dynamics of quiet upright stance result in relative movements of the body segment under the contact point increasing with the distance from the ankle pivot point. Alternatively, one could argue that by being closer to the head a high contact point provides an orientational referent for a maximum number of body segments and joints relevant to the control of posture and balance in upright standing. We found evidence in both groups that interpersonal touch indeed has a stabilizing effect in terms of reductions in body sway. Generalized over both groups, the sway improvement was effective irrespective for contact location but did not apply equally well to all the body locations and combinations assessed. A more superior location at shoulder level tended to result in greater reductions in postural sway than contact at waist level and two simultaneous contact points of which one was located at the high back at shoulder level generated the greatest proportional reductions. Our study demonstrated that both patients with Parkinson's disease and with chronic hemiparetic stroke benefit from passive deliberately light interpersonal touch provided by a care provider. A contact location at shoulder level induced the greatest reductions in postural sway variability compared to a No contact control condition. Interpersonal touch may reduce the risk for falls and play a major role in the rehabilitation of balance control disorders in patients with either Parkinson's disease or chronic hemiparetic stroke.

In Johannsen et al. (2018), we compared the effects of active and passive interpersonal touch support on body sway in mildly impaired, chronic hemiparetic stroke patients and a group of unaffected controls. The results showed that the stroke patients possess similar responsiveness to individual fingertip light touch feedback and interpersonal touch in terms of proportional sway reductions compared to the control participants. No difference between the active and passive interpersonal touch modes were found in both groups, which contrasts with findings reported in Steinl et al. (2018), where collaborative interpersonal touch was more effective in reducing body sway in older adults. The study indicated that the effects of light touch and interpersonal touch are robust but cannot be generalized from healthy older adults to hemiparetic stroke patients without consideration of moderating functional constraints of the individual and the specific postural context, such as postural degrees-of-freedom and positioning of the contact relative to the individual.

Based on the previous studies, where we found interpersonal light touch reduced the variability of sway during quiet standing, we expected that the effect of interpersonal light touch would also generalize to situations in which both individuals are walking. In Schulleri et al. (2017), we aimed to investigate whether deliberately light interpersonal touch at the head is a way to facilitate the control of body sway during walking in children and adolescents with cerebral palsy (CP) and with typical development (TD). The effect of interpersonal touch was assessed in terms of gait speed, temporal gait variability and head and trunk velocity sway. Our results did not turn out exactly as expected, but our study yielded some interesting findings. The participants with CP showed reduced head sway with apex interpersonal touch, opposite to an increase in thoracic sway with interpersonal touch. The TD group, however, responded differently to the testing conditions than the CP group. They showed reduced sway for head and trunk in all interpersonal touch conditions compared to walking alone even in the paired walking condition, in which no interpersonal touch was applied. In contrast, the CP group did not demonstrate a behavioural change in the presence of the therapist. The TD individuals were much more responsive in terms of reductions in head and trunk sway, which may be an expression of reduced sensitivity regarding the social affordances of the interpersonal touch situation in the individuals with CP, which could indicate that the ability to adapt behaviour according to external, social constraints is restricted in CP. We speculated that apex interpersonal touch applied regularly in balance support situations by a carer might improve the habitual locomotor pattern in CP.

Interim conclusions

We demonstrated in group samples of populations of balance-impaired individuals, such as patients with Parkinson's disease, patients with hemiparesis following stroke, children and adolescents with cerebral palsy and older adults, that deliberately light interpersonal touch provides the same benefits to postural stability as in the normal and typically developed population. In Schulleri et al. (2017), we also demonstrated how interpersonal light touch can be used to directly influence the locomotor behaviour of the contact receiver provided the person is sensitive to the social affordances of the context of the tactile interaction.

7 Robotic light interpersonal touch

In recent years, several types of robotic devices have been introduced to neurorehabilitation settings to facilitate upper limb movement as well as lower limb locomotor therapy (Poli et al., 2013). The basic purpose of these devices is to support the weight of individuals' limbs or whole body and to guide limb movements through as many movement repetitions and cycles as possible, more than could be achieved by a therapist's sole manual support in the same amount of time. It is indisputable, however, that current technical solutions for robotic locomotor rehabilitation are still limited in their benefit to the patient. For example, in traditional robot-assisted treadmill walking, in which leg movements are support by the robot, upper body kinematics are significantly altered compared to walking without the robotic-assistance (Swinnen, Baeyens, et al., 2014; Swinnen, Beckwee, et al., 2014). Koopman et al. (2013) suggested that current robot-assisted locomotor therapy lacks a balance training component by forcing lateral pelvic motion towards the centre of the treadmill thereby causing a non-physiological gait pattern without the need for lateral balance control. On the other hand, future generations of robotic gait trainers may become more able to provide locomotor support during overground walking (Vallery et al., 2013). Therefore, with respect to the future design of robotic-assistive devices for locomotor training, Pennycott et al. (2012) demands higher levels of active participation and movement challenge to the patient.

The development of a robotic systems that are able to provide light touch support to a human individual, instead of exclusively mechanical body weight support, will be the next stepping stone in the endeavour of designing effective rehabilitation robotic for balance and locomotion. Patients might also perceive such systems providing light touch support as much less intimidating because the robot would not take hold of the patient's body, an important factor in the patients' acceptance of healthcare robotics (Broadbent et al., 2010).

In Johannsen et al. (2020), we used the maximum forward reaching paradigm described in Steinl and Johannsen (2017) as a model to assess the effect of forms of light interpersonal touch provided by a robotic device on healthy participants' balance performance. Changes in spontaneous maximum forward reaching behaviour and body sway were assessed as a function of the robotic system's mode of control (follower vs anticipation) with respect to the contact receiver's movements. Compared to the body sway in the baseline or end-state as well as the achieved reaching amplitude, robotic interpersonal touch was as efficient as interpersonal touch provided by a human. Beneficial deliberately light interpersonal touch for balance support during maximum forward reaching is easily provided by a robotic system even when it is mechanically uncoupled to the human contact reveicer. This effect does not rely on the robotic system's capability to predict the future position of the contact receiver's wrist. The effects the uncoupled robotic interpersonal touch were comparable to human interpersonal touch on most parameters. As the robotic system itself was not designed for any form of "social" cognition or explicit haptic communication, our study nevertheless demonstrated that robotic interpersonal touch can be used to implicitly "nudge" human contact receivers to alter their postural strategy by adapting to the implicit constraints of the robotic system without any decrements in their postural performance during maximum forward reaching.

Interim conclusions

The main ambition was here to demonstrate that designing a robotic system for supporting an individual's postural stability and balance control is possible but needs to consider an users' spontaneous responsiveness and ability to coordinate with the robotic device. For example, while a normal individual may easily adapt to the requirements of a specific robotic device, a person with a balance impairment of some kind may be respond differently or not at all. In other words, limitations of successful human-robot interaction may be present not only on the side of the robotic engineering solution but also on the human side.

8 General conclusions and discussion

In the previous four sections, I have presented four areas of research concerned with how the availability of light touch shapes and modifies the control of body balance in dynamic settings. In this sense, I hope that it has become obvious that I take an ecological point of view regarding how postural control with light touch is adapted to the perceived task-constraints imposed by the current postural context. According to this understanding, touch is not just an additional sensory channel, which provides reliable information about own body sway relative to the environment. Instead my presumption is that touch is a complex signal, which can only be interpreted for the control of body balance, if beliefs, expectations and predictions about how the environment is behaving with respect to our own behaviour are taken into consideration from the perspective of the postural control systems perspective but from a scientific point of view.

Context-dependent mechanisms of light touch balance control

In the traditional studies by Lackner and colleagues (Holden et al., 1994; Jeka & Lackner, 1994), the postural context was generally unambiguous as the light touch reference location always remained earth-fixed and participants were usually tested in static and steady postural state conditions. Therefore, any tactile signal could be fully attributed to own body motion. Most of the research presented above investigated light touch for balance control in more dynamic situations. For example, optimization of balance compensation following mechanical perturbations, unexpected or voluntary transitions between central postural sets with and without light touch utilization, as well tactile interactions with animated contacts such as haptic or robotic devices or other human individuals showing their own motion dynamics.

In our very first study on the light topic of light touch balance control, we observed already that the provision of passive light touch at shoulder level imposed subtle direction-specific constraints on body sway (Johannsen et al., 2007). In other words, it became apparent that deriving afferent information from keeping light touch with a reference does not just provide sensory cues that augments self-motion perception. Instead, that keeping light touch imposes a sensorimotor task that could be called supra-postural in its nature, possibly involving high-level cognitive and sensorimotor processes. The conclusion that I have arrived at in terms of how light touch shapes balance control, comprises a “dual-mode model”, which consists of two interacting modes of proactive and reactive balance control. The first mode reduces body sway proactively, for example by means of temporarily increased postural stiffness, based on the intention to use tactile feedback as component of the central postural set for balance control in the immediate future. Proactive, feed-forward stiffening would facilitate distinction between self-motion and motion of the contact by minimizing the self-imposed disturbance of the haptic signal. As the proactive “clamp” on body sway is released following a transient stabilization, which results in an improved tactile signal to noise ratio, the integration of own sway-related tactile information into the feedback-based postural control process enables the second mode, which establishes a new postural steady-state.

It seems fair to say that the overlap between both modes needs to be flexible and to be adapted to a current postural task and its situational demands and constraints. The point that I hope to make is, that this model does not represent a model about multisensory integration mechanisms in combination with optimal feedback control but represents a model about postural strategy adjustments, which could entail altered sets of multisensory integration. For example, in the mechanical perturbation situation, a short period of increased postural stiffness based on an expectation of the strength and timepoint of the perturbation would shorten the initial phase of the postural compensation until longer-term stabilization based on augmented sensory feedback becomes dominant. A similar strategy could facilitate transition between state with and without light touch feedback. The body sway overshoot observed in the tactile transition studies following removal of the contact might be a consequence of abrupt release of postural stiffness when implementing a central postural set without tactile feedback. The observed return to baseline no contact sway is then implemented in the following seconds.

An interpersonal haptic interaction may then present itself as a special case where postural stiffness interventions are applied more frequently in order to avoid upsetting the current state of interpersonal postural coordination. In the case of light interpersonal touch for balance support, preventing a disturbance of the haptic signal or even a perturbation of the stability of the interaction partner may be a high-ranking goal of the joint postural task in order to achieve the predominant objective of keeping light interpersonal touch constant. This phasic postural stiffening may be one reason, why steady-state interpersonal touch does not result in constant but changing lead-follower relationships between the two interaction partners in non-dynamic postural interactions.

In contrast to states of quiet standing without specific postural task-relevant instructions, we have observed clear leader-follower relationships in situations, where either the contact receiver (Steinl & Johannsen, 2017) or the contact provider (Steinl et al., 2018) were explicitly instructed to perform a secondary task, for example, maximum forward reaching of the contact receiver in Steinl and Johannsen (2017) and the passive stability support of the contact receiver’s body sway by the contact provider (Steinl et al., 2018). In all three paradigms, we found a clear lead of the contact receiver by the provider of the contact taking the follower role. The communality in these situations was that the contact receiver was only able to predict the contact receiver's motion to a marginal degree and therefore had to adopt a more reactive mode of response. Potentially, the dominance of visual feedback involved may have cued a more consciously controlled action context for the contact provider and therefore delayed postural response times. Returning to the issue of phasic postural stiffness increases during the described tactile interpersonal postural interactions, it seems more likely that the contact receiver was more inclined to reduce sway proactively to enable easier provision of touch by the partner, while the provider would continuously require more postural flexibility.

The proposed "dual-mode" model of postural control during haptic interactions with environmental factors is a generalization and possibly extension of the account of supra-postural tasks imposing task-relevant constraints on the postural control system. This account is, however, not limited to precision requirements affording increased accuracy in a non-postural task and therefore requiring cognitively controlled behaviour, which may be facilitated by the postural system. For example, the flexibility but also context dependency of light touch utilization was illustrated in Kaulmann et al. (2018), where it did not seem to matter in terms of body sway reduction, whether the difficulty of a visual signal detection task was influenced by the variability of fingertip light touch directly or whether difficulty in the visual task depended on body sway so that light touch had only an indirect influence. In the first condition, body sway may have served reduction in variability of the contact force at the fingertip, while in the second condition light touch may have facilitated body sway. Importantly, the mere presence of fingertip light touch without any functional coupling to the visual detection task did not result in similarly reduced sway. Put differently, any relationship to performance in a perceptual task irrespective of the functional hierarchy of body sway control resulted in sway reduction below the level achieved by the presence of light touch alone.

The account of light touch control of body balance, that I am therefore proposing, more generally considers the context-specific valence of certain features of haptic interactions as the major factor influencing the selection of a postural control strategy. What I intend to say by this is probably better illustrated by the deliberately light interpersonal touch during walking paradigm (Schulleri et al., 2017). When walking in a pair with an individual with CP especially, the contact provider has to meet a challenge, in which local precision needs to be sacrificed in favour of global variability. In other words, in this situation the contact provider needs to increase their motion variability, for example in terms of continuously adjusting the relative position of the contacting arm and hand to the trunk or head motion of the contact receiver, in order to "hit a moving target" when applying the haptic contact. Thus, precision with respect to the applied tactile contact as the most relevant action goal cannot be achieved by reduction of motion variability alone, for example increased stiffness, but needs to be ascertained by an opposite strategy of reduced stiffness, that is increased compliance to the motions of the contact partner. To drive the point home once more, light touch interactions with the environment in a balance context are not just means to augment self-motion detection and feedback control of body sway. Light touch involvement creates a new postural context, from the details of which the postural control system will infer how the application of light touch is assisted and whether and how it might assist body sway control. Future studies should more systematically investigate the factors affecting light touch usage, that might be called cognitive, such as participants' expectations and beliefs and the effects of instructions.

In specific circumstances, light interpersonal touch may be a promising strategy for patient guidance in clinical settings. For example, manual support provided by healthcare professionals to stroke patients during training of walking is a routine activity in physical therapy. The specific features of these manual support techniques, however, are usually not elaborated in much detail. The Liten Up patient handling approach promoted by the Accidents Compensation Corporation of New Zealand (ACC, 2003) is probably a very good example of an empirically evaluated training programme aiming to reduce the physical load imposed on the carer by a patient's body weight during today-to-today interactions. While such a training programme may be effective as a combined package in reducing the frequency of musculoskeletal injuries in carers, the individual handling procedures themselves are not backed by empirical evidence and underlying assumptions are often based on "common sense" insights. On the other hand, manual therapeutical approaches such the Alexander Technique and the Feldenkrais Method (Jain et al., 2004) or the Kinaesthetics care conception for nurses (Betschon et al., 2011; Gattinger et al., 2017; Hantikainen et al., 2006) comprise interpersonal movement perception and communication via touch and the haptic modality. Clinical applications of the deliberately light interpersonal touch approach for balance support share the philosophy that softer tactile interactions in contrast to mechanically more demanding grasping and holding may be facilitative to the role of the therapist or carer. In contrast, however, the deliberately light interpersonal touch approach investigated in our present of previous work represents an empirical bottom-up approach that is more concerned with the social and movements dynamics of the haptic interactions between two individuals and the mechanisms that drive the spontaneous behaviours. In the future, this approach might be evaluated not only from a purely movement science perspective but in the light of its therapeutical application, for example in terms of patients' motor learning and functional independence.

Involvement of cortical processes

Not identifying a cortical region, which when disrupted by transcranial magnetic stimulation altered the processing of light touch feedback for balance control, was a disappointment. Of course, we would expect that disruption of the primary somatosensory areas would ameliorate the effect of light touch on balance control due to simple sensorial degradation, but this has not been tested yet. Instead, our main research interest was aimed at any context-specific touch-integrating brain regions. The underlying assumption was that light touch contact with an external reference would cause an interpretation of body sway in spatial terms within an ego-centric frame of reference. Local contact forces, as expressed for example by skin stretch at the contact, were supposed to be perceived in the context of the proprioceptive chain of the contacting limb distally to proximally, such as from the fingertip to the shoulder. Studies of demonstrated the light touch utilization for balance control seems to establish a local, possibly trunk-centred, frame of reference directly associated to the touch contact location (Assländer et al., 2018; Franzen et al., 2011). Findings reported by Dupin et al. (2018) pointed in a similar direction, that a redial trunk-centred frame of reference is involved in the integration of tactile stimulation at the hand and hand motion. It seems, however, that body sway velocity and direction with light touch in terms of a spatial relationship to the contact location are not represented in the parietal lobes. These observations contrast with suggestions that the parietal lobes are generally involved in remapping touch locations into an external spatial reference frame (Azanon et al., 2010). However, it seems that the reference frame underlying perception of touch location is influenced by whether or not voluntary head motion is involved. Moving the head imposes a head-centred frame of reference, while if the head is kept stationary, a body-centred reference frame seems to dominate for touch localization (Pritchett et al., 2012). Serino et al. (2015) postulated diverse reference frames for the representation of peripersonal space depending on the specifically involved body segment and motor actions. Therefore, one could argue that the instructions to keep the head in a fixed posture on the trunk in combination with utilization of light touch for balance control invokes a body-centred reference frame, that may not be represented in spatiotopic terms within the posterior parietal cortex but in terms of somatotopic relationships within other cortical regions, such as the anterior, middle and posterior insula regions. For example, Simmons et al. (2013) reported that activity in the insula regions may integrate attentional exteroception, somatosensory interoception and emotional awareness. Similarly, Stern et al. (2017) suggested that interoceptive sensibility is determined by the activity in several brain regions monitoring the body's autonomic homeostatic state, such as the insula, sensorimotor regions, the occipital cortex, and limbic areas. Perhaps, these areas generalize their functional role to represent also how external and internally generated forces acting onto the body influence its postural equilibrium state. These bodily state representations could include cutaneous somatosensory information from the fingertips or other skin portions within the context of the postural situation to monitor the body's current state of stability.

An unexpected finding was that disruption of the right-hemisphere posterior parietal cortex by transcranial magnetic stimulation with a continuous theta burst stimulation protocol resulted in generally reduced body sway irrespective of the availability of fingertip light touch (Kaulmann, Hermsdorfer, et al., 2017). The causes of why right-hemisphere posterior parietal cortex disruption would lead to increased postural stiffness is not straightforward to understand. A possible interpretation would be spontaneously postural stiffening due to agonist-antagonist co-contractions resulting in less sway variability. In the follow-up study by Kaulmann et al. (2020), we hoped to test this hypothesis directly by the application of mechanical balance perturbations. Our expectation was that increased postural stiffness should lead to shorter displacement amplitude and faster postural stabilization, but we did not observe these changes after PPC disruption. Instead, we observed altered intra-session adaptation to the mechanical perturbation (Schmid & Sozzi, 2016). Assuming that cTBS disruption of the rPPC was effective and that our observations were not random phenomena, one could draw the conclusion that the role of rPPC involvement on balance control is rather context- and task-specific. In both studies, rPPC disruption seemed to "better" postural performance with respect to reduced body sway (Kaulmann, Hermsdorfer, et al., 2017) and higher adaptation rate to a mechanical perturbation (Kaulmann et al., 2020). In more general terms, it therefore seems that the rPPC may exert an influence, which enables greater flexibility of the postural control system. A role of the rPPC in postural adaptation was also suggested by Young et al. (2020), who demonstrated a reduced postural lean after-effect on an inclined surface following non-invasive PPC stimulation by transcranial direct current stmulation (tDCS). Thus, while an obvious connection does exist between postural joint stiffness regulation and functional circuitry of the basal ganglia and the cerebellum (Diener et al., 1989; Diener et al., 1992; Hemami & Moussavi, 2014), it appears that the rPPC provides a modulatory signal, which alters postural adaptability according to the current situational demands. Nevertheless, at this point this interpretation is purely speculative and further, more-targeted and theory-guided research is necessary to gather the appropriate evidence.

*Well, … s*till many questions regarding the mechanisms behind light touch balance control remain unanswered and beg theory-guided and more targeted research efforts. It is my hope that this treatise provides sufficient background information, suggests opportunities and provides incentives for the continuation of this line of research. From my point of view, the "translational gap" between research into the fundamental principles of light touch processing in complex situations and applied research into the development of assistive devices to facilitate behavioural performance in similarly complex situations for instance is very narrow. This means that societal impact for the betterment of the living conditions and quality of life can be achieved in a straight forward manner, for example in the interest of the elderly or neurological patient populations with movement disorders.

9 Summary

This Habilitation treatise has summarized one-and-a-half decades of post-doctoral research into the effect of haptic interactions with the environment on the control of body balance in humans. It covers four areas of research, from single person light touch utilization for balance via light interpersonal touch and its clinical application to human-robot interactions. The current understanding of human balance control is described in detail to establish the theoretical and empirical background, which forms the basis of the interpretation of the light touch phenomena in the balance domain. The treatise ends with an integrative discussion of the "mechanisms of action" and implications of light tactile support of body balance control.

10 Dedication

I have received scientific, technical and emotional support by a number of individuals in those years during which the research presented above has been conducted. Therefore, I like to state that I feel indebted from the core of my heart to all students, colleagues and collaborators, friends and members of my family, who have accompanied me and enabled my research throughout all those years. In order to protect their identity, I am not listing all of them with their individual names. I like to mention, however, that I owe special gratitude to both my mentors Prof. Alan Miles Wing (University of Birmingham) and Prof. Joachim Hermsdörfer (Technical University of Munich), who provided invaluable advice and professional guidance throughout those years that I was a member of their labs and beyond.

11 Cumulus and formal criteria for a cumulative habilitation

Guidelines for the preparation of a cumulative habilitation at the Department of Sport and Health Sciences were adopted by the department council on 08 March 2016. In the following, I describe how my habilitation treatise meets the relevant criteria.

The present cumulative habilitation treatise consists of the following 17 original, peer-reviewed papers. Nine are first authorships, 7 are last authorships and one paper is a middle authorship. All papers have been published in international, english-language, peer-reviewed journals regarded as influential in the field and that have high scientific standards. None of the publications were either related to or part of my doctoral dissertation. All papers are related to the higher-level topic of light touch control of body balance and represent a progression of specific research questions into this topic. The implications of this research programme are also related to the neurophysiology of balance control in general, social behavioural neuroscience and interpersonal interactions, the therapy and manual handling of patient populations as well as human-robot interactions.

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13 List of abbreviations

2PD	Two-point discrimination
CNS	Central nervous system
CoM	Centre-of-Mass
CoP	Centre-of-Pressure
CP	Cerebral palsy
CT	C-type tactile
cTBS	Continuous theta burst stimulation
IPM	Inverted pendulum model
IPG	Inferior parietal gyrus
EEG	Electroencephalography
EMG	Electrical myography
FBI	Full body illusion
fMRI	Functional magnetic resonance imaging
MSD	Mass-spring-damper
NIRS	Near-infrared spectroscopy
PET	Positron emission tomography
PID/PD	Proportional-integral-derivative/proportional-derivative
PIVC	Parieto-insular vestibular cortex
PPC	Posterior parietal cortex
SVV	Subjective visual vertical
TD	Typical development
tDCS	Transcranial direct current stimulation
TMS	Transcranial magnetic stimulation

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Effects and after-effects of voluntary intermittent light finger touch on body sway

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ABSTRACT

Effects of light touch on body sway have usually been investigated with some form of constant contact. Only two studies investigated transient sway dynamics following the addition or withdrawal of light touch. This study adopted a paradigm of intermittent touch and assessed body sway during as well as following short periods of touch of varying durations to investigate whether effects and after-effects of touch differ as a function of touch duration. In a modified heel-to-toe posture, 15 blindfolded participants alternated their index finger position between no-touching and touching on a strain gauge in response to low- and high-pitched auditory cues. Five trials of 46 s duration were segmented into 11 sections: a 6-s no-touching period was followed by five pseudo-randomly ordered touching periods of 0.5-, 1-, 1.5-, 2-, and 5-s duration, each of which was followed by another 6-s no-touching interval. Consistent with previous research, compared to no-touching intervals sway was reduced during touch periods with touch durations greater than 2 s. Progressive reductions in sway were evident after touch onset. After touch withdrawal in the 2-s touch condition, postural sway increased and returned to baseline level nearly immediately. Interestingly, in the 5-s touch condition, reductions in sway persisted even after touch withdrawal in the medio-lateral and antero-posterior plane for around 2.5 s and 5.5 s, respectively. Our intermittent touch paradigm resulted in duration-dependent touch effects and after-effects; the latter is a novel finding and may result from a more persistent postural set involved in proactive sway control.

1. Introduction

Lightly touching a reference object with the finger tip reduces postural sway even though the level of contact force is not sufficient to provide mechanical support [1]. It has been proposed that cutaneous afferent information from the contact provides cues that indicate own body sway [2]. Numerous studies have investigated the nature of this touch effect [3–12]. However, previous studies on the effect of light skin contact on body sway have focused on steady state contact only; except two [13,14] studies have probed the time course of body sway subsequent to touch onset or withdrawal.

The postural control system reweights all available sensory channels in order to optimize the sensorimotor control of stance in altered sensory environments [15]. Gain of a sensory channel is dynamically adjusted depending on a current estimate of its reliability as a reference for own body motion [16,17]. This dynamic function of gain adjustment is non-linear with regard to sensory perturbations [18,19]. Fast adaptation of the postural control system to the addition or withdrawal of light touch is critical in real life situations, as we may face intermittent availability of a support such as a handrail or furniture when moving through our environment. It is therefore important to study stabilization effects and after-effects of intermittent touches with varying durations, in order to see their impacts on postural control.

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Postural stabilization with finger tip tactile feedback has been shown to be a fast process. Rabin and colleagues [13] probed the time course of the light touch effect with a paradigm where finger tip light touch had to be established abruptly. They reported that upon contact body sway is exponentially reduced with a time constant of 1.6 s. In a more recent study, Sozzi and colleagues [14] adopted a paradigm with actively as well as passively initiated, abrupt addition or withdrawal transitions of visual or haptic afferent information. In the active transitions of haptic cues from no-touch to touch, they reported a latency of the onset of sway decrease of around 1.3 s with a time constant of 0.8 s. With regard to an after-effect following touch withdrawal, they observed a shorter latency of the onset of sway increase of just 1 s with a time constant of 0.8 s.

What these two studies above did not investigate, however, was whether the duration of touch exposure affects the dynamics of its after-effects on sway. Therefore, the aim of our current study, with an intermittent touching paradigm, was to investigate changes of body sway during as well as following short periods of touch of varying durations: 0.5, 1, 1.5, 2, and 5 s.

Based on previous studies [13,14], we expected that light touch contact is required to last between 1.5 and 2 s before a reduction in sway will become visible. Sozzi et al. [14] documented that a finite amount of time is necessary for central integration process after transition of touch contact. During this time the touch signal has to pass through several stages of processing [20], in which the signal must be disambiguated within the specific postural context and interpreted in an egocentric frame of reference. If postural adjustments follow the force signal by approximately 300 ms [2,21], it is reasonable to assume a period of 150–200 ms signal processing within supraspinal circuits. Based on the findings of Sozzi et al. [14], we assumed that sway would return to baseline levels following withdrawal within a time frame similar or shorter than the time required to integrate the touch signal.

2. Materials and methods

2.1. Participants

Fifteen healthy adults (eight females and seven males; average age 20.6 SD 2.64 years) gave their written informed consent, as approved by the Institutional Review Board of Chung Shan Medical University Hospital, to participate in the study. All of them were right-handed and reported no musculoskeletal and neurological abnormalities that could have influenced their standing balance.

2.2. Apparatus

A force plate (Bertec FP4550-08, USA) measured the six components of the ground reaction forces and moments to determine the medio-lateral and antero-posterior components of Centre-of-Pressure. A dual-axis strain-gauge (RMAX SN110336-1, Taiwan), which measured normal and lateral shear forces, formed the circular touch plate (5 cm diameter) with a smooth surface. In response to a high-pitched or low-pitched auditory cue, participants either made fingertip contact with the touch plate, mounted on a stand at waist level to the front right of the participants, or withdrew contact from the plate. Three infrared cameras (MotionAnalysis HAWK, USA) captured the motion of two reflective markers, one placed on the tip of participant's index finger and one on the edge of the touch plate. All signals were sampled at 100 Hz.

2.3. Procedure

Participants were asked to hold their index finger of the dominant hand above the touch plate while keeping the outstretched arm in a comfortable posture. Participants stood with bare feet in a modified heel-to-toe stance (the non-dominant heel touching the side of big toe of the dominant foot). Participants were then instructed to close their eyes, and to stand relaxed but as still as possible without speaking.

A single trial lasted for 46 s and consisted of a 6-s no-touching period (1st NT) followed by five touching periods of 0.5-, 1-, 1.5-, 2-, and 5-s duration (0.5, 1, 1.5, 2, and 5 T) in a pseudo-randomized order. Each of the five touch periods was followed by a 6-s no-touching period (2nd to 6th NT). The beginning and end of each trial was cued separately to indicate the starting and ending of data collection.

Trials were started when participants were ready. On hearing a high-pitched tone, participants flexed their index finger at the metacarpal-phalangeal joint to initiate light finger contact. On a low-pitched tone, participants extend their index finger just above the touch plate. Practice trials familiarized participants with the experimental protocol. Participants performed five standing trials and were allowed to rest for 30 s between trials.

2.4. Data analysis and statistics

All data underwent low-pass filtering with second-order Butterworth filter and 6 Hz cut-off frequency. According to the vertical touch force detected by the strain gauge, the onset and offset of each touching period was determined. Afterwards, data were divided into bins of 500 ms duration in order to standardize the number of data points for the sway measure extraction in different duration conditions. Due to the narrow bin width, we chose to analyse sway in terms of Centre-of-Pressure velocity (dCOP) as its variability measure would be less susceptible to voluntary low frequency drift than COP position. Also, a velocity-dependent signal resembles postural control better than position or acceleration under experimental conditions of sensory manipulation [22]. The standard deviation (SD) of dCOP in the medio-lateral ($dCOP_{ml}$) and antero-posterior ($dCOP_{ap}$) directions were calculated separately for the respective bins of interest and averaged for each duration condition across each of the five trials of a participant.

Using statistical software (SPSS 18.0, Chicago, IL, USA), firstly, we examined whether the recurring light touch would result in accumulated effects across a trial despite the interruptions. The change of sway across the no-touching periods irrespective of the inserted touch duration conditions, i.e., the last seven bins of the 1st NT and the first seven bins of the 2nd to 6th NT, was examined by two-way ANOVA (bin $\times$ sequence). Secondly, in order to examine touch effects two-way ANOVA (transition $\times$ duration) was conducted to compare the second to last NT bin before touch onset and the last bin of each touch duration conditions (0.5, 1, 1.5, 2, and 5 T). The bin just before touch onset was not chosen because during this bin the high-pitched cuing tone had occurred and the touching movement was in preparation. ANOVAs were followed up with simple contrasts to examine touch effects within each touch duration condition. Furthermore, the touch effects were fitted with linear regressions as a function of the five non-linear touch durations. Finally, for the specific duration conditions with significant touch effects, sway evolution after touch onset and withdrawal was evaluated by comparing the values of the respective touch bins with the 99% confidence interval (CI) of the first 11 bins of the 1st baseline NT. The significance level was set at 0.05.

3. Results

Overall, 52 out of 375 touch sections were excluded from data analysis, among which 21 had an average touch force greater than

1.4 N, 27 had an actual touch duration that deviated by more than 200 ms from the experimentally set duration (i.e., severely delayed response latencies to the auditory cue). In four touch sections the finger accidentally missed the contact plate when touch had to be established. The mean vertical contact forces were 0.67 N with SD 0.32 N. Delays due to participants' latencies in response to the auditory cue meant that the actual touching periods were slightly shorter or longer than the set periods. The actual duration for the conditions of 0.5 T, 1 T, 1.5 T, 2 T, and 5 T were 576 ms (range 480–700), 1012 ms (range 880–1140), 1540 ms (range 1450–1680), 2041 ms (range 1980–2140), and 4992 ms (range 4870–5050), respectively. Fig. 1 represents the dCOP fluctuations and touch force components of a sample trial.

Fig. 2 illustrates SD of $dCOP_{ml}$ and of $dCOP_{ap}$ for each of the six no-touching periods irrespective of the touch duration conditions. Postural sway on the medio-lateral axis was slighter greater and more variable than on the antero-posterior axis due to the modified heel-to-toe stance. The two-way ANOVA revealed no significant main effect of bin ($dCOP_{ml}$ $F_{6,84} = 0.785$, $p = 0.512$; $dCOP_{ap}$ $F_{6,84} = 1.552$, $p = 0.227$), suggesting that the 6-s no-touching periods in between touching periods were long enough for resetting of sensory integration. No main effect of sequence ($dCOP_{ml}$ $F_{5,70} = 0.58$, $p = 0.582$; $dCOP_{ap}$ $F_{5,70} = 1.759$, $p = 0.133$) was shown, signalling no accumulated effect of the history and number of previous intermittent touch periods.

The two-way ANOVA on the touch effects in the medio-lateral plane (Fig. 3, left panel) revealed a main effect of transition ($F_{1,14} = 5.889$, $p = 0.029$, partial $\eta^2 = 0.296$). No other effects were found. The simple contrasts revealed significant touch effects for 2 T ($F_{1,14} = 7.244$, $p = 0.018$, partial $\eta^2 = 0.341$) and 5 T ($F_{1,14} = 5.064$, $p = 0.041$, partial $\eta^2 = 0.266$). Compared to the second to last bin in the preceding NT, postural sway in the last bin of 2 T and 5 T decreased by 17.5% and 18.0%, respectively. The averaged data give hint of a trend towards further sway reductions with increasing touch duration, and the linear regression fitting indicated a slope of -0.093 cm/s ($p = 0.067$, $R^2 = 0.013$).

The two-way ANOVA on the touch effects in the antero-posterior plane (Fig. 3, right panel) revealed a main effect of transition ($F_{1,14} = 9.912$, $p = 0.007$, partial $\eta^2 = 0.415$). No other effects were found. The simple contrasts revealed borderline touch effects for 2 T ($F_{1,14} = 3.774$, $p = 0.072$, partial $\eta^2 = 0.212$) and significant touch effects for 5 T ($F_{1,14} = 9.405$, $p = 0.008$, partial $\eta^2 = 0.402$). Compared to the second to last bin in the preceding NT, postural sway in the last bin of 2 T and 5 T decreased by 17.8% and 26.1%, respectively. The averaged data give hint of a trend towards further sway reductions with increasing touch duration, and the linear regression fitting indicated a slope of -0.069 cm/s ($p = 0.041$, $R^2 = 0.017$).

As for 2 and 5 T touch effects and after-effects are shown for both $dCOP_{ml}$ and $dCOP_{ap}$ in Fig. 4. The evolution of sway across the respective bins is compared to the 99% CI of the baseline NT. Progressive reductions in sway were evident after touch onset (Fig. 4, left panels). Both the 2 and 5 T duration conditions have decreased below the 99% CI over the time course of the 0.5- to 1.5-s bin after touch onset and both duration conditions progress in parallel to the 2-s bin after which the 2 T condition ceases. After touch withdrawal (Fig. 4, right panels), we see a sudden increase in sway in the first 0.5-s bin as compared to the last bin during touch in both duration conditions. However, the gradual increase in sway does not mirror the time course of sway reduction after touch

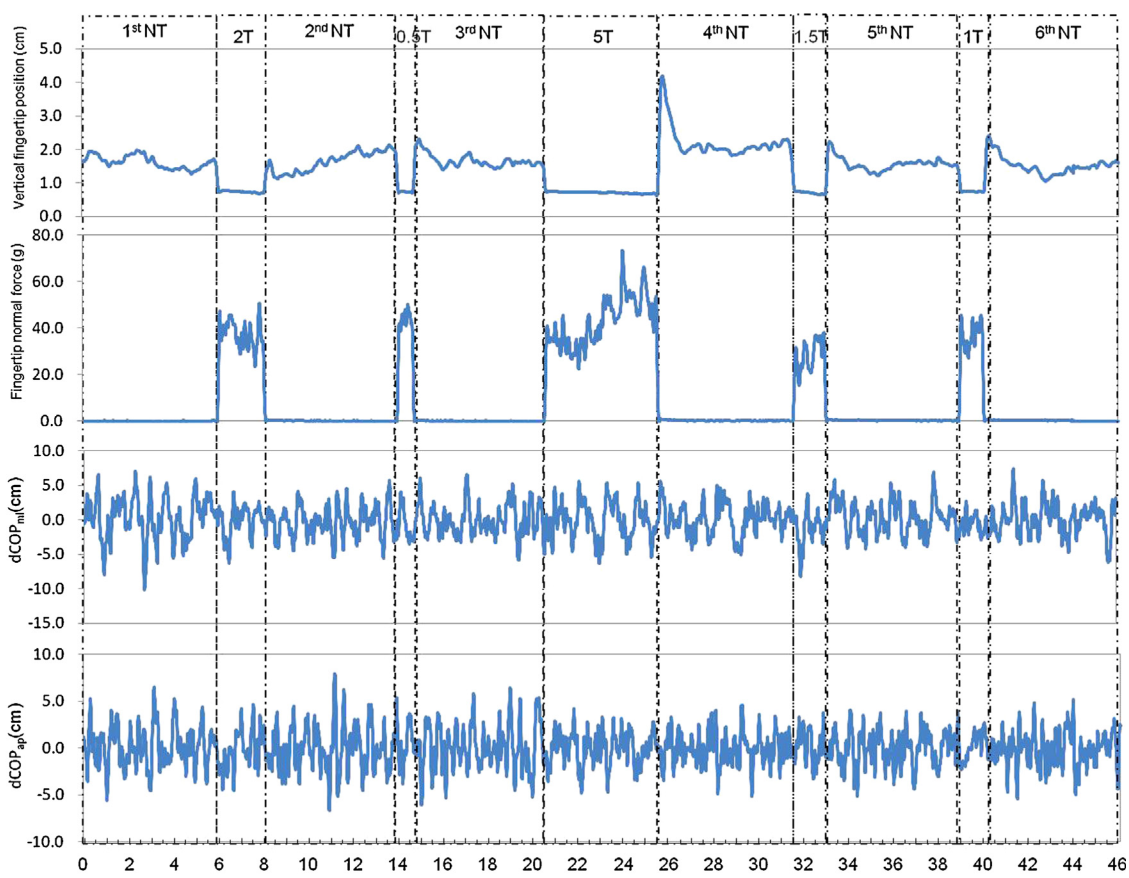

Fig. 1. A representative trial showing vertical fingertip position, fingertip normal force, and Centre-of-Pressure velocity in the medio-lateral ($dCOP_{ml}$) and antero-posterior plane ($dCOP_{ap}$) during 46 s. The 46-s trial was consisted of five touching periods of 0.5-, 1-, 1.5-, 2-, and 5-s (0.5, 1, 1.5, 2, and 5 T) running in pseudo-random order, and six 6-s no-touching periods in between (NT).

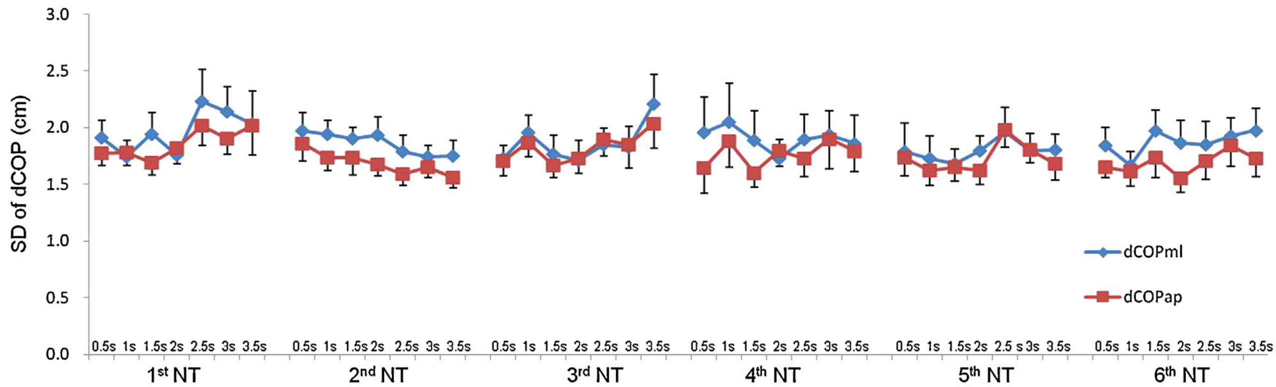

Fig. 2. The mean and between-subject SE of SD of $dCOP_{ml}$ and of $dCOP_{ap}$ as a function of sequence during no-touching periods, i.e., the last seven bins of the 1st NT and the first seven bins of the 2nd to 6th NT periods.

onset as the two duration conditions do not progress in parallel anymore. After touch withdrawal, in 2 T postural sway increased and returned to baseline level immediately in the medio-lateral plane and after 0.5 s in the antero-posterior plane. Interestingly, in 5 T sway had returned to baseline level for the 3-, 4- and 4.5-s bins after touch withdrawal but dropped below baseline at the 5.5-s bin in the medio-lateral plane. In the antero-posterior plane in 5 T, sway remained below baseline except for the 3-s bin. Bonferroni-corrected directed-hypothesis post hoc single comparisons for the 2 and 5 T conditions ($\alpha < 0.0045$; 0.1/22 = 0.0045) indicated that for 5 T exclusively were bins significantly below baseline: in the medio-lateral plane the 2-s bin and in the antero-posterior plane the 0.5- and 1-s bins.

4. Discussion

Up to now, only two studies [13,14] assessed the transient response of light touch on the control of body sway by adopting a paradigm of abrupt addition or withdrawal of haptic information of long durations. In our present study, using an intermittent touching paradigm, we examined how a train of short duration light touch periods alters sway. Our results show progressive reductions in sway after touch onset in the 2 and 5 T duration conditions, following an exponential decrease reaching asymptote within 2–3 s after onset and thus are in good accordance with the exponential decay functions previously reported [13,14]. Postural sway decreased to lower than 99% CI of baseline no-touching intervals after 0.5 to 1 s after touch onset. At the final bin, sway reductions amounted to 17.5–17.8% with exposure duration of 2 s (actual range 1980–2140 ms) and by 18.0–26.1% with exposure duration of 5 s (actual range 4870–5050 ms).

Inconsistent with our prediction, the time course of sway increase after touch withdrawal did not mirror the inverted sway reduction following touch onset. Sway did not return to baseline levels within the same time frame but after-effects were shown for the 5 T condition, which lasted for up to 5.5 s in contrast to the 0.5–1 s period observed for sway reduction after integration of touch information. As a qualitative observation, only 5 T showed bins that were significantly below baseline after Bonferroni-correction. This suggests that the reduction in sway with light touch does not depend on the constant presence of a haptic force signal but can be upheld for an additional amount of time. This finding is in contrast to the report of Sozzi and colleagues [14] who observed a shorter duration of sway increase after touch withdrawal compared to sway reduction after touch onset. We believe that this discrepancy rests on differences in the adopted paradigms, i.e., train of intermittent short touch durations versus touch section with durations of 30 s and longer.

These results express two interesting insights with respect to the 5 T condition. The first is that the 2-s duration, although effective in reducing sway, is not of sufficient duration to induce touch after-effects. That the after-effects occur in the 5-s duration could due to the postural control system requiring up to 5 s for establishing a postural set adjusted to the requirements of light touch contact. The second insight is perhaps, that this light touch postural set is kept online depending on the context of the sensorimotor task. Knowing that a period of light touch will be followed by a long no contact interval as in the study by Sozzi et al. [14] might lead to a rapid taking offline of the light touch postural set in order to optimize sensorimotor gains in contrast to the expectancy that it will be still required in the immediate future as in our intermittent touch paradigm. Dealing with intermittent touch intervals might result in the postural control system to adopt a more conservative sensorimotor control strategy for an intermediate time frame with the consequence of persisting lowered sensorimotor gain for the other sensory channels involved

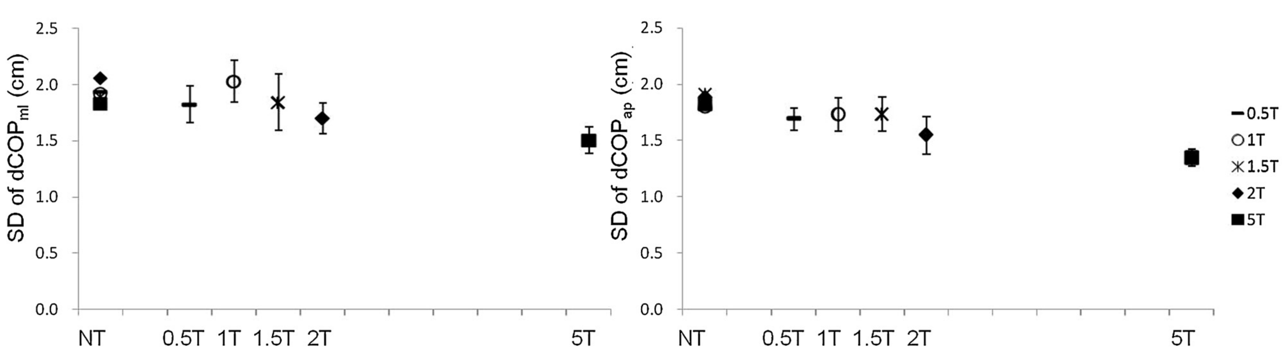

Fig. 3. The transition of SD of $dCOP_{ml}$ (left panel) and of $dCOP_{ap}$ (right panel) from no-touching (the second to last bin just before touch onset) to the steady state of touching (the last bin of each touching condition).

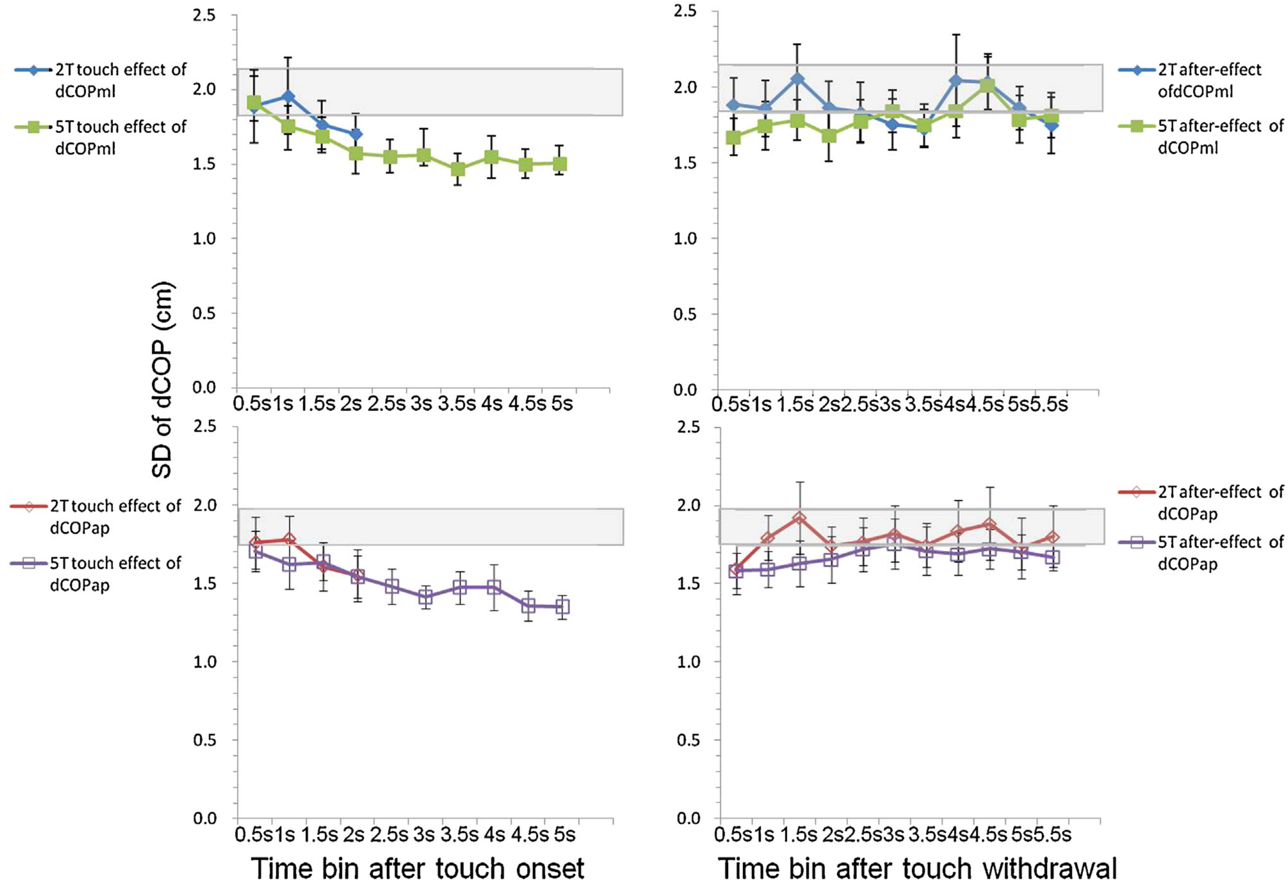

Fig. 4. The changes of SD of $dCOP_{ml}$ (upper panel) and of $dCOP_{ap}$ (lower panel) across the 500 ms bins after touch onset (left panel) and withdrawal (right panel) for the 2 and 5 T conditions. The grey areas indicate 99% CI of postural sway during baseline no-touching interval. Between-subject SE bars are shown.

in standing balance (i.e., vestibular, plantar somatosensory and muscle proprioceptive afferences). For example, Jeka et al. [23] demonstrated faster down-weighting and slower up-weighting of the gain of the visual channel in response to transient changes in amplitude of a wide-field oscillatory visual motion stimulus. They interpreted the longer duration of up-weighting the visual channel as a conservative postural control strategy when confronted with a sensory environment featuring regular transient changes.

On the other hand, it does not appear that participants chose constant multimodal sensory gains across an entire trial as sway in the no contact periods did differ as a function of the preceding touch duration. This means that any subsequent trains of touch durations shorter than 5 s, interrupting the no touch periods of 6-s duration, were not considered sufficiently informative in terms of touch feedback utilization and thus did not suggest sustained sensory gain settings. It is also possible that continuous tactile exploration, e.g., 5-s continuous light touch, leads to enhanced cortical excitability [24], and hence its touch effects may last after touch withdrawal during which the enhanced cortical excitability is still maintained. Another mechanism that may account for the after-effects is the constraining effect brought about by the suprapostural touch task. Keeping the contact finger just above the touch plate and ready for establishing the next touch period may itself form a precision task superordinate to the control of sway leading to decreased sway by active sway constraining in the absence of finger tip force feedback [25].

That intermittent touch effects did not accumulate across a trial suggests that the 6-s no-touching intervals between intermittent touches were sufficient for a wash-out and therefore do not cause difficulties when investigating the effects of intermittent touch. Furthermore, the experimental conditions of five different touch durations were performed in random order, and our findings suggested a linear trend towards further sway reductions with increasing touch duration. Therefore, we believe that our findings of the positive postural stabilization effects of intermittent touch would remain if a longer no-touching interval were adopted. However, our study design was limited by the lack of touch duration between 2 and 5 s. Based on our results, future studies might focus on touch durations within but also beyond this range. Further, a systematic variation of the durations of the inter-touch no contact periods will be important to shed more light on the occurrence of longer-duration touch after-effects following touch withdrawal.

A specific aspect of our experimental setup was that the surface of the contact plate was relatively smooth. One could argue that the low friction resulted in exceptionally low shear forces and reduces tactile sensation at the fingertip and therefore somehow affected our results. Jeka and Lackner [26] demonstrated, however, that the light touch effect on sway is not dependent on the contact surface having rough or smooth characteristics.

In conclusion, our new intermittent touch paradigm not only confirms the postural stabilization effects of touch, but also revealed touch after-effects. The postural stabilization effects provided by light touch have been demonstrated on pathological populations who suffered from postural instability due to various etiologies [3–11]. There is a proposition that the paradigm of light finger touch may represent a potential treatment for patients suffering from postural instability [3,8], probably as a sensory prosthesis or due to exercise-related benefits. Our findings of postural stabilization effects during and after intermittent touch provide further insights into this potential treatment paradigm, which may be especially important for patients who are weaning from constant touch during balance retraining.

Conflicts of interest

None.

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Short communication

Disruption of contralateral inferior parietal cortex by 1 Hz repetitive TMS modulates body sway following unpredictable removal of sway-related fingertip feedback

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HIGHLIGHTS

- We examined time course of sway after passive light touch onset and removal transitions.
- We observed involuntary sway overshoot after light touch removal.
- rTMS over left-hemisphere inferior parietal gyrus reduced sway overshoot after light touch removal.

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ABSTRACT

Contact with an earth-fixed reference augments sway-related feedback and leads to sway reduction during upright standing. We investigated the effect of repetitive transcranial magnetic stimulation (rTMS) over the left hemisphere inferior parietal gyrus (IPG) as well as middle frontal gyrus (MFG) on the progression of sway following right-hand finger tip contact onset and removal. In two experimental sessions, 12 adults received 20 min of 1 Hz rTMS stimulation at 110% passive motor threshold over the left MFG and left IPG, respectively. Before and after each stimulation interval, participants' body sway was assessed in terms of antero–posterior Center-of-Pressure (CoP) velocity. Passive touch onset and removal were timed at random intervals by controlling the vertical position of a contact plate. Progression of sway was evaluated across 6 s before to 6 s after each contact event. Following both contact onset and removal, a temporary increase in sway above baseline without contact was observed. After removal overshoot was especially prominent. While steady-state sway was not altered by stimulation, rTMS over the left IPG reduced overshoot compared to pre-stimulation; thus, improving sway progression on haptic deprivation. We discuss our finding in the light of altered transient postural disorientation due to intermodal sensory conflict, illusion of backwards falling and tactile attention capture.

1. Introduction

Cutaneous information from the fingertips can be utilized to augment sensory feedback for the control of spontaneous body sway [1]. In order to achieve this, any fingertip directional information needs to be processed in the context of the entire kinematic proprioceptive chain from the distal contact to the proximal segments of the trunk [2,3]. Perhaps due to the greater number of degrees-of-freedom available for arm posture as well as the specific constraints of the postural task, the specific interpretation of fingertip feedback for sway control appears to be computationally more complex than the interpretation of visual stimulation [4].

The middle frontal gyrus (MFG), presumably the dorsolateral prefrontal cortex (DLPFC), of the ipsilateral right hemisphere seems to be involved in the processing of fingertip feedback for sway control. In two studies, Bolton et al. [5,6] demonstrated modulation of cortical somatosensory evoked potentials (SSEPs) when right-hand fingertips signalled sway-related information. The application of continuous theta burst repetitive transcranial magnetic stimulation (cTBS) over the right MFG attenuated this difference in SSEPs between conditions with sway-relevant and irrelevant fingertip afferent information [6]. However, no corresponding changes in spontaneous sway that might indicate altered processing of fingertip feedback for sway control were reported [6].

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It seems unlikely, however, that the ipsilateral, right-hemisphere MFG is the only region involved in sway control with light fingertip contact. Within the hierarchy of the cutaneous sensory system, right-hand fingertip vibratory and light touch stimulation activates first of all the contralateral primary and secondary somatosensory cortices [7] but also the contralateral inferior parietal lobule, presumably due to involved tactile attention [8]. Further, repetitive TMS over the contralateral parietal cortex impairs relative finger position sense of the right hand [9]. In order to gain further insight into the contralateral cortical processing of tactile cues for sway control, we investigated body sway with right-hand fingertip contact as a function of the disruption of two contralateral cortical regions by rTMS. Bolton et al. [5] reported modulation of early SSEP components (P50, P100) within the contralateral inferior parietal gyrus (IPG; CP3) during quiet stance with earth-fixed fingertip contact. Therefore, we expected that rTMS over the IPG would result in performance decrements such as a reduced benefit of fingertip contact on steady-state sway. In addition, we expected that transient sway stabilization following transitions of touch addition or removal would be delayed by rTMS over IPG due to disrupted integration of tactile information. As a control region, we selected the contralateral MFG (F3) of which any involvement in fingertip tactile processing for sway control has not been reported in contrast to ipsilateral MFG [5,6].

2. Material and methods

2.1. Participants

Twelve naive, healthy, right-handed individuals took part (4 females, 8 males, aged between 24 and 41). All gave written informed consent and reported not to take any medication or drugs that might affect cortical excitability or altered cognitive functions. Participants had neither neurological, psychiatric, or other relevant medical diagnoses, nor any other contraindication to TMS. The study protocol was carried out in accordance with the ethical research standards of the declaration of Helsinki and was approved by the Medical Research Ethics Committee of the Technische Universitat München.

2.2. Procedure

Blindfolded participants stood without shoes in quiet but relaxed normal bipedal stance on a force plate (Bertec 4060FP; sampling frequency 600 Hz). The right arm was held in an elbow-flexed posture which enabled the index and middle fingers of the right hand to contact a plate positioned directly in front on the participant's midline. The contact plate was instrumented with a force-torque sensor (ATI Industrial Automation Nano 17; sampling frequency 200 Hz; Fig. 1a) and mounted on top of a vertically oriented linear motor (Firgelli Technologies L12). The linear motor enabled contact to be established and removed automatically from below the finger tips. Participants were instructed not to push actively against the plate but to let contact occur passively by keeping their fingers and hand in place resisting any upward movement only. Acoustic anticipation of touch onset and removal due to actuator noise was prevented by earplugs in addition to ear protectors.

Two experimental sessions were scheduled at least seven days apart. During each session, participants' body sway was assessed before (baseline) and after (post-stimulation) rTMS application (Fig. 1b). At the start of each session, participants performed one practice trial without earplugs and blindfold. For each sway assessment, block 6 trials of 120 s duration were recorded. Each trial contained 5 contact onset-removal pairs in which, at a random time point, the contact surface moved upwards until a switch was activated by finger contact. Following a random contact period of 7–20 s, the surface moved downwards and paused for another random 7–20 s no-contact interval.

2.3. rTMS application

Participants were seated in a reclined chair and instructed to relax with eyes closed. Subjects wore earplugs and a tight-fitting EEG cap (Easy CapR) with the extended version of the international 10–20 system drawn on it referenced to the vertex (Cz). Cz was identified as the intersection of the interaural line and the connection between nasion and inion. A PowerMag stimulator (Mag and More GmbH) connected with a figure-of-eight coil (70 mm diameter) was used for delivering biphasic pulses. Motor evoked potentials (MEPs) were recorded from the right first dorsal interosseus (FDI) muscle by surface electromyography (EMG) using Ag–AgCl conductive adhesive electrodes. The 'motor hotspot' of the FDI muscle representation was identified as the scalp position, where single TMS pulses at mean capacitor output intensity consistently induced MEPs in the relaxed muscle. Using a staircase procedure (T.M.S. Motor Threshold Assessment Tool 2.0) the resting motor threshold (RMT) was determined.

Subsequently, for 20 min, 1 Hz rTMS was applied at 110% of the individual's RMT [10] either over the IPG (CP3) or over the MFG (F3) in the left hemisphere [11]. For IPG stimulation, the coil was placed over CP3 tangentially to the scalp with the handle pointing posterolaterally at a 45-degree angle to the sagittal plane and clamped in place with a mechanical arm [9,12]. For MFG stimulation, the coil was placed similarly, but at a 90-degree angle to the sagittal plane over F3 [13]. To ensure a locally consistent stimulation, the position of the coil was marked on the EEG cap, and the participants' heads were fixed by a stabilizing, pellets filled vacuum cushion wrapped around the neck. The order of stimulation locations was randomly assigned across the two sessions.

2.4. Data processing and analysis

The data of the force-torque transducer were transformed from 200 Hz to 600 Hz and merged with the force plate data. Data were digitally low-pass filtered at 10 Hz (dual-pass, 4th-order Butterworth). Center-of-Pressure (CoP) position was differentiated to yield rate of change parameters (dCoP) in order to remove low frequency drift in CoP and to allow selection of a relatively narrow temporal bin width. The standard deviation (SD) of antero–posterior (AP) dCoP was extracted across 24 bins of 500 ms duration each from 6 s before to 6 s after a contact transition (onset: transition from no contact to contact; removal: transition from contact to no contact). We defined sway within the bin from 1 to 0.5 s before contact onset as baseline sway and used this as a reference for any subsequent changes in sway. Data processing and extraction was conducted by MATLAB (MathWorks, 7.13 (2011b). AP sway progression was then statistically analysed by repeated-measures ANOVAs with (1) rTMS stimulation (baseline and post-stimulation), (2) location of stimulation (MFG and IPG), and (3) touch presence or time course of transient sway stabilisation (time bins) as within-subject factors, respectively (SPSS 20). A p-value of .05 was applied for the evaluation of statistical significance.

3. Results

Fig. 1c shows a sample trial of AP body sway across 120 s for an individual participant before rTMS stimulation. Across all participants, the average peak normal contact force was 1.52 N (SD = .85). Not every participant benefitted from light contact in terms of a reduction in body sway. We performed an outlier analysis by calculating the average steady-state sway reduction between pre-stimulation baseline sway and sway with light contact and determined the thresholds of the 99% confidence interval across all participants and time points. Three participants with no steady-state sway reduction (mean = .57 mm/s, SD = 1.12) were excluded from statistical analysis on the grounds that they did not process fingertip tactile feedback in the expected way. The remaining 9 participants showed an average sway reduction of −1.37 mm/s (SD = .62) with light contact.

3.1. Effects on steady-state sway

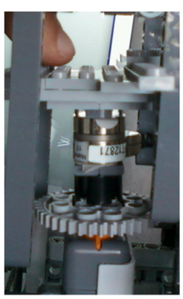
In order to test for steady state sway reduction with finger contact, contrasting body sway within the second-to-last bin (1 s to 500 ms) before touch onset (baseline) to the last extracted bin (5.5–6 s) after onset indicated a significant steady-state sway reduction with fingertip contact ($F(1,8) = 20.66$, $p = .002$, partial $\eta^2 = .72$). Neither rTMS stimulation, stimulation location nor any interaction between main effects were significant. The complementary comparison between the second-to-last bin before contact removal to the last extracted bin (5.5 –6 s) after contact removal indicated a significant increase in sway without fingertip contact ($F(1,8) = 27.28$, $p = .001$, partial $\eta^2 = .77$). Fig. 2 shows relative changes in steady-state sway from before to after contact onset and from before to after contact removal.

3.2. Effect on transient sway following contact onset or removal

Fig. 3a shows the evolution of the sway difference between baseline steady-state sway and sway in each bin from 0.5 to 4.5 s after touch onset for each single assessment block. Sway following touch onset increased until 2 s after touch onset, before it began to settle to levels below baseline ($F(8,64) = 6.85$, $p < .001$, partial $\eta^2 = .46$). Except for the time course, no other main effects or interactions were significant for the touch onset transition.

Regarding the time course of sway following contact removal (Fig. 3b), a rapid increase was observed until peak sway occurred between 2 s and 2.5 s after removal followed by a decrease toward baseline sway without contact ($F(8,64) = 4.85$, $p < .001$, partial $\eta^2 = .38$). In addition, the interaction between rTMS stimulation, stimulation location, and time course were significant ($F(8,64) = 2.48$, $p = .02$, partial $\eta^2 = .24$). Post-hoc comparisons indicated that following IPG rTMS, the increase in sway after contact removal was lower compared to before stimulation at 2.5 s and 3.5 s (both $Fs(1,8) \geq 7.26$, both $ps \leq .03$, both partial $\eta^2 \geq .48$). A tendency was found at 1 s ($F(1,8) = 4.82$, $p = .06$, partial $\eta^2 = .38$). No differences were found when comparing before and after stimulation for MFG rTMS. The evolution of the sway difference between baseline steady-state sway and sway in each bin from 0.5 to 4.5 s after touch removal for each single assessment block is presented in Fig. 3b.

A



B

	Baseline testing	rTMS	Post-stimulation testing
Session 1	6 trials 12 minutes	1 Hz 1200 pulses 20 minutes	6 trials 12 minutes
Session 2	6 trials 12 minutes	1 Hz 1200 pulses 20 minutes	6 trials 12 minutes

C

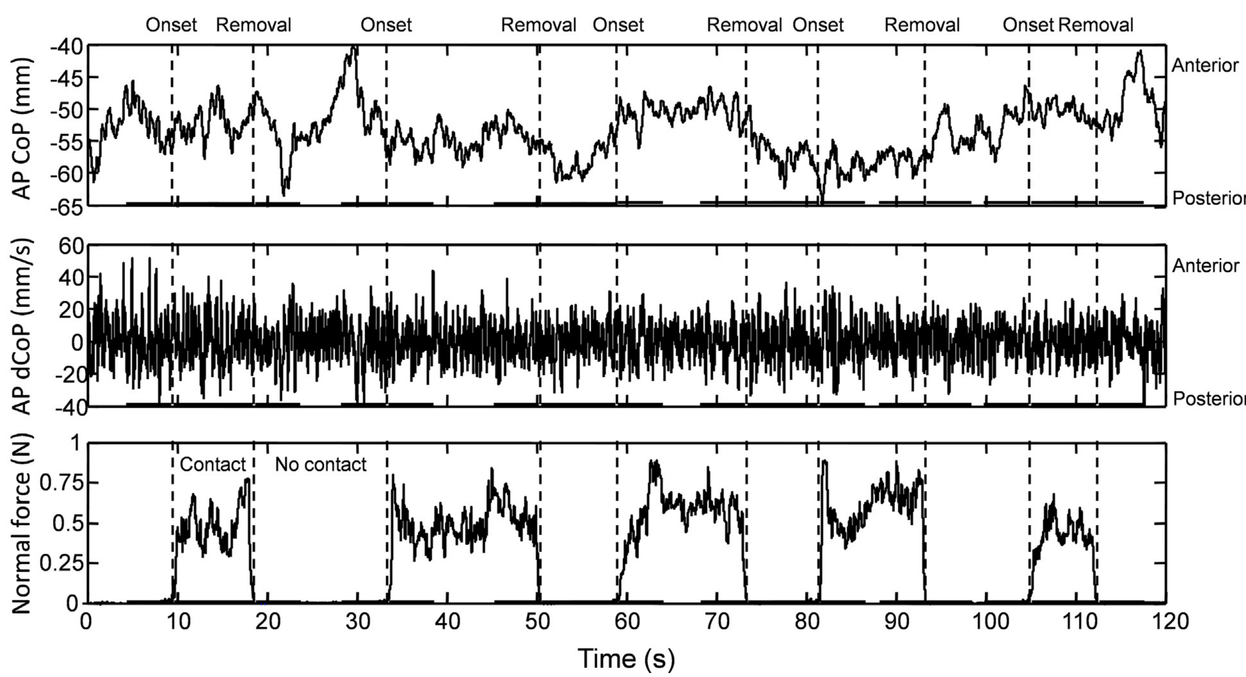


Fig. 1. (A) The fingertip contact plate and force-torque sensor mounted on top of a vertically aligned linear motor. (B) The design of the study. Each rTMS session was performed on a separate day at least a week apart. Order of stimulation locations was randomized across participants. Balance testings took place before and after each stimulation period. (C) A sample trial for single participant. AP CoP position (upper panel), AP CoP rate of change (middle panel) and normal contact force (lower panel) are shown across 120 s. Dashed lines indicate time point of a contact event (onset/removal). Thick black lines indicate the 6 s duration divided into bins of 500 ms duration before and after a contact event.

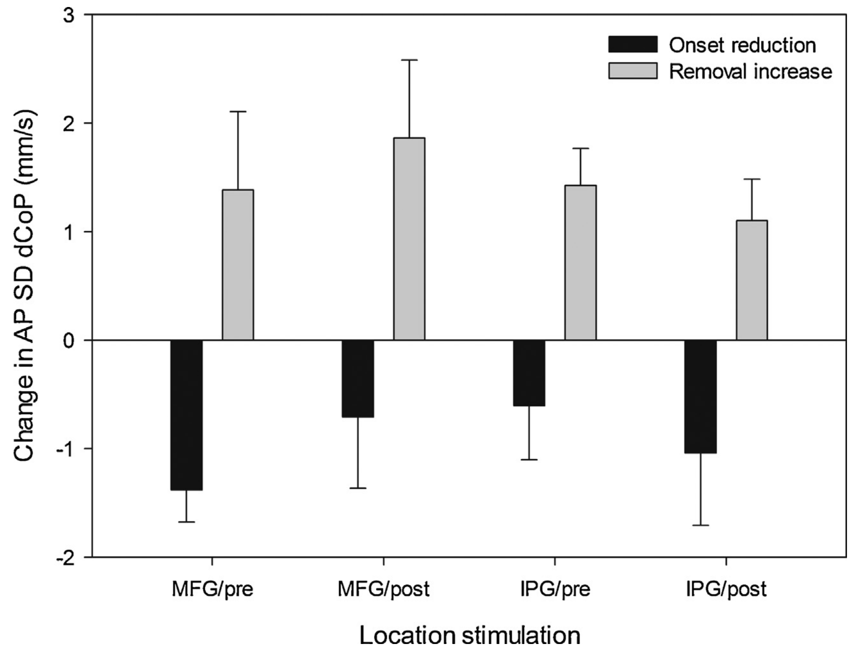


Fig. 2. Difference in steady-state body sway from before to after contact onset (black bars) and from before to after contact removal (light grey bars). Error bars indicate standard error of the mean. MFG: middle frontal gyrus, IPG: intraparietal gyrus, pre: before rTMS, post: after rTMS.

4. Discussions

We expected smaller steady-state sway reduction with light fingertip contact following rTMS over contralateral IPG compared to contralateral MFG. In addition, we assumed that the dynamics of the transient response following unpredictable and abrupt contact onset as well as removal would be altered by IPG rTMS. These expectations were only partially met by our data. Steady-state sway both with or without light contact was not affected by rTMS irrespective of the stimulation location. Similarly, the integration of the tactile signal in terms of the transient response following onset was also not altered by any rTMS condition. Bolton et al. [6] disrupted activation under right-hemisphere, ipsilateral MFG by cTBS and found an alteration of task-specific SSEPs, while actual sway was not affected. We infer that disruption of either the left- and right-hemisphere MFG or the left-hemisphere, i.e., contralateral IPG does not affect steady-state sensorimotor control of body sway. The transient response following contact removal, however, was modulated by IPG rTMS. In general, we observed an increase and overshoot in sway relative to baseline as the main characteristic of the transient responses following contact onset as well as removal. With respect to this phenomenon, rTMS over the left-hemisphere IPG dampened the sway overshoot subsequent to removal transitions.

The initial overshoot might have resulted from abrupt contact causing a slight perturbation to the fingers of the right hand. As peak overshoot did not occur immediately after the contact event, however, the possibility of a mechanical perturbation seems unlikely. Instead, we observed a gradual build-up, which reached maximum within 1–2.5 s following the contact event. Thus, overshoot occurred relatively late in sway progression, and therefore is more likely to express a long-latency process involved in both sway control and tactile perception. Vuillerme et al. [14] demonstrated that utilizing light touch for sway control demands attention and Sozzi et al. [4] argued in favor of increased computational load when integrating or removing light tactile feedback for sway control. In this light, overshoots may express transient capture of attention by the respective tactile event [15], which could have led to short-term degradation of postural control and increased sway. Hagen and Pardo [8] suggested that the IPG is involved in directed tactile attention. Therefore, the reduced overshoot at contact removal following contralateral IPG disruption might be a consequence of reduced tactile attention capture and less interference between tactile perception and sway control. Attentional interference involving the IPG may also explain why Bolton et al. [5] found a suppression of the early P100 component of SSEPs when fingertip feedback signals sway-related information. The implication is that IPG is involved in two distinct functions in the current postural context: directed attention at onset and removal of tactile feedback as well as another function more specific to sway control.

It appears reasonable that sudden deprivation of a strongly weighted tactile signal for sway control leads to acute intermodal conflict between all relevant sensory channels. If the removal overshoot was a direct consequence of the extent of detected intermodal conflict then a lower overshoot might express degraded sensitivity to an actual intermodal discrepancy. Not many studies investigated disorientation following abruptly altered sensory conditions. Peterka and Loughlin [16] demonstrated the emergence of transient, involuntary 1 Hz body oscillations following abrupt cessation of support surface sway referencing, which they attributed to processes of acute sensory reweighting and—integration, which resulted in the production of overcorrective torque. A phenomenon that seems to be exacerbated in older adults during the reintegration of proprioceptive cues after the termination of sway referencing [17] and muscle vibration [18,19]. Similarly, Teasdale et al. [20] investigated the effect of onset and removal transitions in the visual channel (opening and closing the eyes to an acoustic signal) on the change in sway for young and older adult participants. Their data also indicate transient increases in sway subsequent to a transition relative to the later steady-state sway, especially in older adults. As a cause for greater susceptibility in older adults, they regarded reduced effectiveness of central integrative processes for error detection and postural set reconfiguration after altered sensory conditions [19,20].

The left-hemisphere IPG could play a role in the detection of intermodal conflict based on distinct unimodal estimates of self and external motion. A related explanation of our finding is that the sudden disappearance of fingertip feedback could be interpreted by the postural control system as uncontrolled backwards

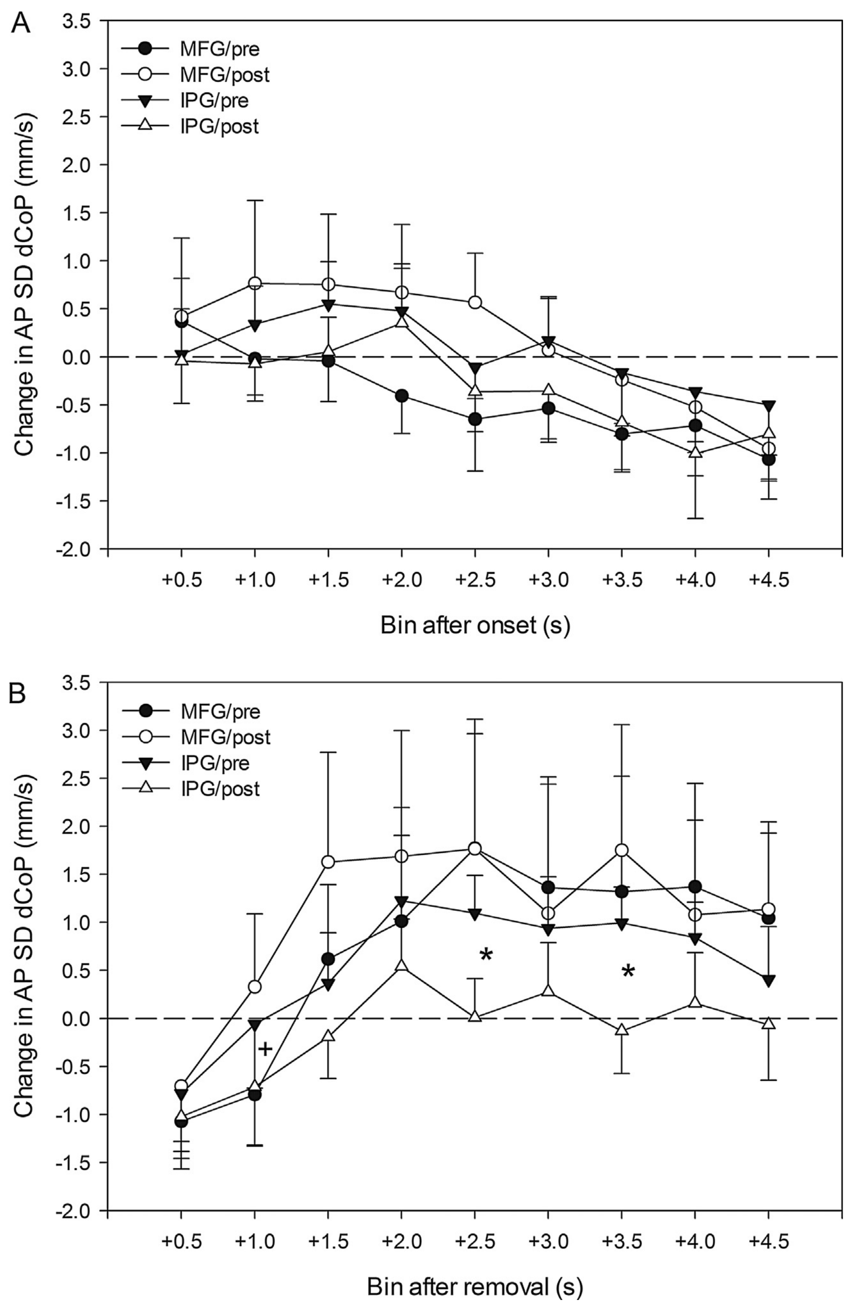


Fig. 3. (A) The time course of sway across 9 bins of 500 ms duration after contact onset. The black squares indicate the overall average across stimulation locations and stimulation. (B) The time course of sway across 9 bins of 500 ms duration after contact removal. The black squares indicate the average across stimulation locations before rTMS and the open squares the average across stimulation locations after rTMS. Error bars indicate standard error of the mean. SD: standard deviation, MFG: middle frontal gyrus, IPG: intraparietal gyrus, pre: before rTMS, and post: after rTMS.

sway, which would trigger compensatory postural adjustments and increase sway. This sensory illusion, respectively, misinterpretation, might be served by neural mechanisms represented within the contralateral IPG and become attenuated by local rTMS, thus softening its impact on sway control. Brandt and co-workers suggested inhibitory reciprocal intermodal interaction (specifically vestibulo–visual interaction) as a mechanism to facilitate self-motion perception in ambiguous contexts [21,22]. These interactions may be susceptible to local deactivations in associated regions by rTMS. For example, Gratton et al. [23] showed how cTBS over the left-hemisphere DLPFC alters functional connectivity in the brain, especially in fronto–parietal networks.

Apparently, we did not stimulate locations within the left hemisphere specific for the integration of fingertip afferences in a trunk-centered frame of reference. Locations more superior to IPG might be more appropriate targets. For example, remapping of touch location on the skin as well as proprioceptive information about arm configuration in a spatial frame of reference involves higher-order cognitive processes located in multimodal brain areas such as the right-hemisphere posterior parietal cortex [PPC, [24]]. Alternatively, it could be that the left hemisphere is not at all involved in postural task-specific processing of fingertip afferences. There is currently no evidence, however, for example in the case of patients with left and right hemisphere parietal lesions, which argues for an exclusivity of the right parietal cortex regarding the benefits of light fingertip contact on body sway. Nevertheless, if this were the case then we expect reduced light touch benefits in both the dominant and non-dominant hand after the right PPC rTMS. A follow-up experiment ought to pursue this hypothesis by testing the change in the effects of hand dominance on body sway from before to after stimulation of either the left- or right-hemisphere PPC. An improvement would also be the application of real-time neuronavigation for the placement of the TMS coil during repetitive stimulation, for example by choosing coordinates for locations directly implicated in intermodal remapping in egocentric frames of reference.

5. Conclusion

Unintended overshoot in body sway following unpredictable removal of sway-specific fingertip feedback is dampened after disruption of the left-hemisphere IPG by rTMS. Altered activity within the IPG may simultaneously reduce both tactile attention capture, illusion of backwards falling or transient postural disorientation by perceived intermodal conflict in effect leading to an optimized progression of sway on haptic deprivation.

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Effects of Maintaining Touch Contact on Predictive and Reactive Balance

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Johannsen L, Wing AM, Hatzitaki V. Effects of maintaining touch contact on predictive and reactive balance. *J Neurophysiol* 97: 2686–2695, 2007. First published February 15, 2007; doi:10.1152/jn.00038.2007. Light touch contact between the body and an environmental referent reduces fluctuations of center of pressure (CoP) in quiet standing although the contact forces are insufficient to provide significant forces to stabilize standing balance. Maintenance of upright standing posture (with light touch contact) may include both predictive and reactive components. Recently Dickstein et al. (2003) demonstrated that reaction to temporally unpredictable displacement of the support surface was affected by light touch raising the question whether light touch effects also occur with predictable disturbance to balance. We examined the effects of shoulder light touch on SD of CoP rate (dCoP) during balance perturbations associated with forward sway induced by pulling on (voluntary), or being pulled by (reactive), a hand-held horizontal load. Prior to perturbation, SD dCoP was lower with light touch, corresponding to previous findings. Immediately after perturbation, SD $dCoP_{AP}$ was greater with light touch in the case of voluntary pull, whereas no difference was found for reflex pull. However, in the following time course, light touch contact again resulted in a significantly lower SD dCoP and faster stabilization of SD dCoP. We conclude that shoulder light touch contact affects immediate postural responses to voluntary pull but also stabilization after voluntary and reflex perturbation. We suggest that in voluntary perturbation CoP fluctuations are differentially modulated in anterioposterior and mediolateral directions to maintain light touch, which not only provides augmented sensory feedback about body self-motion, but may act as a "constraint" to the postural control system when preparing postural adjustments.

INTRODUCTION

The maintenance of standing balance involves a mixture of reactive and predictive control processes (Massion 1992, 1994). Sensory cues from unexpected imposed forces or torques producing disturbances in the position of the center of mass (CoM) relative to the base of support (BoS) result in multisegmental postural adjustments. These stiffen the musculoskeletal structure allowing ground reaction forces to oppose the applied forces and torques (Balasubramaniam and Wing 2002; Nashner and McCollum 1985) and tend to restore CoM over BoS. The amplitude of the postural response scales with the applied force and the onset latency is typically 70–100 ms, which is sufficient for supraspinal, possibly cortical, pathways (Diener et al. 1988).

Predictable disturbances to balance caused by voluntary movement, such as raising of the arms (Bouisset and Zattara 1987; Cordo and Nashner 1982) or forward displacement of a mass (Wing et al. 1997), are associated with anticipatory postural adjustments. These result in ground reaction forces that can lead the focal movement by ≥100 ms and serve to reduce the impact of the voluntary movement on standing posture (Benvenuti et al. 1997; Bouisset and Zattara 1987). Setting of these adjustments may involve an internal forward model that predicts the consequences of the focal movement (Massion 1994). Alternatively, anticipatory postural adjustments may be based on an inverse model, which could be trained by feedback error learning (Kawato and Wolpert 1998).

Quiet standing involves a series of minor postural adjustments that result in fluctuations of the center of pressure (CoP). The discrepancy between the CoP and the vertical projection of the CoM is proportional to the acceleration tending to restore the CoM to a position centered over the BoS (Winter et al. 1996). The resulting changes in body position are termed body sway (Nashner 1971). Sway increases when sensory inputs (e.g., vision) are reduced or degraded, and this indicates the importance of feedback in limiting sway. Normal feedback routes can be augmented in several ways. For instance, auditory or vibratory signals that are directionally linked to postural sway result in reduced CoP fluctuations (Chiari et al. 2005; Dozza et al. 2005a,b; Wall et al. 2001). Light touch (LT), in which one digit rests gently (for example, 1 N contact force) against a stable environmental referent, also reduces postural sway in quiet standing (Clapp and Wing 1999; Jeka and Lackner 1994). Sensory augmentation by light touch can be very effective, for instance, completely suppressing the increased sway associated with leg muscle vibration during quiet standing (Lackner et al. 2000). Light touch is more effective when fingertip contact is maintained in the plane of greater sway (more unstable direction) (Rabin et al. 1999). Light touch contact when the finger is held in position by an external clip is even more effective in reducing sway than free finger contact (Krishnamoorthy et al. 2002; Rogers et al. 2001).

One possible reason that light touch reduces sway is that it provides a time-advanced cue to sway. This view receives support from the finding of a correlation between contact force and CoP with finger contact force leading CoP by 250–300 ms in mediolateral (ML) (Jeka and Lackner 1994, 1995; Rabin et al. 1999) and anterioposterior (AP) (Clapp and Wing 1999; Rabin et al. 1999) directions. Thus according to a feedback control account, a change of the fingertip forces occurs before the EMG response; this shows a 150-ms lead over CoP, which in turn leads sway by 150 ms (Barela et al. 1999; Jeka and Lackner 1995; Rabin et al. 1999). Recently it was shown (Rabin et al. 2006) that light touch effects develop rapidly in quiet standing over the first 2–3 s of contact with a downward

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trend in the absolute error of the CoP detectable in the initial 200 ms. In this study, the correlation between finger tip shear force and CoP was at a maximum when CoP lagged shear force by 320 ms.

Light touch contact need not be restricted to the finger but also works when applied to other body segments. For example, light touch contact with shoulder and leg has been shown to be effective in reducing postural sway (Rogers et al. 2001). Light touch to the head or neck can be more effective in reducing body sway than light touch at the finger tip (Krishnamoorthy et al. 2002). These and other findings suggest that, to reduce sway, the postural system makes use of two types of sensory information from light touch contact: one related to the provision of a fixed reference point in space (Reginella et al. 1999), the other related to the information provided by transient forces developed between the body part and the contact surface (Krishnamoorthy et al. 2002; Rogers et al. 2001).

The majority of studies reporting reduction of postural sway with light touch used a paradigm in which participants kept a static, upright standing posture. It is unclear what are the relative contributions of predictive and reactive elements in the control of balance in this situation. It would be interesting to know whether light touch contributes equally to both aspects of control. Another question is whether light touch is effective for transient disturbances to balance, that is, the dynamic aspects of postural control. A recent study sought to determine whether light touch results in facilitation of postural reflexes triggered by transient balance perturbations (Dickstein et al. 2003). In this study, light touch during quiet standing was combined with sudden 6 cm backward translation of BoS at one of three different velocities. Although no reliable effects of light touch were observed on the latency of postural reflexes after the perturbation, their gain, as indexed by CoP rate relative to BoS velocity, increased with light touch. Moreover, it was noted that light touch tended to act as a "constraint" on postural adjustments in that the latter evidenced AP and ML components that served to maintain light touch contact. Thus with light touch, the rate of the (forward) CoP_{AP} response was reduced and the rate of the (rightward) CoP_{ML} response increased, these changes in COP components being compatible with increased trunk movement toward the location providing light touch contact. Given the increase in gain of the postural reflex with light touch, it is interesting to ask whether, after perturbation, postural sway and CoP fluctuations were reduced; however, this study did not evaluate this (Dickstein et al. 2003).

In the present study, we examined light touch effects on CoP fluctuations both during and after reflex postural response to sudden-onset perturbations to balance. In addition, we compared the effect of light touch on postural reflexes with its effect on anticipatory postural adjustments associated with voluntary perturbations to balance. The two contrasting dynamic contexts involved a voluntary, self-imposed balance perturbation (pulling on a manipulandum with the right hand; hence anticipatory) compared with an externally imposed perturbation of balance (being pulled by the manipulandum; hence reactive). To reduce the situational demands and avoid potential bimanual conflict if light touch had required use of the other hand, light touch contact was applied to the left arm near the shoulder rather than to the left hand.

On the basis of the previous study showing light touch contact effects on the gain of the postural reflex in reactive balance (Dickstein et al. 2003), we expected light touch would enhance the postural response at the onset of the external perturbation. In contrast, given that anticipatory postural adjustments tend to minimize postural disturbance associated with voluntary movement (Bouisset and Zattara 1987), assuming these are already optimal, we hypothesized that light touch contact would cause no further improvements of the anticipatory postural adjustments. We did not therefore expect that light touch would have an effect on stabilization after perturbation associated with voluntary compared with reactive postural responses.

METHODS

Participants

Eleven right-handed adults served as participants [age: 30.1 ± 11.6 (SD) yr]. None reported any neurological or musculoskeletal disorders. All gave their informed consent, and the experiment had the approval of the local ethical committee on testing human participants.

Apparatus

Participants stood in stockinged feet on a force platform (4060H, Bertec) used to measure the six components of the ground reaction forces and torques to determine the AP and ML components of CoP fluctuations (see Fig. 1).

Participants held a manipulandum (M) in precision grip (using the digit pads of the thumb and 1–3 fingers) with the right hand at waist height. The manipulandum comprised two one-dimensional (1D) force transducers (F250, Novatech Hastings) configured in orthogonal orientations to allow simultaneous recording of both the horizontal load force (load) acting on the manipulandum and the normal grip force (grip) exerted by the participant on the manipulandum. Two horizontal steel cables attached to the manipulandum and, over pulleys, to two counter weights kept the manipulandum at a constant position in space. An additional weight (15 N) could be added to one of the counter weights to produce a forward load acting on the manipulandum.

A second pair of two orthogonally mounted 1D force transducers (F250, Novatech Hastings) was mounted on a rigid horizontal support bar fixed to a vertical stand. These transducers were adjusted to apply light touch to the left arm of the participant near the shoulder through a flat wooden endplate covered with a layer of fine grit sandpaper, which provided a textured contact surface. The force transducers recorded normal and shear forces at the left arm aligned with ML and AP directions of the force platform.

Participants' movement kinematics of the trunk and the right hand were registered using a six-camera optoelectronic motion tracking system (VICON 512, Oxford Metrics) with reflective markers attached to the neck (spinal bone C7), right shoulder, and right wrist. EMG activity was recorded over right lateral gastrocnemius (GAS), first dorsal interosseus of the right hand (1DI) and biceps (BIC), and triceps (TRI) of the right arm. The EMG signals were amplified (gain: 3,400) by means of battery-powered amplifiers near the electrodes and passed through a unit gain isolating amplifier.

Data from the force platform, force transducers and EMG electrodes were sampled at 1,080 Hz. Kinematics were sampled by the camera system at 120 Hz via a 16-bit analog interface (VICON datastation, Oxford Metrics).

Procedure

Participants were instructed to stand upright, with eyes closed, head facing forward, as still as possible with 12-cm ML separation of the

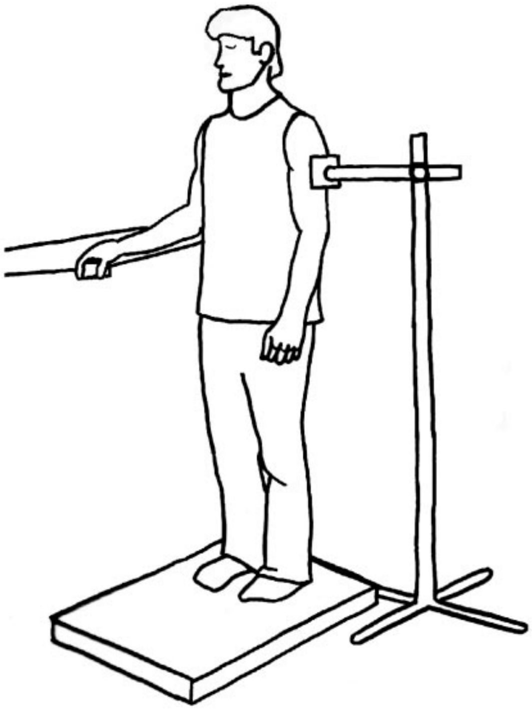

FIG. 1. Drawing of the experimental setup. The participant stood with eyes closed on a force plate. Precision grip (opposed thumb and fingers) was used to lightly grasp the manipulandum, which was attached through a pulley system to a basket behind the participant. Light touch to the left arm of the participant near the shoulder was applied through a flat wooden endplate covered with a layer of fine grit sandpaper, which provided a textured contact surface. In different blocks of trials, a weight was dropped into the basket, and the participant resisted the pull by holding on and steadying the manipulandum (reflex pull), or the participant pulled on the manipulandum to lift the basket and weight off the support (voluntary pull). Three reflective markers were attached to the participant to permit motion tracking of the neck C7, right shoulder, right wrist. Surface EMG was used to record muscle activity of 1st dorsal interosseus of the right hand, the biceps and triceps of the right arm and lateral gastrocnemius of the right leg.

inner border of their heels. A template was used to mark the positions of the feet on the platform so that this posture could be maintained throughout the experiment. They were instructed to hold the manipulandum under two contrasting horizontal *loading conditions* that were tested in separate blocks. During reflex pull, the manipulandum was subject to an added forward load caused by the weight (15 N) being released at an unpredictable time. The weight was released manually on each trial with slightly varying height which resulted in a variable maximum load ($Load_{max}$) and load rate ($dLoad_{max}$). At the beginning of each trial, participants kept light contact with the manipulandum so that they could detect the sudden load onset. During voluntary pull, participants started the trial with fingers near but not in contact with the manipulandum. In their own time, they then gripped and pulled the manipulandum horizontally to quickly lift the 15 N weight off a support. During reflex pull as well as voluntary pull trials, participants were required to keep a steady hold on the manipulandum until the trial ended. These loading conditions were combined with two *touch conditions* involving either light touch or no touch at the left shoulder. In the light touch condition, participants were instructed to use the minimum force required to keep contact; no concurrent feedback was provided about the contact forces.

Twenty trials of 10-s duration were run in each of the four conditions (reflex pull/light touch, reflex pull/no touch, voluntary pull/light touch, voluntary pull/no touch). Each condition was tested in two blocks of 10 trials. The blocked sets of the four conditions were randomized but participants had to complete at least one block of 10 trials in each condition before the second block in the same condition was presented. During each trial the perturbation of standing balance by reflex pull or voluntary pull occurred between 3 and 7 s after trial onset. Prior to data collection in each trial, participants were instructed to close their eyes and to say when they were ready to commence the trial.

Analysis

ML and AP components of the kinematics and the kinetics were digitally low-pass filtered at 10 Hz (dual pass 4th-order Butterworth filter) and differentiated to obtain rate based measures (dLoad, dGrip, dNormal, dShear, $dCoP_{AP}$, $dCoP_{ML}$) that afforded stable zero-valued baselines prior to perturbation. EMG recordings were band-stop filtered between 48 and 52 Hz and subsequently rectified to obtain the EMG envelope. Afterward, the EMG envelopes were also low-pass filtered at 10 Hz (dual pass 4th-order Butterworth filter).

Individual data streams were analyzed using custom interactive waveform measurement software written in Labview (7.1) and Matlab (7.0). Times of onset for dLoad and dGrip as well as onset of $dCoP_{AP}$ and $dCoP_{ML}$ and of EMG envelopes (GAS, 1DI, BIC, TRI) were determined using a cut-off threshold of 4 SD above baseline (before perturbation). Relative onset times for each variable were then computed by subtracting the dLoad onset time. After rejecting those relative onset times with an absolute value >5 s, the remaining relative onset times were analyzed using three-way repeated-measures ANOVA (SPSS 11.5) with muscle, loading condition, and touch as independent factors.

To investigate the time course of the balance response, each trial was segmented into periods of 1-s duration starting from 2 s before to 4 s after onset of dLoad and the within-trial mean (AV) and within-trial SD of dCoP in AP and ML directions were determined for each time segment. For the statistical analysis of AV and SD, the time course was subdivided into two phases: *baseline* before perturbation, preceding dLoad onset ($t < 0$ s), and after *perturbation* following dLoad onset ($t \geq 0$ s). The data for each phase were then subject to repeated-measures ANOVA with *loading* and **touch conditions** as the primary independent factors and *direction* and *time* (2 time intervals for 1st phase; 5 time intervals for 2nd phase) as additional factors. SD CoP data for the second (perturbation) phase were linearized by computing the natural logarithm (ln) before the statistical analysis.

To describe the stabilization of posture following the perturbation for each single trial, we determined the fit of an exponential decreasing function [$x(t) = C + A*e^{-t/B}$] to the reduction in SD of $dCoP_{AP}$ and $dCoP_{ML}$ across the five 1-s time intervals after the perturbation. The three function parameters C (asymptote), A (intercept x_0), and B (time constant) were determined using a least-squares estimation algorithm and were subsequently averaged for each experimental condition and each participant.

To characterize touch forces during light touch, touch force rates dShear and dNormal were analyzed at the left shoulder in terms of AV and SD using two-way repeated-measures ANOVA with *time* (2 time intervals for 1st phase; 5 time intervals for 2nd phase) and *loading* as independent factors.

RESULTS

In the following, we first present data relating to the efficacy of the experimental paradigm, then consider the effects of touch on maintenance of balance.

Effects of loading condition on pulling responses

The experimental conditions resulted in an overall average $Load_{max}$ of 26.5 ± 2.2 (SD) N for voluntary pull and 18.3 ± 0.7 N for reflex pull. Overall average $dLoad_{max}$ was 178.9 ±

49.8 N/s for voluntary pull and 103.3 ± 10.1 N/s for reflex pull. The difference between loading conditions was reliable for both variables [both $F(1,10) \geq 22.37$, both $P \leq 0.001$]. There was no difference in $\mathrm{Load_{max}}$ or $\mathrm{dLoad_{max}}$ as a function of Light touch contact.

Figure 2 shows illustrative data (gray traces) aligned on dLoad onset from three single trials as well as the average data (black line) for all 20 trials of a single participant performing a voluntary pull and a reflex pull, both with light touch contact. Figure 2*A* shows measures from the manipulandum and the right hand, whereas *B* shows measures relating to the postural response and right shoulder contact. Inspection of the traces reveals broadly similar responses in the two conditions (but note the lower $\mathrm{dLoad_{max}}$ and reversed sign of the wrist velocity in reflex pull). However, there is a marked contrast in timing with responses in voluntary pull occurring with or slightly before dLoad onset, whereas in reflex pull the responses clearly follow dLoad onset. The *lowest two panels* of Fig. 2*B* show considerable variation in the touch force at the right shoulder, both within and between trials. AV normal force over all participants in the two loading conditions was 2.9 ± 1.4 N with no significant difference between voluntary and reflex pull conditions. Also, the absolute AV shear force (mean = 0.8 ± 0.5 N) was not reliably different between the two loading conditions.

An analysis of the relative onset times for dGrip and $\mathrm{dCoP_{AP}}$ and for each of the four muscles supported the contrast in timing of the response between voluntary pull and reflex pull evident in Fig. 2. Onset of dGrip showed a significant effect of loading condition [$F(1,10) = 329.22$, $P < 0.001$] but no effect of touch and no interaction between loading and touch. During voluntary pull, dGrip onset preceded dLoad onset on average by 46 ± 38 ms while it was delayed on average by 157 ± 46 ms in reflex pull. Onset of $\mathrm{dCoP_{AP}}$ was also affected by loading [$F(1,10) = 131.01$, $P < 0.001$] and touch [$F(1,10) = 8.43$, $P = 0.02$], but there was no interaction. The average delay of $\mathrm{dCoP_{AP}}$ onset relative to dLoad onset was 9 ± 48 ms in light touch and 13 ± 56 ms in no touch in voluntary pull. Relative

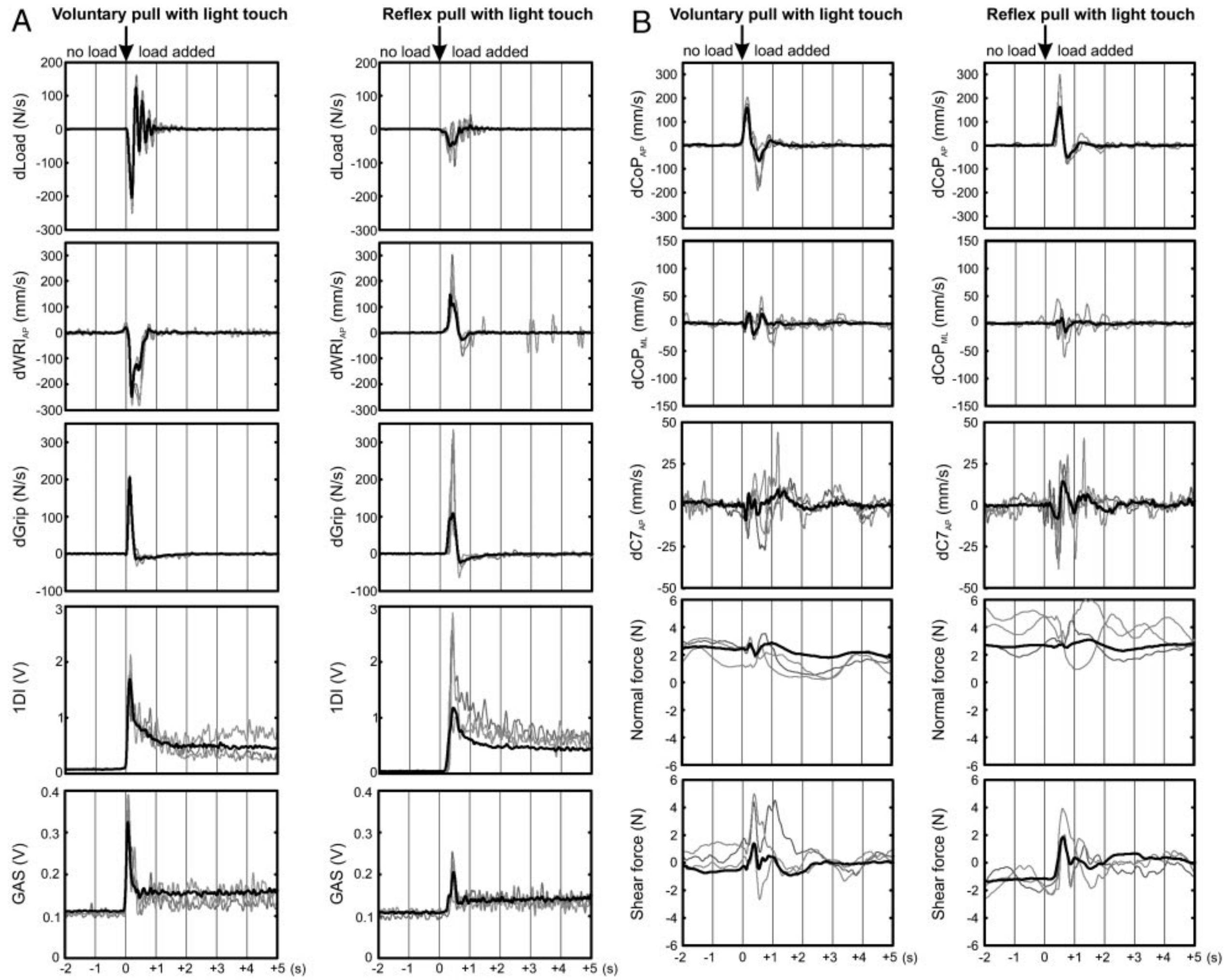

FIG. 2. Illustrative data (gray traces) for each dependent measure from 3 trials of a single participant with light touch contact at the left shoulder during either voluntary (*left*) or reflex (*right*) pull. The thick black line represents the average of the same participant across all 20 trials in the 2 loading conditions. *A*: measures from the manipulandum (M) and the right hand; load force rate (dLoad), wrist anterioposterior (AP) velocity, grip force rate (dGrip), 1st dorsal interosseous and gastrocnemius lateralis electromyogram (EMG). At the manipulandum, an increased transducer voltage signified decrease in load and increase in grip. *B*: measures relating to the postural response and right shoulder contact; CoP velocity in both directions, C7 velocity in the AP direction, shear force, and normal force rates. In the AP direction, a positive sign signified forward directional shift of CoP, and leftward shift in the mediolateral (ML) direction (toward light touch contact surface at the left shoulder). For the contact forces, a positive sign indicated a forward directed shear force in the AP direction, while in the ML direction a positive sign indicated leftward increasing normal force. Each trace is aligned at *time 0* with the onset of dLoad for that trial.

onset of $dCoP_{AP}$ in reflex pull was much later with an average latency of 264 ± 55 ms in light touch and 290 ± 49 ms in no touch.

Analysis of relative onset times of the EMG revealed a significant effect of loading [$F(1,10) = 109.52$, $P < 0.001$] and of muscle [$F(3,30) = 11.36$, $P < 0.001$] but no effect of touch and only an interaction between loading and muscle [$F(3,30) = 5.36$, $P = 0.004$]. Both 1DI and TRI exhibited relative onset times that were similar to those for dGrip. Average 1DI onset times (VP: −58 ± 45 ms; RP: 153 ± 39 ms) slightly preceded the onset of dGrip, whereas average onset of TRI (VP: −32 ± 43 ms; RP: 198 ± 63 ms) occurred slightly later. The average onset of GAS (VP: 0 ± 83 ms; RP: 283 ± 130 ms) coincided closely with the onset of $dCoP_{AP}$. The average onset of BIC was delayed by 131 ± 167 ms) during voluntary pull and by 296 ± 120 ms in reflex pull.

Figure 3 shows the time course of AV dCoP in AP and ML directions for voluntary and reflex pull. Both AV $dCoP_{AP}$ and AV $dCoP_{ML}$ start and end at zero, but after perturbation, AV $dCoP_{AP}$ exhibits a clear maximum (forward directed movement), whereas AV $dCoP_{ML}$ shows a small minimum (rightward directed movement). The time course of AV $dCoP_{AP}$ was

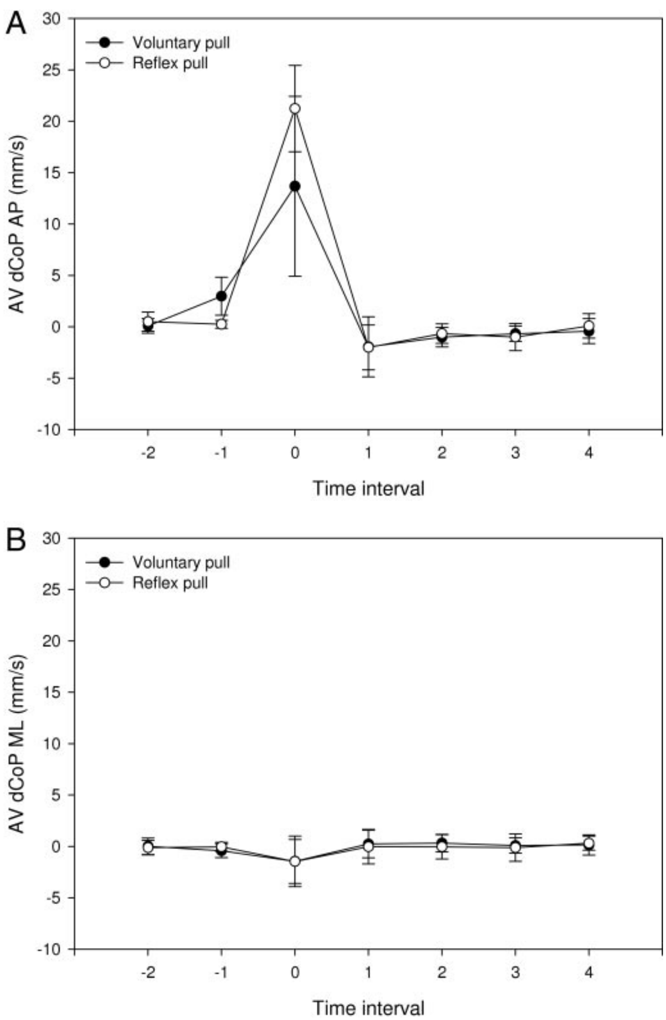

FIG. 3. Temporal evolution of the average CoP rate in the AP (*top*) and ML (*bottom*) directions (AV dCoP) during voluntary and reflex pull for 1-s intervals from 2 s before to 4 s after perturbation. The data have been collapsed over the 2 touch contact conditions. Error bars represent standard deviations.

similar between loading conditions except for an anticipatory response and reduced maximum during voluntary pull. No effect of loading was evident for AV $dCoP_{ML}$.

Effects of shoulder contact on balance

The effect of light touch contact on balance was analyzed over the full time course of all the trials in terms of the SD of the dCoP in both AP and ML directions for voluntary and reflex pull loading conditions. Figures 4 and 5 show the effect of touch on SD $dCoP_{AP}$ and SD $dCoP_{ML}$. The SD data exhibit maxima after perturbation and then decrease geometrically back toward the preperturbation baseline. A summary of the main effects and interactions for the four-way repeated-measures ANOVAs on SD dCoP with factors direction, loading, touch, time is given in Table 1.

Before perturbation, SD dCoP was significantly reduced by touch in both loading conditions for both directions. However, in the AP direction the effect of touch on SD dCoP was smaller during voluntary pull than reflex pull, whereas the opposite pattern was found for the ML direction. Also, SD dCoP was generally lower in the ML than AP direction as well as lower on reflex pull compared with voluntary pull. A significant interaction between loading and time reflected an increase in SD dCoP during voluntary pull in the 1-s interval before the perturbation that was not seen during reflex pull and was more pronounced in the AP direction (anticipatory response). Finally, the difference between voluntary and reflex pull was larger for SD $dCoP_{AP}$ than SD $dCoP_{ML}$ as indicated by a significant interaction between direction and loading.

Touch had a general effect on SD dCoP during the stabilization phase over the five time periods taken from dLoad onset ($t \geq 0$ s; see Table 1). Analysis of the log-linearized SD dCoP time course starting immediately after the perturbation demonstrated that SD dCoP gradually dropped over time and was reduced with Light touch in both AP and ML directions. However, the reduction of SD dCoP with light touch was greater in the ML direction. The effect of touch increased over time and was greatest at the last time interval of the stabilization phase. Again as in the baseline phase before the perturbation, SD $dCoP_{ML}$ was generally smaller than SD $dCoP_{AP}$.

The fit of the exponential decreasing function on the reduction of AP and ML SD dCoP during the stabilization phase after the perturbation was generally found to be quite satisfactory. Table 2 shows the average parameter estimates for the SD dCoP reduction function during stabilization for each experimental condition. The intercept parameter A (at x_0) indicated greater SD dCoP in the period immediately after dLoad onset in the AP compared with the ML direction [$F(1,10) = 107.48$, $P < 0.001$]. There were no effects of loading or touch on the intercept nor was there any interaction between touch and direction. However, a significant crossover between the two loading conditions as a function of direction was found with the intercept tending to be smaller during voluntary compared with reflex pull in the AP direction but larger in the ML direction [$F(1,10) = 8.52$, $P = 0.015$]. Further, a significant interaction between loading condition and touch contact reflected a tendency for the intercept to be greater with light touch during voluntary pull, whereas the opposite was true for reflex pull [$F(1,10) = 10.12$, $P = 0.01$]. Finally, there was a significant two-way interaction between direction, touch, and

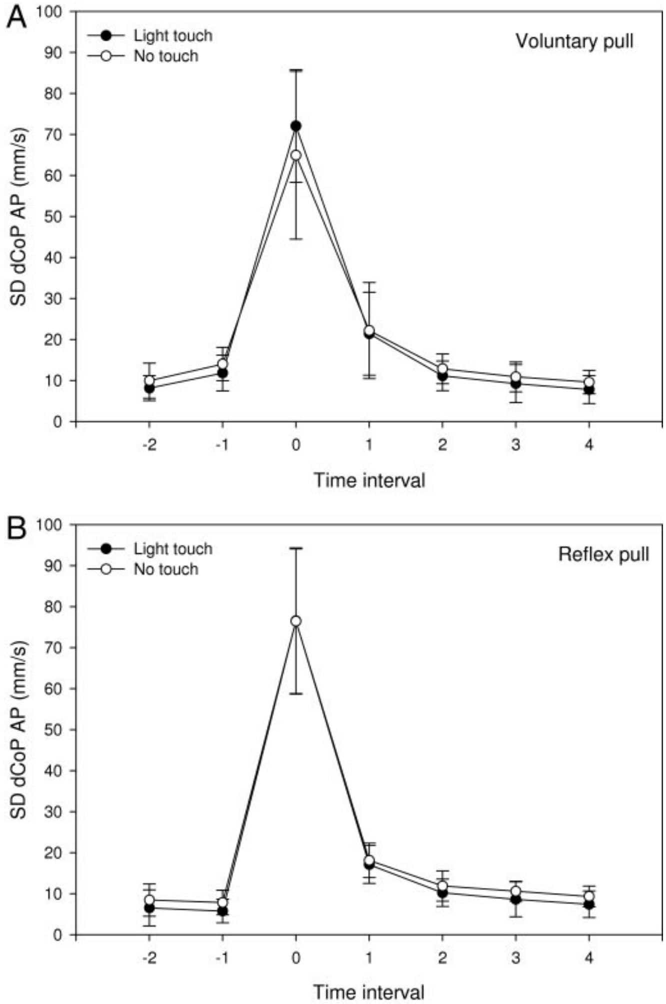

FIG. 4. Temporal evolution of the SD of CoP rate in the AP direction (SD $dCoP_{AP}$). Each point represents the average (over subjects and replications) of SDs computed on single trial data over 1-s intervals from 2 s before perturbation at *time 0* (defined by dLoad onset) to 4 s after. *A*: comparison of voluntary pull with and without light touch contact. *B*: same contrast for the Reflex pull condition. Error bars represent standard deviations

loading with a noticeably greater intercept during voluntary pull with light touch in the AP direction [$F(1,10) = 12.53$, $P = 0.005$; see dLoad onset interval in Fig. 3].[1]

The postperturbation reduction in SD CoP is captured by the time constant parameter *B*. Light touch resulted in a significantly shorter time constant for SD CoP during stabilization [$F(1,10) = 11.84$, $P = 0.006$]. This effect was more pronounced in the ML direction as indicated by a significant interaction between direction and touch [$F(1,10) = 6.45$, $P = 0.03$]. Further, a significant two-way interaction between direction, touch, and loading [$F(1,10) = 5.47$, $P = 0.04$] demonstrated that light touch reduced the time constant of the reduction for reflex pull exclusively in the ML direction, whereas for voluntary pull, the touch effect was comparable in both directions. Generally, the time constant for the reduction of SD CoP_{AP} was slightly shorter for reflex than voluntary pull. The opposite was true for the ML direction as expressed by a significant interaction between load and direction [$F(1,10) = 19.20$, $P = 0.001$]. Further, the effect of direction was significant [$F(1,10) = 44.64$, $P < 0.001$] with a shorter time constant in the AP direction.

Under light touch conditions, the asymptote parameter *C* showed a significantly lower final value for SD CoP during stabilization [$F(1,10) = 46.24$, $P < 0.001$]. Moreover, the asymptotic value for SD CoP_{ML} was lower than for SD CoP_{AP} [$F(1,10) = 108.69$, $P < 0.001$].

Maintenance of shoulder contact

Figure 6 shows the mean force rate of the touch contact in the AP (AV dShear) direction. In general it will be observed that the AV function starts and ends at zero with fluctuations at and after perturbation. Before perturbation, there was no difference in AV dShear between voluntary pull and reflex pull. The same was true for AV dNormal (not shown).

After perturbation during stabilization, AV dShear showed a significant two-way interaction between direction, loading, and time [$F(4,40) = 5.43$, $P = 0.001$]. The sign of AV dShear was

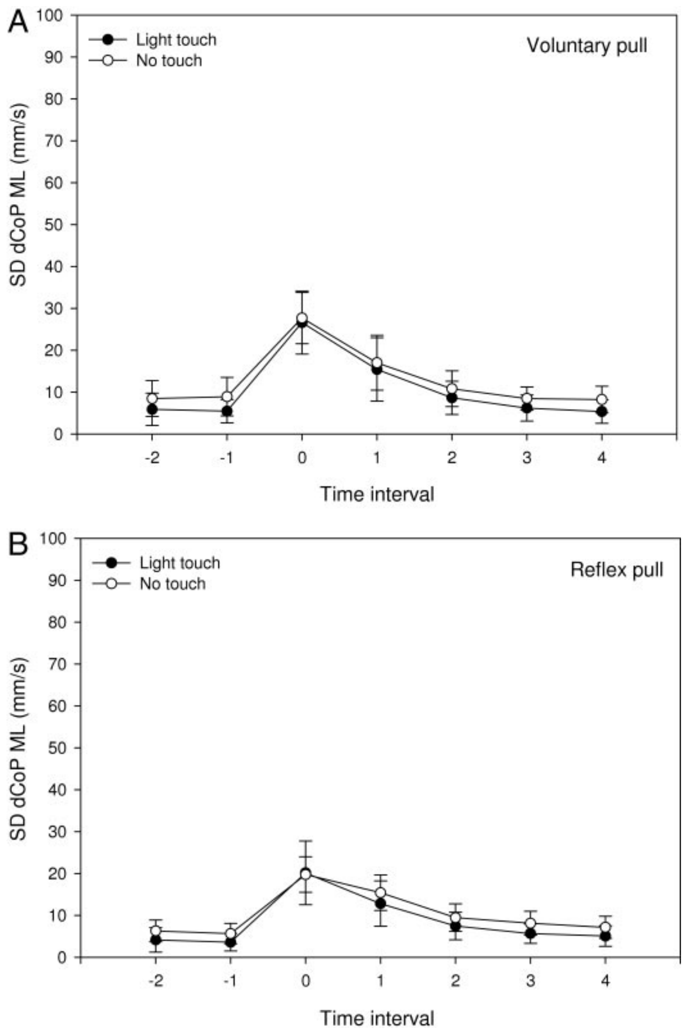

FIG. 5. Temporal evolution of the SD of CoP rate in the ML direction (SD $dCoP_{ML}$) for 1-s intervals from 2 s before to 4 s after perturbation. *A*: comparison of voluntary pull with and without light touch contact. *B*: same contrast for the reflex pull condition. Error bars represent standard deviations.

[1] An additional analysis of peak-to-peak amplitude in each time period was performed to take explicit account of the biphasic form of the dCoP function. The results at t_0 closely followed those reported for the intercept.

TABLE 1. *Effect table (F-values) for influence of direction, loading, touch, and time on SD dCoP*

Phase	Main Effects				Interactions										
	Direction	Loading	Touch	Time	Direction × Loading	Direction × Touch	Direction × Time	Loading × Touch	Loading × Time	Touch × Time	Direction × Loading × Touch	Direction × Loading × Time	Direction × Touch × Time	Loading × Touch × Time	Direction × Loading × Touch × Time
Baseline ($t < 0$ s)	63.35**	92.09**	30.23**	4.47	5.66*	1.45	12.87*	0.43	13.49*	0.56	7.42*	24.4**	0.06	0.41	0.85
Stabilization ($t > 0$ s)	147.98**	2.41	15.36*	257.23**	17.15*	7.30*	88.17**	0.08	0.81	19.1**	0.71	14.06**	0.49	0.63	1.11

SD, standard deviation, dCoP, SD of center of pressure rate. *$P < 0.05$; **$P < 0.001$.

different for voluntary pull and reflex pull. AV dShear was initially directed forward for voluntary pull and backward for reflex pull, indicating contrasting tendencies to sway forward and backwards at perturbation. Subsequently AV dShear changed in opposite directions for voluntary pull and reflex pull. In both loading conditions, the time course of AV dShear resembled an oscillation which started at perturbation and continued during stabilization after perturbation. In contrast, no difference between loading conditions was evident for AV dNormal. The time course of AV dNormal also tended to resemble a fluctuation that started at perturbation with force directed into the contact and tended to oscillate during stabilization. However, this effect was not statistically reliable.

DISCUSSION

Light touch with a static environmental contact point during quiet standing reduces body sway and associated fluctuations in ground reaction forces, as indexed by CoP variation (Holden et al. 1994; Jeka and Lackner 1994; Jeka et al. 1997). Light touch contact has also been shown to affect the postural response to forward sway produced by unpredictable backward displacement of the support surface (Dickstein et al. 2003). We used an unimanual pulling paradigm to investigate the time course of light touch contact effects on standing balance after predictable self-imposed (with voluntary pull) or unpredictable externally imposed (with reactive pull) perturbation. In both cases, perturbing forces at the right hand in the region of 20 N produced forward directed movement of CoP_{AP}. We predicted light touch conditions, involving shoulder contact with a fixed reference, would improve the efficiency of reactive components of the response in reflex pull, facilitating earlier reduction in sway compared with no touch conditions. In contrast, we did not expect any effects of light touch on stabilization after voluntary pull under the assumption that feed forward anticipatory postural adjustments would already optimally stabilize body balance after the perturbation.

Our paradigm was successful in eliciting contrasts between voluntary pull and reflex pull in terms of reliably later grip, postural kinetics, and muscle responses in the reflex pull condition. Both loading conditions resulted in oscillatory fluctuations in the AP and ML components of the rate of center of pressure (dCoP). We took as our primary analysis measure the SD of $dCoP_{AP}$ and $dCoP_{ML}$ computed over successive 1-s windows. In light touch conditions, the magnitude of the horizontal light touch contact normal force at the left shoulder averaged ~2.7 N, which was somewhat greater than the 1 N vertical force threshold employed in previous studies (e.g., Holden et al. 1994). However, the shear force component of light touch contact in the AP direction relevant to the direction of perturbation was only 0.8 N and so small enough to be deemed "light."

First considering variability, it may be observed that SD of both $dCoP_{AP}$ and $dCoP_{ML}$ were significantly reduced with light touch immediately *before* the perturbation which replicates previous studies (Clapp and Wing 1999; Jeka and Lackner 1994) despite the use of an observation window of only 1-s duration in the present study. This result confirms a recent report (Rabin et al. 2006) of the sensitivity to light touch effects of CoP measures based on short observation windows. In the stabilization period *after* both voluntary pull and reflex

TABLE 2. *Mean parameter estimates of the SD dCoP reduction function during stabilization in relation to sway direction and touch*

	Voluntary Pull				Reflex Pull			
	AP		ML		AP		ML	
	LT	NT	LT	NT	LT	NT	LT	NT
Intercept (A; at x_0)	64.26 ± 15.22	55.28 ± 19.66	24.12 ± 6.54	23.79 ± 6.95	63.70 ± 16.38	67.14 ± 19.35	17.08 ± 6.32	17.44 ± 4.02
Time constant (B)	0.58 ± 0.19	0.73 ± 0.45	1.27 ± 0.51	1.38 ± 0.60	0.50 ± 0.10	0.47 ± 0.08	1.58 ± 0.70	2.48 ± 1.34
Asymptote (C)	7.52 ± 3.62	8.82 ± 3.11	4.34 ± 2.79	6.33 ± 3.01	7.08 ± 2.41	8.47 ± 2.37	3.86 ± 1.67	4.69 ± 1.73

AP, anterioposterior; ML, mediolateral; LT, light touch; NT, no touch.

pull perturbations, we found SDs of $dCoP_{AP}$ and $dCoP_{ML}$ were also significantly lower with light touch. Further, light touch resulted in faster stabilization of SD dCoP under both loading conditions. More specifically, during voluntary pull light touch shortened the time constant of SD dCoP reduction in both directions, whereas during reflex pull, this was the case for SD $dCoP_{ML}$ only.

At perturbation, SD $dCoP_{AP}$ was greater with light touch in the case of voluntary pull, whereas no difference was found for reflex pull. This increase in SD $dCoP_{AP}$ at perturbation for voluntary pull with light touch represents a marked reversal of the light touch effect in the baseline period. Because SD $dCoP_{AP}$ was greater *at perturbation* with light touch for voluntary pull but reverts equally quickly to the same level *after perturbation*, it further emphasizes that light touch contact facilitates sway suppression faster in the case of voluntary pull.

The finding of a reduction in SD dCoP in voluntary pull is surprising in the sense that it indicates that the response to a self-imposed perturbation can be further improved with additional tactile sensory feedback about current postural sway despite anticipatory postural adjustments preceding the perturbation. A possible explanation could be that anticipation of the perturbation with light touch contact results in a different set of postural adjustments which more quickly restore the desired postural state.

There was a significant effect of loading condition on SD $dCoP_{AP}$ and SD $dCoP_{ML}$ before perturbation. Before perturbation variability was less in reflex pull than in voluntary pull. One possible reason is that these two conditions were not equivalent in terms of light touch contact. During reflex pull, participants lightly contacted the manipulandum with their digits to be able to detect load onset, whereas in the case of voluntary pull, contact with the digits was only made at the beginning of the pull. Thus in a sense, the initial postural state during reflex pull afforded two sources of light touch contact, the right hand and the left arm. Dickstein (2005) reported that bilateral light touch with the index fingers of both hands is more efficient in reducing postural sway than unilateral light touch with only one index finger as commonly used. Krishnamoorthy et al. (2002) reported that the exact positioning of unilateral light touch contact on the body affects the degree of sway reduction; however, the effect of bilateral light touch at homologous and nonhomologous body positions has not been reported. Although not designed for this purpose, our study is the first that indicates sway decreases with additional sources of light touch contact positioned on nonhomologous, bilateral parts of the body. The mechanism underlying integration of multipoint light touch contact information is one that we feel deserves further study.

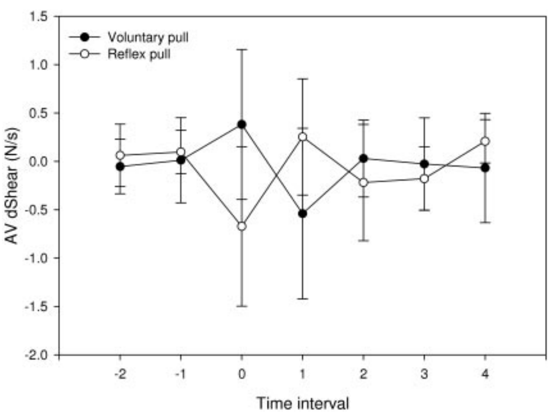

FIG. 6. Temporal evolution of the average shear force rate (AV dShear) during voluntary and reflex pull for 1-s intervals from 2 s before to 4 s after perturbation. Error bars represent standard deviations.

The reduction of AV $dCoP_{AP}$ and the tendency for reduced SD $dCoP_{AP}$ variability seen during perturbation with voluntary pull compared with reflex pull (see Figs. 3 and 4) stand in contrast to a significantly higher load force rate (dLoad) in the voluntary pull condition. This suggests that predictive balance processes, which are indexed by the increase in AV and SD $dCoP_{AP}$ and CoP_{ML} in the 1-s window before perturbation, effectively reduce the postural adjustment required to keep balance during the voluntary pull perturbation (even though the perturbation magnitude is greater in voluntary pull than reflex pull).

Our interpretation of loading effects on AV and SD dCoP emphasize a contrast between the predictability of balance perturbation due to voluntary action with changes that lead load onset, and the more uncertain effects of an imposed disturbance, with changes after load onset. However, what is not clear from our results is whether these effects are fixed or develop, for instance, with familiarity with the task. Although we employed a blocked design that might have lent itself to an analysis of trial effects, relatively few trials per block were run so our study design would be insensitive to such effects and thus this question is left for future research.

It is interesting to note differences between voluntary pull and reflex pull in the time course of the average shear forces (AV dShear) during light touch after perturbation (see Fig. 6). A possible explanation is provided by Yamazaki et al. (2005), who investigated CoP sway, but also bilateral trunk and thigh muscle activity, during rapid bilateral, asymmetrical changes of arm posture (right shoulder flexion and left shoulder extension) during upright stance. These rapid upper extremity movements resulted in a biphasic clockwise-anticlockwise upper trunk rotation, which were counteracted by early hip and thigh muscle contraction, confirming earlier findings of anticipatory postural adjustment preceding voluntary arm movements (Bouisset and Zattara 1981, 1987; Marsden et al. 1981). Further, large ML CoP variations were observed that were related to the rotation of the trunk. These results suggest that the antiphase directional fluctuations in shear force for voluntary and reflex pulls at the shoulder contact (see Fig. 6) originate from small counter-directional trunk rotations caused by the opposite directions of the right wrist movements.

Suppression of trunk rotation to maintain light touch might have "constrained" the postural control system, which limited the execution of the self-imposed voluntary pull perturbation. Thus the requirement of maintaining light touch contact may have acted as an additional task constraint to the postural control system during the preparation and execution of voluntary movement. A similar conclusion can also be drawn from the finding of Dickstein et al. (2003) that light touch results in a shift of mediolateral CoP velocity toward the contact plate during an external perturbation, an effect that was increased with stronger contact forces exerted by the participant.

Light touch contact is traditionally considered to enhance self-motion perception of body movements during upright stance and therefore to result in a reduction of postural sway. The location of light touch contact might serve as a spatial referent that affects body position sense based on proprioceptive information (Rabin and Gordon 2004; Reginella et al. 1999) and as a direct source of self-motion perception through transient shear forces at the contact point in the absence of a fixed spatial referent (Krishnamoorthy et al. 2002; Rogers et al. 2001). Both accounts imply that postural adjustments follow any force changes at the contact point (Jeka and Lackner 1995; Rabin et al. 2006). In contrast, if anticipatory postural adjustments are performed to maintain light touch, then one would expect postural responses to precede any changes of the contact force. However, such a temporal relation between light touch contact force and postural adjustments has not been reported up to now. We are now investigating if the effect of light touch contact switches from a facilitating to a "constraining" function under different postural and stimulus contexts and if the temporal relation between contact forces and postural responses is inverted.

The nature of the neural mechanisms contributing to the stabilization of upright stance is topic of continuing controversy. On one hand, it has been proposed that the intrinsic stiffness properties of the ankle are sufficient to ensure postural stability during quiet upright stance (Winter et al. 1998, 2001, 2003). On the other, Loram and Lakie (Lakie et al. 2003; Loram and Lakie 2002a,b) suggested that some neural mechanisms need to be actively involved in the control of ankle stiffness to modulate postural sway and to keep body balance stable (see also Morasso and Schieppati 1999). Further, this active neural intervention in the control of postural sway was assumed to be anticipatory (Fitzpatrick et al. 1996; Lakie et al. 2003; Loram and Lakie 2002a,b). In this context, the reduction of CoP variability during light touch as reported in the present article might be attributed to control mechanisms increasing ankle stiffness through muscle cocontraction, thereby reducing the movement degrees of freedom at the hip and ankle level to maintain light touch contact. Future research could examine ankle plantar and dorsi-flexor activity to evaluate this hypothesis.

Nevertheless, it is unclear on which neural level the presumed modulation of postural sway might take place. An account favoring lower level processing might assume that light touch at the shoulder changes the gain of the postural feedback loop by augmenting proprioceptive information relative to vestibular information during perturbation of upright stance with closed eyes (e.g., Ishida et al. 1997). However, light touch contact at the shoulder might serve as a stimulus that is processed by supraspinal neural circuits. For example, Jeka and Lackner (1995) inferred from the timing differences between finger tip contact forces, postural sway, and leg muscle activity that either long-latency reflex pathways or conscious anticipation might be involved in the control of postural sway.

The view that the facilitating function of light touch in postural control derives from its function as a "constraint" also implicitly assumes higher level neural processing of the light touch contact. For example, Riley et al. (1999) demonstrated that light touch only reduces variability of postural sway if it is declared relevant to the task. Thus only participants who were instructed to precisely control light touch showed a reduction of sway variability compared with a no touch condition. This finding was subsequently corroborated in a study (McNevin and Wulf 2002) that showed that only an external attentional focus that is directly related to the finger lightly touching an object resulted in reduction of postural sway. Thus light touch may be considered a constraint to the postural system in the sense that it defines a limit on body sway to preserve light touch. If keeping light touch close to a set value or within a certain range acts as the goal of a "supra-postural task," predictive control processes, the function of which is to reduce variability of light touch, can be assumed to be in operation. Also during self-imposed perturbations, the predictive control processes will serve to anticipate the disturbance and minimize the impact on balance. Maintaining light touch during a voluntary movement increases the coordinative complexity of the postural task thereby abolishing the facilitating effect of light touch.

In our present study, we used a static spatial referent to provide light touch contact at the left shoulder. Our findings can be interpreted as light touch affecting participants' postural responses during the perturbation as a constraint of the postural goal state that has to be taken into account when preparing appropriate postural adjustments in response to a perturbation of balance. To discriminate between an account that assumes that light touch improves stability through sensory feedback of small transient shear forces at the contact spot and an account that suggests that light touch imposes a constraint to the postural control system during a perturbation of balance, it seems reasonable to follow the suggestion of Krishnamoorthy et al. (2002), who demonstrated that given strong enough shear forces the effect of light touch is not linked to the availability of a fixed spatial reference. We are currently testing the effect of attaching a tactile stimulator to apply shear forces to the skin which would provide information about body sway. By adjusting the spatial gain of the stimulator the informational nature of the force feedback on balance stability will be explored.

We conclude that the immediate response to a voluntary perturbation and the stabilization after both a voluntary and a reflex perturbation are altered by light touch. To maintain light touch, CoP fluctuations are differentially modulated in AP and ML directions after a voluntary perturbation. Thus light touch influences CoP fluctuations not only by providing a sensory spatial reference but also by constraining the movements of the body after a perturbation.

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GRANTS

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Human bipedal instability in tree canopy environments is reduced by "light touch" fingertip support

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Whether tree canopy habitats played a sustained role in the ecology of ancestral bipedal hominins is unresolved. Some argue that arboreal bipedalism was prohibitively risky for hominins whose increasingly modern anatomy prevented them from gripping branches with their feet. Balancing on two legs is indeed challenging for humans under optimal conditions let alone in forest canopy, which is physically and visually highly dynamic. Here we quantify the impact of forest canopy characteristics on postural stability in humans. Viewing a movie of swaying branches while standing on a branch-like bouncy springboard destabilised the participants as much as wearing a blindfold. However "light touch", a sensorimotor strategy based on light fingertip support, significantly enhanced their balance and lowered their thigh muscle activity by up to 30%. This demonstrates how a light touch strategy could have been central to our ancestor's ability to avoid falls and reduce the mechanical and metabolic cost of arboreal feeding and movement. Our results may also indicate that some adaptations in the hand that facilitated continued access to forest canopy may have complemented, rather than opposed, adaptations that facilitated precise manipulation and tool use.

Increasing recent evidence is challenging the long held concept that the evolution of bipedalism in early hominins was a key factor that resulted in a permanent shift from arboreal to terrestrial life. Instead ancestral bipedal hominins appear to have continued to exploit tree canopy habitats well into our own genus *Homo* (e.g. refs 1–4). However, human bipedal stance is an inherently unstable posture[5] and the forest canopy is highly dynamic, which presents serious challenges for bipedal balance and movement. Together these raise fundamental questions about how bipedal hominins managed to exploit arboreal habitats with increasingly modern morphologies, which are central to understanding the environmental influences that have shaped modern human anatomy.

In humans the Centre of Mass (CoM) of the body lies further forward than the ankles so that, even in quiet standing on a stable support, the muscles of the calf must exert a torque to stop the body toppling forwards[5]. This torque may need to be significantly increased when balance is disturbed. Moreover, the central nervous system perceives self-motion and motion of the environment simultaneously via sensory cues from vision, the vestibular system, proprioception in the leg muscles, and tactile information from the soles of the feet[6]. Deprivation of one of these sensory cues or conflicting messages between them causes conspicuous instability and body sway[7,8]. Thus bipedalism for early hominins exploiting tree canopy habitats would have been particularly challenging since branches typically flex under the body mass of large animals, which destabilises the body[9,10]. The visual environment of forest canopy is also dynamic and unpredictable as branches move in the wind and under the weight of other animals. This feature of forest canopy has not been considered previously as influential on arboreal balance in humans or other primates. However, it has been established that irregularly-moving virtual visual environments are particularly challenging to human balance, and cause the central nervous system to initiate inappropriate muscle activation patterns while it distinguishes between movement of the body and movement of the environment[11,12].

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Non-human great apes can circumvent the problems of arboreal balance by hanging from their long hands or by using arboreal bipedalism, stabilised by their long prehensile toes that can grip to oppose the toppling moment experienced when standing on a thin branch[9, 13]. In contrast many early hominins had short hands that were unsuitable for prolonged suspension[14–17] and while the foot of *Ardipithecus ramidus* and the Burtele foot (BRT-VP-2/73, Woranso-Mille, Ethopia) indicate that these species had some residual gripping ability[18, 19], the foot morphology of *Australopithecus afarensis* and the Laotoli footprints suggests that from 3.66 million years ago (MYA) many hominins were exploiting the forest canopy with essentially modern human-like feet that were not capable of gripping branches[20, 21].

The aim of this study was to investigate how modern humans achieve bipedal stability in a controlled environment that embodies both the physical and visual challenges of forest canopy environments. We propose that a sensorimotor mechanism based upon "light touch" could have enabled early bipedal hominins to counter the physical and visual challenges of forest canopy.

Light touch refers to sensory feedback received from the surface of the fingers[22]. It has been suggested that bipedalism could be mechanically more stable in the forest canopy than on the ground because it would be possible for bipeds to use their hands to hold branches to aid balance[23]. But the flexibility of most available hand supports in forest canopy is instead likely to destabilise the body if significant loads are placed upon them. In contrast, touching a surface very lightly with a finger, similar to the touch used in Braille letter recognition[24] provides a sensorimotor feedback strategy that can enhance human stability without the large forces required for mechanical support. This light touch effect is produced by cutaneous receptors sensing small differences in shear forces through skin deformation[5, 25] and it substantially increases postural stability on firm supports during quiet standing and after a perturbation to balance[8, 25–27]. It has also been shown to improve gait efficiency by reducing muscle activity in the lower leg during treadmill walking before and after a gait perturbation[28]. However, no study has assessed if it is beneficial in environments where participants are exposed simultaneously to challenging physical and visual environments, and where the structures available for the provision of tactile information are highly flexible.

To investigate human stabilisation mechanisms in arboreal-like habitats we studied body sway and leg muscle activity of human participants standing barefoot on a cantilevered springboard. We first compared body sway responses and leg muscle activation levels of humans in quiet standing on the springboard when it was firm (where a solid chock replaced the springs) and when it was compliant. For compliant trials we then applied a mechanical vertical perturbation to the springboard to destabilise the participants. Secondly, we tested the effects of light touch on the participant's postural stability and muscle activity levels during quiet standing and stabilisation after the mechanical perturbation. For all conditions we further exposed participants to different visual environments using a visual display system onto which images were back-projected (see Fig. 1). The visual environments were a static visual environment (SVE) consisting of a single frame of a video of the branches of a leafy tree, a dynamic visual environment (DVE) in which they viewed the video of the swaying branches, and wearing a blindfold in which there was no visual environment (NVE).

Results

Impact of dynamic physical and visual environments on quiet standing. The effect of viewing the dynamic visual environment was to destabilise the body as much as having no visual information available (Table 1; Fig. 2a and Fig. 3). When standing on the solid support the variability of antero-posterior body sway velocity measured at the 7th cervicular vertebrae in the neck (SD dC7, hereafter 'body sway'), which is a strong indicator of the postural control system's "effort" to stabilize balance[29], was 22% greater for both DVE and NVE compared to the SVE (Fig. 3). Moreover, the interaction between challenging visual and support conditions further decreased stability since for NVE and DVE conditions body sway was significantly increased when standing on the compliant support compared to the solid support (20% for both conditions), whereas participants were equally stable on both surfaces when they viewed the SVE (Table 1; Fig. 3). Activity levels in all tested muscles were unaffected by visual and support conditions in quiet standing.

Impact of dynamic visual environments after a support perturbation. The participants body sway increased by 400% when they experienced the mechanical perturbation (Fig. 2a). Body sway began to stabilise after 2 seconds (i.e. in the 2nd time bin after the perturbation), but had not returned to the level of quiet standing after 6 seconds (Table 2 and Fig. 2a). The perturbation also affected muscle activation levels in the vastus lateralis (Fig. 4a) and the rectus femoris (the latter via an interaction with time) (Fig. 4b). Activity in the vastus lateralis was significantly higher for the stabilisation phase when viewing the DVE than when viewing the SVE (29%) or when participants were blindfold (22%) (Fig. 4a). Peak levels of rectus femoris muscle activity occurred within the perturbation time bin for the SVE and DVE, but one to two seconds after the perturbation (time bin 1) when no visual information was available (Fig. 4b). Thereafter, rectus femoris activity was lowest for the SVE trials. It reduced rapidly in the NVE trials to the same final level as for the SVE, but was significantly higher for the DVE immediately following the perturbation (15%) and in the 3rd and 4th time bins after it (26% and 45% respectively) (Fig. 4b).

The effect of light touch. The average resultant light touch force for the hand was 0.29 N (SD: 0.31 N) during the quiet standing phase and 0.33 N (SD: 0.33 N) during the stabilization phase of the trials. Lightly touching the compliant hand support reduced body sway in quiet standing by 24% compared to when participants had to balance without touch (Table 1, Fig. 5a). After the perturbation on the compliant support body sway was affected by the interaction between light touch and vision (Table 2): light touch significantly reduced sway for all visual conditions, however its effect for the DVE and NVE was substantially larger (22% and 29% respectively) than

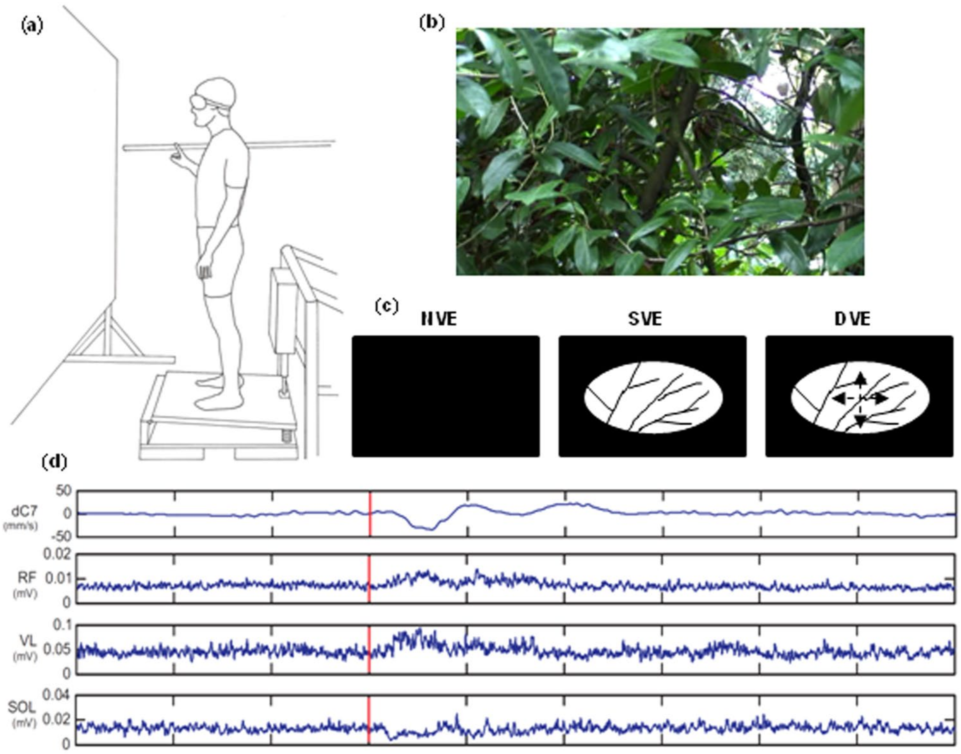

Figure 1. The experimental setup. (**a**) a participant on the springboard. The right arm is in a raised posture in contact with the compliant hand support. An actuator system is tracking the motion of the springboard ready to deliver a vertical perturbation. The participant wears goggles restricting their field of view to a back projection screen. (**b**) A still frame from the video used for the experiment. (**c**) The three visual environment conditions: NVE: no visual environment (participants wore a blindfold); SVE: static visual environment (a still frame from the video of the branches of a leafy tree); DVE: dynamic visual environment (the video of the branches). (**d**) Participant-averaged sample data traces for the velocity of body sway (dC7) and EMG from the rectus femoris (RF), vastus lateralis (VL) and soleus (SOL) muscles from 3 s before to 6 s after the perturbation. The solid red vertical line indicates the time point of the perturbation.

Position/ Measure	P value					
	Touch F(1,6)	Vision F(2,12)	Compliance F(1,6)	Touch x Vision F(2,12)	Touch x Compliance F(1,6)	Vision x Compliance F(2,12)
Body sway	<0.001	<0.001	0.03	NS	NS	0.02
EMG_{RF}	0.02	NS	NS	NS	NS	NS
EMG_{VL}	NS	NS	NS	NS	NS	NS
EMG_{SOL}	0.04	NS	NS	NS	NS	NS

Table 1. Results from a GLM comparison of muscle activation and postural sway measures for quiet standing conditions in the antero-posterior direction. Body sway was measured at the level of the base of the neck (7th cervicular vertebrae). NS: no significant difference. Three-way interactions were not studied due to the sample size.

for the SVE (11%), while its impact in DVE and NVE did not differ significantly from each other (Fig. 5b). Light touch had no effect on the time it took the participants to stabilise (Table 2).

Light touch also affected muscle activity levels. Soleus muscle activity was significantly elevated (6%) in light touch compared to no touch trials in quiet standing (Table 1, Fig. 5c), but there was no effect after the perturbation (Table 2). This effect was reversed for the thigh muscles. Light touch significantly reduced rectus femoris activity levels in quiet standing (Table 1) and both rectus femoris and vastus lateralis activity levels (the latter via an interaction with time) after the perturbation (Table 2), compared to no touch trials. Rectus femoris activity was 23% less in light touch trails than in no touch trials during quiet standing (Fig. 5d). After the perturbation it was 32% less active with light touch than without (Fig. 5d). Vastus lateralis ranged from 30–32% less active in light touch than no touch trials after the perturbation (Table 2, Fig. 5e).

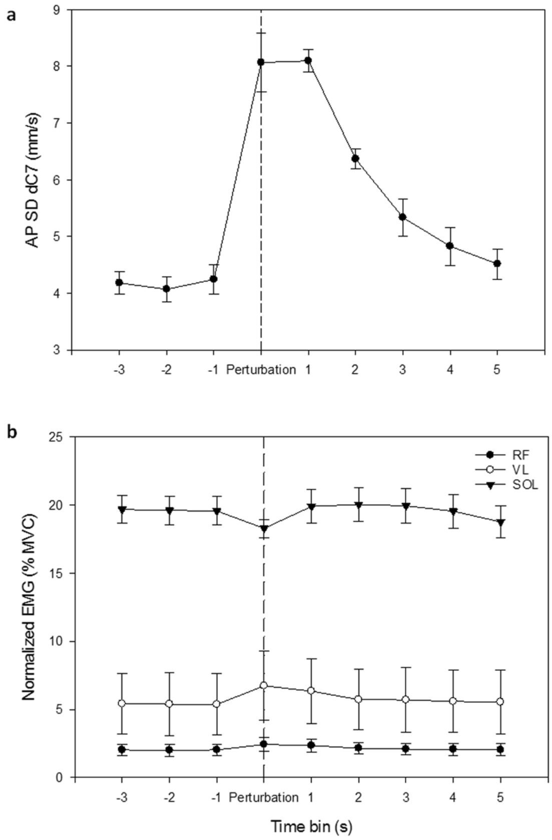

Figure 2. Time series for all participants across all support and visual conditions for measured variables. (**a**) the variability of antero-posterior velocity of body sway (SD dC7) and (**b**) average muscle activity (normalised to the participant's maximum voluntary contraction, MVC). The muscles are rectus femoris (RF), vastus lateralis (VL) and soleus (SOL). Time equates to 1 s time bins from 3 s before to 6 s after the perturbation (see methods for description of time bins). Error bars represent standard errors.

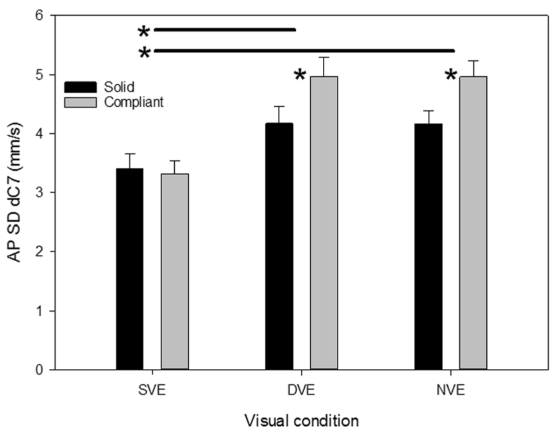

Figure 3. The effect of different visual environments and substrate compliance on antero-posterior (AP) body sway during quiet standing before the mechanical perturbation. Body sway is presented as the standard deviation (SD) of the variability of body sway. SVE: static visual environment; DVE: dynamic visual environment; NVE: No visual environment. *$P < 0.05$. Horizontal black lines with asterisks indicate significant post-hoc comparisons between visual conditions depicted on the horizontal axis, while single asterisks indicate significant post-hoc comparisons between both surface conditions within a specific visual condition.

Discussion

We present novel empirical data that quantifies the impact of the physical and visual challenges characteristic of forest canopy on postural stability in modern humans. We show that light touch makes a substantial difference to

Measure	P value					
	Touch F(1,6)	Vision F(2,12)	Time F(5,30)	Touch x Vision F(2,12)	Touch x Time F(5,30)	Vision x Time F(10,60)
Body sway	0.001	<0.001	0.001	0.002	NS	NS
EMG_{RF}	0.03	NS	0.01	NS	NS	0.04
EMG_{VL}	0.03	0.009	0.01	NS	0.04	NS
EMG_{SOL}	NS	NS	0.03	NS	NS	NS

Table 2. GLM comparison of muscle activation and postural sway in the antero-posterior direction for post perturbation conditions. Body sway was measured at the level of the base of the neck (7th cervicular vertebrae). NS: no significant difference. Three-way interactions were not studied due to the sample size.

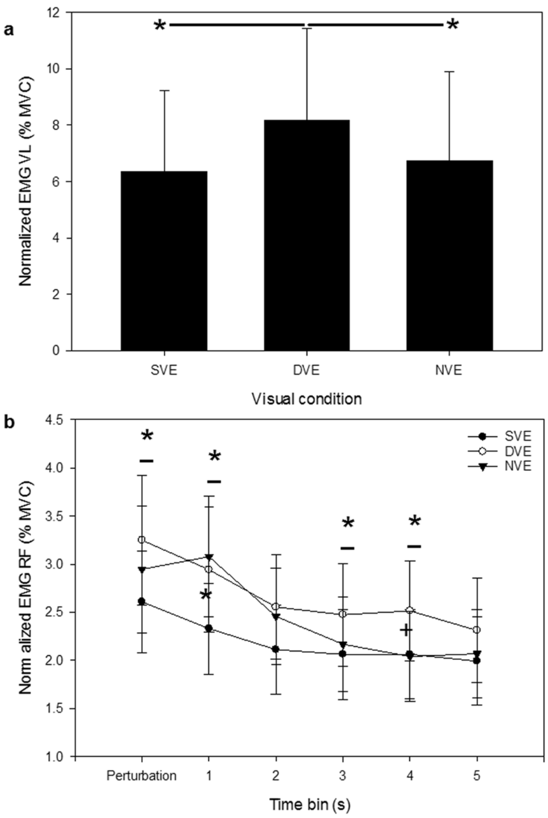

Figure 4. Influences on muscle activity after the perturbation. (**a**) The effect of different visual environments on vastus lateralis muscle activity after the mechanical perturbation. (**b**) The effect of visual environments and time on rectus femoris activity levels after the mechanical perturbation. Muscle activity is presented as the percent of the participants' maximum voluntary contractions (MVC). Horizontal black lines with asterisks indicate significant post-hoc comparisons between visual conditions. SVE: static visual environment; DVE: dynamic visual environment; NVE: No visual environment. See methods for description of time bins. *$P < 0.05$.

human's basic ability to balance in forest canopy environments. Such ability could have underlain our ancestor's success in arboreal locomotor, foraging and predator avoidance behaviours.

The results firstly confirm that, like virtual abstract visual environments[11, 12], the visual environment of forest canopy does significantly destabilise humans. The impact on postural stability of the dynamic forest environment combined with standing on the compliant support was as severe as when the participants wore a blindfold. When the visual, vestibular and somatosensory senses provide unreliable and potentially conflicting information, central mechanisms can employ multi-sensory re-weighting to prioritise the input that offers the most reliable source of sensory information about own body sway[30]. The similarity in the sway response for being blindfold and viewing the dynamic visual environment suggests that vision was 'downweighted' in the latter to reduce its destabilising impact, amounting to a de facto deprivation of visual feedback. For forest canopy conditions however, this clearly creates a problem because vestibular and proprioceptive information on body sway will also be compromised by the compliance of available weight bearing branches; indeed we found that body sway was significantly higher for both DVE and NVE on the compliant support compared to the stiff support (Fig. 3). It also had a substantial effect on thigh muscle activity since rectus femoris was 40% more active (averaged over all

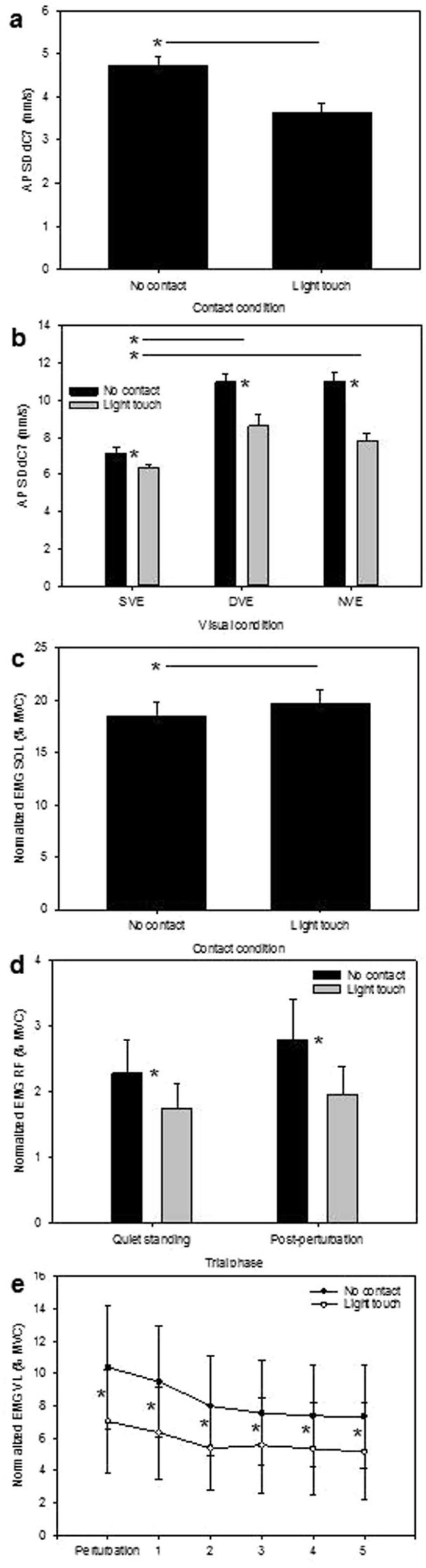

Figure 5. The effect of light touch on: (**a**) body sway in quiet standing, (**b**) body sway after the perturbation according to visual condition, (**c**) soleus activity in quiet standing, (**d**) rectus femoris activity in quiet standing and after the perturbation and e) vastus lateralis activity after the perturbation, according to time. MVC: maximum voluntary contractions. SVE: static visual environment; DVE: dynamic visual environment; NVE: No visual environment. See methods for description of time bins. *$P < 0.05$. Horizontal black lines with asterisks indicate significant post-hoc comparisons between major conditions depicted on the horizontal axis, while single asterisks indicate significant post-hoc comparisons between subconditions within a specific major condition.

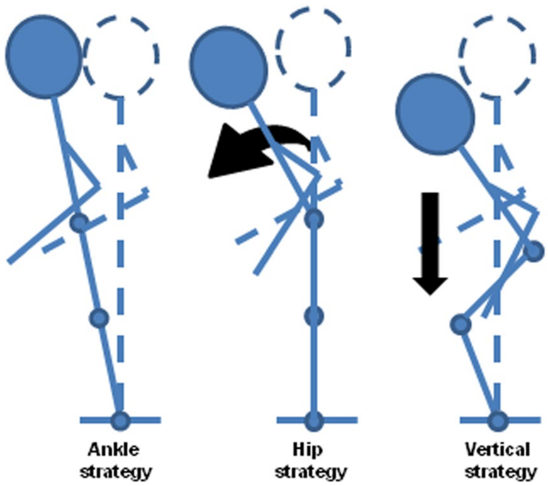

Figure 6. Postural response strategies for maintaining balance following a perturbation (modified after[35]). In the ankle strategy the toppling moment on the body is countered by a torque around the ankle joint, with the upper body acting as a single inverted pendulum. In the hip strategy a double inverted pendulum is created by an additional torque at the hip. In the vertical strategy all lower limb joints flex to control vertical height to counter the toppling moment.

time bins) and vastus lateralis 29% more active in DVE than SVE after the perturbation, and 17% (RF) and 22% (VL) more active in the DVE than the blindfold condition (Fig. 4a and b). Thus, the central nervous system in this experiment reacted in a similar way to its response in virtual visual environments[11,12]; by initiating inappropriate muscle activation patterns while it distinguished between movement of the body and movement of the naturalistic environment.

Previous studies have shown that postural response strategies for maintaining balance following an external perturbation differ according to the environmental context, such as the nature of the threat to postural stability, the available sensory feedback and the specific features of the support[31]. The 'ankle strategy' is the most common strategy for controlling body sway in the anterior-posterior direction[32] particularly on large, flat supports. In this strategy the toppling moment is countered mainly by the production of a torque around the ankle joint, with the upper body behaving as a single inverted pendulum[33]. In the forest canopy this might be encountered when standing along a single large but flexible branch; along two narrower branches (one per foot) or on 'webs' of intermingled narrow branches that form small, flexible platforms (all these are commonly used by wild orangutans[13]). When the postural context becomes more challenging a hip strategy may be also recruited, which creates a double inverted pendulum allowing for additional leaning of the upper body to stabilise body sway[34], or a vertical strategy where the hip, knee and ankle flex to control vertical height[35] (Fig. 6). In forest canopy these are likely to be elicited by standing astride one or more narrow branches. In this study, the soleus was the most active muscle in all trials (Fig. 2b), which suggests that the ankle was important in maintaining postural equilibrium throughout the experiment. However, the increased activity levels for RF and VL immediately after the perturbation (Fig. 4), and particularly in the DVE compared to both SVE and NVE show that forest canopy environments are sufficiently challenging to warrant more complex hip and vertical stabilisation mechanisms, which probably relate to an immediate response strategy to dampen the vertical oscillations of the branch after it has been disturbed.

We subsequently tested whether light touch on a compliant hand support might provide sufficient additional proprioceptive and cutaneous feedback from the fingers to reduce the destabilising effect of the dynamic physical and visual environment without destabilising the body by displacing the hand support. Even though we made the hand support in this study highly flexible (with an effective stiffness of only 1.17 N/mm at the participant's point of contact with the pole), we found that light touch reduced body sway by 24% in quiet standing, independent of the visual and support conditions (Fig. 3). After the perturbation it significantly reduced sway for all visual conditions, but particularly for the dynamic visual environment and when blindfolded (22% and 29% respectively, Fig. 5b). In studies in which participants are able to lightly touch a solid hand support, it is commonly found that light touch decreases postural sway to the level found for the non-challenging condition, such as in dark compared to lit conditions[36,37]. In this study, light touch countered just under two thirds (61%) of the combined destabilising effect of the dynamic visual environment and compliant support on postural sway (measured as the difference between the dynamic and static visual environment for no touch trials after the perturbation, Fig. 5b). This presumably reflects the complexity of forest canopy that creates multiple, simultaneous challenges for the sensorimotor system.

It might be thought that the participants could have been balancing with light mechanical touch rather than sensory support from the fingertips[38]. A rough calculation clearly shows this was not the case. The 7th cervical vertebra moved with a peak velocity of 0.03 m sec^{-1} in the second immediately after the perturbation. This created a maximum possible displacement within that second of 16 mm at the height of the CoM, and hence caused a maximum toppling moment of about 12 Nm (calculated as displacement in m sec x gravity x participant's body mass). In contrast, the hand forces, also exerted within 1 s after the perturbation, averaged around 0.3 N.

The fingertip touched the hand support at shoulder height (1.5 m), and therefore acted at a large mechanical advantage. Nevertheless it would only have generated a mean restoring moment of about 0.5 Nm; just 4% of the toppling moment caused by the participants body sway. Thus the sway reductions seen with light touch cannot be explained by the amount of mechanical support and must have been caused largely by augmented sensory feedback giving faster feedback that the person was overbalancing, so allowing the ankle and hip muscles to respond quicker and reduce the forces they need to apply.

The relative impact of light touch was most powerful for activity in the upper leg where it reduced rectus femoris and vastus lateralis muscle activity by >30% after the perturbation(Fig. 5d and 5e). Activity levels in these muscles were lower overall than the soleus. However, our use of a flat surface rather than a branch structure was a simplified experimental paradigm that enabled us to compare relative underlying stabilisation mechanisms in different environmental conditions by ensuring that participants did not fall in the most challenging trials. It will nevertheless likely underestimate the instability and muscle activation levels that would be generated around the hip when standing on curved branch-like structures, which are likely to generate greater hip contribution and vertical strategies to control the stability of body balance. Thus if these muscles are representative of the quadriceps as a whole, and particularly if the light touch effect extends to locomotion, then light touch could be central not just to enhancing balance and avoiding falls in forest canopy habitats, but to significantly reducing the mechanical and metabolic cost of arboreal bipedality.

No study has directly quantified whether non-human great apes employ light touch to balance in the forest canopy. Orangutans exhibit higher levels of arboreal bipedality than the other great apes but the forelimbs appear to support more than their own weight in the majority of bipedal bouts[13, 23]. Chimpanzees also rely strongly on gripping feet and hand assistance[39] during arboreal bipedality, suggesting neither species regularly utilises light touch in this behaviour. This may be because the long length of their arms, hands and fingers (particularly in orangutans) means that they can reach further to find suitable supports for the hands. They can also grip multiple small branches at once, which should provide a stiffer hand contact than the single flexible branches likely available to short-handed bipeds. Gorillas however are more similar to hominins in their short finger proportions (when scaled to body mass)[40] and in their foot morphology[41]. Moreover they have the largest body mass of all great apes, which will increase the compliance of the branches used to bear their weight. This indicates they may experience somewhat similar balance challenges as hominins in forest canopy, and may therefore also employ light touch during arboreal bipedalism. Unfortunately of all the modern great apes, their arboreal locomotion is the least well documented.

The extent to which the performance of ancestral hominins in the forest canopy was compromised by climbing and clambering with modern foot morphology is currently unknown[42]. There is, however, strong evidence that *Au. afarensis* exploited both terrestrial and arboreal habitats[43], despite possessing transverse and medial arches in the foot[21] and modern human foot function, albeit less strongly expressed than in ourselves[20]. Other australopithecines such as *Au. africanus* (3–2 MYA) and *Au. sediba* (1.98 MYA) have also been shown to be competent terrestrial bipeds that retained a significant degree of arboreality[2, 3]. Within *Homo*, *Homo naledi* (date unknown) combines adaptations of the shoulders and hands that appear well suited for climbing with human-like features of the feet and lower limbs[44–46], while cross sectional bone strength measurements on the humerus and femur indicate that *Homo habilis* (*circa* 1.8 MYA) also combined terrestrial bipedalism with frequent arboreal behaviour[47]. Nevertheless, ancestral hominins that retained short hands[40] whilst undergoing adaptation of the feet for terrestrial bipedalism would need to evolve mechanisms to counter the instability caused by both the physical and visual dynamics of forest canopy if they were to maintain exploitation of forest resources without grasping feet. We suggest light touch, as a sensorimotor strategy, could have substantially enhanced balance stability without pedal grip to have improved safety, decreased the risk of falls, and decreased (or at least prevented large increases in) the mechanical and metabolic cost of arboreal locomotion.

In this scenario light touch would enhance balance during horizontal locomotion along and between flexible branches in the tree canopy, during foraging and in other arboreal behaviours. Nevertheless, hominins would still have needed to transition between the forest canopy and the ground using vertical climbing and descent. The curved phalanges, that are present in at least some fossil hominins (such as *Au. afarensis, Au. sediba* and *H. naledi*[14, 45]), may well have enhanced efficacy and safety in this behaviour.

Our results may also have implications for the evolution of hominin hand morphology and sensorimotor functions of the central nervous system. Although all apes are capable of making contact between the tip of their thumb and their fingers, and thus forming precision grips, the ability to form pad to pad precision grips in which objects are held delicately yet securely between the proximal pulp surfaces of the thumb and the finger tips is present only in humans[48–50]. It has commonly been asserted that the precise manipulative hand morphology required for lithic tool use could have only been attainable after the hands had been freed from the constraints of arboreal locomotion. However, there is increasing evidence that early hominins, such as *Au. africanus* and *Au. afarensis*, were capable of forceful pad to pad precision grasping, even prior to the appearance of stone tools in the archaeological record[15, 51]. We suggest that hominin fingers might have been under selective pressure for light touch as an aid to balance in parallel with selection for the ability to perform forceful precision grips. Indeed, it is highly likely that fingers capable of using light touch are linked functionally to fingers capable of generating the high precision forces required for tool use because both rely on mechanoreceptive afferent fibers in the glabrous skin of the hand for tactile acuity. These fibers include fast adapting types associated with Meissner corpuscules and Pacini receptors, and slow adapting types associated with Merkel receptors[52]. Meissner corpuscles are thought to be particularly associated with tool use because they provide important sensory feedback for the effective control of grip and are especially numerous in the fingertips[53, 54]. Both fast adapting receptor types and Merkel cells have been shown to be integral to light touch, because their small receptive fields enable transmission of spatial details with a relatively high resolution[52, 55, 56].

Comparisons between primates so far have, however, not revealed differences in the size or density of Meissner Corpuscules that would reflect human's unique precision grip abilities[53, 54]. This may be due to methodological issues, for example, the use of an elderly human sample may have distorted the results since tactile acuity, particularly at the fingertip, deteriorates with age[52]. In addition Merkel cells were not studied, but these are also numerous in the fingertips and exhibit greater sensitivity than fast adapting receptors to non-uniform spatial features on objects (gaps, edges and curvatures)[57]. They are therefore considered to be critical for form and texture perception at the fingertip[52, 57]. However, a comparison of younger and older adult humans also showed that while reduced tactile sensitivity correlated with increased contact forces during light touch stance, the sway reduction by light touch itself did not vary with the contact force[52, 56]. Together these observations indicate that tactile sensitivity alone does not predict ability for the utilization of light touch for balance or for precision grip. Such features must be viewed in parallel with the higher order cognitive functions that process motor and tactile information of the hands (e.g. see ref. 58), integrate this information with other sensory cues, and/or resolve conflicting sensory messages in a specific postural context.

In summary, our data allow a unique insight into the sensory ecology of ancestral bipedal hominins. They add weight to the argument that exploitation of arboreal resources situated on peripheral, flexible branches would have been possible for hominins, despite their increasingly modern foot morphology. They may also indicate that some adaptations in the hand that facilitated continued access to forest canopy habitats may have complemented, rather than opposed, adaptations that facilitate precise manipulation and tool use.

Methods

Participants and apparatus. Seven, right-handed males served as participants [age 23 ± 2 (SD) years, height 180 ± 5 (SD) cm, weight 70 ± 3 (SD) kg]. None reported any musculoskeletal or neurological disorders and all refrained from alcohol for 24 hours before the experiment. All participants gave written informed consent and the study was approved by the research ethics committee in College of Life and Environmental Sciences at the University of Birmingham. All experiments were performed in accordance with their guidelines and regulations.

The participants stood barefoot on the cantilevered springboard that was limited to compliant motion in the vertical direction (Fig. 1) and was positioned 90 cm in front of a back-projected visual display. They stood with their feet spaced hip-width apart, at a self selected foot angle. The effective stiffness of the springboard was 4.08 N/mm (when loaded centrally), which is within the range of branch stiffnesses found in tropical forest trees[59]. Complaint branches can deflect in all directions, particularly in windy weather. However when loaded by the weight of a large bodied ape, by far the greatest deflections will be in the vertical direction. The participants stood facing the fulcrum of the springboard so they could not see when we applied a mechanical perturbation to the free end of the springboard at a random time in the experiment to destabilise the body.

Body sway was recorded by a 12 camera optoelectronic motion capture system (Oqus, Qualisys, Sweden) which tracked the position of a reflective marker attached to the skin overlying the 7th cervical vertebrae (C7). Electromyographic (EMG) data was recorded for the soleus (SOL), vastus lateralis (VL) and rectus femoris (RF) muscles in the right leg. Data was collected via Ag–AgCl surface electrodes with a 10 mm diameter conductive area and an inter-electrode distance of 20 mm (Dual Electrode, Noraxon, Scottsdale, AZ, USA, after[60]. EMG signals were transmitted wirelessly to an amplifier system (ZeroWire, Aurion, Italy), amplified ($\times 1000$), digitized, sampled at 1 kHz and stored together with the kinematic and CoP data for off-line analysis. All data streams were synchronised using a single common trigger and recorded using a single analogue to digital converter. The EMG system recorded at 1000 Hz and the camera system recorded at 200 Hz.

Once the participants were standing quietly the mechanical perturbation was introduced via a computer controlled linear actuator system (XTR 2504, Copley Controls Corp, USA) connected to the free end of the spring board via a single axis force transducer (F250, Novatech, UK). The actuator was controlled via custom software written within Labview 2009 (National Instruments, Newbury, UK) which used feedback from the sensor in order to track the movement of the board in a zero force mode. Once triggered the actuator applied a vertical load of 100 ± 3 (SD) N displacing the board vertically down, thereby generating a substantial downwards and small backwards perturbation to the participants. The actuator's TTL trigger signal was simultaneously recorded with the force and EMG data via the analogue to digital converter within the motion tracking system allowing the onset of the perturbation to be determined.

During the experiments participants wore goggles that limited their visual field to 76 ± 10 (SD)° in the horizontal plane and 72 ± 10 (SD)° in the vertical plane centred on their direction of forward gaze; this ensured that the visual display screen encompassed their foveal field of view and also part of the forward projecting extra-foveal field. The video provided a two dimensional representations of branch movement that eliminated the use of possible stereoscopic cues. We applied both these constraints to ensure that the most salient visual cues affecting balance were present in the participants' fields of view. Thus, although the impact of visual stimuli differs depending on whether environmental motion is detected within the central or the peripheral visual field, it is forward (foveal) vision that has the highest acuity and that underlies the detection of rapid object movements[61]. Also, the detection of the direction of movement of an object, and the time to contact, are provided by information extracted directly from the optical flow field without the necessity of stereoscopic cues[62]. Radial optic flow as caused by a looming object or during forward locomotion, however, has an effect on balance only when presented in the centre of the visual field[63, 64]. In contrast, laminar optic flow (parallel flow lines) has been shown to have impact on balance irrespective of the region in the visual field, both in the centre and the periphery[63, 64]. Laminar optic flow is much more prominent in the video for the DVE condition due to left-right, up-down and diagonal movements of the tree branches. For example, a prominent leafy branch in the centre foreground of the image moved with a lateral amplitude of 463 ± 38 (SD) mm at an inclination angle of 34 ± 8 (SD)° to the horizontal. Overall, the branches shown in the video moved in a multi directional pattern with semi regular frequency of

0.4 ± 0.05 (SD) Hz. Thus, the DVE condition induced considerable environmental noise in the visual channel and therefore should have had a pronounced effect on body sway.

To study the role of light touch a 21 mm diameter carbon fibre hand support was mounted at the shoulder (acromion) height of each participant on a 6 degree of freedom force sensor (Delta, ATI, Apex, NC, USA) in a cantilever arrangement, and positioned 310 mm to the side of the centre line of the spring board in parallel with the participant, to replicate an adjacent branch at approximately shoulder level height (Fig. 1). The attenuating effects of light touch are greater when the finger is positioned in the plane of greatest sway[27]. The participants therefore made contact with the hand support by placing the index finger of their right hand on a marker situated 450 mm in front of their body, which was found to be a comfortable location for all participants. When loaded at this point the pole had an effective stiffness of 1.17 N/mm. The contact point was not visible to the participant due to the visual field restrictions.

For the light touch experiments, participants were asked to maintain a "light touch with the hand support with just enough force to maintain contact". During tests with no touch conditions the participants were asked to maintain a similar posture to the touch conditions but to move their hand slightly to the side in order to avoid making contact with the pole. All possible combinations of the three variables (vision, touch and compliance) were tested resulting in 12 conditions, with 10 trials of each condition. Trials were presented in two counterbalanced blocks. Before testing, participants were encouraged to practise standing on the board as it was perturbed, although none asked to experience more than 4 perturbations. Participants were instructed to stand relaxed on the springboard without moving and were asked to say when they had reached a stable postural state, at which point the data recording for each trial would start. Each trial lasted 20 seconds. For trials on the compliant surface the participants were subjected to a single perturbation at a randomized time interval between 5 and 14 seconds after the start of the recording to ensure they were unaware of the exact timing of the perturbation.

Data Analysis. Individual data streams were analyzed using custom interactive software written in MatLab R2008a (Mathworks, Inc., Natick, MA, USA). The antero-posterior (AP) component of the kinematics of a marker placed on the 7^{th} cervicular vertebrae was digitally low-pass filtered at 10 Hz (dual pass 4th-order Butterworth filter) and differentiated to obtain rate-based measures of change per second (dC7).

To investigate the time course of the balance response following the springboard perturbation, each trial was segmented into bins of 1 s duration from 3 s before to 6 s after the onset of the perturbation. The within-bin standard deviation (SD) of the rate of change parameters was determined for each time bin in the AP direction[65]. The time course data for each trial was then divided into two phases: the baseline phase before the perturbation ($t < 0$ s) and the stabilisation phase after the perturbation ($t > 0$ s).

The onsets of the perturbations were determined using the actuators TTL trigger signal linked to the EMG. EMG recordings were band-pass filtered between 10 and 500 Hz, rectified and smoothed by a moving average with 15 ms width to obtain the EMG envelope. The EMG envelopes were normalized against the respective muscle activities during maximum voluntary contraction (MVC) of each specific muscle obtained for each participant[60]. Mean EMG values were extracted for the base-line phase before the perturbation[8]. For the post-perturbation stabilization phase muscle contraction onsets were defined as the time point at which the rectified EMG amplitude increased by 4 standard deviations above a mean baseline period[65].

The data for the base line and stabilization phases was subjected to repeated-measures analysis of variance (ANOVA), performed in SPSS 16. The significance threshold was set to $P = 0.05$ after Greenhouse-Geisser correction. Vision, compliance and touch conditions were primary independent factors and time course was an additional factor for the post perturbation analysis. Significant interactions were explored with post hoc single comparison.

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Author Contributions
S.K.S.T., L.J., S.R.L.C. and A.M.W. designed the study. G.R.M. advised on visual aspects of experiments. A.V.C., W.I.S., R.H.C. and A.R.E. advised on conceptual and biomechanical aspects. S.R.L.C. and L.J. collected the data. L.J., S.R.L.C. and S.K.S.T. analysed the data. S.K.S.T. and L.J. wrote the main manuscript. All authors contributed to the manuscript and gave final approval for publication.

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Interpersonal Light Touch Assists Balance in the Elderly

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ABSTRACT. Previous researchers have shown that light touch contact with a fixed object reduces body sway, whereas light touch with a moving object entrains and increases sway. Given the importance of interpersonal touch and, for example, its use in care for the elderly, it is interesting to ask whether light touch contact between two people reduces or increases sway? The authors measured ground reaction forces and trunk movements in 5 pairs of older participants (*M* age = 65.1 years, *SD* = 4.2 years) during quiet standing, when contacting another person using light touch at the index finger and during light touch with a fixed object. Postural sway was reduced in both light touch conditions, by 13% with interpersonal light touch and by 31% with the fixed object. A small but significant positive correlation in sway with near 0 phase lag during interpersonal light touch may reflect the role of anticipation in maintaining light touch. The authors conclude interpersonal light touch affords an interesting new paradigm for the study of balance.

Keywords: aging, body balance, interpersonal coordination, light touch

The risk of falls increases over age 65 years, with age-related decline in balance control being a likely contributing factor (Lord, Ward, Williams, & Anstey, 1993). Thus, caregivers usually walk close to an elderly person and lightly hold the elbow to allow support in case of a fall. It is likely that caregiver contact facilitates detection and a quick response, in case the older person exhibits excessive body sway suggesting an impending fall.

Balance during quiet standing is improved, especially in the elderly, by lightly touching a stable environmental referent, even though the contact provides no mechanical support (Baccini et al., 2007), or if the contact point can move in space (Krishnamoorthy, Slijper, & Latash, 2002). Passive light touch also improves balance (Rogers, Wardman, Lord, & Fitzpatrick, 2001). It is thought that improvements derive from shear forces at the skin that augment other cues to postural changes resulting in earlier or more accurate postural adjustments (Jeka & Lackner, 1995). We examine whether interpersonal light touch, such as that provided by a caregiver contacting an older person's arm, may benefit older people's balance. As an alternative outcome, we envisaged interpersonal light touch might result in an increase of postural sway. Previous researchers have shown light touch with a moving object can entrain and increase sway (Jeka, Oie, Schoner, Dijkstra, & Henson, 1998). Thus, we thought sway might be increased with movements of the contact point due to the other person's sway.

Method

Participants were 10 older adults (*M* = 65.1 years, *SD* = 4.2 years; 9 women, 1 man) who provided informed consent. Pairs of participants were tested in quiet, narrow, and bipedal standing, facing forward on separate force platforms (Bertec 4060H, Columbus, Ohio, USA) placed side by side, 0.5 m apart. The taller person stood on the left, extending the right arm forward, whereas the other person extended the left. Body sway was recorded with eyes open or closed under three touch conditions: (a) contact with the underside of a plate (30 × 20 cm) mounted on top of a single-column pedestal (i.e., light touch), (b) contact with the adjacent hand of the other participant (i.e., interpersonal light touch),[1] and (c) no contact with the hands pointed forward (i.e., no contact). In all three touch conditions, the arm posture was kept the same and the hand of the extended arm was usually held in supination. The exception was that during interpersonal light touch, the left person kept the right hand in pronation to touch the fingers of the right person from above. Participants were instructed to stand as still as possible but relax without speaking. Practice trials were performed to ensure that participants were aware of the need to keep touch contact as light as possible without losing contact.

Six 15 s trials each were recorded per condition, and the conditions were randomly ordered. Force platform data were sampled at 1,000 Hz to determine anterior–posterior and medial–lateral components of center-of-pressure (COP) fluctuations. Body movement at C7 was sampled at 200 Hz by optical motion tracking (Qualisys Oqus, Gothenburg, Sweden). Data were low-pass filtered at 10 Hz and differentiated to yield two separate rate of change measures of sway (*d*C7, *d*COP).

As a measure of within-trial sway variability the standard deviation of the two rate of change measures were calculated (*SD d*C7, *SD d*COP) and averaged across directions and trials for each condition. Mixed repeated-measures analysis of variance (ANOVA) with touch and vision conditions and sway direction as within-subject factors and position (standing on the left vs. right) as between-subject factor were computed for *SD d*C7 and *SD d*COP, with significance levels set at $p = .05$. The results for the two measures were similar, although *SD d*COP (*M* = 10.96 mm/s, *SD* = 2.83

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mm/s) was numerically larger than *SD d*C7 ($M = 5.76$ mm/s, $SD = 1.28$ mm/s). For brevity, we report sway in terms of *SD d*COP.

Coordination of sway between participants in each pair was investigated using between-individual cross-correlation functions for the interval +1,000 ms (right person leads) to –1,000 ms (left person leads). The largest absolute cross-correlation and corresponding time lags were extracted and averaged for each experimental condition. The correlation coefficients were Fisher-Z transformed and subjected to a similar mixed repeated-measures ANOVA, with touch and vision conditions and sway direction as within-subject factors.

Results and Discussion

Sway was greater with eyes closed than open, $F(1, 9) = 20.39$, $p = .001$, partial $\eta^2 = .69$ (see Figure 1A). Lower sway with eyes open reflected the benefit of early correction of sway afforded by vision (Paulus, Straube, & Brandt, 1984). In both cases, but more clearly in the eyes closed condition, light touch contact reduced sway, $F(2, 18) = 58.26, p < .001$, partial $\eta^2 = .87$. Thus, even in the presence of vision, sway benefited from interpersonal light touch. Overall, light touch reduced sway by 31.0% ($SD = 5.5\%$) and the reduction was 13.4% ($SD = 5.1\%$) with interpersonal light touch.

Previous research has shown light touch with an external referent improves postural stability in situations in which balance is dynamically perturbed (Johannsen, Wing, & Hatzitaki, 2007) or during treadmill walking (Dickstein & Laufer, 2004). Therefore, we would expect that the effect of interpersonal light touch on sway during quiet standing would also generalize to walking. This could have implications in assisting those with poor balance. For example, caregivers for the elderly tend to use light touch at the elbow in order to offer mechanical support in case of a fall. Our results suggest that light touch by a caregiver might offer not only improved assistive reactions by the caregiver if a fall should occur, but also lowered probability of falling because of reduced sway. Further research into this potentially important application is warranted.

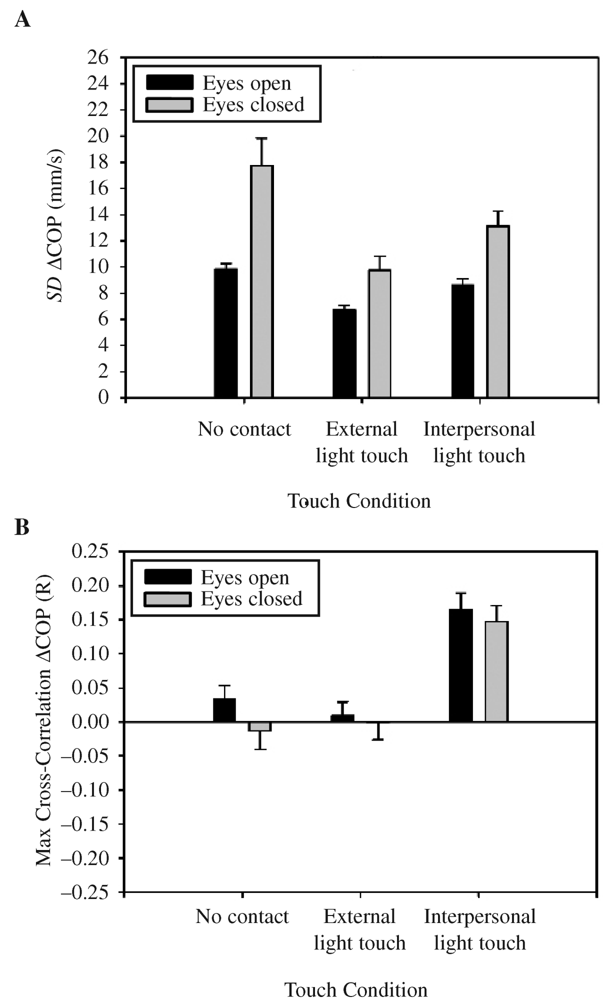

FIGURE 1. The variability of center-of-pressure (COP) rate of change (**A**) averaged over both directions of sway is shown as a function of the touch and vision conditions. The raw cross-correlation coefficients (**B**) expressing the interindividual temporal coordination of the COP rate of change averaged over both directions of sway shown as a function of the touch and vision conditions. Error bars indicate standard errors.

Force feedback from the fingertips during interpersonal light touch is likely to be important as a cue that contributes to reduction of postural sway. We propose that the reduction of postural sway occurs either as a direct consequence of the fingertip feedback (Jeka & Lackner, 1995) or as an indirect consequence of the constraints of the suprapostural task of keeping the fingertip in constant light contact (Riley, Stoffregen, Grocki, & Turvey, 1999). In the second case, it may be that feedback from the fingertip would be used to drive movements of the upper limb directly. In this case, feedforward postural adjustments related to such upper limb movements (Bouisset & Zattara, 1987) may be the basis for the link between light touch and balance.

The sway reduction during interpersonal light touch was smaller than in light touch with a fixed referent, although the shear forces might be expected to be of similar strength. This may reflect the greater variability of tactile input because of the other person's sway compared with the steadier tactile signal during fixed referent light touch. We observed a reliable difference between the touch conditions in terms of interpersonal coordination of sway, $F(2, 8) = 18.99$, $p = .001$, partial $\eta^2 = .83$ (see Figure 1B). Between-individual cross-correlations were significantly greater for interpersonal light touch ($M\ r = .22$, $SD = .05$) compared with both no contact and light touch (both $M\ r \leq .03$, both $SD \leq .05$; both $t \geq 6.02$; both $p \leq .004$).

Why do postural adjustments of the two individuals show a reliable correlation during interpersonal light touch? It is possible that one individual starts to follow the other's lead.

However, the overall phase lag was not reliably different from zero ($M = 50.0$ ms, $SD = 89.6$ ms). Nonetheless, the variability of the phase lag was high. This raises the question of whether the zero phase lag might have been an artifact of averaging large (e.g., 300 ms is typical of the lag between fingertip shear force and postural adjustments; Jeka & Lackner, 1994) positive and negative phase lags? However, inspection of the individual data points failed to reveal bimodality with consistent large phase lags varying in sign. Zero phase lag could be explained if both individuals independently adjust their posture using the common fingertip shear force signal, assuming equal postural reflex delays. However, given that the finger shear forces experienced by the two participants are in opposite directions, this would predict negative not positive correlated sway. Thus, as a working hypothesis, we suggest that the zero lag positive correlation arises from a strategy of coordinating sway (in predictive fashion) to reduce the shear force experienced at the fingertip. In ongoing research, we are exploring this idea and the possibility that the correlation may develop with experience of interpersonal light touch.

In conclusion, we suggest that interpersonal light touch represents an interesting experimental paradigm to investigate implicit and explicit factors of interpersonal joint action with relevance to situations in everyday life, including clinical and therapeutic contexts.

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NOTE

1. Informal testing with a miniature force-torque transducer (ATI Nano17, Apex, North Carolina, USA) attached to the index finger of 1 participant of a pair showed that fingertip contact forces during interpersonal light touch averaged less than 1 N.

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Contrasting effects of finger and shoulder interpersonal light touch on standing balance

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Johannsen L, Wing AM, Hatzitaki V. Contrasting effects of finger and shoulder interpersonal light touch on standing balance. *J Neurophysiol* 107: 216–225, 2012. First published September 28, 2011; doi:10.1152/jn.00149.2011.—Sway is reduced by light nonsupporting touch between parts of the body and a fixed surface. This effect is assumed to reflect augmentation of sensory cues for sway by point-of-contact reaction forces. It has been shown that movement of the contact surface can increase sway relative to an earth-fixed contact. Light touch contact with another person, for example, holding hands, affords a moving contact due to partner sway. We asked whether interpersonal light touch (IPLT) would increase sway relative to standing alone. We expected effects on sway to vary as a function of the site of contact and the postural stability of each partner. Eight pairs of participants, standing in either normal bipedal or tandem Romberg stance with eyes closed and using IPLT (finger to finger or shoulder to shoulder) or no contact, provided 4 trials of 30-s duration in each of 12 posture-touch combinations. Sway (SD of the rate of change of upper trunk position at C7) was reliably less with IPLT compared with no contact, with two exceptions: in normal stance, shoulder contact with a partner in tandem stance, and in tandem Romberg stance, finger contact with a partner in the same stance, increased sway. Otherwise, the reduction in sway was greater with shoulder than with finger contact. Measures of interpersonal synchronization based on cross-correlations and coherence analysis between the partners' C7 movements suggest different control factors operate to reduce sway in IPLT with the hand or shoulder contact.

interpersonal postural coordination; body sway

FINGERTIP LIGHT CONTACT with a static environmental reference point produces significant reduction in the variability of postural sway despite the provision of only minimal mechanical support (Holden et al. 1994). This has been demonstrated in young and older adults (Jeka and Lackner 1994; Tremblay et al. 2004) as well as in patient groups with sensory impaired balance (Dickstein et al. 2001; Jeka et al. 1996; Lackner et al. 1999). Shear forces from the tactile contact, in combination with information about contact location, derived from the distal-to-proximal proprioceptive chain, are thought to provide cues to body sway (Rabin et al. 1999). In general, forces at the fingertip include both normal and tangential components. Depending on finger orientation, these may be identified with anterior-posterior (AP) or left-right (LR) sway and afford cues to which the central nervous system (CNS) responds with adjustments to reduce the sway in each direction. Evidence that this is the basis for the light touch attenuation of body sway during upright standing includes small, but reliable, correlations between body sway and contact forces and torques. Cross-correlation time lags typically indicate a 250- to 300-ms lead of the tactile signal over subsequent postural adjustments (Clapp and Wing 1999; Jeka and Lackner 1994; Rabin et al. 2006).

The use of tactile feedback from light touch to reduce sway requires that the tactile sensory signal reflects own movement rather than that of the contact surface, and this may not necessarily be the case. For example, Jeka et al. (1997) showed that light finger contact with an oscillating reference increases postural sway, relative to the static contact, in synchrony with the movement of the reference. At higher frequencies of reference surface oscillation (>0.4 Hz), coherence with body sway decreases and phase lag increases (Jeka et al. 1997, 1998), suggesting that the feedback process, driven by both the velocity and position of the contact point, is limited by participants' sensory motor lag introducing a low-pass filter effect. Moreover, removing the shear forces at the fingertip, by linking the movement of the contact point to body sway (sway-referenced light touch), increases sway, again indicating the importance of this form of tactile feedback in reducing body sway (Reginella et al. 1999).

A common form of light touch contact arises in a social context, when partners hold hands. We recently demonstrated that light fingertip-to-fingertip contact reduces postural sway in pairs of older adults (Johannsen et al. 2009). However, the reductions were less (13%) than those observed when light contact was kept with a static external reference (31%). The smaller reduction in sway during such interpersonal light touch (IPLT) may reflect the ambiguity of the tactile feedback signal regarding own body sway due to the partner's sway. We were therefore interested in testing the effect of differing degrees of partner sway on own body sway during IPLT. In the present study we sought to assess the effect of each individual's stance (normal bipedal vs. tandem Romberg), the contact site (distal finger vs. proximal shoulder), and the similarity of joint posture in young adults. Because both individuals were always standing side by side, we expected that shoulder-to-shoulder contact would make precision control of body sway more critical in the LR direction to reduce the risk of mutual destabilization due to the lower number of postural degrees of freedom available compared with standing with finger contact. Thus we assumed greater sway reduction with shoulder than finger contact, particularly in the LR direction. We also expected that inter-

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personal light touch would become more effective with an increasing own contribution to the sensed contact force signal, and we therefore predicted that the reduction in sway would be less when the partner was in tandem compared with bipedal stance.

METHODS

Participants

Sixteen healthy adult participants were tested in eight pairs. In six pairs (mean 32.7 yr, SD 11.9 yr; 6 females, 6 males; 2 same-sex pairs, 4 mixed-sex pairs), light touch involved skin-to-skin contact between the two individuals. In the remaining two pairs (mean 45.6 yr, SD 11.2 yr; 2 mixed-sex pairs), additional trials were run in which the contact forces and torques between the participants were recorded using a miniature load sensor. All participants were recruited as an opportunity sample from the students and staff of the local research institute. Although pairings were allocated at random determined by participants' availability for testing, individuals in a pair were likely to know each other, but none of the pairs constituted a couple in an established relationship. Written informed consent was obtained from all participants, and the study was approved by the University of Birmingham Ethics Committee.

Apparatus

The data acquisition setup consisted of a 12-camera optoelectronic motion capture system (Qualisys Oqus, Gothenburg, Sweden), 2 separate force platforms (model 4060H; Bertec, Worthington, OH), and a miniature load sensor (F/T sensor Nano17, 6 degrees of freedom; ATI Industrial Automation, Apex, NC). Body movements at C7 were sampled at 200 Hz, whereas data from the force platforms and the miniature force transducer were sampled at 1,200 Hz. Each platform measured the six components of the ground reaction forces and moments to determine the AP and LR components of center of pressure. The force platforms and load sensor were connected through a single multiplexed analog-to-digital converter (ADC) to the motion capture system, which synchronized the beginning and end of data acquisition for each experimental trial. In those two pairs of participants where the contact forces and torques between the participants were recorded, the sensor was placed between the skin contact surfaces by attachment to the person on the left with double-sided adhesive tape. Two lightweight 15-mm tubular plastic rods with reflective markers were attached to the load sensor and allowed position and orientation tracking of the sensor in three-dimensional space for mapping the sensor's local force readings onto the global (force plate and kinematic) reference frame. The total weight of the force sensor assembly was 18 g.

Procedure

Participants stood side by side, oriented in the same direction, with eyes closed and heads facing forward. The side on which the taller person was standing was randomized. Participants were instructed to stand as still as possible in a relaxed manner without speaking.

Body sway was recorded in 12 different interpersonal joint posture conditions made up of 3 experimental factors: *1*) individual stance posture, *2*) similarity of the interpersonal joint postures, and *3*) form of IPLT contact. Individuals were tested in two stance postures. In normal bipedal stance, participants kept a narrow-base standing posture with an ~5-cm interheel gap. In tandem Romberg stance, participants placed the nondominant foot in front of the dominant, keeping a heel-to-toe gap of ~5 cm. Similarity of the interpersonal joint posture was varied by fully permuting the two stance postures between both individuals. Thus, four interpersonal joint postures were performed by a pair. For the statistical analysis of interpersonal coordination, the number of interpersonal joint postures was reduced to three by averaging the two different interpersonal joint postures for each pair.

Figure 1 illustrates the interpersonal light touch and stance conditions. In IPLT conditions, contact between individuals was established either between the index fingertips of one hand or between the arms at a contact point near the shoulder. In "finger contact," the person on the left force platform extended the right arm at the elbow, while the person on the right extended the left arm. The elbow of the extended arm was kept in contact with the torso at waist level while the other arm was brought across the stomach so that the other hand made contact with the crook of the extended arm. The person on the left kept their right hand in pronation to touch the finger of the person on the right from above, while the latter kept his or her hand in supination. During "shoulder contact," arm postures were similar except that participants kept their hands apart and moved slightly closer together so that contact was established with the outer surface of the arm at the shoulder. Practice trials were performed to ensure that participants were experienced in keeping touch as light as possible without losing contact. During the practice trials, the experimenter served as partner with each participant to provide verbal feedback about the appropriate IPLT force level and to demonstrate the required standing postures. Quantitative feedback about touch force was not given to participants. Finally, as a control condition, participants' performance was also tested during "no contact," when body posture was exactly the same as in the other two conditions except without physical contact between the participants.

Four trials (30 s each) were recorded for each of the 12 interpersonal joint postural conditions, resulting in a total of 48 trials for each pair of participants. The three interpersonal contact conditions were tested in blocks of 16 trials. The sequence of these blocks was ordered randomly. Within each block, four miniblocks of four trials each occurred in random order for each interpersonal joint posture.

Data Reduction and Statistical Analysis

All time series data were low-pass filtered at 10 Hz and differentiated to yield rate-of-change measures of sway (dC7, dCoP). Data analysis focused on sway, SD dC7 (with SD dCoP, which yielded similar findings; see Supplemental Material). (Supplemental data for this article is available online at the *Journal of Neurophysiology* website.)

The proportional change in SD dC7 sway during each of the eight IPLT conditions was calculated relative to the corresponding normal bipedal and tandem Romberg baselines without IPLT contact. On each axis, within-trial estimates of proportional sway change were averaged for every participant across trials for each experimental condition and subjected to repeated-measures ANOVA with stance posture, IPLT condition, and interpersonal joint postural similarity as within-subject factors. Significance levels were set at $P = 0.05$ after Greenhouse-Geisser correction.

Cross-correlation functions were calculated for two specific purposes. First, we aimed to analyze postural coordination between paired individuals for dC7 fluctuations in the AP and LR directions. Second, we used the data from the two pairs of participants, where a miniature load sensor recorded the components of contact forces, for cross-correlating each individual's dC7 fluctuations with the force fluctuations in both horizontal directions. Cross-correlation functions were computed for lags ranging from +1,000 ms (shorter person leads; contact force leads dC7) to −1,000 ms (taller person leads; dC7 leads contact force). The largest absolute cross-correlation and corresponding time lag were extracted and were averaged for each experimental condition. All cross-correlation coefficients were Fisher *Z*-transformed (Fisher 1915) before statistical analysis was carried out. The between-individual cross-correlation coefficients were subjected to repeated-measures ANOVA with touch and interpersonal joint

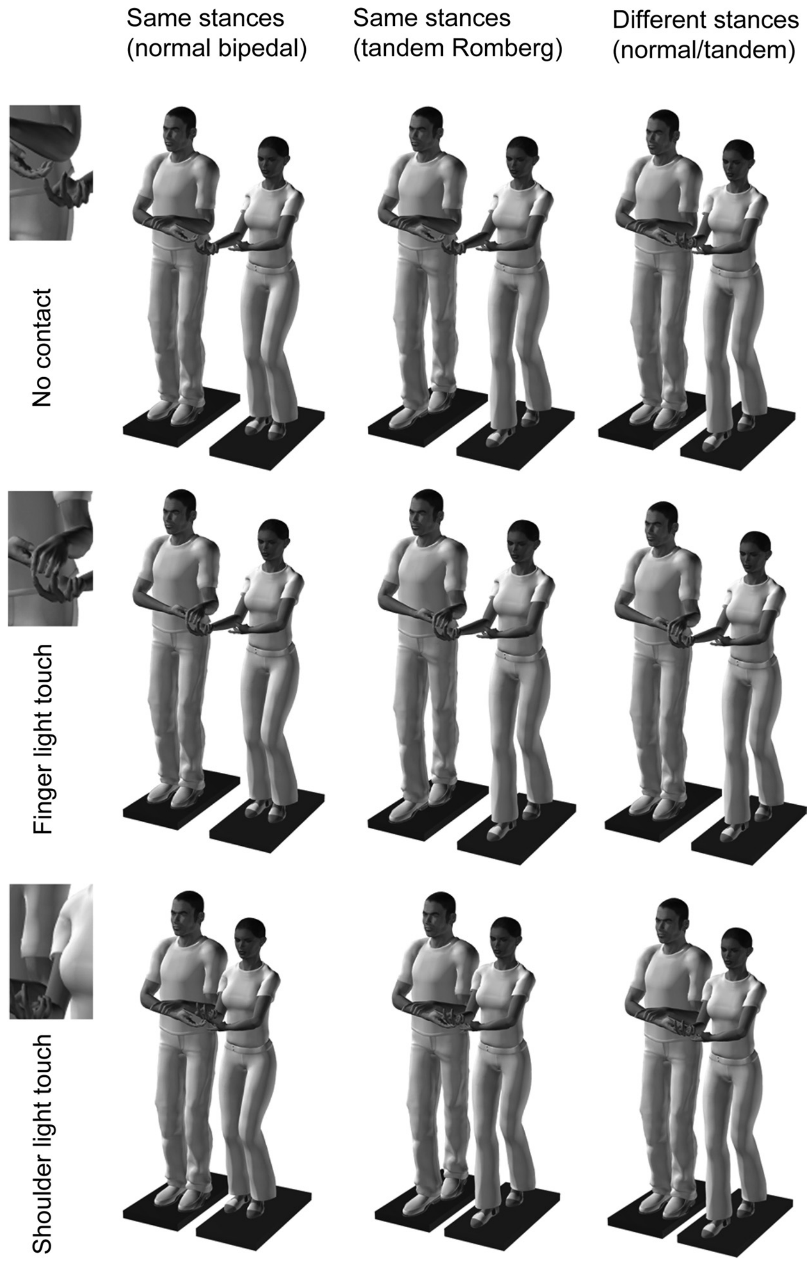

Fig. 1. Illustrations showing 9 of the 12 interpersonal joint postures tested in the study. The remaining 3 stance postures not shown were the reversed different stances configuration.

posture conditions as within-subject factors, whereas the sway contact force cross-correlation coefficients were tested with stance posture, IPLT condition, and similarity of interpersonal joint posture as within-subject factors.

To extend the time domain analysis of interpersonal synchronization during IPLT, we estimated the magnitude squared coherence and the cross-power spectral density using Welch's method (1967) on the C7-position time series. Our primary aim was to gain information on the sway frequency range at which the relation between the two individuals in a pair was strongest. For this the frequency spectrum was segmented into bins of 0.0244-Hz step size. Frequency bins from 0 to 0.1 Hz were excluded from the subsequent frequency peak extraction algorithm to avoid the inclusion of very slow drift effects commonly observed in quiet normal bipedal standing. Therefore, the frequency range considered extended from 0.1 to 10 Hz. The frequency bin with the peak magnitude coherence was found, and the corresponding relative phase angle was extracted from the cross-power spectral density distribution for every single trial. The peak magnitude coherence estimates were Fisher Z-transformed and subjected to repeated-measures ANOVA for each direction of sway with touch and interpersonal joint posture conditions as within-subject factors. The peak coherence frequency bins and corresponding relative phase estimates were analyzed similarly apart from the Fisher Z transformation. All data processing and analysis were performed in Matlab 7.5 (The MathWorks, Natick, MA) and SPSS 16 (IBM, Somers, NY). In this article, we only report those main effects and interactions that were at least marginally significant. Nonsignificant effects and interactions are not mentioned.

RESULTS

Contact Forces

The force transducer recordings in two pairs of participants indicated that the average normal force during finger contact was lower (both normal stance: mean 0.47 N, SD 0.21 N; both tandem stance: mean 0.55 N, SD 0.27 N; different stances:

mean 0.54 N, SD 0.25 N) compared with shoulder contact (both normal stance: mean 0.78 N, SD 0.18 N; both tandem stance: mean 0.89 N, SD 0.19 N; different stances: mean 1.01 N, SD 0.39 N). The same was true for peak normal force during finger contact (both normal stance: mean 1.07 N, SD 0.21 N; both tandem stance: mean 1.25 N, SD 0.27 N; different stances: mean 1.21 N, SD 0.31 N) and shoulder contact (both normal stance: mean 2.01 N, SD 0.57 N; both tandem stance: mean 3.17 N, SD 0.27 N; different stances: mean 2.69 N, SD 0.57 N). Finally, the standard deviation of contact force across a trial was much lower during finger contact (both normal stance: mean 0.19 N, SD 0.003 N; both tandem stance: mean 0.20 N, SD 0.01 N; different stances: mean 0.20 N, SD 0.01 N) than during shoulder contact, where an increase in the force variability over the stability of joint stance postures was apparent (both normal stance: mean 0.33 N, SD 0.14 N; both tandem stance: mean 0.59 N, SD 0.07 N; different stances: mean 0.48 N, SD 0.12 N). Supplemental Fig. S1 shows two examples of the contact force fluctuations recorded between two paired individuals in different stances during finger and shoulder contact, respectively.

Postural Sway

Figure 2 shows illustrative dC7 fluctuations in two participants in different interpersonal joint postures during a single trial for each touch condition. The overall average dC7 values for no contact, finger light touch, and shoulder light touch conditions were 9.5, 8.0, and 7.0 mm/s, respectively. AP sway was greater than LR sway in normal stance (6.0 vs. 4.5 mm/s) and less in tandem stance (8.0 vs. 12.0 mm/s). The effects of contact conditions and joint posture interacted differently according to sway direction. In the following we present the results in terms of change relative to the no contact condition separately for each direction. The statistical analysis of body sway variability in terms of SD dC7 included data from all 12 participants where the miniature force transducer was not used. The pairwise temporal coordination between two participants was therefore analyzed in six pairs.

Figure 3 shows proportional sway change relative to the no contact baseline averaged across all individual participants as a function of the touch condition and the similarity of interpersonal joint posture for each stance. Negative values indicate a

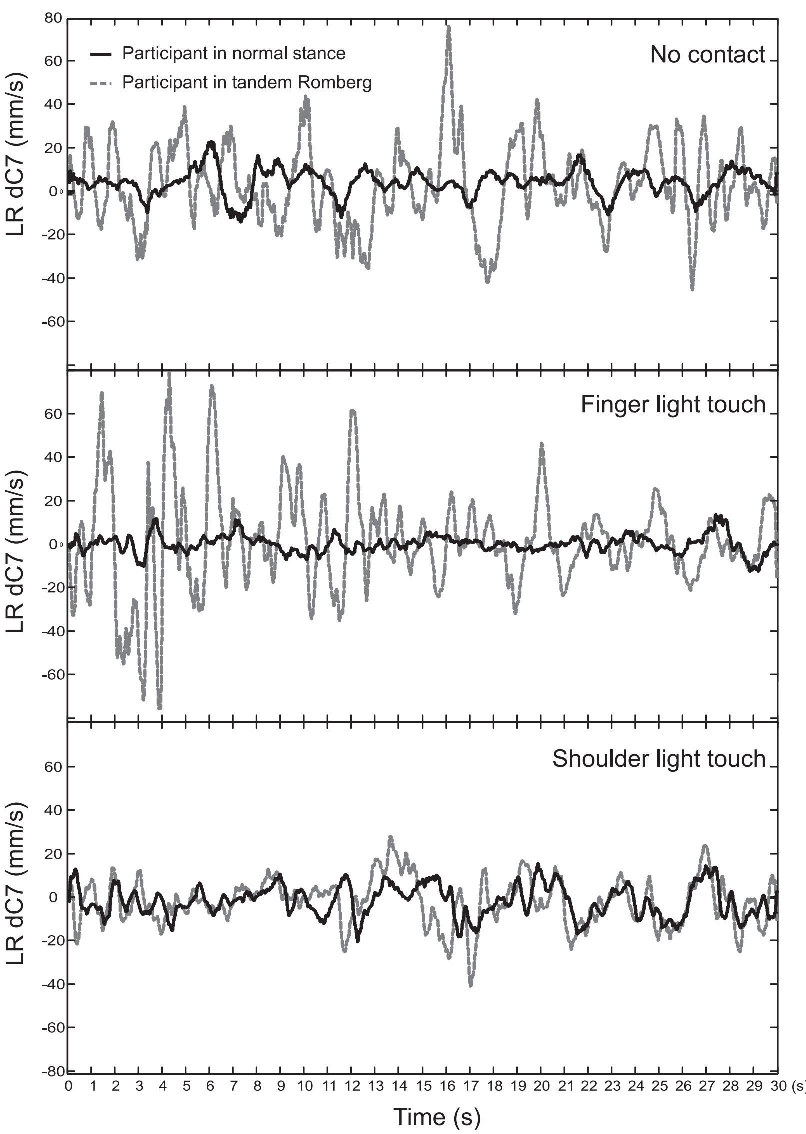

Fig. 2. For each interpersonal light touch condition, data from a single trial illustrating dC7 fluctuations on the left-right (LR) axis in 2 participants in different stances is shown.

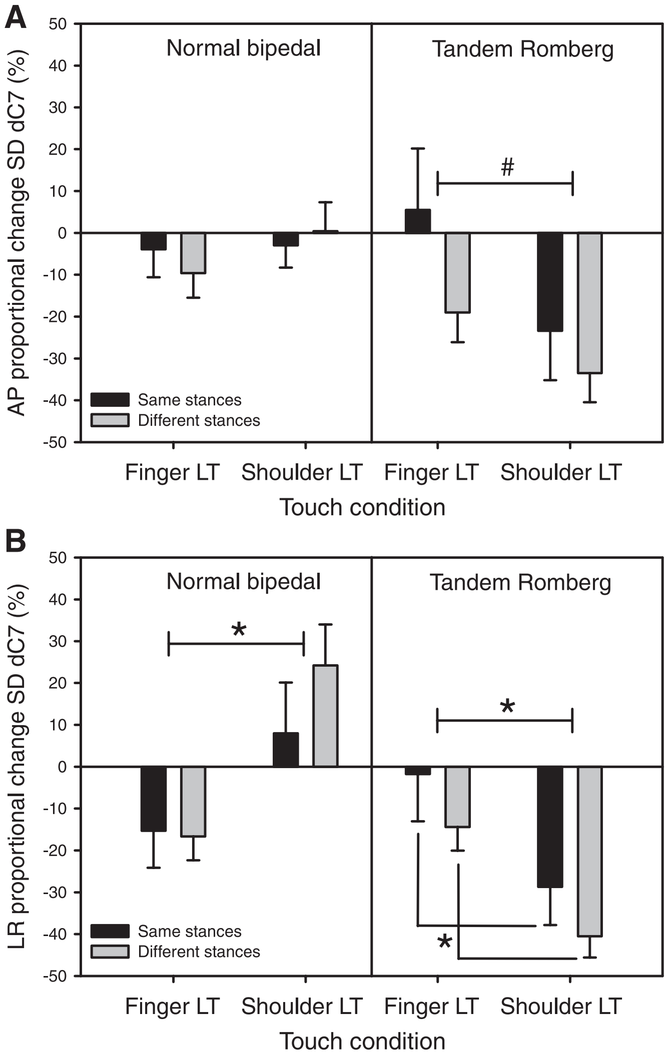

Fig. 3. Proportional change in the standard deviation (SD) dC7 in each interpersonal light touch condition relative to the no contact baseline as a function of stance posture and interpersonal postural similarity. Negative values indicate a reduction in sway, whereas positive values indicate an increase in sway. *A*: proportional sway change in the anterior-posterior (AP) direction. *B*: proportional sway change in the LR direction. Error bars indicate SE. $*P < 0.05$ indicates a significant single comparison. $\#P < 0.1$ indicates a marginally significant single comparison. LT, light touch.

reduction in sway, whereas positive values indicate an increase. In the AP direction, proportional sway change differed significantly between the two similarity conditions [$F(1,10) = 7.22$, $P = 0.02$, partial $\eta^2 = 0.42$]. When partners kept different stance postures, greater reductions in sway were observed for each of the two individuals in a pair than when both partners kept a similar stance. None of the remaining main effects or interactions reached statistical significance except for the interaction between stance posture and touch condition [$F(1,10) = 5.27$, $P = 0.05$, partial $\eta^2 = 0.35$]. Post hoc comparisons showed that in tandem Romberg stance, shoulder IPLT tended to cause greater proportional sway reductions than finger IPLT, while no difference between the IPLT conditions was apparent for normal bipedal stance. On the LR axis, a main effect of stance posture [$F(1,10) = 5.49$, $P = 0.04$, partial $\eta^2 = 0.35$] was observed. Tandem stance showed greater reductions than normal stance. Furthermore, the interaction between stance posture and touch condition was significant [$F(1,10) = 41.49$, $P < 0.001$, partial $\eta^2 = 0.81$]. Finally, there was a tendency for an interaction between stance posture and similarity of interpersonal joint posture [$F(1,10) = 3.64$, $P = 0.09$, partial $\eta^2 = 0.27$]. Post hoc comparisons showed that in tandem Romberg stance, shoulder IPLT induced greater proportional sway reductions than finger IPLT, which actually caused a slight increase when both partners were tandem Romberg. In normal stance, however, shoulder IPLT caused an increase in sway for individuals. Finally, a different interpersonal joint posture tended to be more beneficial in terms of proportional sway reductions for the person in tandem Romberg stance, whereas it tended to lead to greater proportional sway in normal stance.

Interpersonal Postural Coordination

Time domain. Across all three joint posture conditions,[1] absolute peak cross-correlation coefficients indicating spatio-temporal coordination of postural adjustments in terms of dC7 between two paired individuals for the no contact control condition averaged 0.19 (SD 0.03) on the AP axis and −0.20 (SD 0.04) on the LR axis. In the finger contact condition, the average absolute peak cross-correlation coefficient was 0.22 (SD 0.03) on the AP axis and 0.22 (SD 0.02) on the LR axis. Finally, in the shoulder contact condition, the average cross-correlation coefficient was 0.26 (SD 0.03) on the AP axis and 0.51 (SD 0.04) on the LR axis.

In the AP direction, the cross-correlation coefficients differed as a function of touch condition [$F(2,10) = 28.27$, $P < 0.001$, partial $\eta^2 = 0.85$] and similarity of interpersonal joint posture [$F(2,10) = 20.43$, $P = 0.005$, partial $\eta^2 = 0.80$]. A significant interaction was found between touch condition and similarity of interpersonal joint posture [$F(4,20) = 5.73$, $P = 0.01$, partial $\eta^2 = 0.53$]. Post hoc comparisons demonstrated that in both finger and shoulder IPLT, significantly higher interpersonal synchronization occurred compared with no contact. Also, coefficients were generally higher when both partners were in normal bipedal stance. When both participants were standing in normal bipedal stance with both finger and shoulder contact, cross-correlation coefficients exceeded the highest absolute boundary of the 95% confidence interval of any no contact condition. In the LR direction, the cross-correlation coefficients differed as a function of touch condition [$F(2,10) = 112.58$, $P < 0.001$, partial $\eta^2 = 0.96$]. An interaction between touch condition and similarity of interpersonal joint posture was also present [$F(4,20) = 5.93$, $P = 0.02$, partial $\eta^2 = 0.54$]. Post hoc comparisons showed again that in shoulder contact, cross-correlation coefficients were greater compared with no contact. Furthermore, during shoulder contact coefficients were higher when both partners kept a different stance compared with both partners in tandem Romberg stance. All shoulder contact conditions exceeded the highest absolute boundary of the 95% confidence interval of any no contact condition. This means that only shoulder contact resulted in increased synchronization in the LR direction irrespective of the specific condition of interpersonal joint postural similarity. Cross-correlation coefficients as a function of touch

[1]In every single trial (100%), the Fisher *Z*-transformed absolute peak cross-correlation coefficient exceeded the confidence interval threshold for a zero correlation of 0.03 [= $1.96(1/5{,}999^{0.5})$].

condition, similarity of interpersonal joint posture, and sway direction are shown in Fig. 4.

In the AP direction, the corresponding time lags of the peak cross-correlations that surpassed the critical threshold were 26 ms (SD 305; taller participant leads) for finger contact with both partners in the same stance, 283 ms (SD 215; taller participant leads) for both partners in bipedal shoulder contact, and 119 ms (SD 237; taller participant leads) for shoulder contact with both partners in different stances. Single comparisons against zero showed that the lead of the taller participant in normal bipedal shoulder contact was significant [$t(5) = -3.22$, $P = 0.02$]. In the LR direction, time lags for peak cross-correlations surpassing the critical threshold during shoulder contact were 33 ms (SD 351; taller participant leads) for both partners in normal bipedal stance, 7 ms (SD 71; shorter participant leads) for both partners in different stances, and 66 ms (SD 219; taller participant leads) for both partners in tandem Romberg stance. No main effects or interactions as a function of touch condition or similarity of interpersonal joint posture were found for any of the two sway directions. None of these LR time lags were significantly different from zero lag.

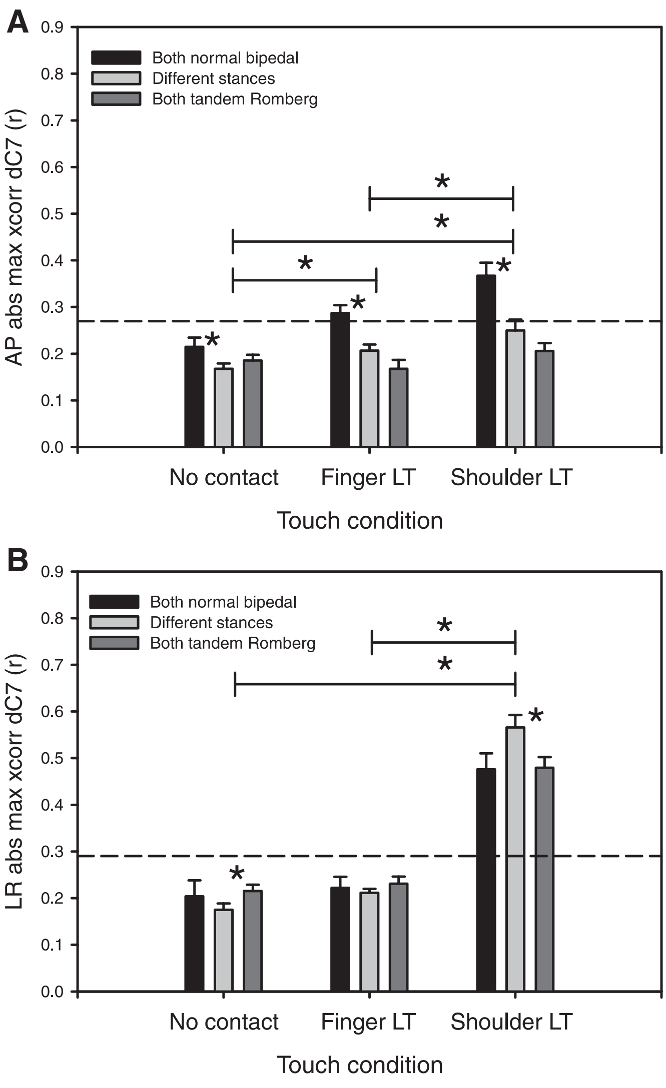

Fig. 4. Peak absolute cross-correlation coefficients between 2 individuals of a pair for dC7 as a function of interpersonal light touch condition and interpersonal postural similarity. *A*: cross-correlation coefficients for dC7 in the AP direction. *B*: cross-correlation coefficients for dC7 in the LR direction. Error bars indicate SE. *$P < 0.05$ indicates a significant single comparison. The dashed lines indicate the upper and lower boundary derived from the absolute maximum 95% confidence interval for any of the 3 no contact conditions. Abs max xcross, absolute maximum cross-correlation.

Frequency domain. In the AP direction, magnitude squared coherence differed with touch condition [$F(2,10) = 7.80$, $P = 0.01$, partial $\eta^2 = 0.61$] and similarity of interpersonal joint posture [$F(2,10) = 4.21$, $P = 0.05$, partial $\eta^2 = 0.46$]. Post hoc comparisons showed that coherence was significantly higher during shoulder contact (mean 0.32, SD 0.02) compared with no contact [mean 0.27, SD 0.03; $F(1,5) = 17.97$, $P = 0.008$, partial $\eta^2 = 0.78$] and tended to be higher compared with finger contact [mean 0.29, SD 0.04; $F(1,5) = 4.49$, $P = 0.09$, partial $\eta^2 = 0.47$]. Also, both partners in normal bipedal stance resulted in higher average coherence than both partners in tandem stance. Similar to peak cross-correlations, shoulder contact with both partners in normal bipedal stance exceeded the highest absolute boundary of the 95% confidence interval of any no contact condition. In the LR direction, peak coherence differed only as a function of touch condition [$F(2,10) = 25.31$, $P = 0.002$, partial $\eta^2 = 0.84$]. Post hoc comparisons indicated that shoulder contact (mean 0.50, SD 0.09) resulted in higher coherence than both finger contact and no contact [finger contact: mean 0.29, SD 0.03; no contact: mean 0.27, SD 0.02; both $F(1,5) > 21.78$, $P < 0.002$, both partial $\eta^2 > 0.81$]. During shoulder contact, all three interpersonal joint postures exceeded the highest absolute boundary of the 95% confidence interval of any no contact condition.

Interpersonal relative phase did not differ between experimental conditions in either the AP (mean 1.05 deg, SD 10.87 deg) or the LR direction (mean 2.23 deg, SD 24.51 deg). On the other hand, the variability of interpersonal relative phase was not the same across the touch conditions in both directions of sway [AP: $F(2,8) = 5.80$, $P = 0.05$, partial $\eta^2 = 0.59$; LR: $F(2,8) = 28.23$, $P = 0.005$, partial $\eta^2 = 0.88$]. Post hoc comparisons showed that the variability of interpersonal relative phase during shoulder contact (AP: mean 66.00 deg, SD 18.88 deg; LR: mean 31.85 deg, SD 21.99 deg) was significantly lower than during finger contact (AP: mean 97.52 deg, SD 12.04 deg; LR: mean 85.13 deg, SD 13.34 deg) or no contact (AP: mean 90.39 deg, SD 16.72 deg; LR: mean 105.04 deg, SD 33.95 deg).

The average peak coherence frequency as a function of touch condition and sway direction is shown in Fig. 5. Overall peak coherence frequency was at 2.94 Hz (SD 0.40 Hz) in the AP direction. No main effects or interaction between touch conditions and similarity of interpersonal joint posture were found. In contrast, in the LR direction, peak coherence frequency was different between the touch conditions [$F(2,10) = 8.74$, $P = 0.02$, partial $\eta^2 = 0.64$]. Post hoc comparisons indicated a significant shift in the peak coherence frequency in shoulder contact to a lower frequency band compared with finger contact and no contact.

Relationship Between Contact Force and Sway

For our analysis of the relationship between fluctuations of dC7 and the horizontal contact force components, we utilized the absolute peak cross-correlation coefficient and its corre-

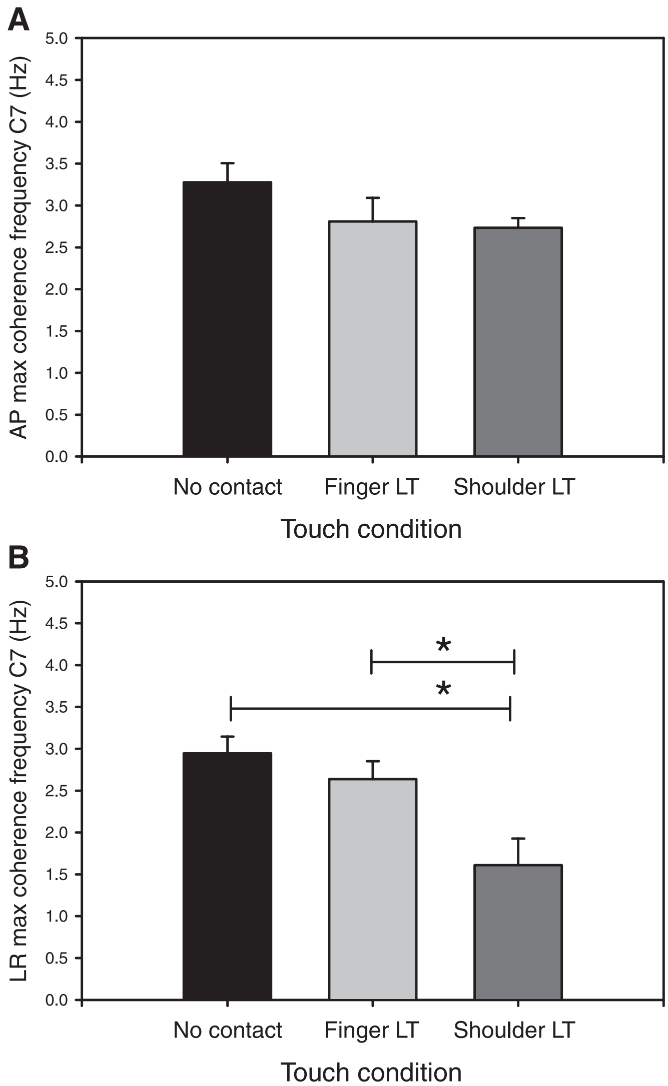

Fig. 5. Frequency of peak magnitude squared coherence between 2 individuals' C7 fluctuations as a function of interpersonal light touch condition and sway direction. *A*: frequency of peak coherence in the AP direction. *B*: frequency of peak coherence in the LR direction. Error bars indicate SE. $^*P < 0.05$ indicates a significant single comparison.

sponding time lag (Fig. 6).[2] On the AP axis, only the effect of touch was significant [$F(1,3) = 44.89$, $P = 0.007$, partial $\eta^2 = 0.94$] due to shoulder contact leading to stronger cross-correlations than finger contact. In the LR direction, coefficients showed main effects of touch condition [$F(1,3) = 16.37$, $P = 0.03$ partial $\eta^2 = 0.85$] and stance posture [$F(1,3) = 24.31$, $P = 0.02$, partial $\eta^2 = 0.89$] as well as an interaction between touch condition, stance posture, and similarity of interpersonal joint posture [$F(1,3) = 27.50$, $P = 0.01$, partial $\eta^2 = 0.90$]. Tandem Romberg stance resulted in higher coefficients than normal bipedal stance, and shoulder contact resulted in higher coefficients than finger contact. Post hoc comparisons explained the three-way interaction with an increase in the cross-correlation coefficients when in normal bipedal stance with shoulder contact to a partner in tandem Romberg compared with contact with a partner in the same normal bipedal stance.

No main effects or interactions were found for the cross-correlation time lags on either axis. Overall averages expressed a lead of the force signal by 71 ms (SD 171 ms) on the AP axis and a lead by the person by 343 ms (SD 304 ms) on the LR axis. Both measures, however, were not significantly different from zero lag.

DISCUSSION

In quiet standing, the variability of sway is reduced by light touch contact with a static environmental referent (Holdenet al. 1994; Jeka and Lackner 1994). The reduction in sway is generally assumed to reflect the use of tactile feedback from the contact. If the contact surface moves, sway increases, as if no allowance were made for its movement (Jeka et al. 1997; Wing et al. 2011). Light touch contact with a moving surface

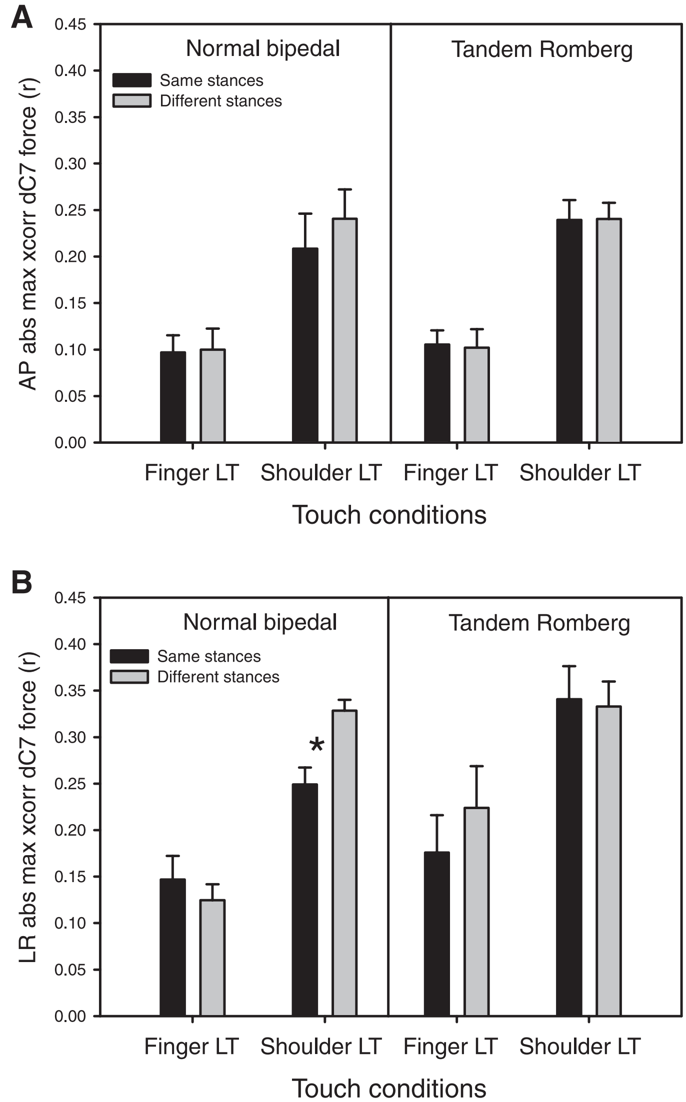

Fig. 6. Peak absolute cross-correlation coefficients between each individual's dC7 and horizontal contact force component fluctuations in each interpersonal light touch condition as a function of stance posture and interpersonal postural similarity. *A*: cross-correlation coefficients for the AP direction. *B*: cross-correlation coefficients for the LR direction. Error bars indicate SE. $^*P < 0.05$ indicates a significant single comparison.

[2]In every single trial (100%), the Fisher *Z*-transformed absolute peak cross-correlation coefficient exceeded the confidence interval threshold for a zero correlation of 0.01 [= 1.96(1/35,999$^{0.5}$)].

commonly occurs in a social context in hand holding. In the present study we examined how IPLT effects on sway may depend on contact location and on the stance of young adults. We predicted alterations in the extent to which an individual would utilize the force signal for the fine-tuning of postural adjustments, and this would depend on the variability of the signal as well as the demands for precise direction-specific modulation of sway. Thus we expected that sway would be reduced less if the partner stood in tandem stance, compared with a partner in bipedal stance, due to the generally increased variability of the touch signal from the partner. This should apply especially to situations in which both individuals adopted differing stance postures (asymmetrical interpersonal stance posture: one person in normal bipedal, one in tandem Romberg). Our expectation for this situation was that the more stable individual would contribute less to the variability of the touch signal and therefore would receive less specific feedback about own body sway and, as a consequence, would show smaller sway reductions. Finally, given the requirement of precision control of body sway, we assumed that shoulder-to-shoulder contact would force participants to constrain their sway actively to compensate for the lower number of postural degrees of freedom of this more proximal contact.

We found that IPLT reduced sway of an individual in both normal bipedal and tandem Romberg stance. The proportional reduction of sway during finger contact (9–15%) was comparable to the figure we obtained for older participants using three-finger IPLT (Johannsen et al. 2009). However, in the present study, differences between finger and shoulder contact were apparent depending on stance posture. In normal bipedal stance, sway was reduced mainly during finger contact for both directions of sway. In the LR direction, shoulder-to-shoulder contact even increased sway. Interestingly, these particular conditions were the ones in which an individual in tandem Romberg showed the greatest reductions in sway during IPLT. A slight increase in sway, however, was observed in the AP direction when both partners were standing in tandem Romberg and were keeping finger contact.

The above alterations in sway may be invoked by a force feedback control loop where the sway-related contribution to the afferent touch signal facilitates more efficient subsequent postural adjustments through improved perception of own sway (Jeka and Lackner 1994). For example, during light touch with an inanimate, earth-fixed reference point, the force signal variability can be attributed exclusively to own sway. During IPLT with a partner in the same stance posture, each partner may contribute equally to the signal, whereas in an asymmetrical joint posture, the individual in the less stable stance might cause the greater amount of variability. According to this notion, a higher contact point, such as during shoulder contact, would result in greater proportional sway reductions due to an increase in sway-related shear forces (Krishnamoorthy et al. 2002; Rogers et al. 2001), yielding a more direct indication of body sway. Further support for this hypothesis comes from stronger cross-correlations between contact force components and sway, especially in the LR direction, irrespective of finger or shoulder contact in tandem Romberg stance. Overall, in tandem Romberg stance, sway reduced progressively with light touch contact with another individual, with a more proximal contact point at the shoulder and with a larger own contribution to the force signal relative to a more stable partner. In contrast, in finger contact, the additional shoulder and elbow degrees of freedom decouple the linkage with body sway.

We believe, however, that the force feedback hypothesis only partly explains the mechanisms at work during IPLT. In addition to the force feedback, with shoulder contact individuals may have chosen a control strategy that involves a greater degree of constraint on active sway to keep touch light (Riley et al. 1999). The requirements for such a strategy would be less during finger IPLT, because intrinsic touch precision would presumably be greater due to additional limb movement degrees of freedom (Rabinet al. 2008). Similar to the hypothesis of constraints on active sway, feed-forward reductions in sway have been suggested to result from a “suprapostural” task goal such as keeping the contact force below a specified threshold, for example, below 1 N (Riley et al. 1999). In this regard, one might speculate whether minimization of the contact force could have become an implicit suprapostural task goal that affords proactive sway control. Besides keeping the contact force light, however, partners may also have tried to minimize the variation of the contact point in spatial coordinates to facilitate interpersonal coordination. In situations where postural control served precise performance in a secondary task such as visual gaze fixation or manual aiming, direction-specific minimization of sway at the cost of increased sway in the perpendicular direction has been demonstrated (Balasubramaniam et al. 2000; Mitra 2004; Stoffregen et al. 1999). On the other hand, we did not find any indications for a similar effect during IPLT; that is, reduced sway on the LR axis at the cost of sway increases in the AP direction. This observation may indicate that during IPLT, the precision demands are qualitatively different from gaze fixation or manual aiming. For example, the specific contact point location might not have been represented in an allocentric spatial reference frame. It appears more likely that the contact point was located in an egocentric frame of reference given that the notion of a change from an allocentric to an egocentric trunk-based reference frame during individual light touch has been proposed in a recent publication by Franzen et al. (2011).

Fingertip light touch increases interpersonal synchronization of postural adjustments during rhythmic externally paced voluntary sway at a low (0.24 Hz) frequency (Sofianidis et al. in press) as well as during quiet standing (Johannsen et al. 2009). In quiet standing, increased positive cross-correlations indicated sway in the same direction with near-zero phase lag in contrast to two control conditions without IPLT where cross-correlations were not significant. One possibility to interpret this zero phase lag would be mutual light touch entrainment, similar to keeping contact with an oscillating reference (Jeka et al. 1997). On the other hand, fingertip exposure to a periodic haptic driving stimulus that replicates a natural sway pattern increases postural sway compared with a control condition without contact (Wing et al. 2011). We conclude from that particular observation that IPLT is more than a process of mutual sway entrainment.

In the present study, we also found evidence for increased interpersonal postural coordination, although it was modulated by the sway direction, the interpersonal stance symmetry, and the site of IPLT. For AP sway, interpersonal coordination appeared to be present only in symmetrical normal bipedal stance for both finger and shoulder contact, whereas for LR sway, increased interpersonal coordination was present during shoulder contact only irrespective of the stance and the interpersonal stance symmetry. The more proximal contact site at the shoulder gave rise to stronger interpersonal coordination, possibly due to a more direct transmission of the trunk's sway or a more fixed alignment with fewer postural degrees of freedom. A coherence analysis confirmed the findings in the time domain but provided additional insight as well. Although the frequency of the peak coherence was not different between touch conditions in the AP direction, with an average frequency bin at ~3 Hz, shoulder contact almost halved peak coherence frequency on the LR axis. In addition, the variability of relative phase was also lower during shoulder contact. This longer oscillatory period may indicate a qualitative change in the nature of interpersonal coordination during shoulder contact. Whereas a period of 300 ms during finger contact but also no contact may express commonalities between individuals with respect to "natural" sensory feedback-driven adjustments of sway, a period of 600 ms in duration may be caused by a postural strategy that facilitates tracking of each other's swaying movements. For example, task-specific combinations of postural control synergies including situations with light touch contact have been reported previously (Krishnamoorthy et al. 2004). During visually guided LR weight shifting, the control synergy is shifted from the hip to the ankle muscles to achieve the precision requirements of the task (Hatzitaki and Konstadakos 2007). An ankle strategy may therefore dominate shoulder-to-shoulder contact as suggested by the reduction in the primary peak coherence frequency.

This study confirmed the zero lags for the conditions with significant cross-correlations between individuals except for a lead of the taller individual with respect to AP sway with shoulder contact and both partners in normal bipedal stance. This observation, however, contrasts with the shorter interindividual lags on the LR axis in the same stance situation and may indicate a functional dissociation between both sway directions in terms of the interpersonal coordination. A mechanical support strategy between both individuals may be considered as a possible explanation for the observed increases in the contact force levels and the relatively short time lags during shoulder contact. This "tripod strategy" achieved by leaning and oscillating toward the partner in an anti-phase coordination pattern should result in negative interpersonal cross-correlations, which we did not observe. Although we recognize that short-term mechanical effects, i.e., mutual perturbations, may increase apparent interpersonal coordination, we do not believe that individuals in a pair chose a mechanical support strategy. Had they done so, force levels would be expected to be higher and within-trial force variability lower during shoulder than finger contact. In addition, we would expect much higher interpersonal cross-correlations than the ones actually found. In our opinion, a mechanical support strategy would have been most likely during shoulder contact, when both partners were in tandem Romberg stance. In contrast, we observed increasing force levels with increasing within-trial standard deviation of contact force, which indicates that keeping shoulder contact becomes more demanding with less stable joint stance postures. We suggest that both participants' sway may also have been influenced by the shared perception of the contact point's directional motion. A spatial tracking mechanism would enable prediction of the contact point position, and thus more precise control of the contact forces, because the difference in acceleration between both individuals would be minimized. In addition, participants may have extracted their partners' movements from the variability of the force signal and the position change that could not be accounted for by an internal source such as own body sway.

It is remarkable that on the LR axis, a dissociation between reduction of sway and interpersonal in-phase coordination seems present for finger IPLT (significant reductions in sway without interpersonal coordination) in contrast to shoulder IPLT, where both effects appear to be associated. We suggest, therefore, that in the case of finger IPLT, where an additional rotational degree of freedom is available on the LR axis, feedback from the fingertip might be used to drive movements of the upper limb for minimizing contact force directly. If the contact force and thus the touch feedback is minimized, however, it appears contradictory that the remaining signal would still be sufficient to inform about own body sway and result in improved postural adjustments. In this case, feed-forward postural adjustments related to such upper limb movements (Bouisset and Zattara 1987) might be the basis for the link between IPLT and reduction in sway. Thus employing the upper limb's extra movement degree of freedom in finger IPLT may reduce the coupling between touch feedback and subsequent postural adjustments and also the coordination between both partners.

In conclusion, the present study investigated in young adults the effect of light touch contact between two individuals on each individual's control of body sway as a function of the skin contact site, each individual's stance posture, and the interpersonal stance symmetry. Reliable reductions in sway were found during both forms of IPLT. Distal IPLT at the fingertip, however, differed from more proximal IPLT at the shoulder with respect to the proportional reduction in sway depending on an individual's and the partner's stance posture. In-phase interpersonal postural coordination with near-zero lag became evident during shoulder-to-shoulder contact on the LR axis. We propose two different mechanisms for maintaining IPLT during shoulder and finger contact.

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GRANTS

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

AUTHOR CONTRIBUTIONS

L.J., A.M.W., and V.H. conception and design of research; L.J. performed experiments; L.J. analyzed data; L.J., A.M.W., and V.H. interpreted results of experiments; L.J. prepared figures; L.J. drafted manuscript; L.J., A.M.W., and V.H. edited and revised manuscript; L.J., A.M.W., and V.H. approved final version of manuscript.

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Robotic Light Touch Assists Human Balance Control During Maximum Forward Reaching

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Objective: We investigated how light interpersonal touch (IPT) provided by a robotic system supports human individuals performing a challenging balance task compared to IPT provided by a human partner.

Background: IPT augments the control of body balance in contact receivers without a provision of mechanical body weight support. The nature of the processes governing the social haptic interaction, whether they are predominantly reactive or predictive, is uncertain.

Method: Ten healthy adult individuals performed maximum forward reaching (MFR) without visual feedback while standing upright. We evaluated their control of reaching behavior and of body balance during IPT provided by either another human individual or by a robotic system in two alternative control modes (reactive vs. predictive).

Results: Reaching amplitude was not altered by any condition but all IPT conditions showed reduced body sway in the MFR end-state. Changes in reaching behavior under robotic IPT conditions, such as lower speed and straighter direction, were linked to reduced body sway. An Index of Performance expressed a potential trade-off between speed and accuracy with lower bitrate in the IPT conditions.

Conclusion: The robotic IPT system was as supportive as human IPT. Robotic IPT seemed to afford more specific adjustments in the human contact receiver, such as trading reduced speed for increased accuracy, to meet the intrinsic demands and constraints of the robotic system or the demands of the social context when in contact with a human contact provider.

Keywords: interpersonal light touch, robotic assistance, body balance, forward reaching

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INTRODUCTION

If robotic systems are envisaged as the solution to future shortages in clinical staff and caregivers when aiming to augment patients' mobility by a provision of balance support, they must show responsiveness to the social constraints and demands, which govern any physical interaction between a patient and a human carer. Therefore, principles of human–human interactions during physical interactions need to be extracted and evaluated in terms of their transferability to human–robot interactions. When caregivers and therapists routinely provide physical assistance to balance-impaired individuals, they attempt to prevent long-term habitual dependency of a patient on external balance aids and other forms of support. Thus, a therapist aims to adopt an optimum level of postural assistance that maximizes a patient's movement autonomy ("assist-as-needed"). One possible approach is the provision of deliberately light interpersonal touch (IPT), which reduces body sway in quiet standing in neurological patients with impaired postural stability (Johannsen et al., 2017). In such an interpersonal postural context, the contact receiver (CR) experiences haptic contact passively with little or no possibility to influence the interaction due to their greater motion-task constraints compared to those of the contact provider (CP). Not only the movement degrees of freedom available to each individual during IPT, but also the relative postural stability of both partners determines the strength of the interpersonal postural coordination and the individual benefit of IPT, with more enhanced postural stability in the intrinsically less stable person (Johannsen et al., 2012).

To explore the interdependencies between CR and CP during IPT in more detail, we evaluated performance in maximum forward reaching (MFR) with and without light IPT applied to the ulnar side of the wrist of blindfolded CR's extended arm intended to provide a social haptic cue and impose social coordinative constraints on both the CR and the CP (Steinl & Johannsen, 2017). Interestingly, IPT reduced sway more effectively when the CP had the eyes closed and their perception of CR's motion was based on haptic feedback alone. In contrast, IPT with open eyes did not result in reduced sway compared with a condition in which IPT was not provided (Steinl & Johannsen, 2017). Minimization of the interaction forces and their variability at the contact location during IPT might act as an implicit task constraint and shared goal between both partners (Knoblich & Jordan, 2003). This goal might afford predictive sway control in each individual and consequently led to in-phase interpersonal postural coordination with an average zero lag but also minimization of the variability of the interaction force (Johannsen et al., 2009, 2012).

In the present study, we intended to contrast the effects of human IPT (hIPT) on CR's postural performance against the effects of two different modes of robotic IPT (rIPT) and expected specific costs and benefits on body sway and postural performance due to the robotic response modes. Similar to hIPT, rIPT was applied in a "fingertip touch" fashion to CR's wrist without any mechanical coupling or weight support. The robotic system either followed a participant reactively or predicted a participant's movement trajectory. As the coupling between two humans with IPT in terms of the interaction forces is intrinsically more noisy due to each individual's motion dynamics and response delays, we expected that a predictive mode of the robotic system would result in a less noisy haptic coupling and therefore enhance performance in the MFR task, such as greater reaching distance with less body sway. In addition, the reactive mode of the robot was supposed to be advantageous over hIPT due to the fixed response delay, which would enable participants to extract own movement-related information from the interaction forces for balance control.

METHODS

Participants

We tested 10 healthy young adults (average age = 28.5, *SD* 3.35 years, 3 females and 7 males) as CR performing a MFR task. Participants were not affected by any neurological or orthopedic indications. Participants were recruited as an opportunity sample from students of the university. The study was approved by the ethical committee of the medical faculty of the TU Munich and all participants gave written informed consent.

Equipment and Experimental Procedure

CRs stood blindfolded on a force plate (Bertec 4060, Columbus, OH, USA; 500 Hz) in normal bipedal stance (lateral distance between feet was 24 cm) performing the MFR task. CR's body sway was determined in terms of the anteroposterior (AP) and mediolateral (ML) components of the Center of Pressure (CoP), as derived from the six components of the ground reaction forces and moments. Before the start of a trial, CRs stood in a relaxed manner, the right arm extended at shoulder height to reach horizontally above a height-adjusted table. The table provided emergency mechanical support in case of a balance loss but, apart from that, touching the tabletop's surface was not allowed. It also served as a lower boundary constraint keeping participants' hand movements on the same height and preventing drastic changes in the postural strategy, such as increased knee bend, to better enable contact provision in the rIPT conditions. Any explicit instructions for a specific movement strategy were not provided, but CRs had to remain static for at least 5 s (baseline) until an auditory signal cued the start of an MFR trial and then to reach quickly but safely as far forward as possible by bending the torso (Figure 1a).

One healthy adult, male CP applied light IPT to the CR's ulnar side of the wrist with the right extended arm. During IPT, the CP stood in bipedal stance between the CR's force plate and the table, parallel to the reaching direction facing the CR orthogonally. The CP kept the eyes open to receive visual cues of the CR's motion as would the robotic systems by optical motion

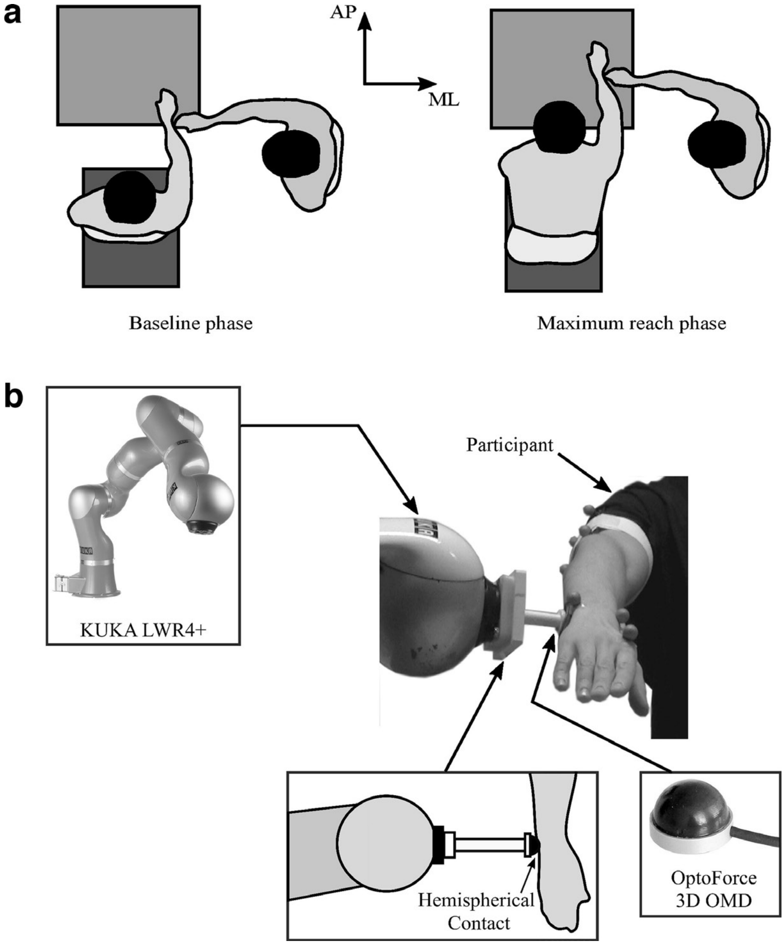

Figure 1. Experimental setup. (a) Execution of the maximum forward reach task with human interpersonal touch (hIPT) support. (b) Robotic IPT without mechanical coupling in hybrid force-position control.

tracking and the CP did wear a thin rubber glove to provide a tactile sensation for the CR similar to rIPT where the end effector of the robot had a rubber surface.

In a pilot experiment, 12 participants were tested in a similar experimental setup but with a force-torque transducer (ATI Nano 17, Apex, NC, USA; 500 Hz) embedded in a wrist bracelet of the extended arm. It was used to acquire the forces and moment in three directions at the contact location during hIPT. Force recordings indicated an average absolute normal interaction force of .15 N (*SD* 0.14) between the CR and the CP, which is lower than the .3 N applied in the rIPT conditions. By the CP being required to grasp a rod mounted onto the force-torque transducer, the wrist bracelet created an unnatural interpersonal link so that it was not used in the hIPT condition of the present study.

During the robotic IPT conditions, a single KUKA LWR4 +manipulator (Augsburg, Germany) served as the CP. The CR's wrist was tracked by the end effector of the robotic system without any mechanical coupling keeping the relative orthogonal distance constant (Figure 1b). The robotic system provided contact via a hemispherical rubber pad attached to a force sensor (OptoForce 3D OMD, OnRobot. Odense, Denmark; 500 Hz) at the end of an "artificial finger." The CR's wrist position was tracked by an optoelectronic motion capture system (OptiTrack, NaturalPoint, Corvallis, OR, USA; 100 Hz) by placing three reflective markers on the CR's right hand (one on the caput ulnae/processus styloideus radii/basis, and two on the ossa metacarpi). The robotic control scheme required high control frequencies to avoid unstable behaviors (Siciliano et al., 2009) and therefore was also controlled at 500 Hz. Hence, motion-tracking data were upsampled in real time to match the robot control frequency.

Three modes of IPT provision were contrasted: hIPT, rIPT with reactively following the participant's movements (rIPTfollow), rIPT with anticipation of the participant's movements (rIPTanticip). The three IPT conditions were assessed in blocks of five trials. The order of the blocked conditions was fully randomized, and each single trial lasted 20 s. Out of a total of 200 trials, 13 trials failed to track the CR's hand and are excluded from the analysis.

Data Reduction

All data post-processing was conducted in Matlab (2016b) (Mathworks, Natick, MA, USA). Kinematic and force-torque sensor data were spline-interpolated to 500 Hz and subsequently merged with the force plate recordings. The data were smoothed using a generic dual-pass, 4th order Butterworth low-pass filter with a cut-off frequency of 10 Hz. CoP and marker data were differentiated to yield velocity. Each trial was segmented into three phases of the MFR (baseline phase, reaching phase, and MFR end-state; Figure 2) based on the AP position of the CR's wrist marker. Reach onset

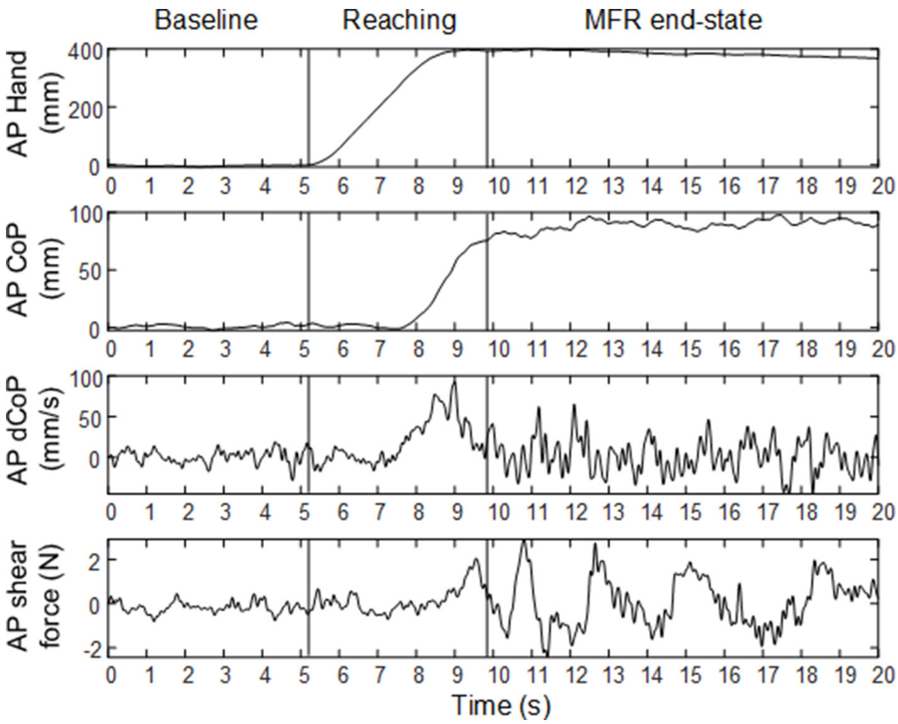

Figure 2. Typical profiles of kinematic and dynamic variables illustrated by data of a single trial for a single participant. Maximum forward reaching (MFR) of the hand marker divided into three phases and the corresponding Centre-of-Pressure (CoP) position, CoP velocity (dCoP), and horizontal shear force in the anteroposterior (AP) direction. Especially, dCoP and the shear force panels well demonstrate the balance challenge imposed by holding a static but unstable posture in the MFR end-state.

was determined as the first frame that exceeded four standard deviations of AP wrist position within the initial 3 s baseline. Stop of forward reaching was determined as the velocity zero-crossing closest to 95% of the absolute maximum reach distance. For the description of the motion dynamics in the MFR end-state, time series data until the end of a trial was used (>10 s).

Several performance measures were selected to characterize participants' movement patterns. Reaching performance was analyzed in the horizontal plane only with two parameters: amplitude and directional angle. Maximum amplitude was determined as the difference between the wrist's average position in the baseline phase and in the MFR end-state. As additional characteristics, curvature in terms of the normalized path length (path length/amplitude), the average and standard deviation of reaching velocity were extracted. In order to quantify the efficiency of balance control, we determined the horizontal CoP amplitude and variability in the MFR end-state as well as calculated the standard deviation of CoP velocity (SD dCoP) as a variability measure for both directions in each reaching phase, as velocity information is predominant for body sway control (Delignières et al., 2011, Jeka et al., 2004; Masani et al., 2003, 2014).

In order to evaluate a potential speed-accuracy tradeoff, we calculated an Index of Performance (IoP) for the control of CoP in the AP direction based on a modification of Fitts and Peterson's IoP (Bootsma et al., 2004; Fitts & Peterson, 1964). Duarte et al. (Danion et al., 1999; Duarte & Freitas, 2005) applied Fitts' law to the balance domain. The unit of the IoP is bit/s (bitrate) and expresses the informational "throughput" of a participant during the movement. An increased IoP resembles greater processing "bandwidth." The IoP was derived from the Index of Difficulty (IoD) over movement time (MT). The IoD is equal to the base 2 logarithm of double the CoP amplitude (ACoP) over the effective dispersion of CoP (WCoP) in the MFR end-state. The averaged standard deviation of CoP position in the AP direction was used as a measure of CoP dispersion. An increased IoD would indicate greater amplitude for a given CoP variability or reduced CoP variability at a given amplitude.

$$\textit{Index of Difficulty}\ (IoD) = \log 2\left(\frac{2*ACoP}{WCoP}\right)$$

$$\textit{Index of Performance}\ (IoP) = \frac{IoD}{MT}$$

Statistical Analysis

The statistical analysis was conducted in R 3.6.1 (RStudio v1.1.456). All performance parameters were log-linearized before statistical analysis to approximate normal distribution. A linear mixed model with IPT condition as four-leveled within-subject factor including participant as random effect was applied using maximum likelihood estimation (lmer function of the lme4 package v1.1–21). For each performance parameter, an α level of .05 was used to test for statistical significance of the main effect of IPT condition. In case that error probability fell below the alpha level, three additional post-hoc comparisons were computed (Helmert contrasts): (1) contrasting the two robotic IPT conditions (rIPTanticip vs. rIPTfollow), (2) contrasting human against robotic IPT (hIPT vs. both rIPT combined), (3) and all IPT combined against No IPT. We applied a corrected α level of .017 to evaluate the statistical significance of the individual contrasts. Effect sizes (d) for mixed-effects models were calculated for the pairwise comparisons (Brysbaert & Stevens, 2018; Westfall et al., 2014).

Robotic Control

Both the robot end-effector position and the interaction force were actively controlled using a hybrid force-position controller based on the prediction of the CR's wrist motion. A Linear Kalman Filter (LKF; Kalman, 1960) with a constant velocity model was exploited to generate a reference for the participant's wrist trajectory. A constant velocity LKF assumes that the motion is generated by the discrete linear system

$$\begin{aligned} s(t+1) &= \begin{bmatrix} p_{KF}(t+1) \\ v_{KF}(t+1) \end{bmatrix} = \begin{bmatrix} I & \Delta t I \\ 0 & I \end{bmatrix} \begin{bmatrix} p_{KF}(t) \\ v_{KF}(t) \end{bmatrix} + \eta \\ &= Fs(t) + \eta, \end{aligned}$$

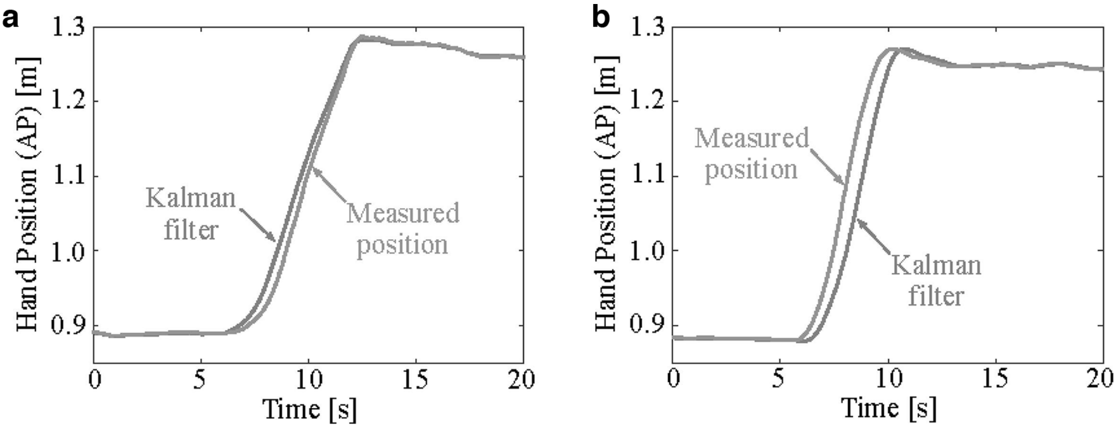

Figure 3. Kalman filtered hand position during maximum forward reaching (MFR). (a) Predicted and measured hand position during MFR for anticipatory robotic interpersonal touch (rIPT) in the anteroposterior (AP) direction. (b) Estimated and measured hand position during MFR for rIPT in follower mode in the AP direction.

where the state vector $s(t)$ contains the Kalman-estimated wrist position $p_{KF}(t)$ and velocity $v_{KF}(t)$, I is an identity matrix, Δt is the sampling time, and η is an additive Gaussian noise. The LKF predicts the next state $s_{KF}(t) = [p_{KF}(t)^{T} \ v_{KF}(t)^{T}]^{T} = Fs(t-1) + y(t)$, where the correction term $y(t)$ is computed as in Kalman (1960) and it depends on the measured wrist position. In our setup, the correction term was set to $y(t) = 0$ until a new measure of the wrist position was available. In this way, the predicted position $p_{KF}(t)$ controlled the robotic system and realized the two different robotic modes. To implement the rIPTfollow mode, the position $p_{KF}(t)$ (position error: AP – .010218 m, ML – .004994 m; Figure 3b) predicted by the LFK at the actual time instant t was used to generate the control command described above. In this way, the robotic system followed the wrist position with one sample delay (10 ms). To generate the rIPTanticip mode, the LKF was exploited to make a one step prediction of the wrist position. In particular, the predicted future position $p_{KF}(t+1) = \mathrm{F}\, P_{KF}(t)$ (position error: AP – 0.012256, ML – .007164 m; Figure 3a) was used to generate the control command. In this way, the robot was anticipating the human motion by one sample (10 ms), thereby leading the movement execution.

The robotic system was controlled to exert a maximum of 1 N force along the ML and vertical directions (force-controlled directions), while tracking the hand motion along the AP axis (position-controlled direction). The force $f_m = [f_{m,x} f_{m,y} f_{m,z}]^{T}$ measured at the contact point and the CR's Kalman-estimated wrist position $p_{KF} = [p_{KF,x}\, p_{KF,y}\, p_{KF,z}]^{T}$ were used to define the desired position of the robot end-effector as $p_x = p_{KF,x} + k_f(f_{m,x} - f_{des})$ and $p_z = p_{KF,z} + k_f(f_{m,z} - f_{des})$. The desired contact force f_{des} was set to .3 N and the gain k_f was set to .00004 m/N, thus regulating the robot motion at the speed of 2.5 mm/s for $f_{m,i} - f_{des} = 1N$ at the update cycle. For the AP direction, the desired robot position was $p_y = p_{KF,y}$. Roughly speaking, the presented controller was adding a delta of position $k_f\,(f_m - f_{des})$ to ML and vertical directions if the measured force was different than f_{des} = .3 N. If the measured force was larger than .3 N, the delta of position was negative and the robot moved slightly back to reduce the force. If the measured force was smaller than .3 N, the delta of the position was positive and the robot pushed slightly against the CR's wrist to remain in contact. In this way, the end-effector kept in contact with the user's wrist while maintaining low interaction forces. The forces were not different between the two rIPT modes. As expected, the average contact force was only slightly higher than the prespecified value of .3N (mean force = .32 N, SD 0.05; rIPTfollow: mean = .31, SD 0.04; rIPTanticip: mean = .32 N, SD 0.05).

TABLE 1: Summary of All Statistical Tests and Comparisons

		Main Effect	Pairwise Comparison		
	Variable	IPT Condition $F(3,30)$; p	No IPT vs. IPT	hIPT vs. Both rIPT T (30); p; d	rIPTAnticip vs. rIPTFollow
Reaching performance	Reaching amplitude	2.32; .10	-	-	-
	CoP displacement amplitude	0.99; .41	-	-	-
	Angular deviation	**3.17; .04**	−2.04; .05; .13	−2.29; .03; .20	*n.s.*; .05
	Curvature index	**24.88;** <.001	**8.52; <.001; .54**	*n.s.*; .06	*n.s.*; .19
	AV reaching velocity	**11.41;** <.001	**5.02; <.001; .21**	**3.00; .006; .18**	*n.s.*; .03
	SD reaching velocity	**14.48;** <.001	**4.40; <.001; .14**	**4.88; <.001; .22**	*n.s.*; .04
Body sway (*SD* dCoP)	Baseline (AP)	**8.81;** <.001	**4.53; <.001; .28**	2.40; .02; .21	*n.s.*; .05
	Baseline (ML)	2.79; .06	-	-	-
	Reaching (AP)	**17.97;** <.001	**5.97; <.001; .26**	**3.99; <.001; .24**	*n.s.*; .16
	Reaching (ML)	**11.96;** <.001	**4.98; <.001; .16**	**3.08; .004; .14**	*n.s.*; .10
	MFR end-state (AP)	**4.22; .01**	**3.53; .001; .20**	*n.s.*; .03	*n.s.*; .03
	MFR end-state (ML)	1.63; .20	-	-	-
Efficiency of body sway control	Index of Difficulty (CoP)	1.09; .37	-	-	-
	Index of Performance (CoP)	**6.99; .001**	**4.20; <.001; 0.15**	*n.s.*; 0.10	*n.s.*; 0.001

Note. IPT = interpersonal touch; hIPT = human IPT; rIPTanticip = robotic IPT anticipating; rIPTfollow = robotic IPT following; SD dCoP = standard deviation of centre-of-pressure velocity; AP = anteroposterior; ML = mediolateral; MFR = maximum forward reach. + = marginally significant; n.s. = not significant. Main effect α level is .05, α level for the single comparisons is .017. Significant effects are printed in bold.

RESULTS

Table 1 summarizes the statistical results of all main effects and single comparisons. The MFR amplitudes for the trajectories of hand and CoP in the horizontal plane were not affected by any IPT condition. All three IPT conditions resulted in comparable amplitudes for the hand (rIPTfollow: mean = 35.1 cm, SD 3.9; rIPTanticip: mean = 35.4 cm, SD 4.5; hIPT: mean = 35.8 cm, SD 5.1; No IPT: mean = 36.8 cm, SD 4.6) and CoP (rIPTfollow: mean = 6.7 cm, SD 2.3; rIPTanticip: mean = 6.5 cm, SD 2.8; hIPT: mean = 6.3 cm, SD 2.7; No IPT: mean = 6.5 cm, SD 2.7) compared to No IPT.

Average (Figure 4a) and the variability of planar reaching velocity (Figure 4b) of the wrist were lower in both rIPT conditions compared to hIPT and in all IPT conditions compared to No IPT. The directional angle of reaching in the horizontal plane tended to show less deviation from the AP axis in the rIPT conditions than in hIPT and No IPT (Figure 4c). The planar curvature index in terms of the normalized path length indicated straighter reaching in all three IPT conditions compared to No IPT (Figure 4d).

In the ML direction in the baseline phase and the MFR end-state, sway variability was not different between the four IPT conditions (Figure 5). During the reaching, however, ML sway variability was reduced in both conditions involving rIPT compared to hIPT and all three IPT conditions compared to No IPT. In the AP direction on the other hand, all three IPT conditions showed reduced sway compared to No ITP across the baseline phase, the reaching, and the MFR end-state. In addition, rIPT showed less sway than

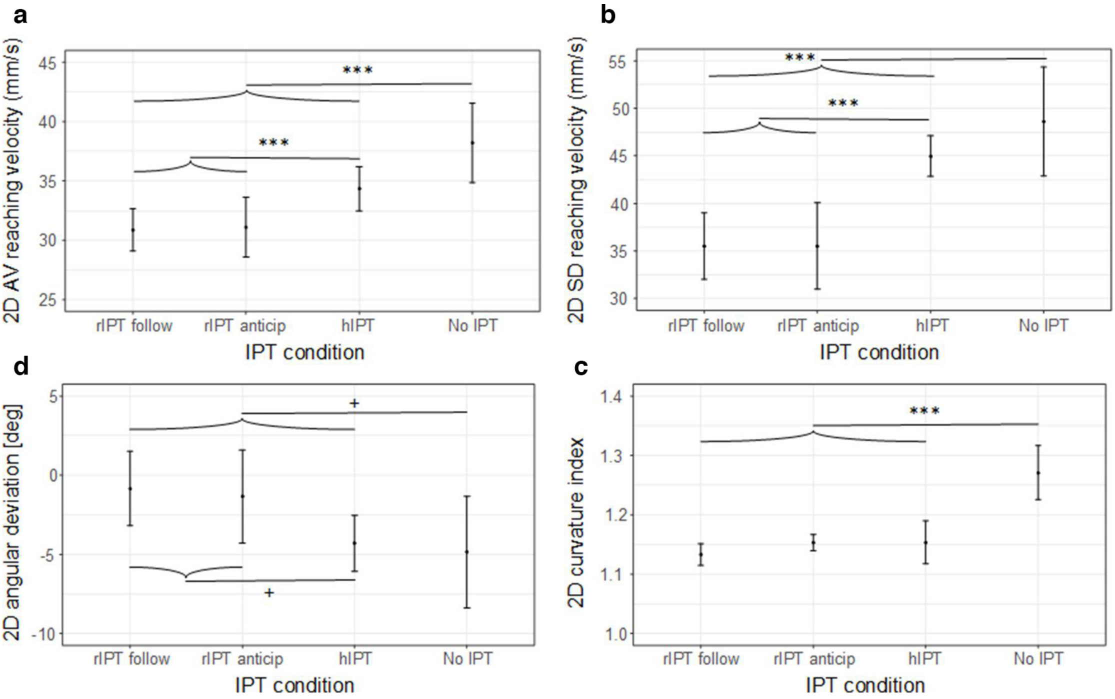

Figure 4. Parameters of reaching performance as a function of the interpersonal touch (IPT) condition: (a) average planar velocity of the hand, (b) variability of planar velocity of the hand, (c) average deviation from a straight line linking the start to the end positions, (d) curvature index in terms of the normalized path length of reaching. Error bars show the standard error of the mean across participants. Horizontal brackets indicate significant within-subject post-hoc single comparisons (+$p < .05$ and $p > .017$; ***$p < .001$). hIPT: human IPT; rIPTanticip: anticipatory robotic IPT; rIPTfollow: robotic IPT in follower mode.

hIPT during reaching and a tendency for a reduction in the baseline phase (Figure 5).

The IoD did not differ between the four IPT conditions (Figure 6a), while the IoP indicated a lower bitrate in the three IPT conditions compared to No IPT (Figure 6b).

DISCUSSION

Our study contrasted the effects of deliberately light IPT received by a robotic system on the control of movements and body balance during MFR in healthy young adults. Changes in spontaneous MFR behavior and body sway were assessed as a function of the robotic system's mode of control (follower vs. anticipation) with respect to the CR's movements. Although we assumed that participants would not be able to consciously perceive any difference between the anticipatory and follower rIPT modes, we nevertheless expected subtle, spontaneous alterations in their MFR behavior indicative of a performance facilitation at best or a disruption in the worst case. Unexpectedly, no differences between the two rIPT modes were observed. In addition, rIPT demonstrated effects comparable to hIPT with respect to body sway in the baseline, reaching phase, and MFR end-state. All three IPT conditions resulted in increased stability in the AP direction. The achieved amplitude, however, was not different from the amplitude achieved without IPT.

Reaching Performance and Body Sway

Augmentation of perceived self-motion relative to the environment by light, mechanically nonsupportive tactile contact with an earth-fixed reference improves body equilibrium and postural control (Holden et al., 1994; Jeka &

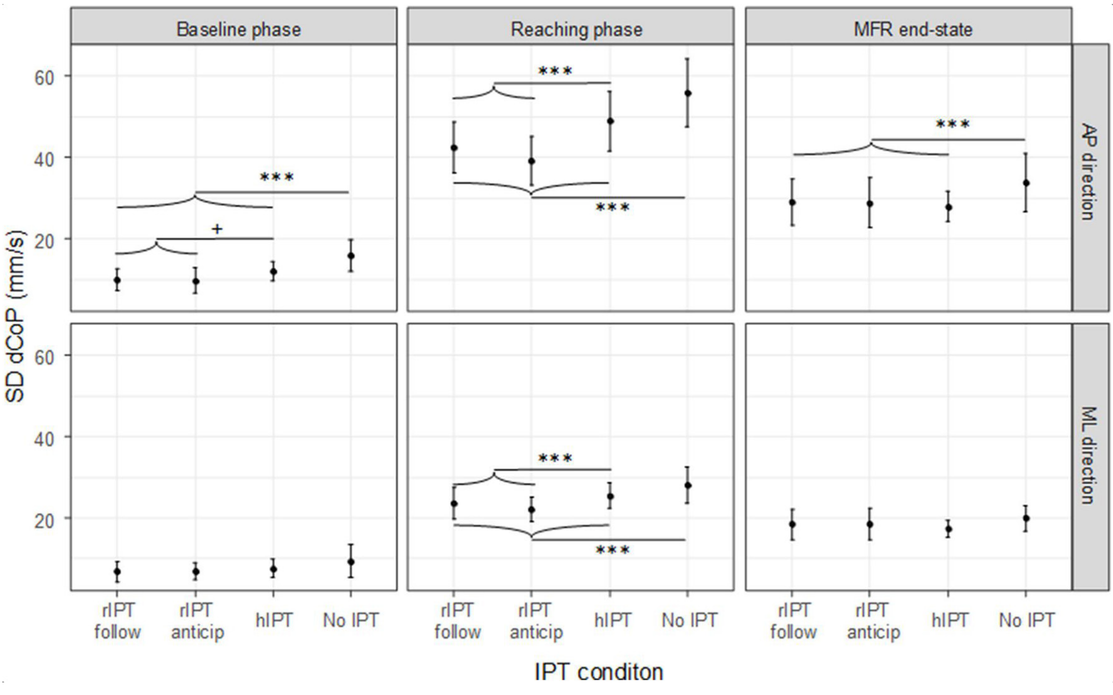

Figure 5. Body sway in terms of the standard deviation of Centre-of-Pressure velocity (SD dCoP) as a function of the interpersonal touch (IPT) condition in the anterior-posterior (AP) and mediolateral (ML) direction in all three phases of the maximum forward reaching (MFR) task. Error bars show the standard error of the mean across participants. Full horizontal brackets indicate significant within-subject post-hoc single comparisons (+$p < .05$ and $p > .017$; ***$p < .001$). hIPT: human IPT; rIPTanticip: anticipatory robotic IPT; rIPTfollow: robotic IPT in follower mode.

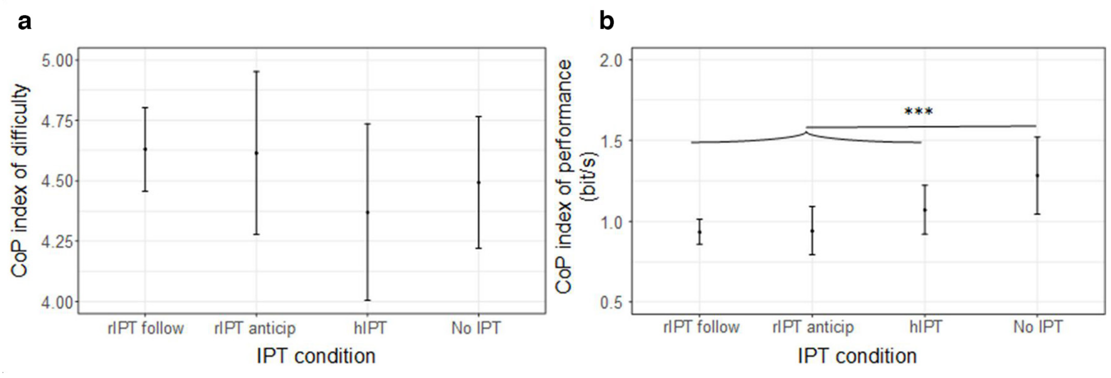

Figure 6. Index of difficulty (a) and Index of performance (b) for CoP motion in each IPT condition for both directions. Error bars show the standard error of the mean across participants. Full horizontal brackets indicate significant within-subject post-hoc single comparisons (***$p < .001$). hIPT: human IPT; rIPTanticip: anticipatory robotic IPT; rIPTfollow: robotic IPT in follower mode.

Lackner, 1994). Touch can also be utilized to stabilize body sway when the tactile contact is received passively (Krishnamoorthy et al., 2002; Rogers et al., 2001). Lightly touching an oscillating contact shows strong effects in terms of body sway entrainment and can be used to drive an individual's body sway in quiet standing (Jeka et al., 1997, 1998; Verite et al., 2013; Wing et al., 2011). Therefore, it may be possible that even in a more dynamic postural context such as the MFR task in the present study, motion of the light touch contact "attracts" swaying motion of the body. For example, subtle forward motion of the contact could be wrongly interpreted as backward sway so that a forward adjustment would follow the contact's lead. This effect could have been more pronounced in the rIPTanticip than the rIPTfollow condition. Although, we have not found any evidence to support this assumption.

An increased MFR amplitude would demonstrate improved confidence in the ability of keeping own body balance stable while approaching one's forward limits of stability (Duncan et al., 1990; Maki & McIlroy, 2006). As we did not observe any difference in reaching amplitude between all four conditions, it also means that IPT provided by a robotic system or a human did neither disrupt nor distract the human CR. This observation corresponds to the previous study, in which hIPT also did not affect reaching distance (Steinl & Johannsen, 2017). On the other hand, a general reduction in MFR velocity and its variability was an obvious change in their behavior when IPT was provided by the human partner or the robotic system. As body sway was reduced in these situations too, these adjustments could reflect a trade-off between speed and accuracy (Fitts, 1954). Participants may have effectively controlled sway variability more carefully to fulfill the task goal of MFR with IPT support in the face of either "hardware" constraints imposed by technical limitations of the robotic system or social constraints imposed by the human partner (Bardy et al., 1999; Scholz & Schöner, 1999). The fact that the IoP indicated narrower informational throughput during the reaching movement in the three IPT conditions compared to No IPT, however, could mean that all IPT conditions were burdened with an additional processing load. Possibly due to a shift in participants from less to more reactive, feedback-dependent postural control, CRs increased their movement time to adjust their motion more precisely to the current position of the robotic end-effector or the human partner and/or to allow the same to stay in better contact with their own wrist.

Human–Robotic Movement Coordination

Haptic interactions between caregiver and patient play an prominent role in cooperative and collaborative human-human sensorimotor interactions in physical rehabilitation (Sawers & Ting, 2014). More recently, Haarman et al. (Haarman et al., 2017) investigated the balance-assistive forces applied by therapists to the pelvis of patients during gait training. Using force-torques sensors, they quantified the predominant corrective forces applied by the therapists in the mediolateral direction to both sides of the hips at about 9N, amounting to approximately 2% of participants' body weight. Compared to the forces imposed by the robotic systems in our current study, the forces applied by the therapists are still by magnitudes greater.

In a cooperative physical human–human interactions, the relationship between interaction forces and movement kinematics is important for communicating intended movement direction (Mojtahedi et al., 2017; Sawers et al., 2017; Takagi et al., 2018). Gentry and Murray-Smith (2003) described the influence of haptic signals used for coordination and synchronization in human dancing. Hoelldampf et al., 2010 used interaction forces to adjust and optimize the robot's motion in a system designed for human–robot interactive dancing. Similarly, Chen et al. (2015, 2017) developed a mobile robotic system responsive to interaction forces to practice dance stepping with a human partner. Response gain and compliance of the robot's effectors altered human upper body posture and human–robot coordination. Interestingly, the majority of human partners perceived the robots as following their movements (Chen et al., 2015).

"Assist-as-needed" (Cai et al., 2006) robotic devices will provide corrective forces only if a participant's limb movement kinematics hit the walls of a predefined "virtual tunnel" (Duschau-Wicke et al., 2010) aiming to keep an individual's body or limbs within an initially defined "normal" range. In contrast to this kind of "positive" force feedback, our deliberately light interpersonal touch paradigm could be considered to act with "negative" force feedback. This means that if participants stray from a reaching trajectory, they will perceive a reduction in touch, which might cue them to perform a subtle correction, such as moving toward the contact, with the intention to keep a constant force and to minimize contact force variability. In this sense, the robotic system in our study was controlled according to a similar principle and we believe it imitated the behavior of the CR and CP more naturally. The reaching trajectories were not prespecified within the robotic system but emerged as a compromise between the CR and the respective CP so that the CR's movements remained unconstrained physically.

In this context, it is remarkable that rIPT led to straighter forward reaching trajectories with least amount of medial drift. This could mean that a robotic system is a better haptic "communicator" in the sense that it made participants "listen" more closely to the haptic feedback they received. The dynamics of the robotic system were not independent but a direct consequence of CR's movements. Despite the lack of any real "social cognitive" capabilities of the robotic system, this fact can nevertheless be interpreted as highly precise responsiveness, which a human CP could never match. Possibly, participants interpreted rIPT as a more reliable spatial reference and therefore adjusted their reaching movements more in a feedback-driven manner.

Influence of Visual Feedback for CP

The provision of hIPT involved visual feedback or optical tracking of CR's body and movements. In human pairs, the presence of visual feedback with habitual visual dominance is likely to turn the CP into a follower of CR's movement (Steinl & Johannsen, 2017). Assessing human–human as well as human–robot interactions in a single degree of freedom object manipulation task, Groten et al. (2009a, 2009b) characterized inter-agent dominance as a function of the interaction force with dominance between both partners varying flexibly. Generally speaking, in most physical interactions between 2 human individuals leader-follower relationships are not necessarily fixed. It seems to be the case, however, that the more adaptive individual, for example the person on whom fewer requirements to fulfill specific movement constraints are imposed, is more likely to take a follower role (Skewes et al., 2015). This interpretation implies that in hIPT the CP coordinated the movements in a reactive fashion as well, potentially in follower mode due to visual dominance.

Limitations

The results of our study are subject to limitations, such as small sample size limiting not only the possibility to generalize our findings to a wider population of older adults or patients with disturbed body balance. Similarly, our experimental setup and task represent a specific laboratory situation that imposed specific constraints onto participants. As a consequence, the generalizability of our findings to other postural tasks and daily life activities is restricted too. Another limitation is the lack of force recordings in the hIPT condition. As we do not know the absolute interaction forces applied between the CP and CR, it could mean that IPT had not been applied in a light fashion and therefore potentially influenced the CR's movements in some way. We believe, however, that touch had being applied lightly in our hIPT condition as the overall movement pattern observed in hIPT was not dramatically different from either the No IPT or the rIPT conditions in the present study as well as hIPT in a pilot experiment, where the interaction forces and torques were being recorded. Usually, hIPT tended to fall in between No IPT and rIPT, which implies that the mechanical coupling between both partners was not much stronger than in the rIPT conditions. The possibility remains that phases occurred during which contact between the CP and CR was not present. One way to evaluate the movement coupling between both partners would be the recording of both partners' movement dynamics. Unfortunately, our setup was limited to the acquisition of only CR's motion for the lack of a second force plate and more extensive motion capture coverage.

CONCLUSIONS

Beneficial deliberately light IPT for balance support during MFR is easily provided by a robotic system even when it is mechanically uncoupled to the human CR. This effect does not rely on the system's capability to predict the future position of the CR's wrist. As the robotic system itself was not designed for any form of "social" cognition or explicit haptic communication, our study nevertheless demonstrates that robotic IPT can be used to implicitly "nudge" human CRs to alter their postural strategy for adapting to the robotic system without any decrements in their postural performance during MFR.

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KEY POINTS

- Robotic light touch supports human balancing performance during forward reaching.
- Human participants seem to adapt to the specific affordances of robotic light touch support.
- Subtle differences in the relative time lags between the robotic modes of interaction did not result in behavioral effects.

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Deliberately Light Interpersonal Touch as an Aid to Balance Control in Neurologic Conditions

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Abstract

Purpose: We aimed to quantify the benefit of externally provided deliberately light interpersonal touch (IPT) on body sway in neurological patients.

Design: IPT effect on sway was assessed experimentally across differing contacting conditions in a group of 12 patients with Parkinson's disease and a group of 11 patients with chronic hemiparetic stroke.

Methods: A pressure plate recorded sway when IPT was provided by a healthcare professional at various locations on a patient's back.

Findings: IPT on the back reduced anteroposterior body sway in both groups. Numerically, IPT was more effective when applied more superior on the back, specifically at shoulder level, and when applied at two contact locations simultaneously.

Conclusion: Our findings demonstrate the benefit of deliberately light IPT on the back to facilitate patients' postural stability.

Clinical Relevance: Deliberately light IPT resembles a manual handling strategy, which minimizes load imposed on healthcare professionals when providing balance support, while it facilitates patients' own sensorimotor control of body balance during standing.

Keywords: Interpersonal coordination; light touch; body balance control; hemiparetic stroke; Parkinson's disease.

Introduction

A frequent consequence of neurologic conditions, such as Parkinson's disease or hemiparetic stroke, is a compromised control of body balance leading to an increased fall risk (Forster & Young, 1995), which demand support often provided manually by a healthcare professional (HCP) such as a nurse or physiotherapist in clinical settings. The main characteristic of manual support is partial transfer of the patients' body weight onto the HCP, thereby increasing the risk of work-related musculoskeletal injury. Waters and Rockefeller (2010) consider therapeutic manual patient handling tasks as especially risky due to the longer durations of supporting a patient's weight. Aside from a few exceptions, recommending manual support provided at the waist and upper trunk of a patient (ACC, 2003; Potter & Perry, 2005), many rehabilitation guidelines on stroke and Parkinson's disease do not give specific advice on manual handling (AGS, 2001; Keus et al., 2004; McInnes, Gibbons, & Chandler-Oatts, 2005; NICE, 2013; NSF, 2010; SIGN, 2010) but recommend that HCPs undertake some training in physical handling techniques (NSF, 2005).

To reduce the risk of musculoskeletal injury in carers, standardized handling methods such as "minimal lifting" and "no lifting" policies have been suggested (WorkCover, 2006). On the other hand, mechanical transfer of body weight is not required to improve postural stability during quiet standing. Often it can be sufficient to receive tactile sway feedback by lightly contacting an external reference (contacting force <1 N; Jeka, 1997). In this context, a distinction is made between "active" contact, where the receiver extends a limb and keeps it at the contact location, and "passive" contact, where contact is provided externally to the leg, the shoulder, the head, or the neck (Krishnamoorthy, Slijper, & Latash, 2002; Menz, Lord, & Fitzpatrick, 2006; Rogers, Wardman, Lord, & Fitzpatrick, 2001). Reductions in sway tend to be greater the more superior the contact point is located (Rogers et al., 2001), and the combination of

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two simultaneous, bilateral contact points facilitates sway reductions more than a single unilateral contact (Dickstein, 2005).

Sway reduction has also been demonstrated when two people make light touch contact (interpersonal touch [IPT]; Johannsen, Guzman-Garcia, & Wing, 2009; Johannsen, Wing, & Hatzitaki, 2012). This finding has clinical implications as an HCP might support a patient through light touch at arm, shoulder, or on the back. Indeed, specific nursing concepts have incorporated light tactile support to facilitate patients' sensorimotor performance (e.g., Hatch, & Maietta, 2003; see also Betschon, Brach, & Hantikainen, 2011). Nevertheless, empirical evidence in favor of specific manual handling techniques and quantitative assessments of changes in patients' balance control during the provision of interpersonal contact is not provided.

In contrast to active light contact with an earth-fixed reference, IPT resembles an ambiguous signal as it contains both self-imposed and externally imposed dynamics. To utilize this signal for sway control, one needs to be able to distinguish self-motion from any external sources. To what degree this ability is compromised in neurological disorders such as Parkinson's disease and hemiparetic stroke is not clear. Therefore, in this study, we aimed to investigate whether deliberately light IPT benefits both groups of patients during upright quiet standing. In line with previous results in the literature for passively applied external light touch, we assumed that more superior locations result in greater reductions in sway and that the provision of two contact points increases the overall benefit.

Methods

Data were acquired in two opportunity samples of patients with Parkinson's disease and patients with chronic hemiparesis following stroke. Participants were recruited and tested at the National Institute of Conductive Education in Birmingham, UK, where they attended regular therapy sessions. All testings took place in the morning before the start of such a session.

Participants gave written informed consent for taking part. The research project was approved by the University of Birmingham Ethics Committee. Effects of "passive" IPT on sway were assessed as a function of the specific contact locations on participants' back of the upper body either at a single location or at two combined locations at the same time. Both groups were exposed to slightly different protocols with respect to the specific IPT test conditions. The reason was that, in addition to the three contact locations that were applied in both studies (No contact, Low back, and High back), we were also interested in exploring specific dual-contact situations without strictly repeating each condition in each patient group. A comparison between the two patient samples was not intended.

All participants received instructions to stand quietly in normal bipedal stance on a force plate (Wii-Fit balance board, Nintendo, Redmond, WA) with their hands by their side. The Wii-Fit has been reported as an accurate data acquisition device (e.g., Bartlett, Ting, & Bingham, 2014). The device recorded sway in terms of Center-of-Pressure (CoP) position, which represents the net forces and moments generated by the neuromusculoskeletal system during standing. Each experimental trial lasted 20 seconds, and two trials were recorded for each IPT condition. The sequence of IPT conditions was randomized for each participant. Output from the balance device was sampled at 40 Hz and provided anteroposterior (AP) and mediolateral (ML) components of CoP motion. To eliminate noise not attributable to postural adjustments, CoP data were smoothed using a moving average with a window width equivalent 100 milliseconds in Matlab 7.5 (Mathworks, Natwick, MA) and CoP was differentiated (dCoP) to remove low-frequency drift due to voluntary weight shifts. Body sway variability was expressed as the standard deviation of dCoP (*SD* dCoP). For every participant, data were averaged across the two trials for each IPT condition.

A chair was placed in front with the backrest toward participants so that they could reach forward to stabilize themselves. Also, a therapist stood next to the participant (on the nonparetic side in the case of the stroke patients) to provide emergency support if required. The two IPT providers, who were trained with contact force feedback in our lab, stood always behind a participant within arm's reach to apply the different forms of IPT using all four fingers of the right hand. During IPT, the provider adjusted position of the contacting hand according to the body motion of a participant to keep steady contact. Participants were instructed to stand relaxed, not to move their limbs during a trial, and not to attempt to lean against the contact provider. Touch was always applied through light clothing adequate for the subsequent therapy session. Quantitative measurements of contact force were not practical as the data acquisition took place off-site.

In the first experiment, 12 participants with Parkinson's disease were recruited (mean age = 69.3 years, *SD* = 7.7; six women, six men). An average Hoehn and Yahr score (Hoehn & Yahr, 1967) of 3.08 (*SD* = 0.76) indicated that participants showed mild-to-moderate symptoms of Parkinson's disease with still noticeable balance impairments. All were able to stand unsupported for at least 20 seconds with their eyes closed. At the time of testing, all were on their respective medication to reduce the Parkinson's symptoms. We assumed that the medication increased responsiveness to the IPT stimulus, whereas the deprivation of visual feedback

Table 1 Participants' clinical and demographic information

Group	Participant	Age (years)	Gender		Time Since Diagnosis (years)	Hoehn and Yahr Scale
Parkinson	1	81.3	Male		2	2
Parkinson	2	65.2	Female		11	3
Parkinson	3	61.2	Male		4	2
Parkinson	4	70.8	Female		5	2.5
Parkinson	5	74.1	Male		14	4
Parkinson	6	62.7	Female		12	4
Parkinson	7	78.4	Female		5	2.5
Parkinson	8	71.6	Male		14	4
Parkinson	9	61.4	Female		3	3
Parkinson	10	56.5	Female		6	3
Parkinson	11	74.2	Male		14	4
Parkinson	12	73.7	Male		3	3
Group	**Participant**	**Age (years)**	**Gender**	**Lesioned Hemisphere**	**Time Since Lesion (years)**	**Rivermead Mobility Index**
Stroke	13	82.9	Male	Left	2.3	7
Stroke	14	71.3	Male	Left	7.6	8
Stroke	15	77.9	Female	Left	4.3	11
Stroke	16	76.7	Female	Right	1.5	7
Stroke	17	42.8	Female	Right	0.4	8
Stroke	18	58.4	Female	Right	2.0	9
Stroke	19	69.1	Male	Left	2.9	9
Stroke	20	72.8	Female	Right	0.9	7
Stroke	21	48.6	Male	Left	2.5	7
Stroke	22	34.1	Male	Left	1.6	11
Stroke	23	69.7	Female	Left	0.7	7

increased patients' dependency on the tactile stimulus. Table 1 provides clinical and demographical information for all individuals in both groups of participants.

Sway was assessed under six different conditions. Five entailed some form of IPT: (a) No contact, (b) IPT over the spine at waist level (Low back), (c) IPT over the spine at shoulder level (High back), (d) IPT over the spine at waist level and at the right elbow (Low + elbow), (e) IPT over the spine at shoulder and waist level simultaneously (High + low), and (f) IPT on the back close to the left and right shoulders simultaneously (Dual high). All IPT locations are shown in Figure 1.

In the second experiment, 11 participants with chronic hemiparetic stroke volunteered to take part (mean age = 64 years, *SD* = 15.9; six women, five men). An average Rivermead Mobility Index (Collen et al., 1991) of 8.27 (*SD* = 1.56) indicated that participants were able to stand unsupported but unable to walk longer distances without help. For a number of participants in this group, however, standing unsupported with eyes closed posed too challenging. Therefore, participants were tested with eyes open, which allowed all participants to stand unsupported for at least 20 seconds. Stroke participants were tested under the following IPT conditions: (a) No contact, (b) No contact but with assurance that the experimenter would provide contact in case the participant would become unsteady (No contact+), (c) IPT at the nonparetic elbow, (d) IPT over the spine at waist level (Low back), (e) IPT over the spine at shoulder level (High back), and (f) IPT over the spine at shoulder level and at the nonparetic elbow (High back + elbow).

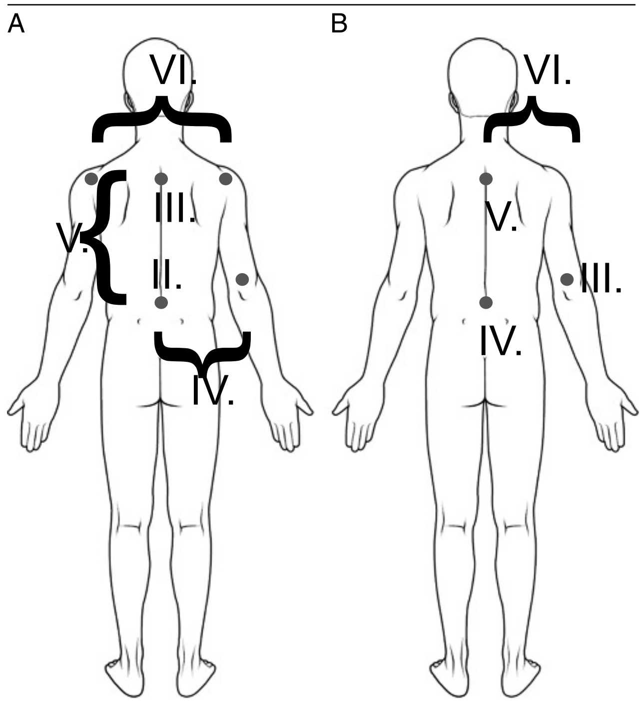

Figure 1. Haptic contact locations on the back of the patients with Parkinson's disease (A; conditions II–VI) and the back of the patients with chronic stroke (B; conditions III–VI). The conditions not shown did not involve haptic contact.

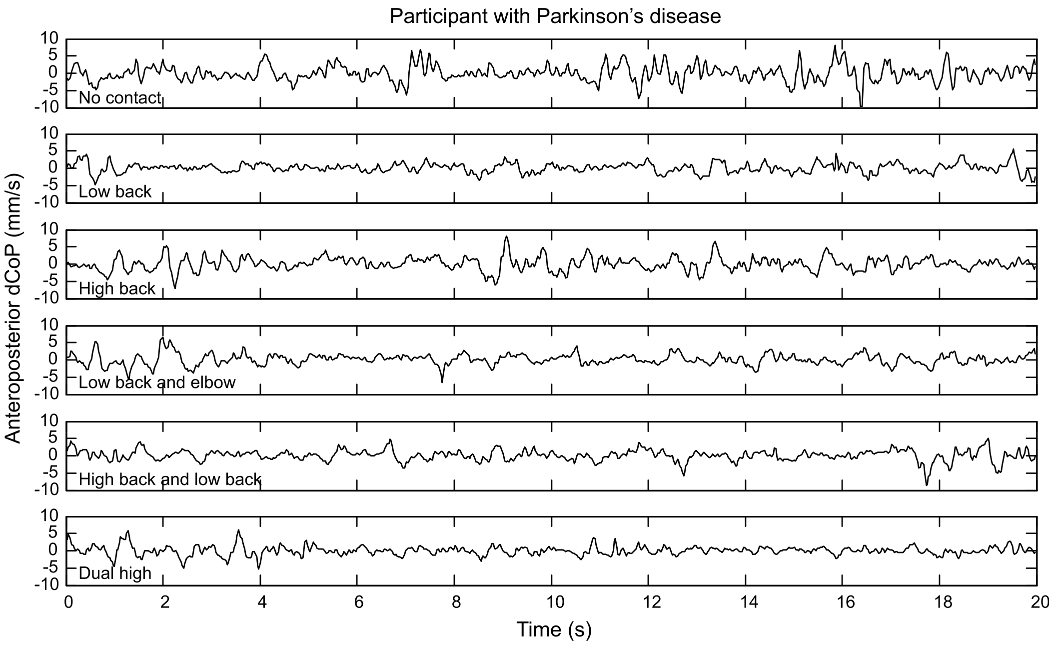

Figure 2. Anteroposterior Center-of-Pressure rate of change (dCoP) time series of a single participant with Parkinson's disease for each of the six interpersonal touch conditions.

One-way repeated-measures analyses of variance (ANOVAs; SPSS 20, IBM Corporation, Somers, NY) were computed on *SD* dCoP in both direction of sway with IPT conditions as within-subject factor. A Greenhouse–Geisser corrected level of significance at p = .05 was used. Post hoc single comparisons were performed between IPT conditions when necessary.

Results

Parkinson's Disease

Figure 2 shows illustrative AP dCoP time series of a Parkinson's patient for each of the six IPT conditions. Clearly, dCoP was less variable in the IPT conditions compared to the No contact condition. Average AP sway was 80.9 mm/s (*SD* = 52.1) and 25.3 mm/s (*SD* = 15.4) in the ML direction. Variability of body sway was different between the six IPT conditions in the AP direction ($F[5,55]$ = 4.65, p = .02, partial η^2 = 0.30) and tended to be different in the ML direction ($F[5,55]$ = 2.60, p = .10, partial η^2 = 0.19). On the AP axis, all post hoc single comparisons between each IPT condition and No contact were significant (all $F[1,11] \geq 5.59$, all $p \leq .04$, all partial $\eta^2 \geq 0.34$). Each of the IPT conditions led to a reduction in sway variability compared to No contact. The greatest reduction (26%) was present during IPT with two contact points at shoulder level (Dual high), followed by Low back with simultaneous elbow contact (22%), High back with Low back contact (19%), and both single Low and High back, respectively (both 16%).

On the ML axis, only Low back with simultaneous elbow contact (14%) resulted in a significant sway reduction ($F[1,11]$ = 4.67, p = .05, partial η^2 = 0.30). Tendencies for a sway reduction were found for High back with simultaneous Low back contact (15%) and Dual high back contact (20%; both $F[1,11] \geq 3.30$, both $p \leq .10$, both partial $\eta^2 \geq 0.23$). Figure 3 shows *SD* dCoP in both directions of sway for each IPT condition for the Parkinson's

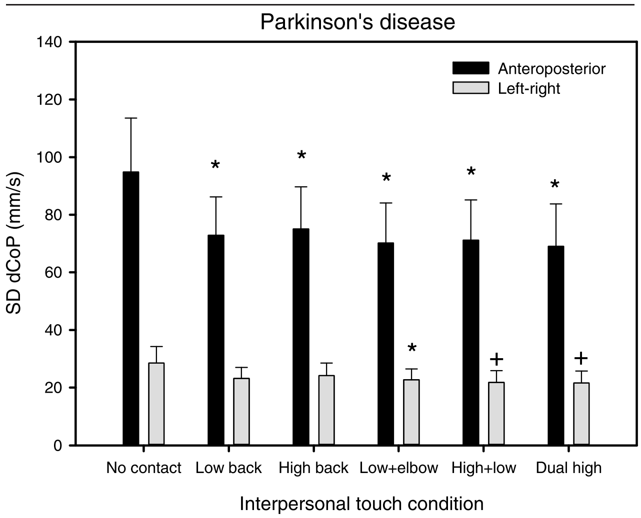

Figure 3. Variability in sway for the patients with Parkinson's disease as a function of the respective interpersonal touch (IPT) condition and the direction of sway. Low back: IPT over the spine at waist level, High back: IPT over the spine at shoulder level, Low + elbow: IPT over the spine at waist level and at the right elbow, High + low: IPT over the spine at shoulder and waist level simultaneously, Dual high: simultaneous IPT on the back close to the left and right shoulders. *Significant sway reduction compared to No contact (p < .05); +Tendency for sway reduction compared to No contact (p < .10).

group. A comparison between the subgroups of patients' with Hoehn–Yahr score Stage 2 (below 3) and Stage 4 did neither show a group difference in body sway nor an interaction between group and IPT condition. Similarly, Spearman correlations did not indicate any associations between patients' Hoehn–Yahr score and the reduction in sway in the Dual high back condition for both directions of sway.

Hemiparetic Stroke

Figure 4 shows illustrative AP dCoP time series of a chronic hemiparetic stroke patient for each of the six IPT conditions. Especially, the two IPT conditions, which included contact at the high back, showed less variable dCoP. Average AP sway was 62.3 mm/s (*SD* = 27.9) and 18.4 mm/s (*SD* = 7.2) in the ML direction. We averaged the two IPT conditions without touch contact as sway variability was not different between the two. In the AP direction, the main effect of IPT condition was significant ($F[4,40] = 3.83$, $p = .03$, partial $\eta^2 = 0.28$). Post hoc single comparisons between the No contact conditions and those IPT conditions involving touch demonstrated that IPT at the nonaffected elbow, at the lower back, and at the higher back, respectively, and High back with simultaneous and elbow contact resulted in reduced AP sway variability (all $F[1,10] \geq 4.76$, all $p \leq .05$, all partial $\eta^2 \geq 0.32$). The reduction in sway was greatest during IPT over the spine at High back (15%), followed by contact at High back with simultaneous contact at the nonaffected elbow (14%), and both Low back contact (11%) and elbow contact (11%), respectively, showing the least reduction.

The tendency for a main effect of IPT condition was found in the ML direction as well ($F[4,40] = 2.76$, $p = .07$, partial $\eta^2 = 0.22$). Sway in the ML direction was reduced with High back contact ($F[1,10] = 5.84$, $p = .04$, partial $\eta^2 = 0.37$) and tended to be reduced in the other IPT conditions (all $F[1,10] \geq 3.80$, all $p \leq .08$, all partial $\eta^2 \geq 0.28$). Again, High back resulted in the greatest sway reduction (20%), followed by High back with simultaneous elbow contact (18%), Low back (17%), and finally single elbow contact (16%). Spearman correlations showed that stroke patients' Rivermead Mobility Index and the amount of reduced sway during High Back contact were not significant for both sway directions. Figure 5 shows *SD* dCoP in both directions of sway for each IPT condition for the participants with hemiparetic stroke.

Discussion

We evaluated the benefit of deliberately light IPT on sway during quiet standing in patients with Parkinson's disease and in patients with chronic hemiparetic stroke. We found evidence in both groups that IPT stabilizes body sway. Generalized across both groups, sway reductions were effective irrespective of contact location but did not apply equally well to all the body locations and combinations assessed. A more superior location at shoulder level tended to result in greater reductions in postural sway than contact at waist level. Two simultaneous contact points of which one was

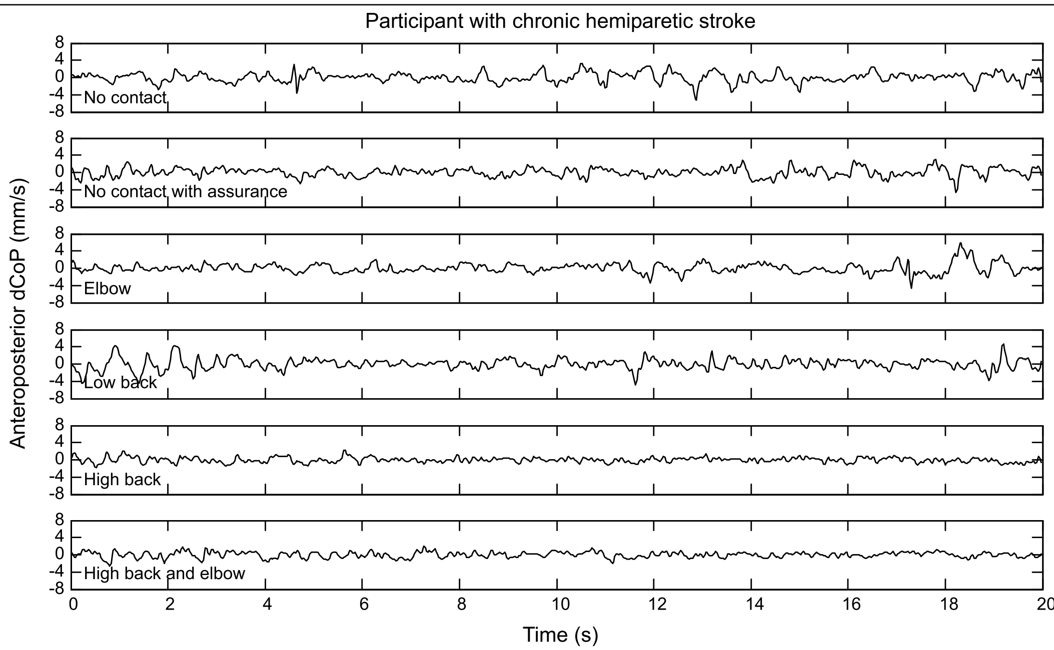

Figure 4. Illustrative Center-of-Pressure rate of change (dCoP) traces of a single participant with chronic hemiparetic stroke for each of the six interpersonal touch conditions.

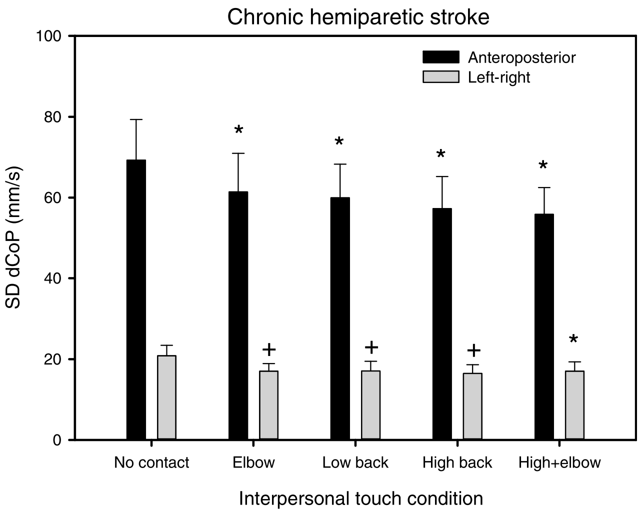

Figure 5. Variability in sway for the patients with chronic hemiparetic stroke as a function of the respective interpersonal touch (IPT) condition and the direction of sway. No contact+: No contact but with assurance that the experimenter would provide contact in case the participant would become unsteady, Low back: IPT over the spine at waist level, High back: IPT over the spine at shoulder level, High back + elbow: simultaneous IPT over the spine at shoulder level and at the nonparetic elbow. *Significant sway reduction compared to No contact ($p < .05$); +Tendency for sway reduction compared to No contact ($p < .10$).

located at the high back at shoulder level generated the greatest proportional reductions. We reason that the inverted pendulum-like dynamics of upright standing result in increasing relative motion of the contacted segment as well as shear forces under the contact point with increasing distance from the ankle pivot. Alternatively, a high contact point might provide a vertical reference for a greater number of body segments and joints.

Impaired balance function has been observed in Parkinson's disease during dynamic activities such as reaching, turning, and walking, but increased sway has also been reported for quiet upright standing with eyes closed (Blaszczyk, Orawiec, Duda-Klodowska, & Opala, 2007; Viitasalo et al., 2002). In stroke, the spectrum of impaired balance control ranges from severe inability to keep an upright body orientation to more subtle deficits affecting the production of appropriate postural adjustments during upright standing (Holt, Simpson, Jenner, Kirker, & Wing, 2000; Kirker, Jenner, Simpson, & Wing, 2000). For those individuals who achieve an upright standing posture, prominent impairments increased postural sway (Corriveau, Hebert, Raiche, & Prince, 2004; Titianova & Tarkka, 1995) and an asymmetrical distribution of body weight in the frontal plane with the Center-of-Mass (CoM) kept mostly above the nonparetic foot (Bohannon & Larkin, 1985; Leonard, 1990). With IPT provided, the stroke patients benefitted from a single contact at waist or shoulder level in the AP as well as the ML directions, whereas the patients with Parkinson's disease benefitted in the AP direction only. This observation might be linked to an aspect specific to the Parkinson's patients' control of sway in the ML direction with closed eyes (Blaszczyk et al., 2007) and may indicate less reliable use of the IPT signal for ML sway control in Parkinson's patients. As contact was applied onto clothing, it is possible that a single contact did not provide sufficient tactile information to the patients with Parkinson's disease.

Our study is the first to show quantitative reductions in sway in patients with Parkinson's disease as well as hemiparetic stroke during passively received deliberately light IPT. Reductions in sway with active light touch have been demonstrated in patient populations with impaired balance control caused by peripheral sensory loss in the visual, vestibular, or somatosensory modality (Baccini et al., 2007; Dickstein, Shupert, & Horak, 2001; Jeka, Easton, Bentzen, & Lackner, 1996; Lackner et al., 1999). Cunha, Alouche, Araujo and Freitas (2012) reported that individuals who suffered a stroke can also use fingertip light contact to reduce body sway, a finding replicated by Rabin, Chen, Muratori, francisco-Donoghue and Werner (2013) in patients with Parkinson's disease. Consequently, Baldan, Alouche, Araujo and Freitas (2014) suggested that the maintenance of the fingertip lightly touching an external reference augments somatosensory information for the individuals with poor balance and thus could be used as a strategy to improve balance control during intervention programs. Unfortunately, the potential of light touch to reduce long-term fall risk in patients with Parkinson's disease as well as patients with stroke has not been assessed to date. In this study, we did not attempt to quantify the intensity of the contacting force. This is a methodological short-coming but which should not have any consequences to the actual application of IPT in a clinical context. Any measurement apparatus between the contact provider and receiver will diminish the contact provider's haptic sensitivity and ability to quickly adjust the contacting force to the sway excursions of the receiver. Lacking immediate force feedback, the exact amount of the contacting force has no relevance to the HCP. More important is the deliberation of the contact provider to keep the contact "light" and not to accept any transfer of the receiver's body weight onto the extended limb. Similarly, the provider would not apply pressure onto the receiver's body, as this would likely perturb the receiver's body balance. The contact providers in this study were practiced in the application of IPT by adjusting the contacting force and the position of contacting hand according to a participant's trunk swaying motion. They were not actively damping a participant's sway, as this would have created a tight mechanical coupling between both individuals. To train HCPs unfamiliar with deliberately light IPT, however, quantification of the contacting force is reasonable to allow contact

Key Practice Points

- Deliberately light interpersonal touch (IPT) without mechanical weight transfer to the provider reduces postural sway in receivers with a neurological condition.
- It is recommended to apply more superior and multiple contact points to increase the benefit.
- Applying deliberately light IPT in a secure environment might not only provide assurance to balance impaired individuals but also resemble a stimulus that facilitates sensorimotor processes for balance control.
- Contact providers are required to attend to cutaneous feedback from their fingertips to adapt the contacting force to receivers' body sway to avoid weight transfer and involuntary postural perturbations.

providers to experience the requested amount of the contacting force at about 1 N.

Our study provides "proof of concept" for using deliberately light IPT to improve standing balance in populations with balance impairments due to neurological disorders. Key aspects of IPT are IPT that remains light and does not restrict a patient's own movement and postural degrees of freedom. IPT, however, not only reassures a patient but also facilitates optimization of postural control based on internal sensory afferences. An HCP should adjust the provision of IPT to the capability of a patient. Therefore, IPT has implications for balance rehabilitation not only with respect to a situation-specific reduction in fall risk but also in the sense that a patient does not receive more physical balance support than is actually needed. Thus, IPT is best suited for a controlled situation and environment such as in a balance therapy session where the utilization of touch feedback provided by the therapist can be emphasized and the adverse consequences of a fall can be minimized.

Our study did not evaluate changes in patients' general, long-term fall risk when deliberately light IPT is exercised in inpatient rehabilitation settings regularly. Current nursing concepts such as "Kinaesthetics," however, promote a light touch approach during manual patient handling (Hatch & Maietta, 2003) without a priori investigation of specific handling techniques. In addition, published nursing guidelines that recommend specific manual patient handling techniques suggest that an HCP supports a balance-impaired patient at the shoulder, the arm, or the waist to reduce fall risk and to offer reassurance. It is argued that their techniques are biomechanically safe and do not require taking a patient's weight, but empirical evidence is not provided (ACC, 2003; Potter & Perry, 2005). We believe, therefore, that our study is a first step toward evidence-based manual handling techniques and recommend that future studies evaluate whether deliberately light IPT applied in clinical routine and rehabilitation results in long-lasting improvements in balance ability and reduced fall risk in patients with neurological disorders. In addition, future studies ought to investigate the critical factors of IPT provided to a balance-impaired patient during more complex, dynamic postural activities such as walking.

To summarize, our study demonstrates that both patients with Parkinson's disease and with chronic hemiparetic stroke benefitted from deliberately light IPT provided by an HCP. We suggest that a contact location on the back at shoulder level is a preferable location for facilitating a patient's balance control based on augmented sensory processing of own body sway. Additional benefits are provided by a second simultaneous contact location. Future studies need to evaluate the potential of regularly applied deliberately light IPT for reducing long-term fall risk in patients with neurological conditions.

Acknowledgments

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Authors' Contribution

All authors declare that they participated in the study design, the data collection and analysis, and the preparation of the submitted manuscript and that they all have seen and approved the final version.

Competing Interest

The authors have no conflicts of interest to disclose.

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LETTER TO THE EDITORS

Body sway during quiet standing post-stroke: effects of individual and interpersonal light touch

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Dear Sirs,

Individuals with hemiparetic stroke have an elevated falls risk [1, 2]. In clinical practice, therefore, evidence-based strategies for the augmentation of sensorimotor control of posture are required that facilitate patients' performance without the provision of too much mechanical body weight support. Lightly touching an earth-fixed external reference point generally improves stability of posture [3]. In stroke patients, however, the effect of light touch (LT) on balance have been demonstrated [4] but are less certain. For example, Boonsinsukh et al. [5] showed that LT provided by a cane stabilizes mediolateral trunk sway during walking, while Ijmker et al. [6] could not find any evidence for optimized walking with LT of a handrail. It seems, therefore, that the utilization of LT is not as straight forward in stroke as it might be in other balance-impaired populations.

Another strategy observed in daily life is light touch provided by a caregiver. In older adults, light collaborative ("active") fingertip-to-fingertip interpersonal touch (IPT) results in sway reductions in quiet standing [7]. Deliberately light IPT received to the back ("passive") also reduces sway in stroke patients [8]. In the present study, the effects of LT as well IPT were contrasted between older adults with and without chronic hemiparetic stroke.

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Seven chronic hemiparetic stroke patients [6 female, 1 male; age: 61–69 years; time since lesion: >1 year; 5 ischemic, 2 hemorrhagic; lesioned hemisphere: 4 left, 3 right; paresis range: 3–5 (arm), 2.5–4 (leg); Berg-Balance-Scale: 44–51; Rivermead Mobility Index: 7–11; Modified Rankin Scale: 2–3] and 11 healthy older adults (4 female, 7 male; age range: 63–77 years; Berg-Balance-Scale: 50–56) were recruited from the community. All participants were right-handed and able to stand unsupported. Individuals with other neurological pathology, orthopaedic or rheumatic conditions or who were unable to follow verbal instructions were not included.

Participants stood with open eyes in comfortable, normal bipedal quiet stance on a force plate (Bertec 4060FP; 200 Hz; normal footwear) and performed 4 blocks of 10 stance trials (duration: 20 s) in random order: no contact (NC), fingertip LT (fLT), active fingertip IPT (aIPT) and passive elbow IPT (pIPT). During all trials, one contact provider stood perpendicular to the participant on the side of the dominant arm (unaffected arm in stroke) to ensure participants' safety and to apply continuous IPT when instructed. Participants held their arm in a default elbow-flexed posture enabling the tip of the extended index finger to contact a height-adjustable stand positioned in front. Sway data were low-pass filtered (4th order dual-pass Butterworth with 10 Hz cut-off) and differentiated to express body sway in the anteroposterior (AP) and mediolateral (ML) directions as the standard deviation of Centre-of-Pressure rate of change. Mixed multifactorial ANOVAs with contact conditions as within-subject factor and group as between-subject factor were calculated. An α level of $p<0.05$ was used after Greenhouse-Geisser correction.

Body sway varied in response to the contact condition in both groups (Fig. 1). In the healthy controls, sway was reduced compared to the control condition in both directions of sway (both $p \leq 0.02$; fLT: AP -35%, ML -22%; aIPT: AP -11%, ML -11%; pIPT: AP -6%, ML -12%).

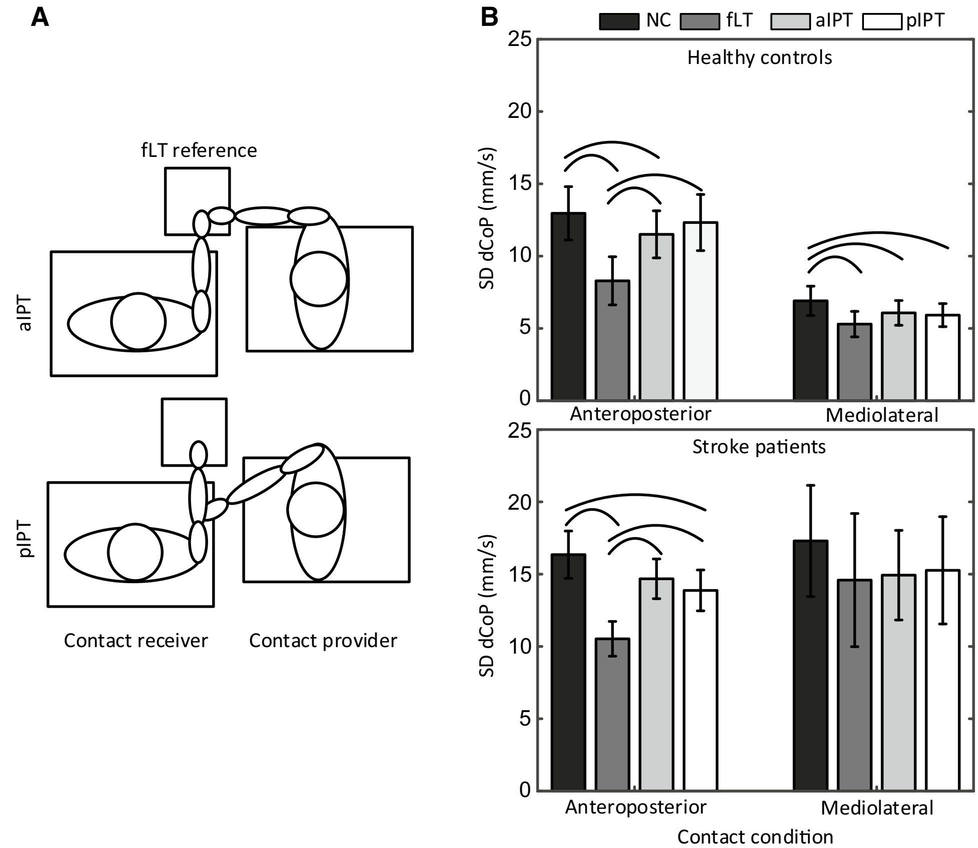

Fig. 1 **a** Interpersonal stance configuration for the light collaborative, "active" fingertip-to-fingertip interpersonal touch (aIPT; upper panel) and the "passive" elbow interpersonal touch (pIPT; lower panel) conditions. **b** Bar plots of the variability of Centre-of-Pressure rate of change (SD dCoP) for both groups of participants in both directions as a function of the light touch contact condition. Horizontal arcs indicate significant post hoc single comparisons ($p < 0.05$). *NC* no contact, *fLT* fingertip light touch to stand reference; error bars show the standard error of the mean across participants

This occurred for the stroke patients in the AP direction only ($p = 0.02$; fLT: AP −32%; aIPT: AP −8%; pIPT: AP −15%).

Our results showed that in the AP direction mildly impaired, chronic hemiparetic stroke patients possess similar responsiveness to LT and IPT in terms of proportional sway reductions comparable to the control participants and previous reports in older adults [7]. Although quite capable, our sample of stroke patients still showed relative instability despite the availability of touch in the ML direction, which indicates a limitation. Interestingly, we found no difference between the two IPT conditions in both groups, which contrasts with recent findings for balance exercises in older adults, where collaborative IPT was more effective [9]. Paresis of the proximal segments of leg and the hip could have interfered with improved postural stability in the frontal plane [10, 11] in some of our stroke participants. In general, our study indicates that the effects of light touch are robust but cannot be generalized from healthy older adults to hemiparetic stroke patients without consideration of moderating functional constraints of the individual and the specific postural context, such as postural degrees-of-freedom and positioning of the contact relative to the individual [12]. Despite the positive responsiveness to light haptic augmentation in our stroke patients, it is known that some severely impaired stroke patients, e.g., showing contraversive pushing behaviour, do not utilize haptic feedback and resist passive interpersonal support [13]. Nevertheless, patients and clinicians alike should be encouraged to apply light touch balance support strategies that were safely possible for the augmentation of mechanically unsupported postural control.

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Compliance with ethical standards

Conflicts of interest The authors declare that they have no conflict of interest.

Ethical standards The study accorded to the ethical principles laid down in the 1964 Declaration of Helsinki and its later amendments and was approved by the University of Birmingham Research Ethics Committee. All participants gave their informed consent prior to their inclusion in the study.

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Consolidation of the postural set during voluntary intermittent light finger contact as a function of hand dominance

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Abstract— **Light fingertip contact with an earth-fixed referent decreases body sway. In a previous study Johannsen et al. (2014) demonstrated longer return-to-baseline of body sway for intermittent contacts of more than 2 seconds duration. This indicates that sway reduction with light tactile contact involves postural control strategies independent of the availability of tactile feedback and may depend on the intention to control body sway with light touch feedback. In the present study, we investigated the effect of hand dominance on post-contact return-to-baseline to probe for potential inter-hemispheric differences in the utilization of light finger contact for sway control. Twelve healthy, right-handed young adults stood in normal bipedal stance with eyes closed on a force plate with an earth-fixed referent directly in front. Acoustic signals instructed onset and removal of intermittent light touch. We found that return-to-baseline of sway following longer contact durations is affected by hand dominance with the dominant hand resulting in a slower return to No-contact levels of sway. Our results indicate that the light touch postural set is more persistent and might need longer to disengage when established with the dominant hand or takes longer to consolidate when established with the non-dominant hand.**

I. Introduction

In daily life, we often establish intermittent haptic contact with objects in our environment to orientate ourselves and to yield stability of body balance. For example, walking down the aisle on a moving train carriage, we move from handhold to handhold prepared to counter any unexpected perturbations. Or when we cross an unlighted room, we haptically move from contact to contact to gain an estimate of our position and to augment our sense of spatial orientation.

Light fingertip contact with an earth-fixed reference leads to a reduction in body sway [1]. Only a few studies have addressed the time course of sway before and after a contact transition [2, 3, 4]. Sway stabilization with light touch is a time-consuming integrative and attention demanding process [2, 3, 5].

In terms of a multimodal sensory strategy, it seems rather costly if the postural control system switches between different multisensory sets each time intermittent contact is established or removed [6]. Instead, while anticipating upcoming contact intervals and thus the imminent availability of reliable haptic feedback, keeping a multisensory set including the haptic channel temporarily active might offer an advantage with respect to the costs of switching the postural sets [7]. For example, Bove and colleagues (2006) demonstrated that the intention to establish contact within less than 5 seconds leads to reductions in body sway before contact is established. Schiepatti and colleagues [8] proposed that transient anticipatory processes are involved in the preparation of the central postural set to the context of stance control with light contact. Investigating intermittent touch with only short contact durations, Johannsen et al. (2014) demonstrated that contact durations of more than 2 s result in slower recovery of reduced sway to baseline levels after contact removal. These observations indicate that the integration of fingertip contact requires no less than about 2 seconds and is likely to involve not only bottom-up sensory processing but also top-down, “intentional” control of body sway and tactile attention.

The two hemispheres of the human brain might play different roles in the control of body sway with and without light touch [9, 10]. In the present study we not only aimed to replicate previous findings with intermittent but longer contact durations, we also intended to probe for differences between the dominant and non-dominant hemispheres regarding their influence on switching the postural set in right-handed participants during phases of intermittent light touch.

II. Methods

Participants
Twelve healthy young adults (mean age = 25.8, SD = 2.6; 7 woman and 5 men) were recruited for the current study. Inclusion criteria were (1) right hand dominance and (2) no balance impairment. All participants were informed about the study protocol and signed a written informed consent was provided. The study was approved by the Clinical Research Ethics committee of the Technical University of Munich.

Procedure
Participants stood barefoot in normal bipedal stance. After the height of the stand was adjusted to each participant’s waist level, participants were asked to hold their index finger of the dominant hand above a touch plate while keeping the outstretched arm in a comfortable posture. We instructed participants to close their eyes, and to stand relaxed but as still as possible without speaking.

Trials were started when participants indicated that they were ready. On hearing a high-pitched tone, participants flexed their index finger at the metacarpal-phalangeal joint to initiate light finger contact. On a low–pitched tone, participants lifted their index finger just above the touch plate. Before testing participants could practice the task in order to

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familiarize themselves with the experimental protocol. Afterwards they performed at least 6 trials with 30 s break in between hands.

After participants finished sway testing, we assessed the tactile discrimination threshold of each hand's index fingertip using 13 orientation gratings with a gap width ranging from 0.35 mm to 5.50 mm [11]. Participants had to judge whether gratings were aligned straight or orthogonal with the fingertip. Gratings were applied manually for about two seconds. Testing protocol consisted of a staircase procedure which ended either after ten successful reversals or a total of 50 grating presentations. The final tactile acuity threshold was derived from the average of the last 10 presentations.

Apparatus

A force plate (600 Hz; Bertec FP4060-10, USA) measured the six components of the ground reaction forces and moments to determine the antero-posterior (COP_{ap}) and medio-lateral (COP_{ml}) components of Centre-of-Pressure. In response to a high-pitched or low-pitched auditory cue, participants either made or withdrew fingertip contact with a touch plate (3 cm diameter), mounted on a stand at waist level to the front of the participants. A force-torque transducer (ATI Nano17, USA) measured the normal and horizontal shear forces applied to the touch plate with a rate of 200 Hz. We measured body kinematics (60 Hz; Zebris, Germany) in terms of trunk motion with three acoustic markers placed at wrist, shoulder and hip.

Each balance testing consisted of 2 blocks of at least 6 trials per hand (range=6 to 8 trials; blocked, randomized order: dominant hand, non-dominant hand). Every balance trial contained four auditorily triggered active transitions between No-touch and Touch ("onset") and Touch and No-touch ("removal"). The acoustically cued intermittent active contact durations were 1 s, 1.5 s, 10 s and 20 s in randomized order. Every No-contact interval was at least 10 seconds long. Onset and removal time points were randomized resulting in total trial durations of at least 130 s.

Data reduction and statistical analysis

All data were interpolated to 600 Hz and merged before low-pass filtering with a fourth-order Butterworth filter (10 Hz cut-off frequency) and differentiated to yield rate of change. According to the vertical touch force as detected by the force-torque sensor, onset and removal time points of each touch period were determined. For comparisons between contact durations participants' *actual* contact durations were sorted into the following categories: T1 (0.8 s – 1.6 s), T2 (2.0 s – 2.6 s), T10 (8.0 s – 13.0 s) and T20 (18.0 s – 22.0 s). Trial segments with other contact durations were discarded. Subsequently, the T1 and T2 categories were averaged and subsumed under "short" duration conditions, while T10 and T20 were averaged and combined as "long" contact durations for statistical analysis.

Non-discarded trial segments were divided into bins of 500 ms duration from 5 s before to 5 s after a contact transition. Sway within each bin was quantified in terms of the standard deviation (SD) of the Centre-of-Pressure velocity in the anterior-posterior ($dCOP_{ap}$) direction. Sway parameters were averaged for each duration condition of all trials a participant performed.

Using SPSS 18.0 software (Chicago, IL, USA), repeated-measures ANOVAs were performed with time course across a range of 500 ms bins, contact duration and contacting hand as within-subject factors.

In order to characterise the return of sway to the No-contact baseline following contact removal, we fitted linear regressions across three time bins: 0.5 s before removal, 0.5 s and 1 s after removal. Statistical analysis of regression slope and zero-offset was conducted with repeated-measures ANOVAs with contact duration and contacting hand as within-subject factors. Level of significance was set to p=.05 after Greenhouse-Geisser correction. Effects with estimated effects sizes of partial $\eta^2>0.14$ were considered large.

III. Results

Statistical analysis of the tactile discrimination thresholds revealed no significant differences between the dominant and non-dominant hands ($p = 0.33$), which suggests that hand dominance did not influence tactile sensitivity of the respective hand. Figure 1 shows the tactile sensitivity thresholds for the index finger of both hands.

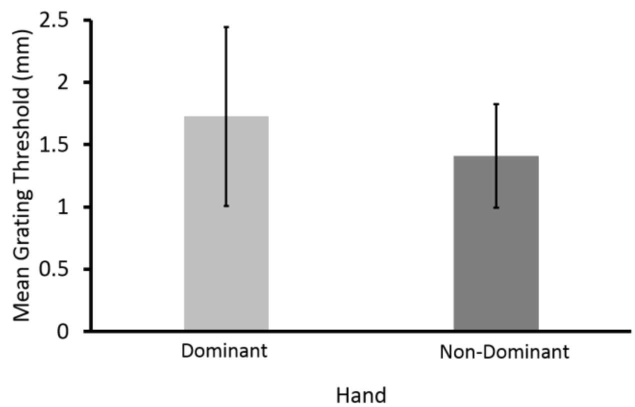

Figure 1. Tactile sensitivity threshold in terms of the just noticeable gap width for the dominant (light grey) and non-dominant (dark grey) hand. Error bars indicate standard error of the mean.

Figure 2 shows average sway progression from 5 s before to 5 s after contact onset and Figure 3 shows average sway progression around contact removal for short (upper panel) and long (lower panel) contact durations. Sway is oscillating close to the No-contact baseline before contact is established. After the onset of touch, sway transiently rises above and then begins to drop below the baseline. Similarly, sway with light touch is noticeably below the baseline before contact is removed. Following contact removal, sway once again overshoots the No-contact baseline and then settles towards it.

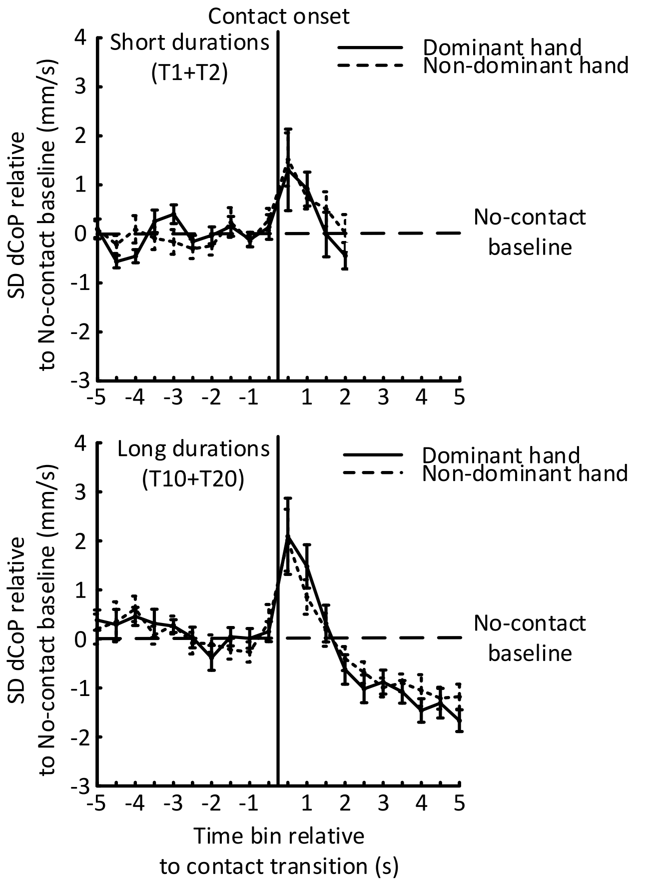

Figure 2. Average time course of sway across 500 ms bins from 5 s before to 5 s after contact onset for the short durations (upper panel) and long durations (lower panel) for the dominant (bold line) and non-dominant hand (dashed line). Error bars indicate standard error of the mean.

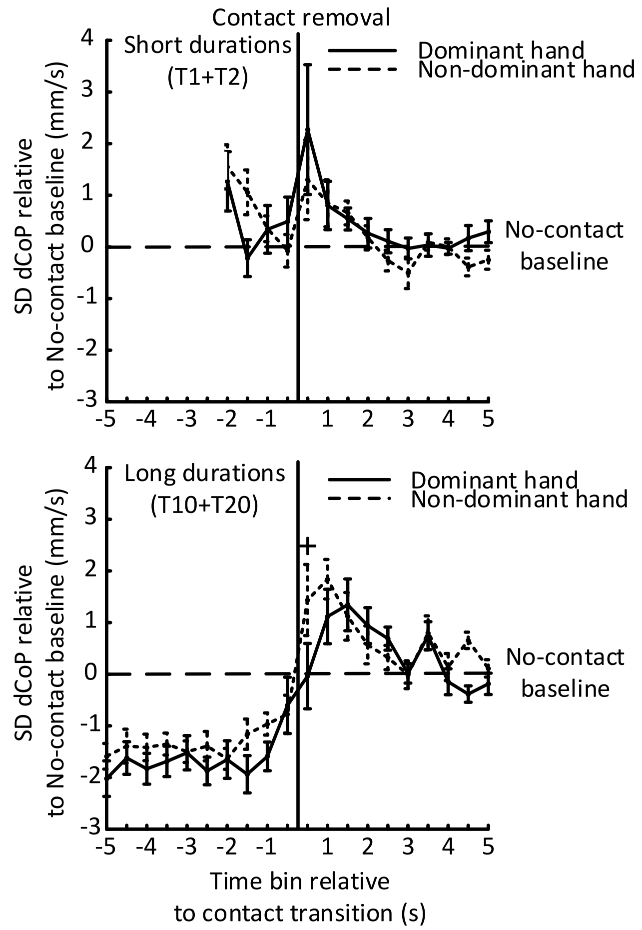

Figure 3. Average time course of sway across 500 ms bins from 5 s before to 5 s after contact removal for the short durations (upper panel) and long durations (lower panel) for the dominant (bold line) and non-dominant hand (dashed line). Error bars indicate standard error of the mean. The cross indicated the tendency of a difference between both hands (p>.1).

Although steady-state sway with light touch of the dominant hand (time bins from 5 s to .5s before contact removal) appears lower compared to the non-dominant hand, the two contact conditions were statistically not different ($p>.25$, partial $\eta^2=.12$).

The increase in sway after removal of long duration light touch appears less rapid with the dominant hand compared to the non-dominant hand. In order to assess the return-to-baseline of sway after contact removal (including the overshoot), we examined the time course of sway during the removal transitions. Focussing on the range from 0.5 seconds before to 1.5 seconds after. We found statistical significant interactions of between hand and contact duration ($F(1,11)=6.83$, $p=.02$, partial $\eta^2=.38$) as well as between hand, contact duration and time course ($F(3,33)= 4.18$, $p=.03$, partial $\eta^2=.28$). Post-hoc single comparisons showed a strong difference between the dominant and non-dominant hand at the 0.5 s time bin after long duration contact removal ($F(1,11)=3.47$, $p=.08$, partial $\eta^2=.24$) with lower sway after contact removal of dominant hand.

Sway overshoot after removal of the non-dominant hand had progressed further during this period, almost reaching peak overshoot, compared to the dominant hand. Peak overshoot, although numerically lower following contact with the dominant hand, was not affected by limb dominance (….).

Analysis of the linear regression parameters showed significant interactions between contact durations and hand for the regression slope ($F(1,11)=6.89$, $p=.02$, partial $\eta^2=.39$) and offset ($F(1,11)=6.70$, $p=.03$, partial $\eta^2=.38$). For both slope and offset after short duration contact, post-hoc single comparisons did not show differences between hands. After long duration contact, however, previous contact with the dominant hand resulted in a lower slope ($F(1,11)=5.55$, $p=.04$, partial $\eta^2=.34$) and offset ($F(1,11)=4.81$, $p=.05$, partial $\eta^2=.30$) compared to the non-dominant hand. Figure 4 shows linear regression slope and offset of the sway progression following contact removal for short and long contact durations as a function of the hand tested.

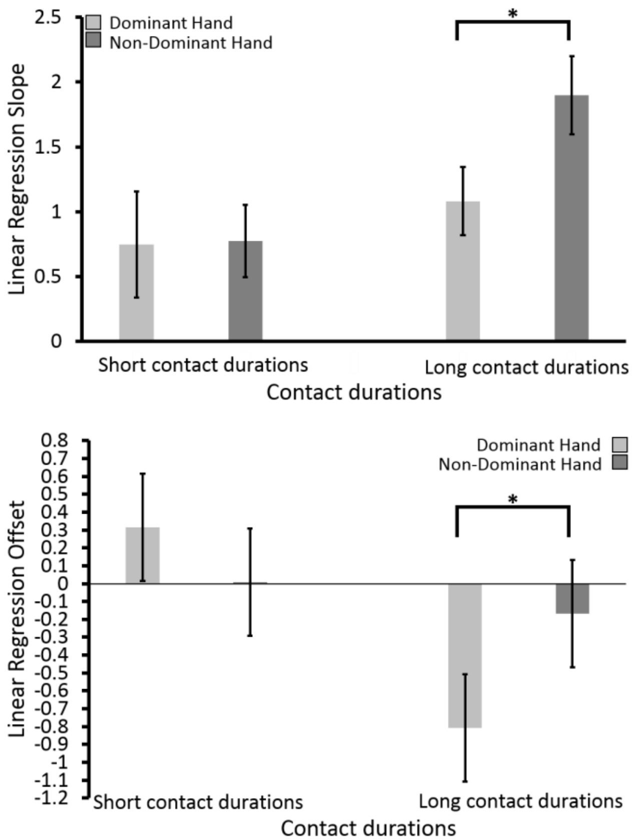

Figure 4. Linear regression slope (upper panel) and offset (lower panel) for short and long contact durations for the dominant (light grey bars) and non-dominant (dark grey bars) hand. Error bars indicate standard error of the mean. An asterisk indicates a significant comparison between hands ($p<0.05$).

IV. Discussion

Actively removing intermittent light touch at the fingertip leads to a rapid increase in sway within 500 ms after contact removal for contact durations shorter than 2.5 seconds irrespective of the contacting hand. Similarly, contact at the fingertip of the non-dominant hand also shows rapid increase for longer durations. In contrast, more persistent contact with the dominant hand results in delayed sway return-to-baseline.

In our present study, the general progression of sway during a contact removal transition is in line with the previous study of Johannsen et al. [4]. They showed that short contact durations initiate a reduction in sway but do not result in a significant reduction. A delayed return-to-baseline only occurred for contact durations longer than 2 seconds. Contact durations longer than 5 seconds, however, were not tested. Therefore, our present study tested longer contact durations, which ought to more likely result in steady-state sway with light contact. Indeed, we found that the sway progression after touch removal increased at a lower rate but only when longer duration touch was established with the dominant hand. With the non-dominant hand, contact resulted in a rapid sway increase similar to the short contact durations.

A central question to be answered is whether the less rapid, more gradual return of sway to No-contact levels after removal of the dominant hand resembles a functional advantage or disadvantage? It could be that a rapid return expresses a fast readjustment in the multisensory strategy of the postural control system. The instantiation of a new postural set involving the haptic channel could result in inter-sensory conflict between an information-deprived haptic channel and the other senses. The sway overshoot observed could be a consequence of the sudden deprivation of a highly weighted tactile signal leading to acute intermodal conflict. For example, following abrupt cessation of long-term support surface sway referencing, Peterka and Loughlin demonstrated the emergence of transient, involuntary 1 Hz body oscillations, possibly due to over-corrective torque production [12].

It seems more reasonable to delay postural set switching until the likelihood is high that the haptic channel will provide reliable feedback for an extended period. Once such a steady state has been reached it also seems reasonable to keep this set active and delay disengagement, if further contact periods are expected to occur in the near future. This reasoning seems to apply to the pattern we observed for the dominant hand. As we tested right-handed participants it implies that the dominant left hemisphere is involved in this strategy. In a previous study, we observed that disruption of the left-hemisphere inferior parietal gyrus (IPG) by repetitive transcranial magnetic stimulation (rTMS) inhibited sway overshoot following unexpected, passive removal of light contact [4]. This could mean that the left IPG plays a role in the detection of multisensory conflict or the directing of tactile attention. This is in correspondence with reports by Ishigaki and colleagues [13], who suspected involvement of the left primary somatosensory and posterior parietal cortices in the processing and integration of steady-state right hand light touch. On the other hand, we disrupted the left and right PPC by cTBS and did not find any alterations in sway progression following removal of active light touch [10]. Nevertheless, all-in-all the evidence suggests that the left-hemisphere plays some role in the control of body sway with light haptic feedback form the contralateral, right hand, for example in the consolidation of an adequate central postural set.

Why did the non-dominant, left hand not demonstrate a delayed return-to-baseline similar to the dominant, right hand? One possibility is that consolidation of the central postural set for the light touch with the non-dominant hand has a longer time constant. For example, our participants might have been more used to explore the environment with their dominant hand.

An aftereffect on postural sway following an extended duration of lightly gripping a cane was reported by Oshita and Yano [14]. They investigated the effect of lightly touching a cane on postural sway and ankle- joint muscle activity. They found decreased sway and decreased co-contraction of the ankle joint muscles when the cane was gripped lightly. These reductions were also present after lifting off the cane from the ground. In interestingly, their participants used the left hand to grip the cane, presumably the non-dominant hand. Oshita and Yano did not assess varying contact durations but 30 s contact only. It seems that also light contact with the non-dominant hand can lead to slow return-to-baseline of body sway. Perhaps contact durations of more than 20 s duration are the prerequisite.

To conclude, the occurrence of a delayed return-to-baseline of sway following removal of fingertip light touch is affected by hemispheric lateralization. While the dominant hand showed a delayed return-to-baseline effect after long contact durations, it was not observed when the non-dominant hand was used for contact. This difference cannot be explained by differences in the tactile sensitivity of the contacting index fingers of the two hands. Instead, the effect could rely on a difference in the rate of consolidation of a light touch postural set, with faster consolidation when tactile feedback is processed in the dominant hemisphere.

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LETTER TO THE EDITORS

Serving performance in a suprapostural visual signal detection task: context-dependent and direction-specific control of body sway with fingertip light touch

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Dear Sirs,

When upright stance body sway is increased during horizontal oscillatory smooth pursuit, it may indicate interference between oculomotor and sway control, potentially due to an efferent oculomotor signal [1]. In specific contexts, however, body sway reduction has also been reported during smooth pursuit [2]. Riccio and Stoffregen [3] argued that the postural control system also takes into account an individual's behavioural goals, such as performance in a "suprapostural" task, especially when the task imposes visual demands in contrast to cognitive demands [4]. Therefore, sway may be dampened proactively to reduce self-imposed variability and to improve oculomotor accuracy during visual tracking or reduce retinal slip in a visual discrimination task [2, 5, 6]. Similarly, precision control of fingertip light touch (LT) with an earth-fixed reference, which most reliably reduces body sway [7], has been considered a suprapostural task [8]. The interpretation of proactive sway control assisting fingertip LT is corroborated by observations that body sway may be reduced for intermittent periods when LT is absent, but nevertheless relevant to the postural context [9–11]. Is a natural sensorimotor congruency always required to elicit task-related sway adaptation or does it generalize to more complex sensorimotor stimulus–response mappings? Our present study adopted a "biofeedback" approach, in which the perceptual difficulty in a visual signal detection task (VSDT) was coupled (implicit feedback coupling, IFC) to either body sway directly or to the contact force during fingertip light touch. In both situations, we expected that body sway would be reduced proactively to ease the difficulty of the VSDT.

Ten healthy right-handed young adults (4 females, 6 males; age = 26.7 yrs, SD 6.0) faced a flat-screen display (Samsung UE40D6500) in tandem stance. A force plate (600 Hz; Bertec FP4060-10) recorded body sway in terms of centre-of-pressure (CoP) fluctuations. A single Landolt-C was presented as the VSDT target, randomly changing the direction of its opening every 2 s while continuously oscillating horizontally (0.09 Hz) across the entire width of the display. Participants were instructed to press a response button with their non-dominant hand as fast as possible when the opening of the Landolt-C pointed upwards. The dominant arm was held in a default elbow-flexed posture, enabling the extended index fingertip to contact a force–torque transducer (200 Hz; ATI Nano17) on a height-adjustable stand positioned in front. VSDT perceptual difficulty varied in terms of the amplitude of random vertical target jitter. Body sway was assessed in four IFC conditions: (1) LT with independent jitter (LT-IJ), (2) LT with jitter depending on LT contact force (LT-CF), (3) LT with jitter depending on body sway (LT-BS), and (4) no contact with jitter depending on body sway (NT-BS). IFC conditions were tested in randomly ordered blocks of five trials (120 s duration). Further details of the experimental setup are provided in the online methods supplements (Figs. 2 and 3). CoP was low-pass filtered (4^{th}-order dual-pass Butterworth with 10 Hz cut-off) and differentiated to express body sway as the standard deviation of CoP velocity (dCoP). Repeated-measures

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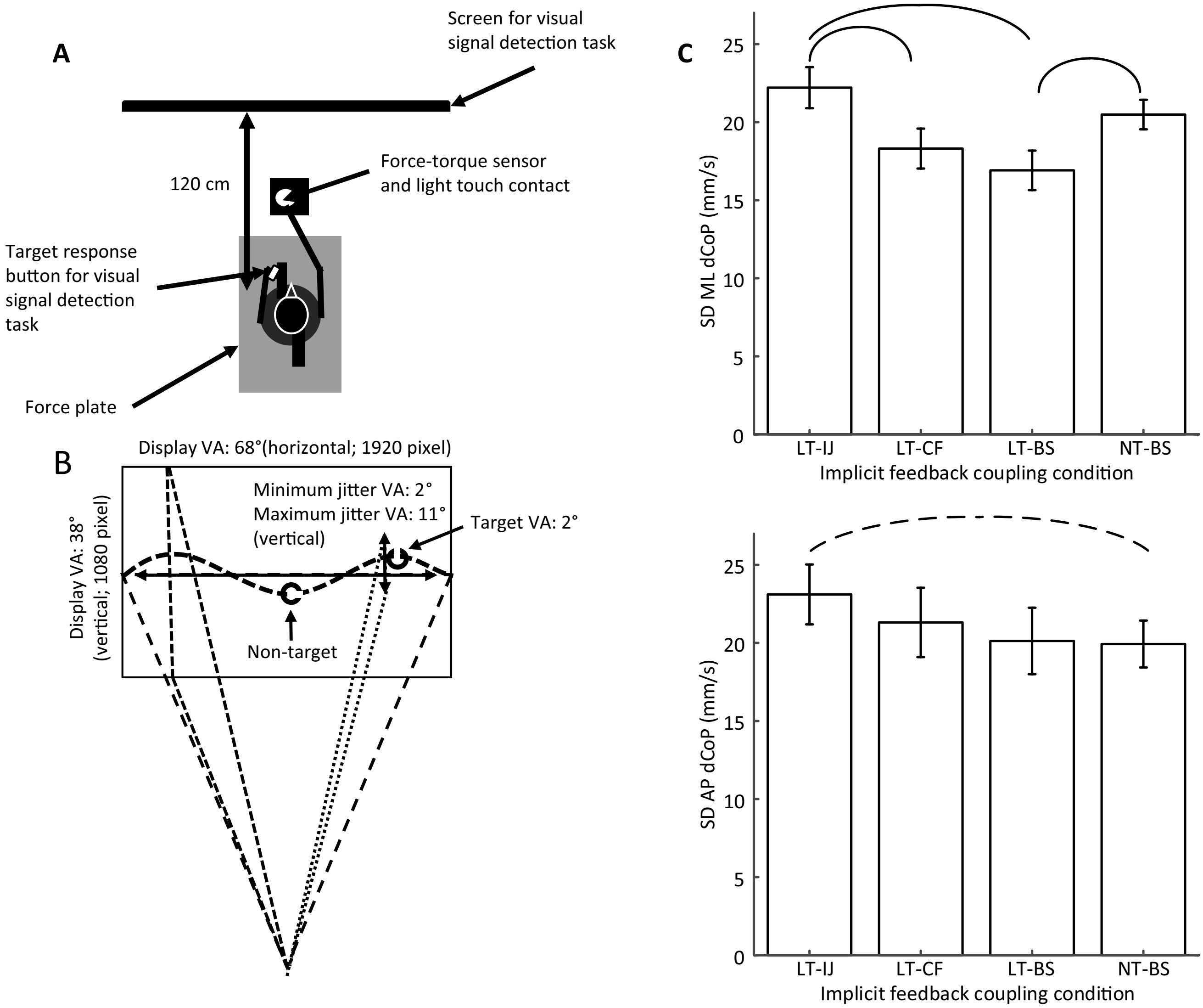

Fig. 1 **a** The experimental setup showing an individual in tandem stance on a force plate in front of the display screen with fingertip light touch of the dominant hand and a response button in the non-dominant hand. **b** Schematic of the stimulus display. A Landolt-C oscillated horizontally along a double sine-wave trajectory across the entire width of the display at a constant velocity of approximately 14°/s changing the direction of its opening every 2 s. Participants had to gaze track the target to press the response button when its opening pointed upwards. Random jitter of variable amplitude in the vertical direction disrupted the visibility of the Landolt-C opening, thereby affecting the difficulty of the visual signal detection task. Current jitter amplitude depended on the current fingertip contact force or current body sway. *VA* visual angle. **c** Variability of mediolateral (ML; upper panel) and anteroposterior (AP; lower panel) body sway velocity (SD dCoP) in each implicit feedback condition (IFC). LT-IJ: fingertip light touch with independent maximum jitter amplitude; LT-CF: jitter amplitude dependent on light touch fingertip contact force; LT-BS: jitter amplitude dependent on body sway with additional fingertip light touch; NT-BS: jitter amplitude dependent on body sway without additional fingertip light touch. Error bars indicate the standard error of the mean. Straight horizontal arcs indicate significant post hoc single comparisons ($p<0.05$), and he dotted horizontal arc indicates a statistical tendency ($p<0.10$)

ANOVA was calculated with IFC condition as within-subject factor. An alpha level of $p<0.05$ was used after Greenhouse–Geisser correction. Post hoc single comparisons were Bonferroni-adjusted.

The proportion of hits in the VSDT task was 67% in LT-IJ, 80% in LT-CF, 77% in LT-BS, and 59% in NT-BS. Average LT force was 0.85 N (SD 0.17) with no difference between the IFC conditions with LT. Resulting body sway differed between the IFC conditions ($F(3,27)=12.74$, $p<0.001$; Fig. 1). Reduced mediolateral sway was found in both LT-CF and LT-BS compared to LT-IJ (both $p \leq 0.007$) and in LT-BS compared to NT-BS ($p=0.003$). No difference between the IFC conditions was observed for anteroposterior sway ($p=0.12$). Nevertheless, there was a tendency for a difference between LT-BS and LT-IJ ($p = 0.09$).

Our results demonstrate a direction-specific reduction in mediolateral body sway below a level achieved by LT sway-related feedback augmentation alone if an implicit feedback coupling is present. Similar direction-specificity of sway control has been reported in visuomanual aiming [12]. In visual search involving saccadic eye movements instead of smooth pursuit, Chen et al. [13] showed that LT improved search performance. Demands of the visual search task, however, reduced sway independent of LT availability so that two processes seemed to act in parallel [13]. Similarly, in our current study, both direct (LT-CF) and indirect (LT-BS) involvement of fingertip contact in an IFC condition minimized sway, which implies either that no control hierarchy existed for whole body sway and fingertip contact (integration of both control processes) or that the hierarchy can be reversed flexibly (one facilitating the other) if it serves the implicit goal of reduced perceptual noise and enhanced performance within the context of our suprapostural VSDT.

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Compliance with ethical standards

Conflicts of interest The authors declare that they have no conflict of interest.

Ethical standards The study accorded to the ethical principles laid down in the 1964 Declaration of Helsinki and its later amendments and was approved by the Technical University of Munich Ethics Committee. All participants gave their informed consent prior to their inclusion in the study.

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Disruption of right posterior parietal cortex by continuous Theta Burst Stimulation alters the control of body balance in quiet stance

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Abstract

Control of body balance relies on the integration of multiple sensory modalities. Lightly touching an earth-fixed reference augments the control of body sway. We aimed to advance the understanding of cortical integration of an afferent signal from light fingertip contact (LT) for the stabilisation of standing body balance. Assuming that right-hemisphere Posterior Parietal Cortex (rPPC) is involved in the integration and processing of touch for postural control, we expected that disrupting rPPC would attenuate any effects of light touch. Eleven healthy right-handed young adults received continuous Theta Burst Stimulation over the left- and right-hemisphere PPC with sham stimulation as an additional control. Before and after stimulation, sway of the blindfolded participants was assessed in Tandem-Romberg stance with and without haptic contact. We analysed sway in terms of the variability of Centre-of-Pressure (CoP) rate of change as well as Detrended Fluctuation Analysis of CoP position. Light touch decreased sway variability in both directions but showed direction-specific changes in its dynamic complexity: a positive increase in complexity in the mediolateral direction coincided with a reduction in the anteroposterior direction. rPPC disruption affected the control of body sway in two ways: first, it led to an overall decrease in sway variability irrespective of the presence of LT; second, it reduced the complexity of sway with LT at the contralateral, non-dominant hand. We speculate that rPPC is involved in the active exploration of the postural stability state, with utilisation of LT for this purpose if available, by normally inhibiting mechanisms of postural stiffness regulation.

Introduction

Keeping light contact ('light touch', LT) with objects in our environment augments the sensory feedback about the body's relative orientation in space and leads to reductions in body sway (Jeka & Lackner, 1994). In order to integrate haptic information from the fingertips into the postural control loop, the central nervous system (CNS) may require interpretation of a local contact signal within the context of the body's overall proprioceptive state. This includes both arm posture and stance configuration, which could involve transformations of the haptic signal into an egocentric reference frame.

The posterior parietal cortices may be central components of a distributed network of neural circuits for the processing of somatosensory and proprioceptive information in ego-centric frames of reference (Longo *et al.*, 2010; Medina & Coslett, 2010; Bolton, 2015). For example, Azanon *et al.* (2010) showed that disruption of the right posterior parietal cortex (rPPC) impairs conscious position judgements of tactile stimuli on the left forearm relative to the face. With respect to the processing of haptic information for the control of body sway, Franzén *et al.* (2011) suggested that the postural control system has switched from a global to a local trunk-centred reference frame after light touch has been integrated into the postural control loop. Thus, right-hemisphere PPC (rPPC) seems like a good candidate to test for involvement in the processing of a fingertip signal within an egocentric reference frame for the control of body sway.

Light touch of the dominant hand during quiet standing involves processing in the dominant left-hemisphere. Bolton *et al.* (2011) demonstrated that when the somatosensory feedback of the right hand contains sway-related information, brain activity at the left inferior parietal lobe caused by somatosensory-evoked potentials is modified by the specific postural context. In addition, Johannsen *et al.* (2015) investigated repetitive Transcranial Magnetic Stimulation (rTMS) over the left inferior parietal gyrus (IPG) to assess how stimulation affects the progression of sway before and after passive

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onset and removal of right-hand fingertip contact. They found that rTMS over the left IPG reduced overshoot of sway after contact removal, which indicates that this brain area may influence sensory reorganisation for sway control, for example in terms of directed tactile attention (Johannsen *et al.*, 2015). There is evidence, however, that regions exist also in the non-dominant, right hemisphere for the processing of ipsilateral touch in the context of upright stance. Bolton *et al.* (2012) reported that disruption of the right prefrontal cortex alters the processing of right-hand somatosensory-evoked potentials during contact with an earth-fixed reference.

Nevertheless, in the two stimulation studies reviewed above steady-state sway with light touch was not affected, which raises the question if disruption of another region such as the PPC changes the light touch effect during steady-state sway and if the rPPC in particular is contributing to the processing of touch irrespective of the haptically stimulated body side. The aim of this study was therefore to investigate the involvement of cortical processes represented within both posterior parietal cortices in the processing of haptic afferents for the control of balance. Assuming similar asymmetries between the hemispheres in terms of the processing of tactile input within spatial reference frames, as observed with respect to the distribution of spatial attention (Azanon *et al.*, 2010) to the environment, we expected that disruption of the rPPC alters the integration of haptic afferences of both hands for sway control. In contrast, we expected that left-hemisphere PPC (lPPC) disruption would lead to an altered integration of touch of the contralateral hand only.

Methods

Participants

Eleven healthy right-handed young adults (mean age = 25.45, SD 2.73; six women and five men) were recruited for the current study. Inclusion criteria were (i) right-hand dominance, (ii) no neurological or musculoskeletal disorders, (iii) no balance impairment and (iv) no reported cases of epilepsy. All participants were informed about the study protocol and signed a written informed consent. The study was approved by the Clinical Research Ethics committee of the Technical University of Munich.

Procedure

The experimental protocol was divided into three sessions. As a first session prior to the stimulation sessions a high resolution anatomical brain scan, consisting of a T1 MPRAGE (3T whole-body scanner, Signa HDx; GE Healthcare, Milwaukee, WI, USA) was carried out at the University Hospital Großhadern, Center for Sensorimotor Research. The brain scan was used in the following sessions for real-time neuronavigation in order to locate the respective stimulation area.

Each TMS session consisted of a balance pre-test, the application of TMS and a balance post-test. The balance tests required blindfolded participants to stand on a force plate (600 Hz; Bertec FP4060-10, Columbus, OH, USA) in quiet Tandem-Romberg stance, while actively initiating and ceasing finger contact with an earth-fixed referent in response to an acoustic signal. The earth-fixed contact reference point was placed in front of the participants. They held one arm slightly angled in front of the body and reaching straight forward. The other arm remained passive with the hand touching the stomach in order to prevent subjects from using arm movement to correct their body balance. Each balance testing consisted of six trials of at least 130 s (blocked, randomised order: three with the dominant hand, three with the non-dominant hand). Durations of the single trials varied due to the randomisation of the length of the interval between contact events. Tandem-Romberg stance posture was adjusted according to the contacting hand. When the dominant hand contacted the reference point, the leg on the same side took the rear tandem position. When the contacting hand changed, so did the position of the feet. Participants were instructed to stand relaxed and not flex their knees to lock legs in position.

Each balance trial had six auditory triggered active transitions between No-touch and Touch (lowering the finger to the contact; ‘onset’) and Touch and No-touch (raising the finger of the contact; ‘removal’). Every contact phase was at least 8 s long. Time points of contact onset and removal were randomised. We instructed participants to lightly press onto a contact plate downwards with a force around 1N. Before testing began, they practiced light touch in order to get a feeling for the applied force. Participants did not receive feedback about the contact force during a trial to avoid any attentional distractions and to prevent contacting from becoming an explicit precision task.

Body kinematics (4 Oqus 500 infrared cameras; 120 Hz; Qualisys, Göteborg, Sweden) and forces and torques at the reference contact location (6DoF Nano 17 force-torque transducer; 200 Hz; ATI Industrial Automation, Apex, NC, USA) were assessed. To capture body motion, reflective markers were placed at contacting fingertip, wrist, shoulders, C7, Sternum, hip and ankle.

During the TMS we applied continuous Theta Burst Stimulation (cTBS) of an intensity of 80% of the passive motor threshold for 60 s over the rPPC or lPPC (Fig. 1A; PMD70-pCool; MAG & More, Munich, Germany). This protocol is widely used and stimulation effects can last from 20 min up to 1 h (Staines & Bolton, 2013). A staircase procedure was used to determine the passive motor threshold. In order to define the cTBS target areas, we used the MNI coordinates reported in Azanon *et al.* (2010), who stimulated the right-hemisphere human homologue of macaque ventral intraparietal area. We therefore expected that cTBS would disrupt activity in the Superior Parietal Lobule (SPL; Area 7A) and Intraparietal Sulcus (IPS) of the respective hemisphere. Stimulation locations were targeted using real-time neuronavigation software (TMS Neuronavigator; Brain Innovation, Maastricht, The Netherlands). During stimulation, participants were seated comfortably on a reclined chair facing a wall and keeping their head straight. Participants needed five steps from the seat to the force plate. They had to cover this distance with their eyes closed in order to preserve any aftereffects of the stimulation as best as possible.

Testing took place on two non-consecutive sessions with at least 1 day in between stimulation. The order of stimulation locations was randomised across participants with sham stimulation being always the first stimulation in the second TMS session. Sham stimulation was executed over the same target locations as for the cTBS (PMD70-pCool-Sham; MAG & More). The location alternated across the sequence of participants, so that odd and even numbered participants received lPPC or rPPC sham stimulation respectively. Six participants received a lPPC/rPPC order and five a rPPC/lPPC order of stimulation.

Data processing and statistical analysis

The data of the force-torque transducer as well as the kinematic motion capture system were interpolated to 600 Hz and merged with the force plate data. Data were digitally low-pass filtered with a cut-off frequency of 10 Hz (dual-pass, 4th-order Butterworth). Center-of-Pressure (CoP) position was differentiated to yield rate of change

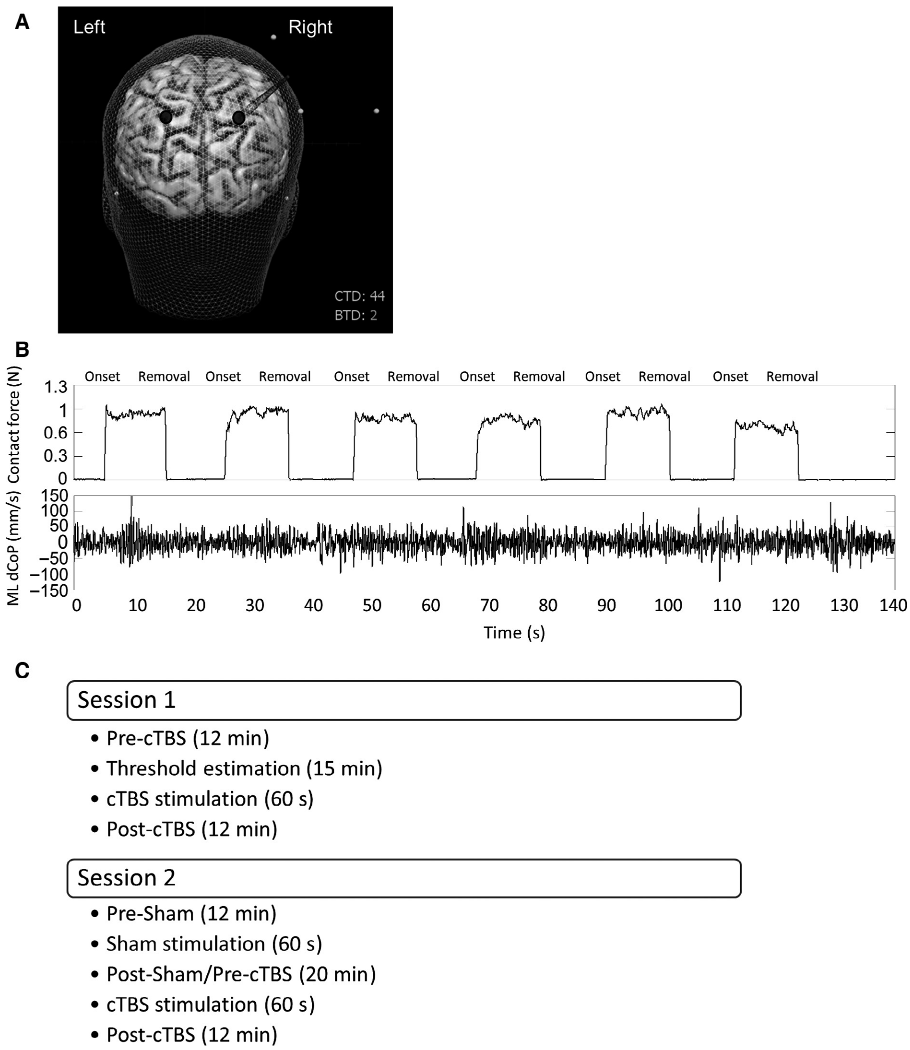

FIG. 1. (A) An illustration of real-time neuronavigation for a participant. Black circles mark the stimulation location in the left and right PPC. (B) A sample trial for single participant. Normal contact force and mediolateral CoP rate of change are plotted across the time course of 140 s trial. (C) Generic overview of the two stimulation sessions.

parameters (dCoP) in order to remove low frequency drift. Based on the Normal force detected by the force-torque sensor, the onset and offset time points of each touching period was determined. In order to represent the time course of sway from 5 s before to 5 s after a contact event (onset/offset), the sway time series was segmented in to temporal bins of 500 ms duration. The standard deviation (SD) of anteroposterior (AP) and mediolateral (ML) dCoP was extracted for each bin. Data processing and extraction was conducted by a custom processing toolbox (MATLAB 2014b). Figure 1B shows the progression of contact force and sway velocity over one trial.

In order to characterise the fluctuation dynamics of body sway in non-transitory, steady postural states, segments of 5 s duration centred in between contact events were extracted from the time series of CoP position. These steady-state segments were appended in order to create time series of at least 25 s duration for Detrended Fluctuation Analysis (DFA) (Peng *et al.*, 1995; Amoud *et al.*, 2007; Duarte & Sternad, 2008). We followed the basic algorithm as described by Peng *et al.* (1995) and obtained the DFA scaling exponent α as the slope of the linear regression of the log-log scaled detrended fluctuation plot as a function of a temporal window width of up to 10 s duration.

Sway in the anteroposterior and mediolateral directions and the scaling exponents were statistically analysed using 4-factorial repeated-measures ANOVA with (i) contacting hand (dominant vs. non-dominant hand; ipsilateral vs. contralateral hand relative to stimulation side), (ii) location of stimulation (rPPC, lPPC and Sham), (iii) effect of stimulation (Pre- and Post-cTBS) and (iv) time course for onset and offset events (time bins) as within-subject factors. In order to test for steady-state effects, time bins 4.5–3.5 s before the contact event and the three last extracted time bins (4–5 s) after the contact event were contrasted for both each respective event type. For statistical significance a Greenhouse-Geisser corrected *P*-value of smaller 0.05 was used. A similar analysis was conducted for the derived contact force. All statistical analyses were carried out using SPSS (IBM SPSS Statistics 21).

Results

Contacting force at the fingertip

Overall, average fingertip contacting force was 2.33 N. Statistical analysis of the average contacting force and its variability did not reveal any effect of hand dominance, location of stimulation, effect of stimulation or any interactions between these factors.

Variability of body sway during contact transitions

Figure 2 shows the progression of sway variability over the time course of 5 s before a contact transition to 5 s after in bins of 500 ms duration before and after cTBS for each of the three stimulation locations. Before onset of fingertip contact, sway variability of the mediolateral direction is high and drops gradually to a lower level after contact is initiated ($F_{19,190} = 19.55$, $P < 0.001$, $\eta^2 = 0.66$). Sway variability remains low as long as contact is kept. Briefly after fingertip contact is removed, variability rises to higher, pre-contact levels ($F_{19,190} = 40.18$, $P < 0.001$, $\eta^2 = 0.80$). A similar progression of sway can be observed in the anteroposterior direction (onset $F_{19,190} = 16.83$, $P < 0.001$, $\eta^2 = 0.63$; offset $F_{19,190} = 16.91$, $P < 0.001$, $\eta^2 = 0.63$).

In terms of the general effect of touch, comparisons between the time bins from 4.5–3.5 s before a contact event and the three last extracted time bins after the same contact event revealed a reduction in body sway variability with touch by 21% in the mediolateral direction (onset: $F_{5,50} = 36.96$, $P < 0.001$, $\eta^2 = 0.79$; removal: $F_{5,50} = 122.49$, $P < 0.001$, $\eta^2 = 0.93$) and by 22% in the anteroposterior direction (onset: $F_{5,50} = 56.12$, $P < 0.001$, $\eta^2 = 0.85$; removal: $F_{5,50} = 51.87$, $P < 0.001$, $\eta^2 = 0.84$).

Regarding the effect of cTBS on sway variability, we found an interaction between stimulation location and stimulation effect in the mediolateral direction ($F_{2,20} = 6.12$, $P = 0.02$, $\eta^2 = 0.38$). We performed *post hoc* ANOVAS for each stimulation location and found general sway reductions after cTBS for both the onset ($F_{1,10} = 5.14$, $P = 0.05$, $\eta^2 = 0.34$) and removal phases ($F_{1,10} = 5.28$, $P = 0.04$, $\eta^2 = 0.35$) after rPPC stimulation but after either lPPC or sham stimulation. In the mediolateral direction, stimulation over the rPPC decreased the sway variability in all phases with and without fingertip contact by 8%. In contrast, sway variability was not reduced by lPPC (3% increase) or sham stimulation (1% increase). In the anteroposterior direction, a similar numerical trend could be observed (rPPC: 8% decrease; lPPC: 3% decrease; sham: 2% increase). However, the interaction between stimulation location and stimulation effect was not significant ($F_{2,20} = 1.78$, $P = 0.20$, $\eta^2 = 0.15$). Figure 3 shows sway variability averaged across all time bins (both onset and removal transitions combined) as a function stimulation location and effect for the mediolateral (Fig. 3A) and the anteroposterior direction (Fig. 3B).

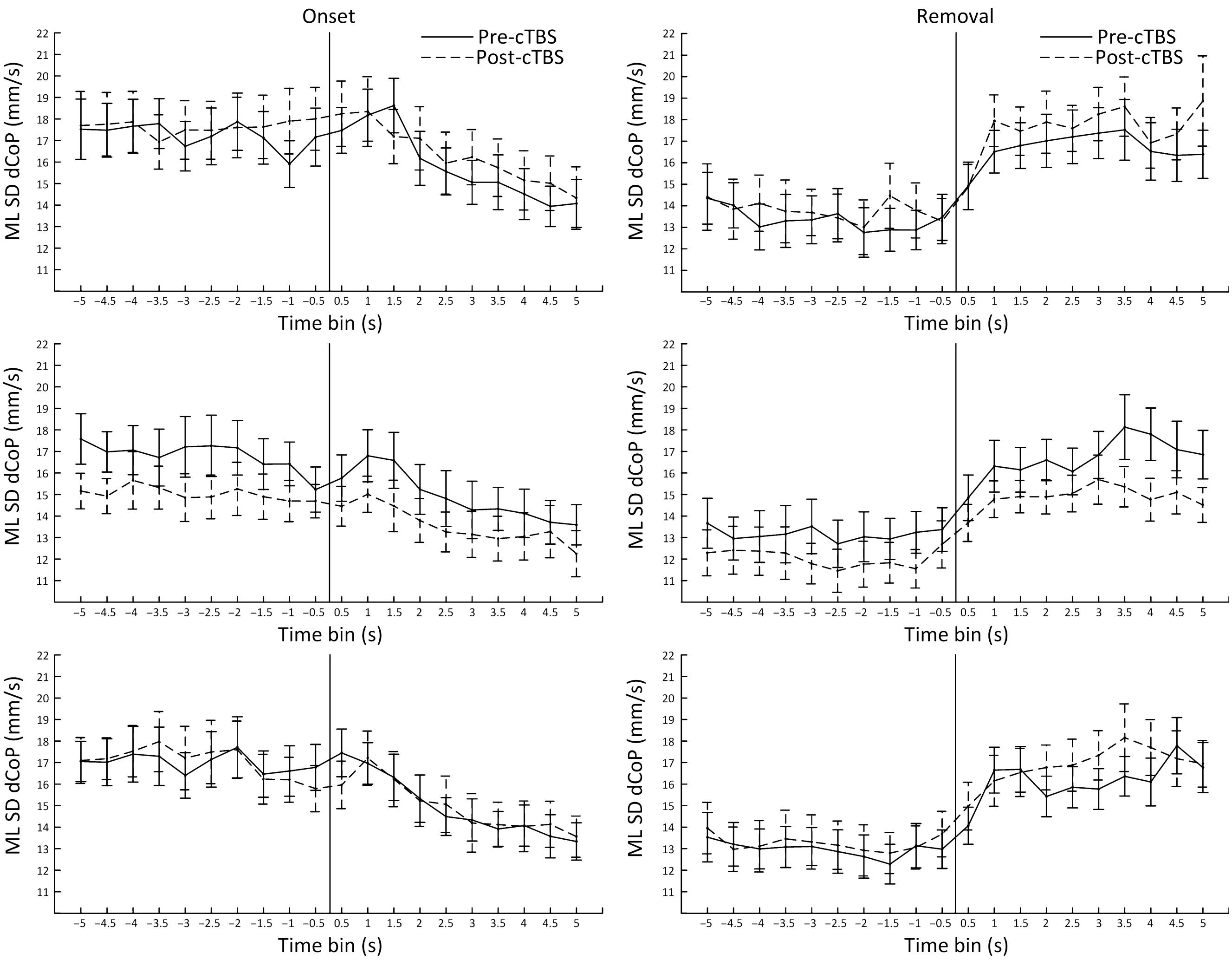

FIG. 2. The time course of mediolateral sway across 20 bins of 500 ms width at contact onset and removal. The black lines indicate body sway variability before cTBS and the dashed lines following cTBS. Error bars indicate standard error of the mean. PPC, posterior parietal cortex.

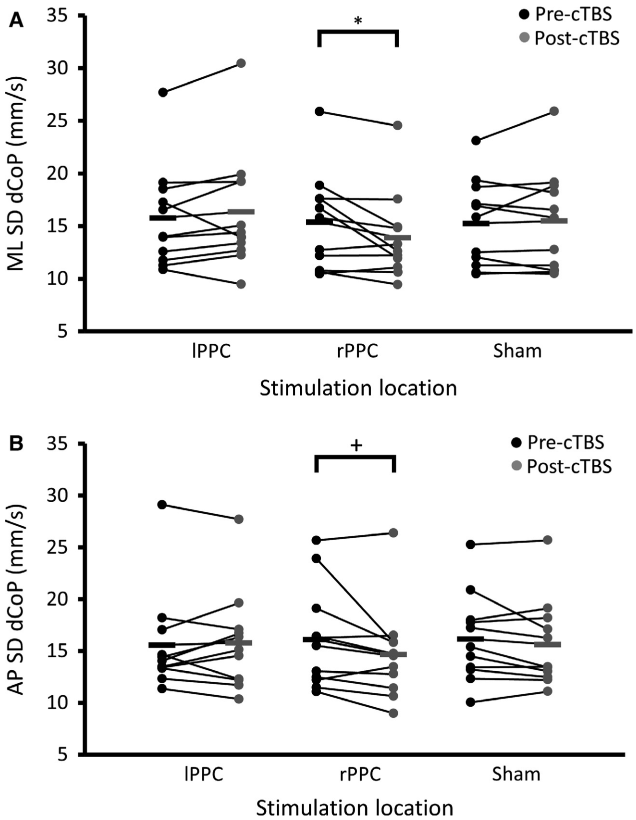

FIG. 3. Grand averaged body sway variability as a function of stimulation location before (light grey points) and after (dark grey points) cTBS for the mediolateral (A) and anteroposterior direction (B). Horizontal bars indicating the mean value averaged across all participants. *: $P < 0.05$. +: $P < 0.10$. lPPC, left posterior parietal cortex; rPPC, right posterior parietal cortex.

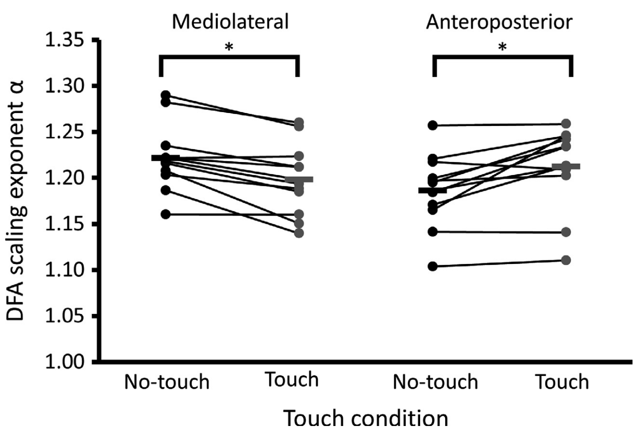

FIG. 4. Scaling exponent as a function of light touch contact for the mediolateral and anteroposterior direction. Horizontal bars indicating the mean value averaged across all participants. *: $P < 0.05$.

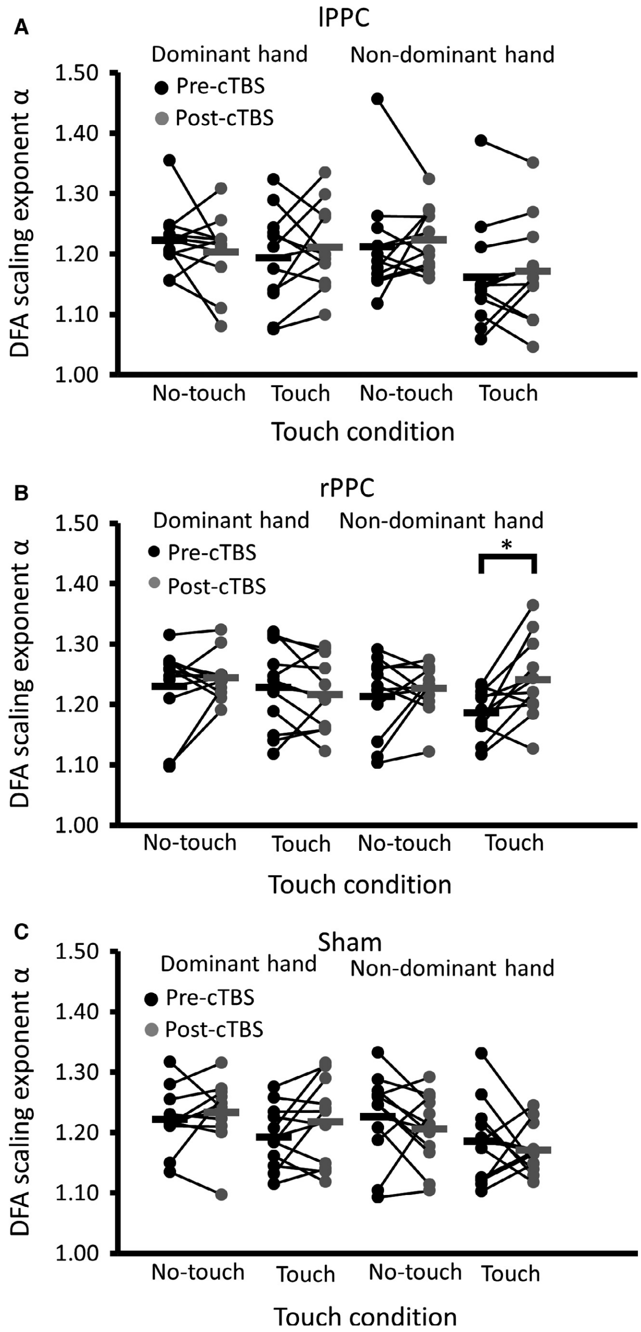

FIG. 5. Scaling exponent as a function of touch contact with the dominant and non-dominant hand before (black points) and after (light grey points) cTBS for (A) Left PPC stimulation, (B) Right PPC stimulation and (C) Sham stimulation. Horizontal bars indicating the mean value averaged across all participants. *: $P < 0.05$. lPPC: left posterior parietal cortex. rPPC, right posterior parietal cortex.

Sway fluctuation dynamics

Detrended fluctuation analysis of sway for the mediolateral direction revealed that fingertip touch decreased the scaling exponent α in the DFA plots compared to No-touch (Fig. 4A; $F_{1,10} = 18.91$, $P < 0.001$, $\eta^2 = 0.65$). In contrast, the scaling exponent α increased with touch in the anteroposterior direction ($F_{1,10} = 9.59$, $P = 0.01$, $\eta^2 = 0.49$).

Furthermore, we found a marginally significant four-way interaction between touch, hand, stimulation location and stimulation effect in the mediolateral direction ($F_{2,20} = 2.77$, $P = 0.10$, $\eta^2 = 0.22$). *Post hoc* single comparisons expressed that rPPC stimulation increased the scaling exponent α with contact of the non-dominant hand ($F_{1,10} = 6.06$, $P = 0.03$, $\eta^2 = 0.38$; Fig. 5B). In contrast, lPPC and sham stimulation resulted in no difference in this contact condition (Fig. 5A and C).

Discussion

We aimed evaluate the effects of disruption by cTBS of the PPC in both hemispheres on the processing of fingertip light touch for body sway control in Tandem-Romberg stance. Surprisingly, after stimulation of the rPPC, the general level of sway variability was decreased. This encompassed all trial phases including those in which light fingertip contact was applied and body sway reduced by the augmented sensory feedback. Light touch changed the sway dynamics in a direction-specific manner in favour of the mediolateral direction. In the mediolateral direction, however, a second effect of rPPC disruption became visible. After the stimulation, the sway dynamics degraded in those phases in which light contact was kept with the non-dominant, contralateral hand.

The general reduction after rPPC disruption appears like an unexpected improvement in sway. Reduced sway variability, however, does not necessarily mean that individuals possess a greater degree of stability in terms of the ability to compensate a balance disturbance. For example, variability is adjusted by the postural control system according to the demands of a specific supra-postural task and seems to be necessary for flexible reactions to external perturbations (Balasubramaniam *et al.*, 2000). It can be argued that the reduction in sway reflects an unfavourable effect in terms of participants becoming less adaptive and less able to compensate for unexpected perturbations (Lipsitz, 2002) after rPPC disruption. Possibly, rPPC disruption resulted in an increase in overall postural stiffness by muscular co-contractions and therefore showed reduced body sway variability (Maurer & Peterka, 2005).

If disruption of the rPPC results in increased stiffness, then the question remains which functional aspect of body sway control does the rPPC represent? We propose a functional equilibrium between the process that controls body stiffness and the process that actively explores the own body's current state of stability in the context of the specific postural configuration and orientation (Riccio *et al.*, 1992). Control of stiffness plays a crucial part when interacting with the environment, for example to gain postural support or when anticipating external perturbations. In the absence of an external perturbation, active stability state exploration would probe for any deviation from the body's equilibrium point by registering the forces and torques required to counteract any environmental dynamics exerted onto the body. Possibly, the rPPC is involved in this active exploration process.

Yadav & Sainburg (2014) propose a distinction between two neural systems for limb control, one for predictive control of arm movements and the other for control of arm stiffness (impedance). The former system is attributed to the dominant (left) hemisphere in right-dominant participants, while the latter to the non-dominant (right) hemisphere (Yadav & Sainburg, 2014). Several studies in stroke patients have implied that the right hemisphere may dominate the control of body sway (Rode *et al.*, 1997; Peurala *et al.*, 2007; Tasseel-Ponche *et al.*, 2015). Assuming that stiffness control by the right hemisphere generalises from the non-dominant arm to the control of body sway, our results suggest that stiffness control and active exploration are two processes coordinated within the right hemisphere. If the rPPC contributes to active exploration, the question remains, which right-hemisphere regions control stiffness. It is likely that the rPPC is part of a network, which is distributed across several brain regions responsible for maintaining a functional equilibrium (Bolton, 2015). Studies reveal a wide spread of different cortical areas involved in the control of balance ranging from the prefrontal cortex, primary motor cortex and the parietal cortex (Mihara *et al.* 2012) to the basal ganglia (Visser & Bloem, 2005). Functions of the basal ganglia include muscle tone regulation and control of automatic postural responses and patients with dysfunction in that area often show axial stiffness, gait freezing or co-contraction (Visser & Bloem, 2005). Thus, the basal ganglia seems like a good candidate to be involved in stiffness or impedance control. The prefrontal, primary motor and parietal cortices might form the exploratory processes for balance control.

Our results show reduced variability of sway with light touch in both directions. Although apparently a similar effect occurred in both directions, there might be differences between mediolateral and anteroposterior sway as the complexity measure of sway dynamics showed opposite changes for both directions. While the scaling exponent α decreases with light touch in the mediolateral direction, it rises in the anteroposterior direction (Fig. 4). In both directions the scaling exponent α was > 1, which is interpreted as a non-stationary signal with low long term self-similarity and reduced complexity. 1/f noise ($\alpha \sim 1$) is associated with a high complexity and is present in many natural, healthy, unperturbed systems (Duarte & Zatsiorsky, 2001). Deviations from this complexity range might result in pathophysiological disturbances (Duarte & Zatsiorsky, 2001; Peng *et al.*, 1995). Perhaps, the generally > 1 scaling exponent α in our study is an expression of the increased postural challenge caused by the stance position with eyes closed. Although the scaling exponent α does not decrease to a value close to or below 1, a reduction could be observed in the mediolateral direction at the cost of an increase in the anteroposterior direction with light touch.

It might be possible that with light contact the dynamics of sway became more direction-specific. Participants stood in Tandem-Romberg stance, which introduces imbalance especially in the mediolateral direction. Therefore, this direction might have become more task-goal relevant in terms of the utilisation of the haptic signal for the control of sway. These effects in the mediolateral direction occurred despite the contact point being orientated along the orthogonal, anteroposterior direction. Effects might be even stronger if the contact point is positioned along the mediolateral axis (Jeka *et al.*, 1998). We placed the contact point on the midline to enable quick switching between the two hands as two force-torque sensors were not available to us for placement of one contact point on each side. The sway dynamics do not show a general effect of rPPC disruption. Instead, results show an increase in the scaling exponent α after disruption of the rPPC with fingertip contact of the non-dominant, contralateral hand. It might be that the disruption led to a non-optimal integration of haptic information for body sway control. Ishigaki *et al.* (2016) demonstrated that processing of a haptic signal when it contains information about body sway relative to an earth-fixed reference reduces cortical activity in the contralateral left-hemisphere parietal lobe as determined by EEG. Unfortunately, they did not assess the effect of contact with the non-dominant (left) hand. We would expect similar contralateral activity reductions in the right-hemisphere parietal lobe.

We did not find an increase of the scaling exponent α in the dominant hand after lPPC disruption. It might simply be that we missed the adequate target location in the left-hemisphere parietal lobe to induce any disruptive effects. It might also be possible, however, that differences between the hemispheres exist with respect to the processing of tactile feedback for sway control. In a previous study, we did not find any disruptive effects of rTMS over the left IPG and left middle frontal gyrus on steady-state body sway with LT (Johannsen *et al.*, 2015). It may be that a disruption of the left-hemisphere was compensated by other brain regions for example the rPPC.

Figure 6 summarises a simple functional model of interhemispheric interaction, which could underlie our effect patterns. Assuming that rPPC is part of a neural architecture which controls active

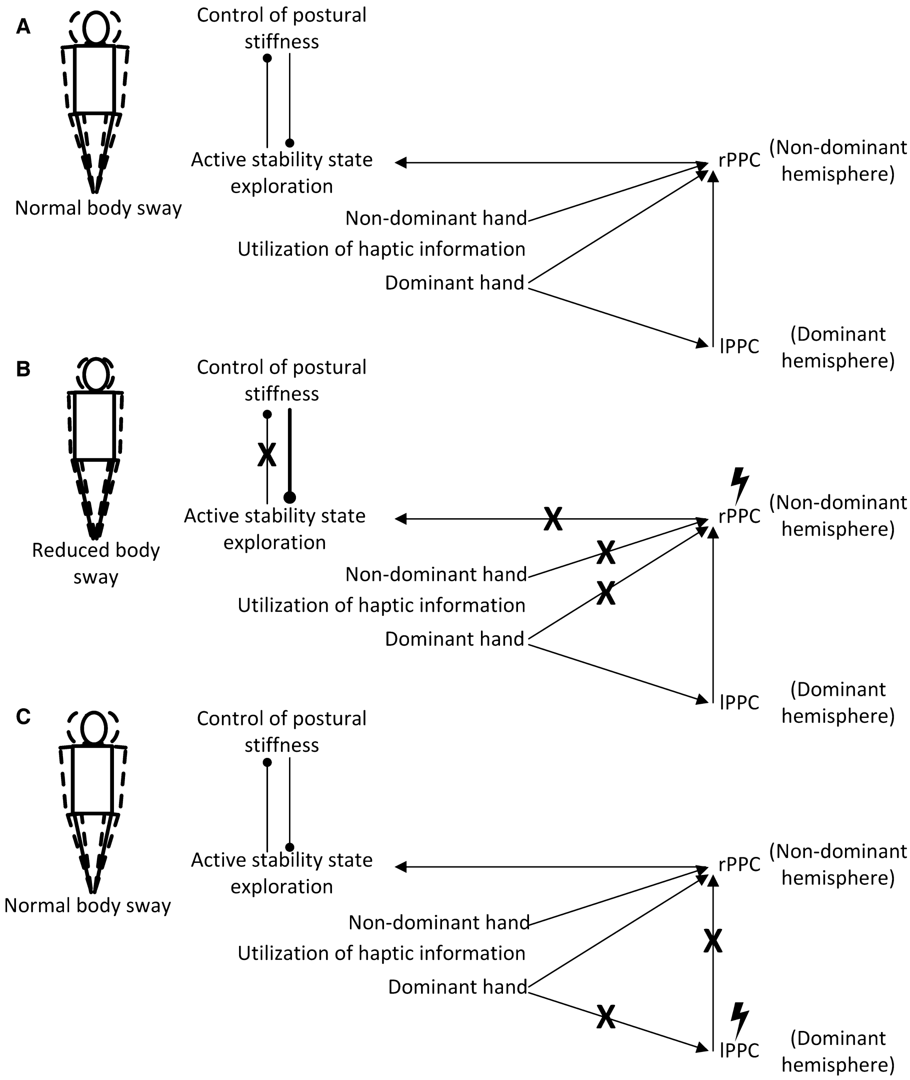

FIG. 6. A simplistic functional model of interhemispheric interactions for active stability state exploration. (A) No cTBS disruption. (B) cTBS over the right parietal cortex. (C) cTBS over the left parietal cortex. lPPC: left posterior parietal cortex. rPPC, right posterior parietal cortex. Lightning symbol: cTBS disruption. X: dysfunction.

exploration of the postural stability state opposed by other structures which regulate postural stiffness, rPPC might utilise the haptic signal at the fingertips for this task. rPPC may be disposed to processes haptic information in ego-centric reference frames (Longo *et al.*, 2010; Medina & Coslett, 2010) from both hands, while lPPC processes and relays haptic information from the contralateral hand only (Fig. 6A). If rPPC is disrupted by cTBS, active stability state exploration may be impaired leading to reduced body sway (Fig. 6B). In addition, the utilisation of haptic information for sway control from both hands may be affected. In terms of the sway dynamics, a deficit becomes apparent for the contralateral (relative to rPPC), non-dominant hand as the left-hemisphere can still process and relay in a signal from the contralateral (relative to lPPC), dominant hand. Finally, if lPPC is disrupted by cTBS (Fig. 6C), only processing of the dominant hand's haptic information is impaired, which can be compensated by rPPC's own access to ipsilateral haptic information. For example, Borchers *et al.* (2011) reported a stroke patient, who demonstrated a proprioceptive deficit for both hands after a right postcentral lesion. Ishigaki *et al.* (2016), however, did not report bilateral activity changes during quiet stance with light touch but exclusively in the dominant hemisphere contralateral to the contacting hand. As both hemispheres were undisturbed physiologically in their experiment, it may be that any ipsilateral activity changes in the right hemisphere were suppressed.

Continuous TBS over the right or left PPC had no effect on the applied finger force and its variability. Even though average contacting force exceeded 1N, we still consider it a light touch as the applied forces were still not sufficient to provide mechanical support. Moreover, we argue that the light touch in our experiment is a more natural evolving light touch as we tried to avoid turning it into an explicit precision task by including online force feedback. It might be possible, however, that the applied touch in our experiment is processed differently than light touch of lesser than 1N. Jeka & Lackner (1994) reported that feedback delays between fingertip forces and postural adjustments were much longer and the coupling weaker -with contact below 1 N compared to contact with unconstrained forces showing shorter time lags and stronger coupling between fingertip forces and postural adjustments. In this respect the latter might resemble classical supraspinal, long-latency reflexes. Average contact forces in the unconstrained condition in Jeka &

Lackner (1994), however, exceeded 4 N, which is at least twice the amount of contact forces in our present study. Whether the processing of haptic feedback below 1 N or above 4 N is linked with a continuous functional gradient or whether a discontinuity exists between these two ranges is unknown to date and worth further investigation. As contact forces in our present study are closer to the 1 N range, we suggest that the haptic signals in our study should still be considered 'light' but we cannot exclude the possibility that this was the reason disruption of the PPC led to no changes in the level of sway specifically with light touch.

In conclusion, we replicated the traditional effect of light touch on body with decreased sway variability but showed direction-specific changes in its complexity. Moreover, we showed that overall sway variability decreases, in addition to the light touch effect, while the sway complexity increases when utilising haptic information from the non-dominant, contralateral hand after rPPC disruption. We speculate that an increase in postural stiffness could result from lowered inhibition of stiffness regulation by a disrupted process, which is engaged in actively exploring the body's stability state. We propose a simple functional model of interhemispheric interactions, which could explain our results pattern by the assumption of an asymmetry between the rPPC and lPPC regarding bilateral utilisation of haptic information for the control of body sway.

Conflict of interest

The authors declare no conflict of interest.

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Data accessibility

The experimental data will be accessible via the institutional media repository of the Technical University Munich (https://mediatum.ub.tum.de).

Author contributions

DK and LJ contributed equally to the study design, data collection, data analysis and manuscript preparation. JH contributed to study design and manuscript preparation.

Abbreviations

AP, Anteroposterior; CoP, Center-of-Pressure; DFA, Detrended Fluctuation Analysis; IPG, Inferior Parietal Gyrus; ML, Mediolateral; PPC, Posterior Parietal Cortex; SD, Standard Deviation; TBS, Theta Burst Stimulation; TMS, Transcranial Magnetic Stimulation.

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RESEARCH ARTICLE

Stabilization of body balance with Light Touch following a mechanical perturbation: Adaption of sway and disruption of right posterior parietal cortex by cTBS

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Abstract

Light touch with an earth-fixed reference point improves balance during quite standing. In our current study, we implemented a paradigm to assess the effects of disrupting the right posterior parietal cortex on dynamic stabilization of body sway with and without Light Touch after a graded, unpredictable mechanical perturbation. We hypothesized that the benefit of Light Touch would be amplified in the more dynamic context of an external perturbation, reducing body sway and muscle activations before, at and after a perturbation. Furthermore, we expected sway stabilization would be impaired following disruption of the right Posterior Parietal Cortex as a result of increased postural stiffness. Thirteen young adults stood blindfolded in Tandem-Romberg stance on a force plate and were required either to keep light fingertip contact to an earth-fixed reference point or to stand without fingertip contact. During every trial, a robotic arm pushed a participant's right shoulder in medio-lateral direction. The testing consisted of 4 blocks before TMS stimulation and 8 blocks after, which alternated between Light Touch and No Touch conditions. In summary, we found a strong effect of Light Touch, which resulted in improved stability following a perturbation. Light Touch decreased the immediate sway response, steady state sway following re-stabilization, as well as muscle activity of the Tibialis Anterior. Furthermore, we saw gradual decrease of muscle activity over time, which indicates an adaptive process following exposure to repetitive trials of perturbations. We were not able to confirm our hypothesis that disruption of the rPPC leads to increased postural stiffness. However, after disruption of the rPPC, muscle activity of the Tibialis Anterior is decreased more compared to sham. We conclude that rPPC disruption enhanced the intra-session adaptation to the disturbing effects of the perturbation.

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Abbreviations: CNS, Central Neuro System; CoG, Centre of Gravity; CoM, Centre of Mass; CoP, Centre of Pressure; cTBS, continuous Theta Burst Stimulation; dCoP, differentiated Centre of Pressure; IPG, Inferior Parietal Gyrus; LT, Light Touch; PPC, Posterior Parietal Cortex; rTMS, repetitive Transcranial Magnetic Stimulation; rPPC, right Posterior Parietal Cortex; TMS, Transcranial Magnetic Stimulation.

Introduction

The main objective for the control of body posture and balance is to stabilize upright standing against the pull of gravity or any other external forces and to prevent the body from toppling over. This is achieved by keeping the Centre of Mass' (COM) vertical projection onto the ground (Centre of Gravity, CoG) within the support boundaries. In order to maintain balance, the Central Nervous System (CNS) relies on sensory feedback processed by the visual, vestibular and somatosensory systems [1]. However, in addition to its primary senses the CNS is also able to use information from secondary afferent channels, such as the skin, as long sway-related information is conveyed. Light touch (LT) with an earth-fixed reference point has been shown to decrease sway variability and improve balance during quite stance [2] but also in dynamic situations, such as when compensating an either foreseeable or unpredictable external perturbation. For example, Dickstein and colleagues [3] demonstrated that Light Touch facilitates the scaling of postural compensation in response to horizontal support surface translations. Furthermore, Light Touch results in faster stabilization and reduced body sway following both externally and self-imposed body balance perturbations [4]. Imposing the sudden release of a backward load to the trunk, Martinelli et al. [5] reported that Light Touch reduced and slowed Centre-of-Pressure (CoP) displacement as well as decreased activity in the lower limbs' Gastrocnemius muscles under challenging sensory conditions. Johannsen and co-workers [6] also provided evidence for the benefit of Light Touch in dynamic postural contexts by exerting abrupt backward perturbations onto participants standing on a compliant springboard under different conditions of visual feedback. The utilization of Light Touch stabilized balance and decreased thigh muscle activity by up to 30%, which indicates that Light Touch optimizes mechanical and metabolic costs of balance compensation following a perturbation to a compliant support surface [6].

Although responses to postural perturbations are faster than voluntary movements, the observation that long-latency reflexes are sensitive to the postural context suggests involvement of supraspinal neural circuits including the cerebral cortex [7]. Several studies implied a role of cortical neural circuits in the control of posture when anticipating a perturbation to body balance. Cortical potentials preceding self-initiated perturbations, as well as predictable external perturbations show differences in amplitude as well as temporal characteristics [8], which might represent adjustments in a central set prior to the onset of a known perturbation. Depending on alterations in the cognitive state, such as changes in the cognitive load or attentional focus, initial sensory-motor conditions, prior experience and prior warning of a perturbation influences the central set enabling adaptations of the postural response to a perturbation [7]. Several cortical areas have been identified for playing a role in the control of balance, mainly the primary motor cortex, the somatosensory cortex and the posterior parietal cortex (PPC). For example, the primary motor cortex is responsible in the regulation of induced postural responses of the lower limbs [9]. Taube et al. [9] applied a single pulse TMS paradigm to demonstrate that corticospinal projection to the soleus muscle facilitates long-latency responses following abrupt backward translations of the support. Similarly, the sensorimotor cortex has been reported to play a role not only in the integration and in processing of sensory information, but also in adjusting the central set to modify externally triggered postural responses [7]. In addition, involvement of the supplementary motor area in motor planning and preparation for an adequate response to perturbations has been reported [10–12]. Contrasting balance perturbations caused by horizontal translations of a support surface with and without an auditory pre-warning, Mihara et al. [10] used functional near-infrared spectroscopy to demonstrate that both the left-hemisphere supplementary motor area and the right-hemisphere posterior parietal cortex increased activation, when preparation for the upcoming perturbation was possible. This observation argues for an involvement of both areas in the anticipation and probably also compensation of an expected postural imbalance. Likewise, An et al. [13] who investigated the contribution of the sensory motor cortex and the PPC to recovery responses following unpredictable perturbations during standing or walking. Both areas showed a suppressed activity in the alpha band during periods of balance recovery [13]. The significant role of the posterior parietal cortex in the stabilization of balance is further corroborated by Lin et al. [14]. They showed that a lesion in the posterior parietal cortex following stroke leads to reactive postural control deficit, such as impaired recruitment of paretic leg muscles and a more frequent occurrence of compensatory muscle activation patterns compared to controls. Lin et al. [14] concluded that the PPC is part of a neural circuitry involved in reactive postural control in response to lateral perturbations.

Regions of the cerebral cortex are also involved in the processing and integration of the sensory information from the fingertips when utilizing Light Touch for postural control. Ishigaki et al. [15] demonstrated involvement of the left primary sensorimotor cortex and the left posterior parietal cortex in stance control with light tactile feedback. Johannsen et al. [16] investigated how rTMS over the left inferior parietal gyrus (IPG) influences sensory re-organization for the control of postural sway with light fingertip contact. They reported that rTMS over the left IPG reduced overshoot of sway after contact removal, which indicates that this brain region may play a role in inter-sensory conflict resolution and adjustment of a central postural set for sway control with contralateral fingertip contact.

Assuming that an ego-centric reference frame would be the basis of interpreting and disambiguating fingertip Light Touch for sway control in a quiet upright stance with transitions between postural states with and without Light Touch feedback, we investigated the effects of disrupting the left- and right hemisphere PPC using continuous Theta Burst Stimulation (cTBS) [17]. We expected that disruption of the right Posterior Parietal Cortex would impair integration of Light Touch into the postural control loop and attenuate the effect of Light Touch on body sway. These expectations were not confirmed but we demonstrated that rPPC disruption influenced the complexity of body sway with Light Touch of the non-dominant, contralateral hand [17]. In addition, disruption of the rPPC resulted in an overall sway reduction and altered complexity irrespective of the presence of Light Touch. A possible reason could be that rPPC disruption increased overall body stiffness due to lower limb muscular co-contractions and thus reduced body sway [18]. Sway reduction does not mean, however, that participants are intrinsically more stable. Variability is a means of the postural control system to achieve a specific task goal while at the same time being more able to react flexibly to possible external balance perturbations [19]. Thus, it can be argued that the reduction in sway reflects an unfavourable effect in terms of participants becoming less adaptive and less able to compensate unexpected perturbations [20] after rPPC disruption.

Taking into account the well documented light-touch-related facilitation of balance stabilization, following an external perturbation [3,4,5,6] we implemented a perturbation paradigm to assess the influence of rPPC disruption on dynamic stabilization of body sway with and without Light Touch. In previous studies, however, perturbations consisted either of a single constant force or of variable forces but in a blocked design, making perturbations much more predictable, enabling adjustment to a central postural set. In our current study, we intended to make it much more difficult for the participants to predict the force of an upcoming perturbation. Therefore, we randomized three forces on a trial-by-trial basis within a block of either Light Touch or no touch. We hypothesized that the benefit of Light Touch would be amplified in the more dynamic context of an external perturbation to balance, improving the compensation response. We also expected that the immediate response to a perturbation and sway stabilization in terms of its time constant would be affected expressing an increase in postural stiffness following rPPC disruption.

Methods

Participants

Thirteen healthy right-handed young adults (age = 26 ± 2 (SD); 10 women and 3 men) were recruited for this study, using the faculties own blackboard. Inclusion criteria were (1) no neurological or musculoskeletal disorders, (2) no balance impairment and (3) no known history of epilepsy or reported seizures. All participants were informed about the study protocol and signed a written informed consent. The study was approved by the Clinical Research Ethics committee of the Medical School of the Technical University Munich.

Study protocol, apparatus and experimental procedure

The study protocol comprised of two single TMS sessions in the balance lab. The order of stimulation locations (rPPC or sham TMS) was randomized across participants. Stimulation sessions were separated by at least 24 hours. Each experimental testing session consisted of three parts: a balance pre-test, 60 seconds of cTBS and a balance post-test. During the pre- and post-test participants stood in Tandem-Romberg stance on a force plate (600Hz; Bertec FP4060-10, Columbus, Ohio, USA), with their eyes blindfolded and instructed to stand quietly but relaxed and not to attempt to minimize body sway.

Participants were required either to keep light haptic fingertip contact with their dominant hand to an earth-fixed reference point or to stand without fingertip contact. Participants practiced keeping Light Touch with the reference point prior to the start of the experiment receiving verbal feedback about the strength of the contact force until they felt comfortable maintaining Light Touch below 1 N. During the experiment, however, participants did not receive feedback about contact force to prevent contacting from becoming an explicit, attention-demanding precision task. The earth-fixed contact reference point was placed in front of the participants. They held one arm slightly angled in front of the body and reaching straight forward. The other arm remained passive at the side of their body (Fig 1).

Body kinematics (4 Oqus 500 infrared cameras; 120 Hz; Qualisys, Göteborg, Sweden) and forces and torques at the fingertip reference contact location (6DoF Nano 17 force-torque transducer; 200 Hz; ATI Industrial Automation, Apex, USA) were also acquired. To capture body motion, reflective markers were placed at the contacting fingertip, wrist, elbows, shoulders, C7, Sternum, hip, knees and ankles. Additionally, surface EMG (1kHz) of the Gastrocnemius, Soleus and Tibialis Anterior of the posterior supporting leg was recorded to measure muscle activity (Trigno Wireless PM-W05, Delsys, Natic, MA, USA).

During every single standing trial, a robotic arm (KUKA LBR4+, Augsburg, Germany) exerted a push to participants at their right shoulder in medio-lateral direction. In order to make the next perturbation force as unpredictable as possible, the force of a lateral push was exerted with either 1%, 4% or 7% of their respective body weight in a randomized order in a block consisting of 6 trials (2 trials for each push force). Using a percentage of the body weight for every single participant, results in different absolute forces for the participants. However, relative force of the push for the perturbation is equalized for across participants. Table 1 shows the absolute peak push forces in N for the conditions averaged over all participants.

A testing session consisted of 4 blocks before the cTBS application (pre-test) and 8 blocks after (post-test). The blocks alternated between Light Touch (LT) and No Touch (NT) conditions. For a comparison between sway before and after the cTBS application, sway was averaged across the NT and LT blocks respectively (pre-test: NT = blocks 1+3, LT = blocks 2+4; post-test: NT = blocks 6+8+10+12; LT = blocks 5+7+9+11). Duration of a single trial was 20 seconds, with the lateral push always applied at 4.5 seconds after the start of a trial (Fig 2).

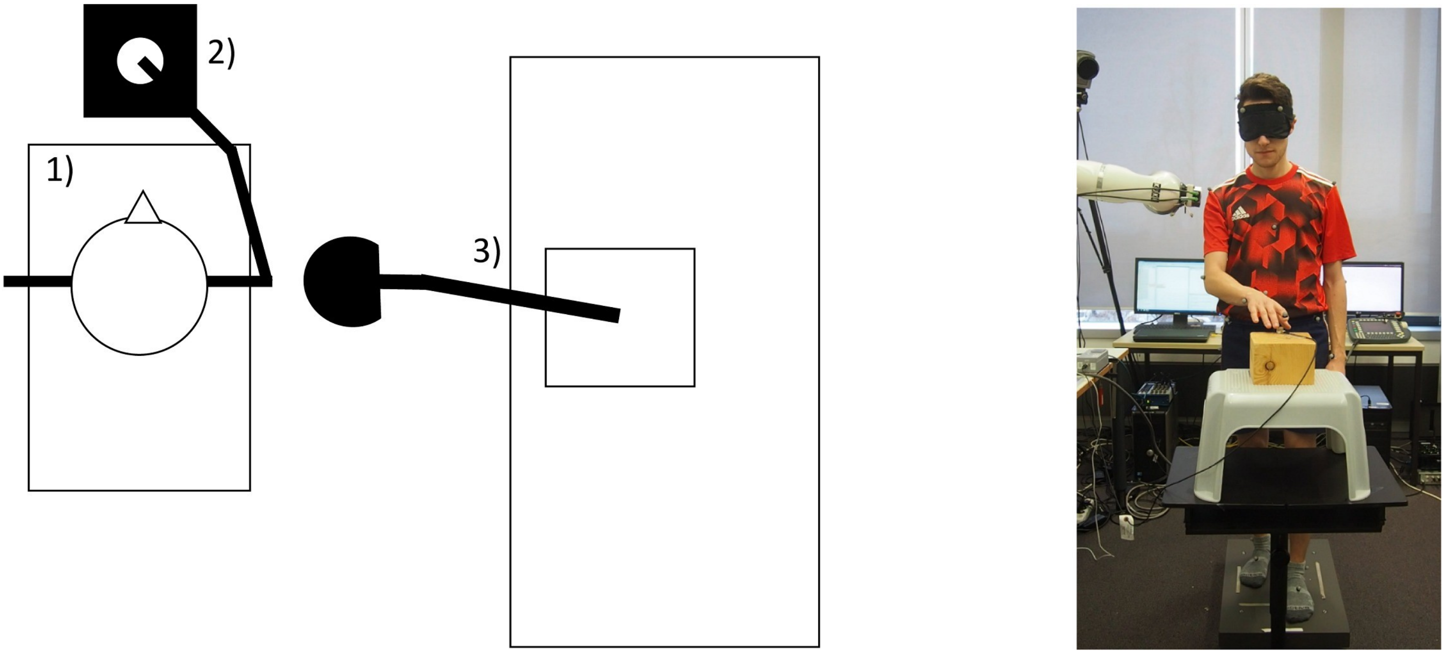

Fig 1. Experimental set up as seen from above. (1) Force plate, (2) contact reference point on a waist high stand and (3) Robotic arm mounted on a table.

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Neuronavigation and TMS protocol

During cTBS stimulation, participants were seated comfortably on a reclined chair facing a wall and keeping their head straight. We applied continuous Theta Burst Stimulation (cTBS) of an intensity of 80% of the passive motor threshold for 60 seconds over the rPPC (PMD70-pCool; MAG & More, Munich, Germany). This protocol is widely used and stimulation effects can last from 20 minutes up to 1 hour (Staines & Bolton [21]. The passive motor threshold was determined by registering the motor evoked potential (MEP) at the musculi interossei dorsales manus of the left hand following a single TMS pulse over the hand representation of the right-hemisphere primary motor cortex. A staircase procedure was used to adjust the pulse intensity until a 50µV MEP could be elicited reliably [22].

Sham stimulation was applied over the same target location as for the cTBS using a sham coil powered at similar intensities, which produced no focussed magnetic induction but created similar acoustics and tactile sensation. (PMD70-pCool-Sham; MAG & More, Munich, Germany).

High-resolution anatomical brain scans were acquired before the study at the University Hospital Großhadern, Center for Sensorimotor Research and consisted of a T1 MPRAGE (3T

Table 1. Push forces averaged over all participants broken down by force push condition and stimulation protocol.

% of Body Weight	Stimulation Protocol	Force (N)
1	Sham	2.99
1	Stim	2.89
4	Sham	6.95
4	Stim	6.01
7	Sham	11.56
7	Stim	10.06

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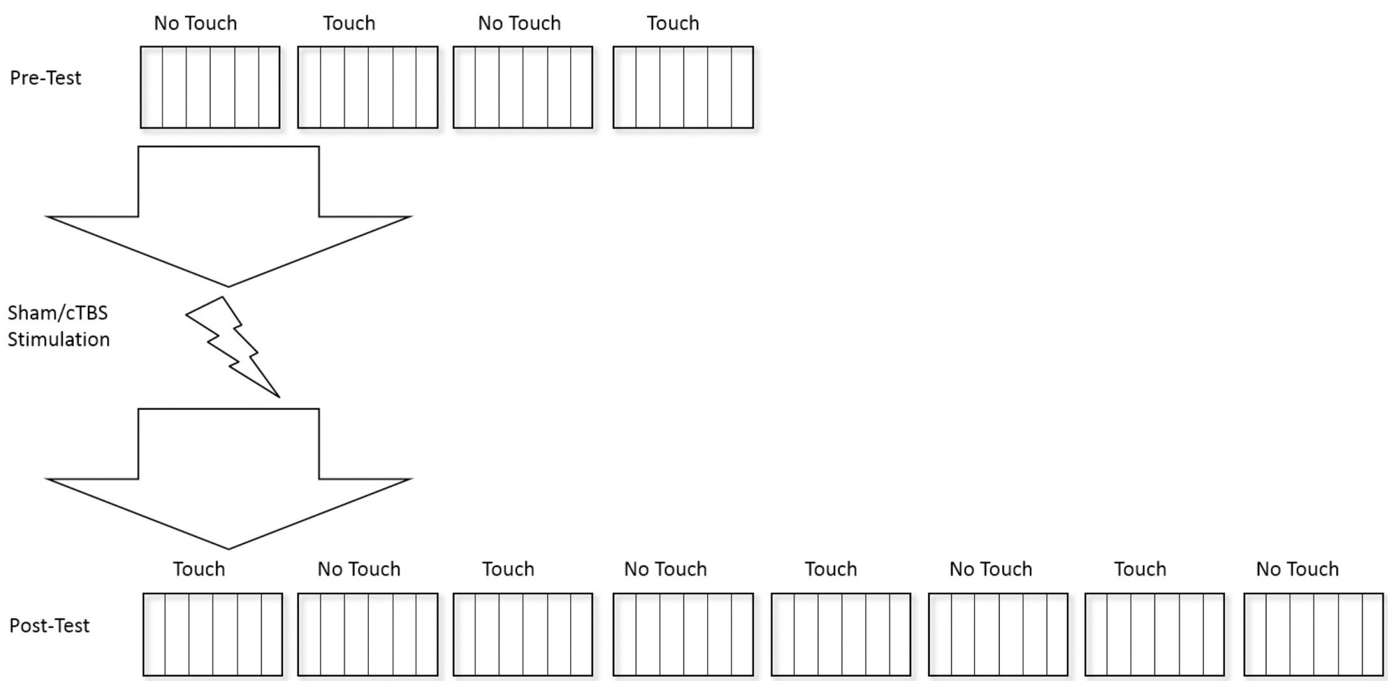

Fig 2. Experimental process. Rectangle boxes represent blocks, separated by lines representing single trials.

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whole-body scanner, Sigma HDx, GE Healthcare, Milwaukee, Wisconsin, USA). In order to define the cTBS target area, we used MNI coordinates (x = 26, y = 258, z = 43) reported in Azañón et al. [23] (2010), who stimulated the right-hemisphere human homologue of macaque ventral intraparietal area. We therefore expected that cTBS would disrupt activity in the Superior Parietal Lobule (SPL; Area 7A) and Intraparietal Sulcus (IPS) of the right hemisphere. Stimulation locations were targeted using real-time neuronavigation software (TMS Neuronavigator, Brain Innovation, Maastricht, Netherlands).

In order to localize the stimulation area for each individual participant, the high-resolution scan was co-registered and normalized to the MNI template.

Data processing and data reduction

All data processing was performed using customized functions scripted in Matlab 2018b (Mathworks, MA, USA). Centre-of-Pressure (CoP) data of the force plate was digitally low-pass filtered with a cut-off frequency of 10 Hz (dual-pass, 4^{th}-order Butterworth). CoP position was differentiated to obtain CoP rate-of-change in m/s(dCoP). In order to characterize balance recovery, we followed a similar approach as applied in Johannsen et al. [4]. The standard deviation of the medio-lateral dCoP (SD dCoP) was calculated for each of 13 temporal bins of 1 s duration before and after the moment of the perturbation. A period of 3 s duration before the perturbation served as an intra-trial sway baseline. Across the 10 post-perturbation bins demonstrating stabilization, we fitted from an exponential decreasing non-linear regression $x(t) = C + A^{*}e^{\left(-\frac{t}{B}\right)}$, from which we obtained the function parameters A (intercept), B (time constant) and C (asymptote). The intercept is derived from the body sway at perturbation (t = 0) and therefore reflects the immediate effect of the perturbation. The time constant represents the rate of stabilization of body sway after the perturbation with shorter time constants indicating faster stabilization. The third parameter, the asymptote, indicates the level of steady-state long-term stabilization.

EMG recordings were band-pass filtered between 10 and 500 Hz, rectified and smoothed by a moving average with 15ms width to obtain the EMG activity envelope of a muscle. For each muscle we extracted peak amplitude, indicating the amount of phasic activity directly following a perturbation and the area-under-the-curve of the activity envelope as an indication of the tonic activity across an entire trial serving as an indication of general muscle activation. EMG activity was then normalized to the first baseline block for NT and LT respectively and percentage of change from baseline was calculated.

Statistical analysis

Data of the robotic device was checked for failures to deliver a forced push with an abrupt impact and immediate withdrawal of the end-effector. Trials in which the robotic arm only continuously shoved participants were excluded. Only successful force pushes were included in the data analysis. Overall there was a success rate of 87%.

Only trials with exponential fits of greater than 75% explained variance were included in the subsequent statistical analysis. In total, 15% of trials did not reach this threshold and were excluded from the statistical analysis. In order to identify possible non-responders to the cTBS stimulation we applied a k-means cluster analysis. K-means cluster analysis is a unsupervised learning algorithm that tries to cluster data based on their similarity, once the amount of desired clusters is defined. We defined 2 clusters (Responder vs. Non-responder) that we wanted data to be grouped into. Data for the intercept, time constant, asymptote, peak amplitude and area under the curve were pooled together and clustered in the two groups of either responders or non-responders. We identified two possible non-responders, leaving us with 11 participants for the statistical analysis. Prior to analysis data was log transformed to fit normal distribution. Parameters were then analysed statistically using a linear mixed model, with four repeated-measures factors (1) hand contact (Touch vs. No Touch), (2) stimulation session (cTBS vs. Sham), (3) Test (pre- vs. post-stimulation) and (4) force push (1% vs 4% vs 7%): (Variable~Stimulation_Session+Hand_Contact+Test+Force_Push+Stimulation_Session*Hand_Contact+Stimulation_Session*Test+Stimulation_Session*Force_Push+Hand_Contact*Test+LT*Force_Push+Test*Force_Push+Stimulation_Session*Hand_Contact*Test+Stimulation_Session*Test*Force_Push+Stimulation_Session*Hand_Contact*Force_Push+Hand_Contact*Test*Force_Push+Stimulation_Session*Hand_Contact*Test*Force_Push + (1 |Subjects)) (Table 2). Fixed effects were "Hand_contact", "Stimulation_Session", "Test" and "Force_Push". Force push was treated as continuous, the others as factors. A post-hoc analysis was carried out to clarify the effects of stimulation session on muscle activity. A linear model with three repeated-measures factors (1) Test (pre- vs. post-stimulation), (2) hand contact (Touch vs. No Touch) and (3) force push (1% vs. 4% vs. 7%) was carried out for both stimulation sessions (sham and cTBS) respectively: (Variable~Test+Hand_Contact+Force_Push+Test*Hand_Contact+Test*Force_Push+Force_Push*Hand_Contact+Test*Hand_contact*Force_push + (1|Subjects)).

We also performed an analysis to investigate progression of sway over time with three repeated-measures factors (1) Block (progression over time), (2) hand contact (Touch vs. No Touch) and (3) stimulation session (cTBS vs. Sham): (Variable~Stimulation_Session+Hand_Contact+Block+Stimulation_Session*Hand_Contact+Stimulation_Session*Block+Stimulation_Session+Hand_Contact*Block+LT+Block+Stimulation_Session*Hand_Contact*Block+Stimulation_Session*Block+Stimulation_Session*Hand_Contact+Hand_Contact*Block+Stimulation_Session*Hand_Contact*Block + (1 |Subjects)) (Table 3). We also

Table 2. Results for Centre of Pressure and EMG.

Measure	P value								
	Light Touch F (1,231)	Test F (1, 231)	Push Force F(2, 231)	Light Touch x Test F(1, 231)	Stimulation protocol x Test F(1, 231)	Test x Push Force F(1, 231)	Light Touch x Push Force F (2, 231)	Light Touch x Stimulation Protocol F(1, 231)	Stimulation protocol x Light Touch x Test F(1, 231)
	Centre of Pressure								
Intercept	< .01	< .001	< .001	< .05	NS	NS	NS	NS	NS
Slope	NS	NS	< .05	NS	NS	NS	NS	NS	NS
Constant	< .001	< .001	< .001	< .001	NS	NS	NS	NS	NS
	Tibialis Anterior								
EMG Integral	< .001	< .001	NS	< .05	< .001	NS	NS	< .05	NS
Peak Amplitude	< .001	< .01	< .01	NS	NS	NS	NS	NS	NS
	Gastrocnemius								
EMG Integral	NS	< .01	NS	NS	NS	NS	NS	NS	NS
Peak Amplitude	< .001	< .001	NS	NS	< .05	NS	NS	< .05	NS
	Soleus								
EMG Integral	NS	NS	NS	NS	NS	NS	NS	NS	NS
Peak Amplitude	< .05	NS	< .001	NS	NS	NS	NS	NS	NS

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performed a post-hoc analysis with specific focus on the first four blocks before the stimulation (Variable ~ Stimulation_Session + Hand_Contact + Block + Stimulation_Session*Hand_Contact + Stimulation_Session*Block + Hand_Contact*Block + Stimulation_Session*Hand_Contact* Block + (1 | Subjects)), investigating whether stimulation protocol had an influence in the pre-test already. This would hint at a session effect rather a stimulation effect.

For statistical significance, a p-value of 0.05 was used. Statistical analysis was carried out using the lme4 package in R-statistics (R version 3.4.0). Model estimates of the two main linear mixed models can be found in the supporting information.

Results

General sway analysis

Fig 3 shows illustrative data of one participant, averaged over all conditions. After the perturbation, the C7 body marker is deflected laterally accompanied by an excursion of the differentiated CoP signal. EMG activity of the Gastrocnemius rises to produce the required torque to compensate the perturbation. As a result, the CoP is accelerated into the opposite direction and C7 returns to the baseline position. EMG activity and CoP settle at pre-perturbation levels again until the end of the trial.

CoP stabilization

Light Touch improved the immediate sway response to the perturbation compared no touch (Table 2). As can be seen in Fig 4, participants showed lower intercepts independently of the type of stimulation. Post hoc analysis revealed a significant effect of block, which is the progression over all 12 blocks (Table 3).

Table 3. Results for analysis of gradual decrease.

Measure		P value						
		Stimulation Protocol F(1,238)	Light Touch F (1,238)	Block F (1,238)	Stimulation Protocol x Light Touch F(1,238)	Stimulation protocol x Block F(1,238)	Light Touch x Block F(1,238)	Stimulation protocol x Light Touch x Block F(1,238)
		Centre of Pressure						
	Intercept	NS	< .001	< .05	NS	NS	NS	NS
	Slope	NS	NS	NS	NS	NS	NS	NS
	Constant	NS	< .001	< .001	NS	NS	NS	NS
		Tibialis Anterior						
	EMG Integral	< .001	< .001	< .001	< .05	< .001	NS	NS
	Peak Amplitude	< .001	< .001	< .001	< .05	< .05	NS	NS
		Gastrocnemius						
	EMG Integral	< .05	NS	NS	NS	NS	NS	NS
	Peak Amplitude	< .01	< .001	< .001	NS	NS	NS	NS
		Soleus						
	EMG Integral	NS	NS	NS	NS	NS	NS	NS
	Peak Amplitude	< .05	< .05	< .05	NS	NS	NS	NS

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The effect can be derived from Fig 4 as well, showing a gradual decrease over time. Additionally, stronger lateral push forces resulted in higher intercepts (Fig 5A).

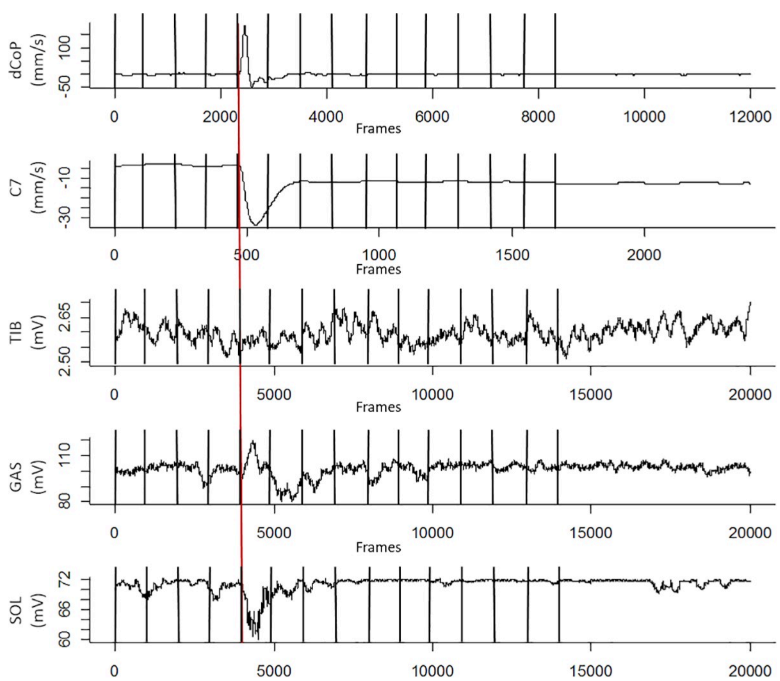

Fig 3. Illustrative data of one participant averaged time course over all conditions of sway (ML dCoP (mm/s)), the C7 marker (mm/s), and the muscle response of the Tibialis Anterior (mV), Gastrocnemius (mV) and Soleus (mV). The red line indicates the time of perturbation. Black vertical lines represent time bins of 1 second.

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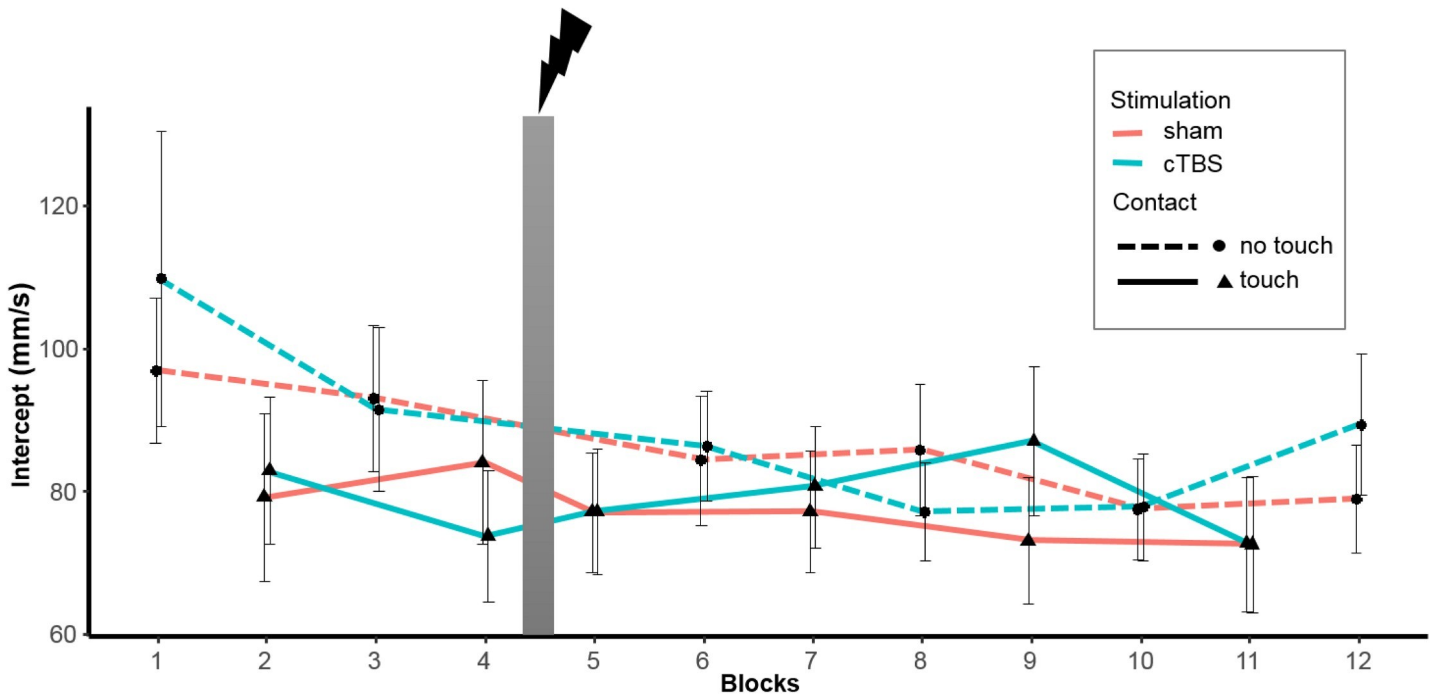

Fig 4. Progression of averaged intercept of the body sway at perturbation as a function of contact condition (Touch/No Touch) and stimulation protocol (sham/cTBS). Wide grey vertical line represents stimulation (Blocks left to it are pre-test, blocks right to it are post-test). Error bars indicate standard error.

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The compensation time constant was only affected by push force. Similar to the immediate effect of the perturbation on sway, steady-state asymptote was reduced with Light Touch Independently of the type of stimulation (Table 2). Stronger pushing forces lead to a more variable postural steady state as indicated by higher asymptotes (Fig 5B). Asymptote showed a decrease of 15% in both the 1% and 7% force push condition and 20% decrease in the 4% force push

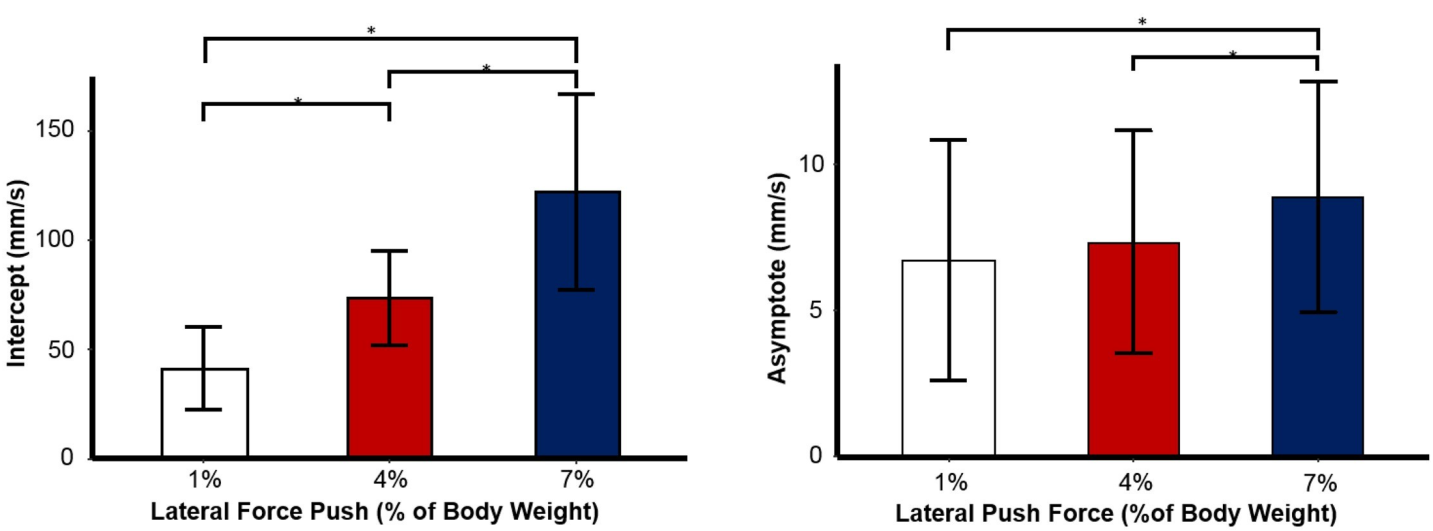

Fig 5. A) Averaged Intercept of the body sway at perturbation as a function of lateral push force (% of Body Weight). B) Averaged Asymptote of the body sway at perturbation as a function of lateral push force (% of Body Weight). Error bars indicate standard error.

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compared to the pre-test. In addition, the asymptote also showed an interaction between Light Touch and intra-session testing (Table 2). We see the highest value during no touch in the pre-test. Asymptote values decrease in the post test even without Light Touch. However, we also see that with Light Touch asymptote values are already decreased in the pre-test. Even though with Light Touch asymptote values do not decrease further compared to the pre-test, there is a significant difference between post-test levels ($p = .003$), with smaller asymptote values when utilizing Light Touch (Fig 6). Post hoc analysis revealed again a gradual decrease over time, independently whether Light Touch was established or not ($p < .001$) (Table 3).

EMG

Tibialis Anterior activity was affected by Light Touch and intra-session testing. Interactions between intra-session testing and stimulation protocol as well as between Light Touch and intra-session testing were found. General Tibialis Anterior activity decreased with the utilization of Light Touch. We saw that the highest level of general muscle activity (EMG integral) was expressed in the pre-test of the no touch condition, but decreased in the post-test. During the pre-test with Light Touch Tibialis Anterior activity already showed a lower level compared to no touch. Post hoc analysis of the two stimulation protocols revealed a significant effect of test (pre vs. post) for the Tibialis Anterior ($p < .001$) (Fig 7). Similar to the progression of sway we found gradual decrease of muscle activity over the progression of the 12 blocks (Figs 8 and 9). Post hoc test of the first four blocks before stimulation revealed no significant effect of stimulation session, showing that stimulation session is indeed an effect of the utilized stimulation rather than a general difference between sessions. Post hoc test did reveal a significant effect of Light Touch ($p < .001$) and Block ($p < .05$).

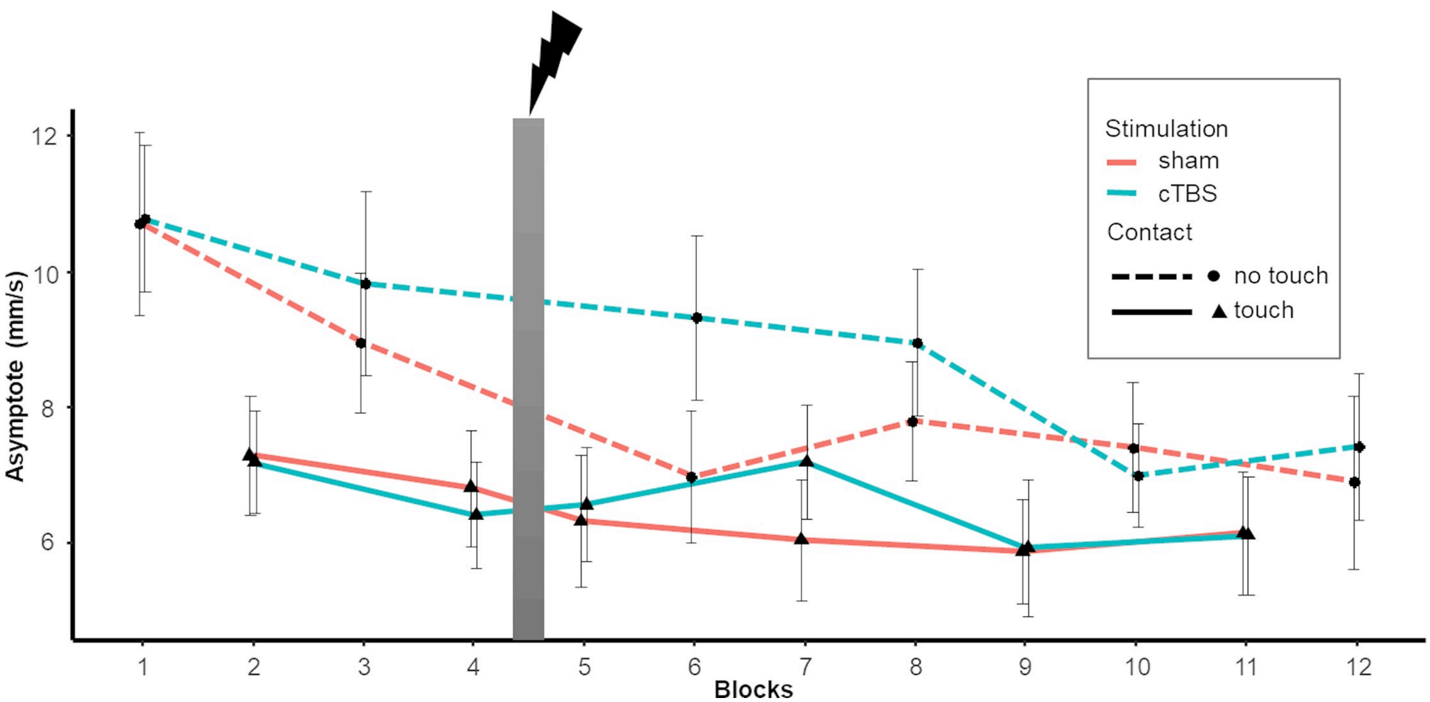

Fig 6. Progression of averaged asymptote of the body sway at perturbation as a function of contact condition (Touch/No Touch) and stimulation protocol (sham/cTBS). Wide grey vertical line represents stimulation (Blocks left to it are pre-test, blocks right to it are post-test). Error bars indicate standard error.

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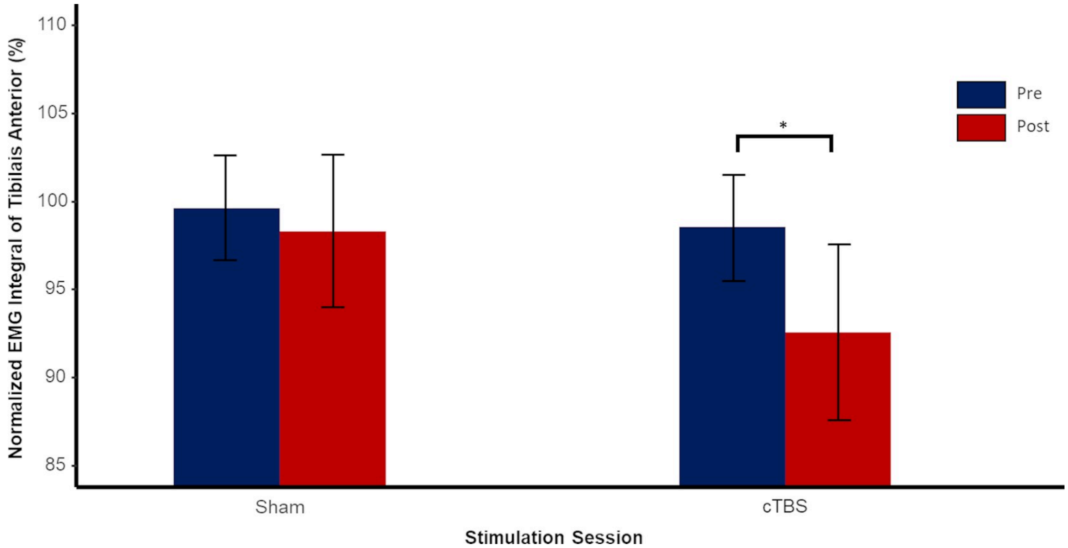

Fig 7. Normalized EMG Integral of Tibialis Anterior as a function of Test (Pre/Post) and stimulation protocol (sham/cTBS). Error bars indicate standard error.

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Looking at the decrease in percentages, we see that in the 1% and 7% force push condition EMG integral decreases 13% and 11% respectively, while the 4% force push condition shows a greater decrease with 16%. Interestingly, cTBS stimulation showed greater decreased levels of muscle activity of the Tibialis compared to sham. Following sham stimulation muscle activity is decreased by 11% but after cTBS we saw a decrease of 16%. As can be derived from Table 3 post hoc analysis showed a significant interaction of stimulation protocol and intra-session testing.

In terms of peak amplitude of muscle activity directly following the perturbation, Gastrocnemius, Tibilais and Soleus all showed lower peak activity amplitudes with Light Touch

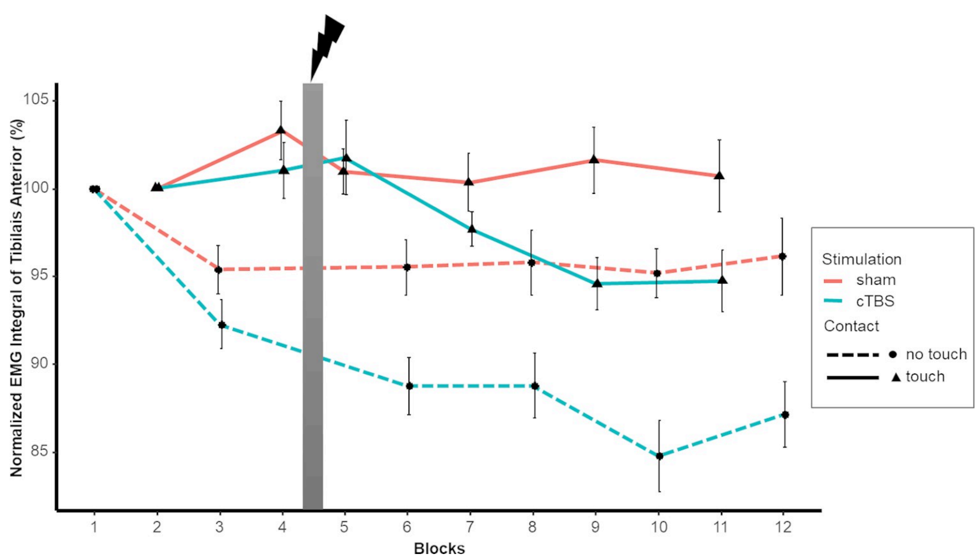

Fig 8. Normalized EMG Integral of Tibialis Anterior as a function of contact condition (Touch/No Touch) and stimulation protocol (sham/cTBS). Wide grey vertical line represents stimulation (Blocks left to it are pre-test, blocks right to it are post-test). Error bars indicate standard error.

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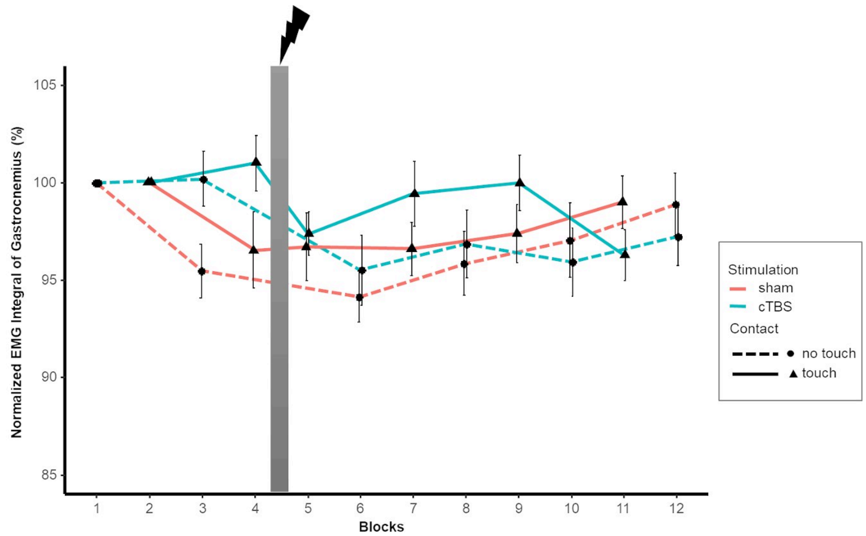

Fig 9. Normalized EMG Integral of Gastrocnemius as a function of contact condition (Touch/No Touch) and stimulation protocol (sham/cTBS). Wide grey vertical line represents stimulation (Blocks left to it are pre-test, blocks right to it are post-test). Error bars indicate standard error.

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compared to No Touch (Table 2). Finally, a significant interaction between stimulation protocol and intra-session testing was observed for peak amplitude of the Gastrocnemius. Post-hoc analysis showed a differences between stimulations protocols. There was a significant effect of test for Gastrocnemius $p < .01$ for the cTBS stimulation, while after sham no effects were found. Similar to the stimulation effects of the EMG integral, we see a decrease of peak activity after cTBS stimulation, while it stays the same after sham.

Discussion

Our study pursued two main objectives. The first was to investigate whether light fingertip contact improves balance compensation following a perturbation unpredictable in its relative force so that generation of a context-specific central postural set would be hindered. The second was to assess the role of the right posterior parietal cortex for the control of postural stiffness by disrupting the rPPC using continuous theta burst stimulation. We expected strong effects of light fingertip contact on body sway and muscle activations before, at and after a perturbation indicative of Light Touch feedback resulting in improved postural stability. Disruption of rPPC, on the other hand, was expected to hinder facilitation of sway stabilization with Light Touch but also affect the immediate response to a perturbation and sway stabilization by induced greater postural stiffness.

Facilitation of body sway control with light touch

Baseline sway before a perturbation was reduced by Light touch in line with previous studies assessing steady-state postural sway [1]. At the perturbation, Light Touch reduced the immediate response as well as the asymptotic post-perturbation steady state. In addition, activity of the Tibialis Anterior and Gastrocnemius was reduced with Light Touch. Similar results were found when investigating Light Touch benefits on balance stabilization following a sudden backward perturbation [5,6]. Light Touch led to smaller amplitudes of CoP displacement and decreased muscle activity of the Gastrocnemius. Martinelli et al. [5] argued that usually large body oscillations are prevented primarily through torque production around the ankles and that smaller displacement during Light Touch in return requires less muscle activation to produce smaller required correcting torque. Decreased general muscle activity (EMG Integral) in Tibialis Anterior across an entire perturbation trial agrees with this interpretation.

Against our expectations, Light Touch did not reduce the time constant of compensation following a perturbation. This observation contrasts with previous findings [4,5,6]. Johannsen and colleagues [4] observed shorter stabilization time constants with Light touch following both self-imposed as well as externally imposed perturbations. Similarly, Martinelli at al. [5] found reduced CoP sway during stabilization with Light Touch. However, their Light Touch effects for stabilization were limited to the most challenging conditions without vision while standing on a compliant surface. In all previous perturbation studies, that assessed the effect of augmented self-motion feedback with Light Touch, participants were tested in a normal bipedal stance posture with the perturbation in the antero-posterior direction [3,4,5,6]. In our present study, participants kept a tandem Romberg posture with a perturbation in the medio-lateral direction. Failed generalization of the Light Touch benefit to the time constant of balance stabilization in the context of the present study could indicate that the benefits of Light Touch for active stabilization could be highly context-specific. A central postural set represents the sensorimotor context of a postural task including the available sensory channels and current mechanical constraints [24]. Stance with Light Touch will also resemble a specific central postural set adjusted to the current task requirements such as the inclusion of a specific spatial frame of reference centred at the contacting finger or the trunk depending on the task [25,26]. If the postural context involves a balance perturbation, the task set will also represent the anticipated consequences of a known perturbation as well as any appropriate postural responses. For example, exposure to a sequence of horizontal support-surface perturbations with the same amplitude and velocity results in an appropriately scaled initial response of the agonist muscle, in contrast randomizing perturbations with respect to amplitude and velocity will result in a default response, partly determined by the strength the preceding perturbation [27]. In our current study, participants had to alternate between central postural sets with and without finger Light Touch in blocks of six trials each. Within each block the sequence of the perturbation forces was randomized and therefore unpredictable in its magnitude. The absence of any indications of Light Touch facilitation of dynamic stabilization in the current study implies a distinction between context-invariant or context-sensitive elements of a central postural set. Context-sensitive or rate-of-change-dependent components, such as an adequate compensation strategy following a perturbation, might have been excluded from the Light Touch central postural set or alternatively were impossible to implement due to the unpredictability of the experienced perturbations. It should be noted here that we did not find a direct influence of Light Touch in terms of shorter stabilization of the time constants. However, participants with a lower intercept but a constant time constant would reach their steady state sway earlier. In this regard, it might be possible that a strategy that even further decreases the time constant was deemed redundant, given that participants already reached their steady state faster.

Disruption of the rPPC did not interfere with the processing of fingertip haptic feedback for the stabilization of body sway following a perturbation. This confirms our previous study, where we showed that disruption of the rPPC did not affect the integration and utilization of Light Touch in a quiet stance context [17]. The present study generalizes this observation to more dynamic postural contexts involving external perturbations. This leaves us with a conundrum as the rPPC has been considered an important brain area that represents peri-personal space [28] and performs coordination transformation processes for mapping local tactile stimulation into hand-centered, head-centered, or trunk-centered spatial frames of reference [29,30]. Thus it seems likely that disruption of the rPPC does not alter the postural effects of

Light Touch sensory augmentation. As for the reason why, it is possible that a central postural set for the control of body sway with Light Touch makes use of more limb-cantered body representations without involvement of a predominantly spatial reference frame or egocentric representation. Dolgilevica and colleagues [31] proposed a conceptual framework which emphasizes the role of body representations such as the postural configuration of the body as well as the size and shape of body segments in the spatial localization of touch. In a previous study, we observed effector-specific differences between participants' dominant and non-dominant hand in terms of sway after-effects following sudden removal of a Light Touch reference [32]. The after-effect, that is the time to return to no touch baseline sway, was prolonged when the dominant hand was used to keep the Light Touch contact. As our participants were all right-handed, the observation implies that involvement of the left-hemisphere delayed switching between sets by keeping the Light Touch central postural set active for longer [32]. Thus, the control of body sway with Light Touch but without visual feedback may rely more on representations of somatotopy in the secondary somatosensory cortex [33] than representations of external space in the posterior parietal cortex.

Control of postural stabilization following the perturbation

In our previous cTBS study involving a quiet stance situation, we found that disruption of the right PPC leads to a decrease of the general sway variability [17]. We attributed this reduction in sway to a disrupted process for the continuous exploration of the body's postural state [34] resulting in reduced inhibition of a process controlling postural stiffness [34]. Therefore, we expected that the postural perturbation paradigm of the present study would provide us with more direct evidence of an increase in postural stiffness following disruption of the rPPC. For example, reduced body sway in a steady postural state as well as a more rigid response to the lateral push, such as a reduced immediate effect of the perturbation on body sway but a prolonged time constant of stabilization, could be indicative of increased postural stiffness with reduced flexibility. The influence of postural stiffness on compensation of a balance perturbation has previously been shown by Horak and colleagues [35] testing Parkinson's patients, whose rigidity has been lowered by levodopa replacement therapy. Following support-surface translations these participants expressed less resistance and faster Centre-of-Mass displacement.

Jacobs and Horak [7] assumed that contextual cues of an impending perturbation are used to optimize anticipatory postural adjustments. Based on that assumption, Smith et al. [36] analysed the effects of support translations on anticipatory postural adjustments testing how different amplitudes of support surface translations in combination with different cuing conditions influences optimization of anticipatory postural adjustments. Displacement amplitude was either cued by means of repetitive, blocked perturbations, or a random sequences of displacement amplitudes of uncued perturbations was delivered. In the blocked sequences, CoP under the feet showed a slower initial displacement following perturbations as compared to the random sequences. The authors interpreted the result as supporting the notion that postural control is optimized when contextual cues are given prior to the perturbation. The exposure to similar perturbations across trials in a block, however, may have induced optimization of postural responses by adaptive motor control processes and not through contextual cues alone [36]. Coelho et al. [37] investigated whether optimized postural responses are a result of contextual cuing or whether they are dependent on motor experience. They were able to show that block sequence of perturbations leads to the generation of more stable automatic postural responses in comparison to the serial and random perturbation sequences. During block sequence perturbation lower body sway amplitude, decreased displacement velocity and longer delays of activation onset of leg distal muscles were found. They interpreted these results as optimized postural responses in the block sequence due to adaptive processes underlying repetitive perturbations over trials rather than to processing of contextual cues [37]. To better understand how the postural control system adjusts postural responses following a specific type of perturbation, Kim et al. [38] exposed participants to forward trunk pushes of 5 different strengths in randomized order and estimated the gradual scaling of the sensory feedback gain. After comparing the observed feedback gain scaling to perturbations expressed following support surface translations [39], they concluded that the postural control system seems to select a feedback gain set according to the current postural context as characterised by the type of a perturbation and biomechanical constraints. Although Kim et al. [38] favoured a feedback gain interpretation, they could not exclude the possibility of situation-specific changes in dynamic parameters such as joint stiffness and damping.

In our present study we found results indicative of an adaptive process in terms of lower leg muscle activity and steady state sway, with a general decrease over time, independently whether Light Touch was used or not. This supports the idea that exposing people repetitively to a perturbation leads to an optimization of the postural response. Interestingly, this adaptive process was present although participants were perturbed to a randomized sequence of three different force pushes within one block. Given the range of the perturbations with a small, medium and strong force push, one possibility is that instead of finding three strategies against the perturbation force, the postural control systems settled for a compromise across the three forces and prepared for a medium configuration. If this were the case we would expect to see greater improvement, respectively greater decrease of muscle activity and postural sway in the medium force push condition. Looking at the decrease in percentages, this was the case. While in the small and strong force push condition we see a reduction in the EMG integral of the Tibialis of 13% and 11% respectively, the medium force push condition shows the highest decrease with 16%. Similar results can be found for the asymptote, with a decrease of 15% in both the small and strong force push condition and 20% decrease in the medium force push.

Unexpectedly, cTBS stimulation resulted in more decreased levels of activity of the Tibialis anterior and peak activity of the Gastrocnemius compared to sham stimulation. This observation contrasts with tonic activity of the Gastrocnemius, where activity stayed relatively the same over time, independently of the type of stimulation. Sozzi and colleagues [40] investigated the individual role of the lower leg muscles during standing in tandem Romberg stance and reported roles of the muscles specific to individual balancing functions. They concluded that while the soleus supports the body against gravity, the Tibialis Anterior and the peroneus stabilize the body in the medio-lateral direction. This supports our conclusion that the greater reduction in Tibialis anterior activity is tied to an improved postural adaptation following cTBS of the rPPC.

The decrease of muscle activity in the Tibialis Anterior should not be mistaken as a direct influence of the rPPC disruption on muscle activity, but rather as a result of a centrally mediated adaptation of postural control to the challenges of a perturbation. If we assume that reduced lower leg muscle activity indicates an experience-dependent optimization of the postural adjustments, then we can conclude that rPPC disruption enhanced anticipation of the disturbing effects of the perturbation. In Kaulmann et al. [17], we argued that rPPC may be involved in a process with generates postural sway to actively explore the postural stability state, which might normally interact with a postural stiffness control process in a reciprocal inhibitory manner. Thus, cancellation or disruption of a process represented in the rPPC for exploring the postural state might lead to a clearer feedback-dependent signal used for the prediction of the effects of an externally imposed external perturbation and the optimization of any compensatory responses.

There is ample evidence, however, that points to the role of brain areas other than the cerebral cortex in the adjustment of postural responses to external perturbations of balance. For example, Thach and Bastian [41] reported that the cerebellum is involved in the adaptation of response magnitude, as well as in the tuning of the coordination of postural responses based on practice and knowledge. This was in line with Horak and Diener [42], who demonstrated that patients with cerebellar lesions are unable to scale the magnitude of their postural responses to predicable amplitudes of surface translations. Also involvement of the basal ganglia in postural responses following external perturbations as illustrated by Parkinson's disease resulting in the inability to modify postural responses to a perturbation [43]. For example, healthy subjects are able to change postural synergies immediately after a single exposure, while individuals with Parkinson's disease require several trials to adjust their responses [44]. Thus, we do not claim that the rPPC is exclusively involved in the adaptation to a postural perturbation but that the region nevertheless resembles an important component of a network of brain regions controlling postural stiffness and adaptation.

Limitations

We have no direct indicator of the neural effect induced by cTBS stimulation at the target cortical area. Therefore, we cannot assume without reservation that cTBS did indeed cause local inhibition of the rPPC as the region, being primarily involved in sensorimotor integration for movement control, does not project directly to end-effector specific areas in the primary motor cortex that could have validated its effectiveness. Therefore, the evidence presented by our study for a role of the rPPC in the adaptation of postural responses to unpredictable perturbations must be considered as circumstantial only. A subsequent study needs to follow-up our observations by being more properly designed to evaluate sensorimotor learning of the perturbations and which validates the disruption of rPPC by cTBS using a different probe task, for example assessing visual attention.

Conclusion

We found a strong effect of Light Touch, which resulted in improved stability following an unpredictable perturbation. Light Touch decreased the immediate sway response, as well as the steady state sway following re-stabilization. Decreased sway is accompanied by reduced muscle activity of the ankle Tibilais Anterior. We assume that the improved sway response lead to increased stability, which required less torque production around the ankles in order to stabilize the body. However, we did not find an improvement of the time constant in response to the perturbation with Light Touch. This contrasts with studies that investigated the benefit of Light Touch when compensating a perturbation in the sagittal plane, while standing in normal bipedal stance. The lack of improvement might be a result of a different postural context or the unpredictability of the force of the perturbations. We observed a gradual decrease of muscle activity, which is indicative of an adaptive process in terms of lower leg muscle activity, following exposure to repetitive trials of perturbations. This supports the idea that exposing people repetitively to a perturbation leads to an optimization of the postural response. Given the range of the perturbations we suspect that the postural control system settled for a compromise across the three different perturbation forces and prepared for a medium configuration. This is supported by the notion that we see greater decrease of muscle activity in the medium force push condition. Regarding the effects of the disruption of the rPPC we were not able to confirm our hypothesis that disruption of the rPPC leads to increased postural stiffness. However, we did find an unexpected effect of cTBS stimulation in terms of improvements of the aforementioned adaptive process. After disruption of the rPPC muscle activity of the Tibialis

Anterior is decreased even greater, compared to sham. From that we can conclude that rPPC disruption enhanced the intra-session adaptation to the disturbing effects of the perturbation.

Supporting information

S1 File.
(DOCX)

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Writing – review & editing: Leif Johannsen.

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ORIGINAL RESEARCH

Deliberately Light Interpersonal Contact Affects the Control of Head Stability During Walking in Children and Adolescents With Cerebral Palsy

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Abstract

Objective: To evaluate the potential of deliberately light interpersonal touch (IPT) for reducing excessive head and trunk sway during self-paced walking in children and adolescents with cerebral palsy (CP).

Design: Quasi-experimental, proof-of-concept study with between-groups comparison.

Setting: Ambulant care facility, community center.

Participants: Children and adolescents (N=65), consisting of those with CP (spastic and ataxic, n=26; Gross Motor Function Classification System I–III; mean age, 9.8y; 11 girls, 15 boys) and those who were typically developed (TD, n=39; mean age, 10.0y; 23 girls, 16 boys).

Interventions: IPT applied by a therapist to locations at the back and the head.

Main Outcome Measures: As primary outcomes, head and trunk sway during self-paced walking were assessed by inertial measurement units. Secondary outcomes were average step length and gait speed.

Results: CP group: apex and occiput IPT reduced head velocity sway compared with thoracic IPT (both $P=.04$) irrespective of individuals' specific clinical symptoms. TD group: all testing conditions reduced head velocity sway compared with walking alone (all $P\leq.03$), as well as in apex and occiput IPT compared with paired walking (both $P\leq.02$).

Conclusions: Deliberately light IPT at the apex of the head alters control of head sway in children and adolescents with CP. The effect of IPT varies as a function of contact location and acts differently in TD individuals.

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Severe gait deficits in individuals with cerebral palsy (CP) lead to an increased fall risk, with disabilities in activities of daily living and reduced social participation.[1] During walking, the motion of the trunk as the heaviest segment of the body strongly affects the locomotor pattern and requires active balance control.[2] Individuals with CP show a severe gait disorder in combination with noticeable abnormalities in trunk motion, which may be a genuine deficit and specific cause for gait instability in CP.[3,4] Impaired gross motor function is associated with a greater thorax range of motion during walking in CP.[5] Heyrman et al[6] reported that children with spastic diplegia and only mildly impaired gross motor function still show increased lateral bending of the trunk during gait, while more severely impaired children demonstrate an increased motion amplitude in all 3 spatial planes.

Any trunk motion during walking will perturb head orientation and thus cause significant vestibular stimulation unless neck

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articulation minimizes head motion. Compensatory head-on-trunk articulation during walking primarily serves head stability.[7] Minimizing head motion may therefore be a major goal of the postural control system during walking in order to align the horizontal semicircular canals of the vestibular system to the earth horizontal for facilitating the integration of vestibular and visual information.[8]

It is an open question how trunk control can be improved in children with CP. Vision and vestibular feedback play an important role, but they are not the only afferent signals that can be used for locomotor control. Somatosensory afferences as well as proprioceptive feedback are also used for controlling the gait cycle and body balance.[9] A review by Pavão et al[10] indicated that there was a lack of research on the benefit of somatosensory feedback for balance control in individuals with CP.

Researchers have become increasingly interested in the effect of nonplantar light tactile feedback on body control when contacting an external reference. The effect of light touch during standing and walking has been described in several patient populations.[11] In addition to the single-person concept of haptic sensory augmentation, interpersonal touch (IPT) is a category of haptic interactions very relevant and frequently used in clinical situations. Deliberately light IPT results in reduced sway and increased coordination of trunk sway between 2 individuals during quiet standing as well as voluntary swaying.[12,13] IPT reduces sway in patients with chronic stroke as well as Parkinson disease.[14] More rostral IPT (at shoulder level) reduces sway to a greater amount than more caudal (low back) locations,[14] which is analogous to single-person effects of light touch on body sway.[15,16] The observation that more cranial IPT results in more reduced sway could be caused by a clearer signal resulting from a greater sway amplitude at the contact point. Alternatively, an increased resemblance between the haptic and vestibular signals could facilitate a more accurate stability state estimation.[17]

This proof-of-concept study aimed to investigate the effect of IPT on the control of trunk sway and gait during walking in children and adolescents with CP. To assess the effects of IPT on locomotion without confounding movement impairments caused by CP, we tested age-matched typically developed (TD) participants. We hypothesized that reinforcement of the head as an inertial guidance platform[8,18] by IPT at more rostral locations would benefit the control of head and trunk sway in participants with and without CP.

Methods

Participants

A convenience sample of 26 children and adolescents (mean age ± SD, 9.8±4.5y; mean height ± SD, 134±22cm; mean weight ± SD, 34.3±18.5kg) with CP were recruited at 3 therapeutic institutions (Schön Klinik Harlaching, München; Phoenix Pfennigparade, München; Petö Institute, Budapest). Participants with CP needed a Gross Motor Function Classification System (GMFCS)[18] level of III or higher to participate. Individuals were excluded if any other impairments were reported that could either affect locomotion or communication. Another convenience sample of 39 TD individuals (mean age ± SD, 10.0±4.4y; mean height ± SD, 144±25cm; mean weight ± SD, 38.5±17.5kg) were recruited from the community as a control group. Table 1 shows the demographic and clinical information of all participants. The study was approved by the medical ethical committee of the Technical University of Munich, and all participants or their guardians gave written informed consent.

List of abbreviations:

CP	**cerebral palsy**
GMFCS	**Gross Motor Function Classification System**
HVS	**head velocity sway**
IPT	**interpersonal touch**
TD	**typically developed**
TVS	**trunk velocity sway**

Experimental procedure

Each participant took part in a single 45-minute testing session. After demographic and medical data were collected, the child was familiarized with an inertial motion tracking system.[a] Four sensors of the system (60Hz) were fastened to both lower legs laterally, the sternum, and the forehead. After 2 practice trials, each participant walked at a self-chosen pace in a straight line for a distance of 10m between 2 measured floor markings, 6 times per testing condition. Participants were tested in 5 testing conditions in randomized order. IPT was applied by either a physical therapist or a conductor, who was trained in conductive education, in 3 conditions, while in the remaining 2 control conditions participants walked without IPT. The 5 testing conditions were as follows: (1) walking alone; (2) walking with the physical therapist/conductor peripherally visible (paired walking); (3) IPT on the thoracic spine (between the scapulae); (4) IPT below the occiput; and (5) IPT slightly dorsal of the apex of the head. An overview of the IPT locations is presented in figure 1A.

Data reduction

Orientation of the inertial sensors in all 3 planes was processed unfiltered by a custom processing toolbox in Matlab (2014a).[b] Phases of steady-state walking were extracted by manually segmenting trials based on sensor data from the dominant leg to exclude turning points, gait initiation, and stopping from analysis. Gait speed and average step length were determined by dividing the walking distance by the time needed to cover it and the number of all steps detected during this period.

Head velocilty sway (HVS) and trunk velocity sway (TVS) were measured as the SD of the angular velocity of the respective sensor's orientation. To prevent angular flip-overs between $-180°$ and $180°$ from distorting the velocity sway measure, sensor orientation angles were cosine-transformed before differentiation ($\cos(\alpha)/s$; fig 1B). A direction-unspecific velocity sway measure was calculated for each sensor by taking the square root of the sum of squares of the velocity sway on each of the 3 axes of a sensor.

Statistical analysis

Statistical analysis was performed in IBM SPSS statistics 23.[c] All extracted parameters (gait speed, step length, HVS, TVS) were statistically analyzed using a mixed 2-factorial repeated-measures analysis of variance, with group as the between-subject factor (2 levels: CP vs TD participants) and testing condition as the within-subject factor (5 levels). Because of the participants' range in demographic parameters such as age, height, and weight, we used independent *t* tests as well as chi-square tests to assess differences in the sample averages and distributions between both participant

Table 1 Demographic and clinical information of all participants

Group	Participant	Age (y)	Height (cm)	Weight (kg)	Sex	Dominance	GMFCS	Symptom I*	Symptom II†
TD	1	14	175	60	M	R	NA	NA	NA
TD	2	11	149	37	F	R	NA	NA	NA
TD	3	13	160	52	M	L	NA	NA	NA
TD	4	15	186	68	M	L	NA	NA	NA
TD	5	17	169	53	F	R	NA	NA	NA
TD	6	11	149	41	F	L	NA	NA	NA
TD	7	13	165	58	F	R	NA	NA	NA
TD	8	9	146	32	F	R	NA	NA	NA
TD	9	6	126	25	F	R	NA	NA	NA
TD	10	6	126	26	F	R	NA	NA	NA
TD	11	9	151	42	F	R	NA	NA	NA
TD	12	7	123	25	M	R	NA	NA	NA
TD	13	8	137	35	F	R	NA	NA	NA
TD	14	11	159	38	F	L	NA	NA	NA
TD	15	14	170	50	M	R	NA	NA	NA
TD	16	9	140	30	M	R	NA	NA	NA
TD	17	8	128	22	F	R	NA	NA	NA
TD	18	12	152	46	M	R	NA	NA	NA
TD	19	11	148	38	F	R	NA	NA	NA
TD	20	5	111.5	20	M	R	NA	NA	NA
TD	21	17	176	63	F	R	NA	NA	NA
TD	22	12	180	50	M	L	NA	NA	NA
TD	23	13	165	46	F	R	NA	NA	NA
TD	24	11	150	44	M	R	NA	NA	NA
TD	25	10	148	37	M	R	NA	NA	NA
TD	26	13	166	59	F	R	NA	NA	NA
TD	27	4	110	18	M	R	NA	NA	NA
TD	28	17	188	83	M	R	NA	NA	NA
TD	29	18	170	60	F	R	NA	NA	NA
TD	30	8	130	28	F	R	NA	NA	NA
TD	31	5	116	22	F	R	NA	NA	NA
TD	32	19	174	65	M	L	NA	NA	NA
TD	33	6	107	16	F	R	NA	NA	NA
TD	34	3	100	17	M	L	NA	NA	NA
TD	35	6	120	20	F	R	NA	NA	NA
TD	36	4	108	21	F	L	NA	NA	NA
TD	37	6	119.5	20	F	R	NA	NA	NA
TD	38	6	124	17	F	R	NA	NA	NA
TD	39	4	102	16	M	L	NA	NA	NA
CP	1	7	116	17	M	R	I	4	2
CP	2	6	116	26	F	NA	III	4	2
CP	3	4	111	19	M	L	II	1	1
CP	4	6	118	18	F	R	I	1	1
CP	5	7	113	18	F	R	II	4	1
CP	6	4	107	15	F	R	II	2	1
CP	7	6	110	17	M	L	II	2	1
CP	8	6	121	26	F	L	I	1	1
CP	9	5	99	15	M	R	II	4	2
CP	10	12	145	43	F	NA	II	2	1
CP	11	10	141	44	F	L	II	2	1
CP	12	8	119	22	M	R	III	2	1
CP	13	9	139	27	F	NA	II	4	2
CP	14	14	162	44	M	L	II	3	1
CP	15	10	145	56	F	L	I	2	1
CP	16	12	141	29	M	L	III	2	1
CP	17	9	135	34	M	L	I	2	1
CP	18	13	164	61	M	L	I	2	1

(*continued on next page*)

Table 1 (*continued*)

Group	Participant	Age (y)	Height (cm)	Weight (kg)	Sex	Dominance	GMFCS	Symptom I*	Symptom II†
CP	19	10	145	38	M	R	II	2	1
CP	20	18	159	51	F	R	I	4	1
CP	21	8	112	20	F	L	II	2	1
CP	22	7	110	19	M	NA	III	2	1
CP	23	12	150	39	M	R	II	3	1
CP	24	19	171	84	M	R	I	1	1
CP	25	18	172	71	M	L	II	1	1
CP	26	18	163	38	M	L	II	2	1

Abbreviations: F, female; L, left; M, male; NA, not available; R, right.
* Symptom I: 1, unilateral; 2, bilateral leg; 3, bilateral arm; 4, bilateral complete.
† Symptom II: 1, spastic; 2, ataxic.

groups. The TD group tended to be taller by about 10cm ($t_{63}=1.70$, $P=.09$; $\chi^2_3=8.25$, $P=.04$). Therefore, we included height as a covariate in all analyses encompassing a comparison between both groups. Greenhouse-Geisser−corrected P values were used as a conservative statistical criterion. Level of significance was set to $P=.05$. Bonferroni-corrected post hoc comparisons between conditions were conducted as appropriate to resolve interactions between group and testing condition.

Additional statistical analyses were performed between subgroups of the CP participants according to GMFCS level (I/II/III) and impairment categorizations (spastic/ataxic; plegia: unilateral/bilateral leg/bilateral arm/bilateral complete). No differences between subgroups of the individuals with CP were found with respect to age, height, or weight with the exception that the individuals with ataxic CP were numerically younger and shorter (both $P\geq.11$).

Results

Gait speed and stride duration

Spontaneous gait speed was slower in the CP group (mean ± SD, 1.03±.29m/s; $F_{1,63}=13.60$, $P=.001$, partial $\eta^2=.19$) than in the TD group (mean ± SD, 1.32±.26m/s). An interaction between group and testing condition was found ($F_{4,252}=15.36$, $P<.001$, partial $\eta^2=.21$). In the CP group, the participants did not change their gait speed in any of the testing conditions. In contrast, the TD group walked slower in all 4 conditions compared with walking alone (mean ± SD, 1.41±.27m/s; all $P\leq.002$). Gait speed was still slower in occiput IPT (mean ± SD, 1.25±.26m/s) compared with thoracic IPT (mean ± SD, 1.30±.26m/s) and paired walking (mean ± SD, 1.34±.27m/s; both $P\leq.02$).

Average step length was shorter in the CP group (mean ± SD, 50±10cm; $F_{1,63}=13.84$, $P<.001$, partial $\eta^2=.20$) compared with the TD group (mean ± SD, 62±11cm). We also found an interaction between the group and testing condition ($F_{4,252}=9.30$, $P<.001$, partial $\eta^2=.14$). While no differences between testing conditions were found for the CP group, in the TD group step length was shorter in all 4 test conditions involving the physical therapist/conductor compared with walking alone (mean ± SD, 65±11cm; all $P\leq.03$). Thoracic (mean ± SD, 60±12cm) and occiput IPT (mean ± SD, 59±12cm) showed still shorter step length relative to paired walking (mean ± SD, 63±12cm; both $P\leq.006$).

For step length and gait speed, no general differences between subgroups or interactions with the testing condition were found for the subdivisions of the participants with CP. Exceptions were GMFCS level I tending to show the fastest gait speed (mean ± SD, 1.17±.27m/s), followed by level II (mean ± SD, 1.02±.22m/s) and level III (mean ± SD, .82±.41m/s; $F_{2,23}=2.52$, $P=.10$, partial $\eta^2=.19$).

Head and trunk velocity sway

HVS was greater in the CP participants ($F_{1,63}=15.98$, $P<.001$, partial $\eta^2\geq.21$) compared with the TD group (fig 2A). TVS only tended to be greater in the CP participants than the TD group ($F_{1,63}\geq3.04$, $P=.09$, partial $\eta^2\geq.05$) (fig 2B). For HVS and TVS, interactions were found between group and testing condition (both $F_{4,252}\geq3.54$, both $P\leq.03$, both partial $\eta^2\geq.06$). In the CP group, HVS was reduced in the occiput and apex IPT conditions compared with thoracic contact (both $P\leq.04$). Concerning the trunk, the thoracic IPT condition tended to show more TVS than apex IPT ($P=.06$). In the TD group, all other conditions showed less HVS compared with walking alone (all $P\leq.03$). In addition, occiput and apex IPT were still lower than paired walking (both $P\leq.02$). For the trunk, both apex and thoracic IPT tended to show lower TVS compared with walking alone (both $P\leq.09$).

The CP subgroups differed in terms of HVS, but no interactions between testing conditions and subgroups were found for either HVS or TVS. As an exception, an effect of GMFCS level on TVS was present ($F_{2,23}=3.60$, $P=.05$, partial $\eta^2=.25$). The participants with GMFCS level III showed the most variable TVS (mean ± SD, .45±.15), followed by level II (mean ± SD, .29±.17) and level I (mean ± SD, .21±.15).

Discussion

We aimed to investigate whether IPT at the head is a way to facilitate the control of body sway during walking in children and adolescents with CP and with typical development. The effect of IPT was assessed in terms of step length and gait speed as well as head and trunk velocity sway. In general, the CP and TD groups differed in gait speed and average step length. The TD group walked faster with longer average steps and less head and trunk velocity sway than the CP group. This is not unexpected since it is well known that individuals with CP show reduced gait speed with longer stride duration and increased postural instability.

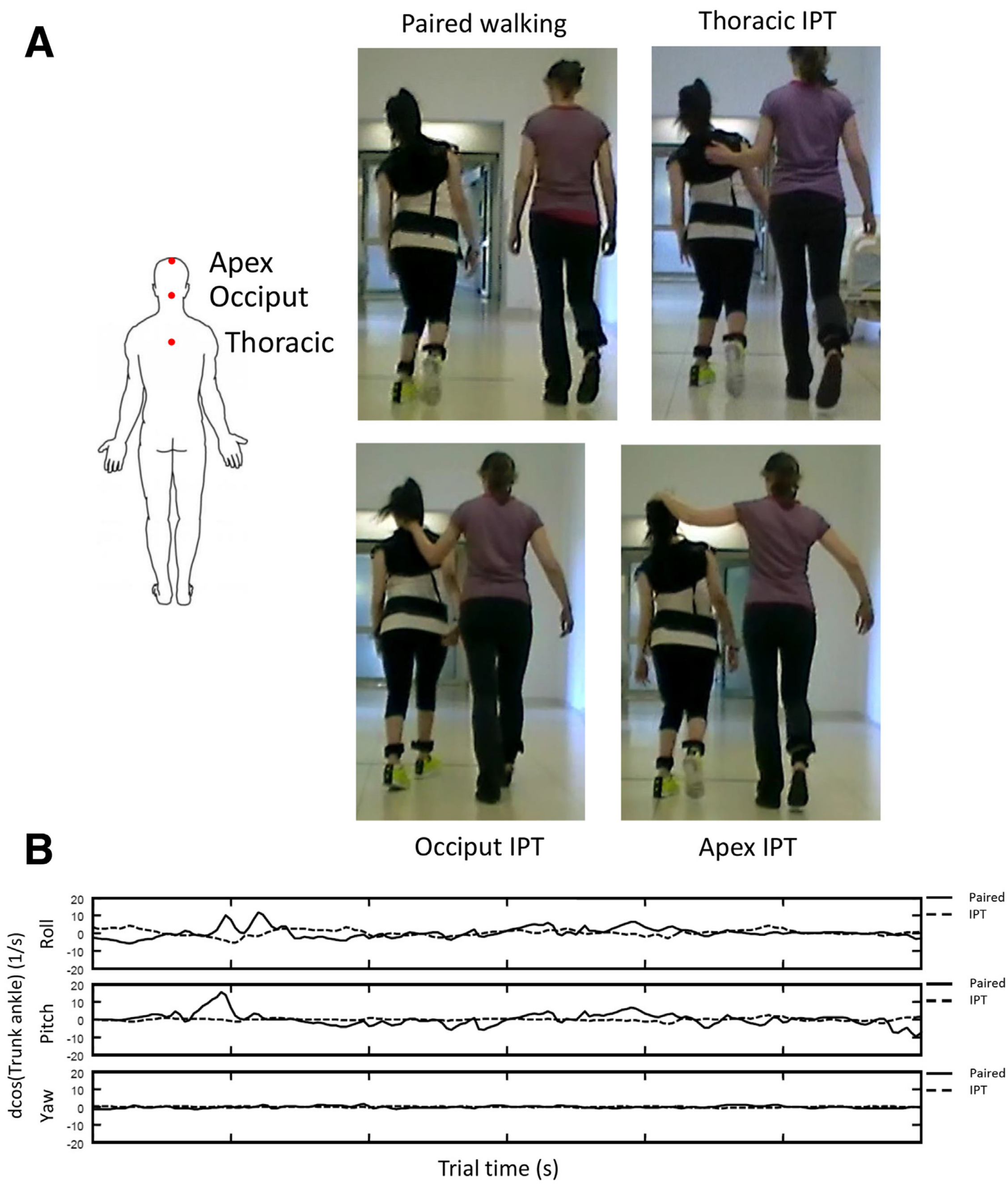

Fig 1 (A) Four of the 5 testing conditions demonstrated on an individual with CP (left) by a therapist (right). Deliberately light IPT was provided to 3 contact locations: thoracic, occiput, and apex (experimental conditions; control condition: paired walking). The individual with CP is wearing trunk and pelvis parts of an inertial measurement unit sensor suit (not a thoracolumbosacral orthosis). (B) Illustrative inertial measurement unit sensor traces of a single CP participant. The 3 panels show transformed trunk angular velocity around a sensor's roll, pitch, and yaw axes for paired walking (straight line) and thoracic IPT (dashed line). To prevent angular flip-overs between $-180°$ and $180°$ from distorting the variability measure, sensor orientation angles were cosine-transformed before differentiation ($\cos(\alpha)/s$). Abbreviation: dcos, differentiated cosine.

Although our results did not exactly turn out as hypothesized, our study yielded some interesting findings. The participants with CP showed less HVS with apex and occiput IPT in contrast to thoracic IPT. Numerically, these 2 conditions tended to differ from the 2 control conditions (walking alone, paired walking) in opposite directions, with reduced HVS during apex IPT. Nevertheless, it shows that the location at which IPT is applied to the receiver's body does matter in CP. In contrast, the TD group showed the lowest HVS in occiput and apex IPT compared with both walking alone and paired walking. Further, while the CP group did not walk with measurably changed speed, the TD group walked with reduced speed by taking shorter average steps in the IPT conditions.

We assumed that IPT at the head facilitates the role of the head as an inertial guidance platform for locomotion, improves control of trunk sway, and optimizes gait in CP. In this respect, only the TD group behaved in correspondence with our expectations. The TD group showed the least HVS in both head contact conditions

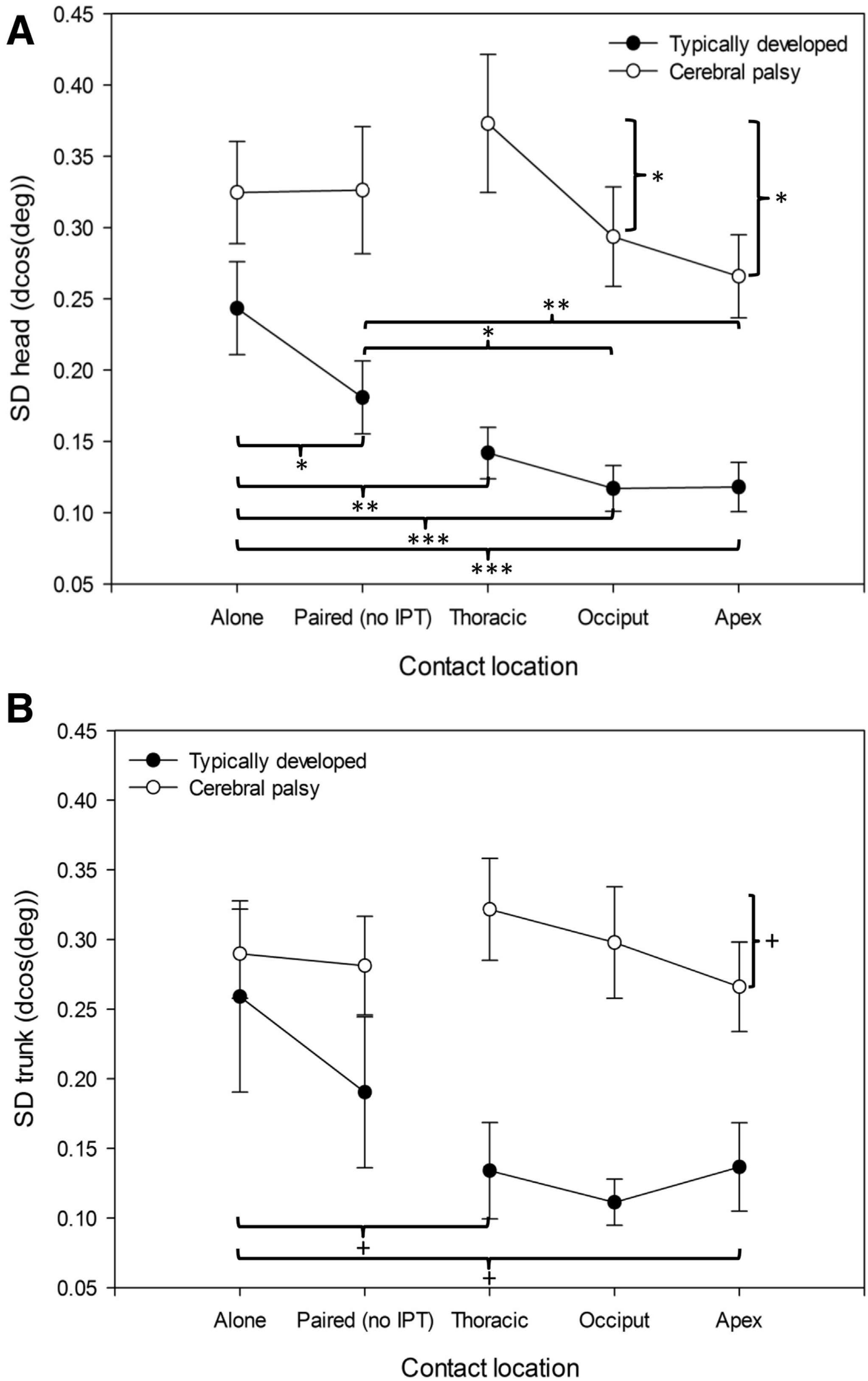

Fig 2 The average head (A) and trunk (B) velocity sway as a function of testing condition and group, expressed as the resultant, direction-unspecific SD of the angular velocity of the respective sensor. Error bars represent the SEM. Brackets and asterisks indicate statistically significant differences ($^{+}P<.10$; $^{*}P<.05$; $^{**}P<.01$; $^{***}P<.001$) between testing conditions (experimental conditions: thoracic, occiput, and apex; control conditions: alone and paired walking).

and a small corresponding reduction in TVS. This indicates that the control of head sway became more influenced by a headcentric sensory signal compared with thoracic IPT or walking without IPT.

The CP group did not demonstrate any effect of the presence of the physical therapist/conductor. In contrast, the TD participants reduced HVS during paired walking, which may be the result of some form of "social facilitation," perhaps by some form of spontaneous interpersonal entrainment of the stepping pattern between the physical therapist/conductor and participant. The difference between the groups could mean that the CP group was insensitive to or unable to comply with the social demands and constraints of interpersonal coordination.

With respect to human ontogenetic locomotor development, it was proposed that selective control of the neck's movement degrees of freedom is a key feature of a mature upper body gait pattern.[19] Wallard et al[20] observed an "en bloc" head-on-trunk strategy with increased head angle variability in the frontal plane during walking in children with CP, and proposed that it might express an "en bloc" compensatory strategy by deliberate reduction of the neck's movement degrees of freedom. Because we found subtle effects of apex IPT in the CP group, we speculate that apex IPT may still be a therapeutic approach to open up a habitual "en bloc" strategy and to enable the exploration of neck articulation as well as the benefits of actively stabilized head orientation. Advocates of a "hands-off" approach[21] emphasize unrestricted self-exploration of the movement repertoire by the patient. We perceive deliberately light IPT as a married form between "hands-on" and "hands-off" because of the low contact forces involved and the absence of active restriction. The "guidance" in IPT is considered less physical but more implicit to the social context.

We did not find any differences between symptom subgroups among the participants with CP, which indicated that differences in symptoms did not alter the susceptibility to IPT and its social context. Visual inspection of our data showed that the responsiveness of the individuals with CP showed a high degree of interindividual variability. Since only 2 IPT providers were involved in data collection, it is unlikely that variability in the way IPT was applied caused this. Instead, factors within the individuals with CP must be the reason—for example, current motor competence in the control of trunk sway and neck articulation. The observation that more impaired individuals with CP, as indicated by their GMFCS level, performed worse was to be expected. It shows, however, that the capacity to respond to IPT is not determined by the general impairment level.

Study limitations

It might appear as a limitation, that the sway variability measures used in our study do not represent positional variability. Variability of angular velocity, however, is more closely related to the control of body balance during locomotion. Differentiation of a signal acts as a high-pass filter, which removes low-frequency drift, which could occur in the absence of any positional control. For example, Allum and Carpenter[22] recommended measurements of trunk angular velocity as a means to differentiate between specific control deficits of body balance.

We did not restrict our recruitment to participants with CP showing specific symptoms, although this could have made our results more generalizable for this symptom subgroup. Our intention was to evaluate the general feasibility of IPT in a wide spectrum of symptoms. The present study aimed to advance the understanding of the "mechanisms of action" of IPT for balance support during walking in individuals with CP, and thus was designed as a single-session, proof-of-concept study. The long-term benefits of deliberately light IPT during locomotor training in CP remain speculative at this point and therefore require a properly designed multisession intervention study.

Conclusions

Deliberately light interpersonal contact applied to the apex of the head results in a reduction of HVS compared with thoracic IPT during walking in children and adolescents with CP, irrespective of their symptoms. This implies that the effect of IPT depends on the location at which it is applied in individuals with CP. The CP group, however, did not act in the same way as the TD group. TD individuals were much more responsive in terms of reductions in HVS because of the presence of the therapist and the application of IPT. The difference may be an expression of reduced sensitivity regarding the social affordances of the IPT situation in individuals with CP, which could indicate a restriction of the ability to adapt behavior to external social conditions. Further research is still required to assess any longer-term benefits of IPT in individuals with CP.

Suppliers

a. Xsens MTw; Xsens Technologies BV.
b. Matlab (2014a); MathWorks, Inc.
c. IBM SPSS statistics 23; IBM Corp.

Keywords

Cerebral palsy; Locomotion; Postural balance; Rehabilitation; Therapeutic touch

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Full length article

Interpersonal interactions for haptic guidance during maximum forward reaching

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ABSTRACT

Caregiver–patient interactions rely on interpersonal coordination (IPC) involving the haptic and visual modalities. We investigated in healthy individuals spontaneous IPC during joint maximum forward reaching. A 'contact-provider' (CP; n = 2) kept light interpersonal touch (IPT) laterally with the wrist of the extended arm of a forward reaching, blind-folded 'contact-receiver' (CR; n = 22). Due to the stance configuration, CP was intrinsically more stable. CR received haptic feedback during forward reaching in two ways: (1) presence of a light object (OBT) at the fingertips, (2) provision of IPT. CP delivered IPT with or without vision or tracked manually with vision but without IPT. CR's variabilities of Centre-of-Pressure velocity (CoP) and wrist velocity, interpersonal cross-correlations and time lags served as outcome variables. OBT presence increased CR's reaching amplitude and reduced postural variability in the reach end-state. CR's variability was lowest when CP applied IPT without vision. OBT decreased the strength of IPC. Correlation time lags indicated that CP retained a predominantly reactive mode with CR taking the lead. When CP had no vision, presumably preventing an effect of visual dominance, OBT presence made a qualitative difference: with OBT absent, CP was leading CR. This observation might indicate a switch in CR's coordinative strategy by attending mainly to CP's haptic 'anchor'. Our paradigm implies that in clinical settings the sensorimotor states of both interacting partners need to be considered. We speculate that haptic guidance by a caregiver is more effective when IPT resembles the only link between both partners.

1. Introduction

Balance control requires successful integration of self-motion information from multiple sensory modalities [1]. The human postural control system is able to derive self-motion not only from its primary motion detectors but also from actively acquired or passively received light skin contact with the environment [2,3]. Haptic information also stabilizes quiet stance when it originates from a non-weight-bearing contact that possesses motion dynamics of its own, i.e. another human (interpersonal touch; IPT) [4]. Deliberately light IPT is intended to involve small forces only, in order to minimize the mechanical coupling and to maximize the informational exchange [5]. Sway reductions with IPT may emerge from mechanically and informationally coupled adaptive processes and responsiveness in both partners [5].

When joint action partners coordinate their movements they may share information but also face differences in task-relevant knowledge and roles. For example, a blind person receives tactile, visual or verbal cues from the guiding partner. Spontaneous interpersonal postural coordination (IPC) has been demonstrated in diverse joint tasks [6]. For example, implicit observation of a partner in a joint precision task improved manual performance as well as IPC [7]. Verbal communication in a joint problem solving task also influences IPC regardless of whether visual information about the partner was available [8], perhaps mediated by shared speaking patterns [9]. Finally, haptic interactions provide powerful sensory cues for IPC [10]. Coordinative processes supporting goal-directed joint action can result in the emergence of spontaneous leader-follower relationships, for example in a visual, periodic collision avoidance task [11]. In situations such as quiet stance IPT, however, no clear leader-follower relationship has been reported, also not in situations with asymmetrical stance postures with one person intrinsically more stable than the partner [4,12,13].

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A well-established clinical task to assess body balance control is the Functional Reach (FR) [14]. Maximum forward reaching (MFR) challenges the control of body sway as the body's Centre-of-Mass (CoM) approaches the physical limits of stability so that the likelihood of balance loss increases with reaching distance [15]. We assumed that joint action in an asymmetric interpersonal postural context, such as the MFR task with one partner more intrinsically stable, would be more adequate than quiet stance to investigate spontaneously emerging leader-follower relationships. According to the ecological principles of interpersonal affordances [16], we aimed to create dependencies between two individuals by asymmetries in the intrinsic postural stability and in the knowledge of the joint postural state based on the available sensory feedback. We expected that additional haptic feedback, for example as either an additional object or IPT, would increase reach distance but also stabilize body sway in the reaching person (contact-receiver; CR). Further, we anticipated that spontaneous IPC, specifically the leader-follower relationship, is altered by the haptic feedback available to CR as well as by the visual feedback available and the instructions given to the person providing IPT (contact-provider; CP). Although CR would be the main actor performing the MFR, we assumed that CR would become more dependent on CP, when CP was able to perceive the scene.

2. Methods

2.1. Participants

Twenty-two healthy participants (average age = 26.3 yrs, SD = 4.1; 17 females and 5 males; all right-handed for writing) were tested. Participants with any neurological or orthopaedic indications were excluded. Two naïve, healthy young adults provided IPT

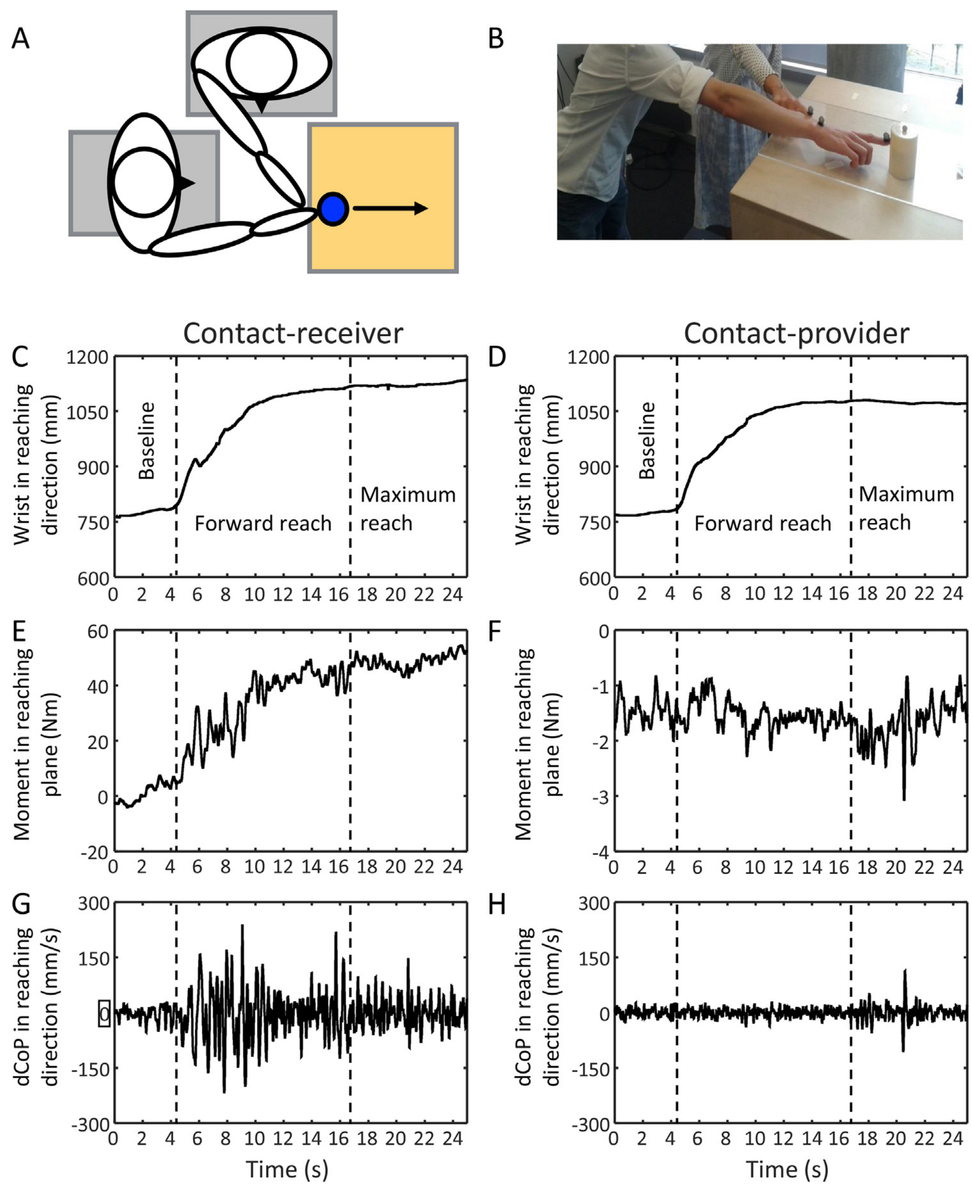

Fig. 1. (A) The stance configuration of the experimental setup at the beginning of a trial. Upon a signal by the experimenter the contact receiver will start the forward reach pushing the object as far out as possible. (B) The contact provider keeping light contact with the receiver's wrist. (C) Position of a receiver's wrist in the reaching direction across single trial. The dashed lines indicate the beginning and end of the forward reach phase. (D) Position of a providers's wrist in the reaching direction across the same trial. (E) Moment in the plane parallel to the reaching direction exerted by the receiver. (F) Corresponding moment exerted by the provider. (G) Receiver's Centre-of-Pressure (CoP) velocity in the reaching direction. (H) Corresponding CoP velocity of the provider.

to all CRs. Participants were recruited as an opportunity sample from students of the university. The study was approved by the local ethical committee and all participants gave written informed consent.

2.2. Experimental procedure

Six conditions were combined from the task requirements imposed on CR and CP. CR stood blindfolded on a force plate in bipedal stance to perform MFR with or without tactile feedback at the fingertips by touching a light object (OBT; weight = 59.3 g). CR was instructed to reach as far forward as possible or asked to shove OBT instead, which was placed upon a fibreglass plate (kinetic coefficient of friction = 0.33). OBT could move in any direction and therefore afforded manual precision. Before the start of a trial, CR was instructed to stand in a relaxed manner, the dominant right arm extended at shoulder height to reach horizontally above a table. The table was adjusted to each individual to avoid surface contact.

CP stood orthogonally to CR in bipedal stance on a force plate placed ahead of CR in the reaching direction (Fig. 1a) and provided light IPT during CR's reach with the right extended index finger contacting CR's medial wrist (Fig. 1b). The visuotactile interpersonal context (VIC) consisted of three conditions: IPT with open or closed eyes and CP tracking the motion of CR's wrist with the extended index finger visually but without IPT. Before the start of a single trial, CP kept his contacting finger close to the wrist of CR waiting for the specific task instructions.

Each condition was assessed in blocks of 10 trials for a total of 60 trials in fully randomized order. A single trial lasted 25 s consisting of three phases: baseline (5 s static posture), self-paced forward reaching (cued by experimenter) and reach end-state (static posture until trial end).

Two force plates (Bertec 4060H, OH, USA; 600 Hz) oriented in parallel measured both individuals' six components of the ground reaction forces and moments to calculate anteroposterior (AP) and mediolateral (ML) components of the Centre-of-Pressure (CoP). In addition, a four-camera motion capture system (Qualisys, Göteborg, Sweden; 120 Hz) tracked markers on both individuals at the following locations: right index finger, right wrist, left and right shoulders, 7th cervical segment.

2.3. Data reduction and statistical analysis

Motion data were spline interpolated to 600 Hz and subsequently merged with the kinetic data. Time series data were smoothed using a generic dual-pass, 4th-order Butterworth lowpass filter (cutoff = 10 Hz). After differentiation, trials were segmented into three movement phases based on the AP position of CR's wrist marker (Fig. 1c). Reach onset was determined as the first frame that exceeded 4 standard deviations of wrist position within the initial 3 s. Stop of forward reaching was determined as the velocity zero-crossing closest to 95% of the absolute maximum reach distance. Reach performance was analysed in the horizontal plane. Average reach amplitude, direction, curvature (normalized path length = path length/straight line length) of the trajectory from baseline position to maximum reaching end-state as well as the average and standard deviation of reaching velocity were extracted. Velocity information is the predominant source for body sway control [17], therefore postural control in the maximum reach end-state was extracted as the standard deviation of CoP velocity (SD dCoP) in both directions (Fig. 1g). Similarly, standard deviation of the wrist velocity (SD dWrist) expressed reaching stability and precision in both directions. For each phase, IPC was estimated in terms of the cross-correlation function (time lag range: ±3 s) between both participants' moments as recorded by the force plates in the plane parallel to the reaching direction (Fig. 1e–f). The largest absolute cross-correlation coefficient and corresponding time lag were extracted. Coefficients were Fisher Z-transformed for statistical analysis. Two-factorial repeated

Table 1
Statistical effect table. OBT: light object; IPT: interpersonal touch; ML: mediolateral; AP: anteroposterior; n.s.: not significant; Italics: marginal significance. P-values are rounded to two or three decimals respectively.

	Condition		Presence of OBT	Visuotactile interpersonal context	Interaction between OBT and visuotactile interpersonal context
	Interpersonal contact		No IPT	IPT	IPT
Trial phase	Parameter		$F_{1,21}$; p; partial η^2	$F_{2,42}$; p; partial η^2	$F_{2,42}$; p; partial η^2
Reaching performance					
Forward reaching	Horizontal amplitude		4.80; 0.04; 0.19	n.s	n.s
	Directional angle		n.s	n.s	n.s
	Horizontal velocity		19.67; 0.001; 0.48	n.s	n.s
	Variability of horizontal velocity		12.87; 0.002; 0.38	n.s	n.s
	Curvature		n.s	n.s	n.s
Control of body balance and posture					
Reach end-state	Variability of wrist velocity	ML	n.s	n.s	n.s
		AP	14.56; 0.001; 0.41	n.s	3.59; 0.04; 0.15
	Variability of CoP velocity	ML	36.50; 0.001; 0.64	n.s	n.s
		AP	13.65; 0.001; 0.39	*2.95; 0.06; 0.12*	n.s
Interpersonal postural coordination					
Complete trial	AP wrist	Coefficient	4.49; 0.05; 0.18	11.64; 0.001; 0.36	n.s
		Time lag	n.s	n.s	n.s
	AP moment	Coefficient	6.45; 0.02; 0.24	n.s	n.s
		Time lag	n.s	n.s	3.84; 0.03; 0.15
Forward reaching	AP wrist	Coefficient	6.75; 0.02; 0.24	10.40; 0.001; 0.33	n.s
		Time lag	n.s	5.34; 0.01; 0.20	3.55; 0.05; 0.15
	AP moment	Coefficient	13.21; 0.002; 0.39	n.s	n.s
		Time lag	n.s	n.s	n.s
Reach end-state	AP wrist	Coefficient	n.s	9.69; 0.001; 0.32	n.s
		Time lag	4.25; 0.05; 0.17	n.s	n.s
	AP moment	Coefficient	n.s	4.63.; 0.02; 0.18	n.s
		Time lag	n.s	n.s	n.s

measures ANOVAs with OBT (2 levels) and VIC (3 levels) as within-subject factors were calculated. Significant findings were detected at a Greenhouse-Geisser-corrected $p < 0.05$.

3. Results

Table 1 presents the statistical results for all extracted parameters.

3.1. Forward reaching performance

Fig. 2a shows the amplitude of CR's reach as a function of the VIC and OBT presence. Without OBT the amplitude of reaching was 37.9 cm (SD = 7.0). OBT increased reach distance to 38.9 cm (SD = 6.5). The average reach direction indicated a slight medial deviation of 5.9° (SD = 7.0). Horizontal wrist velocity was reduced from 46.5 mm/s (SD = 19.2) to 40.9 mm/s (SD = 17.8) with OBT. Likewise, the variability was reduced from 54.2 mm/s (SD = 26.5) to

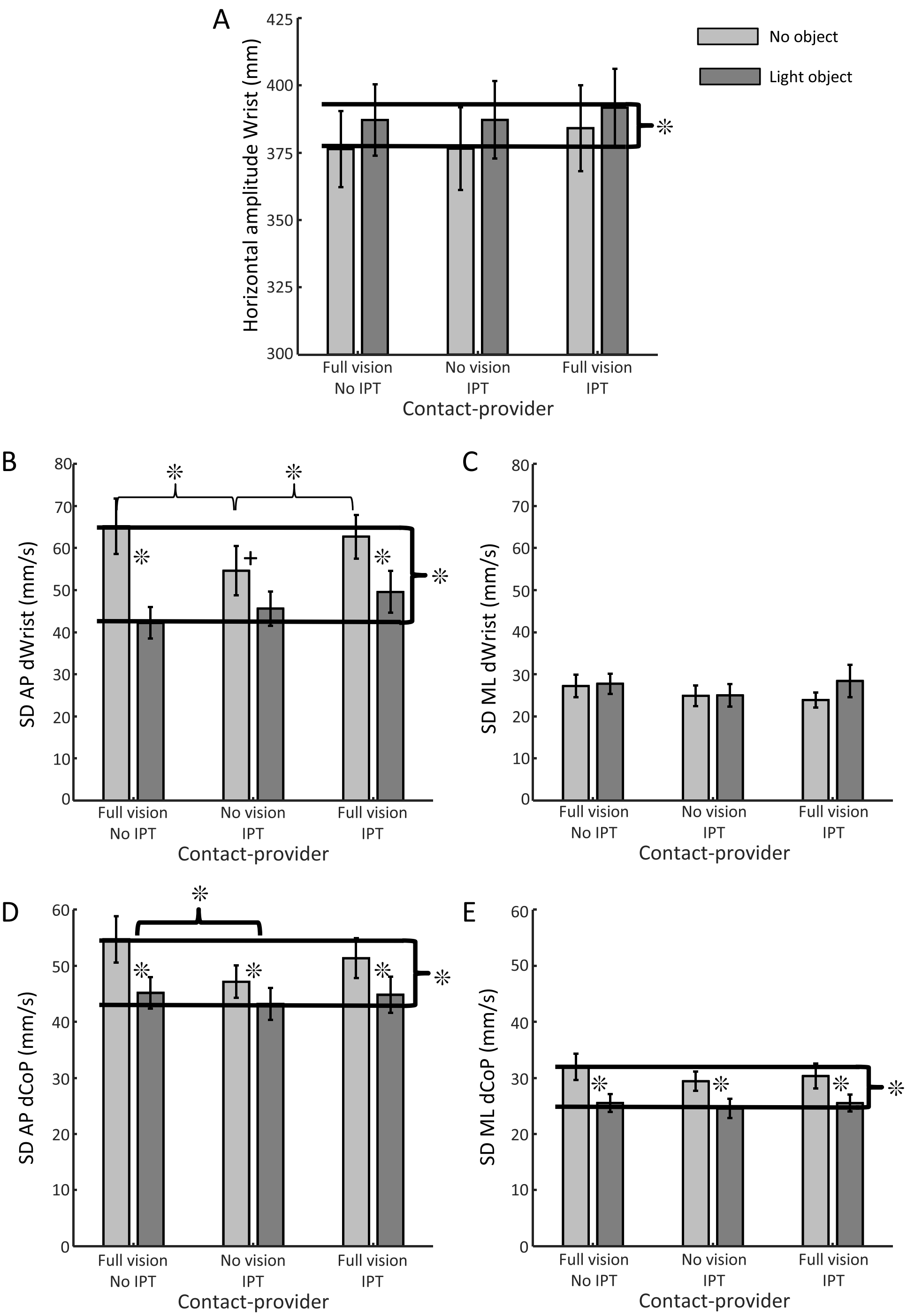

Fig. 2. (A) The horizontal amplitude of the contact receiver's wrist as a function of the presence of the light object (OBT) and visuotactile interpersonal context. The standard deviation of the contact receiver's wrist velocity in the anteroposterior (B) and mediolateral (C) directions during the reach end-state. The standard deviation of the contact receiver's CoP velocity in the anteroposterior (D) and mediolateral (E) directions during the reach end-state. Bold vertical brackets indicate an effect of OBT presence. Bold horizontal brackets indicate a single comparison between visuotactile interpersonal contact conditions averaged for the OBT factor. Thin horizontal brackets refer to a single comparison between not-averaged specific visuotactile interpersonal context conditions. Error bars indicate the between-subject standard error of the mean. The asterisk indicates $p < 0.05$ and the cross indicates $p < 0.1$. IPT: interpersonal touch.

42.5 mm/s (SD = 14.4) with OBT. Curvature indicated a slightly curved trajectory (average = 1.7, SD 0.8), which was not affected by OBT or VIC.

3.2. Postural control in the reach end-state

The reach end-state lasted on average 10.4 s (SD = 3.0). Separating wrist velocity into its AP and ML components resulted

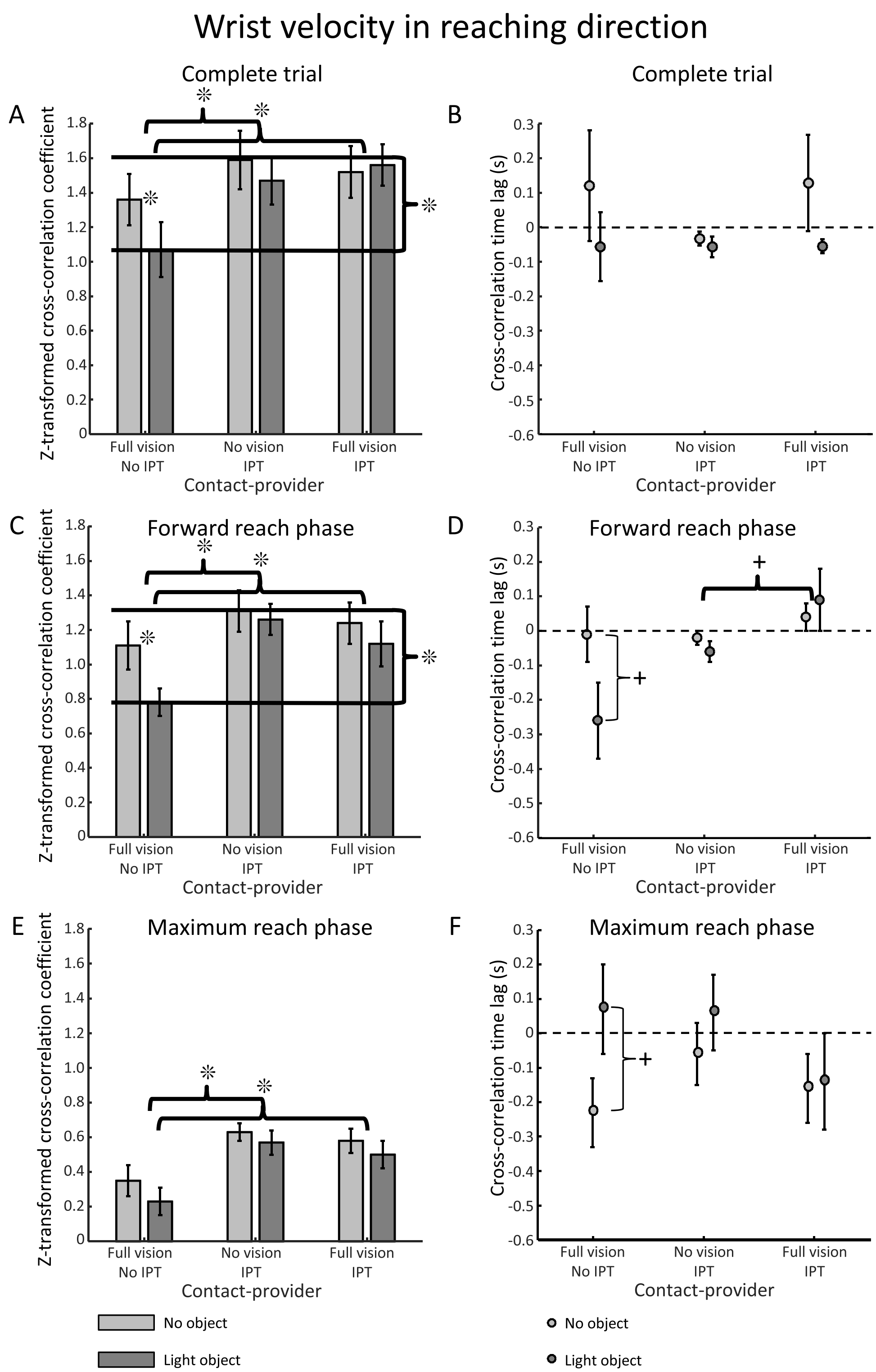

Fig. 3. Left panels show the average Fisher Z-transformed cross-correlation coefficients of the wrist velocity in reaching direction as a function of the presence of the light object (OBT) and visuotactile interpersonal context in (A) the complete trial, (C) reaching phase and (E) maximum reach end-state. Right panels show the cross-correlation time lags as a function of the visuotactile interpersonal context and the object presence in (B) the complete phase, (D) reach phase, (F) and maximum reach end-state. Bold vertical brackets indicate an effect of OBT presence. Bold horizontal brackets indicate a single comparison between visuotactile interpersonal contact conditions averaged for the OBT factor. Thin horizontal brackets refer to a single comparison between not-averaged specific visuotactile interpersonal context conditions. Error bars indicate the between-subject standard error of the mean. The asterisk indicates $p < 0.05$ and the cross indicates $p < 0.1$. IPT: interpersonal touch.

in an effect of OBT and an interaction between OBT and VIC on AP SD dWrist. OBT reduced AP SD dWrist in general (Fig. 2b). Post-hoc single comparisons indicated that IPT without visual feedback and without OBT resulted in a reduction compared to the other two VIC conditions (Fig. 2b).

SD dCoP was reduced by the presence of OBT in both directions (Fig. 2d–e). A tendency of an effect of VIC was found in the AP direction. Single comparisons showed that the IPT condition with visual feedback reduced SD dCoP compared to visual tracking.

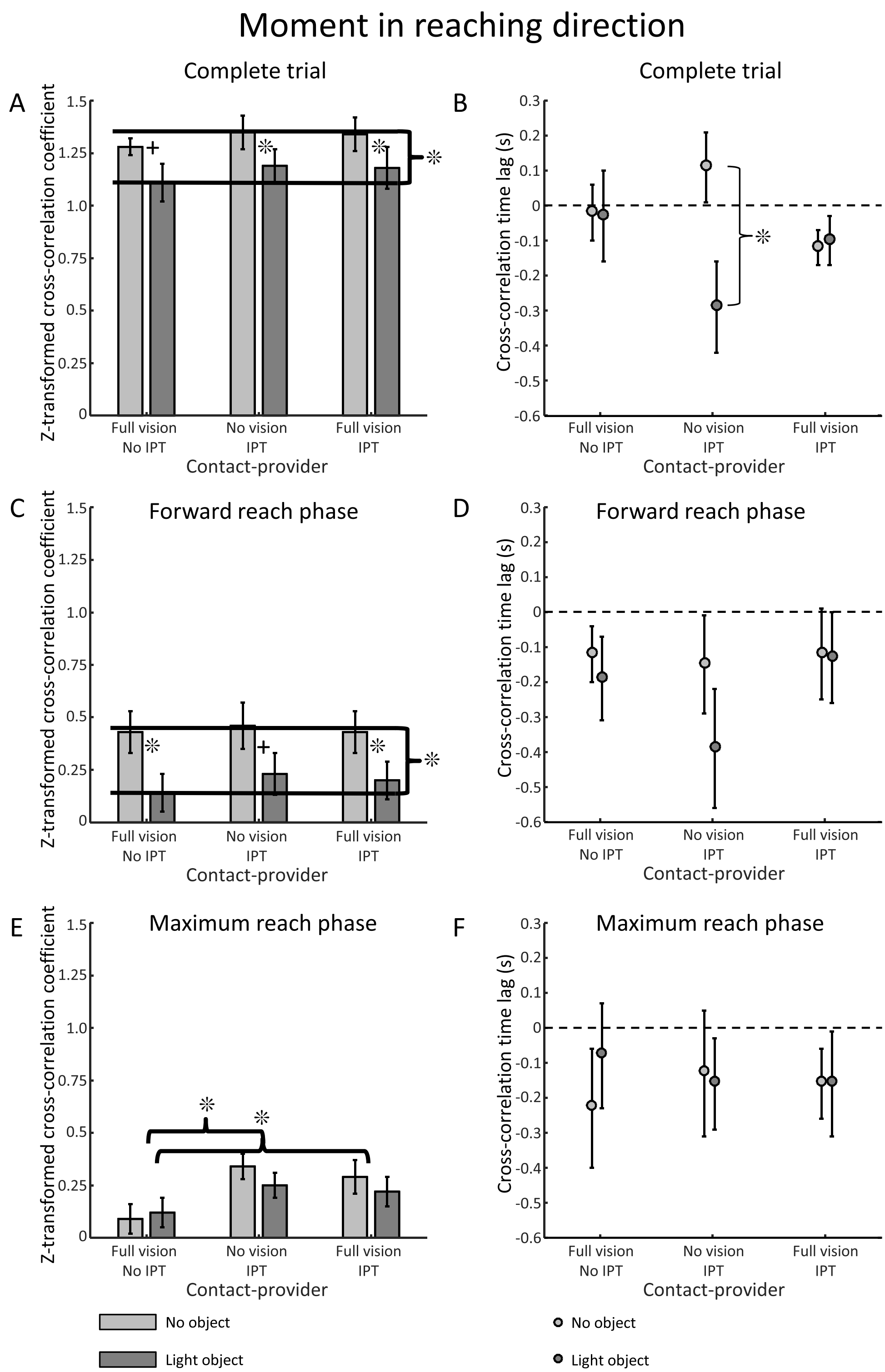

Fig. 4. Left panels show the average Fisher Z-transformed cross-correlation coefficients of the moments in reaching direction as a function of the presence of the light object (OBT) and visuotactile interpersonal context in (A) the complete trial, (C) reaching phase and (E) maximum reach end-state. Right panels show the cross-correlation time lags as a function of the visuotactile interpersonal context and the object presence in (B) the complete phase, (D) reach phase, (F) and maximum reach end-state. Bold vertical brackets indicate an effect of OBT presence. Bold horizontal brackets indicate a single comparison between visuotactile interpersonal contact conditions averaged for the OBT factor. Thin horizontal brackets refer to a single comparison between not-averaged specific visuotactile interpersonal context conditions. Error bars indicate the between-subject standard error of the mean. The asterisk indicates $p < 0.05$. IPT: interpersonal touch.

3.3. Interpersonal coordination

Fig. 3 shows the Fisher-Z-transformed coefficients and time lags of the peak cross-correlations between the wrist velocities of CR and CP in the reaching direction for the complete trial (Fig. 3a–b), the forward reaching (Fig. 3c–d) and the reach end-state (Fig. 3e–f).

Across the complete trial, both OBT and the VIC affected the strength of IPC (Fig. 3a). Single comparisons indicated that in visual tracking, coefficients were weakest compared to the other two IPT conditions. Time lags tended close to zero (average = 8 ms, SD = 457; Fig. 3b). In the forward reaching, coefficients were lower compared to the complete trial but affected in a similar manner (Fig. 3c). The time lags were affected by the VIC and showed an interaction with OBT. Single comparisons indicated that in the condition with IPT and visual feedback, CP tended to show a slight lead ahead of CR (average = 69 ms, SD = 338) compared to IPT without visual feedback, where the interpersonal relationship tended to be reversed (average = 41 ms, SD = 115). In visual tracking, OBT tended to result in CP lagging behind CR by about 263 ms (SD = 528; Fig. 3d) in contrast to a zero lag without OBT (average = 10 ms, SD = 397). In the reach end-state, visual tracking resulted in the weakest IPC compared to the two conditions involving IPT (Fig. 3e). The time lags showed an effect of OBT presence with OBT resulting in zero lags (average = 6 ms, SD = 595) compared to a lead by CR when OBT was absent (average = 151 ms, SD = 454; Fig. 3f).

Fig. 4 shows the Fisher-Z-transformed coefficients and corresponding time lags of the peak cross-correlations between CR and CP for the moments in the plane parallel to the reaching direction across the complete trial (Fig. 4a–b), forward reaching (Fig. 4c–d) and in the reach end-state (Fig. 4e–f).

OBT decreased the strength of IPC (Fig. 4a). Regarding the time lags, single comparisons showed that an interaction between OBT and VIC was caused by the presence of OBT to alter the interpersonal timing when CP provided IPT without vision (Fig. 4b). With OBT, CP followed CR by 286 ms (SD = 62), while in the absence of OBT, CP was 112 ms (SD = 486) ahead of CR. In the other two VIC conditions time lags showed a lead of CR about 70 ms (SD = 400). In forward reaching, coefficients were generally lower relative to the complete trial. Similarly, OBT presence reduced the strength of IPC (Fig. 4c). Time lags indicated that CP followed CR by about 184 ms (SD = 614; Fig. 4d). In the maximum reach phase coefficients were still lower than during forward reaching. An effect of VIC was found (Fig. 4e). Single comparisons indicated that visual tracking showed the weakest IPC compared to the other two conditions. Overall, the time lags averaged around 155 ms (SD = 697; Fig. 4f).

4. Discussion

We aimed to understand the spontaneous IPC for balance support in maximum forward reaching and intended to modulate the leader-follower relationship by creating asymmetric interpersonal dependencies. CR, deprived of visual feedback and in the less stable postural state, was supposed to rely more strongly on CP when no alternative source of haptic information was available. On the other hand, CP's responsiveness to CR was expected to vary with the visuotactile interpersonal context in terms of visual feedback and the IPT instruction.

OBT influenced the reaching performance of CR. The precision demands (speed/accuracy) were greater with OBT as expressed by CR's reduced and less variable reaching speed. In the reach end-state, increased amplitude with OBT (Fig. 2a) coincided with reduced AP wrist and SD dCoP (Fig. 2b,d). Our results confirm previous observations that a target object in the FR task facilitates performance [18,19]. Despite low friction of the fibreglass surface, the interaction with OBT could have resulted in haptic feedback at the fingertips facilitating control of balance [3] and resembling a non-rigid, haptic 'anchor' as conceptualized by Mauerberg-deCastro and colleagues [20].

Contact between the hands ought to have resulted in better interpersonal coordination and synchronization. Indeed, an increase in strength of IPC between the hands occurred in the two IPT conditions. Nevertheless, mechanical coupling between the hands is unlikely as IPT provided support to CR's arm in terms of vertical friction only. The absence of an effect of the VIC on SD dWRI in the ML direction indicates that IPT did not constrain CR's forward reaching. This is corroborated by the observation that the movement trajectories were also not influenced by IPT. In contrast in the reach end-state, both AP wrist and CoP velocity showed selectively reduced variability during IPT without visual feedback. For SD dCoP this difference was independent of the presence of OBT (Fig. 2d). It seems that the benefit of IPT appeared predominantly when CP was not able to observe CR visually. Summation of OBT and IPT should have resulted in greatest improvements in reach distance and balance stability. The lack of a summation effect of the two haptic modes [21] as observed in individual, passively received light touch [22] suggests that the two sources were not integrated. Reliability estimates or the contextual information of the two sources could have been too divergent [23]. While CR participants have experience in contacting environmental objects during stance, the social content of IPT could have made it incompatible with the OBT signal. Perhaps the variability reductions with IPT may result from social facilitation [24] with the requirement that CP attends exclusively to CR's local dynamics.

Individuals achieve joint goals by switching between symmetrical and asymmetrical modes of IPC depending on the constraints of their complementary roles. Skewes et al. [25] investigated how people trade synchronization and complementarity in a continuous joint aiming task. Interestingly, when the level of difficulty in the complementary task became too high for one partner of the dyad, this person became less adaptive to their partner's requirement thus taking the 'leader' role in the joint task. In addition, partners synchronized better with an irregular, but adaptive partner, than with a completely predictable one [25]. OBT presence and the VIC altered the strength and temporal coordination between both individuals during IPT across the complete trial and during forward reaching. OBT reduced the cross-correlation coefficients between both individuals (Figs. 3a, 3c, 4a, 4c). OBT was more relevant to CR than to CP, therefore this difference expresses CR's responsiveness to the interpersonal context. For example, being engaged in a precision task, restricted CR's adaptability, which could explain why CR was 'leader' in the majority of testing conditions.

With respect to IPC of the postural responses, CP used to follow CR's motion by up to 200 ms when visual feedback was involved (Fig. 4b,d). Thus, visual processing in CP's task requirements seems to have resulted in a reactive mode. While the nature of the IPT signal is local, with eyes open CP may have attended to the global scene and involuntarily experienced visual dominance [26]. Although vision dominates in bisensorial contexts, latencies to visual stimuli in these situations are typically delayed compared to touch or audition [27]. In the condition without visual feedback for CP but constant IPT, the presence of OBT made a big difference (Fig. 4b). Removing OBT, which deprived CR of a competing tactile signal, seems to have caused CR to a focus on the IPT signal, thereby turning CP into the 'leader'. During forward reaching (Fig. 4d), however, once more time lags indicated CP as the 'follower'. Naturally, the reaching phase did not contain the transition points such as initiation and stop. It is reasonable that these two events are central to successful IPC. Perhaps, in the IPT condition without visual feedback and in the absence of OBT at CR's fingertips, CR's motion onset was triggered by CP.

According to our present results, a caregiver needs to take into account the context-dependent responsiveness of a patient. If a caregiver intends to guide a patient haptically, the caregiver needs to ascertain that two prerequisites are met: the patient has no competing tactile signal available and the therapist deliberately refrains from adopting a reactive mode based on vision. This still needs to be tested in realistic patient-caregiver settings.

5. Conclusions

We described the effects of visual and haptic sensory information on interpersonal postural coordination in an asymmetrical maximum forward reach joint action paradigm. We observed temporal movement coordination between a 'contact-provider' and a 'contact-receiver' to depend on the presence of an external object and the visuotactile interpersonal context. Interpersonal postural coordination was strongest when deliberately light IPT was provided without the presence of an additional object at the contact-receiver's fingertips. As the leader-follower relationship between both partners was also modified by the visuotactile interpersonal context of the contact-provider, the sensorimotor states of both partners have to be considered of equal importance. We speculate that IPT is a promising strategy for patient guidance in clinical settings. More research is needed before its implementation as a patient manual handling tool.

Conflict of interest

There are no conflicts of interest for any of the authors.

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Full length article

Interpersonal interactions for haptic guidance during balance exercises

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ABSTRACT

Background: Caregiver–patient interaction relies on interpersonal coordination during support provided by a therapist to a patient with impaired control of body balance.
Research question: The purpose of this study was to investigate in a therapeutic context active and passive participant involvement during interpersonal support in balancing tasks of increasing sensorimotor difficulty.
Methods: Ten older adults stood in semi-tandem stance and received support from a physical therapist (PT) in two support conditions: 1) physical support provided by the PT to the participant's back via an instrumented handle affixed to a harness worn by the participant ("passive" interpersonal touch; IPT) or 2) support by PT and participant jointly holding a handle instrumented with a force-torque transducer while facing each other ("active" IPT). The postural stability of both support conditions was measured using the root-mean-square (RMS) of the Centre-of-Pressure velocity (RMS dCOP) in the antero-posterior (AP) and medio-lateral (ML) directions. Interpersonal postural coordination (IPC) was characterized in terms of cross-correlations between both individuals' sway fluctuations as well as the measured interaction forces.
Results: Active involvement of the participant decreased the participant's postural variability to a greater extent, especially under challenging stance conditions, than receiving support passively. In the passive support condition, however, stronger in-phase IPC between both partners was observed in the antero-posterior direction, possibly caused by a more critical (visual or tactile) observation of participants' body sway dynamics by the therapist. In-phase cross-correlation time lags indicated that the therapist tended to respond to participants' body sway fluctuations in a reactive follower mode, which could indicate visual dominance affecting the therapist during the provision of haptic support.
Significance: Our paradigm implies that in balance rehabilitation more partnership-based methods promote greater postural steadiness. The implications of this finding with regard to motor learning and rehabilitation need to be investigated.

1. Introduction

Falls and fall related injuries in older adults are a public health issue [1,2]. Balance exercises, however may reduce falls risk [3]. In balance rehabilitation, a physical therapist (PT) manipulates the provision of sensory cues during sensorimotor training to facilitate motor learning, and control of body balance [4–6].

The factors governing sensorimotor interactions between therapist and client, however are poorly understood [7]. Interpersonal sensorimotor interaction can be classified into cooperation and collaboration [8]. In contrast to collaborative interactions that do not integrate a priori role assignments, roles are assigned a priori to each participant in cooperative interactions. For example during balance exercises, this can lead to an allocation of sub-tasks, such as provision of haptic balance support by a therapist and reception by the client involved in the balancing task [9].

Additional tactile feedback is a reliable approach to augment control of body balance [10]. In the traditional paradigm ("active" light touch), a participant is controlling the upper limb directly, which is contacting the external haptic reference [11]. Hereby, the movement degrees of freedom of the contacting limb are used for precision control of the contact force with the control of body sway as a separate process [12]. In addition to the haptic feedback signal, the output of fingertip control could serve as a signal to control sway [13]. In non-manual,

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"passive" light touch, the contact is delivered to a participant's body segment. A participant is less able, to control the precision by which the contacting force is applied [13]. Here, the movement degrees of freedom available to a participant for controlling the contact force are limited by the current postural degrees of freedom, thereby creating a direct equivalence between control of body sway and precision of the contact.

Passive light touch with an earth-fixed reference results in proportional sway reductions in the range of 20%–30% [13]. This is similar to what has been reported in studies involving fingertip light touch [i.e. 14]. Interpersonal fingertip touch (IPT) leads to lesser sway reductions of around 9–15% [9,14–17]. The reason for this diminished effect could lie in the fact that the contact reference is not earth-fixed but shows own motion dynamics, which might make disambiguation of the haptic signal in terms of own sway-related feedback more challenging. Johannsen et al. [9] assessed "passive" IPT in neurological patients as well as chronic stroke and reported sway reductions between 15%–26%. In stroke patients, passive, trunk-based IPT [9], nevertheless, seemed more beneficial than fingertip IPT [16].

In our study, we directly contrasted the effects of active and passive support modes on body sway in a therapeutic setting. We measured the interaction forces between a physiotherapist and participants and characterized the interpersonal postural coordination (IPC) between both partners. We predicted that the participant would demonstrate the greatest sway reductions when passive IPT was provided to the trunk with no involvement in contact precision control. We increased the sensory challenges imposed by the balance task (foam surface, eyes closed, pitch head movement) and assumed that with increasing difficulty, the benefit of IPT would increase as well potentially in interaction with the specific IPT mode.

2. Methods

2.1. Participants

Ten older adults without significant neurological or orthopedic history, between the age of 71 and 86 years (mean age 79 yrs, SD = 5; 5 females, 5 males; all right-handed for writing) participated in this study. One PT (16 years of experience) provided support.

2.2. Recruitment and exclusion criteria

Participants were recruited from a sample of screened healthy elderly subjects from a preliminary study [18]. This study was approved by the Institutional Review Board of the University of Pittsburgh.

2.3. Demographic data

Participants completed the Activities-specific Balance Confidence Scale (ABC) questionnaire [19] and the Functional Gait Assessment [20] prior to the experiment. The participants reported a balance confidence level between 74% and 100% (mean 94%, SD = 8). The Functional Gait Assessment (FGA) is a modification of the Dynamic Gait Index (DGI) that uses higher level gait tasks [20]. Participants achieved scores between 17 and 30 in the FGA (mean 26, SD = 5).

2.4. Experimental design

Participants performed 2 sets of 6 randomized balance exercises during two different conditions: passive support (PS) and active support (AS) (Fig. 1). In the PS condition, the PT who was in bipedal stance with full vision, stood behind the participant and lightly held on to an instrumented handle mounted on the back of the participant's vest and applied stronger support only when he felt the participant required firmer assistance to maintain upright balance. In the AS condition, the PT and the participant faced one another and simultaneously held on to the handle. Participants were instructed to stand as stable as possible with their arms crossed in front of their waist (PS) or to stand as stable as possible while holding on to a handle (AS). For each set of six balance conditions participants completed a partial factorial design of the conditions (see Fig. 1D). These exercises were chosen across a range of difficulty based on a preliminary study [18].

2.5. Instrumentation

The participant and PT stood on separate force platforms (Bertec, Columbus, Ohio, USA) that measured ground reaction forces and moments at a sampling rate of 120 Hz (see Fig. 1A and B). A tri-axial load cell (DSA-03 A TecGihan, Japan) was mounted to a custom-made handle and bracket which was secured to the back of a support vest worn by the participant to measure forces during the PS condition (see Fig. 1A). Force plate and load cell data were collected by the same data acquisition system (National Instruments, Austin, TX). During the AS condition, the handle was removed from the vest and a second handle was attached to the bracket for the participant's use (see Fig. 1B).

2.6. Procedure

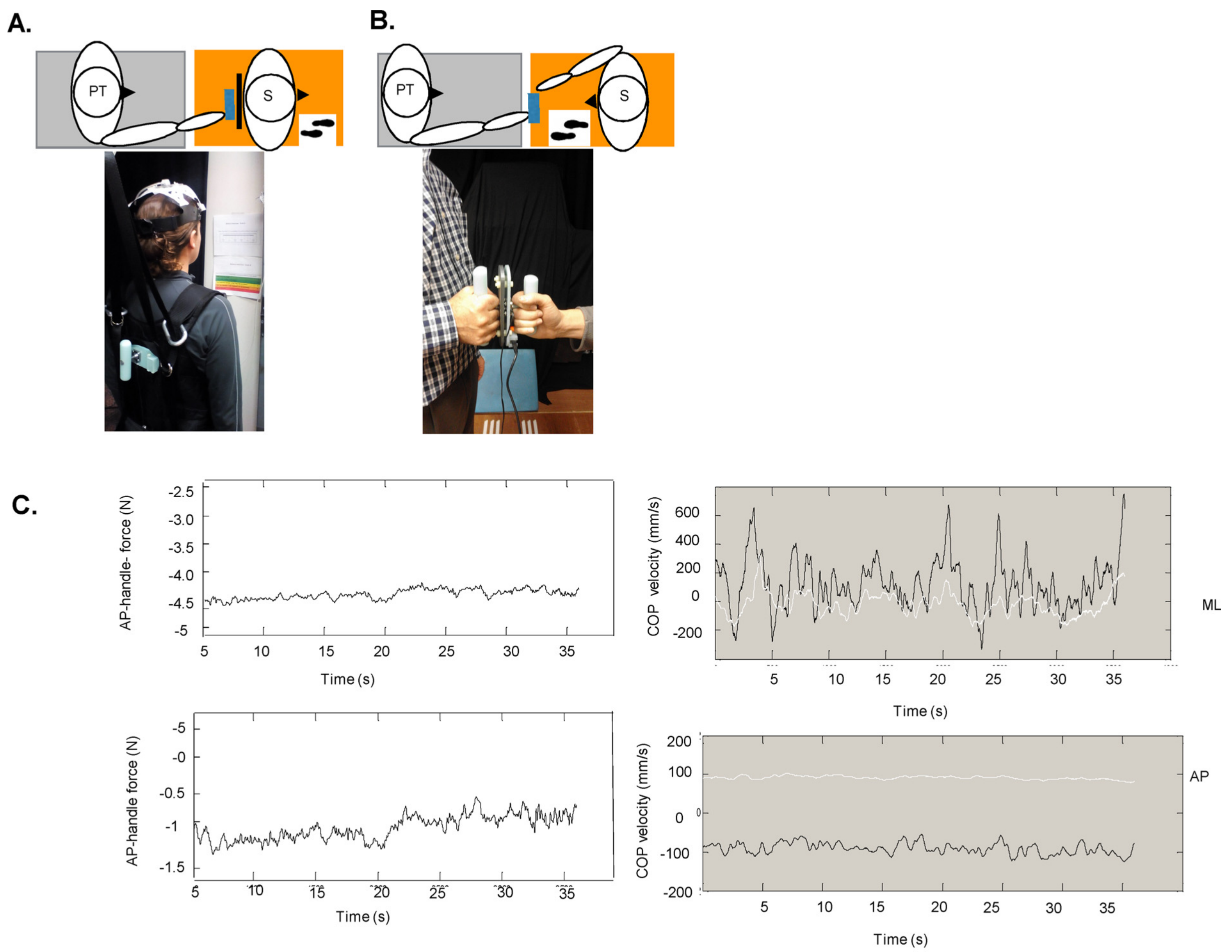
Participants stood in semi-tandem stance by placing their feet so that the medial borders were touching, and moving their dominant foot backward by a half of foot length [21]. During the foam surface conditions, participants stood on foam (AIREX Balance Pad S34-55, height 6 cm, length 51 cm, width 40 cm). During the pitch condition, participants moved their head over a total range of 30 degrees at 1 Hz by following a metronome [22]. Trials lasted 30 s and participants wore a safety harness.

2.7. Data reduction and statistical analysis

The force platform and load cell data were transformed into center of pressure (COP) and handle force measurements, respectively, using calibration equations The antero-posterior (AP) and medio-lateral (ML) components of the COP and the AP component of the handle force were extracted. All data time series were smoothed using a dual-pass, 4^{th} order Butterworth lowpass filter (cutoff = 10 Hz). COP data were numerically differentiated to produce COP velocity measures. Velocity information is the predominant source of body sway control [23] therefore the root-mean-square of the AP and ML COP velocity (RMS dCOP) were the primary postural control measures. The IPC was estimated by computing the cross-correlation functions between both participants' COP velocity time series.

Cross-correlations were computed within a range of minimum and maximum time lags between ± 3 s. We used the standard MATLAB cross correlation function which measures the dependence between two signals [24,25]. The largest maximum (in-phase behavior) and minimum (anti-phase behavior) cross-correlation coefficients and corresponding time lags were extracted. The cross-correlation coefficients were Fisher Z-transformed for statistical analysis.

SPSS version 23 was used for statistical analysis. A linear mixed model analysis with support mode (2 levels: active and passive) and condition (6 balance exercises) effect as well as the support * condition interaction was performed. For the estimation of the model we used a maximum likelihood method. Postural sway parameters (RMS) were analyzed including subject as a random effect while IPC parameters (correlation coefficients, lags) and forces were analyzed using only fixed effects. A diagonal covariance structure was used for repeated effects in the mixed model [26]. An alpha level of 0.05 was used for level of significance, and post-hoc comparisons were computed using Sidak adjustment.



D.

Surface		Vision	Head movement
1.	firm	eyes open	still
2.	firm	eyes closed	still
3.	firm	eyes closed	pitch
		Break	
4.	foam	eyes open	pitch
5.	foam	eyes closed	still
6.	foam	eyes closed	pitch

Fig. 1. (A & B) The stance configuration of the experimental setup at the beginning of a trial with the physical therapist on the grey force plate and the subject in semi-tandem on the orange force plate in the passive intermittent support mode (A) and in the active continuous support mode (B). The instrumented handle is represented by the blue rectangle in the schematic. Time series plots of the antero-posterior (AP) handle force (left) and AP and medio-lateral (ML) COP velocity of the physical therapist (light) and subject (dark) in active support mode during a foam surface, eyes closed and pitch movement trial (right) (C). The subject performs six balance exercises with increasing difficulty (D) (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

3. Results

3.1. Postural control

3.1.1. Sway velocity in AP direction

Significant support ($F(1,58.5) = 22.8$, $p < 0.001$) and condition ($F(5,28.5) = 80.6$, $p < 0.001$) effects were found for participant RMS dCOP in the AP direction (Fig. 2). The passive support led to higher sway velocity production. The sensory conditions generated progressively increased sway velocity (see Fig. 2A).

3.1.2. Sway velocity in ML direction

Analysis of the RMS dCOP in the ML direction generated similar support ($F(1,57.5) = 51.3$, $p < 0.001$) and condition ($F(5,25.9) = 59.2$, $p < 0.001$) effects as in the AP direction, but there was also a significant interaction between condition and support ($F(5,25.8) = 3.90$, $p = 0.001$) (Fig. 2B). The interaction indicates that there was greater difference in the amount of sway velocity between

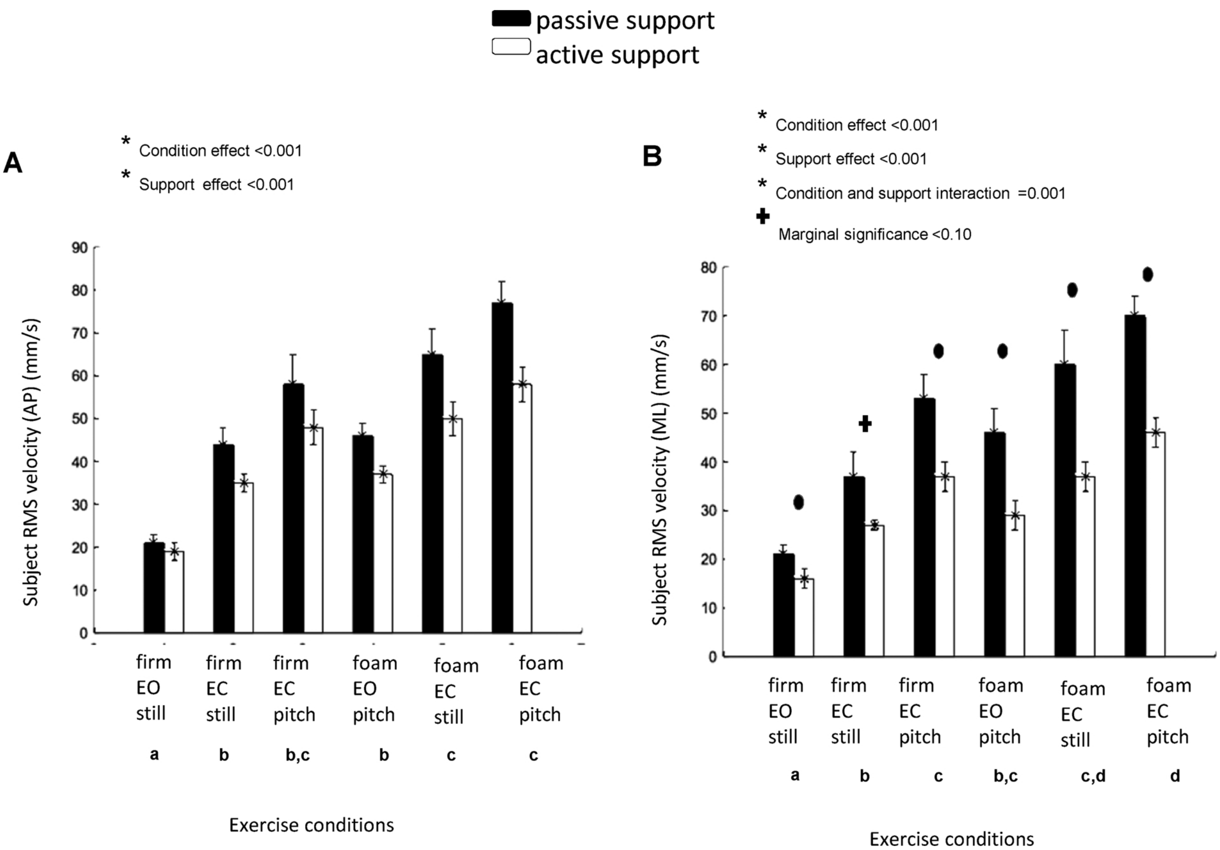


Fig. 2. The RMS COP velocity as a function of the exercise conditions and the support provision (passive/active) in AP (A) and in ML direction (B). Letters show the pairwise comparison between conditions; the same letters express conditions are not significantly different from each other. Bold dots indicate the significant support differences within each condition. Error bars indicate the standard error of the mean.

passive and active support conditions as the balance conditions became more challenging. The difference in sway velocity ranged from approximately 18.5 mm/s in the firm surface, eyes open, head still condition to 58 mm/s during the foam surface, eyes closed, head pitch condition.

3.2. Handle forces

3.2.1. Average AP handle force

A significant effect of support mode ($F(1,46.2) = 8.22$, $p = 0.01$) on the average handle force was found (Fig. 3A). A mean force of 1.7 N (SD 0.5 N) in the posterior direction on the handle was observed during the passive support trials. During the active support trials, the forces of the PT and participants counteracted one another on average, with a mean force of 0.01 N (SD = 0.05 N) towards the PT. A significant effect of sensory condition ($F(5,22.2) = 4.0$, $p = 0.01$) was found. Larger posterior forces on the handle were exerted during the foam, eyes closed, and passive support conditions compared with much smaller force exertion during the other conditions. During the active support trials, a pattern emerged in which the force was directed toward the participant in the easier conditions, and toward the PT in the foam, eyes closed conditions. Lateral forces were also minimal (see Fig. 3A).

3.2.2. Variation in AP handle force

The magnitudes of variation of handle forces applied between the PT and participant, as measured by the standard deviation of the time series, are shown in Fig. 3B. A progressive increase in variation in forces occurred as the sensory conditions became more difficult ($F(5, 21.8) = 18.4$, $p < 0.001$).

3.3. Interpersonal coordination of postural sway

3.3.1. Minimum cross correlation coefficients between participant and PT

Fig. 4 displays the minimum (i.e. anti-phase) cross correlation coefficients between the COP velocity of the PT and participant. A significant condition effect was found in both the AP (Fig. 4A, $F(5,29.7) = 9.2$, $p < 0.001$) and ML directions (Fig. 4B and $F(5, 37.1) = 3.9$, $p = 0.01$). In the AP direction, IPC anti-phase behavior was larger in the eyes closed conditions. In the ML direction, there was less anti-phase IPC in the firm surface, eyes open, head still condition.

3.3.2. Maximum cross correlation coefficients between participant and PT

The maximum (i.e. positive) cross-correlations were greater in absolute magnitude than the minimum (negative) cross-correlations, indicating that the in-phase IPC was more prominent than the anti-phase IPC. The IPC in-phase behavior of the COP velocity in the AP direction demonstrated significant support, condition and interaction effects (Fig. 4C). Lower average interpersonal cross-correlation coefficients were found in AS 0.28 (SD 0.02) than in PS 0.34 (SD 0.02) in the AP direction ($F(2,101.8) = 13.4$, $p < 0.001$), which indicated greater strength of the in-phase IPC in the passive mode. The sensory conditions differed ($F(5,34.2) = 20.8$ $p < 0.001$), which showed increasing IPC during the more difficult sensory conditions, similar to the pattern of results of the RMS dCOP. A significant interaction between support and exercise mode ($F(5,34.2) = 2.7$ $p = 0.04$) demonstrated greater IPC during the active support mode for the firm surface, eyes open, head still condition, in contrast with greater IPC during the passive support mode for all other conditions. The in-phase coordination in the ML directions showed a significant condition effect only ($F(5,35.8) = 14.24$ $p < 0.001$).

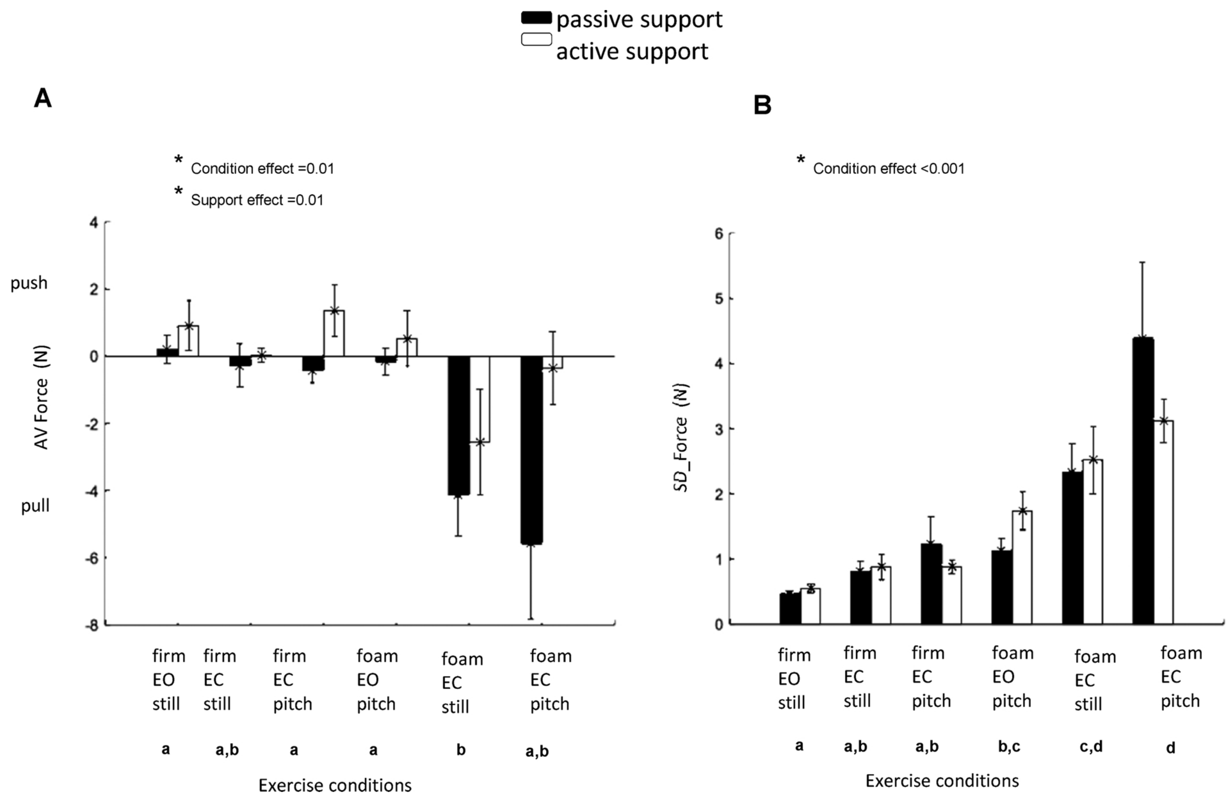


Fig. 3. The average (AV) of the handle force as a function of the exercise conditions and the support mode (passive/active) (A) as well as the standard deviation (SD) of the handle force as a function of the exercise conditions and the support mode (passive/active) (B). Letters show significant pairwise differences between conditions; same letters express that conditions are not significantly different from each other. Error bars indicate standard error of the mean.

3.4. Time lags in IPC between participant and PT

We found a significant support mode effect ($F(1,90.6) = 6.6$, $p = 0.02$; passive mean = −287 ms SD = 13 ms; active mean = 210 ms SD = 13 ms) (Fig. 5A; anti-phase IPC). The PT led in all but the third sensory condition (AS) and followed in all but the second and third sensory conditions (PS). Fig. 5C (in-phase IPC) demonstrates a pattern in which the PT was always the follower (AS: mean = 159 ms SD = 17 ms; PS: mean = 323 ms SD = 21 ms) with the exception of the easiest sensory condition (firm, EO, still) in active mode.

4. Discussion

We aimed to contrast the effects of two different modes of client participation in the provision of interpersonal light touch balance support by a therapist to balance-challenged older adults.

4.1. Postural control

In both directions, the active support mode resulted in less participant sway velocity compared with the passive support mode. Proportional sway velocity difference between both modes was 32% of passive condition, which is similar to passive LT sway reductions with an earth-fixed reference or fingertip LT [14,27]. An interaction between support mode and sensory condition for sway in the ML direction indicated that the active support mode provided a greater benefit with greater sensory disruption. The observation that more active participation in the control of contact force precision resulted in reduced sway under conditions of greater sensorimotor destabilization was unexpected as in previous studies the comparative proportional benefit of passive trunk-based IPT on body sway tended to be greater than IPT at the fingertips.

The difference between the two IPT modes in this study could rest on stronger and less ambiguous haptic feedback from the grasp of the handle or processes of anticipatory postural control and voluntary force precision control in the active IPT mode. Wing et al. [28] investigated the coupling between grip force during one-handed precision grasp on a manipulandum and concurrent postural adjustments in anticipation of dynamic and static loads during horizontal pulling and pushing. They demonstrated a functional linkage between grip force adjustments anticipating changes in load force on the manipulandum and ground reaction torque in anticipation of self-imposed balance perturbations due to the pushing and pulling motion. They suggested that an efferent signal controlling grip force could facilitate the prediction of upcoming postural load and appropriate postural adjustments [28]. Further, minimization of the interaction force and its variability could have resembled the goal of a so called “suprapostural” task resulting in proactive, task-adapted body sway reductions [29,30]. As the latter mechanism might apply to fingertip IPT too, we speculate that an efferent grip force control signal contributing to anticipatory postural control facilitated postural stability primarily in this study instead.

By facing the participant in active mode, the therapist might have received clearer social cues about postural destabilization of the participant that facilitated internal simulation of a participant’s sway dynamics for the anticipation of instabilities and need for support [31]. For example, the sight of another person can improve an individual’s ability to compensate for imbalance [32].

4.2. Handle forces

It needs to be considered that in the passive IPT mode, strength of the contacting force was upregulated intermittently based on the therapist’s visuotactile assessment of a participant’s current state of postural stability. In the easier sensory conditions, the interaction forces remained relatively low, which possibly indicates the relative absence of active stabilization of participants’ sway by the therapist. The

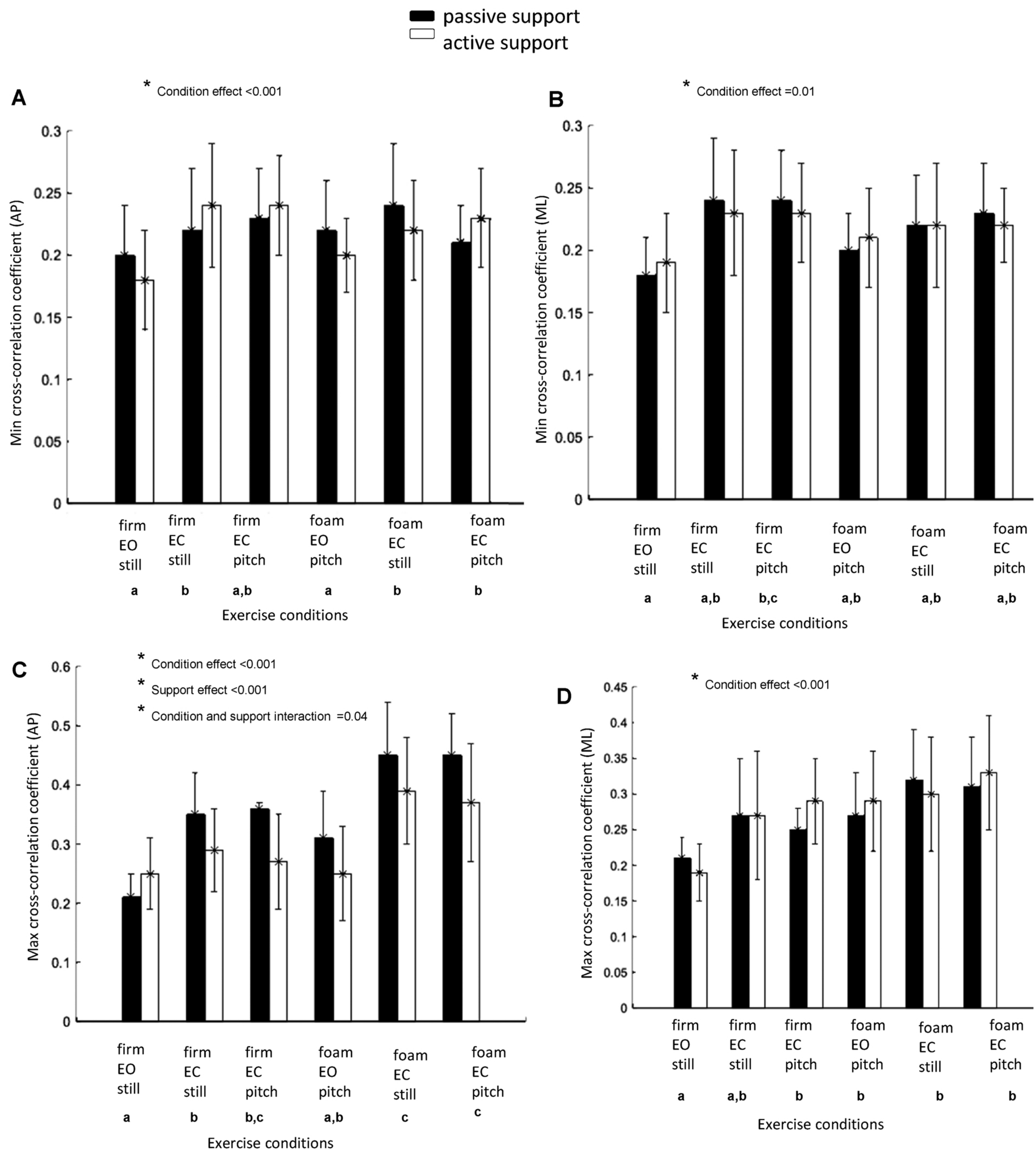


Fig. 4. Upper panels show the average minimum cross-correlation coefficients of the CoP velocity as a function of the exercise conditions and the support mode (passive/active) in AP (A) and ML (B) direction. Lower panels show the average maximum cross-correlation coefficients of the CoP velocity as a function of the exercise conditions and the support mode (passive/active) in AP (C) and ML (D) direction. Statistical results refer to the Z-transformed cross-correlations. Minimum cross-correlations represent negative values and are shown rectified for better visual understanding. Error bars indicate the standard error of the mean. Letters show significant differences between conditions; same letters express conditions that are not significantly different from each other.

interaction forces fell into the range from 4 N to 6 N in the two most challenging conditions (foam surface), which could imply more continuous in addition to stronger haptic support.

Nevertheless the stronger haptic support with passive IPT did not result in less variable body sway compared to the active mode in the two most challenging conditions. As the variability of the interaction force was comparable, we can ascertain that the average interaction forces are not affected by an averaging artefact of extreme values.

Despite less physical support by the therapist, the balance reduction is still greater in the active mode, which corroborates our conclusion that participants received additional cues facilitating of body sway control.

4.3. Interpersonal coordination of postural sway

In the AP direction of sway spontaneous in-phase in both active and passive IPT was the prominent IPC pattern, which confirms observations in previous studies [14,15]. IPC was strongest in the two most challenging sensory conditions and in the majority of sensory conditions passive IPT resulted in stronger IPC than active IPT, with the exception of the easiest condition. Possibly, active stabilization of the participant by the therapist was applied less frequently in the easiest

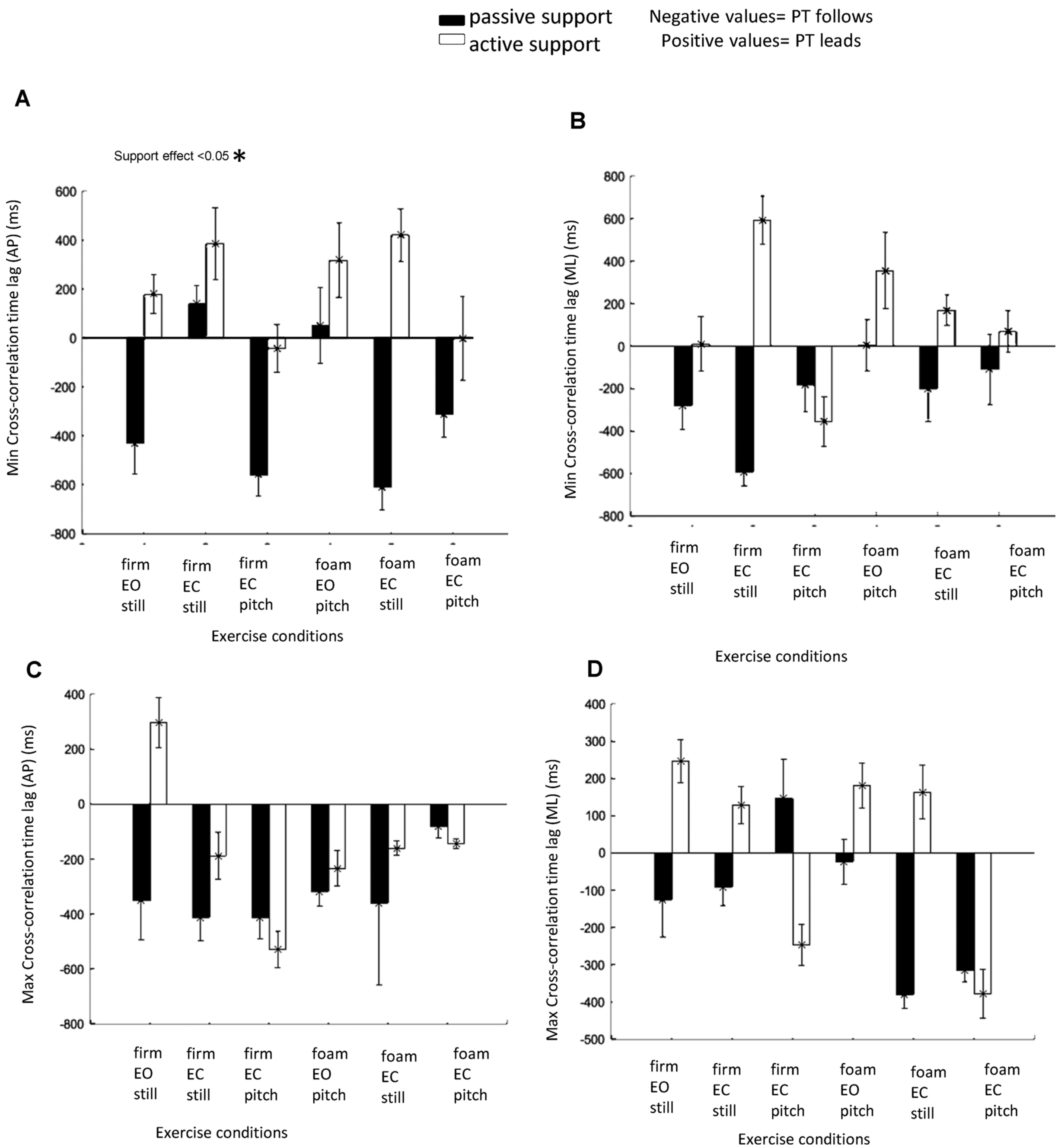


Fig. 5. Upper panels show the minimum average cross-correlation lags of the CoP velocity as a function of the presence of the exercise conditions and the support mode (passive/active) in AP (A) and ML (B) direction. Lower panels show the maximum average cross-correlation lags of the CoP velocity as a function of the presence of the exercise conditions and the support mode (passive/active) in AP (C) and ML (d) direction. Statistical results refer to the Z-transformed cross correlations. Error bars indicate the standard error of the mean.

sensory condition with passive IPT, therefore causing weaker IPC, compared with the active IPT mode, in which stronger interpersonal entrainment [33] could have driven IPC. Fingertip IPT has been reported to result in lower cross-correlation coefficients compared to shoulder IPT [17], which might indicate that the involvement of a greater number of movement degrees of freedom in both partners interpersonal haptic interactions amounts to generally weaker IPC.

The corresponding time lags of the maximum in-phase cross-correlation coefficients demonstrated an average lead of 164 ms by the participant's over the therapist's body sway fluctuations. This is surprising as previous studies reported zero lags [14,17,34]. In these studies, however, visual feedback of the partner's body sway was not available or restricted to peripheral vision, which could have allowed haptic feedback to dominate the IPC. In this current study, the therapist kept open eyes permanently to observe a participant's body sway. We speculate, that visual dominance caused the therapist to automatically adopt a reactive follower mode [35,36]. We observed a similar leader-follower relationship in a forward reaching task, when visual feedback was available to the contact provider [37].

5. Conclusion

We described the effects of passive and active involvement for balance support in a therapeutic context. The passive mode demonstrated increased strength of the interpersonal coordination and the active mode decreased the postural sway of the participant to a greater extent. We suggest balance training could be more effective when both partners face each other. Being more involved in the interaction might enable the participant to spend more time in a challenging balance situation searching and practicing a successful postural strategy. This still needs to be further investigated.

Conflict of interest

There are no conflicts of interest for any of the authors.

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Research

Light touch for balance: influence of a time-varying external driving signal

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Sensory information about body sway is used to drive corrective muscle action to keep the body's centre of mass located over the base of support provided by the feet. Loss of vision, by closing the eyes, usually results in increased sway as indexed by fluctuations (i.e. standard deviation, s.d.) in the velocity of a marker at C7 on the neck, s.d. dC7. Variability in the rate of change of centre of pressure (s.d. dCoP), which indexes corrective muscle action, also increases during upright standing with eyes closed. Light touch contact by the tip of one finger with an environmental surface can reduce s.d. dC7 and s.d. dCoP as effectively as opening the eyes. We review studies of light touch and balance and then describe a novel paradigm for studying the nature of somatosensory information contributing to effects of light touch balance. We show that 'light *tight* touch' contact by the index finger held in the thimble of a haptic device results in increased anteroposterior (AP) sway with entraining by either simple or complex AP sinusoidal oscillations of the haptic device. Moreover, sway is also increased when the haptic device plays back the pre-recorded AP sway path of another person. Cross-correlations between hand and C7 motion reveal a 176 ms lead for the hand and we conclude that light tight touch affords an efficient route for somatosensory feedback support for balance. Furthermore, we suggest that the paradigm has potential to contribute to the understanding of interpersonal postural coordination with light touch in future research.

Keywords: sensory; motor; balance

1. INTRODUCTION

Upright bipedal stance is an inherently unstable posture. In the sagittal plane, centre of mass (CoM) of the body naturally lies in front of the ankle, so the tendency is for anterior (forward) sway. In the frontal plane, the bridge-like frame formed by the legs and pelvis with CoM in the middle appears more stable, however, the slightest sideways displacement in either direction results in sway in that direction. In either case, muscle action (for example, the calf muscles in the case of forward sway or left or right hip abductor in the case of lateral sway) is required to arrest and reverse the sway. Thus, standing is a matter of correcting the tendency to sway by activating appropriate muscles.

Changes in the level of muscle activation to maintain upright stance produce fluctuations in ground reaction forces and torques, and these may be used as an index of the degree of control required for balance. Typically, the measure employed is the variability of the rate of change of the centre of pressure (s.d. dCoP), where CoP reflects the point through which the net forces and torques act in anteroposterior (AP) and mediolateral (ML) directions [1]. Normally, multiple sensory cues, including vision, vestibular sensation, proprioception (leg muscle) and tactile sensations (soles of the feet) are available to the nervous system for the detection of sway. Reduced or absence of input to any of these sensory channels, but especially vision, often results in increased sway excursions that require greater muscle activation to compensate, and this is indexed by greater s.d. dCoP. However, in the case of loss of vision, light touch contact with just the tip of one finger with a stable environmental surface restores sway (or s.d. dCoP) to the level associated with full vision. In this paper, we review a series of studies from the last 15 or so years identifying sensory factors contributing to light touch attenuation of sway. We then describe results from a new paradigm, involving movement of the contact point, which we use to explore the underlying mechanism of light touch contributions to balance.

(a) *Review of studies of the effect of light touch on sway*

In the first of many publications relating to the light touch paradigm, Lackner and co-workers [2,3] asked participants to stand heel to toe (tandem Romberg stance) which increases ML sway with the right-hand lightly touching a waist-high force transducer on one

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side. The transducer was connected to provide an audible warning if the vertical force exerted by the participants exceeded 1 N. In one condition the eyes were open and in the other they were closed. Side-to-side sway in the frontal plane was reduced with light touch to the same low level whether the eyes were open or closed. In an additional condition ('force touch'), participants were allowed to use as much normal force as they wished, typically three or four times more than in light touch. This condition yielded low levels of sway similar to light touch. However, the two conditions differed in the correlation between shear force at the finger and ML CoP fluctuations. In the light touch condition, the peak correlation coefficient was +0.6 and the lag a little over 350 ms, with shear force at the hand leading CoP. In the force touch condition, the correlation was larger and the lag was smaller. The authors suggested that, in the light touch condition, the shear force at the finger drove postural corrections to maintain light touch with low normal force. The efficiency of this additional tactile feedback route, compared with normal proprioceptive and tactile channels available in the absence of vision, resulted in reduced sway. In contrast, it was suggested that the force touch condition reduced sway by physically stabilizing balance. However, the degree to which arm weight is more or less taken on the transducer does not itself determine whether the force condition results in mechanical stabilization. Instead, this depends on how stiff the kinematic chain is linking the arm to the body. We suggest that an alternative possible account of the differing lag between light and force touch is that the heavier contact provides clearer sensory information about sway allowing faster and more accurate compensatory balance adjustments.

The initial demonstrations by Lackner and colleagues of the benefits of light touch on balance involved the tandem Romberg standing posture, which is particularly unstable in the frontal plane. Subsequently, Clapp & Wing [4] obtained a similar effect of light touch on balance in the sagittal plane with normal bipedal stance. They also observed a positive correlation between hand shear force and CoP with lag of 350 ms in support of a tactile feedback loop reducing sway. If the effect of light touch on balance derives from additional feedback, reducing feedback should remove the benefits. Reginella *et al.* [5] asked participants in normal stance to make light touch contact with a vertical surface using the index finger. While the authors found that light touch contact reduced sway when the vertical surface was fixed, they found AP sway with contact surface that moved simultaneously with own AP sway, was as large as, or larger than, the sway with no contact. This result suggests that reducing the information about sway relative to the contact point provided by the tactile signal impaired the ability to compensate for AP sway. An even more direct approach to manipulate tactile feedback was taken by Kouzaki & Masani [6], who observed that sway reduction with light touch disappeared when the hand was anaesthetized using a compression block on the upper arm.

Further evidence of the utilization of tactile feedback from light touch in standing balance was demonstrated in a paradigm in which the contact surface was oscillated rather than being fixed. Jeka *et al.* [7] asked participants standing in tandem Romberg posture to place their finger lightly (vertical force less than 1 N) on a surface that oscillated sinusoidally with a peak velocity of 65 mm s^{-1} in the frontal plane at one of a set of frequencies ranging from 0.1 to 0.8 Hz (amplitude ranged from 18 to 2.25 mm). For oscillations up to 0.5 Hz, spectral analysis of the sway showed the presence of the contact surface oscillation frequency. This indicates that oscillation of the finger entrained the postural corrections in a 1 : 1 manner as would be expected if the tactile input were being used as a reference signal in maintaining posture. Given entrainment, it might be expected that this would have elevated sway variability relative to no contact, however, the study did not include a no-contact condition and so this possibility remains open.

Light touch reduction of sway does not necessarily need to involve the hand and arm but also occurs with 'passive' light touch in which an environmental referent, in the form of a fixed flexible contactor covered in soft textured material, touches the skin. Rogers *et al.* [8] asked participants to stand with flexible soft contact of this kind at the leg or shoulder. They found that sway was reduced in both cases, and was reduced more with the shoulder than with the leg contactor, presumably because a given degree of sway results in greater variation in force, or excursion of the contactor, at the higher point on the body. Furthermore, Rogers *et al.* [8] found that the sway reduced more when both contactors were applied together, indicating summation of information from the two sources. In analogous manner, greater reduction of sway with two compared to one contact point was also demonstrated by Dickstein [9] for 'active' light touch with two versus one index finger making light touch contact.

In the studies reviewed so far, the participants' task was to stand quietly. However, standing balance is often assessed by the response to dynamic perturbation involving movement of the support surface [10] or a push to the body [11]. It is therefore interesting to ask whether light touch benefits recovery from dynamic perturbations to balance. Johannsen *et al.* [12] examined the effect of passive light touch at the left shoulder on the response to forward sway owing to balance perturbations produced either by voluntary, self-initiated or involuntary, experimenter-imposed pull on the right-hand tending to cause forward sway. They found that balance was restored faster (earlier return of sway to pre-perturbation levels) with passive light touch, both after self-initiated and experimenter-imposed perturbation.

In light touch-assisted balance, the contact surface need not be rigidly fixed to reduce sway. Thus, Rogers *et al.* [8] reported that there was a light touch effect, even in the more variable sway condition (standing on foam) in which there was sliding of the shoulder level contactor. Moreover, Riley *et al.* [13] observed that sway was reduced by contact with a curtain, although this benefit of light touch contact was obtained only when the participant's attention was drawn to keeping the finger in contact without disturbing the curtain. This study illustrates that the forces involved in the light touch effect can be very light. Indeed, in a study by Backlund Wasling *et al.* [14], reduction in sway was obtained with an air jet directed at the pad of the index finger providing the light touch contact. This demonstrates that variation in the spatial location of contact without any shear force is sufficient to achieve the light touch effect. However, in cases of low or zero force it may be that the attention demands of using the light touch information are increased. Thus, Vuillerme *et al.* [15] showed that sway reduction using light touch with a curtain leaves less attentional capacity for a secondary auditory detection task (using reaction time as a measure) compared to standing with full vision, which would, presumably, have yielded comparable levels of sway. Light touch contact with a rigid surface might then be expected to be less attention-demanding than curtain contact, although, to our knowledge, this has not yet been tested. Moreover, other finger touch conditions that result in enhanced reduction in sway, such as force contact (cf. [3]) or holding the finger clipped to the contact surface [16] might also reduce the attention demands compared with light touch contact.

Recently, we have shown [17] that, even a moving contact surface, such as that provided by touching another person, who is also swaying of course, enables reduced sway with light touch contact. Participants stood side by side on two separate force plates with elbows flexed and the right or left hand held out forward of the trunk. They were instructed to stand still and, in different conditions, make light contact with each other's index fingers, with a fixed surface, or no contact. The interpersonal light touch (IPLT) contact condition resulted in a reliable reduction in sway compared with no contact, albeit the effect was smaller than that achieved with contact with the fixed support. One interpretation of this result is that, in IPLT, information about own sway from light touch contact is degraded by the contact surface movements owing to the sway of the other person. However, another aspect of the task, the need to follow the sway of the other person to maintain contact, may also be a factor in sway being greater than that with a fixed point of contact. This view is consistent with the finding that the ground reaction forces of the two participants were reliably correlated in the IPLT contact condition. Recently, we obtained further support for the role of contact point movement in a study [18] which included a shoulder to shoulder IPLT condition. We found that under conditions in which one person was in stable bipedal stance and the other was in unstable tandem stance, IPLT contact increased the sway of the person in stable stance and decreased that of the other person in unstable stance.

(b) *Outstanding issues*

In summary, the various studies of light touch suggest that force and position information associated with light touch contact provide cues to balance that can be combined with other cues, such as visual or vestibular information, to determine the current postural state and take action to move towards a desired, more stable, state. The touch information appears to be used to control body sway more efficiently than if there were no contact. The efficiency may be owing to more accurate, or earlier, motor commands owing to improved sensory information indicative of own body sway. Factors affecting that efficiency include contact attributes (e.g. spatial location, active versus passive, force level) and also attention. Sway can be entrained by oscillating light touch contact, and, if there is IPLT contact with another person, whether sway increases or decreases appears to depend on the relative stability of each individual.

Sway reduction with IPLT is a very interesting finding as this form of contact between people is widely observed, for example, in holding hands. Of course, the form of finger contact commonly involves a full grasp rather than the single finger contact of the laboratory paradigm. However, in holding hands, the grip and both partners' arms generally appear relaxed, so it is perhaps not unreasonable to suppose that there is normally very little net force between partners and that this joint posture would probably meet the criterion of 'light touch'. Given that IPLT in standing is so common, it is surprising that relatively little is known about the parameters of sensing and control in IPLT and the contribution to balance brought by this mode of light touch contact.

To investigate processing of finger light touch for arm movement and balance, we now describe an experiment in which participants, with eyes closed, made light touch contact between the index finger tip and a haptic device. Unlike most of the studies described above, in which the contact point could move, we followed the study of Krishnamoorthy *et al.* [16] in using a form of contact in which the finger was restrained or clipped to a contact surface. Their apparatus exerted a (normal) pinch force on the finger of 14 N, but with less than 1 N (tangential) pull force by the clip on the finger. As noted earlier, with this form of tight and light contact, Krishnamoorthy *et al.* [16] found that there was a greater reduction in sway than the normal light touch contact. We chose this 'light *tight* touch' form of contact in the hope of improving the efficiency of the linkage between somatosensory input and balance output. We therefore predicted positive cross-correlations between hand and ground reaction force with hand lead time shorter than the previously described 350 ms [3,4]. We measured postural sway (rate of displacement of a C7 marker, dC7) and balance adjustments (rate of change of centre of pressure, dCoP) as a function of a range of actuator movement conditions comprising: (i) No-contact control condition, in which the arm was held with the haptic device thimble on the finger but not connected to the drive mechanism, hence there was no force feedback. (ii) The Earth-fixed reference condition, replicating Krishnamoorthy *et al.* [16], which we expected would result in significantly reduced sway. (iii–v) Sinusoidal trajectories of the haptic device with frequencies of 0.3, 0.5 and superimposed $0.3 + 0.5$ Hz, to replicate and extend the study of Jeka *et al.* [7]; we expected that entrainment in conditions (iii–v) would result in greater sway than in (i,ii). (vi) Biological movement contact condition with haptic trajectory selected from a set of trajectories sampled from other individuals in the no-contact control condition. We considered condition (vi) as capturing one element of IPLT, extending the work of Johannsen *et al.* [17], and predicted that sway would be greater than the fixed condition (ii) but less than in the no-contact condition (i) and the moving support conditions (iii–v) because the non-periodic reference signal would not entrain sway.

2. METHODS

(a) *Participants and procedure*

Nine participants (mean = 26.3 years, s.d. = 4.6 years; five females and four males, all right-handed for writing) were tested while standing in stockinged feet on a force plate (Bertec 4060H, OH, USA) in normal bipedal stance with 5 cm inter-heel gap, eyes closed and head facing forward. The force platform measured the six components of the ground reaction forces and moments to determine the AP and ML components of the centre of pressure (CoP). Participants were instructed to stand as still as possible in a relaxed manner without speaking and eyes closed. The right arm was extended with the elbow in contact with the torso at waist level while the left arm was brought across the stomach so that the other hand made contact with the crook of the extended arm. Written informed consent was obtained from all participants and the study was approved by the University of Birmingham Ethics Committee.

Body sway was recorded in six experimental conditions. In each condition, the participant's dominant index finger was kept in the thimble of a haptic device (PHANToM 1.5 Sensable Technologies, MA, USA). In all except the first condition, which served as a no-contact control condition, the thimble was engaged with the haptic device which was located 40 cm in front of the participant. The hand was held in pronation and the index finger was extended but relaxed so that passive movements of the finger joints by the haptic device were still possible. A virtual plane was implemented at the hip level beyond which the thimble could not be lowered without deliberate effort against a resisting spring force updated at 200 Hz. Participants were instructed to maintain constant 'light touch' on this plane. The thimble was free to move in the ML direction as well as upwards away from this planar barrier. The haptic device, however, controlled the thimble's position in the AP direction in an open-loop mode according to a pre-specified trajectory. The haptic device's force output was always limited to a maximum of 1 N and therefore the thimble could deviate from this trajectory if participants generated a force larger than 1 N.

In each of the five experimental conditions, the haptic device produced one of a number of different categories of thimble trajectory. In the remaining condition, the index finger was also kept in the thimble (mass 19 g) detached from the haptic device. This condition (i) served as a 'no-contact' (no force feedback) control condition with an equivalent 'finger in thimble' sensation. The five conditions with haptic stimulation were: (ii) thimble held by the haptic device at a constant position ('stiff' haptic device) with spring-like force feedback directed towards the specific location on every axis (spring stiffness 0.5 N mm^{-1}); (iii) sinusoidal 0.3 Hz oscillation; (iv) sinusoidal 0.5 Hz oscillation; (v) superimposed 0.3 and 0.5 Hz oscillations (SP); (vi) biological movement with playback of thimble movements during the no-contact control condition of one randomly chosen trial from each of five other individuals who were not taking part in the experiment (BL). The amplitude of the haptic device trajectory in each of the conditions (ii–vi) was scaled so that the standard deviation of the thimble position matched that in the no-contact condition (i). The average peak-to-peak amplitude across the participants was 8.2 ± 2.8 mm. Each experimental condition was tested five times for a total of 30 trials. The no-contact condition was tested first in a block of five trials. The order of the remaining 25 trials was fully randomized. The duration of a single trial was 63 s, however, the first 2 s and the last second were removed so that only 60 s were analysed in each trial. Figure 1 shows the set-up and an illustrative trace for the stimulus input and the sway for each experimental condition.

(b) *Data reduction and statistical analysis*

Data from the force platform, the haptic device and body movements at C7 captured by optical motion tracking (Qualisys Oqus, Sweden) were sampled at 200 Hz. Force platform recordings were processed to determine AP and ML components of CoP fluctuations. All data time series were smoothed using a 100 ms moving average window and differentiated to yield rate of change measures of sway (dC7, dCoP) and thimble velocity. Within-trial estimates of sway (s.d. dC7, s.d. dCoP) were subjected to ANOVA with experimental condition as within-subject factor. Significance levels were set at $p = 0.05$ after Greenhouse–Geisser correction. The coupling between thimble movements and sway in the five conditions (ii–vi) involving the haptic device was analysed by calculating cross-correlation functions in the AP and ML directions. Cross-correlation functions were computed for time lags ranging from +3600 ms (haptic device leads) to −3600 ms (sway leads). The largest absolute cross-correlation coefficient and corresponding time lag were extracted. The cross-correlation coefficients were Fisher-Z-transformed [19] and also subjected to ANOVA with experimental conditions as within-subject factor.

In order to quantify entrainment of body sway in the oscillating haptic stimulus conditions, spectral analysis was performed on the thimble of the haptic device ('driving signal') and also on the C7 and the CoP position time series. For each variable, the fast Fourier transform (FFT) was calculated with a window length (8192 data points) of the nearest power of 2 smaller than the number of total data points per trial, which resulted in a frequency spectrum with a step size of 0.0244 Hz per bin. Frequency bins from 0 to 0.1 Hz were excluded from the subsequent frequency peak extraction algorithm to avoid the inclusion of slow drift effects commonly observed in quiet normal bipedal standing. Three frequency ranges ('harmonics') based on the FFT of the

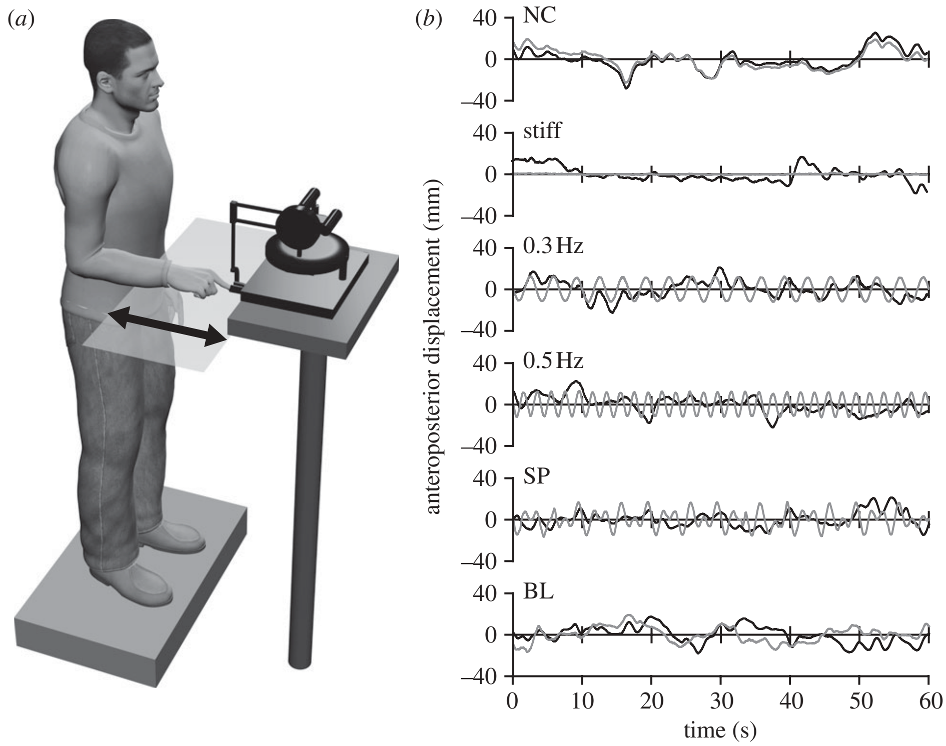

Figure 1. (*a*) The experimental set-up consisting of a participant in normal bipedal stance with the index finger of the right-hand inserted in the thimble of a PHANToM 1.5 haptic device. The haptic device imposed six types of movements onto the finger along the anteroposterior axis (shown by the double-headed arrow). (*b*) Illustrative trajectories and sway (C7) fluctuations for a single trial in each experimental condition for a single participant. The light grey line shows the position change of the finger imposed by the haptic device. The black line represents C7 fluctuations during the haptic stimulation. In the stiff condition, a small amount of movement in the thimble was observed owing to the spring constant (=1 N) used to prevent thimble displacement. NC, no contact; SP, superimposed periodic oscillation (0.3 + 0.5 Hz); BL, playback of spontaneous biological thimble movements previously recorded from an individual not exposed to any resisting forces by the haptic device.

driving signal from the lowest frequency condition (0.3 Hz) were defined for extraction of the respective peak frequency within each range. First, the frequency bin with the peak magnitude for the driving signal was found. Second, the range of the first harmonic was set up by adding and subtracting half the peak bin position. Then, the range for the second harmonic was defined by adding the width of the first harmonic range to the nearest frequency bin greater than the upper bound of the first harmonic and the same was done for the third harmonic range based on the upper bound of the second harmonic range. For every trial, the local peak frequency bin for the driving stimulus was located and the FFT magnitude over this bin as well as the phase were recorded within each of the three frequency ranges. For C7 and CoP, the FFT magnitude and phase were extracted from the same peak frequency bins identified for the driving stimulus. Finally, the frequency bin with the greatest absolute FFT magnitude was identified as the 'primary frequency' and the second largest was identified as the 'secondary frequency'. All data analysis was performed in MATLAB 7.5 (MathWorks, Natick, MA, USA) and SPSS 16 (IBM Corporation, Somers, NY, USA).

3. RESULTS

(a) *Thimble force and movements*

We first characterize the thimble force and movement parameters, as they can be affected by both the original driving stimulus provided by the haptic device but also by the stiffness parameters of the participants' contacting finger. No difference was found between the experimental conditions with respect to the average force exerted by the haptic device, in contrast to variability of force where differences were apparent ($F_{4,32} = 8.77$, $p = 0.003$, $\eta^2 = 0.52$). The s.d. of force was lowest for both the stiff and 'biological' conditions (mean, $M = 0.25$ N, s.d. = 0.13) with a significant increase for the remaining conditions comprising the simple and complex oscillations ($M = 0.43$ N, s.d. = 0.14; $F_{1,8} = 32.83$, $p < 0.001$, $\eta^2 = 0.80$). Also, peak thimble velocity tended to differ between the experimental conditions ($F_{4,32} = 3.24$, $p = 0.09$, $\eta^2 = 0.29$). Peak thimble velocity was lowest in the stiff condition which was induced by the participants' own finger movement exerting a force that temporarily exceeded a maximum 1 N of the haptic device ($M = 10.5$ mm s^{-1}, s.d. = 4.7), followed by the biological condition ($M = 22.0$ mm s^{-1}, s.d. = 9.4), the 0.5 Hz condition ($M = 38.7$ mm s^{-1}, s.d. = 24.7), the 'superimposed' condition ($M = 41.3$ mm s^{-1}, s.d. = 22.7) and the 0.3 Hz condition ($M = 42.7$ mm s^{-1}, s.d. = 41.3). Finally, the experimental conditions were different with respect to the s.d. of thimble velocity ($F_{4,32} = 10.0$, $p = 0.009$, $\eta^2 = 0.56$). Thimble velocity was least variable in the stiff condition ($M = 2.5$ mm s^{-1}, s.d. = 1.13), followed by the biological condition ($M = 5.9$ mm s^{-1}, s.d. = 2.6), the 0.3 Hz condition ($M = 12.2$ mm s^{-1}, s.d. = 4.8) and the superimposed

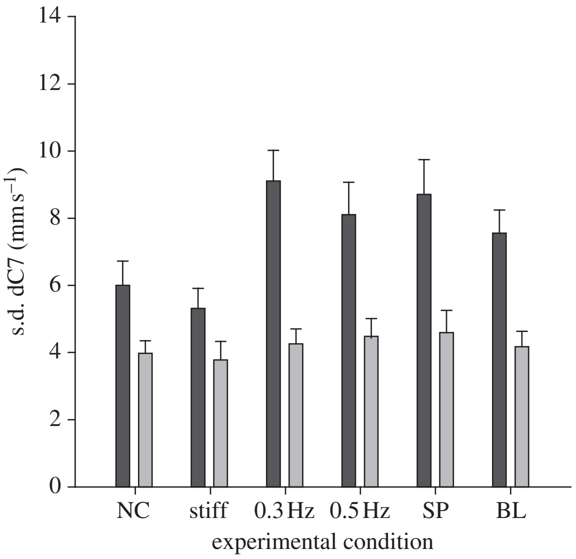

Figure 2. Sway (s.d. of C7 rate of change, s.d. dC7) shown as a function of experimental condition and sway direction. Error bars indicate the standard error of the mean. Black bars, anteroposterior; grey bars, mediolateral.

condition ($M = 14.7$ mm s^{-1}, s.d. $= 10.3$). Variability of thimble velocity was greatest in the 0.5 Hz condition ($M = 15.6$ mm s^{-1}, s.d. $= 11.8$).

(b) *Body sway*

Sway analysis focused on dC7 variability. Sway in terms of s.d. dCoP is relegated to the electronic supplementary material as the results were broadly similar to s.d. dC7. Figure 2 shows dC7 sway for each experimental condition in both directions of sway. Sway was greater in the AP direction ($F_{1,8} = 33.28$, $p < 0.001$, $\eta^2 = 0.81$). In the AP direction, participants' sway showed the greatest variability in the 0.3 Hz condition, where s.d. dC7 increased by 65 per cent compared with the no-contact condition. However, in the ML direction, the sway increased by only 14 per cent. The least amount of sway occurred in the stiff condition, when the haptic device was set to keep the thimble at a specific target position. The relative sway reduction in the stiff condition was 9 per cent in the AP and no change (0.4% increase) in the ML direction compared with the no-contact condition. There was a main effect of experimental condition ($F_{5,40} = 5.79$, $p = 0.008$, $\eta^2 = 0.42$) and a significant interaction between experimental condition and sway direction ($F_{5,40} = 6.31$, $p = 0.004$, $\eta^2 = 0.44$). In the AP direction, *post hoc* comparisons against the stiff condition revealed significantly greater sway for all conditions (all $F_{1,8} >= 5.59$, all $p <= 0.05$, all $\eta^2 >= 0.41$). Furthermore, *post hoc* comparisons against the no-contact condition resulted in a significant increase in sway for the 0.3, 0.5 Hz and superimposed conditions (all $F_{1,8} >= 6.65$, all $p <= 0.03$, all $\eta^2 >= 0.45$) with a tendency for an increase in the biological condition as well ($F_{1,8} = 3.47$, $p = 0.10$, $\eta^2 = 0.30$). In the ML direction, no changes in sway were found compared with either the stiff or the no-contact conditions.

(c) *Cross-correlations*

Figure 3 shows illustrative cross-correlation functions between dC7 and thimble velocity, averaged across all five trials, as well as overall peak cross-correlation coefficients for each of the experimental conditions on AP and ML axes. Concerning dC7 on the AP axis, cross-correlation functions show a single positive peak with slightly positive phase lag, indicating that the haptic device led the sway of the participants during the stiff and the biological conditions. In the periodic oscillation conditions, a positive peak with a slightly positive phase lag can also be seen. The cross-correlation functions, however, show gradually damped oscillation as the lag departs further from the peak. Peak correlation coefficients between the thimble velocity and both sway measures were exclusively positive indicating an in-phase coupling between the two variables. On the ML axis, cross-correlation functions for dC7 showed a single peak at either short positive or negative phase lags, depending on the experimental conditions. Peak coefficients of the cross-correlation function between dC7 and thimble velocity were significantly lower for the ML direction ($F_{1,8} = 51.29$, $p < 0.001$, $\eta^2 = 0.87$). Correlations with dC7 showed no differences between the experimental conditions, although the interaction between experimental conditions and sway direction tended towards significance ($F_{4,32} = 2.87$, $p = 0.08$, $\eta^2 = 0.26$). For the AP axis, *post hoc* comparisons between the stiff and the remaining experimental conditions revealed a marginally lower correlation coefficient for the 0.5 Hz condition ($F_{1,8} = 4.27$, both $p = 0.07$, all $\eta^2 = 0.35$). Peak correlation phase lags were statistically different between the two sway directions for dC7 ($F_{1,8} = 15.30$, $p = 0.004$, $\eta^2 = 0.66$). In the AP direction, dC7 significantly lagged behind thimble velocity by 176 ms (s.d. $= 92$; $t_8 = 5.73$, $p < 0.001$). On the ML axis, average time lags were not reliably different from zero ($M = 62$ ms, s.d. $= 160$).

(d) *Periodic haptic entrainment*

Figure 4 shows the FFT magnitudes for both the position of the thimble of the haptic device and C7 as a function of each of the three periodic driving stimuli as well as the primary and secondary frequency for the 0.3, 0.5 Hz and superimposed conditions. For the 0.3 Hz condition, the thimble consistently showed a primary frequency at 0.29 Hz and for the 0.5 Hz condition, the primary frequency lay at 0.40 Hz. The superimposed condition showed primary and secondary frequency components at 0.28 and 0.45 Hz, respectively. FFT magnitudes at the driving frequency were significantly lower for C7 than for the driving stimulus ($F_{1,8} = 6.58$, $p = 0.03$, $\eta^2 = 0.45$).

4. DISCUSSION

Light touch contact in which the hand is held with controlled force against an environmental surface stabilizes balance [3]. To investigate processing of finger light touch for the control of arm movement and balance, we have described a novel experimental paradigm in which participants stood with eyes

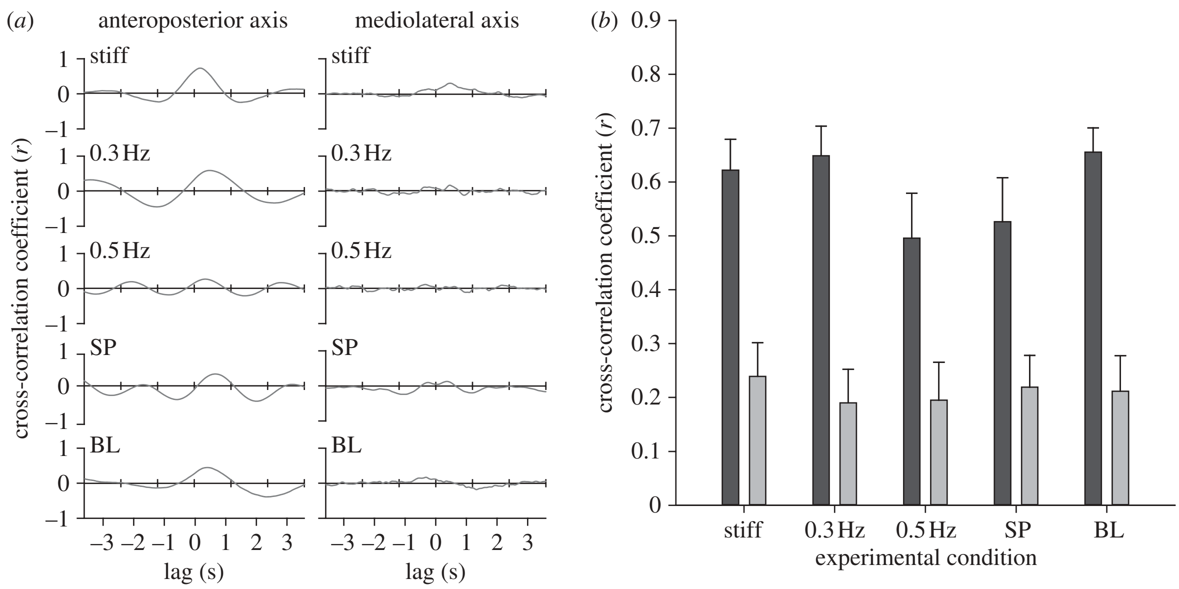

Figure 3. (*a*) Illustrative cross-correlation functions between dC7 and thimble velocity, averaged across the five trials, for each experimental condition and sway direction. Negative phase lags indicate a lead of dC7, while positive phase lags indicate a lead of the thimble velocity. (*b*) Peak correlation coefficients as a function of experimental condition and sway direction (black bars, anteroposterior; grey bars, mediolateral axis).

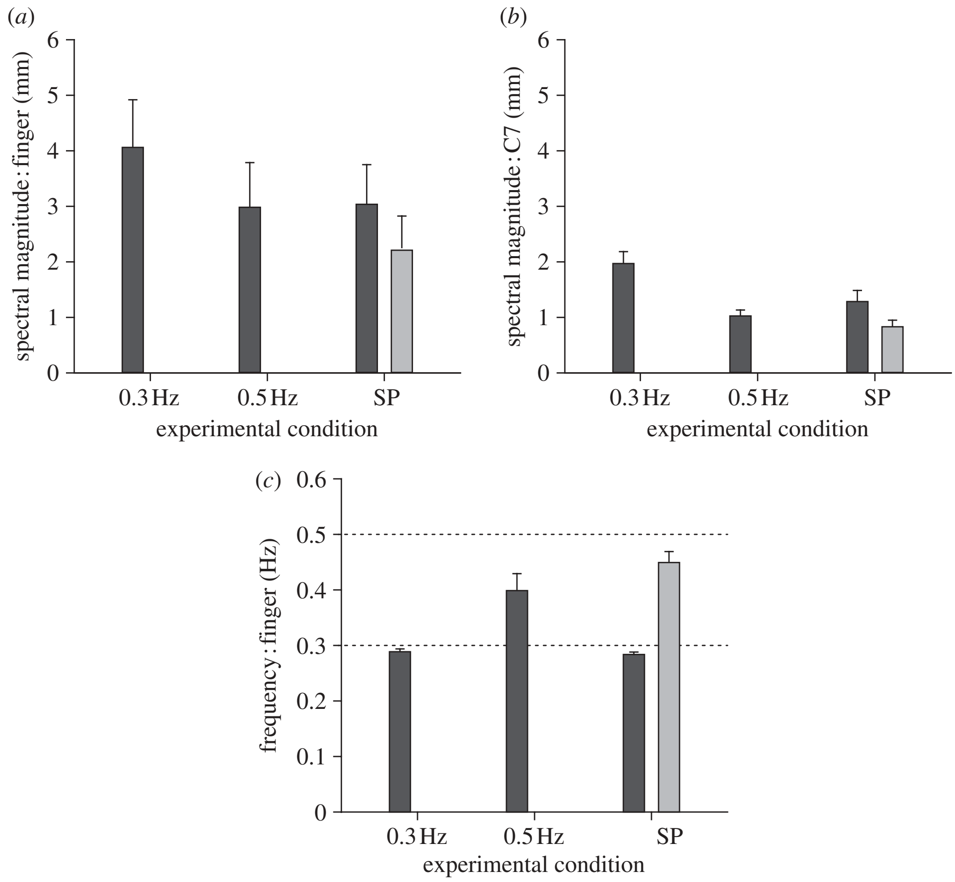

Figure 4. Peak spectral magnitudes over the respective frequency bin for the displacement of the thimble of the haptic device (*a*) and C7 fluctuations (*b*) for the three periodic oscillating conditions. (*c*) Frequency bin of the primary (black bars) and secondary (grey bars) frequencies embedded in the movements of the driving stimulus. The dotted line indicates the idealized target frequencies from which deviations were possible depending on the force exerted by the participant on the thimble.

closed and with the index finger tip placed in the thimble of a haptic device creating light tight contact. We examined two sets of conditions, one set providing balance cues expected to entrain balance and increase sway, the other set providing balance cues expected to reduce sway.

In the conditions expected to entrain and increase sway compared with the no-contact condition (finger still in the thimble, but the thimble disconnected from the haptic device), lower (0.3 Hz) and higher (0.5 Hz) frequency sinusoidal movement of the haptic device resulted in an increase in sway (s.d. dC7). However, in both sinusoidal movement conditions, the analysis comparing frequency components of the thimble of the haptic device and of the sway, showed evidence of entrainment, with similar proportions of power evident in C7 relative to the thimble for both conditions. In the condition with the 0.3 Hz frequency present in the motion path of the thimble, the greatest increase in sway was seen compared with the no-contact condition. Also in the superimposed condition where both 0.3 and 0.5 Hz were present, there was evidence of entrainment by the component frequencies, with similar proportion of power in the combined condition as for the low- and high-frequency conditions. Thus, our results with light tight touch contact replicate and extend the study of Jeka *et al.* [7].

In the conditions with light tight contact expected to reduce sway compared with 'no contact', we observed a reduction in sway in the stiff condition (finger held fixed on all three axes) only. In the biological condition (haptic device replayed the previously recorded sway path of another participant), sway was increased compared with the stiff condition but not as much as in the 0.3, 0.5 Hz or superimposed conditions. This finding of an increased level of sway is in contrast to the result we previously obtained for actual IPLT [17]. While the previous study used a real person as a partner, the haptic device in this experiment produced a pre-recorded 'other' trajectory in an open-loop manner. That is, in the present study, the participant had sole responsibility (as opposed to sharing responsibility in the previous study) of following the movement in order to keep the contact force light. This might have somewhat elevated the sway level in the present study. It also shows, however, that the postural coordination between two partners with finger light touch contact may be subject to more complex internal dynamics than just the passive entrainment by the other person's sway.

In addition to examining sway across all conditions, we also evaluated cross-correlation between the velocity of the haptic device thimble and dC7. In the AP direction, we found strong correlations in the same direction with the thimble velocity leading by 176 ms. This value is considerably less than the previously reported leads of 350 ms. We attribute this to the tight contact between the finger and the thimble resulting in a clear sensation, which Krishnamoorthy *et al.* [16] previously noted appeared more efficient than light contact in reducing sway, although they did not report cross-correlations and lags. The latency value obtained is longer than the 120 ms typically associated with postural reflexes [20], which are usually identified with supraspinal pathways. However, they are in the range of haptic reaction times of 140–190 ms reported in the context of a manipulator control task [21]. Compared with the manipulator task, it would be reasonable to suppose that some additional time would be required for cortically mediated transformations mapping hand coordinates to whole body posture to render effective the postural adjustments evidenced in the dC7 fluctuations.

In all experimental conditions, except for the stiff condition, the driving stimulus provided by the haptic device was along the AP axis. In the stiff condition, a 1 N force was applied on all axes in order to resist any movements of the thimble away from the set position. In general, we would have expected greater cross-correlation coefficients than the ones presently seen in all conditions if ML thimble velocity were determined entirely by ML dC7 sway as suggested by the near-zero phase lags. Thus, it may be possible that the oscillating stimulus on the AP axis also affected ML sway.

The present study demonstrates an advanced methodology for probing the interaction between time-varying tactile stimulation at the index finger and continuous sway adjustments during upright standing. The use of such a programmable haptic device allows, not only the application of complex periodic oscillations in open-loop mode, but, in future could also be used for closed-loop interactions between the haptic device and a participant. For example, the haptic device could be programmed to respond adaptively to the position information from the participant in order to simulate the feedback control understood to be used by the participant. Such a paradigm might be extended further to a two-person task by functionally linking two haptic devices, each one serving as light touch contact for one person, passing information about the sway of the other person. The experimenter could then manipulate the virtual linkage between the two participants to better explore IPLT and its effects on balance.

In conclusion, we have reviewed an active field of research in which light touch contact contributes to the maintenance of stable balance. We have presented a new paradigm which allows the nature of the touch stimulus to be manipulated to increase or decrease sway compared with no-contact conditions. Using the paradigm, we have demonstrated reduced sway with fixed light tight contact compared to no contact, and increased sway when the light tight contact was subject to simple or superimposed sinusoidal oscillations or when it reproduced the sway path of another person. Cross-correlation between finger motion and sway in the AP direction was higher and showed shorter lags than in previous studies in which the contact was light (and potentially free to move) rather than held tightly (but not allowing AP forces above 1 N) as in the present study. We speculate the difference may reflect more efficient (and less attention demanding) processing in using the somatosensory input at the hand. In current studies, we are using this approach to explore quantitative models for the exchange of information between two people who allow joint improvement of their balance in IPLT.

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