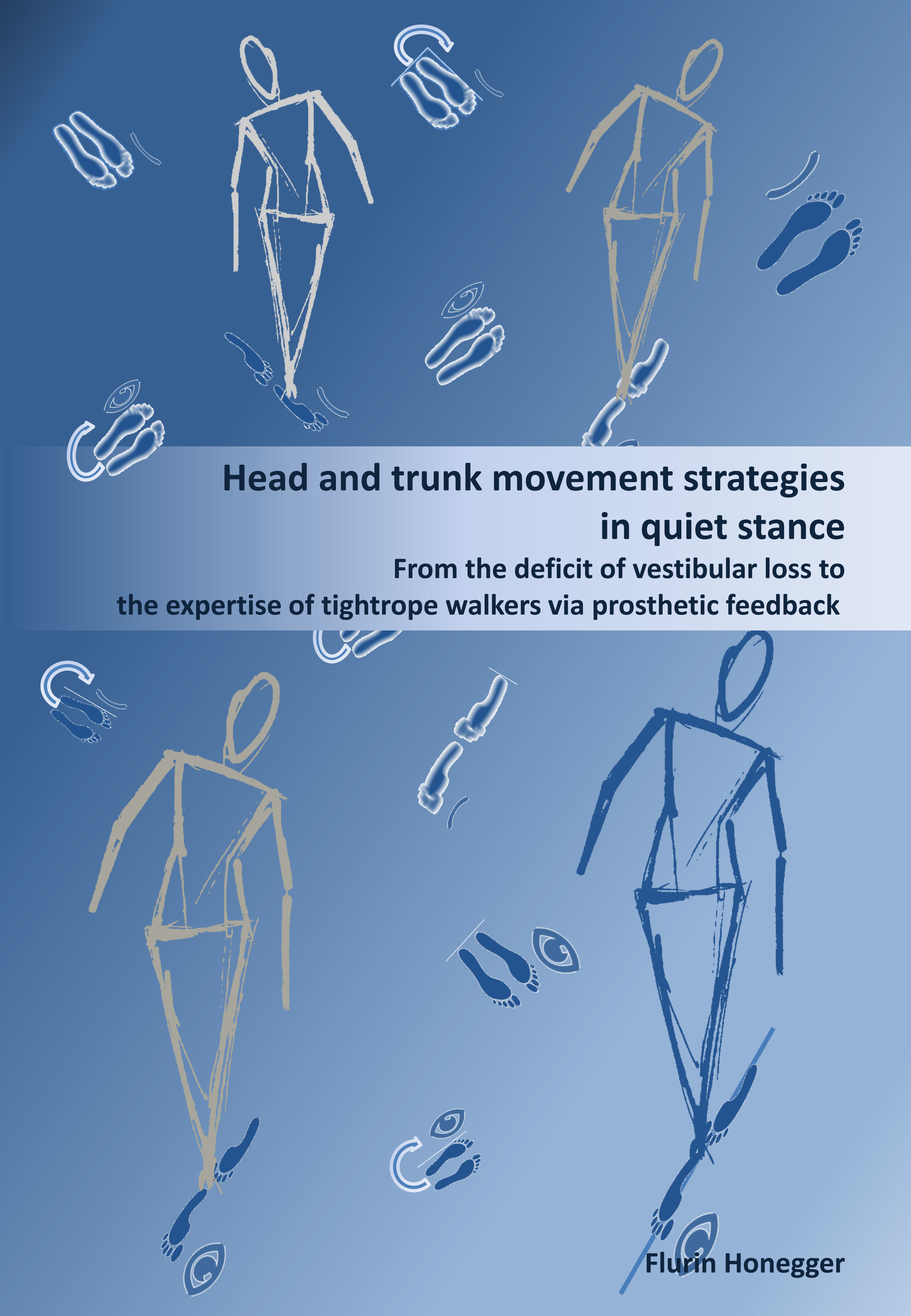

Head and trunk movement strategies in quiet stance

From the deficit of vestibular loss to the expertise of tightrope walkers via prosthetic feedback

Flurin Honegger

Head and trunk movement strategies in quiet stance: from the deficit of vestibular loss to the expertise of tightrope walkers via prosthetic feedback

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Summary

Summary

ABSTRACT

Is the head more locked to the trunk or stabilised in space during quite stance? Does prosthetic vestibular feedback have a positive impact on movement strategies and muscle synergies of those with vestibular loss? Does training in tandem stance lead to improved movement strategies and/or reweighting of sensory inputs? These questions have not been answered to date. This thesis attempted to answer these questions with appropriate, but new, techniques. The coordination of the head with respect to the trunk and pelvis during quiet, feet side by side, stance in normal and vestibular loss subjects was examined as well as the effect of prosthetic feedback on the strategies and synergies used by vestibular loss subjects. Changes in movement strategies and sensory feedback in tight-rope walkers with considerable training in tandem stance (one foot before the other), were also investigated.

Subjects performed the stance tasks under different sensory conditions: with eyes open or closed, and on either a firm or foam surface. Stance was either side by side stance or tandem stance. For one experiment, vibrotactile and auditory balance feedback of trunk sway was used in addition. Subject groups were bilateral vestibular loss (BVL) patients, trained tightrope walkers and age matched controls. Two further groups of young and elderly healthy subjects were used to characterise differences in head movements with aging. In all studies roll and pitch angular velocities were recorded with six body-worn gyroscopes; a set of two worn at the upper trunk, an identical set at the hips and another lighter set worn on a head band. In one study with BVL subjects, another of the lighter gyroscopes was strapped onto the lower leg. For the balance feedback study surface EMGs were recorded from pairs of antagonistic muscles at the lower leg, trunk and upper arm. Data from all experiments was analysed in both time and frequency domains. For the analysis of tandem stance an estimate of centre of mass movement was calculated as well as its time to reach a virtual stability boundary.

The results indicated that under most sensory conditions, two legged, feet side by side stance conditions, head sway at the head for both the roll and pitch direction is greater than at the upper trunk and the pelvis. For low and mid-frequencies (<0.3 Hz) the head is locked to the trunk i.e. there is a tendency for the head and trunk to move as one unit but the head movement is always more than expected from a pure inverted pendulum movement mode. For the BVL subjects the head on trunk locking is more rigid and characterized by higher resonant frequencies. Prosthetic feedback reduced pelvis sway angle displacements in BVL subjects to values of age-matched healthy controls for all stance tasks. Movement strategies in BVL subjects were reduced in amplitudes with feedback but otherwise not changed. Reduced amplitudes are achieved with improved antagonistic muscle synergies. As we observed with feet side by side stance, tandem stance is also multisegmental. Keeping balance while standing on a tightrope appears to require similar intersegmental movement strategies for the head, trunk and pelvis to those used with other, less difficult tandem stance tasks. The difference with respect to untrained normal subjects is that faster trunk movements are used by tightrope walkers as they explore the limits of the base of support. At the same time they reduce relative head and pelvis movements to those of the trunk via changed proprioceptive weightings.

KEY POINTS ADDED TO THE LITERATURE

Coordination of the head with respect to the trunk and pelvis in the roll and pitch planes during quiet stance

The head is locked to the trunk for low frequency motion possibly because motion is just supra-vestibular threshold. For movements above 3 Hz, head movements are anti-phasic with respect to the trunk. Significant age differences were not found.

Coordination of the head with respect to the trunk, pelvis and lower Leg during quiet stance after vestibular Loss

The data indicate that for quiet stance vestibular loss subjects change the characteristics of their head on shoulder motion, reducing relative motion of the head below 3 Hz, and shifting head resonances to higher frequency. Presumably these changes are accomplished with increased use of proprioceptive neck reflexes.

The effect of prosthetic feedback on the strategies and synergies used by vestibular loss subjects to control stance

This study is the first to demonstrate how vestibular loss subjects achieve a reduction of sway during stance with prosthetic feedback: with reduced and better controlled muscle synergies. Thus both body movement and muscle measures should be explored when choosing feedback variables and feedback location for prosthetic devices improving stability of those with a tendency to fall.

Movement strategies in tandem stance: Differences between trained tightrope walkers and untrained subjects

Standing in tandem stance is a demanding coordinative task. Training in the form of tight-rope walking causes a sensory reweighting in the neck and lumbosacral joints, probably as a result of higher trunk velocities used to explore the limits of stability. The similarity of tandem stance on foam to that on the tightrope indicates that the foam tasks could be used for effective training of tightrope walking.

Chapter 1

General introduction and aims of this thesis

Chapter 1

RESEARCH IN POSTUROL CONTROL

Studies on postural control are mostly driven by the believe that careful observation and analysis of body sway during quiet and perturbed stance provided valuable information to characterize and model changes in postural control due to age, pathology and athletic skill (Nardone and Schieppati, 2010). The focus is centred on improved characterization of balance control for clinical assessment and therapeutic purposes (Allum and Shepard, 1999, Nardone and Schieppati, 2010). If this focus can be appropriately directed then the goal preventing falls and accompanying injuries could be achieved.

BALANCE

The balance control system, involved in maintaining balance and preventing falls during our everyday activities, consists of multiple systems constantly acting and reacting with one another. This system is a substantial ruling component for most motor behaviour. Sensory inputs inform the CNS about the orientation of the body in space and about the actual movement environment. The main sensory inputs come from of the vestibular system in the inner ear, vision from the eyes, and sensory receptors in the muscles, skin, and joints. The CNS can then plan balance correcting strategies based on knowledge of the movement environment and prior practice. Thereby, the CNS orchestrates through the efferent system the execution of movement strategies by the musculoskeletal system.

Vestibular Cues

The vestibular contribution to postural control lies in the detection of head movements in space. The two laterally placed systems consist each of three roughly orthogonally oriented semicircular canals that sense angular acceleration and two otolith organs, the utricle and saccule, that sense linear acceleration or the gravity vector and together provide complimentary information (Angelaki and Cullen, 2008). Three types of vestibular reflexes are described in the literature. (Uchino and Kushiro, 2011). With the vestibulo-ocular reflex (VOR), visual inputs are kept sharp on the fovea during head movements (Cremer et al., 1999). The vestibulo-collic reflex (VCR) stabilizes the head in relation to the body (Roberts, 1978, Wilson et al., 1979) and vestibulo-spinal reflexes (VSR) act on the trunk and the limbs to assert a safe erect stance (Keshner et al., 1987).

Visual Cues

Visual cues effect posture through anticipatory information on postural demands. These also feed-back information of the own body's movement as well as that of the environment to the CNS. Though these cues have different thresholds for detecting motion (Brandt et al., 1973), both gaze fixation and peripheral vision influence stabilization of posture (Brandt et al., 1973, Stoffregen, 1985, Straube et al., 1994, Nougier et al., 1998, Berencsi et al., 2005, Slaboda et al., 2013, Wright et al., 2013). The size of the visual field (Dijkstra et al., 1994a, Dijkstra et al., 1994b, Straube et al., 1994), contrast sensitivity and visual acuity (Lord et al., 1991, Anand et al., 2002, Lord, 2006), all influence the postural response. Apart from detecting motion, vision acts primarily as a reference system for the vertical (Nashner et al., 1982, Keshner et al., 1987).

Somatosensory Cues

The sense of the relative position of the individual body-segments (proprioception) is mediated through specialized neuromuscular sensors within the muscles, tendons and joints (Macgillis et al., 1983, Zarzecki et al., 1983). Proprioception mainly relies on the actions of the muscle-spindle receptors situated between skeletal muscle fibres, reacting to muscle length changes due to stretch or contraction (Matthews, 1977, 1982, 1986). It also relies on tendon and joint receptors (Macgillis et al., 1983, Zarzecki et al., 1983). The sense for the body's relation to surfaces or external objects is mediated by cutaneous receptors in the skin. These receptors in the soles of the feet are important for postural control (Diener et al., 1984, Magnusson et al., 1990a, Magnusson et al., 1990b) but the direct participation in balance control is debatable. All these receptors systems adapt with different time-constants which enables proprioception to detect the body's position both for quiet upright stance and during postural perturbations.

Sensory Integration and Weighting and Reweighting

From the foregoing it may be supposed that a person's ability to restore stable balance following a destabilization largely depends on how the central nervous system (CNS) executes motor action within the neurophysiological, mechanical and environmental constraints. One part of this action, sensory, integration concerns how the brain combines sensory input from multiple sensory modalities. The integration process which occurs within the CNS at different levels (Xerri et al., 1988, Jacobs and Horak, 2007) is an unconscious action (Teasdale and Simoneau, 2001, Fabbri et al., 2006, Siu and Woollacott, 2007, Werkhoven et al., 2009, Remaud et al., 2012) and is crucial when information from any of the sensory systems is unreliable (Shumway-Cook and Woollacott, 2000, Redfern et al., 2001, Redfern et al., 2009). This process competes with that of cognitive tasks and has a greater impact in older adults (Rankin et al., 2000, Shumway-Cook and Woollacott, 2000, Mendelson et al., 2010).

An example of this integration occurs with the vestibular system. It is incapable of determining without visual cues whether an acceleration takes place in one direction or a deceleration in the opposite direction (Xerri et al., 1988). When the body is moving at constant velocity, vision is the only reliable cue for determining motion (Allum et al., 1976, Waespe and Henn, 1977). Another example is when a combination of visual and vestibular and proprioceptive cues is necessary to distinguish between self-movements and that of the surrounding (Cullen, 2012). Yet another example is distinguishing between passive and active head movements. For these proprioceptive information is required (Angelaki and Cullen, 2008). It has been demonstrated that interaction between the sensory systems expands the working ranges but also can have a suppressive effect (Xerri et al., 1988, Brandt and Dieterich, 1999, Bronstein, 2004). The availability and reliability of the different sensory systems influences postural sway. Visual cues can reduce the effect of inappropriate proprioceptive and vestibular information (Fransson et al., 1998, Straumann and Bockisch, 2011). A comparable stabilization can be observed when a stationary object is touched during proprioceptive perturbations (Rabin et al., 1999, Lackner and DiZio, 2005, Rabin et al., 2006). Thus the influence that each of the individual sensory cue components exerts on postural control depends on the context. Each contributes with a certain «weight» relative to the context (Peterka, 2002, Peterka and Loughlin, 2004, Goodworth and Peterka, 2009, 2012). Adaption to the change of the importance of the individual sensory cues is termed «sensory reweighting» (Allison et al., 2006, Carver et al., 2006, Bair et al., 2007, Haran and Keshner, 2008, Fetsch et al., 2009, Rinaldi et al., 2009, Jeka et al., 2010, Cuisinier et al., 2011, Eikema et al., 2012, Goodworth and Peterka, 2012).

Chapter 1

POSTUROGRAPHY

Posturography refers to the techniques used to measure and quantify balance control in upright stance. The goal is to investigate the mechanisms of the sensorimotor system maintaining balance. In order to determine and probe the contribution of one or more of the sub-systems involved, balance control under various stance conditions is assessed. Measurements may include body kinematic data collected with accelerometers, gyroscopes, goniometers, or camera based position tracking of passive or active markers to yield CoM and CoG estimates, as well as measurements of ground reaction forces to yield centre of pressures measures (CoP) (Bloem et al., 2003).

Traditionally posturography techniques fall in to two main categories that are called static and dynamic depending on whether the support surface on which the subject stands moves or not (Nashner and Peters, 1990, Visser et al., 2008). Often both techniques are used in a complementary fashion to address a certain topic.

Static posturography focuses on stance conditions such as standing unperturbed on a firm or foam surface with eyes open or closed, and the feet aligned in different positions, with the focus on analysing and quantifying the balance controlling movement strategies that cope with self-initiated destabilizing movements (Horak, 2006). Recording times extend over several tens of seconds and some of the analysis techniques have been standardized (Kapteyn et al., 1983). Though not exclusively so, static posturography is founded on CoP measurements based on ground reaction re-cordings. Over the years, the analysis of CoP has been expanded in numerous experimental studies. The analysis techniques have become more complex and less applicable to clinical studies. Some of these studies (Collins and De Luca, 1993, 1994, 1995, Sabatini, 2000b, Delignieres et al., 2003, Duarte and Sternad, 2008, Teresa Blázquez et al., 2010) were inspired by the concept of ordinary and fractional Brownian motion (Einstein, 1905, 1906, Mandelbrot and Van Ness, 1968). Others used nonlinear time series analyses techniques, including recurrence quantification analysis (RQA), to characterize the deterministic features of CoP data (Yamada, 1995, Riley et al., 1999, Ladislao et al., 2006, Schmit et al., 2006, Basafa et al., 2007, Seigle et al., 2009). Another group focused on heuristic CoP decomposition, rambling and trembling (Zatsiorsky and Duarte, 1999, 2000, Krishnamoorthy et al., 2005, Mochizuki et al., 2006, Monteiro Ferronato and Barela, 2011, Shin et al., 2011, Tahayor et al., 2012, Sarabon et al., 2013, Slomka et al., 2013). Lastly, one group extracted dynamical invariants such as approximate entropies and Lyapunov exponents (Yamada, 1995, Newell, 1998, Sabatini, 2000a, Hong et al., 2006, Roerdink et al., 2006, Costa et al., 2007, Ladislao and Fioretti, 2007, Duarte and Sternad, 2008, Lamoth et al., 2009a) that have roots in the analysis of dynamic and particularly ergodic systems (Walters, 1982, Broer and Takens, 2011). Although highly technical, these studies analysed in general, whether the effects of a set of functionally relevant factors such as visual perception, cognitive task, disease, aging and athletic skills have characteristic imprints in CoP traces (Newell, 1998, Schmit et al., 2005, Ladislao et al., 2006, Schmit et al., 2006, Costa et al., 2007, Duarte and Sternad, 2008, Lamoth et al., 2009a, Seigle et al., 2009, Madeleine et al., 2011, Monteiro Ferronato and Barela, 2011, Ramdani et al., 2011, Shin et al., 2011, Sarabon et al., 2013).

Dynamic posturography is the study of human postural responses to sudden balance perturba-tions. The field has two core areas. In the first work is focused on CoP for clinical testing (Chaudhry et al., 2011), assessment and classification. In the second - multi-measurement posturography – the emphasis is on capturing body motion and muscle activity as completely as necessary to reveal, for example, the impact of neurological deficits on balance and postural control elicited by translating, lifting or tilting a platform along an arbitrary but well defined direction or axis (Allum et al., 1994, Carpenter et al., 1999, Tokuno et al., 2006). Dynamic posturography focusses on reflexes and balance-correcting muscle responses (EMG), analysis of physically interpretable entities such as trunk sway, and its observation time is limited to 1 or 2 seconds (Allum and Shepard, 1999, Visser et al., 2008, Nardone and Schieppati, 2010). A main assumption in the analysis, shared with the analysis of gait cycles, is that it is feasible to average responses from repeated stimuli once the first adapting responses have occurred (Keshner et al., 1987), in order to compare with normal responses and identify clinical abnormalities. For this, obviously the stimulus must be constant in timing and provoke a response with a clear onset and amplitude.

Another focus of dynamic posturography results from using procedures where the supporting surface is constantly tilted, rotated or translated periodically (Pyykko et al., 1991, 1995, Corna et al., 1999a, De Nunzio et al., 2005, Bugnariu and Sveistrup, 2006, De Nunzio et al., 2007, Fujiwara et al., 2007, Cappa et al., 2008, Nardone et al., 2010, Schmid et al., 2011). The modulation frequencies range from as low as 0.01 Hz (Vaugoyeau et al., 2007, Vaugoyeau et al., 2008) up to 24Hz (Pyykko et al., 1991). Analysis techniques depend on the stimulus frequency and amplitude. Particularly for low stimulus frequencies, correlation of body movements with the stimulus profile becomes difficult to interpret. Another problem is that analysis techniques based on averaging original traces as used by Vaugoyeau et al (2007, 2008) become inappropriate because the subtle multilink kinematics became averaged out with such techniques.

For several years body worn sensors have been employed to measure balance control during stance and gait - so called «stance and gait posturography» (Allum et al., 2001, Allum and Carpenter, 2005, Vonk et al., 2010) with protocols complementing standard clinical vestibular assessments (Vonk et al., 2010, Allum, 2012). Easy availability of mobile wireless devices with their own computing power combined with sensors shrinking in size and cost but with increased precision and improved noise, drift, and temperature compensation has opened the field of posturography to wearable sensors (Yang and Hsu, 2010). These sensors are started to influence posturography (Najafi et al., 2010, Lee et al., 2012). With all the advantages implied (mobile equipment, measurements not restricted to a specific room or light conditions, high sampling rates achievable) it is likely that next generation of body worn sensors will provide the necessary momentum in the field to push clinical routine posturography beyond the assessment of centre of pressure CoP (Chaudhry et al., 2011), compete with optical motion capture systems (Takeda et al., 2009), and open the door for real live monitoring (Scanaill et al., 2006).

SIDE BY SIDE STANCE

For normal two-legged side-by-side stance, different movement strategies have been proposed for pitch (AP) sway. One strategy that has attracted a lot of attention due to its simplicity. It is derived from the model of an inverted pendulum moving around the ankle joint, with ankle proprioception providing the main contribution to the postural control (Horak and Nashner, 1986, Fitzpatrick et al., 1992a, Fitzpatrick et al., 1992b, Fitzpatrick and McCloskey, 1994, Hsu et al., 2007). Thereby the theoretical impact from proprioceptive inputs arising from around the knee, hip, and lumbosacral joints is diminished as in Nashner et al. (1982) and Kuo et al. (1998). In contrast, Koozekanani et al. (1983) suggested upright stance is controlled in pitch by multi-segmental movement strategies. More recently Creath et al. (2005), Hsu et al. (2007), Pinter et al. (2008) and Horlings et al. (2009) have supported this suggestion. In-phase and anti-phase, for low frequency and high frequency, respectively, movements between the ankle, hip and lumbosacral joints have been shown to co-exist simultaneously in the sagittal plane, depending on the support surface (Creath et al., 2005, Horlings et al., 2009). Those for roll motion consisting in addition of a strategy of motion of the shoulders about a stable pelvis (Horlings et al., 2009). In view of these strategies, one possible way to reduce trunk sway would be to change from an ankle strategy to an anti-phase hip strategy in order to reduce the motion of the CoM.

Another way to lessen sway would involve reducing or changing the composition of sensory feedback used to generate joint torques (Peterka, 2002, Maurer et al., 2006, Goodworth and Peterka, 2009). Peterka et al. (Peterka, 2002, Maurer et al., 2006, Goodworth and Peterka, 2009) applied small 1 to 4 degrees continuous support-surface perturbations to stance and used the resulting CoM or trunk responses to argue, with the support of modelling techniques, that amplitude response non-linearities (relatively less sway with increasing stimulus amplitude) demonstrated sensory reweighting. This mechanism involves the CNS shifting its reliance on one sensory system (e.g. ankle proprioceptive) to another (e.g. vestibular) depending on the stimulus amplitude (Goodworth and Peterka, 2009), current task instability (Peterka, 2002), and presumably task difficulty. Thus for easier tasks with little spontaneous pitch or roll motion at the shoulders with respect to the pelvis, lumbosacral proprioceptive gains can be set high, but not when shoulder motion is large.

TANDEM STANCE

Most studies describe two-legged stance. Little is known about the movement strategies during tandem stance, though it is used in clinical balance assessments, in studies on balance performance (Nichols et al., 1995, Smithson et al., 1998, Lamoth et al., 2009b, Seino et al., 2009), and in studies on the fusion of different combination of sensory inputs such as touch, vision and hearing (Easton et al., 1998, Clapp and Wing, 1999, Rabin et al., 1999, Kiemel et al., 2002, Oie et al., 2002).

Winter et al. (1993) suggested that for normal side-by-side two-legged stance that the mechanisms maintaining stability in AP and ML (and medial-lateral) directions consisted of independent ankle and hip strategies, with the ML direction dominated by a hip strategy, and AP direction by an ankle strategy. Because in tandem stance the ankle axes are lined up in AP direction, with more instability in the roll plane due to the reduced base of support, Winter et al. (1996) postulated that postural control was then dominated by an ankle strategy for ML sway with a little contribution from the hip strategy. In contrast, the AP sway was dominated by a hip strategy with little contributions from the ankle strategy during tandem stance. However, several authors (Loram and Lakie, 2002, Morasso and Sanguineti, 2002) have questioned the assumptions used by Winter et al. (1996).

When somatosensory information is available, for example, with a firm support-surface, this information plays a role in the control of upright posture (Horak et al., 1990). For tightrope walking the support surface is not completely firm but slightly resilient. Furthermore, the foot is only supported by a small support base. When the foot base is shortened laterally this will lead to different effect on foot somatosensory inputs than shortening in the AP direction. For a short AP support base, Horak and Nashner (1986) demonstrated with dynamic posturography that subjects depend on a hip strategy, in which the trunk and the hip move anti-phasically, to maintain upright postural stability rather than an ankle strategy. Thus the question arises whether a different multi-joint motion also occurs in tandem stance with a shortened support base in the lateral direction, as occurs on a tightrope.

Lamoth et al. (2009b) investigated whether the level of athletic skill is reflected in the control of body sway. The subjects stood in a two legged tandem stance on a narrow plywood strip. They found that as the level of gymnastic skill increased, trunk acceleration variability decreased and stability increased, indicating a more efficient postural control. They found that the differences in skill did not depend on vision or proprioception inputs, suggesting that expert gymnasts exhibited differences in the underlying organization of postural control which appear to be independent of the specific form of sensory information used.

HEAD MOVEMENTS

Few studies have recorded head movements during either normal two-legged stance or tandem stance. Head movements are of interest because the vestibular system receives an input signally rotation of the head with respect to an earth-fixed reference system whenever the head is moved, be it with respect to the trunk, or in tune with ankle movements. Based on these two types of head movement, two theories about sensory control of upright stance have been proposed: top-down and the bottom-up control (Nashner et al., 1982, Maurer et al., 2000). The bottom-up theory, states that upright stance is controlled based on ankle proprioceptive inputs which are confirmed by vestibular inputs as it assumed that the body moves as an inverted pendulum. That is, the head is assumed to be locked to the trunk. When upright stance is controlled top-down; head movements are regulated to be fixed in space, using a combination of vestibular and visual inputs as an earth fixed reference, and the proprioceptive inputs from the neck, hips and ankle joint are used to control body sway. Keshner and Peterson (1995) suggested that the mechanisms underlying head and neck stabilization also included biomechanical components (particularly head inertia), voluntary control, as well as vestibular and proprioceptive neck reflexes. Thus the mechanisms which stabilize the head are related to both the frequency and stimulation direction (roll or pitch) of head motion relative to the trunk (Keshner et al., 1995). Generally through, pitch stabilization of the head in space (top down mode) rather than a head-fixed-on trunk mode appears to be the underlying movement strategy when the support surface is moved (Buchanan and Horak, 1999, Corna et al., 1999b, Akram et al., 2008).

PROSTHETIC BIOFEEDBACK

As indicated in the previous sections, properly integrated visual, vestibular and proprioceptive sensory information is essential in postural control (Day et al., 2002). Age-related declines in vestibular function, visual acuity, proprioceptive sensitivity, and brain function responsible for integrating and processing this information, may contribute to the deficits in postural control often seen in older adults (Bugnariu and Fung, 2007). Loss of vestibular function is well-known to lead to an increased tendency to fall in older persons(Tinetti et al., 1988). Biofeedback for purposes other than reducing a tendency to fall has been applied over 50 years and can be defined as a process in which a person learns to reliably influence two kinds of physiological responses: either responses which are not ordinarily under voluntary control, or responses which ordinarily are regulated, but for which regulation has broken down (Blanchard and Epstein, 1978). So it is not surprising that a number of investigators have developed devices to provide persons with balance problems a replacement for vestibular sensory information by augmenting or using sway information available through other sensory cues. Such «prosthetic systems» generally rely on auditory or vibro-tactile feedback or both, suitably coded with body sway information (Dozza et al., 2005b, Hegeman et al., 2005, Horak et al., 2009, Vichare et al., 2009, Davis et al., 2010, Allum et al., 2011, Goodworth et al., 2011). Opinions differ on the kind of feedback that should be used for an optimal balance prosthesis, but the conclusion that biofeedback has been shown to improve balance during stance and gait is shared among investigators (Dozza et al., 2005a, Hegeman et al., 2005, Wall and Kentala, 2005, Dozza et al., 2007, Horak et al., 2009, Allum et al., 2011, Goodworth et al., 2011). It is however unknown whether, for example, for quiet stance of bilateral peripheral vestibular loss (BVL) subjects, if a change in movement strategy parallels the reduction in sway or whether the increased stability is gained by means of decreased or increased muscle activity.

AIMS OF THE THESIS

Starting with the studies of Horlings et al. (2009) our group characterized the relationship between trunk and pelvis motion during quiet stance in healthy controls and those with vestibular or lower leg proprioceptive loss. A first goal of this thesis was to extend this analysis to include head motion. The study of Chapter 2 follows this aim by analysing head motion of healthy young persons and is guided by the following research questions.

- What is the relationship between head and trunk movements of the body during quiet stance?
- Are there differences with respect to the roll and pitch plane?
- Are head movements highly correlated with the trunk or the pelvis movements?
- Is head on trunk resonance observed?
- Are head movements with respect to the trunk altered under different sensory conditions?

Thereby I could fulfil a secondary aim of this thesis that body wearable sensors can be used to characterise body segment motion. In Chapter 3 the analysis is taken a step further into a comparison of bilateral vestibular loss (BVL) patients with that of age-matched healthy controls. In addition, lower leg information was added to the analysis. This chapter aimed to answer the following three research questions,

- How is the relationship between head and trunk movements in the pitch and roll planes altered during quiet stance under different sensory conditions following vestibular loss?
- If head resonance is observed, is it changed in frequency with vestibular loss?
- How is the relationship between pelvis and leg movement of the body altered during quiet stance following vestibular loss?

With completion of the first two studies we had a characterization of normal and vestibular loss motion strategies during quite stance as a basis for improving motion control with prosthetic feedback. In the study of Chapter 4 changes in stance movement strategies and muscle synergies were examined when bilateral peripheral vestibular loss (BVL) subjects were provided prosthetic feedback of pelvis sway angle. The chapter was focused on bringing more light into the following research questions.

- Do patients need to use a particular type of movement strategy for effective sway reduction?
- Are improvements in balance control using artificial sway position feedback achieved by BVL subjects using the same strategy as without feedback?

- How are muscle synergies changed when biofeedback of pelvis sway is provided to BVL subjects?

Standing on a tightrope is probably the most difficult task imaginable. But the problem we faced is how to compare performance in this task to balance control of untrained (on the tightrope) normal subjects. We could hardly place the untrained on the tightrope. A start was to bring about lateral instability, and simultaneously increase the AP stability, by asking subjects to stand in the tandem stance position, just as tight-rope walkers do. Then, the base of support would be shortest in the lateral direction. Also the tightrope is not a normal support surface, but gives way. Here we used a foam support surface as a «safe» alternative. Finally, we made the situation more difficult by having the subjects close their eyes. In this way we created in pilot experiments an ideal comparison situation to stance on the tightrope With this setup I could pursue the aims of Chapter 5 which were to investigate the movement strategies of trained tight rope walkers with those of age matched healthy controls. This chapter was guided by the following set of questions.

- Does skill with a difficult balancing task such as tightrope walking lead to improved balance for similar but easier tasks using the altered movement strategies of tightrope walkers?
- Are weightings of sensory inputs altered for tightrope walkers and are these altered weightings used for tandem stance balance tasks on a normal surface?
- Are the same two types of movement strategies used in tandem stance, where roll motion is more unstable, as for roll controlling the feet side-by-side position?
- Do tightrope walkers lock their head more to the trunk and shift the head resonance, relying more on neck proprioceptive inputs and decreased vestibular inputs to control head motion?
- Are movement strategies at the neck and lumbosacral joints changed for tandem stance under different sensory conditions?
- Does standing on a tightrope require a different movement strategy compared to standing on a foam support-surface?
- Is lateral control in tandem stance predominant? If so can it provide a stance control situation to study the effect of a feedback device in one control dimension?

CONTRIBUTIONS

Chapter 2, «Coordination of the Head with Respect to the Trunk and Pelvis in the Roll and Pitch Planes during quiet Stance»

FH developed the experimental instrumentation and software, designed the study, supervised experiments, analysed the data, interpreted the data and wrote the final manuscript and created the final figures. GVS recruited and tested subjects and collected data, helped design the study, participated in analysis, and wrote a preliminary draft of the manuscript. JHJA helped design the study and the designed the instrumentation, helped analyse and interpret the data, and helped writing the final manuscript and advised on the form of the final figures.

Chapter 3, «Coordination of the Head with Respect to the Trunk, Pelvis and lower Leg during quiet Stance after vestibular Loss»

FH developed the experimental instrumentation and software, designed the study, supervised experiments, analysed the data, interpreted the data and wrote the final manuscript and created the final figures. HJW recruited and tested subjects and collected data, helped design the study, participated in analysis and wrote a preliminary draft of the manuscript. JHJA helped design the study and designed the instrumentation, helped analyse and interpret the data, helped writing the final manuscript draft and helped rework the final figures.

Chapter 4, «The effect of prosthetic feedback on the strategies and synergies used by vestibular loss subjects to control stance

FH developed the experimental instrumentation and software, designed the study, supervised experiments, analysed the data, interpreted the data and wrote the final manuscript and created the final figures. IMHA recruited and tested subjects and collected data, helped design the study, participated in analysis and wrote a very preliminary first draft. NGAE recruited and tested subjects and collected data, and contributed to the initial study design. KST tested subjects and collected data, participated in the early design. JHJA helped design the study and instrumentation, helped analyse and interpret the data, helped writing the final manuscript and helped rework the final figures.

Chapter 5, «Movement strategies in tandem stance: Differences between trained tightrope walkers and untrained subjects»

FH developed the experimental instrumentation and software, designed the study, supervised experiments, analysed the data, interpreted the data and wrote several drafts of the manuscript and created the final figures. RJMT recruited and tested subjects and collected data, helped design the study, participated in analysis. JHJA designed the study and instrumentation, helped analyse and interpret the data, helped FH in writing the final manuscript.

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Chapter 2

Coordination of the head with respect to the trunk and pelvis in the roll and pitch planes during quiet stance[1]

[1] Adapted from:

Honegger F, van Spijker GJ, Allum JH (2012) Coordination of the head with respect to the trunk and pelvis in the roll and pitch planes during quiet stance. Neuroscience 213:62-71, IF 3.38
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ABSTRACT

This study examined the relationship between head and trunk sway during quiet stance and compared this relationship with that of the pelvis to the trunk. Sixteen younger and 14 elderly subjects participated, performing four different sensory tasks: standing quietly on a firm or foam support surface, with eyes open or closed. Roll and pitch angular velocities were recorded with six body-worn gyroscopes; a set of two mounted at the upper trunk, an identical set at the hips, and another set on a head band. Angle correlation analysis was performed in three frequency bands: below 0.7 Hz (LP), above 3 Hz (HP) and in between (BP) using the integrated angle velocity signals. Angular velocities were spectrally analysed.

Greater head than trunk motion was observed in angle correlations, power spectral density (PSD) ratios, and transfer functions (TFs). Head on trunk motion could be divided for all sensory conditions into a low-frequency (<0.7 Hz) «head locked to trunk» inverted pendulum mode, a mid-frequency (ca. 3 Hz), resonant mode, and a slightly anti-phasic head motion on stabilized trunk, high-frequency (>3 Hz) mode. There was coherent motion between head and trunk but not between head and pelvis. Trunk and pelvis data was consistent with previously reported in-phase and anti-phase movements between these segments. Significant age differences were not found.

This data indicates that during quiet stance body motion increases in order of pelvis, trunk, head and quiet stance involves control of at least two separate links: trunk on pelvis and head on trunk dominated by head resonance. The head is locked to the trunk for low frequency motion possibly because motion is just supra-vestibular threshold. The head is not stabilised in space during stance, rather the pelvis is.

Key words: balance control, multi-segmental movement strategies, vestibular signals, head resonance.

INTRODUCTION

The central nervous system (CNS) employs information from the vestibular, visual and somatosensory systems to control upright stance. These signals originate in a number of different body segments. Thus if there is relative movement between body segments the interpretation of these signals by the CNS is more complex than if there is no relative motion. For example, for the conditions of perturbed stance where body motion is multi-link, the proportion of these inputs to balance control may vary (Black et al., 1983, Allum and Honegger, 1998, Peterka and Loughlin, 2004, Allum et al., 2008). Likewise for unperturbed stance, the relative motion between the upper and lower body segments may vary with an in-phase, inverted pendulum like mode, observed at low frequencies, anti-phasic motion at high frequencies above 3 Hz (Creath et al., 2005, Horlings et al., 2009). Therefore, even the control of quiet stance is more complex than just control of an inverted pendulum. To reduce the sensory complexity this bimodal mode implies it has been suggested that lower body motion is controlled by changing the weighting of sensory inputs as sway amplitudes increases and the upper body motion is predominantly influenced by intrinsic musculo-skeletal mechanisms (Goodworth and Peterka, 2012). This may not prove a sufficient reduction in the complexity for controlling quiet stance as motion of the head needs to be taken into account.

Koozekanani et al. (1983) concluded that considering ankle movements and hip joint motion is crucial for describing balance control during quiet stance. More recently, others have shown that a single segment model (the inverted pendulum model) does not adequately represent control of upright stance; it appears that ankle, hip (or lumbo-sacral) strategies exist simultaneously (Kuo, 1995, Creath et al., 2005, Hsu et al., 2007, Pinter et al., 2008, Horlings et al., 2009), depending on the sway frequency band considered, and the type of surface (firm or foam) on which the subject is standing (Creath et al., 2005, Horlings et al., 2009), the direction (roll or pitch) of sway (Creath et al., 2005, Horlings et al., 2009), and the presence of vestibular or lower leg proprioceptive inputs (Horlings et al., 2009). Missing from these studies is information about how head movements are coordinated within these upper and lower body strategies during quiet stance.

Knowledge about head movements during stance is of interest, as head movements need to be compensated with oppositely directed eyes movement to stabilize gaze for adequate fixation on the environment. Further, the vestibular system is situated in the inner ear and receives input whenever the head is moved. The sensory signal thereby produced can elicit eye movement or muscles counteraction to control posture, provided relative motion between body segments is known. Concerning this relative motion there are two well-known theories: the head is fixed in space or the head is stabilized on the trunk (Horak and Nashner, 1986). When upright stance is con trolled in the first manner, the head is held still, that is servo-referenced to the visual input and the vestibular system output is nulled out. This mode of control has also been observed for various locomotor tasks in humans (Pozzo et al., 1998). Based on the alternative theory, head stabilized on the trunk, the head moves with the body and the vestibular system provides a measure of trunk motion. Both of these modes are highly dependent on the sensory information available and thus should be altered with changes in sensory inputs, for example, by eyes closure or reducing ankle proprioceptive input effectiveness with a foam support surface, particularly if lower body movements are controlled by proprioceptive inputs (Keshner and Kenyon, 2000). If as suggested by Goodworth and Peterka (2012) upper body motions is mostly controlled by intrinsic biomechanics, and not as supposed by vestibular inputs (Keshner and Kenyon, 2000), vestibulo-spinal reflexes would then act only to stabilise the head on the trunk by damping its biomechanical resonance (Keshner et al., 1995, Peterson et al., 2001, Goldberg and Cullen, 2011).

Several studies have suggested that the head resonates at 3 Hz when the body is perturbed. For example, a sudden pitch rotation of the surface on which subjects are standing leads to 3 Hz head oscillations (Keshner et al., 1987). Whole body rotations in the yaw plane also yield 3 Hz (Keshner and Peterson, 1995). Finally directly applied head oscillations in the pitch revealed a resonance at 3 Hz (Viviani and Berthoz, 1975). Given this evidence, it would not be surprising to observe a 3 Hz resonance of the head during quiet stance. There has been some research about head movements during quiet stance. Karlberg and Magnusson (1998) investigated the effect of wearing a neck collar for patients with compensated unilateral peripheral vestibular loss, assuming that the collar would stabilize the head with respect to the trunk. Instead, the neck collar impaired balance in these patients. This finding suggests that head movements independent of shoulder movements are essential for maintaining postural stability. The question arises how much motion is necessary. For example, there is a tendency for head movement to be less than trunk movements when the support is oscillated (Vaugoyeau et al., 2008). As a simultaneous reduction in CoM movements occurred too, the head movements may act to counter the movements of masses having the greatest effect on CoM sway, such as the trunk and pelvis (Corna et al., 1999, Akram et al., 2008).

In this study we measured body movements at the level of pelvis, upper trunk and head in order to gain more insights into how the head moves with respect to the trunk and pelvis in order to maintain balance during quiet stance. We asked the following questions: What is the relationship between head and trunk movements of the body during quiet stance? Are head movements highly correlated with the trunk or the pelvis movements? Is an independent head resonance observed? Are the head movements with respect to the trunk different in the roll and pitch plane? Further, are head movements with respect to the trunk altered under different sensory conditions? These questions build on our previous study on the relationship between trunk and pelvis motion during quiet stance in healthy controls and those with vestibular or lower leg proprioceptive loss inputs (Horlings et al., 2009).

In general, it is known that elderly are less flexible and have more body sway after 65 years of age (Nardone et al., 2000, Gill et al., 2001). Furthermore, healthy older adults generate more head sway than healthy young adults while performing virtual reality tasks in quiet stance (Sundermier et al., 1996, Borger et al., 1999, Loughlin and Redfern, 2001, Sparto et al., 2006), suggesting that older adults rely more on visual cues than young adults and are therefore more unstable with greater head sway. Based on this finding it is reasonable to assume that head with respect to trunk movement strategies are different between young and elderly. Here we also explored if elderly have a different relationship in body sway of the different segments compared to the younger. We found few significant differences and therefore this report concentrated on strategies of the young.

EXPERIMENTAL PROCEDURES

Subjects

Thirty healthy subjects participated in this study: 16 younger (8 F, 8 M aged 22.6 ± 3.1 (mean ± SD)) and 14 elderly (6 F, 8 M aged 68.4 ± 4.5 (mean ± SD)). The latter were recruited at a health club for the elderly in Basel, Switzerland. Subjects had no neurological, vestibular, visual, or orthopaedic problems that could influence balance, and had a body mass index (BMI) in the range of 18-30. All subjects gave witnessed, written informed, consent to participate in the experiments according to the Declaration of Helsinki. The Institutional Ethical Review Board of the University Hospital Basel approved the study.

Procedure

The subjects were asked to stand as quiet as possible during the four stance tasks: standing on both legs on a firm or a foam (F) support surface, with eyes open (EO) and closed (EC). The order of the surface used first was randomised. The block of foam used had a height, width, and length of 10 by 44 by 204 cm, and a density of 25 kg/m^3. The subjects stood without shoes, so different shoe types could not interfere with the measurements. The feet were placed at shoulder width apart and the arms were hanging at the sides of their body. While performing the eyes open tasks, subjects were asked to fixate a point 5 m away. A spotter stood next to the subject, to assist in case balance was lost. Subjects performed each task once, for 180 s. During three trials there was a technical failure in data collection, requiring data to be cut at 97 s (one young subject for ECF), and at 120 s and 150 s (both elderly subjects for EOF).

Measurement systems

Two identical gyroscope-based measurement systems of weight 500 g (SwayStar, Balance International Innovations GmbH, Switzerland) were time synchronised together to measure pelvis and trunk movements. At the pelvis, one system was mounted on a converted motor-cycle kidney-belt fitted around the hips and the other system was mounted on a custom-built tight-fitting shoulder harness. A MTi miniature gyro-enhanced attitude and heading reference system (Xsens

Technologies, Netherlands) attached to a tight-fitting headband (total weight 150 g) was used to measure head movements. The cables connecting the systems to the PCs were swung over the shoulder of the spotters in order not to interfere with the subject's mobility. The data were sampled at a rate of 100 Hz. The SwayStar systems measured angular velocities in the roll (side-to-side) and in the pitch (fore-aft) plane with a 16-bit accuracy over a range of ±327 deg/s. The Xsens system has a slightly lower range of ±300 deg/s and its noise spectrum is about a factor 1000 higher. The noise spectra of the gyroscope systems compared to measured data of the younger subjects are depicted in Fig. 1. Because the noise of the Xsens system approached the spectral content of head signals at 10 Hz (see Fig. 1c), our analysis was limited to signals less than 10 Hz.

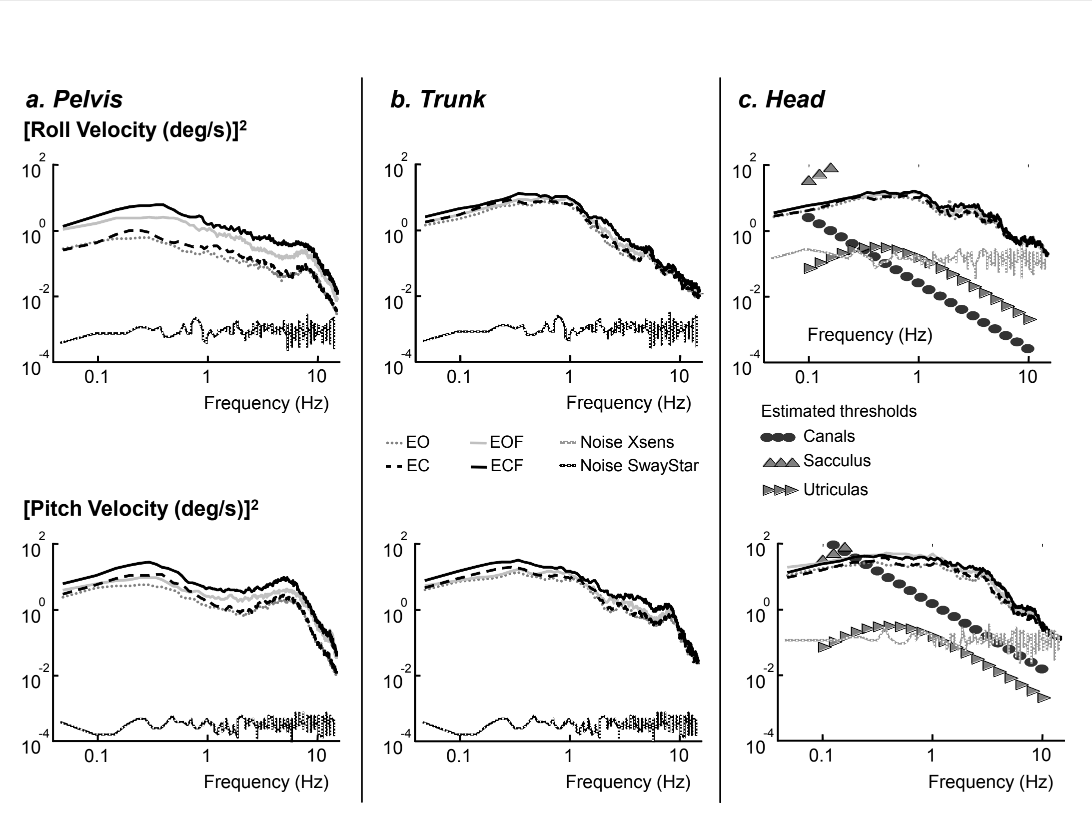

Fig. 1. Power spectral densities. Mean PSD data in each condition of the younger population and sensor noise spectra at pelvis (a), trunk (b) and head (c) level. The upper panels are for the roll plane, and the lower panels in the pitch plane. The head data in c also contains the estimated threshold of the semicircular canals (filled ovals) based on data from Seemungal et al. (2004) and the estimated threshold of the otoliths (triangles) based on the data of Kingma (2005).

Data processing

Time domain: correlations between angle displacements of different segments. Data processing of the angle data after integration of the velocity data involved similar techniques to those described by Horlings et al. (2009).

The low-frequency trend of the angle data (see Fig. 2) was obtained by filtering the data using the denoise function of the «Rice Wavelet Toolbox» (Baraniuk et al., 2002). The function implements a wavelet low-pass filter based on the work of Daubechies (1988) and Donoho (Donoho and Johnstone, 1994, Donoho, 1995). For our study populations, the de-noising process was equivalent to adaptive filtering of frequencies below 0.05 Hz. The original angle data minus the low-frequency trend was further analysed in three separate frequency bands (see Fig. 2): a low-pass (<0.7 Hz) and a high-pass (>3.0 Hz) band and the band in between (0.7–3 Hz). Supplementary to Horlings et al. (2009) we employed a band-pass filter because we observed, as described in the results section, a resonance in this band in ratios of head to shoulder power spectral densities of velocity data (see Figs. 6 and 7) and wished to quantify movement strategies at the frequency of this resonance. The filtering was accomplished using simple 3rd-order Butterworth filters running forwards and backwards through the data, which is equivalent to a 6th-order filter with zero phase shift. Fig. 2b–d shows examples of low-pass (LP), band-pass (BP), and high-pass (HP) filtered data, respectively. Then, total least-squares regression lines (Krystek and Anton, 2007) were fitted to the x–y plots of roll and pitch angles in each frequency band by calculating the main axis of the regression between the variables of interest, as if both variables have equal uncertainty in error.

We examined angle regression slopes with circular statistics(Jammalamadaka and SenGupta, 2001). The statistical calculations used the «Circular Statistic Toolbox» update 2010b published by (Berens, 2009). A short description of the tests can be found in Behrens' publication as well as references to the relevant literature. The Watson–Williams test we used acts as the circular analogue to the two-sample t-test or the one-factor ANOVA, and the Harrison–Kanji test as the circular two-factor ANOVA. Before circular calculations the axial regression slopes were mapped by $\alpha_i \rightarrow (2\cdot\alpha_i) \bmod (2\cdot\pi)$ to a unimodal sample (Fisher, 1993). Rao and Rayleigh tests were used to check the directedness of the data.

Frequency Domain: Power spectral densities (PSD) and transfer functions of velocity data. Estimates of the power spectral (PSD) $P_{x,x}(f)$, cross power densities $P_{x,y}(f)$, transfer functions $T_{x,y}(f)=\dfrac{P_{y,x}(f)}{P_{x,x}(f)}$, coherence functions $C_{x,y}(f)=\dfrac{|P_{x,y}(f)|^2}{P_{x,x}(f)\cdot P_{y,y}(f)}$ and PSD ratios (trunk/pelvis, head/pelvis, head/trunk) $R_{x,y}(f)=\dfrac{P_{x,x}(f)}{P_{y,y}(f)}$ of the angular velocity data were calculated to determine relative segmental motion. For this purpose MATLABs Signal Processing Toolbox Version 6.13 (R2010A) was used. The estimates were based on Welch's averaging method (Welch, 1967), employing a window size of 2048 samples with an overlap of 1024 samples. Group averaged data was smoothed with a Tukey's running median smoothing (Tukey, 1977) known as «(3RSR)2H twice» for display purpose only (see Figs. 1, 3, 6, and 7). Repeated measures two-way ANOVAs (condition x group) with post hoc Bonferroni corrections were performed to analyse differences in PSD, PSD ratios and transfer functions, across 10 frequency bins logarithmically divided across the range 0.15 to 10.03. These frequency bins were: 0.15 Hz (0-0.2930 Hz; mean of 7 samples), 0.2 Hz (0-0.3906 Hz, mean of 9 samples), 0.42 Hz (0.1953-0.6348 Hz, this and all other bins mean of 10 samples), 0.71 Hz (0.4883-0.9277 Hz), 1.39 Hz (1.1719-1.6113 Hz), 2.32 Hz (2.0996-2.5391 Hz), 3.69 Hz (3.4668-3.9063 Hz), 6.13 Hz (5.9082-6.3477 Hz) 7.59 Hz (7.3730-7.8125 Hz), 10.03 Hz (9.8145-10.2539 Hz). We chose 10 bins in order to determine the frequencies at which resonances occurred. To explore whether the PSD ratios and transfer function differed from a value of 1, one sample t tests were performed across the frequency bins.

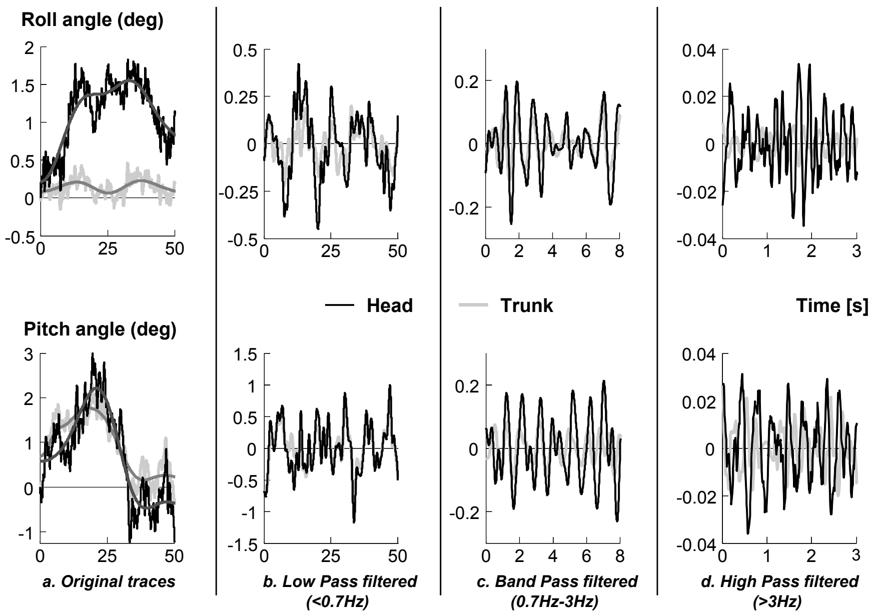

Fig. 2. Original traces of head and trunk movements (Y, EC). (a and b) Fifty-second recording of a younger subject, standing with eyes closed on firm support surface; the upper panels are in the roll plane, and the lower panels in the pitch plane. (a) Original sway traces of head (black) and trunk (grey) with general trend lines. (b) Same trace after removing the general trend and low-pass filtering. (c) Eight-second of the same trace as in (a) after removing the general trend and band-pass filtering. (d) Three-second of the same trace as in (a) after removing the trend high-pass filtering.

RESULTS

Elderly versus Younger

In general there are no differences between the two age groups. For example, Fig. 3 shows PSD data of both groups for the eyes closed firm floor condition. Statistically, as for other conditions, some frequency bins differed from one another for the age-groups. These bins were isolated from one another, and therefore considered of less interest because across neighbouring bins consistent differences were not observed. Given the lack of statistically significant differences across neighbouring bins in PSD data, in the following description, we considered only the results of the slightly larger group of younger subjects.

Head movements with respect to the trunk

Fig. 2a–c depict the filtered angle data of a typical young subject, standing eyes closed on firm support (EC). As seen in this figure and in the population mean PSD plots of Figs. 1 and 3, the amplitudes of head sway were larger than those of the shoulders, i.e. are larger than movements of the upper trunk. Based on estimates of vestibular thresholds across frequencies (see Fig. 1c), head movements were only sub-threshold for semicircular canal systems at low frequencies of pitch. Head

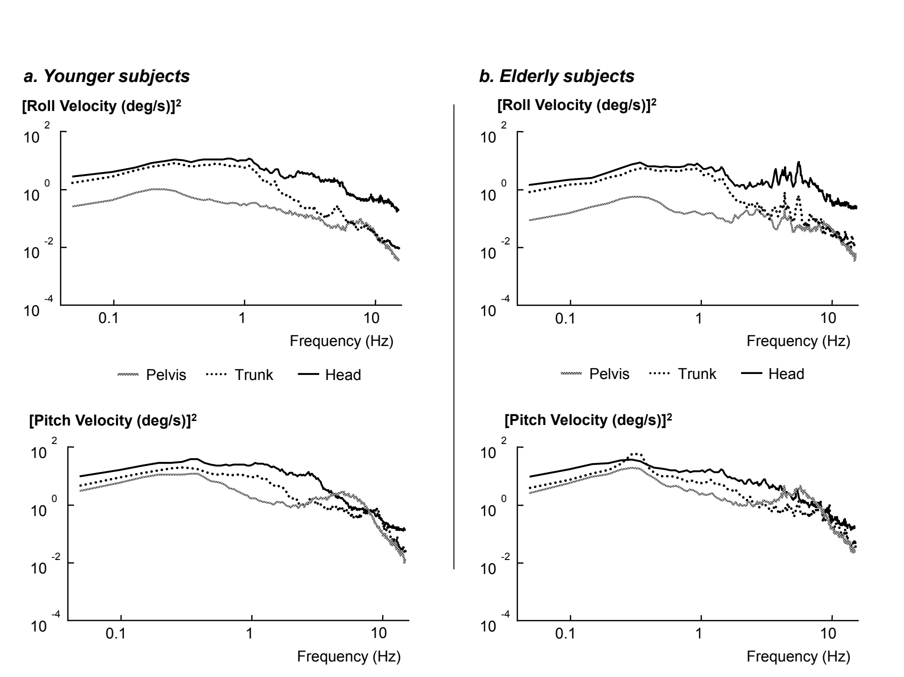

Fig. 3. Power spectral densities: younger vs elderly (EC). This figure shows the mean PSD data at all three measurement levels (pelvis, trunk and head) while standing with eyes closed on firm support surface, of both groups: younger (left) and elderly (right). The upper panels are in the roll plane, and the lower panels in the pitch plane.

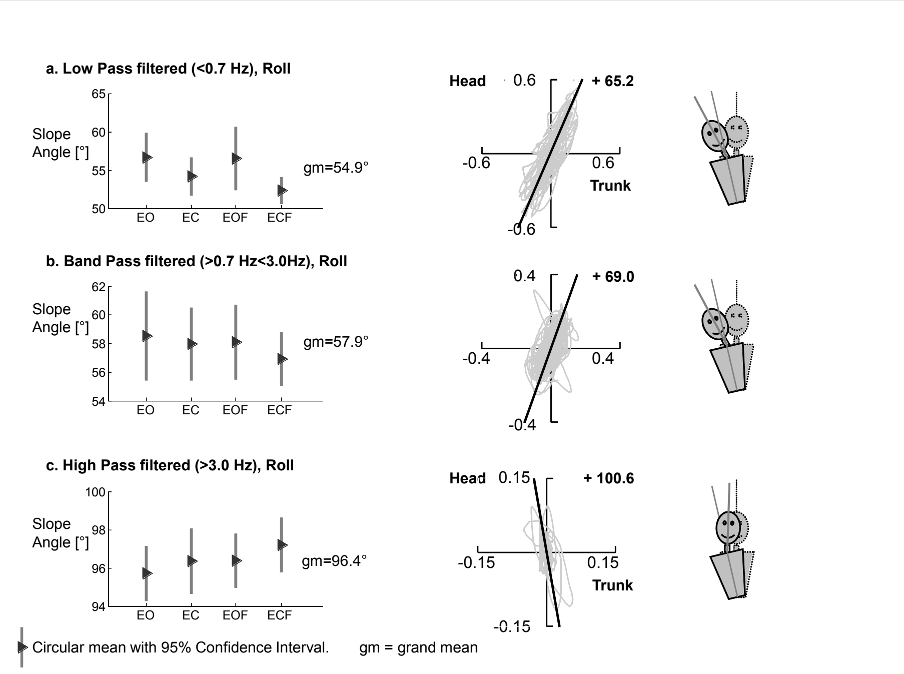

Fig. 4. Regression slopes head with respect to the trunk for roll. This figure shows mean regression slopes for all four standing conditions: low-pass filtered (a), band-pass filter in (b) and high-pass filtered in (c). In the bar plots, the circular means and 95% confidence intervals are depicted. In the middle examples of original regression traces of head with respect to the trunk and corresponding slope lines are shown (same trial as in Fig. 2, eyes closed on a firm surface). The schemas on the right represent the movement of head and/or trunk based on each slope angle.

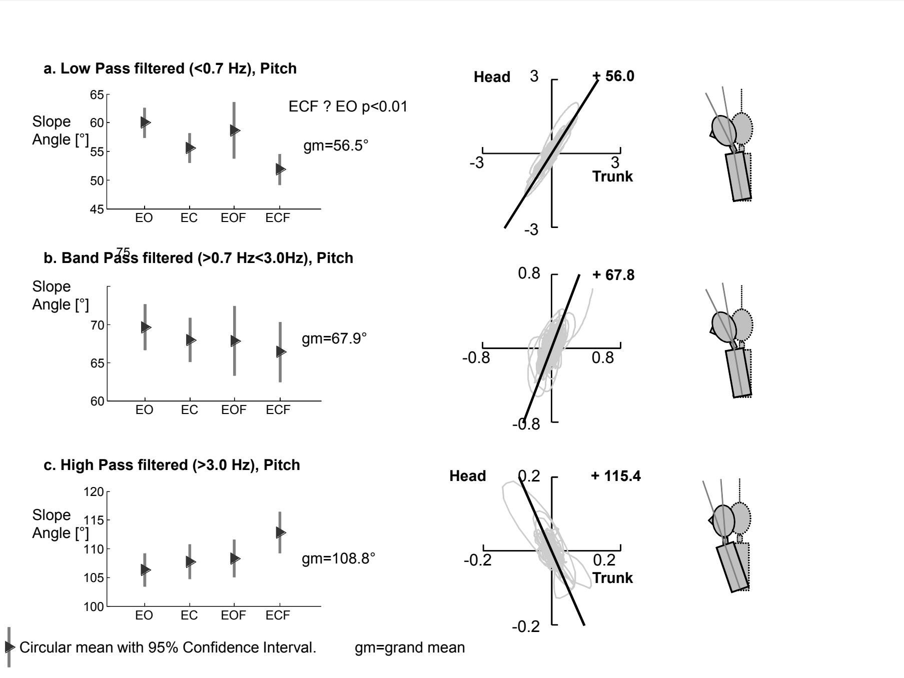

Fig. 5. Regression slopes: Head with respect to the trunk for pitch. This figure shows mean regression slopes for all four standing conditions: low-pass filtered (a), band-pass filtered in (b) and high-pass filtered (c). In the bar plots, the circular means and 95% confidence intervals of the mean are depicted. In the middle examples of original regression traces of head with respect to the trunk and corresponding slope lines are shown (same trial as in Fig. 2, eyes closed on firm surface). The schemas on the right represent the movement of the head based on the slope angle.

movements were always larger than threshold estimates for the utricular system but lower than those for the saccular system.

We confirmed the larger sway of the head with respect to the trunk using the mean values of the least square regression slopes of the roll and pitch angle data subdivided into the LP, BP and HP bands as shown in Fig. 2. Regression slope values for all stance conditions are presented in Fig. 4 for roll and Fig. 5 for pitch. In general the regression angles indicated that the LP pitch and roll angles are nearly in phase (perfect in-phase would have a regression angle of 45°),BP roll and pitch angles are in phase but less so than for the LP data and HP head motion is anti-phase, almost independent of the trunk for roll. These findings and the fact that all regression angles were not in the interval between 0° and 45°, -45°(135°) and 180° is consistent with larger amplitude head movements than shoulder movements. In addition, the LP pitch data showed a main condition effect ($p < 0.01$). A post-hoc test showed only a significant difference ($p < 0.01$) between the EO and eyes closed on foam (ECF) conditions, with only ECF approaching at 52° a perfect in-phase condition of 45°.

Figs. 6a and 7a show the PSD ratios of headwith respect to the trunk. The ratios were in nearly all frequency bins different from unity ($p < 0.05$, only significant data for ECF is marked in the figures). Low frequency roll and pitch ratios were slightly greater than unity, i.e. head motion is slightly greater than trunk motion. This is in agreement with the regression slopes data shown in Figs. 4a and 5a. At approximately 3 Hz a peak was observed in both the roll and pitch data (Figs. 6a and 7a). After 6 Hz the PSD pitch ratios were close tounity again,whereas for roll the ratios increased. For comparison the ECF pelvis to trunk data is shown in Figs. 6 and 7. These ratios replicate those of our earlier study (Horlings et al., 2009). For frequencies above 2.3 Hz, trunk with respect to the pelvis ratios were significantly lower than those of the head to the trunk.

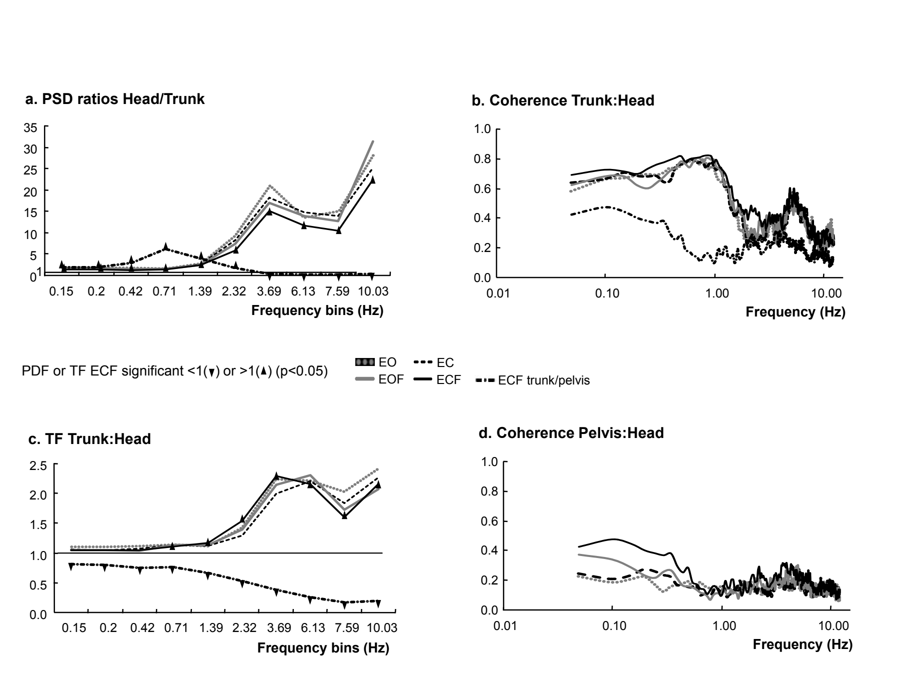

Fig. 6. PSD ratios, coherence and transfer function of head with respect to the trunk in the roll plane. Depicts for all four standing conditions the head to trunk ratios (a) and transfer functions (c) across the 10 frequency bins; and coherence of head to trunk (b), and head to pelvis (d) across the whole frequency spectrum, in the roll plane. Each symbol (N or .) on the ECF head/trunk and ECF trunk/pelvis lines signifies that the PSD ratio or transfer function significantly differs from 1 (binominal test) at that frequency bin (N signifies >1 and . <1).

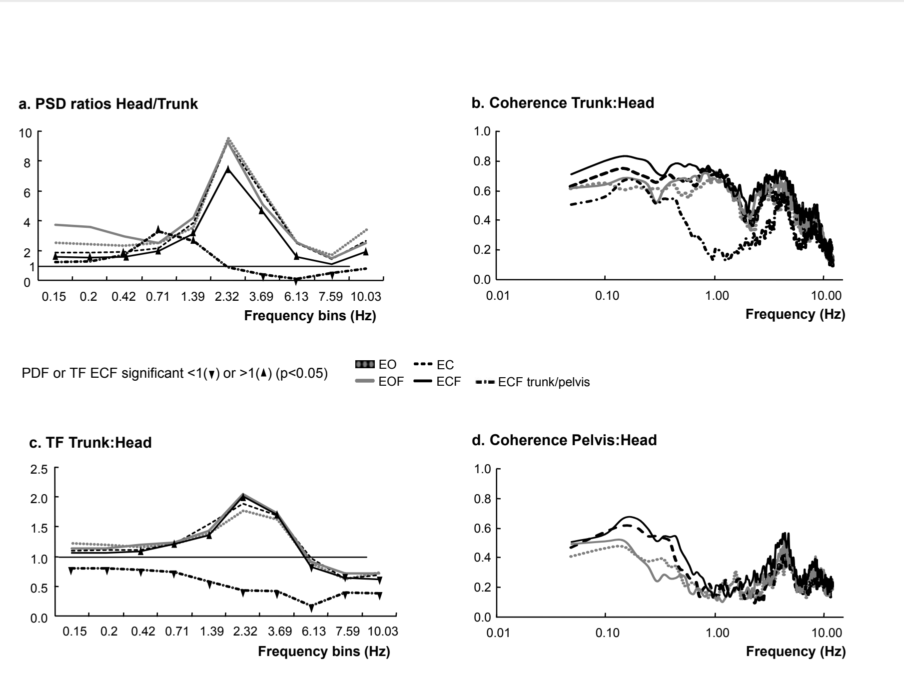

Fig. 7. PSD ratios, coherence and transfer function of head with respect to the trunk in the pitch plane. Depicts for all four standing conditions the head to trunk ratios (a) and transfer functions (c) across the ten frequency bins; and coherence of head to trunk (b), and head to pelvis (d) across the whole frequency spectrum, in the pitch plane. Each symbol (N or .) on the ECF head/trunk and ECF trunk/pelvis lines signifies that the PSD ratio or transfer function significantly differs from one (binominal test) at that frequency bin (N signifies >1 and . <1).

Coherence between trunk and head are shown in Figs. 6b and 7b; in the roll and pitch planes, respectively. The coherence was rather large up to 1 Hz, then fell off and increased after 3 Hz. That is, the coherence was low-est when PSD ratios were highest. Note that the coherence for the pelvis and trunk is lower and fell off around 0.3 Hz and that the coherence between the pelvis and head was even lower (Figs. 6d and 7b). The corresponding transfer functions (Figs. 6c and 7c) followed the trends of the PSD ratios. TF gains between trunk and head were near unity in the low frequencies (below 0.7 Hz), contrasting with those between pelvis and trunk which were less than unity (see Figs. 6c and 7c, and also (Horlings et al., 2009)). Both the pitch and roll TFs had a peak at 3 Hz but at higher frequencies the TFs differed for the roll and pitch planes. Roll TF gain remained approximately constant at a value of two whereas pitch gain decreased below unity. TFs between pelvis and shoulder were, however even lower (see also Horlings et al. (2009)).

In the PSDs for the shoulder there was a clear ordering effect based on task difficulty with the most difficult task ECF showing the largest movement (Fig. 1b). A similar effect with a broader spread was seen for the pelvis (Fig. 1a). At the head, a similar separation was not observed (Fig. 1c). Consequently, PDS ratios of head to shoulder were larger for the easier sensory conditions (EO) in both roll and pitch compared to the ECF conditions (Figs. 6a and 7a, $p < 0.05$). However, a similar separation across sensory conditions was not observed for the TFs.

DISCUSSION

The results of this study add to those of previous studies indicating that the body does not move just as an inverted pendulum during quiet stance. Rather, a dual mode of motion is present in both the pitch (Creath et al., 2005) and roll planes (Horlings et al., 2009) with motion of the upper trunk (at the level of the shoulders) being anti-phasic to that of the pelvis for frequencies above 3 Hz. There is, however, also a tendency for the body to move as an inverted pendulum for low frequency movements below 0.7 Hz (Creath et al., 2005, Horlings et al., 2009). The tendency is greater for pitch and greater for the most difficult sensory condition on a foam support surface with eyes closed. In contrast, in the roll direction on a firm surface the upper trunk moves most and the pelvis is stabilized. Under these circumstances it is important to know how the head moves. Our current results indicate that under most two legged stance conditions, sway at the head in both the roll and pitch is greater than at the upper trunk and the pelvis, thus the head is unlikely to move as the rest of the body as an inverted pendulum in concert with trunk movements even when trunk movements are equal to those of the pelvis. When, however, the body motion is divided into low, mid and high frequency components, we noted a tendency for the head and trunk to move as one unit during low- and mid-frequency sway. The regression slopes of head to trunk angles over these frequency ranges were always greater than 45°, indicating that head movements always leaned more than expected from a pure inverted pendulum mode. A similar result was noted for the trunk and pelvis (Horlings et al., 2009).

The lack of differences between the elderly and young subjects is not surprising. The difference in sway velocities between young and elderly subjects of 70 years of age is small, even for the most difficult stance test we employed (Hegeman et al., 2007). Secondly, the elderly were all without a falling tendency, visiting the fitness centre where we recruited them at least once a week. This result contrasts with results obtained when the support surface is perturbed. Elderly subjects have a different movement strategy here to that of the young due, presumably, to their joint stiffness and the greater sway angles induced by support surface perturbations (Allum et al., 2002) than those during quiet stance.

For low-frequency roll movements both PSD ratios and transfer function values between trunk and head were close to unity. Thus it can be argued that the head is approximately locked to the trunk. The support surface generally had no effect on this «strapped down» mode (Horak and Nashner, 1986). However, under eyes closed conditions on a foam support surface, the head movement is more locked to the trunk. For low-frequency pitch and roll movements the PSD ratios were greater than unity even though coherent transfer functions were not different from unity. This result implies that the CNS was attempting to control the head to move as one unit with the trunk in the presence of independent head movements.

This movement mode of head fixed to trunk is a logical solution to the control of human stance. As indicated in Fig. 1c these movements are equal to or less than the thresholds of vestibular thresholds for pitch rotations sensed by the canal afferents. Under the circumstances sensing head movements would need to depend on otolith systems. Head movements are supra-threshold for utricular but not saccular otolith signals as indicated in Fig. 1c. However, without canal information the CNS cannot distinguish between small inclinations and linear accelerations of the body. Thus locking the head to the trunk would be one solution removing the need to rely on vestibular semi-circular canal signals to control upright balance.

We observed resonances between 2 and 4 Hz which appeared from the corresponding transfer functions for pitch to be damped out at high frequencies as suggested by Keshner (2000). In contrast there was no damping out in the roll plane, rather both PSD ratio and transfer function gains remained high. Whether this implies that these damping processes, controlled by vestibulo-cervical reflexes (Goldberg and Cullen, 2011), are only possible in the pitch plane or the biomechanics are fundamentally different in these two planes remains to be investigated.

The question arises as the appropriate mode of control between the trunk and head for frequencies above 3 Hz. This question is complicated by the presence of resonance peaks of head movements in both the pitch and roll planes. Indeed it has been recently argued that one of the roles of vestibular spinal and cervical colic reflexes is to dampen out oscillations associated with natural resonant frequencies of the head (Goldberg and Cullen, 2011). Although their review was based on animal studies, similar resonant frequencies and damping thereof have been observed in human subjects (Viviani and Berthoz, 1975, Keshner et al., 1987, Keshner, 2000). Damping could presumably be highly necessary because rapid anti-phasic motion of the body both in pitch and roll (Horlings et al., 2009) would excite such a resonance.

The current study showed no changes in the resonant frequencies of pitch and roll when visual inputs were removed and only a small change when ankles were reduced in efficacy when subjects stood on a foam support surface. Thus future studies should investigate sensory participation in these resonances by examining changes with vestibular loss.

In our previous studies (Horlings et al., 2009) we separated motion into low frequency (<0.7 Hz) and high frequency (>3 Hz) components based on a dip in pelvis PSDs values between these cut-off frequencies. However, head motion shows a resonance with respect to the trunk between these frequencies. Therefore it was of considerable interest to examine regressions between trunk and head data in the range 0.7–3 Hz and show that these must pass between in-phase to anti-phase as expected at a resonant frequency. It would be of further interest to examine the mechanisms underlying this resonance, for example by testing patients with vestibular loss. This because it is not clear that the resonance is completely driven by the trunk movements as the coherence is lowest when the TF values are highest.

Although the movement of the pelvis is clearly greater under more difficult sensory conditions (see Fig. 1), that of the trunk is relatively less so, and head movement, though greater than that of pelvis, still less so. In the context of the work of Goodworth and Peterka (2010), this would imply a non-linear gain control dependent on sensory reweighting. Presumably, such changes would be brought about by changes in cervico-colic reflex gains. In contrast to work of Goodworth and Peterka (2010) we did not however observe large changes in TF gains under different sensory conditions. Thus we must assume that, consistent with the high PSD ratios compared with transfer functions (see Fig. 6 and 7), considerable head motion occurs independent of trunk motion.

In the introduction two possible strategies were described for motion during quiet stance: (1) the head stabilised in space, and (2) the head «locked» to the trunk. We could demonstrate that the latter mode of control was found for low- and mid-frequency pitch and roll motion, however we found no evidence that the head is stabilised in space. Rather our evidence and that of Horlings et al. (2009) is that the pelvis is more stabilised in space. We demonstrated that control of head motion with respect to the trunk during quiet stance uses two different strategies simultaneously, just as is the case for trunk and pelvis motion (Creath et al., 2005, Horlings et al., 2009) an inverted pendulum mode for low frequencies and an anti-phasic mode for high frequencies. We had expected that the anti-phasic motion between pelvis and trunk, as well as between head and trunk could result in head and pelvis being in phase. Regression on angles and low coherence indicated that this was not the case. Our current data and that of our previous studies (Horlings et al., 2009) indicate that strategies are generally bimodal between the head and trunk, as between the pelvis and trunk. Motion of the head is dominated by resonances at ca. 3 Hz. The question arises whether the strong biomechanically coupling between head and trunk with considerable head movement due to resonant frequencies of the head on the trunk changes with peripheral vestibular loss.

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GLOSSARY

Roll: angle rotation in the lateral plane
Pitch: angle rotation in the sagittal plane

ABBREVIATIONS

BP, band pass; CoM, centre of mass; CNS, central nervous system; EC, eyes closed; ECF, eyes closed on foam; EO, eyes open; EOF, eyes open on foam; gm, grand mean; HP, high pass; LP, low pass; PSD, power spectral densities; TF, transfer function.

Chapter 3

Coordination of the head with respect to the trunk, pelvis and lower leg during quiet stance after vestibular loss[1]

[1] Adapted from:
Honegger F, Hubertus JW, Allum JH (2013) Coordination of the head with respect to the trunk, pelvis, and lower leg during quiet stance after vestibular loss. Neuroscience 232:204-215, IF 3.38
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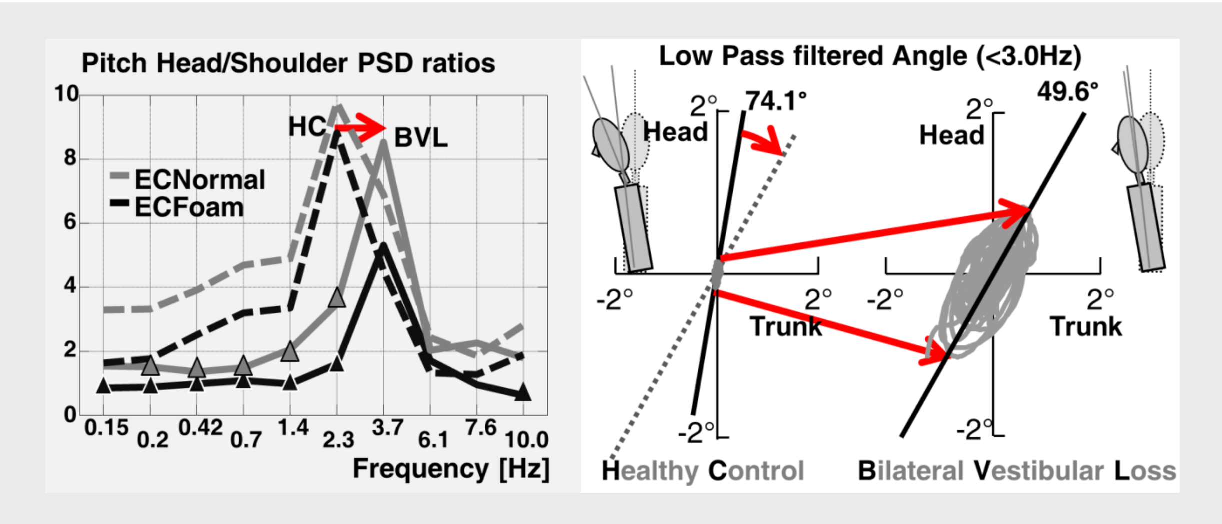
ABSTRACT

This study examined the relationship between head and trunk sway and between pelvis and leg sway during quiet stance in subjects with long-standing bilateral peripheral vestibular loss (BVLs) comparing these relationships to those of age-matched healthy controls (HCs). All subjects performed three different stance tasks: standing quietly on a firm or foam support surface, with eyes closed (ECF or eyes closed on normal) and on foam with eyes open. Data were recorded with four pairs of body-worn gyroscopes to measure roll and pitch angular velocities at the head, upper trunk, pelvis and lower-leg. These velocities were spectrally analysed and integrated for angle correlation analysis in three frequency bands: below 0.7 Hz (low pass, LP), above 3 Hz (high pass, HP) and in between (band pass, BP). For both groups head motion was greater than trunk and pelvis motion except for BVL subjects (BVLs) under ECF conditions. BVLs had greater motion than HCs at all measurement locations for ECF conditions. Angle correlation analysis indicated that the head was almost «locked» to the trunk for BVLs over the LP and BP frequency bands. Head movements for both groups were relatively independent of the trunk in the HP band. Power spectral density ratios, and transfer functions showed a similar result – head relative to trunk movements were less up to 3 Hz in all tests for BVLs. The resonant frequency of head-on-trunk motion was shifted to a higher frequency for BVLs: from 3.2 to 4.3 Hz in pitch, 4.6 to 5.4 Hz in roll. Both groups show greater lower-leg than pelvis motion. These data indicate that during quiet stance BVLs change the characteristics of their head on shoulder motion, reducing relative motion of the head below 3 Hz and increasing head resonant frequency. Presumably these changes are accomplished with increased use of proprioceptive neck reflexes.

GRAPHICAL ABSTRACT



Key words: Key words: balance control, multi-segmental movement strategies, vestibular signals, head resonance, bilateral vestibular loss (BVL).

INTRODUCTION

To maintain the head stable on the trunk during upright stance, the CNS can predominantly employ three kinds of sensory inputs: vestibular, visual and neck proprioceptive. Depending on stance conditions, the contribution of these inputs to balance control can vary (Black et al., 1983, Allum and Honegger, 1998, Peterka and Loughlin, 2004, Allum et al., 2008). For example, with eyes closed (EC), the relative contribution of proprioceptive and vestibular inputs will be higher.

The relative contributions of sensory inputs will be dependent on how the body moves during a postural task. Until recently, it was generally accepted that during quiet stance, body sway was like that of an inverted pendulum (Fitzpatrick et al., 1992, Pinter et al., 2008) with motion only about the ankle joint. In this single segment model, ankle proprioception is assumed to be the main contributor to balance control. The inverted pendulum model is, however, an oversimplification. By demonstrating that not only ankle movement, but also hip joint motion is important for balance control during quiet stance (Pinter et al., 2008), evidence was accumulated that the single segment model could not adequately explain balance control during quiet upright stance. It was observed that both ankle and hip strategies are present simultaneously during quiet stance in separate sway frequency bands with an ankle strategy dominant for low frequencies (Kuo, 1995, Creath et al., 2005, Hsu et al., 2007, Pinter et al., 2008). This observation suggests that at least two types of proprioceptive inputs (ankle and hip) contribute to balance control. Further, because two frequency ranges of sway are involved, the contribution of vestibular inputs will vary too as otolith and canal inputs cover different frequency ranges (Honegger et al., 2012).

Ankle, hip, and trunk movement strategies have been explored for quiet stance in healthy subjects, but there is a lack of information about how head movements are implemented into these strategies. Recently, head movement strategies on the trunk during quiet stance have been explored in healthy subjects (Honegger et al., 2012). For lower frequencies it was found that the head moves faster but in phase with the trunk, but for frequencies above 2 Hz resonances were described (Honegger et al., 2012). There have been, however, to our knowledge, few studies on the effect of sensory loss on head movements during quiet stance, and especially not for those subjects with permanent vestibular loss.

Several studies have examined the role of the vestibular and proprioceptive systems in controlling head movements during perturbations to quiet stance using rapid support surface movements (Horak et al., 1994, Shupert and Horak, 1996, Allum et al., 1997, Allum and Honegger, 1998). It is generally agreed that proprioceptive signals trigger balance corrections and vestibular as well as proprioceptive inputs modulate the resulting muscle response amplitudes (Horak et al., 1994, Bloem et al., 2000, Bloem et al., 2002). With vestibular loss poor control of trunk motion ensues. Interestingly, the motion of the head on the trunk is marginally affected (Allum et al., 1988, Shupert and Horak, 1996), possibly because neck reflexes compensate for the vestibular loss, especially for those with a long standing loss. Similarly, during walking, head movements appear not to be different in amplitude from those of healthy controls (HCs), even if there appears to be a lack of compensation for vertical head movements with each step (Pozzo et al., 1991, Mamoto et al., 2002). Thus the question arises whether vestibular loss subjects can compensate as well for their loss for the smaller movements of unperturbed stance.

The effect of vestibular loss on head movements during stance is of interest, as the vestibular sensory system is situated in the inner ear and normally provides vestibular input to the CNS whenever the head moves (Allum and Honegger, 1998, Runge et al., 1998). This sensory signal elicits eye and head movements maintaining gaze stable in space, the latter leading to a relative head motion with respect to the trunk. Two general theories exist which describe the head on trunk motion: either the head is fixed in space, or the head is stabilised on the trunk (Horak and Nashner, 1986). An additional concept we have proposed is that the head is approximately stabilised on the trunk for low frequency movements, but that, above 2 Hz, head stabilisation is not possible due to a biomechanical resonance (Honegger et al., 2012).

When the head is held still in space and referenced to vestibular and visual inputs, proprioceptive information about trunk motion is not needed to control head position. However, when the head is fixed on trunk, the head moves with the trunk and in this situation either the vestibular system provides a measure of trunk motion or trunk proprioceptive inputs a measure of head motion. Presumably, neck proprioception can be used to regulate locking the head to the trunk. In our previous study (Honegger et al., 2012) we concluded that for low frequencies (below 0.7 Hz) the CNS uses a mode of control in which the head moves in-phase but faster than the trunk, and is only approximately locked to the trunk. We theorised that this mode was preferred during quiet stance because head motion is sub-threshold for the vestibular system in this low frequency range (Honegger et al., 2012). Moving the head slightly faster than the trunk would ensure that neck proprioceptive inputs were continuously active. A recently published concept fitting this hypothesis is that for supra-threshold movements vestibulo-cervical and cervical–spinal information acts to reduce head on trunk resonances (Goldberg and Cullen, 2011). In our previous study (Honegger et al., 2012) we noted the presence of a head resonance in healthy subjects. Motion of the head dominated over that of the trunk at resonances which lie between 2 and 4 Hz. It was concluded that in addition to weak locking of the head on the trunk, a biomechanical resonance was reduced using vestibulo-cervical and cervical–spinal pathways.

The question arises whether head motion becomes unstable on the trunk and more resonant when vestibulo-cervical reflexes (VCRs) are deficient due to vestibular loss. There are different viewpoints which can be applied to this question. One report suggested that patients with vestibular loss have impaired stabilization of the head on the trunk (Bronstein, 1988). On the other hand, locking the head to the trunk in HCs and patients with compensated unilateral vestibular loss using a neck collar leads to greater body sway (Karlberg and Magnusson, 1998), suggesting that head movements greater than trunk movements may aid balance control as noted above. In the case of vestibular loss patients greater head movements would presumably result in greater use of neck proprioceptive signals. However, head movements of vestibular loss subjects could also be more unstable and resonant if vestibular colic reflexes provide increased muscle stiffness and viscosity (Simoneau et al., 2008). Absence of this feedback would lead to decreased stiffness and viscosity and therefore greater head excursions. These theoretical considerations need empirical support which we have attempted to provide here.

How leg motion relative to the pelvis contributes to balance control during stance is less well known than head on trunk motion. It had been shown that the control of pelvis movement with respect to the trunk is much more affected by somatosensory information but not by vestibular information (Creath et al., 2008). Thus, it could be expected that there is little change in leg relative to pelvis movements following vestibular loss, but this is not known. In HCs, Akram et al. (2008) studied the coordination between hip and ankle joints in the pitch plane while subjects stood on a sinusoidally moving platform and showed that the lower body moves in opposite direction (anti-phase) to the pelvis, presumably to minimise acceleration of the centre of mass (Allum and

Honegger, 1998, Aramaki et al., 2001). The question also arises as to how leg motion is controlled with respect to the pelvis during roll. Roll motion has been shown to be highly dependent on stance width of stance, with equal pelvis and leg motion when the feet are placed shoulder width apart (Goodworth and Peterka, 2010).

With these considerations in mind, we investigated the control of head to trunk and lower leg to pelvis motion in the pitch and roll planes during quiet stance in subjects with bilateral vestibular loss (BVL) and compared their balance control with that of HCs. To modify visual information during stance we asked subjects to close their eyes. To modify somatosensory information we asked them to stand on a foam support surface which reduces the efficacy of ankle proprioception. By comparing stance control in BVL patients to healthy subjects, we hoped to gain insights into the role of vestibular inputs on balance control.

We asked the following research questions by comparing body sway of BVL subjects with that of agematched HCs: How is the relationship between head and trunk movements in the pitch and roll planes altered during quiet stance under different sensory conditions following vestibular loss; and, how is the relationship between pelvis and leg movement of the body altered during quiet stance following vestibular loss? Our emphasis was on the first question.

Based on our earlier work on HCs (Honegger et al., 2012) we hypothesised that the amplitude of the head resonance at 2–4 Hz would be the main cause of instability in BVL subjects. We therefore expected to record higher amplitude head resonances with respect to the trunk in BVL subjects than in normal subjects.
Regarding hip to leg movement, we expected small differences between BVL subjects and controls assuming that such motion would be primarily under the influence of proprioceptive control (Horlings et al., 2009).

Answers to the questions above would presumably lead to a better understanding of the multi-segmental balance control strategies in patients with BVL. This information would provide new insights into the sensory contributions of vestibular signals to body movement coordination during quiet stance and in addition facilitate development of prosthetic interventions to improve balance, in patients with balance problems of a wide range of origin.

EXPERIMENTAL PROCEDURES

Participants

Seven (6 M, 1 F, mean age 49.3 ± 3.0 years), BVL patients were recruited from the ORL outpatient clinic of the University Hospital in Basel, Switzerland. Seven healthy subjects served as controls (3 M, 4 F, mean age 49.0 ± 4.3 years). Exclusion criteria for both groups were visual, orthopaedic, or neurological conditions other than BVL that could influence their balance, and a BMI outside the range of 18–30 kg/m^2. All BVL subjects had vestibular ocular reflex (VOR) responses to head horizontal (yaw axis) and vertical (pitch axis) whole body rotations of 80°/s^2 that were lower than the 25% limit of normal responses. Further, caloric irrigation of both ears gave no response. The BVL was in all cases idiopathic but central deficits as cause were excluded with magnetic resonance imaging of the brain. The deficits occurred at least 5 years prior to testing. We therefore assumed that any possible compensation for the bilateral loss had occurred prior to testing. All subjects gave witnessed, written informed consent to participate in the experiments, according to the Declaration of Helsinki. The Institutional Ethics Review Board of the University Hospital Basel approved the study.

Procedure

Subjects were asked to stand as quiet as possible, without shoes, during three stance tasks: standing on both legs with eyes open (EO) on foam (F) and EC on a normal (N) and on a foam support surface for 70 s. A task was repeated a maximum of three times, if balance control was lost prior to 70 s. The longest trial was selected for analysis provided it was longer than 10 s. Otherwise data could not be included because of error requirements of our analysis procedures, described below. This selection was necessary (see Table 1) for the eyes closed on foam (ECF) task. The sequence of the tasks was randomised for support surface to restrict the influence of a possible training effect. The dimensions of the foam support surface used were 10 × 44 × 204 cm. It had a density of 25 kg/m^3. Subjects were asked to stand naturally with their feet separated shoulder width apart, and with their arms hanging alongside their body. They stood without shoes to eliminate the influence of different shoe types on stance. For the EO task, subjects had to fixate their gaze at a point on the wall approximately 5 m away. During all tasks, a spotter stood next to the subjects to prevent a possible fall in case balance was lost.

Measurement systems

Motion of the trunk and pelvis was measured using two identical gyroscope based measurement systems (SwayStar, Balance International Innovations GmbH, Switzerland), with wireless transmission to time-synchronised PCs. One SwayStar unit was mounted on a belt strapped around the hips, the other was mounted between the scapulae on a tightly fitting shoulder harness. Motion of the head and leg was measured using two MTi gyro-enhanced Attitude and Heading Reference Systems (Xsens Techonologies, Enschede, The Netherlands), one attached to a tight-fitting headband and the other to a tight belt fitted around the upper part of the leg shank just below the left knee. The two computers used for recording were synchronized to one another using a trigger signal indicating the start of recordings emitted by each SwayStar system and by one computer. The light cables transmitting the SwayStar recordings start signals and the Xsens data signals to the computers were looped over a converted infusion stand, so as not to impend the subject's mobility. The noise levels of the SwayStar system and those of the MTI system are shown in Fig. 1 in the trunk, respectively, head power spectral densities (PSDs).

Table 1. Patient parameters for standing eyes closed on foam

Patient #	Age[years]	Gender	Duration[s]	Roll[deg]	Pitch[deg]
57[b]	54.4	M	9.6	*2.27*	*4.69*
64	47.5	M	31	*2.61*	*3.32*
66	45.6	F	65	*2.65*	*2.02*
61	50.1	M	70	*1.51*	*2.39*
62	49.3	M	70	*1.42*	*2.05*
63	48.8	M	70	*1.39*	*2.88*
59	52.2	M	70	0.53	*1.44*
Controls	Mean 49.2, range [41,53.2]		Mean 70	<0.88[a]	<1.18[a]

[a] Upper limit 95% confidence intervals for 90% range.

[b] Data not included as record too short.

DATA PROCESSING

Time analysis: angle correlations between different body segments.

Data angle processing comprised three steps: integration of measured velocity data to yield angle data, and then detrending and filtering the angle data. The low frequency trend was removed using the denoise function of the Rice Wavelet Toolbox (Baraniuk et al., 2002) by applying 1-denoise function as an adaptive low-pass (LP) filter affecting frequencies below 0.05 Hz.

This detrended position data were then partitioned into low, mid, and high frequency components. As in our previous study (Honegger et al., 2012) we separated sway based on the shape of the average PSD of the angular velocity for the pelvis (Horlings et al., 2009). That is, the original angle data minus the low-frequency trend, were separated into three frequency bands: a low frequency band (<0.7 Hz), a high frequency band (>3 Hz) and the band in between (0.7–3.0 Hz) – see Figs. 2 and 3. The low and high frequency were set at frequencies just higher and just lower, respectively, of two resonances in pelvis data (Horlings et al., 2009). Similar resonances can be observed in the trunk pitch data of Fig. 1. The filtering was accomplished with 3rd order Butterworth filters running forwards and backwards through the data to achieve zero phase shift. For each frequency band we performed regressions on the angle position data between the head and trunk, and pelvis and leg, using total least-squares regression procedures (Krystek and Anton, 2007), that is, we fitted total regression lines to the x–y plots of roll (x) and pitch (y) angle data in each frequency band. The regressions were defined in terms of a regression slope angle.

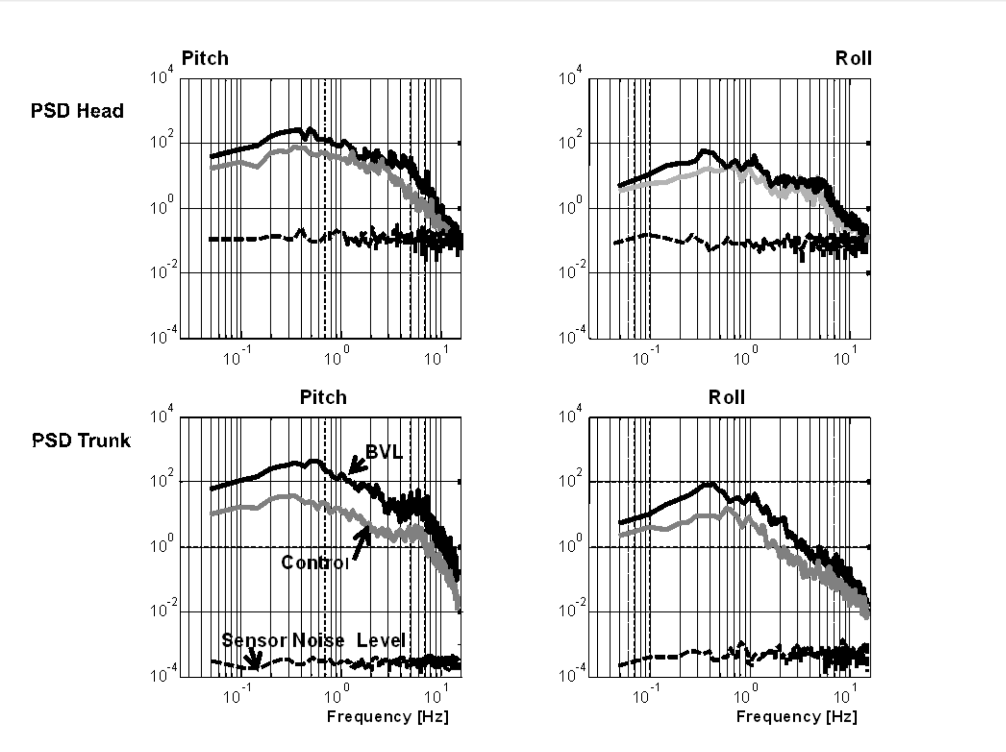


Fig. 1. Power spectral densities of head and trunk movements for BVLs and healthy controls The mean population PSD data of the head and trunk while standing with eyes closed on foam support surface (ECF) are shown for BVLs (black lines) and healthy controls (grey lines). The upper panels show head data, the lower panels trunk data (at the level of the shoulders). The left panels from the pitch plane, the right panels from the roll plane. The noise levels of the measurement instruments (dotted lines) are also shown.

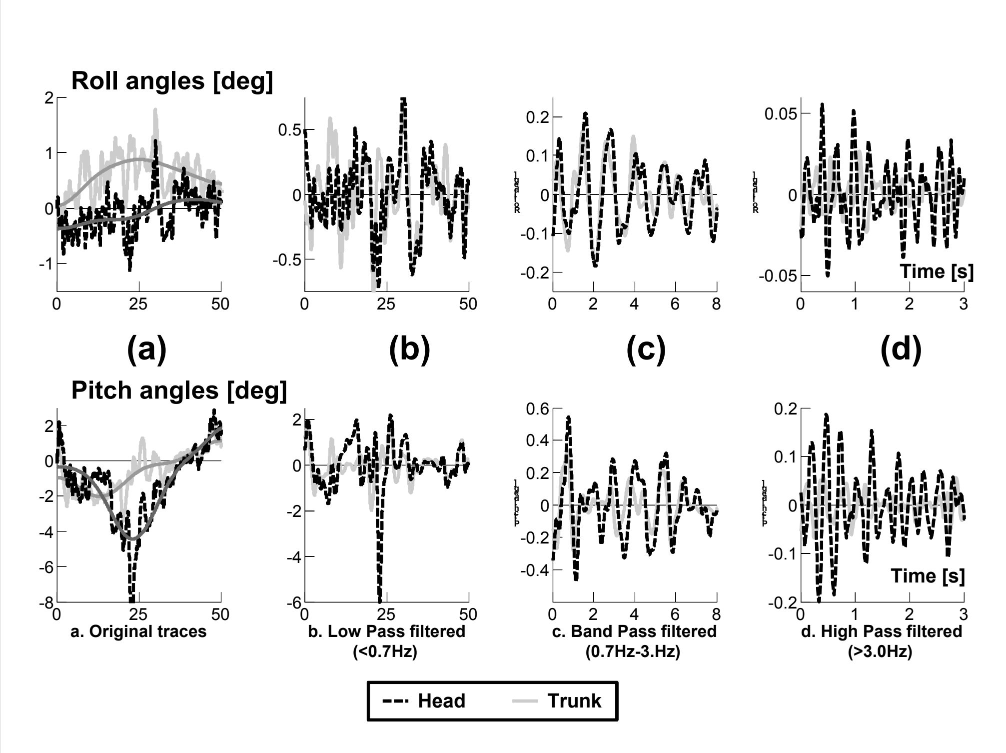


Fig. 2. Original traces of head and trunk movement of a typical BVL subject standing with eyes closed on a foam support surface (ECF). The upper panels for the roll plane, the lower panels in pitch plane. From left to right the figures show: (a) 50 s of original sway traces of head (black dashed lines) and trunk (grey lines) with general trend lines, (b) 50 s of the same traces as in (a) after removing the general trend and after low-pass (LP) filtering, (c) 8 s of the same traces as in (a) after removing the general trend and band-pass (BP) filtering, (d) 3 s of the same traces as in (a) after removing the general trend and after high-pass (HP) filtering.

Frequency analysis: relative amplitudes of motion between different body segments

To determine relative segmental motion, angular velocity data were used to calculate PSDs ratios of head-to-shoulder and pelvis-to-leg PSD values, transfer functions (TFs) and coherence functions. This was done with MATLABs Signal Processing Toolbox Version 6.3 .The estimates were based on Welch's averaging method (Welch, 1967), in which fast Fourier transformation was performed, employing a window size of 2048 samples, that is 20 s of data, with an overlap of 1024 samples (for further details, see Honegger et al. (2012)).

STATISTICAL ANALYSIS

Differences in slope regression angles were examined with the Harrison–Kanji test which is the circular analogue of a two-factor ANOVA. Posthoc differences in group means for each condition were examined with a Watson–Williams multi-sample test for equal means after testing for value directedness and concentration (Rao and Rayleigh tests). For the circular statistic analysis we used the 2010b update of the «Circular Statistic Toolbox» for MATLAB published by P. Berens in (2009).

Because the PSD data were not normally distributed, differences in PSDs of angular velocities, PSD ratios, coherences and transfer function magnitudes (TFMs) between the two groups, BVL and controls and stance conditions (stance tasks) were analysed using the non-parametric Mann–

Whitney test. For this purpose, the data were averaged over 10 frequency bins each of which combined values of 10 frequency samples, except for the two lowest frequency bins. Bins were logarithmically divided across the range of 0.00–10.25 Hz. The frequencies used were 0.15 Hz (mean of seven samples from 0 to 0.29 Hz), 0.2 Hz (mean of nine samples from 0 to 0.39 Hz), 0.42 Hz (0.2–0.63 Hz), 0.71 Hz (0.49–0.93 Hz), 1.39 Hz (1.17–1.61 Hz), 2.32 Hz (2.10–2.54 Hz), 3.69 Hz (3.47–3.91 Hz), 6.13 Hz (5.91–6.35 Hz), 7.59 Hz (7.37–7.81 Hz) and 10.03 Hz (9.81–10.25 Hz).

The significance of peak frequency shifts in the head to shoulder PSD ratios and TFs were inspected based on a function centroid (Weisstein, 2012) method. For each subject and test the centroid frequency $f_{c,test,subject}$ over the range from 1 to 7.6 Hz was calculated as a weighted average of the binned PSD ratios or TFMs: $f_{c,test,subject} = \frac{\sum_{1Hz<f(i)<7.6Hz} w_{test,subject}(f(i)) \cdot f(i)}{\sum_{1Hz<f(i)<7.6Hz} w_{test,subject}(f(i))}$ where $w_{test,subject}$ are either PSD ratio or TFM values.

The method is a development of spectral centroid frequency shift methods described in the literature (Schweitzer et al., 1979, Hary et al., 1982, Sieck et al., 1985, Parker et al., 1988). For the comparison of the frequency centroids and 90% angle ranges we used paired and unpaired samples t-tests as appropriate. For all comparisons of 90% angle ranges where, a priori, we assumed a larger value for the BVL population we used one-tailed t-tests. The statistics software used – IBM SPSS Statistics 19 – provides only two-tailed t-tests. The one-tailed values were derived by dividing the two-sided p-values by 2. For all statistics $p \leq 0.05$ was defined as significant, with corrections for multiple comparisons as necessary.

RESULTS

Upper body movements: BVLs versus controls

One of our main findings was that head, trunk and pelvis movements of BVL subjects had greater amplitudes than those of controls under the ECF stance condition ($p \leq 0.05$). For the pelvis and trunk the greater angle movements of BVL subjects compared to HCs for ECF were equally pronounced in the roll and pitch planes. Greater head angle movements than HC occurred only in the roll plane (Figs. 2–4) under the ECF condition. For all stance conditions trunk amplitudes of BVL subjects were greater than HCs in roll (Fig. 4). These results for the pelvis and trunk confirm our earlier findings for BVL subjects (Horlings et al., 2009).

Vestibulo-spinal compensation for a unilateral peripheral deficit can be judged using pitch and roll sway values of the pelvis for the ECF task (Allum and Adkin, 2003). We employed the same measures to examine the status of compensation of the BVL subjects (Table 1). In Table 1, patient values are listed in order of trial duration and roll sway. The table documents that no BVL was fully compensated because neither roll nor pitch sway amplitudes were within the normal range, except for subject 59. The subject (57) with the poorest compensation was excluded from analysis as the patient could not stand during the ECF task the minimum 10 s required for our analysis techniques. Inclusion of his data would have improved the significance of population differences noted in Fig. 4.

Head movements with respect to the trunk

Fig. 2 shows the trunk and head movements of a typical BVL subject. This figure illustrates our second main finding that head movements of BVL subjects are not larger than those of the trunk. Fig. 4 confirms this result for the complete BVL test population. Roll and pitch head amplitudes of BVL subjects were not greater than those of the trunk under the ECF condition (Fig. 4). Fig. 3 shows angle data in different frequency bands of a typical HC subject standing ECF. As seen in this figure and in the population mean plots of Fig. 4, the amplitudes of head sway were larger than those of the upper trunk for HCs across all frequencies and conditions. These results support our previous results with different age groups of HCs (Honegger et al., 2012). Summarising, in pitch both populations had greater head than trunk movements for the eyes open on foam (EOF) and closed on normal (ECN) conditions, but not BVL subjects for the ECF condition. In roll, only HCs had larger head than trunk movements, the BVL subjects did not have larger head movements.

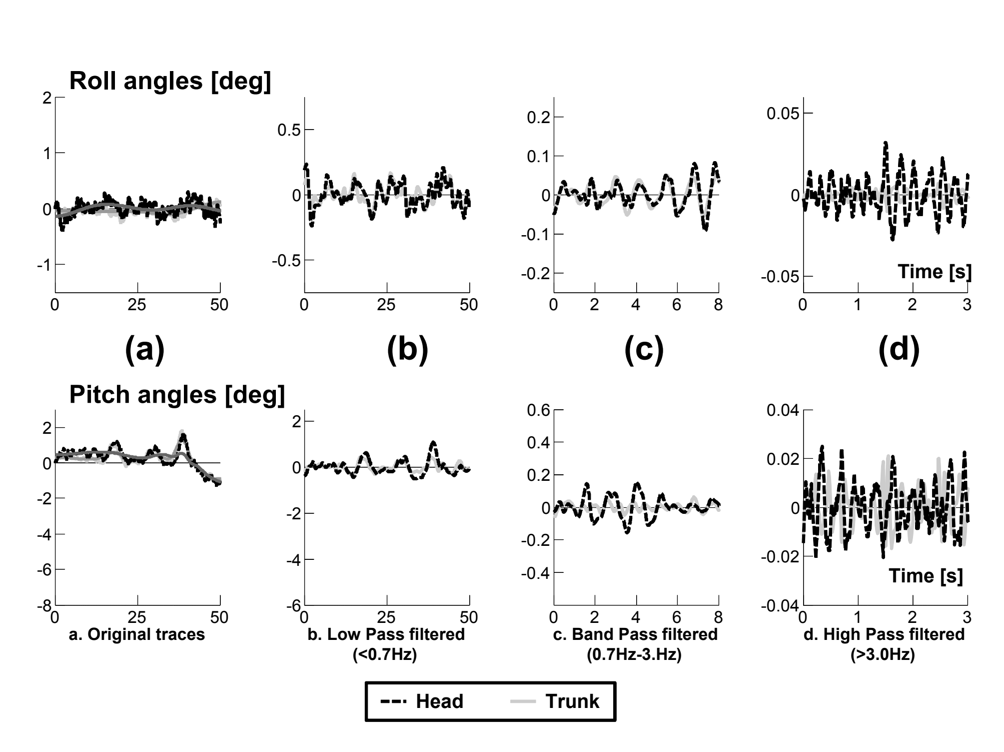


Fig. 3. Original traces of head and trunk movements of a typical healthy control subject standing with eyes closed on foam support surface (ECF). The layout and scaling of the figure is identical to that of Fig. 2.

We examined the amplitude and phase relationships of the head with respect to the trunk using the mean values of least square regression slopes of roll and pitch angle data, subdivided into LP, band-pass (BP) and high-pass (HP) bands. The results are shown in Figs. 5 and 6. Examples of this slope analysis are also shown in these figures for a BVL subject and for a HC subject standing ECF. Regression slope angles for all tasks and both groups are presented in the centre columns of Fig. 5 for the pitch plane, and Fig. 6 for the roll plane. For the LP and BP data the slopes of the regression lines of BVL subjects approached a 1:1 relationship and showed across conditions a significant slope difference with respect to HCs. Circular ANOVA calculations revealed the following group effects for pitch and roll, respectively: LP pitch with $F = 13.02$ and $p < 0.001$, BP pitch with $F = 15.03$ and $p < 0.001$, LP roll with $F = 6.68$ and $p = 0.013$, BP roll with $F = 13.25$ and $p < 0.001$. Significant ($p \leq 0.05$) post hoc group differences were only present under some stance conditions, even though a trend was observed across all stance conditions (see Figs. 5 and 6 centre columns a, b and c). The trends for BVL subjects were towards equal in-phase movement of head and trunk, that is slopes closer to 45°. BVL group means of LP and BP slope values were 45.3° and 46.6° in roll, supporting the evidence in Fig. 4 of equal head and trunk movement amplitudes for BVL subjects (note also the differences between BVLs and HCs in the example plots of Figs. 5 and 6). Perfect in-phase (head locked to trunk) would have a regression angle of 45°. The mean angles observed for LP pitch movements in BVL subjects were 44.8°. BP movements were less in-phase for pitch, 50.4° thus supporting the mean trend shown in Fig. 4 of larger pitch head than trunk movements. For HCs LP and BP slope values were higher, closer to 60°, indicating that the head moves more than the trunk in HCs as shown in Fig. 4. HP movements were slightly anti-phase, for both HCs and BVLs, close to 100° of phase indicating greater head than trunk motion for both groups in the frequency region above 3 Hz. There were no differences between HC and BVL subjects in this frequency band.

The PSD ratios for pitch and roll also showed that head on trunk movement was relatively less up to 3 Hz in all tests for BVL subjects (Fig. 7). For all tasks, the pitch and roll TFMs were less than 1 for low frequencies (≤ 2.32 Hz) in BVL subjects, that is, the transmitted velocity of the trunk to the head movement was less for BVL subjects, as expected from the aforementioned regression plots. The HCs have low frequency (below 1.4 Hz) TFMs in pitch that were slightly greater than 1. This pattern of differences between HCs and BVL subjects was similar but less pronounced for roll (Fig. 7). For frequencies above 1.4 Hz the roll and pitch PSD ratio and TFMs increased for HCs. The increase started above 2.3 for BVL subjects (Fig. 7).

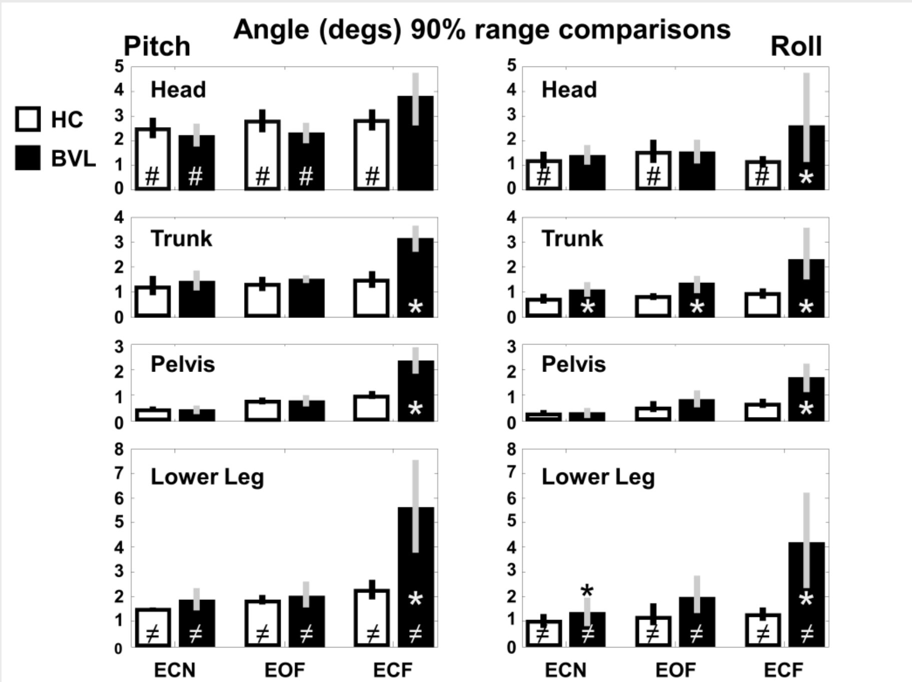


Fig. 4. Mean 90% sway angle range comparisons between BVLs and healthy controls (HCs). This column height shows the mean angle ranges (in degrees) with 95% confidence intervals as vertical bars, for head, trunk, pelvis and leg sway, in pitch (on the left) and roll (on the right) sway directions. The black columns show BVL means, the open columns show HC means. An asterisk (*) indicates mean values of BVLs greater than HCs (p ≤ 0.05), a hash symbol (#) when head motion has a greater amplitude than the trunk, and a not equals symbol (≠) when leg is greater than pelvis motion. Note that the ordinate scale is the same for all segment values.

The PSDs ratios peaked in the pitch plane at 3.2 Hz for HCs before reducing to a plateau value at 6.1 Hz. In contrast, BVL values peaked at 4.2 Hz, a significantly higher frequency ($p < 0.001$). A similar separation of
the resonant peak frequency between HC and BVL subjects was observed in the roll plane. However, the peaks occurred at higher frequencies, 4.6 Hz and 5.4 Hz, respectively, for HCs and BVLs, and were broader (differences $p < 0.001$). For the PSD ratios the differences in peak frequency were significant for all stance conditions in pitch and roll. For the TFMs differences were found for all stance conditions in roll. TFMs in pitch were significantly different for the EOF condition and when pooled across conditions.

Summarising, all three different analysis techniques we employed, 90% angle ranges, correlations of angle data in LP and BP regions, and ratios of PSD and TFM values across frequency, demonstrated that low frequency (≤ 1.4 Hz) head motion relative to the trunk was lower in BVL subjects than in controls. Resonant frequencies, which occurred at frequencies greater than 1.4 Hz, were shifted to higher frequencies in BVL subjects.

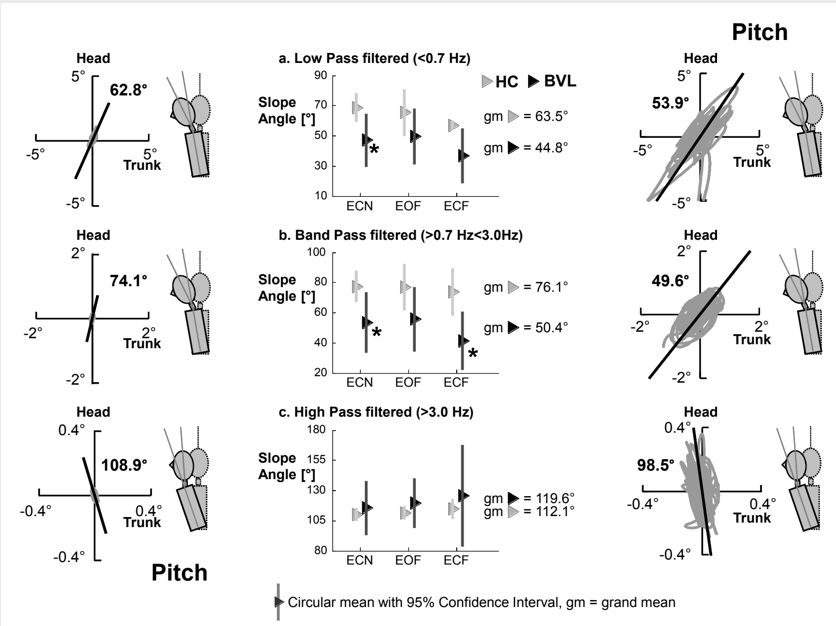


Fig. 5. Regression slopes of head angular movements with respect to the trunk in the pitch plane BVLs compared to healthy controls, with 95% confidence intervals. (a) Low-pass filtered, (b) band-pass filtered, (c) high-pass filtered. In the bar plots, the circular means and 95% confidence intervals are depicted by a triangle on its side and a vertical bar, respectively. Dark triangles mark BVL values, grey those of HCs. Significantly different means ($p \leq 0.05$) are marked with an asterisk (*) symbol. To the left and right examples of regression of head movements with respect to those of trunk and corresponding slope lines are shown for two typical subjects. The numerical values on these plots are the slope angles.

Pelvis movement with respect to the lower leg

The movement strategies of the pelvis with respect to the leg were not different between BVL and HC subjects, involving greater lower-leg motion than pelvis motion (see Figs. 4 and 8). As can be observed in Figs. 4 and 8 the amplitude of leg movements for BVL subjects was greater under ECF conditions.

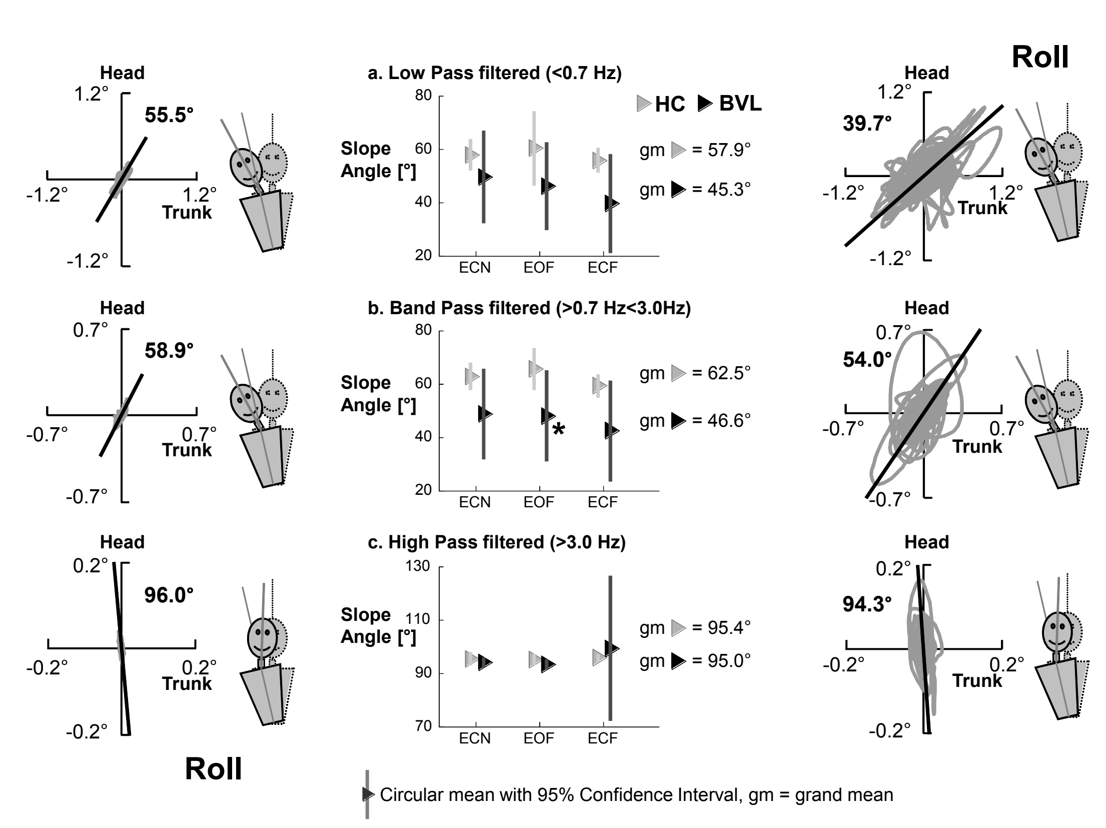


Fig. 6. Regression slopes of head angular movements with respect to the trunk in the roll plane. The layout of the figure is identical to that of Fig. 5.

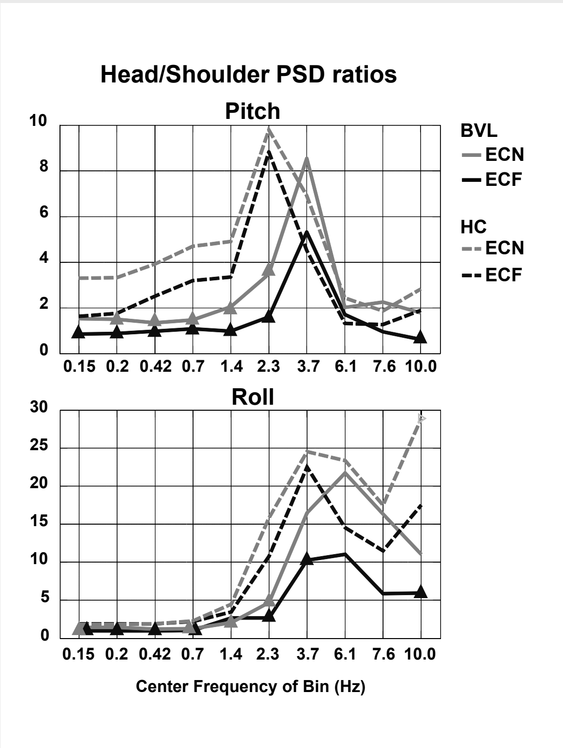


Fig. 7. Mean population PSD ratios of head-to-trunk movements for eyes closed tests in pitch and roll planes across the ten frequency bins listed in the methods section. Each triangle on the BVL bin values signifies that the BVL value significantly differs from that of the HCs for that frequency bin (tested with Mann–Whitney test). Full lines are for BVLs, dashed lines for the HCs.

DISCUSSION

The primary aim of our study was to examine the relationship between head and trunk movements during quiet stance in subjects with bilateral peripheral vestibular loss (BVL), compared to that of HCs. Two strategies have been described concerning head movements during stance. One strategy is to lock the head to the trunk, and for the other, the head is stabilised in space (Horak and Nashner, 1986). The results from this study indicate that BVL subjects, in contrast to HCs, use the strategy of locking the head to the trunk for movements with frequency content less than 3 Hz. Presumably, cervico-collic reflexes are enhanced in BVL subjects to achieve this head on trunk stabilisation.

In HCs the head moves in phase with trunk movement too, but with greater amplitude for all stance conditions. Thus, the presence of a functioning vestibular system does not lead to the head being more stabilised in space than without vestibulo-spinal action, rather the pelvis is more stabilised in space, as these and other results indicate (Horlings et al., 2009, Honegger et al., 2012). The control of head motion with respect to the trunk, for frequencies above 3 Hz, is with a slightly anti-phasic motion with

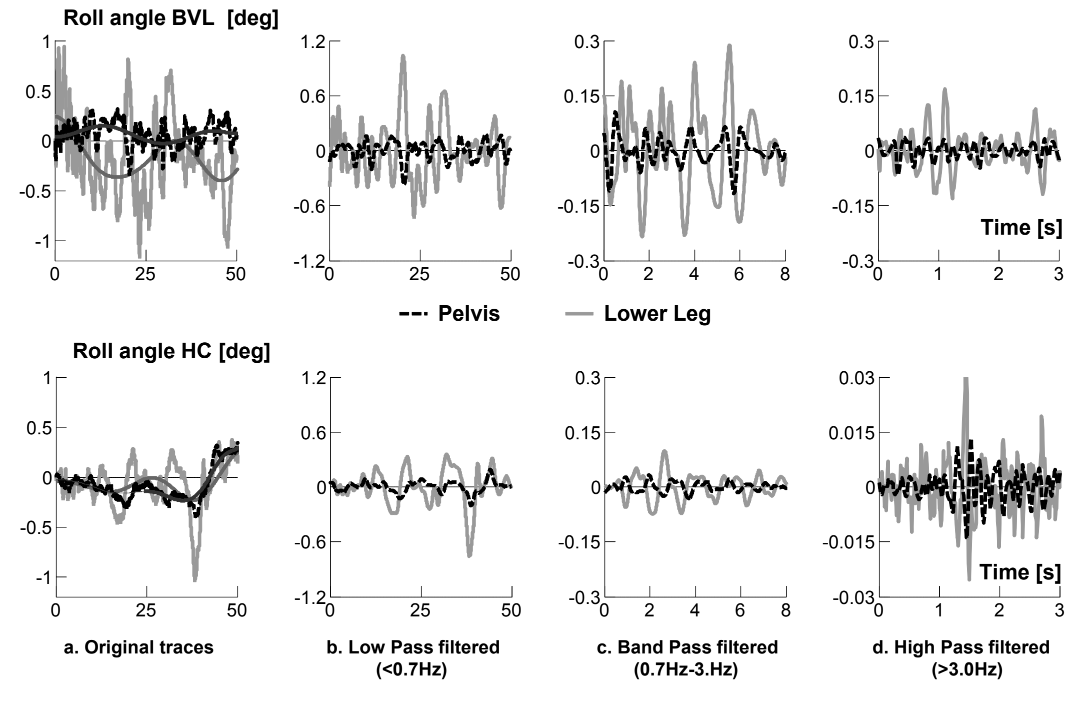


Fig. 8. Original traces of lower leg and pelvis roll movements for typical BVL and HC subjects (same subjects as in Figs. 2 and 3) standing with eyes closed on foam support surface (ECF). The upper panels are for the BVL subject, the lower panels for the HC. From left to right the figures show: (a) 50 s of original recordings of pelvis and lower leg sway with general trend lines, (b) 50 s of the same traces as in (a) after removing the general trend and after low-pass (LP) filtering, (c) 8 s of the same traces as in (a) after removing the general trend and band-pass (BP) filtering, (d) 3 s of the same traces as in (a) after removing of the general trend and after high-pass (HP) filtering. Note the larger leg than pelvis movements in both sets of traces.

respect to the trunk. As this high frequency mode is similar in BVL and HCs, we assume that at high frequencies the vestibular system has little effect on the relative movement of the head on the trunk. The results for HCs duplicate those of our previous studies (Honegger et al., 2012). Here we demonstrate that the head is more strongly coupled to trunk motion in BVL subjects. In both populations, head to trunk strategies are bimodal.

The results question the idea that the CNS of vestibular loss subjects adopts a different response strategy depending on the task at hand. For quiet stance, although trunk motion is greater than normal in BVL subjects, the head motion is not greater than trunk motion. This is in contrast to the greater head than trunk motion of HCs. Thus for several stance conditions this results in BVL head motion that is not greater than normal. A similar situation occurs for perturbed stance in which trunk motion is unstable and greater than that of HCs (Shupert and Horak, 1996, Carpenter et al., 2001). However, the head motion is not greater than normal in well compensated BVL subjects (Allum et al., 1988, Shupert and Horak, 1996, Allum et al., 1997). Likewise, a similar situation exists for gait in which head movements are also not greater than normal (Pozzo et al., 1991, Mamoto et al., 2002). What seems to be lacking during gait is the finely controlled head pitch movement to compensate for the vertical head translation induced with each toe-off event (Pozzo et al., 1991, Mamoto et al., 2002). Locking the head to the trunk is detrimental to gaze stability during gait. For this task compensate for the vertical head translation induced with each toe-off event (Pozzo et al., 1991, Mamoto et al., 2002). Locking the head to the trunk is detrimental to gaze stability during gait. For this task however, efference copy of the head movement on to vestibular nucleus neurons may be used to stabilize gaze (Sadeghi et al., 2012).

Motion of the head during quiet stance in HCs is dominated by resonances at approximately 3–4 Hz. We observed that the resonance was shifted to a higher frequency in BVL subjects. We had expected that the resonance observed in HCs (Honegger et al., 2012) would be enhanced in BVLs due to an inability to use vestibulo-collic reflexes to stabilise the head as suggested by Goldberg and Cullen (2011). In fact the amplitude of the resonant peak in BVL was if anything less in BVL subjects. As suggested above concerning locking the head to the trunk we assume that BVL patients compensate for their loss by enhancing their proprioceptive neck (cervico-collic) reflexes, because in both tasks with EC (i.e. eliminating compensation for the vestibular loss with visual inputs) this result was observed. Enhancement of neck proprioceptive reflexes in BVL subjects was also noted by Pyykko et al. (1991).

Theoretically, the VCR should stabilise the head in space by opposing head movements during unexpected or voluntary trunk movements (Ito et al., 1997) and it should dampen head oscillations that occur due to the heads passive mechanics with respect to the trunk. Here, however, it appears that the cervico-collic reflex alone is more effective in damping motion of the head with respect to the trunk than in combination with the VCR. However, the motion of the trunk and pelvis is highly dependent on vestibulo-spinal reflexes as shown with BVL subjects when visual and lower leg proprioceptive inputs are eliminated, respectively, less effective when standing EC on a foam support surface (ECF task).

The ECF task is extremely effective in quantifying the amount of compensation for a peripheral vestibular deficit (Allum and Adkin, 2003).Generally, stance durations increase and sway decreases as compensation progresses. Other measures of vestibulo-spinal control, for example, responses to support surface perturbations can be employed as measures of compensation too (Allum et al., 1988, Horak, 2009). The subjects in our BVL patient group whose data we analysed were able to stand without falling for at least 30 s during the ECF task. The data of the least well compensated subject who could not stand for more than 10 s during this task were eliminated from the analysis. Well compensated unilateral loss subjects can stand for at least 20 s during the ECF task without falling (Allum and Adkin, 2003). Thus we assume that our analysed subjects were well compensated for their loss even though their loss was acquired as an adult.

Our results also support those of Keshner (2000) who investigated the selection of strategies for head stabilisation in young and old subjects, by having subjects change the head-neck system through actively relaxing or stiffening their neck muscles. Neural (i.e. reflex) damping was present for frequencies up to about 2.5 Hz, thus suggesting that for low frequencies (<3 Hz) cervico-collic reflexes function to stiffen the neck muscles, in order to keep the head locked to the trunk. Summarising, the change in characteristics of head to shoulder motion shown by BVL subjects is consistent with a locking of the head to the trunk by increased use of cervico-collic proprioceptive reflexes. However, as we did not measure neck muscle activity we cannot eliminate the possibility that a general neck stiffening approach is used by BVL subjects in order to reduce the number of joints controlled. Nonetheless, our suggestion for increased use of proprioceptive inputs onto neck motor pathways normally dominated by vestibule-spinal inputs would concur with the observations of Sadeghi et al. (2012) on vestibulo-ocular pathways. These authors showed that absent inputs to vestibular nucleus neurons were partially compensated for with neck proprioceptive inputs and during active head movements with efference copy signals. In followup investigations it might be worthwhile to examine whether a better head stabilisation on the trunk could be achieved with neck proprioceptive rather than vestibulo-spinal signals. Movement strategies between the lower-legs and pelvis were not different between BVL and HC subjects. Movements were larger for the lower-leg than the pelvis under all stance conditions and larger than HCs for the BVL subjects under the eyes-closed-foam condition. This result raises the question of how leg and upper body motion is coordinated. To expand on the current study and answer this question, sensors would need to be mounted on the upper leg in addition to the lower leg and pelvis in order to discriminate between ankle, knee and hip joint motion. Our previous results (Horlings et al., 2009, Honegger et al., 2012) indicate that for low frequencies the trunk and pelvis move as a «floppy» inverted pendulum but our current results indicate that this is not the case for the lower body.

Conflicts of Interest - *The authors report that J.H.J. Allum and F. Honegger worked as consultants for the company producing SwayStar, part of the equipment used in this study.*

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GLOSSARY

Roll: angle rotation in the lateral plane, i.e. side-to-side

Pitch: angle rotation in the sagittal plane, i.e. back-forward

ABBREVIATIONS

BP, band pass; BVL, bilateral vestibular loss; EC, eyes closed; ECF, eyes closed on foam; ECN, eyes closed on normal; EO, eyes open; HC, healthy control; HP, high pass; LP, low pass; PSD, power spectral density; TF, transfer function; TFM, transfer function magnitude; VCR, vestibulo-cervical reflex; VOR, vestibular ocular reflex.

Chapter 4

The effect of prosthetic feedback on the strategies and synergies used by vestibular loss subjects to control stance[1]

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ABSTRACT

Background:
This study investigated changes in stance movement strategies and muscle synergies when bilateral peripheral vestibular loss (BVL) subjects are provided prosthetic feedback of pelvis sway angle.

Methods:
Subjects performed 3 stance tasks: standing eyes closed, on a firm and foam surface, and eyes open on foam. Pelvis and upper trunk movements were recorded in the roll and pitch planes. Surface EMG was recorded from pairs of antagonistic muscles at the lower leg, trunk and upper arm. Subjects were first assessed without feedback. Then they received training with vibro-tactile and auditory feedback during stance tasks before being reassessed with feedback active.

Results:
Feedback reduced pelvis sway angle displacements to values of age-matched healthy controls (HC) for all tasks. Movement strategies were reduced in amplitude but not otherwise changed by feedback within analysed frequency bands. These strategies were not different from those of HCs before or after use of feedback. Low frequency motion was in-phase and high frequency motion anti-phasic. Feedback reduced amplitudes of EMG, activity ratios (synergies) of antagonistic muscle pairs and slightly reduced baseline muscle activity.

Conclusions:
This is the first study demonstrating how vestibular loss subjects achieve a reduction of sway during stance with prosthetic feedback. Unchanged movement strategies with reduced amplitudes are achieved with improved antagonistic muscle synergies. This study suggests that both body movement and muscle measures should be explored when choosing feedback variables, feedback location, and patient groups for prosthetic devices which reduce sway of those with a tendency to fall.

Key words: Prostheses and Implants, Feedback/physiology, Vestibular Diseases, Postural Balance/physiology, Electromyography, Middle Aged.

BACKGROUND

Loss of vestibular function is well-known as a factor underlying an increased tendency to fall in older persons (Tinetti et al., 1988). Furthermore, a number of persons with less than 60 years of age, having either a unilateral or bilateral peripheral vestibular loss (UVL or BVL) have difficulty maintaining their balance particularly for complex gait tasks such as climbing stairs, or for stance tasks for which visual and proprioceptive inputs are reduced, such as standing on a carpet in the dark (Allum and Adkin, 2003). For these reasons, a number of investigators have developed balance prostheses to provide persons with balance problems a replacement for vestibular sensory information on their center of mass sway. Such prosthetic systems generally rely on vibro-tactile or auditory feedback or both modes, appropriately coded with body sway information (Dozza et al., 2005, Hegeman et al., 2005, Horak et al., 2009, Vichare et al., 2009, Davis et al., 2010, Allum et al., 2011, Goodworth et al., 2011). There are variations concerning where the sway measures are taken, and how the sway signals are processed, and at which body location the feedback is provided. However, the general conclusion is that such biofeedback helps UVL and BVL patients improve their balance during stance and gait (Horak et al., 2009, Allum et al., 2011, Goodworth et al., 2011).

There are different opinions on whether position or velocity feedback should be used for balance prostheses. Early results suggest that angle position feedback was more effective than velocity feedback of pelvis sway (Hegeman et al., 2005) and when employed lead to reduction of angle rather than velocity of sway (Davis et al., 2010). Others have used a combination of angle and velocity feedback with success (Horak et al., 2009, Vichare et al., 2009, Goodworth et al., 2011). Another crucial question is whether, for effective sway reduction, patients need to use a particular type of movement strategy. One way to begin answering this question is to examine a patient group such as those with vestibular loss that have similar but exaggerated movement strategies to those of healthy controls (HCs) (Horlings et al., 2009b). For HCs, feedback of pelvis sway angle is known to be effective in improving balance control (Dozza et al., 2005, Horak et al., 2009).

Vestibular loss patients have movement strategies during stance that are similar to those of controls, whereas those with lower-leg proprioceptive loss, for example, have different movement strategies (Horlings et al., 2009b). For this reason, it has been suggested that subjects with vestibular loss might be more responsive to biofeedback modes that function well for healthy controls (Horlings et al., 2009a). Thus the first question this study sought to answer was whether improvements in balance control achieved by vestibular loss subjects using artificial sway position feedback were brought about using the same movement strategies as when no feedback was available. This question is by no means as simple as considering body sway during stance as similar to that of an inverted pendulum, because the upper and lower parts of the body move with two modes simultaneously during stance (Creath et al., 2005, Horlings et al., 2009b). One mode, a low frequency (<0.7 Hz) mode, is like a «floppy» inverted pendulum with mostly in-phase motion of the pelvis and trunk and the other, high frequency (> 3 Hz) mode, is an anti-phase motion of the 2 segments. Between 0.7 and 3 Hz, motion transitions between these 2 modes. Thus the muscle synergies possibly underlying reductions in amplitudes of these movement strategies in these two modes must at least be driven by muscles acting at the ankle joints and at the trunk. If as shown by Goodworth et al.(Goodworth et al., 2011), improvements are only present in the low-frequency inverted-pendulum mode of motion, then changed control of muscle activity at the ankle joint would be sufficient for better balance. The typical changes in muscle synergies observed with vestibular loss in response to rotational perturbations of the support surface cause changes in both modes of motion described above for control of quiet stance (Carpenter et al., 2001). These changes consist of reduced ankle muscle activity but increased trunk muscle activity with respect to responses of healthy controls(Allum and Honegger, 1998, Allum and Carpenter, 2012). Thus the second question we have attempted to answer with this study is how muscle synergies are changed when biofeedback of pelvis sway is provided to BVL subjects.

METHODS

Participants

Fourteen adult subjects (6 bilateral peripheral vestibular loss (BVL) subjects and 7 age-matched healthy control subjects) were included in this study. The BVL subjects were out-patients at the University Hospital of Basel, Switzerland. The BVL subjects ranged from 45 to 52 years of age (Mean 48.8; SD 2.2) and the controls ranging from 41 to 53.2 years had a mean age 49.2 (SD 4.4). Inclusion criteria for the BVL subjects was the absence of vestibular ocular reflex responses to caloric irrigation of both ears, rotating chair (yaw and pitch) responses less than the lower 5% bound of normal subjects, normal auditory evoked potentials and normal magnetic resonance imaging of the brain. Exclusion criteria for the healthy subjects included self-reported sensory, neurological or musculoskeletal impairments that could interfere with balance and inability to stand on one leg, eyes closed, for 20 seconds without falling. All subjects provided informal written consent to participate as required by the ethical committee of the University Hospital of Basel who approved the study.

Testing Procedures and Data Recording

Two gyroscope based systems, SwayStar (Balance International Innovations GmbH, Switzerland), measuring angular velocities in the roll and pitch planes, were used. One SwayStar system was mounted on a belt strapped around the hips in order to measure pelvis movements. The other system was mounted on a plate placed between the scapulae and held in place with a tightly fitting shoulder harness. These angular velocities were sampled at 100 Hz with 16 bit accuracy over a range of 327 deg/s and then transferred wirelessly to a PC which computed angle changes via trapezoid integration (Allum and Carpenter, 2005). Muscle activity was measured with pairs of surface, silver-silver chloride EMG electrodes. These electrodes were placed 3cm apart, along several muscles on the left side of the body: tibialis anterior, soleus, external obliques, paraspinalis at L4-L5, and the medial deltoid. EMG signals were recorded using a preamplifier with a gain of 1000 and a band-pass of 0.7 Hz to 2.5 kHz. The signals were analog band-pass filtered between 60 and 600 Hz, full-ware rectified and low-pass filtered at 100 Hz with a 3rd order Paynter filter prior to sampling at 1 kHz (Gottlieb and Agarwal, 1970).

A BalanceFreedom feedback system (Balance International Innovations) provided biofeedback of pelvis sway to the participants using signals from the pelvis gyroscopes. Actuators for the feedback were mounted on a head band and were active once angle thresholds for activating the vibro-tactile, auditory and visual actuators were exceeded. Feedback thresholds were based on individual values of the 90% ranges of pelvis sway in the pitch and roll directions computed for the 70 sec duration (or less) of each task of the first assessment. If a loss of balance occurred, the task was repeated a maximum of two times and the trial with the largest duration was used. The thresholds in each direction were set at ±40% of the 90% range for vibro-tactile signals (that is, a range equal to 80% of the 90% measurement), ±80% for the acoustic signals and ±150% for the visual threshold. Once activated, each feedback signal remained active as long as its threshold was exceeded. The vibro-tactile activation signal was sent to 1 of 8 vibrators in the headband set at 45˚ intervals around the headband. A vibrator switched on when the sway threshold was reached in the direction of the sway and this direction had the largest sway amplitude. The acoustic feedback consisted of two bone-conducting acoustic actuators placed above the ears at the level of the mastoids. The left actuator was activated at 870 Hz when the acoustic threshold was reached for sway to the left, the right actuator at 500 Hz when swaying to the right, and both conductors with a frequency of 1370 Hz and 250 Hz when swaying backwards and forwards, respectively. The visual feedback served as a flashing warning signal regardless of sway direction (Davis et al., 2010).

Pelvis and trunk sway and EMG signals were recorded for each of the stance tasks. The tasks were performed without shoes, and with the arms hanging alongside the body. Two stance tasks were performed on a foam surface, eyes open and eyes closed and one task eyes closed on a normal surface. First the original assessment was performed. Then, subjects rested for 20 minutes, before 30 minutes of training was provided with biofeedback. The training tasks were the same as the assessment tasks but also included tandem stance on a firm surface, eyes open and closed. After another short pause of 5 minutes, subjects were reassessed on the 3 stance tasks with the feedback active. During all tests, two spotters stood close to, but behind, the subjects in order to prevent a potential fall.

DATA PROCESSING AND STATISTICAL ANALYSIS

Once the original velocity data was integrated into angular data, the low frequency trend, determined by applying the dynamically parameterized denoise function of the «Rice Wavelet Toolbox» (Baraniuk et al., 2002), was subtracted from the original data. The resulting data were subsequently filtered and separated into three frequency bands (see Fig. 1.): low pass (<0.7 Hz), high pass (>3.0 Hz) and band pass (0.7-3.0 Hz). The filtering is based on previous observations of power spectral densities (PSDs) of pelvis sway (Horlings et al., 2009b) – two resonances are observed, one below 0.7 Hz, the other above 3 Hz. This filtering was implemented using simple 3rd order Butterworth filters running forwards and backwards over the data.

To measure pelvis-trunk coordination, total least-squares regression lines (Krystek and Anton, 2007) were calculated in each frequency band with respect to the x-y plots of trunk versus pelvis angle samples for the roll and pitch data (see Fig. 1. and 4.). The slope angles of the regression lines were further processed and visualized by applying circular statistics (Jammalamadaka and SenGupta, 2001). For statistical calculations the «Circular Statistic Toolbox» update 2010b published by P. Behrens (Berens, 2009) was used. For the graphical representation (Fig. 4.) negative slope angles where mapped to the corresponding positive 2nd quadrant. Slope angles represent axial data and are bimodal because for example -70º and 110º represent the same data point ($\tan(\alpha) = \tan(\alpha + 180)$). Additionally, with the help of the MATLABs Signal Processing Toolbox Version 6.13 (R2010A) we calculated power spectral densities (PSDs) and PSD ratios for the EMG data (after 100 Hz low-pass filtering) data. Fast Fourier transformation was performed on a window size of 2048 samples (20.5s) with an overlap of 1024 samples. Shifts in EMG baseline activity due to feedback were examined by considering the first quartile – 25%- values (Q1) of EMG amplitude samples. Changes in EMG amplitude modulation were investigated using the 90% ranges of EMG samples. Data of near falls at the end of the recordings were associated with large changes in EMG activity and were excluded from the analysis.

Differences in slope regression angles between groups and conditions were examined with the Harrison-Kanji test which is the circular analogue of a two-factor ANOVA. Posthoc differences in group means for each condition were examined with a Watson-Williams multi-sample test for equal means after testing for value directedness and concentration (Rao and Rayleigh tests). For the circular statistic analysis we used the 2010b update of the «Circular Statistic Toolbox» for MATLAB published by P. Behrens in (Berens, 2009).

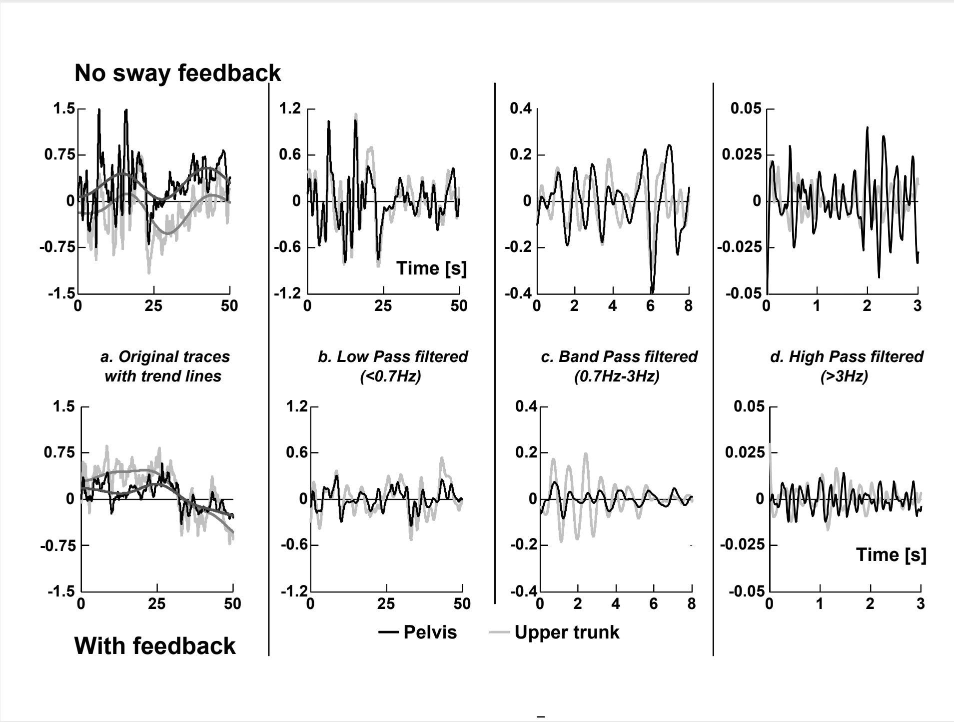

Fig. 1. Improvement in roll sway for a BVL subject when provided biofeedback of pelvis sway The upper four panels show the sway angles in degrees of the pelvis and upper trunk while standing eyes closed on foam with no biofeedback, the lower panels with biofeedback. a. 50 seconds of the original traces of upper trunk and pelvis sway with general trend lines. b. The same traces as in a. after removing the general trend and low pass filtering. c. 20 seconds of the same traces as in a. after removing the general trend and band pass filtering. d. 5 seconds of the same traces as in a. after removing the general trend and high pass filtering.

Because neither Q1 values nor the PSD of EMG data were normally distributed, differences in PSDs and PSD ratios (as a measure of coactivation at the trunk and ankle joint muscles), between the two groups, BVL (with and without feedback) and controls across stance conditions (stance tasks) were analysed depending whether the situation requires a paired or independent samples test using either the non-parametric Mann-Whitney or the Wilcoxon Signed Ranks test . To examine PSD amplitudes and ratios, the data were averaged over 19 frequency bins that combined values of 3 adjacent frequency samples. These bins are equally spaced on a logarithmic scale across the range of 0.00 – 40.04 Hz. The bins were: 1.95 (1.46-2.44 Hz), 2.44 (1.95-2.93),.2.93 (2.44-3.42), 3.91 (3.42-4.39), 4.39 (3.91-4.88), 4.88 (4.39-5.37), 6.84 (6.35-7.32), 8.30 (7.81-8.79), 9.77 (9.28-10.25), 11.23 (10.74-11.72), 13.18 (12.70-13.67), 15.63 (15.14-16.11), 18.07 (17.58-18.55), 21.00 (20.51-21.48), 24.90 (24.41-25.39), 29.30 (28.81-29.79), 34.18 (33.69-34.67), 40.04 (39.55-40.53).

For all statistics p≤0.05 was defined as significant, with corrections for multiple comparisons as necessary.

RESULTS

The dual mode, vibro-tactile-auditory feedback, helped BVL subjects reduce sway angles at the pelvis and the upper trunk. Fig. 1. provides an example of the improvement for a typical BVL subject who had considerable difficulty to stand eyes closed on the foam surface (ECF). Fig. 2. shows the mean BVL population 90% ranges of sway angle and angular velocity at the pelvis for the eyes closed and open on foam compared to mean values of the healthy controls (HC). There was a population improvement which was significant for all ECF measures in the angular position amplitudes of sway.

As shown in Fig. 2., the sway amplitudes for the pelvis of BVL subjects in roll and pitch were significantly reduced with feedback to levels that were not different from those of HCs. There was no change in velocities when BVL subjects stood with feedback eyes open on foam, but the velocities were not different from HCs even without feedback. In contrast, velocities standing ECF were reduced with feedback, but levels were still greater than those of HCs (Fig. 2.).

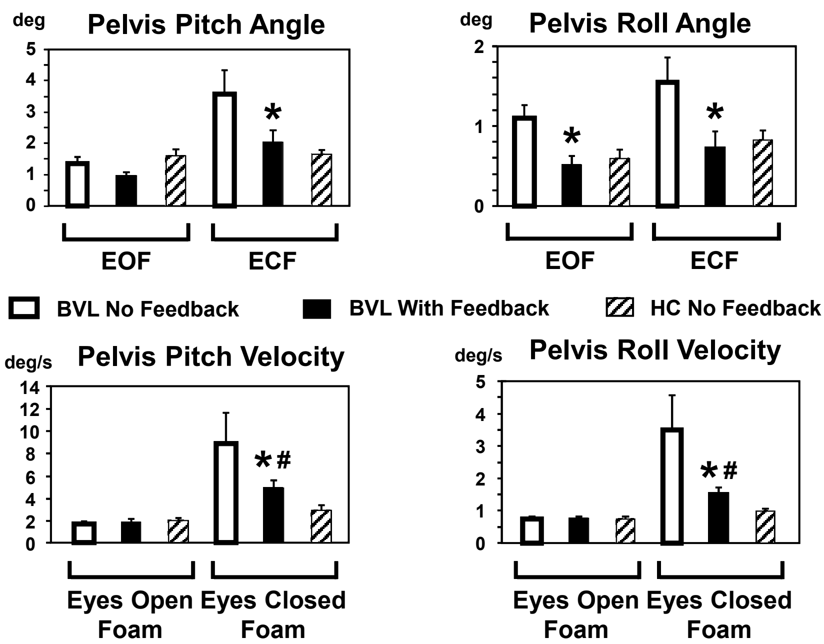

Fig. 2. Population angle and angular velocity means of 90% pelvis sway ranges with and without feedback. The column height represents the mean population pelvis angle or angular velocity for each task. Values are shown for stance tasks on foam with eyes open (SEOF) or closed (SECF). The vertical line above the column indicates the standard error of the mean. BVL stands for bilateral vestibular loss population, HC for the healthy control population. BVL means with feedback marked with * have a significant decrease compared to means with no biofeedback. If the biofeedback means of BVL subjects remained significantly greater than healthy controls with feedback the BVL values are marked with #.

When angle displacements of the trunk and pelvis for the ECF condition were split into different frequency bands, the most consistent improvements across frequencies were noted for the roll direction (significant reduction for all bands except trend for trunk roll) . Nonetheless, a trend for pitch improvements occurred in several frequency bands. However, these pitch trends were only significant for 2 bands (<0.7 Hz trunk and >3Hz pelvis). Thus for both roll and pitch there was no restriction to only low frequency bands (<0.7 Hz) for improvement.

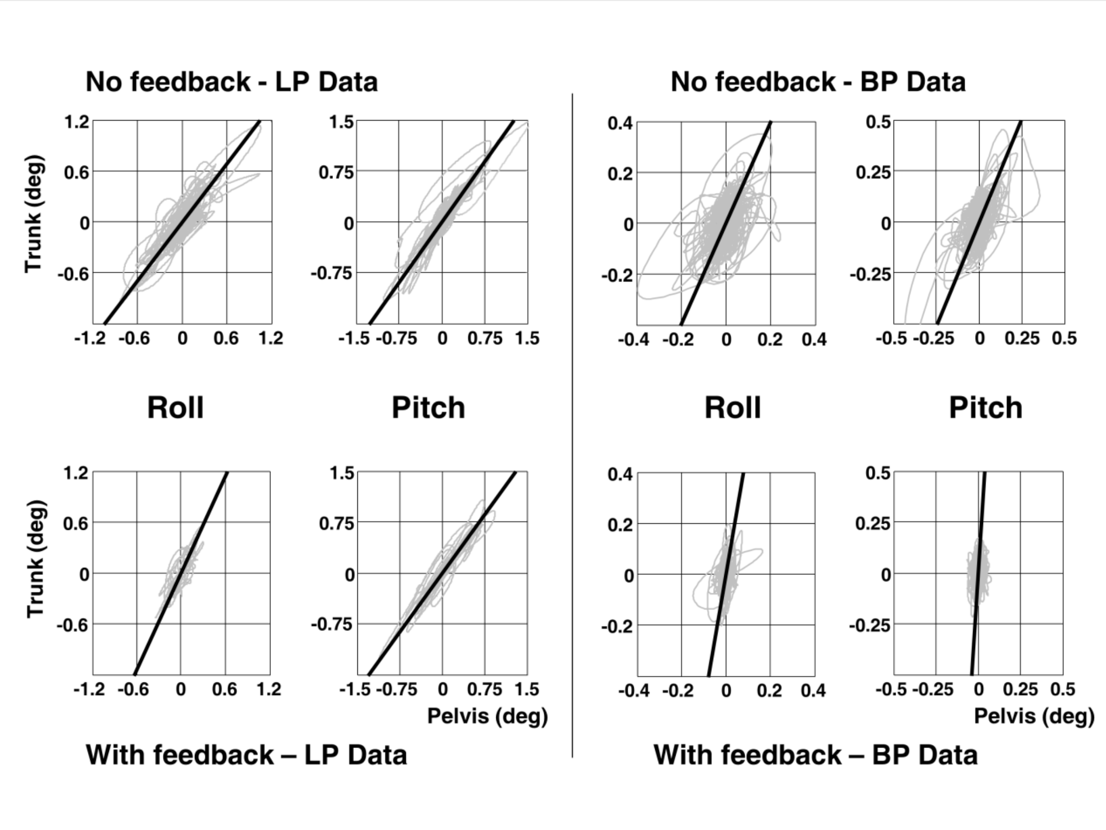

Fig. 3. Regression slopes of trunk versus pelvis movements of a BVL subject standing eyes closed on foam with and without feedback. Regression slopes after low-pass filtering (LP) and band-pass filtering (BP) of trunk and pelvis sway angle traces are shown. The upper panel shows the slopes without biofeedback, the lower ones with biofeedback. On the left are the low pass regressions, on the right the band-pass regressions.

Phases representing movement strategies present between the pelvis and upper trunk were examined using correlation plots of pelvis and trunk angle divided into low (<0.7 Hz), middle (0.7 to 3 Hz), and high (>3 Hz) band widths. Perfectly in-phase movement-synergies between the trunk and pelvis would imply that the upper body moved as an inverted pendulum and correlation plots would have slope lines of 45°. The example low pass plots of Fig. 3. indicate near in-phase low pass (LP) movements, with pitch more in phase than roll. This trend is confirmed by the population values shown in Fig. 4a. The mid-frequency (0.7 to 3 Hz) movements were still in-phase but are restricted to mostly trunk movements on a fixed pelvis (see Fig. 3 left and 4b). High frequency (>3 Hz) movements were characterized by anti-phase motion (regression lines values greater than 90°). When BVL population values of the phase relationships were examined in each frequency band without and with feedback, no changes in these phase characteristics were observed when feedback was provided. Furthermore, the phase relationships did not differ from those of HCs.

The data of BVL subjects described in Figs. 1-4 is consistent with a decrease in amplitude modulation when sway feedback is provided rather than any changes in movement strategies. The question arises how these changes in amplitude modulation are brought about. The patterns of und-

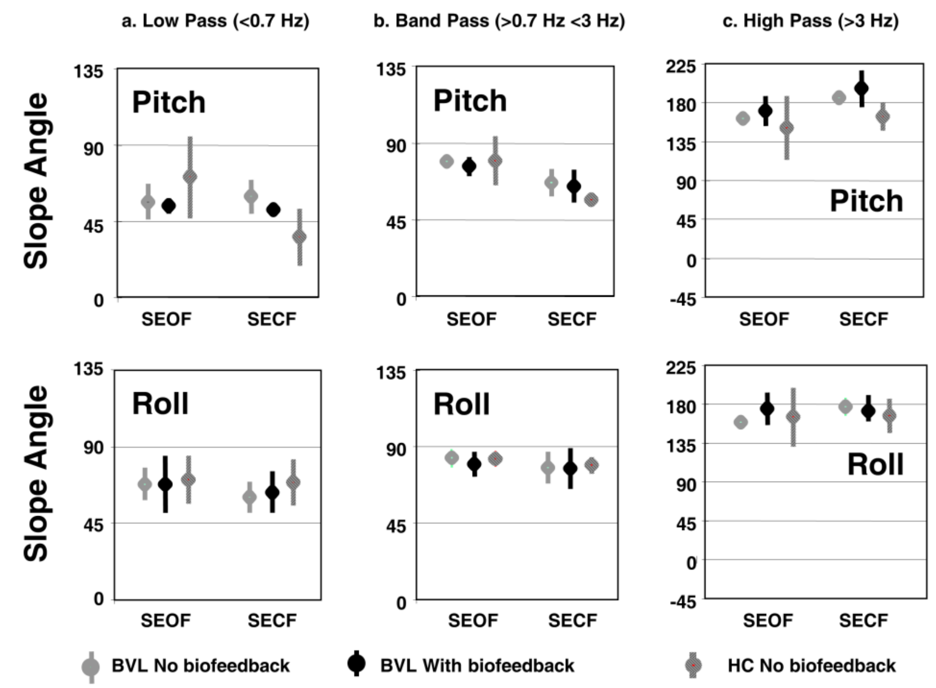

Fig. 4. Mean regression slopes of trunk angles with respect to pelvis angles. Mean regression slopes are depicted for the two foam tasks, eyes open and closed, without and with biofeedback. The upper panels show the slope angles in pitch, the lower panels in roll. a. slope values for low-pass filtered angle values, b. band-pass filtered angles and c. for high-pass filtered values. The bullet symbol marks the mean value of the slope; the vertical line depicts the 95% confidence intervals of the mean.

derlying muscle activity shown in Fig. 6 suggest three mechanisms. One mechanism is to change in the depth of modulation of muscle activity as depicted by the example in Fig. 5.(left). The change in amplitude modulation across frequencies was examined with PSDs of the filtered EMG activity compared on and off feedback (see Fig. 6A). A significant reduction of activity was observed for arm, ankle and trunk muscles across almost all frequency bands except for external obliques for which only a trend ($p<0.1$) was observed. Fig. 6A shows the mean population values for the ankle and trunk muscles. Although there were slight differences between BVL PSD values with feedback and those of healthy controls without feedback (see, for example, paraspinal in Fig. 5.), those differences were not significant, except for the tibialis anterior and deltoid muscles. These exceptions demonstrated lower activity levels in controls (Fig. 6A). A second manner in which movements could be reduced is by lowering the general level of muscle activity and therefore intrinsic muscle stiffness. For this purpose, we used the median level lower 25% of the EMG histograms as a measure of the baseline EMG activity. With feedback, these levels were significantly reduced in the paraspinal and deltoid muscles for ECF conditions. The third method for changed synergies we examined was if coactivation ratios, using the PSD ratios of ankle and trunk muscle activity, changed, as shown in Fig. 6B. There was a trend for the ratio of paraspinals to obliques and tibialis to soleus to be reduced with feedback for ECF conditions. This trend was significant across several frequency bins for the ratio of tibialis anterior to soleus (for details see Fig. 6B). Interestingly, the variances of BVL ratios was markedly smaller with feedback, however, still larger than those of HCs.

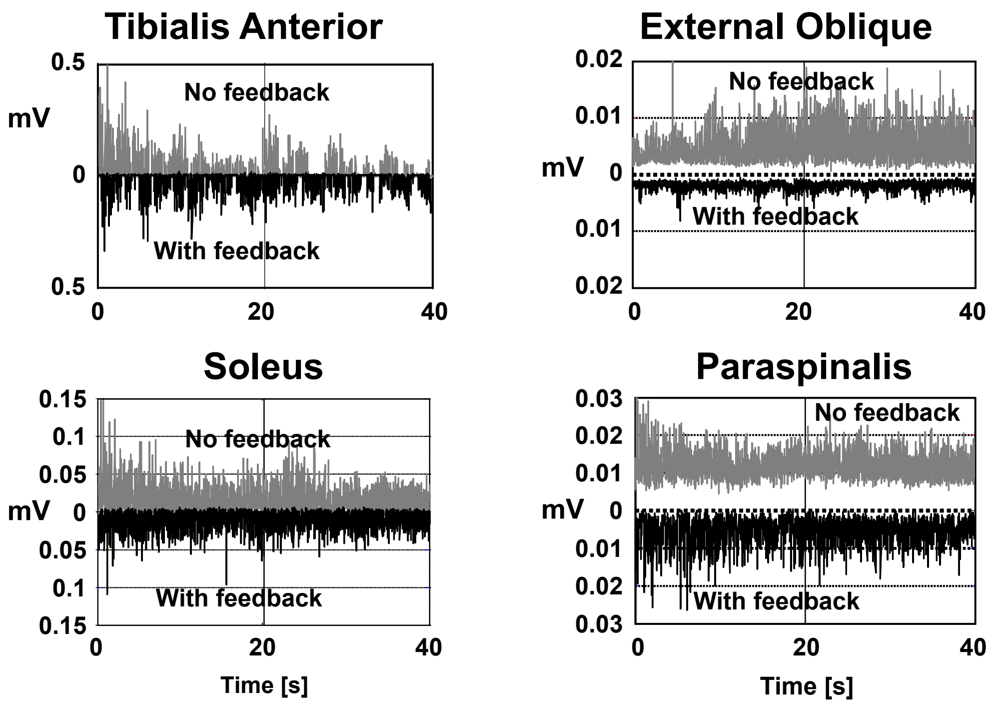

Fig. 5. EMG activity of a BVL subject with and without feedback for the task standing eyes closed on foam. The ankle and trunk muscle activity without feedback is shown in the upper traces of each panel, the activity from the same muscles with feedback is shown inverted in each panel.

DISCUSSION

The results of this study indicate that for sway frequencies up to 10 Hz (the high frequency limit of our movement analysis) combined vibro-tactile and auditory feedback provides improved control of balance for BVL subjects. The improvement was present in both the roll and pitch directions. Our previous research on the direction sensitivity of feedback or balance control during stance has indicated that improvements are generally larger in the pitch direction for healthy subjects but are dependent on the amount of sway (Davis et al., 2010). Therefore we assume that bilateral vestibular loss subjects benefit most by reducing their pitch sway about the pelvis as movements are larger in this direction. We have previously indicated that vestibular loss subjects tend to overreact at the trunk in the pitch and roll directions (Carpenter et al., 2001), and provide insufficient pitch control about the ankle joint (Allum and Honegger, 1998). Furthermore, differences in sensory processing times in these two directions may underlie the poor balance control in the roll direction (Allum et al., 2008). Despite changes in movement amplitudes with feedback, there was no change in balance-

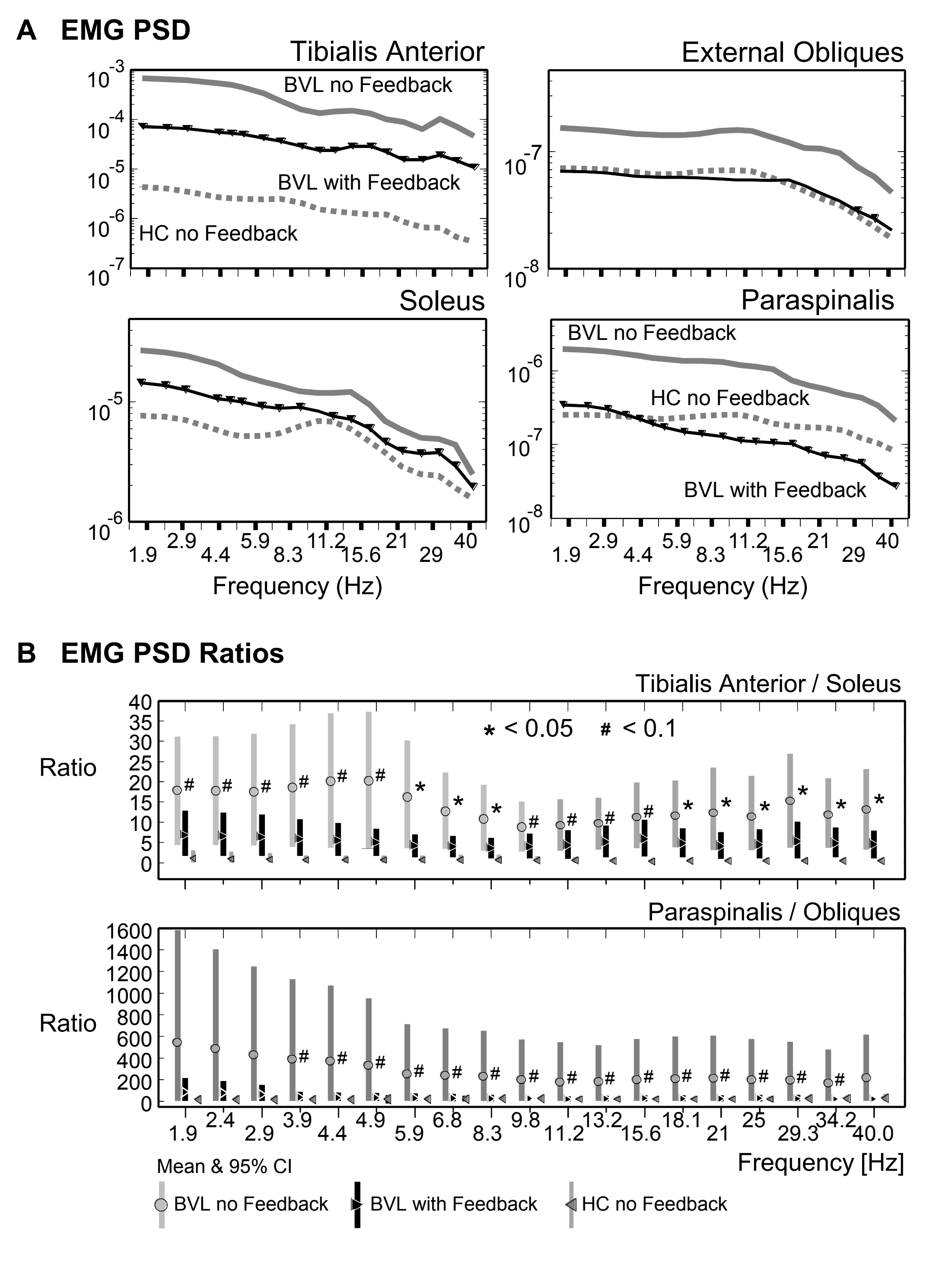

Fig. 6. A: Mean population PSDs of BVL EMG activity on and off feedback compared to mean activity of controls without feedback. The upper grey line is mean BVL activity without feedback, the lower black line has triangles indicating those frequency bins where the PSD values are significantly lower (p<0.05). The grey dotted lines indicate healthy control values without feedback. Only the control values for tibialis anterior were significantly different from BVL subjects with feedback. B: Mean population activation ratios of pairs of ankle (tibialis / soleus) and trunk (paraspinals / obliques) muscles compared for BVL subjects on and off feedback and to controls off feedback. The vertical bars mark the 95% confidence intervals. Asterisks (*) mark significant (p<0.05) differences of BVL subjects on and off feedback, while gate symbols (#) mark trends (p<0.1).

correcting movement strategies employed when feedback of pelvis sway was provided. Moreover, the same movement strategies were used on and off feedback as those used by normal subjects.

The amplitudes of the low frequency (0.7 Hz) both in phase and the high frequency (>3 Hz) anti-phase movement strategies were reduced with pelvis sway feedback by employing three types of muscle action: reduced depth of activity modulation, reduced background activity, and reduced muscle activation ratios. Presumably reduced background activity reduces intrinsic muscle stiffness and therefore induces less extensive sway for the same muscle modulation. However, this action was reinforced by smaller muscle modulation and lower antagonistic activation ratios. To our knowledge, this is the first study to show a direct neural correlation with the improvements in sway using prosthetic feedback. The reduction in EMG activity correlated with sway improvement raises the interesting question whether EMG feedback would provide a better feedback parameter than pelvis position. EMG feedback is widely used to control upper limb prostheses. However, force control is equally as effective in controlling prosthetic limb position (Corbett et al., 2011). Here we used pelvis position as the pelvis CoM is close to that of the whole body CoM and we assumed that this latter variable is controlled by the CNS. It may well be worthwhile considering in future studies a dual feedback mode (CoM position and EMG) for prosthetic feedback, particularly for gait tasks where excessive EMG activity may help predict gait instabilities in the next step.

It was interesting to note that the deltoid activity was also reduced with feedback. Whether this reduction is a direct result of reduced use of the arms to help balance or a reduced startle effect when balance is lost cannot be inferred from our results. The use of the arms to control a body sway when perturbed has, however, a limited effect on the CoM movement (Kung et al., 2010). The movement of the arms due to enhanced deltoid activity is both part of acoustic startle responses and first trial effects to support surface perturbations (Oude Nijhuis et al., 2010). Thus we would assume that BVL subjects became more confident with feedback training and therefore less startled when they realised they might fall.

The reductions in sway we observed were mostly in sway angles, but as sway angle became larger with the task of standing eyes closed on foam, velocity reductions occurred as well. These rapid reactions to the feedback may even have been reflex-like responses as it is hard to envisage the improvements we noted for movements with content greater than 3Hz (see Fig. 1d) and reduced EMG modulation up to 40 Hz (Fig. 6B) being solely due to voluntary reactions. This raises the question about the best way to code sway information in a balance prosthesis in order to produce the most effective feedback. Goodworth et al. (2011) indicated that vibro-tactile feedback provides only low frequency (<0.6 Hz) information on sway. It has been argued that this result occurred because feedback parameters were not fitted individually, rather a population «one-fits-al» approach was used (Loughlin et al., 2011). Our thresholds were determined individually. It is also possible that mainly position information is extracted by subjects from our vibro-tactile feedback signals and velocity information enhanced in higher frequency sway was obtained by the acoustic feedback which increased in volume when sway was larger. The system that Goodworth et al. (Goodworth et al., 2011) used uses 3 rows of vibrators activated by a combination of pelvis angle and angular velocity (Horak et al., 2009, Vichare et al., 2009). It remains to be investigated as to what extent the velocity feedback is used by patients using vibro-tactile feedback. We noted no difficulties for the patients with combining vibro-tactile and auditory feedback into motor commands. The visual feedback we provided was only active for large sway angles and not available under eyes closed conditions.

This study may have implications for other patient groups when these receive biofeedback to reduce abnormal body sway. Here we have emphasized that vestibular loss subjects use the same in- and anti-phase movement strategies with feedback as healthy controls and their movement strategies were similar to those of controls without feedback. We assume that the feedback substituted for the vestibular inputs that normally act to provide appropriate modulation of balance correcting strategies which are presumably triggered by proprioceptive inputs (Allum et al., 2008). Another patient group, those with Parkinson's disease, also showed improvement in roll sway with the identical type of feedback (Nanhoe-Mahabier et al., 2012). Whether the balance correcting strategies of these patients are similar to those of vestibular loss patients is not known. Lower-leg proprioceptive loss subjects use different balance correcting strategies (Horlings et al., 2009a). For such patients, it is an open question whether these patients need to change their balance correcting strategies to those of healthy controls before they can be aided by the feedback schemes of this and other studies (Dozza et al., 2005, Horak et al., 2009, Vichare et al., 2009) or whether the feedback characteristics need to be changed appropriately to fit their abnormal balance correcting strategies.

CONCLUSIONS

This study demonstrated how vestibular loss subjects achieve a reduction of sway during stance with prosthetic feedback. Unchanged movement strategies with reduced amplitudes are achieved with improved antagonistic muscle synergies at the lower legs and trunk. Thus both body movement and muscle measures could be explored as feedback variables for prosthetic devices which aim to reduce sway of those with a tendency to fall.

LIST OF ABBREVIATIONS

BVL: Bilateral vestibular loss; HC: Healthy control; CNS: Central nervous system; CoM: Center of mass; LP: Low pass; BP: Band pass; HP: High pass; EO: Eyes open; EC: Eyes closed; SCN: Standing eyes closed normal floor; SOF: Standing eyes open foam floor; SCF: Standing eyes closed foam floor; PSD: Power spectral density; TF: Transfer function; TFM: Transfer function magnitude

COMPETING INTERESTS

The authors report that J.H.J. Allum and F. Honegger worked as consultants for the company producing SwayStar™/ BalanceFreedom™, part of the equipment used in this study.

AUTHORS' CONTRIBUTIONS

FH developed the experimental instrumentation and software, designed the study, supervised experiments, analyzed the data, interpreted the data and help draft the manuscript. IMHA recruited and tested subjects and collected data, helped design the study, participated in analysis and wrote a preliminary draft. NGAE recruited and tested subjects and collected data, and participated in design. KST tested subjects and collected data, participated in design. JHJA designed the study and instrumentation, helped analyze and interpret the data and draft the manuscript.

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Chapter 5

Movement strategies in tandem stance: Differences between trained tightrope walkers and untrained subjects[1]

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ABSTRACT

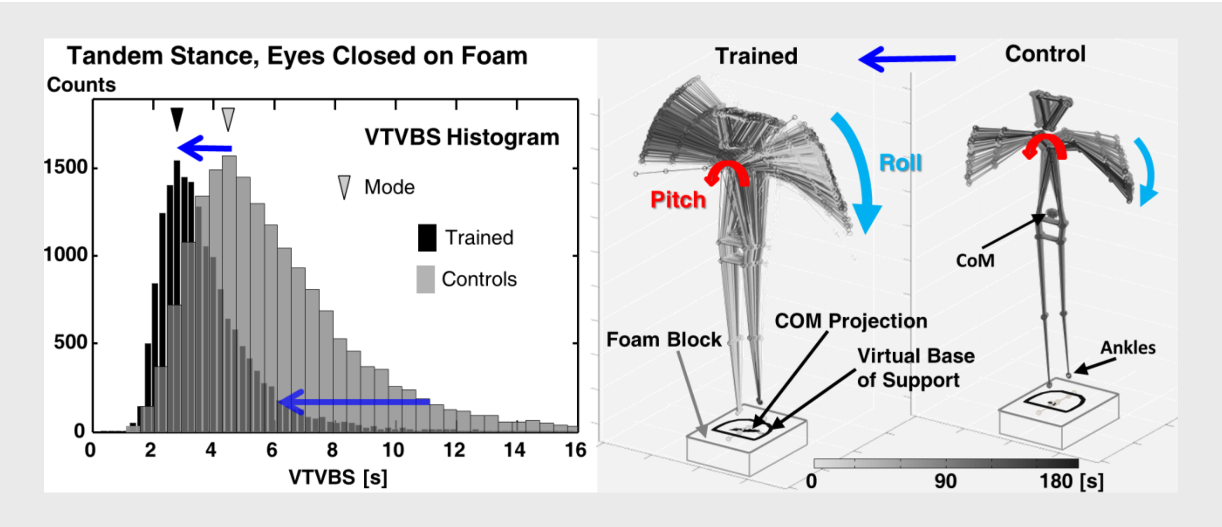
Does skill with a difficult balancing task such as tightrope walking lead to improved balance for similar but easier tasks through altered movement strategies or altered weighting of sensory inputs? We approached this question by comparing data collected during tandem stance (TS) of 7 tightrope walkers with those of 12 untrained control subjects.

All subjects performed 4 TS tasks under different sensory conditions: eyes open or closed, on a normal firm or foam surface (EON, ECN, EOF, ECF); tightrope walkers also on a tightrope (EOR). 90% ranges of head, upper trunk and pelvis angular velocities were measured in pitch and roll. Power spectral densities (PSDs), ratios, and transfer functions between these body segments were calculated. Centre of mass (CoM) excursions and its virtual time to contact a virtual base of support boundary (VTVBS) were estimated.

Gain nonlinearities in the form of decreased trunk to head and pelvis PSD ratios were present with increasing sensory task difficulty for both groups. These ratios were less in trained subjects, though, in absolute terms, trained subjects moved their head, trunk, pelvis and CoM faster than controls, and had decreased VTVBS. The exception, head roll, was unchanged with task or training. Surprisingly, CoM deviations were not less for trained subjects. For the trained subjects, EOR measures were similar to those of ECF.

Standing on a tightrope requires a velocity modification of a principally similar TS movement strategy to that of untrained controls. More time is spent exploring the limits of the base of support with an increased use of fast trunk movements to control balance. Our evidence indicates that this modified strategy results in an increased reliance on neck and pelvis proprioceptive inputs. The similarity of TS on foam to that on the tightrope indicates that the foam tasks could be used for effective training of tightrope walking.

GRAPHICAL ABSTRACT



Key words: tandem stance, sensory reweighting, movement strategies, centre of mass(CoM).

INTRODUCTION

Human upright posture is mechanically unstable during tandem stance. More so, tightrope walkers who are constantly training balance tandem stance on the rope, with a necessity to carefully control lateral sway. Presumably when adapting to these unstable postural conditions, roll sensory inputs are weighted differently than on a normal surface in order to control sway. Also, a different movement strategy could be used by tightrope walkers to control sway excursions on the rope to a minimum compared to a normal flat surface.

Sensory reweighting

Information about deviations from the upright position is accessible to the human brain through several sensory systems located on different body segments. The central nervous system (CNS) integrates these vestibular, visual, and somatosensory inputs, to maintain balance (Mergner et al., 2002, Peterka, 2002, Ravaioli et al., 2005, Goodworth and Peterka, 2009, Seemungal et al., 2009, Barra et al., 2010, Doumas and Krampe, 2010, Goodworth and Peterka, 2012, Sozzi et al., 2012). Depending on the balance task, the proportion of these sensory inputs may vary across body segments and with respect to centre of mass (CoM) motion (Black et al., 1983, Allum and Honegger, 1998, Peterka and Loughlin, 2004, Creath et al., 2005, Cenciarini and Peterka, 2006, Allum et al., 2008, Fetsch et al., 2009, Goodworth and Peterka, 2010, Tjernstrom et al., 2010, Goodworth and Peterka, 2012, Billot et al., 2013). For tightrope walkers, it is an open question whether they use different weightings of sensory inputs to take full advantage of sensory dynamic ranges for the task of tightrope stance and apply these weightings to tandem stance balance tasks on a normal surface.

Sway during stance can be investigated by examining changes in the composition of sensory feedback used to generate joint torques (Peterka, 2002, Maurer et al., 2006, Goodworth and Peterka, 2009). These authors applied small, 1 to 4 degrees of continuous support-surface perturbations to stance and used the resulting CoM or trunk responses to argue, with the support of modelling techniques, that amplitude response nonlinearities (less relative sway with increasing stimulus amplitudes) demonstrated sensory reweighting. This involved shifting reliance on ankle proprioception in favour of vestibular signals with increasing stimulus amplitude (Goodworth and Peterka, 2009, van der Kooij and Peterka, 2011). Thus for easier tasks, such as standing on a firm support surface eyes closed, with little pitch or roll motion of the trunk with respect to the pelvis, lumbosacral and neck proprioceptive gains can be set high, but not when trunk motion is larger; for example when standing on a foam surface, vestibular gains would be set higher. This action would, however, be disadvantageous with vestibular loss (Peterka et al., 2011). In general though, for roll it appeared that sensory weighting was less important for the upper body than for the lower body (Goodworth and Peterka, 2012).

Tandem stance movement strategies

In the past, most emphasis on the control of sway was on the anterior-posterior (AP) direction of two-legged stance with the feet side by side. Several movement strategies have been used to describe AP sway for this posture. The simplified single segment inverted pendulum - introduced for the interpretation of stabilograms (Gurfinkel, 1973) - involves movements around the ankle joint, with lower-leg proprioception providing the main contribution to the postural control (Horak and Nashner, 1986, Fitzpatrick et al., 1992a, Fitzpatrick et al., 1992b, Fitzpatrick and McCloskey, 1994). Proprioceptive inputs arising from around the knee, hip, and lumbosacral joints were initially thought to have little or no influence on balance control (Nashner et al., 1982). In their pioneering paper Koozekanani et al. (1983) suggested, however, that upright stance is controlled in pitch by multisegmental movement strategies. Further experimental and theoretical studies supported the multi-segment view and extended this concept to lateral sway (Day et al., 1993, Kuo, 1995, Hsu et al., 2007, Pinter et al., 2008) . In fact, in-phase and anti-phase movements between the lower (legs) and upper body (trunk) have been shown to co-exist simultaneously in the pitch plane, representing low and high frequency modes of body sway, respectively (Creath et al., 2005, Horlings et al., 2009b). Roll motion was shown to consist of two strategies of motion of the trunk, one with trunk and pelvis moving together and the other with high frequency trunk movements about a relatively stable pelvis (Horlings et al., 2009b). Thus a question arises if the same two types of control mechanisms are used in tandem stance where roll motion is more unstable than in the feet side-by-side position.

Although tandem stance (TS) is regularly used for clinical balance and scientific assessment and studies about balance performance (Nichols et al., 1995, Smithson et al., 1998, Lamoth et al., 2009, Seino et al., 2009), little is known about the movement control strategies during tandem stance. Winter et al. (1993) suggested that for normal, side-by-side, two-legged stance the mechanisms maintaining AP and medial-lateral (ML) stability consisted of independent ankle and hip strategies, with the ML direction dominated by a hip strategy, and the AP direction by an ankle strategy. For tandem stance, Winter et al. (1996) postulated that postural control was achieved with an ankle strategy for ML sway with little contribution from the hip strategy. In contrast the AP sway was dominated by a hip strategy with little contribution from the ankle strategy. However, several authors (Loram and Lakie, 2002, Morasso and Sanguineti, 2002) have questioned the assumptions used by Winter et al. (1996).

For tandem stance on a tightrope it is inherently clear that lateral movements of the rope provide only a limited constraint on support surface movement and balance must be maintained by other means than torques generated at the foot (Otten, 1999). The minimal model abstraction that fits balance control requirements is a two segmented inverted pendulum where the upper (trunk) segment is needed to generate torques to rotate the swaying body back to upright (Paoletti and Mahadevan, 2012). Similarly for tandem stance on a foam support surface, lower-leg somatosensory inputs are less reliable and the possibility of using these inputs efficiently to generate lateral torque is strongly degraded. For this reason one would expect a comparable movement strategy for this condition as standing on a tightrope. However, in contrast to normal stance the base of support in TS is enlarged in the AP direction compared to stance with the feet side-by-side resulting in an enhanced AP stability. In principle, the torque generated by pitching forward or backward can be used to temporarily stabilize roll movements via a coriolis gyroscopic effect. In summary, tandem stance on foam might require a similar movement strategy as on a tightrope but would prove to be a safer means to compare tightrope walkers with controls.

Head movements

Few studies have recorded head movements during either normal unperturbed two-legged stance (Honegger et al., 2012a, Honegger et al., 2012b) and to our knowledge none during tandem stance. Head movements are of interest because the vestibular system tracks rotations and translations of the head with respect to an earth-fixed reference system whenever the head is moved, be it with respect to the trunk or in tune with ankle movements. Based on these two types of head movements, two theories about sensory control of upright stance have been proposed: top-down and the bottom-up control (Nashner et al., 1982, Maurer et al., 2000). The bottom-up theory, states that upright stance is controlled based on ankle proprioceptive inputs which are confirmed by vestibular inputs as it assumed that the body moves as an inverted pendulum. That is, the head is assumed to be locked to the trunk. When upright stance is controlled top-down; head movements are regulated to be fixed in space, using a combination of vestibular and visual inputs as an earth fixed reference, and the proprioceptive inputs from the neck, hips and ankle joint are used to control body sway. Keshner and Peterson (1995) suggested that the mechanisms underlying head and neck stabilization also included biomechanical components (particularly head inertia), voluntary control, as well as vestibular and proprioceptive neck reflexes. Thus the mechanisms which stabilize the head are related to both the frequency and stimulation direction (roll or pitch) of head motion relative to the trunk (Keshner et al., 1995). Generally through, pitch stabilization of the head in space (top down mode) rather than a head-fixed-on trunk mode appears to be the underlying movement strategy when the support surface is moved (Buchanan and Horak, 1999, Corna et al., 1999, Akram et al., 2008). Head movements during stance are changed after vestibular loss (Honegger et al., 2012a). Not only are these larger due to the overall instability, but head motion is anchored more firmly to the trunk leading to a shift towards a higher frequency resonance (Honegger et al., 2012a). Thus it is an open question whether tightrope walkers also lock their head more to the trunk and shift the head resonance, relying more on neck proprioceptive inputs and decreased vestibular inputs to control head motion.

Training effects of tandem stance

Lamoth et al. (2009) investigated if the level of athletic skill is reflected in the control of body sway. The subjects stood in two legged tandem stance on a narrow plywood strip. They found that as the level of gymnastic skill increased, trunk acceleration variability decreased and stability increased, indicating a more efficient postural control. They found that, in contrast to sensory reweighting concepts described above, differences in skill did not depend such reweighting, suggesting instead that expert gymnasts exhibited differences in the underlying postural control strategies which appear to be independent of the specific weighting of sensory information.

Given the aforementioned questions raised on tandem stance strategies, particularly with reference to head and lumbosacral movements, we investigated how these different body segments move while standing in tandem stance under different sensory conditions. We hoped to answer the following questions: Are movement strategies at the neck and lumbosacral joints changed for tandem stance under different sensory conditions? Are there differences in movement strategies or sensory reweighting of trained tightrope walkers in comparison to healthy untrained subjects? Finally, does standing on a tightrope require a different movement strategy compared to standing on a foam support-surface? If we could demonstrate similarities in strategies for difficult tandem stance conditions and standing on a tightrope, this could demonstrate the applicability of a safer situation for those wishing to practise performing tightrope walking.

METHODS

Subjects

In cooperation with Circus Monti (Switzerland), seven trained tightrope walkers were recruited: five females and two males, aged 22.4 ± 8.4 years (mean ± SD) with BMI 20.7 ± 1.8 (mean ± SD). Their experience with the tightrope ranged from 4 to 15 years with 1.5 to 13.5 hours of training per week. Twelve untrained healthy subjects matched for sex (nine females, three males), age (21.9 ± 5.24 years) and BMI (20.5±1.6), provided a comparison population. Exclusion criteria for both groups were: balance problems, neurological diseases, orthopaedic disorders, rheumatologic disorders, operations to the lower limbs and back, hearing problems, recent fractures, severe visual impairment, medication or drugs use and pain at the time of measurement. Practice with tight- or slack-rope walking was an exclusion criterion for the untrained group. All subjects signed an informed consent form according to the Declaration of Helsinki. The study was approved by the Institutional Ethical Review Board of the University Hospital Basel.

Procedure

For the current protocol, tandem stance was defined as standing with one foot directly in front of the other, touching heel-to-toe. All the subjects performed four tandem stance tasks: eyes open, or eyes closed, on a normal firm and on a foam surface (EON, ECN, EOF, ECF). The trained subjects also performed a TS task on the tightrope eyes open. The order of support surface (firm or foam) was randomized, to restrict influence of a training effect, except that the tightrope tasks were always performed last. For the foam task the subjects stood in the middle of the block of foam, which had a height, width and length of 10 by 44 by 400 cm, and a density of 25 kg/m^3. The tightrope was a 7 meters long, consists of twisted steel rope, with a diameter of 12 mm, and was strung 70 cm above the ground. The trained subjects needed to wear ballet shoes on the tightrope. The same make of ballet shoes was used by all subjects. They wore these shoes for all tasks. The untrained subjects stood barefoot. The subjects were not allowed to look at their feet, performing the eyes open tasks while fixating a point 7 m away. The arms were free to move during all tasks except eyes open (EON) on a firm surface. For this task the arms were held hanging loosely at the side of the subject's body, so comparisons could be made with the results of other studies. Subjects chose which foot was in front of the other during tandem stance. One spotter stood slightly behind the subject in order to assist in case off a near fall. All tasks had a duration of 180 seconds or until balance control was lost (i.e. the subject required assistance from the assistant). In this case, the task was repeated a maximum of three times and the longest of 3 attempts was used for analysis.

Measurement systems

Head angular rates in the roll and pitch directions were measured with a micro-mechanical gyro-enhanced attitude and heading reference sensor system (MTi Xsens Technologies, Enschede, Netherlands) attached to an adjustable tight-fitting headband (total weight 150 grams). Because the noise of the Xsens system approached the spectral content of head signals at 10 Hz (see Fig. 3A), our analysis was limited to signals less than 10 Hz.

Body sway at the trunk and pelvis in the roll and pitch directions was measured with two identical fibre-optic based gyroscope systems of weight 500 grams (SwayStar™, Balance International Innovations GmbH, Switzerland). The pelvis SwayStar™ unit was mounted on a converted motor-cycle kidney-belt strapped around the hips at the level of the trochanter, the other, trunk unit was mounted between the scapulae on a tightly fitting shoulder harness. Both systems were firmly but comfortable strapped to the body.

The SwayStar systems measured angular velocities in the roll (frontal) and in the pitch (saggital) plane with a 16 bit accuracy over a range of ±327 deg/s. The Xsens system has a slightly lower range of ±300 deg/s and its noise spectrum is a factor 1000 higher. The noise spectrum of the MTI Xsens systems compared to measured head data is depicted in Fig. 3A. The noise spectrum of SwayStar™ is depicted in Fig. 1 of Honegger et al. (2012b). The two computers used for recording the angular velocity signals from the three sensor systems were synchronised to one another using digital trigger signals indicating the start of recordings emitted by each SwayStar™ system and by one computer. The thin synchronisation cables connecting the measurement systems to the computers were swung over the shoulder of the spotter, in order to prevent the cables giving feedback to the subject and to prevent interference with the subject's mobility. All data was sampled at a rate of 100Hz.

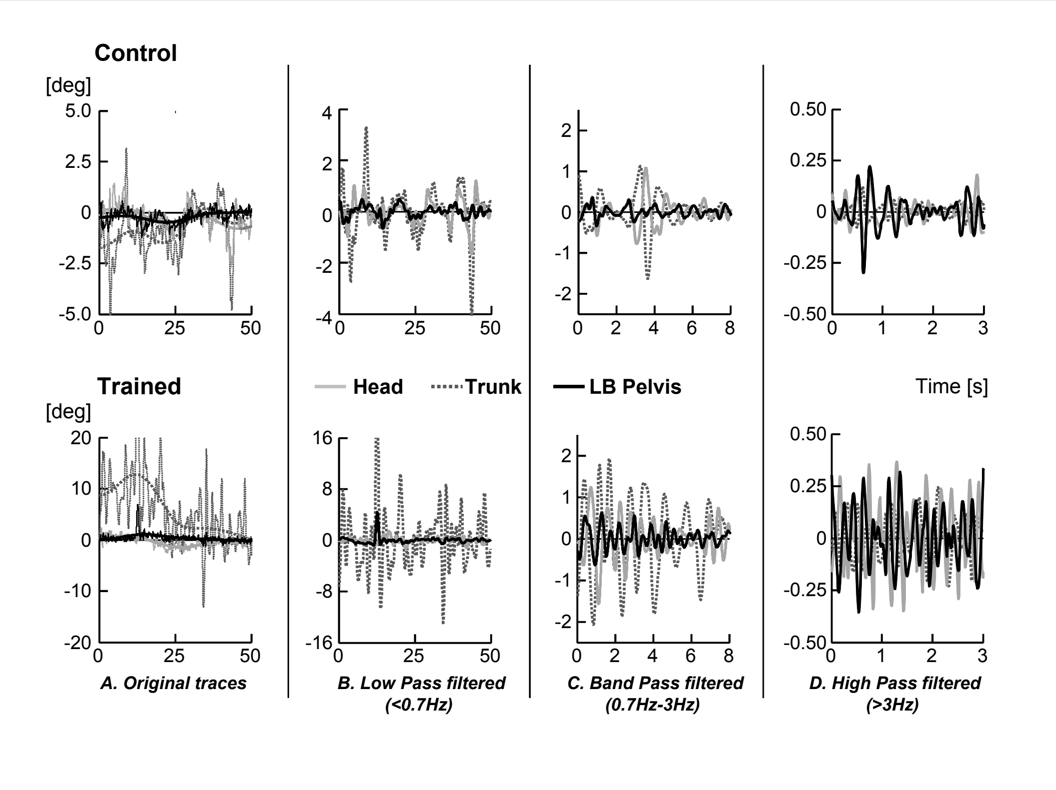


Fig. 1. Typical traces of pelvis, trunk and head movements
Angular displacement during standing on a foam support with eyes closed. Roll data is shown for a typical untrained control subject (upper traces) and a typical trained subject (lower traces). Fig. 1A shows 50 seconds of the original traces of head (grey), shoulder (dark grey dotted) and LB-pelvis (black). A trend line of each original trace is drawn in A, which is removed from the data of Fig. 1B-D. Fig. 1B shows the same range after a low-pass filtering (<0.7 Hz). Fig. 1C shows 8 seconds of the same traces after band-pass filtering (0.7<3 Hz) and Fig. 1D 3 seconds after high-pass filtering (>3Hz). The smaller time intervals were selected for visualization purposes only. Note the larger movements of the trunk for the trained subjects but relatively smaller movements of the head and pelvis. The scales of the ordinates on figures 1A and 1B are four times larger for the trained then for the control subject.

Data Processing

All calculations necessary to prepare and extract data measures were performed with MATLAB and the accompanying Signal Processing Toolbox, both R2010a (The MathWorks, Inc. Natick, MA, USA). Calculation of CoM were performed with the help of the SpacLib V2.2 Matlab library (Legnani, 2009) and the soder function (Jacobs, 1997) found on the website of International Society of Biomechanics (ISB). For the de-trending as described below we relied on the denoise function of the Rice Wavelet Toolbox (Baraniuk et al., 2002). Statistical analysis was then performed with IBM SPSS (Version 20) except for the circular statistic analysis were we used the 2010b update of the ''Circular Statistic Toolbox'' for MATLAB published by P. Berens in (2009).

Frequency Domain

Estimates of the power spectral densities (PSD) $P_{x,x}(f)$, cross power densities $P_{x,y}(f)$, PSD ratios $R_{x,y}(f) = \frac{P_{x,x}(f)}{P_{y,y}(f)}$ and transfer functions $T_{x,y}(f) = \frac{P_{y,x}(f)}{P_{x,x}(f)}$, between pelvis and trunk, and head and trunk were calculated to determine relative segmental motion. The estimates were based on Welch´s averaging method, employing a window size of 2048 samples with an overlap of 1024 samples. Group averaged data was smoothed with a Tukey's running median smoothing (Tukey, 1977) known as «(3RSR)2H twice» for display purpose only (Fig. 3). For further statistical processing the data were averaged over 10 frequency bins each of which combined values of 10 frequency samples, except for the two lowest frequency bins. Bins were equally spaced on a logarithmic scale of the range of 0.00–10.25 Hz. The frequencies used were 0.15 Hz (mean of seven samples from 0 to 0.29 Hz), 0.2 Hz (mean of nine samples from 0 to 0.39 Hz), 0.42 Hz (0.2–0.63 Hz), 0.71 Hz (0.49–0.93 Hz), 1.39 Hz (1.17–1.61 Hz), 2.32 Hz (2.10–2.54 Hz), 3.69 Hz (3.47–3.91 Hz), 6.13 Hz (5.91–6.35 Hz), 7.59 Hz (7.37– 7.81 Hz) and 10.03 Hz (9.81–10.25 Hz). Peak frequency shifts for the resonance structure seen in the head to trunk PSD ratios were investigated using the procedure reported in Honegger et al. (2012a). For each subject and test the centroid frequency $f_{c,test,subject}$ over the range from 1 to 7.6 Hz was calculated as a weighted average of the PSD ratios or transfer function magnitudes (TFM): $f_{c,test,subject} = \frac{\sum_{1Hz<f(i)<7.6Hz} w_{test,subject}(f(i)) \cdot f(i)}{\sum_{1Hz<f(i)<7.6Hz} w_{test,subject}(f(i))}$ where $w_{test,subject}$ are the PSD ratio values.

Time Domain

Ranges of Angular Data

Raw velocity data and its integral (angular position) were used to calculate 90% angle and angular velocity ranges (Fig. 5). Angle and angular velocity 90% ranges within specific frequency bands were also calculated as described below.

Centre of Mass and Ranges

CoM movements were estimated using a 12-body segment (see Fig. 2) adaptation of the 14-body segment model of Winter et al. (2003) used in previous studies (Visser et al., 2008, Oude Nijhuis et al., 2009, Kung et al., 2010). From marker data of Oude Nijhuis et al. (2009) an initial template was built, consisting of an average of four subjects with the same age and BMI as subjects in the current study (mean ± SD for age 24.0±2.1 years and for BMI 21.57±2.28 Kg/m^2). This source template for a subject standing feet side by side was transformed into a template standing in TS. The arms of the original template were in normal position hanging loosely at the side of the body. For all conditions but EON the arms were then laterally rotated up to 10° above the horizontal plane for trained subjects and 10° below this plane for the untrained. These positions were based on pilot study observations (SRF, 2010). For the EON condition, the arms were laterally lifted for both groups 10° away from the body. The template was adjusted to body height of each subject of the current study by means of an affine projection. Each individual template was then kinematically propagated by driving it with the measured angular velocity data. For this purpose, virtual markers where clustered in three rigid segments $\vec{x}_{s,k}(t_0)$: legs and pelvis combined, trunk and arms combined, and head. For the propagation we defined the following connecting anchor points: ground projection of mid-ankles, mid-assis, mid-shoulder. With the measured segments, these built a linked chain that was numerically integrated by the following procedure. Starting at the bottom for each segment the virtual markers $\vec{x}_{s,k}(t_i)$, base anchor $\underline{\vec{x}}_{s,f}(t_i)$ were rotated in correspondence of the pitch and roll angle displacement increments $V_s(t_i) \cdot \Delta t$ relative to a segment fixed coordinate system (rotation in yaw was neglected). After rotation, the segment was translated to shift back the anchor point $\underline{\vec{x}}_{s,f}(t_{i+1})$ to the position in the lower segment or the base $\vec{x}_{s-1,f}(t_{i+1})$. The algorithm can be summarized by the following update formulas:

$$\vec{x}_{s,k}(t_0) = \bar{R}_{s0}(t_i) \cdot \vec{x}_{s,k}(t_i) - \bar{T}_{s0}(t_i), \qquad \underline{\vec{x}}_{s,f}(t_{i+1}) = \bar{R}_{s0}^{-1}(t_i) \cdot V_s(t_i) \cdot \vec{x}_{s,f}(t_0) \cdot \Delta t,$$
$$\vec{x}_{s,k}(t_{i+1}) = \bar{R}_{s0}^{-1}(t_i) \cdot V_s(t_i) \cdot \vec{x}_{s,k}(t_0) \cdot \Delta t - \left(\underline{\vec{x}}_{s,f}(t_{i+1}) - \vec{x}_{s-1,f}(t_{i+1})\right), t_i = i \cdot \Delta t \ .$$

The calculation of the roto-translation $(\bar{R}_{s0}(t_i),\ \bar{T}_{s0}(t_i))$ for each segment s that relates the virtual marker positions $\vec{x}_{s,k}(t_i)$ at a given time t_i with the corresponding in the initial template $\vec{x}_{s,k}(t_0)$ is the key step of this algorithm, necessary because the reference system of the sensors is body fixed (Application 1 in Chiallis (1995)).

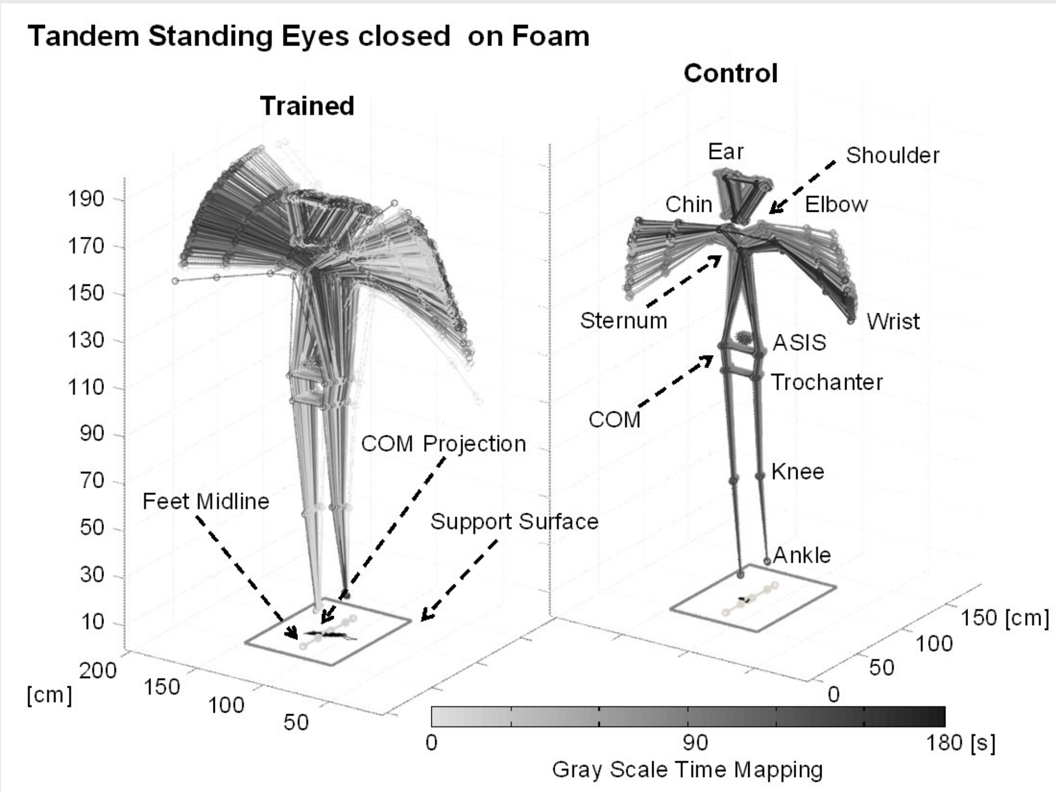


Fig. 2. Stick figure model
Stick figure visualization of the movements of a trained tightrope walker and a control subject during the standing eyes closed on foam task. Plotted are 5 frames per second over the extent of a 180 s task. Note the higher amplitudes of body segment movements of the trained subject.

Virtual Time to Virtual Boundary of Stability (VTVBS), Medians and Ranges

To have an additional spatiotemporal measure for the proximity of the CoM to the boundary of the base of support, the concept of virtual time to collision was adapted to CoM as described in the work of Slobounov et al. (1997) for centre of pressure (CoP). In contrast to Slobounov et al., who calculated the postural stability boundaries for CoP individually for each subject, given subjects' foot geometry, a common virtual boundary of stability (VBS) based on the CoM estimation outlined in the previous section was used. The VBS was defined as follows: first, the total union (thick grey dashed line in Fig. 7D) of all individual convex hulls for the 90% lateral and anterior-posterior ranges of CoM within the x-y-plane was computed for all tasks and subjects (fine black dotted and dashed lines in Fig. 7D are the control and trained subject convex hulls, respectively for the ECF task); second, the lateral mirror image was created of the total union to yield a VBS convex hull (thick black line in Fig. 7D). Given this VBS, VTVBS values for every CoM sample was defined as the time the CoM would need to propagate from its current postion (CoM_x, CoM_y) within the VBS starting with its current velocity ($vCoM_x$, $vCoM_y$) and maintaining its current acceleration ($aCoM_x$, $aCoM_y$) constant until reaching the VBS defined above. Haibach et al. (2007) summarizes all aspects of this calculation in a concise manner. The principle of the calculation is to determine the possible intersection times of the CoM path with the straight line segments of the convex hull (in our case 23) that make up the VBS and take the shortest among these as the VTVBS. VTVBS was then summarized by calculating medians and 90% ranges (Fig. 7B and C). Points inside the VBS can be defined as points that can be controlled by the current movement strategy, whereas points outside correspond to falls or near falls unless a restoration of stability with a different strategy such as jumping occurs. Because we took 90% ranges of CoM to calculate the VBS we excluded from our VTVBS calculation any CoM values located outside the VBS. Exclusions occurred in one EON, in two ECN and ten ECF – 13 out of 76 – recordings. The percentiles of points omitted per recording ranged from 0.04% to 8.56% with an average of 2.34%.

Filtered Time Domain

Phase Estimates between Segments

To estimate the phase relationships between segments the data was processed in three steps: integration of measured velocity data to yield angle data, and then detrending and filtering the angle data. The low frequency trend was calculated using the denoise function of the Rice Wavelet Toolbox as an adaptive low-pass (LP) filter by applying at 0.05 Hz. The filtered output was then subtracted from the angular data to yield de-trended angular positions that were then partitioned into low, mid, and high frequency components. As in our previous study (Honegger et al., 2012b) sway frequency ranges were separated based on the shape of the average PSD of the angular velocity for the pelvis (Horlings et al., 2009a). That is cut-off frequencies, 0.7Hz and 3.0Hz (Fig. 1) were set at frequencies just higher or lower, respectively, of two resonances in pelvis data (Horlings et al., 2009a). Similar resonances can be observed in the pelvis and trunk data of Fig. 3. The filtering was accomplished with 3rd order Butterworth filter running forwards and backwards through the data to achieve zero phase shift. For each frequency band we performed regressions on the angle position data between the head and trunk, and pelvis and trunk, using total least-squares regression procedures (Honegger et al., 2012b). The regressions yielded the regression slope angle and were a measure of phase difference.

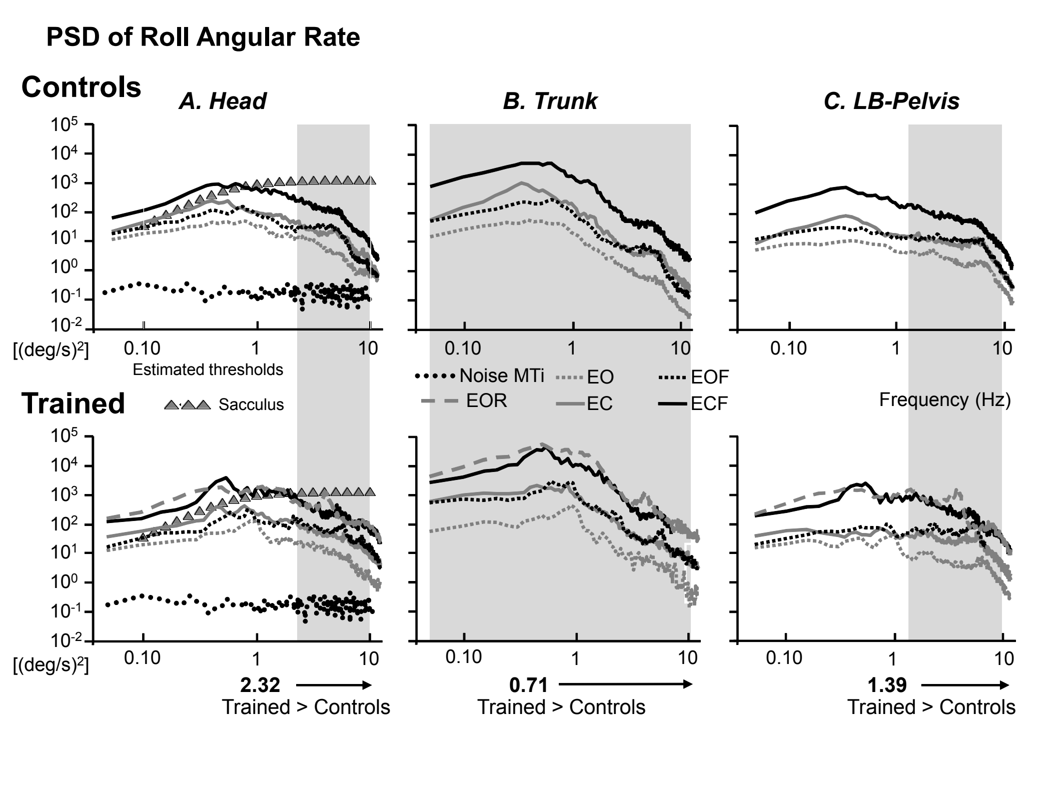


Fig. 3. Mean power spectral densities in roll plane
Mean power spectral densities (PSD) in the roll plane of the untrained (top) and trained group (bottom) for the three different measurement positions: (A) head, (B) shoulders, (C) LB-pelvis across the different tasks. The noise spectrum of the Xsens MTi sensor is indicated in the head PSD plots. The noise spectrum of the SwayStar systems was flat but below the ordinate scale at 10^{-4}. The head data also contain the estimated thresholds of the sacculus (triangles) based on the data of Kingma (2005). Thresholds of the canals (Seemungal et al., 2004) and utriculus (Kingma, 2005) are at the level of the Xsens noise spectra. The grey shaded rectangles indicates the range of frequencies for which a group differences were found and the arrows at the bottom the range where significant post hoc comparisons occur. Notice that the PSDs of the trained subjects are larger than the PSDs of the control subjects.

Statistical analysis

Group and task effects were analysed with generalized linear model based on type III sums of squares taking subject group as the between and task as the within factor (ANOVA). In the majority of cases when Mauchly's test of sphericity was significant, degrees of freedom were corrected using Greenhouse–Geisser estimates of sphericity. Epsilon was always below 0.75 so a switch to the Huynd-Feldt corrections was not necessary. Post-hoc analysis of significant condition effects were performed using paired t-tests. For the comparison of group effects we switched to the nonparametric Mann-Whitney U test. As it compares the sums of ranks, the Mann–Whitney U test is less likely than the t-test to spuriously indicate significance because of the presence of outliers. For the same reason, comparisons between pitch and roll directions were calculated with the Wilcoxon Signed Ranks test (WSR). For the regression slopes (axial data) the directedness was verified with both the Rayleigh test and Rao's spacing test. We used the Watson–Williams test as the circular analogue to the two-sample t-test or the one-factor ANOVA, and the Harrison–Kanji test as the circular two-factor ANOVA (Harrison and GK, 1988, Berens, 2009). In all cases, if not otherwise stated, the significance level was set at 0.05.

RESULTS

All subjects were able to perform the TS tasks EON, ECN and EOF for the whole duration of the trials. Three control and two trained subjects were not able to perform the ECF task and the same two trained subjects the EOR task for the full 3 minutes. When permitted (all tasks except EON) the trained held their arms about 10° above the horizontal, whereas the controls held the arms about 20° lower. Fig. 1 provides example traces of typical subjects performing the ECF task and illustrates our common observation that trained subjects had much greater movement ranges than control subjects particularly for the upper trunk. This effect is particularly clear in the stick figures of Fig. 2 based on our CoM modelling techniques.

Task Effects in general and with respect to the TS EOR task

The PSDs of Fig. 3 and the amplitude plots of Fig. 5 show clear task effects across all 3 segments measured. For all PSD and amplitude variables we analysed, ANOVAs indicated significant task effects ($p<0.01$ in 70% of the analyses, for the remainder $p<0.05$). Post-hoc analysis shows that all ECF values were significantly larger than the values of the EON, ECN, and EOF tasks, especially for the frequency bins in all PSDs of Fig. 3 above 0.42Hz and for the 90% ranges in roll and pitch of Fig. 5. The same holds for the lateral and AP 90% ranges of CoM position, velocity and acceleration shown in Fig. 8.

Post-hoc analysis revealed that if for a given variable there is a significant difference (see, for example, Fig. 5 for the 90% ranges) between the values of two tests, for example between EON and ECN or between EOF and ECF, for the trained group then this observation was paralleled for the control group. Because of this observation and the fact that intersegment phase patterns between LB-pelvis and trunk were the same for both groups - in phase for frequencies below 0.7 and anti-phase for frequency above 3 Hz – and were conserved throughout all tasks, we focused our report here mainly on group differences.

To define differences with respect to the TS EOR task of the trained subjects we inspected the outcome of the post-hoc analysis of repeated measurements ANOVAs ran separately for the trained subjects. Whenever the ANOVA revealed a significant task effect, the EOR values differed significantly from the ECN, EON and EOF ($p<0.05$) values but not from the ECF values ($p \approx 1$).

Frequency Domain

Power Spectral Densities (PSD)

Fig. 3 shows mean roll velocity PSDs of the controls (top) and trained group (bottom) for the three different recording positions: (A) head, (B) trunk and (C) pelvis. For both directions roll and pitch (not shown) ANOVA analysis indicated a significant task effect over all frequency bins. A significant group effect was also found across a large frequency range, especially for the trunk as shown by the grey shaded rectangles in Fig. 3. The arrows along the bottom of Fig. 3 indicate for which frequencies up to 10Hz magnitudes in post-hoc tests were significantly greater for the trained subjects for all tasks except EON. Noticeably, for the trunk, this range (0.71 to 10Hz) was greatest. The same pattern was repeated in pitch PSDs with the exception that the group effect in trunk pitch PSD and post-hoc comparisons were only significant for frequencies above 1.39 Hz.

A MANOVA analysis indicated that in roll there was a trend for trunk PSD to be larger than the head and pelvis PSDs, and the head PSD larger than the pelvis PSD. The pattern was slightly different in pitch. In pitch for the EON, ECN and EOF tasks trunk and head values were of comparable size but both larger than pelvis PSD. For the ECF task in pitch pelvis and head PSD were of similar size but significantly smaller than the trunk PSD.

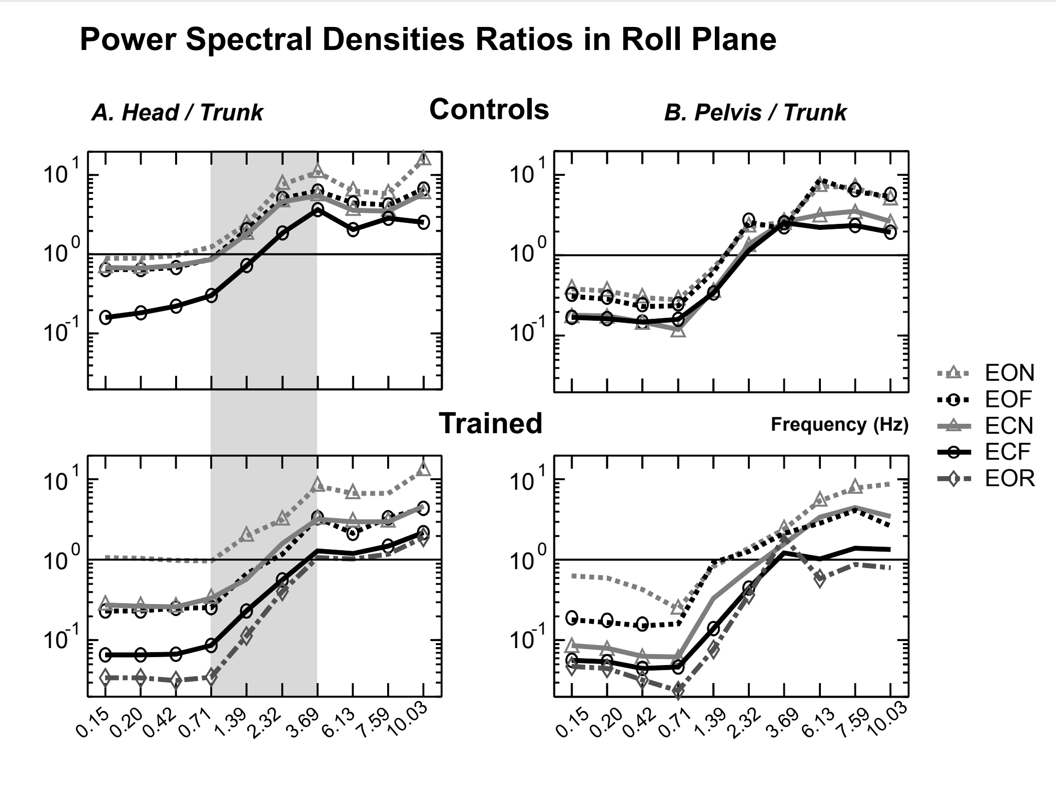


Fig. 4. Mean PSD ratios in roll plane

Mean PSD Ratios of the control (top) and trained group (bottom) in the roll plane: (A) head with respect to the trunk (that is, head/trunk) and (B) pelvis/trunk. The PSD ratio is shown for all conditions across the ten frequency bins. A symbol on the line at the frequency bin indicates that the ratio is significantly different from 1 ($p < 0.05$) at that frequency bin and for the task condition. The horizontal arrow below the plots indicates frequencies for which a group difference was found.

Roll Head/Trunk Ratios

For each pair of recording sights we first examined ratios of PSDs to determine if segments moved as a unit (ratio of PSD equal to unity). As Fig. 4A shows across low frequencies (controls ≤0.71 Hz and trained ≤ 0.42 Hz) the head/trunk PSD ratios were below one (as indicated by the symbols on the lines in figure 4), indicating that the head moved less in relation to the trunk, but across high frequencies (≥3.69 Hz) the ratio was above one, indicating the opposite effect, but in both frequency bands ratios decreased with task difficulty, that is less head movement with increased trunk movement. However, no statistical difference was found between the ECF and EOR conditions in the trained group. There was also a general group effect – region indicated by grey rectangle in Fig. 4 -

for frequencies between 0.71 and 3.69 Hz the head/trunk PSD ratio was significantly less for the trained subjects than for the controls.

Across low frequencies (untrained: ≤ 0.71 Hz: trained: ≤2.32 Hz) the head/trunk transfer function (TF) was less than unity. The TFs of the trained group were smaller than the TFs of the control group at frequencies below 7.59. Furthermore, a general trial condition effect was found with TF gain decreasing with task difficulty. TFs in both groups were different across trial conditions, smaller for the trained subjects. However, TFs for EOR and ECF tasks were not significantly different from each other for the trained subjects.

Pitch Head/Trunk Ratios

The head/trunk PSD ratio was above unity in all frequencies for the control group (data not illustrated). For the trained group the ratio differed from unity only in few frequency bins. No group effect was found. A trend in a task effect was only seen in head/trunk PSD ratios for frequencies below 2.32 Hz. ECF and EOR do not statistically differ. Similar outcomes were seen for the pitch TF.

Head Resonant Frequencies

The ANOVA for the centroid peak frequency showed a significant group effect but no task dependency. Analysis for the pooled data resulted in a mean centroid frequencies for the untrained controls of 4.15 Hz in roll and 3.5Hz in pitch, and for the trained 4.5Hz in roll and 3.8Hz in pitch. The Mann-Whitney U on the differences between trained and controls resonant frequencies revealed a significance level of $p<0.001$.

Roll Pelvis/Trunk Ratios

Similar to the head/trunk ratios, pelvis/trunk PSD ratios were in the low frequencies (untrained ≤1.39 Hz and trained ≤ 2.32 Hz) below unity (see Fig. 4B). In the high frequencies (≥3.69Hz) the control group had pelvis/trunk PSD ratios above one, but the trained group had ratios in the high frequencies (> 2.32 Hz) that did not significantly differ from unity. A group effect was found for frequencies below 2.32 Hz with the pelvis/trunk PSD ratio being smaller for the trained group. In addition a trend in trial condition across frequencies below 3.69 Hz was found with smaller ratios for the most difficult tasks. The ratios for ECF and EOR tasks in the trained group do not differ. The pelvis/trunk TF magnitude was below unity across low frequencies (controls: ≤ 0.71 Hz: trained: ≤ 1.39 Hz). A group effect was found for frequencies below 2.32 with the trained group having a smaller TF in comparison to the control group. Furthermore, a trend was found with the most difficult tasks having lower TF magnitudes. However there was no significant difference between ECF and EOR conditions.

Pitch Pelvis-to-Trunk Gain

The pelvis/trunk PSD ratio was below unity across low frequencies (≤ 3.69 Hz for the untrained group and ≤ 2.32 Hz for trained group). However in high frequencies the PSD ratio did not significantly differ from one. Neither a group nor a condition effect was present. The outcome for the TF was similar but in addition the TF magnitude did not significantly differ from unity.

Time Domain

90% Ranges of Angles and Angular Rates

Group differences were found in trunk roll angle for the most difficult tasks (EOF, ECF) and a trend for ECN (Fig.5A). As we suspected that in-and-out-of-phase modes of motion would mask group difference in trunk roll velocities, we analysed data within specific frequency bands. The amplitude analysis for roll in different frequency components shown in Fig. 6 depicts the results for the ECF task and indicates that such masking indeed occurred. Significant group differences were found for all tasks (except EON) in the mid frequency band (0.7-3Hz) for both roll angle and angular rate of the pelvis and trunk. Differences were observed in the high frequency band (>3Hz) for head, trunk and pelvis. For the ECF task, significant group trunk differences were present in the low frequency angle trend and in the low frequency band (<0.7Hz) for both angle and angular rate.

Significant group differences (trained > controls) were also found in pitch for both angle and angular rate (Fig. 5B) except for trunk and pelvis angle in the EON condition. Pitch 90% ranges were significantly greater than the roll 90% ranges, for angles and angular rates at the pelvis, trunk and head, for trained and control subjects, with two exceptions – head angular rate for the controls in the EON and ECF tasks (Fig. 5).

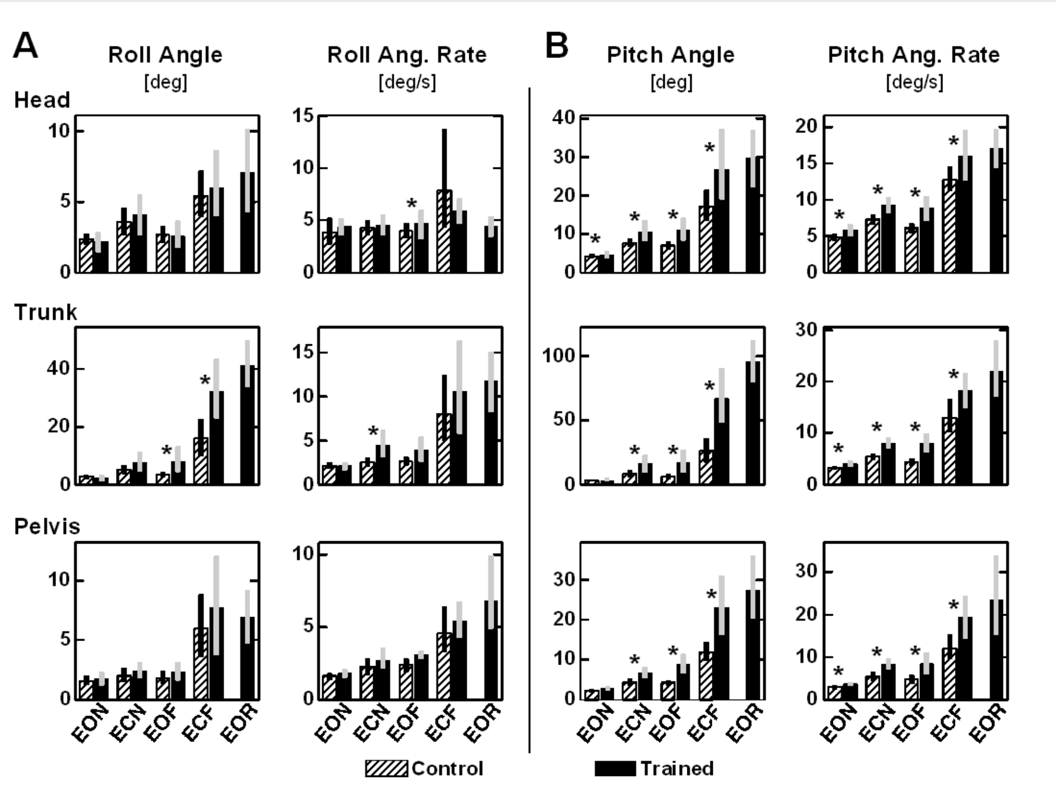


Fig. 5. 90% ranges of angle and angular rate in roll and pitch
Mean 90% roll sway angle and angular rate range comparisons between trained and control subjects for the four different tasks performed. The column height show the mean 90% ranges. The vertical bars indicate the 95% confidence intervals of these means. For both roll on the left and pitch on the right the ranges are displayed for head, trunk and pelvis. The black columns show data of trained and the hatched data of control subjects. An asterisk (*) indicates that the values of the trained subjects are either significantly greater at a significance level at 5% (p<=0.05 Mann-Whitney exact).

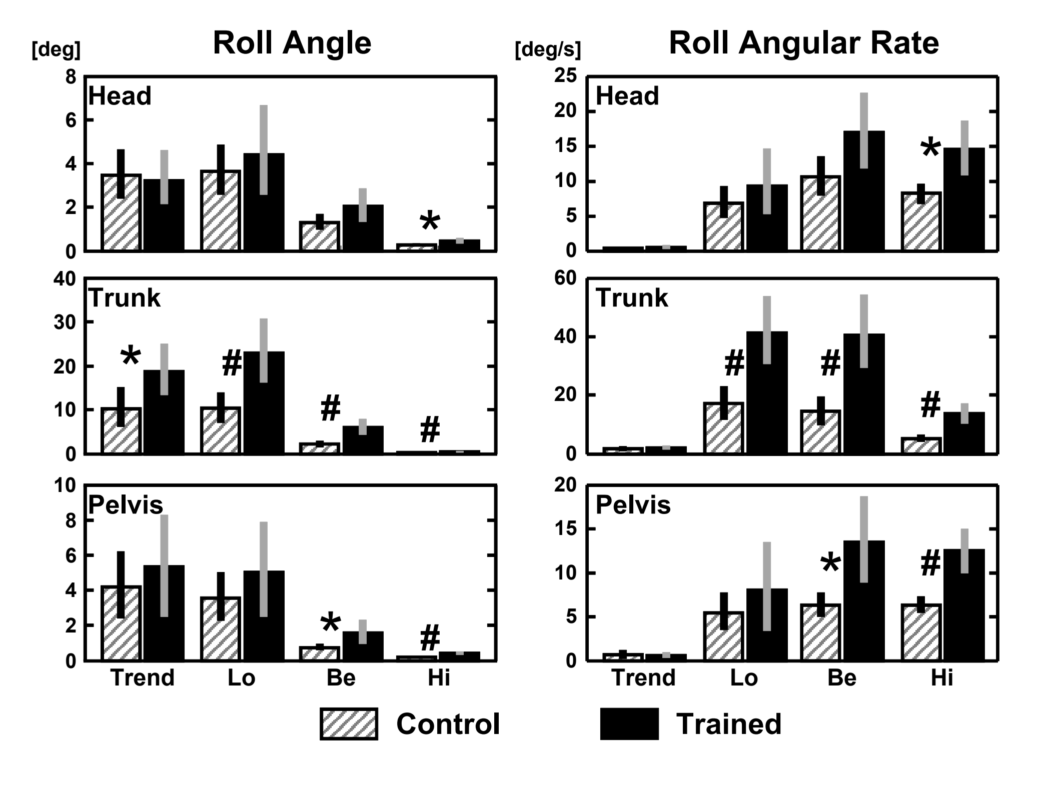


Fig. 6. 90% ranges of roll angle and angular rate for the eyes closed foam task
Mean 90% roll sway angle and angular rate range comparisons between trained and control subjects are shown for four different frequency bands. The column height show the mean 90% ranges for the eyes closed foam task. The vertical bars indicate the 95% confidence intervals of these means. For both roll angle on the left and roll angular rate on the right, the ranges corresponding to the different frequency bands - base trend (Trend), lowpass filtered (Lo), base band filtered (Be) and high pass filtered (Hi) - are displayed for head, trunk and pelvis. The black columns show data of trained and the hatched columns data of control subjects. An asterisk (*) or number sign (#) indicates that the values of the trained subjects are either significantly greater for the trained at a significance level at 5%, respectively, 1% ($p<=0.05/0.01$ Mann-Whitney exact).

Phase Estimates between Segments

The PSD ratios in Fig. 4 as well as the TFs (not shown) show a unity value crossover between 0.7 and 3 Hz with different low and high frequency gains. These findings support our previous analysis of phase differences using regression plots of the recorded samples split in low, crossover and high frequency bands with low and high pass cut-offs at 0.7 and 3 Hz, respectively (Horlings et al., 2009b, Honegger et al., 2012a, Honegger et al., 2012b). Thus, as expected, regression angles for the low frequency band were (in phase) positive angles and those for the high frequency band negative (out of phase). ANOVA analysis revealed a trend for the trunk to head regressions angles to have smaller magnitudes (closer to 45 or 135 for in-and-out-of-phase movements, respectively, for the more difficult tasks (EOF and ECF) post hoc but this task effect was not significant when corrected for multiple comparisons.

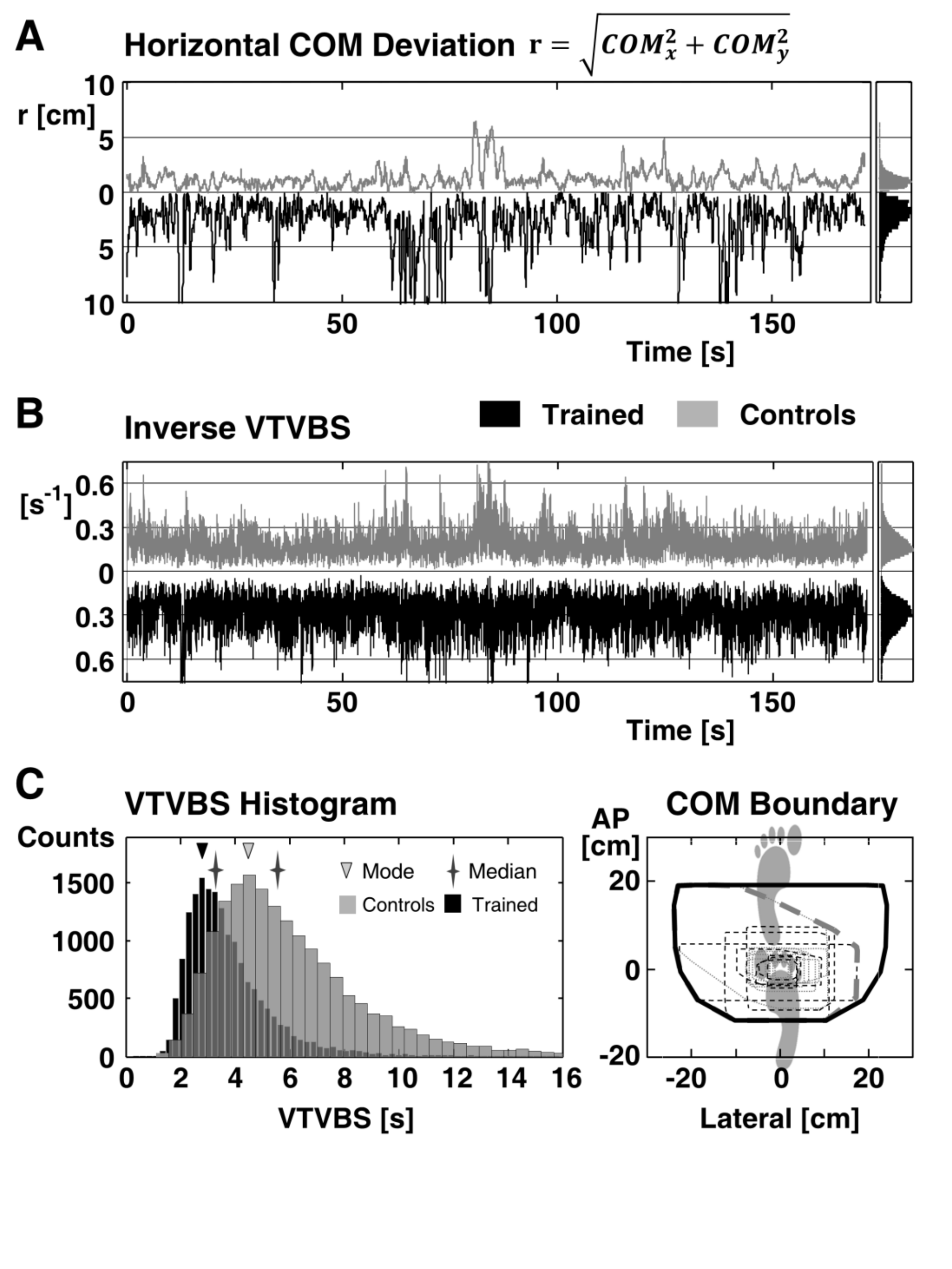


Fig. 7. Typical CoM deviations and VTVBS times

The outcome of model calculation for CoM deviations and VTVBS calculated with the subject data of Fig. 1 – a typical tightrope walker and a control – A shows the horizontal CoM deviation from the mean and B the inverse VTVBS. In C, on the left, histograms for VTVBS are displayed. Note that CoM for the trained shows more excursion and the distribution of VTVBS is more compact for the trained resulting in smaller median (4.48 vs 5.44s) and mode (2.81 vs 3.35s) values. On the right (D) the construction of the virtual base of support (VBS) is depicted. The boundary consists of 23 straight line segments (for details see methods sections).

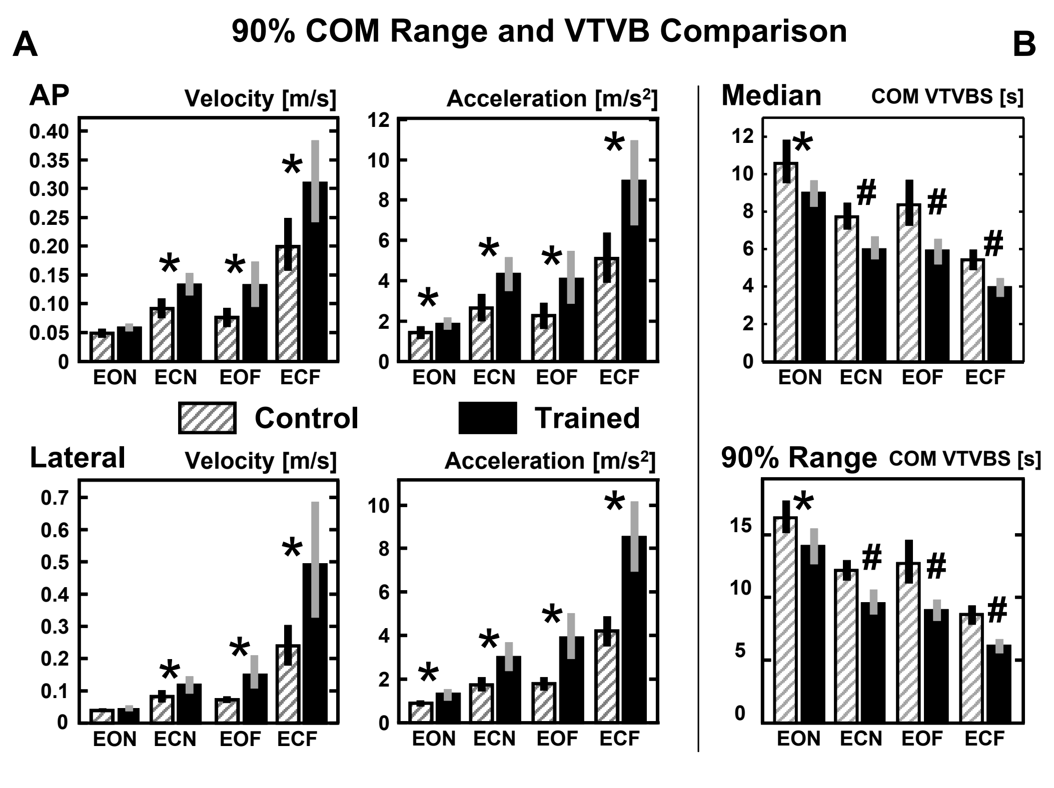


Fig. 8. 90% ranges of roll angle and angular rate for the eyes closed foam task
Average 90% centre of mass (CoM) velocity and acceleration range (part A) and median and 90% range of CoM virtual time to contact the virtual base of support (VTVBS, part B) are compared between trained (tightrope walkers) and control subjects. The column height shows the group averaged values - 90% ranges or median - with 95 % confidence intervals as vertical bars. The anterior posterior (AP) kinematic components are displayed in the upper row of Fig. 6A, the lateral in the lower row. Fig. 6C shows similar statistics for the VTVBS. The black columns show the average trained data, the hatched columns the averages of the controls. An asterisk (*) or numer sign (#) indicates that the values of the trained subjects are either significantly greater for CoM or smaller for VTVBS at a significance level at 5% respectively 1% ($p<=0.05/0.01$ Mann-Whitney exact).

CoM and VTVBS

Based on the larger angles and angular velocities of the trunk for the trained subjects (see Figs. 1, 2, 5 and 6) we expected to observe differences in CoM too. Fig. 7 illustrates the outcome of our model calculation for CoM and VTVBS for the data of a typical tightrope walker and a typical untrained control subject standing TS ECF (same subjects as in Fig. 1). In Fig. 7A the distance of CoM from mean CoM position is displayed. For both subjects CoM position was always inside the VBS (r≤10cm). It can be clearly seen that the trained subject's CoM has more excursions and variance than the control. In Fig.7B the inverse of the $VTVBS$ is plotted for the same subjects because this measure is more intuitively compared with the data of Fig. 7A, being a measure of velocity towards the VBS. It may be observed that values are «faster» for trained subjects. In Fig. 7C histograms of $VTVBS$ are displayed for the same subjects. The distributions are positively skewed – a Pearson Type VI distribution provided best fitsthe data – having greater mean than median and greater median than mode (peak value) with smaller values for the trained subject. Fig. 8B shows the group median and 90% range for the VTVBS for all test conditions. Group differences that were significant in post-hoc analyses are indicated in the figure. The trained subjects always had shorter VTVBS times and short ranges regardless of the task. The shorter VTVBS times indicate that the trained were always closer to the base of support or the velocities of the CoM were greater for a given CoM position.

The acceleration and velocity of CoM movements showed significant group differences as depicted in Fig. 8A. In contrast there were no group differences for the 90% range of CoM amplitude deviations from the mean position. Mean values for the untrained controls were 2.7, 4.0, 3.2, 11.6cm and for the trained 3.0, 4.7, 5.1, 18.1cm for the EON, ECN, EOF, ECF tasks, respectively. ANOVA analysis revealed only a trial effect, no group differences.

DISCUSSION

Common Movement Patterns

Our evidence shows that pelvis, trunk and head movement patterns of trained tightrope walkers during tandem stance (TS) on a tightrope are very similar to those of the same subjects during tandem stance eyes closed on foam (ECF). Further, the intersegmental coordination for ECF is similar to that of tandem stance in other TS sensory conditions (EON, ECN, EOF). These similarities enabled us to compare the movement of trained subjects with those of untrained control subjects. Moreover, as the movement patterns did not differ between the two groups of subjects we could consider how proprioceptive and vestibular inputs were differently weighted by the two groups during TS balance control.

CoM analyses

Despite the obviously faster roll and pitch trunk movements in the trained subjects (Figs. 5 and 6), the excursions of the CoM were not less than those of controls as we originally thought. Our original concept emerged from observations that when untrained subjects are provided biofeedback of lower back/pelvis sway angles during feet together stance, angles are held within tighter bounds using greater trunk velocities (Davis et al., 2010).

Thus the goal of the greater trunk movements in trained subjects in TS appears not to be to control CoM within a more restricted region. CoM and VTVBS results show that CoM is less restricted and the smaller VTVBS indicates a higher CoM proximity to critical base-of-support values. This proximity does not indicate an instability as the trained subjects did not fall. We assume, as discussed below, that it is part of an exploratory behaviour required to maintain sensory weightings for movements over the extent of the base of support and to have a better internal model of stability bounds.

Under the EOF, ECF and EOR conditions TS stability cannot be controlled by ankle torques alone, trunk movements must be used to generate the necessary balance restoring torques. It is counterintuitive, but once the upper trunk is in an unstable position it, must be rotated in the direction of fall to prevent fall. These corrective trunk movements in roll in relation to the pelvis are seen as increased angular velocities. When such a corrective trunk sway movement reaches its maximum angular velocity the pelvis is pushed into an anti-phase stabilising motion (Fig. 1C, D) in roll. This anti-phase mode operates simultaneously with an in-phase mode of trunk and pelvis movements. The trunk to pelvis measures indicated a crossover in characteristics between 1 and 3 Hz. This crossover has been attributed to two simultaneous modes of action: one in which body segments move more in-phase and another when motion is anti-phase as described above (Creath et al., 2005, Horlings et al., 2008). The anti-phase mode is much clearer in TS than in stance with the feet shoulder width apart (Honegger et al., 2012b).

In our CoM modelling we assumed, as did Goodworth and Peterka (2010), that pelvis movement is identical to that of the legs. Under this assumption CoM ML sway was maximally 4.5cm for ECN TS in their study, comparable to 4.0 and 4.7cm for the controls and trained, respectively, in our study. Another group, Forth et al. (2007), modelled CoM data for an «absent vision and sway-referenced support» (SOT 5) of Neurocom™. This task compares with sway pattern for the standing feet together on foam (Allum et al., 2002) but is less challenging than our ECF task as the instability is only in the pitch direction. Their CoM trace SM presented a 90% CoM range of 5.9cm, comparable to that between our EOF and ECF tasks. These comparisons indicate that it was valid as a first approximation to treat pelvis and legs as one unit for CoM calculations.

Pelvis and Trunk in Pitch

We had also not expected that pitch movements along the more stable AP direction would exceed roll movements in magnitude in all quantities measured. An obvious role for the pitch movements would be to help lateral stabilisation by generating a corrective stabilising ML torque via a coriolis effect. The corrective movement patterns in pitch are similar to those seen in roll. The low frequency PSD pelvis/trunk ratios and TF magnitudes are about 5 times larger than the ones in roll but still significantly less than 1. Torque generated at the ankle joints can stabilize pelvis to swing less in the opposite direction when the trunk pitches. All in all, pitch over roll domination seems part of the movement strategy of TS.

Patterns across Tasks

The task comparisons in Fig. 4 indicate that tandem stance intersegmential movement patterns for low (<1Hz) and high (>3Hz) frequency movements do not change with increasing task difficulty, rather changes in relative movement amplitude between segments occur and these are responsible for changes in observed amplitudes of sway shown in Figs. 5, 6 and 8. That is, with increased trunk and CoM sway velocity, relatively less head and pelvis sway amplitude and velocity occurred. Such response non-linearities have been assumed to be caused by sensory reweighting of vestibular and proprioceptive contributions to balance control (Peterka 2002, Peterka and Loughlin 2004). The concept of sensory weighting is based on increased use of vestibular inputs as sway amplitudes increase (Peterka, 2002, Peterka et al., 2011). In this study greater sway amplitudes were induced using a foam support surface. Under this condition greater reliance on vestibular compared to lower leg proprioceptive inputs is mandatory for stability (Horlings et al., 2009b) as proprioceptive inputs are less effective when standing on foam. The question arises whether the trained subjects rely more on vestibular or intersegment proprioceptive gains to reduce the overall transmission of trunk movements to the head and pelvis. This question can probably only be answered with modelling techniques as appropriate patient models are extremely rare (Bloem et al., 2002). Proprioceptive loss normally commences in the extremities and then spreads to more proximal segments, not vice versa.

Differences between Tasks and Groups – Implications for Sensory Reweighting

As task difficulty increased, angular velocities of trunk increased as well (Fig. 4). Because angular velocities increased less for head and pelvis the gains with respect to the trunk decreased. In contrast to low frequency movements, for the high frequency movements the ratio pelvis/trunk velocities decreased less (Fig. 6). If it is assumed that proprioceptive signals from the lumbo-sacral and neck joints to stabilise the head and pelvis more with respect to the trunk, then this result would imply that the gains of these proprioceptive systems would be reweighted both with task difficulty and with training. Our analysis of head resonant frequencies is consistent with the effect of training on proprioceptive weighting because a similar mechanism was observed with vestibular loss subjects (Honegger et al., 2012a). Carpenter et al. (2010) showed in their study that it is likely that postural sway is used by the CNS as an exploratory mechanism to ensure that optimal dynamic ranges are available to the different sensory systems. With this in mind the more rapid trunk movements in trained subjects can be understood as a measure to provide reliable information to the CNS given the velocity dependence of proprioceptive systems (Cao and Pickar, 2011).

Coordination Patterns of the Head

Given the limitation of head roll movements to approximately 5deg and 5deg/s in roll but increasing pitch amplitudes with task difficulty (Fig. 5), the question arises whether this pattern can be summarized as: the head is stabilized in space for roll and locked to the trunk in pitch. Furthermore, the presence of different resonant frequencies for pitch and roll would suggest different movement patterns in roll and pitch. The values reported for the control group in the previous study (Honegger et al., 2012a) – 4.6Hz in roll and 3.2Hz in pitch for an easier task (standing feet side by side shoulder width apart) are comparable to those of the current study. The higher value for roll would be consistent with greater use of proprioceptive feedback in roll, preventing counter rotation of the head due to inertia when the trunk is moved in roll. We speculate that the restriction on head movement in roll is part of the reweighting strategy to avoid excitation of a sacculus triggered otolith reflex. It can be noted in Fig. 3A that head movements are close in amplitude to sacculus thresholds. (Utricle and canal thresholds are much lower than head movement amplitudes observed in the current study (Honegger et al., 2012b)). Thus by restricting roll head movements the saccular vestibular-cervical reflex would not be excited. Super threshold saccular movements may disrupt tracking of lateral tilt angle with the utriculus system, a function critical for tandem stance.

Tightrope stance movement strategy

It was remarkable to us that the movement strategy used while standing on the tightrope (EOR) is similar to that used standing under ECF conditions. Further that the trained subjects did not revert to the movement velocities of controls for the EOF and ECN conditions. The overlap between the EOR and ECF conditions in Figs. 3, 4 and 6 is pronounced. Because the movement strategies during ECF and EOR do not differ significantly from one another, this indicates that the ECF condition could be used in training programs for potential tightrope walkers, with less chance of injuries.

CONCLUSION

Standing in TS is coordinative demanding task were the focus of sensory exploration is shifted to the neck and lumbosacral joints. Movements in pitch and roll are equally important for stability but have different characteristics.

We conclude that keeping balance while standing on a tightrope requires similar inter-segmental movement strategies for the head, trunk and pelvis to those used as those with other, less difficult tandem stance tasks. Different (>3Hz) movement strategies (in- and anti-phase, respectively) are used for high and low frequency movements. Trunk movements are primarily used to control balance with the head and pelvis movements reduced with respect to the trunk. Persons with no experience on the tightrope walking use similar strategies to maintain balance. The difference is the faster trunk movements used by tightrope walkers as they explore the limits of the base of support, yet reducing relative head and pelvis movements via proprioceptive reweighting.

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CONFLICT OF INTEREST

The authors report that F Honegger and JHJ Allum worked as consultants for the company producing the SwayStar™ equipment used in this study.

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GLOSSARY

Roll: angle rotation in the lateral plane, i.e. side-to-side

Pitch: angle rotation in the sagittal plane, i.e. back-forward

ABBREVIATIONS

ANOVA, analysis of variance; CNS, central nervous system; CoM, centre of mass; BMI, body mass index; BP, band pass; ECN, eyes closed on firm surface; ECF, eyes closed on foam ; EON, eyes open on firm surface; EOF , eyes open on foam; HP , high pass; LP, low pass; ML, medial-lateral; PSD , power spectral density; TF, transfer function; TS, tandem stance; VBS, virtual boundary of stability; VTVBS, virtual time to virtual boundary of stability.

Chapter 6

Conclusions

HEAD MOVEMENTS

Our current results indicate that under most two legged, feet side by side stance conditions, sway at the head in both the roll and pitch is greater than at the upper trunk and the pelvis. For low and mid-frequency (<0.3 Hz) the head is locked to the trunk i.e. there is a tendency for the head and trunk to move as one unit during low- and mid-frequency sway but head movements always leaned more than expected from a pure inverted pendulum mode. For the BVL subjects the head locking is firmer. Presumably, proprioceptive neck (cervico-collic) reflexes are enhanced in BVL subjects to achieve this head on trunk stabilization which results also in a shift of the resonant frequency of head-on-trunk motion to a higher frequency for BVLs: from 3.2 to 4.3 Hz in pitch, 4.6 to 5.4 Hz in roll. In the pitch plane this resonance was damped out for high frequencies as suggested by Keshner (2000). In contrast there was no such damping out in the roll plane. Because the pattern is the same for HC and BVL subjects, it is likely that biomechanical differences rather then vestibulo-cervical reflexes (Goldberg and Cullen, 2011) play a major role in this pitch to roll difference, but this has to be further investigated.

The major difference in TS was that head roll movements were unchanged and limited approximately to 5deg and 5deg/s in contrast to increasing pitch amplitudes with task difficulty. Furthermore, the presence of different resonant frequencies for pitch and roll would suggest different movement patterns in roll and pitch. The question arises whether this pattern can be summarized as: the head is stabilized in space for roll and locked to the trunk in pitch. The values are of comparable magnitude to those of the first study. The higher value for roll would be consistent with greater use of neck proprioceptive feedback in roll, preventing counter rotation of the head due to inertia when the trunk is moved in roll. We speculate that the restriction on head movement in roll is part of the reweighting strategy to avoid excitation of a sacculus triggered otolith reflex. It may be noted in that roll head movements are close in amplitude to sacculus thresholds. (Utricle and canal thresholds are much lower than head movement amplitudes observed, see chapter 2 and 3). Thus by restricting roll head movements, the saccular vestibular-cervical reflex would not be excited. Super threshold saccular movements may disrupt tracking of lateral tilt angle with the utriculus system, a function critical for tandem stance. Research with an appropriate selected stimulus profile with a 3D rotating chair, could possibly resolve the speculative character of this interpretation.

NORMAL STANCE

The conclusion of Chapter 2&3 and our previous work (Horlings et al., 2009), and that of other investigators (Creath et al., 2005, Pinter et al., 2008) is, that the dynamics of postural sway cannot be captured using a one-segment inverted pendulum model. For the interpretation of the data two separate on-going processes must be considered. The musculoskeletal keeps the body permanently upright. This activity leads to small oscillation around a temporary reference configuration and is observed as anti-phasic intersegment oscillations in the high frequency (>3 Hz) range shown for pelvis to trunk and trunk to head movements. Carpenter et al. (2010) showed in their study that it is likely that postural sway is used by the CNS as an exploratory mechanism to ensure that optimal dynamic ranges are available to the different sensory systems. This would provide an explanation for the baseline drift and the greater leaning than expected from a pure inverted pendulum mode as observed in Chapter 2&3 and Horlings et al. (2009).

TANDEM STANCE

Keeping balance while standing on a tightrope requires similar inter-segmental movement strategies for the head, trunk and pelvis to those used as those with other, less difficult tandem stance tasks. Movements in pitch and roll are equally important for stability but have different characteristics and the focus of sensory exploration is shifted to the neck and lumbosacral joints. Trunk movements are primarily used to control balance with the head and pelvis movements reduced with respect to the trunk. Persons with no experience on the tightrope walking use similar strategies to maintain balance. The difference is the faster trunk movements used by tightrope walkers as they explore the limits of the base of support. At the same time they reduce relative head and pelvis movements to those of the trunk via proprioceptive reweighting, although in absolute terms the faster trunk movements are paralleled with faster head, pelvis and CoM movements, with the exception of head roll that was unchanged with task or training.

These findings of the comparison with tightrope walkers show also, that when TS is used for clinical balance assessment and scientific studies to score balance performance, one should note that a person who moves more may not be a bad performer!

BIOFEEDBACK

In the introduction, the question was raised whether tandem stance would provide a stance control situation to study the effect of a feedback device in one control dimension, i.e. roll only. Our studies indicate that roll and pitch movements are equally important in tandem stance, that inter-segmental movement strategies are dominant, and therefore a simplification to a roll only treatment is not possible.

Two-legged side-by-side stance seems to be definitely the simplest situation to study the effect of a biofeedback, a stance configuration, for which the seminal study presented in chapter 4 has shown that the kind of feedback provided - thresholded pelvis sway - reduces sway and improves antagonistic muscle synergies in BVL patients under different sensory condition. The main reason for this seems that beside the presence of a low frequency in-phase and a superimposed high frequency anti-phasic mode the low frequency mode dominates in amplitude and the pelvis deviation angles are a good estimator for the tilting angle around the net body axis (termed idiotropic vector in Mittelstaedt (1983, 1999)) which is believed to be controlled by the CNS. The study also shows that feedback information can be fused with other sensory information. From the discussion on tandem stance we understand that pelvis sway is not an efficient estimator for the net body axis. One therefore would expect that a feedback provided similar to the one presented in chapter 4 is ineffective for tandem stance. This is what we actually observed when we tested some older subjects with feedback under the tandem stance condition. They complained that feedback provide during tandem stance information to be irritating and disturbing and in effect the control subjects performed worse with feedback because correcting trunk movements where penalized by the feedback. So probably a feasible feedback under general stance and walking conditions can be only provided by the fused output of several sensors. Most likely the feedback unit has to be multimodal and adapt feedback information depending on movement patterns used. Another common misconception brought two light with the analysis of tandem stance is the intriguing fact that the increased stability should not be equated with less sway or body movement. Actually, stability implies an equilibrium which we can both observe and measure. Static stability and dynamic stability reflects around no movement and some movement, respectively. Besides the «positive» synergetic effects in muscle activities, prosthetic biofeedback makes only sense when it improves the stance and movement characteristics, such that the body when disturbed from an original state of stable motion in an upright position, the return to stable motion is improved and the risk of fall is reduced. For example, for a BVL subject walking in the twilight and stumbling on a stone. One can see there quite a few questions future research and development still has to resolve!.

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Glossary

Glossary

Adaptation:	Any alteration in the structure, form or mechanisms, by which the organism becomes better suited to handle a new or changed environment.
Cervico-collic Reflex (CCR):	A cervical (neck) reflex that stabilizes the head on the body with neck muscle activation.
Cervico-ocular Reflex (COR):	A reflex driven by cervical (neck) proprioceptors. As the COR can complement the VOR under particular circumstances it becomes relevant when considering compensation for vestibular loss.
Circular Statistics:	A subfield of statistics, which is devoted to the development of statistical techniques for the use with data lying on an angular or periodic scale.
Surface Electromyogram (EMG):	A recording of electrical potentials generated by muscle fibres from electrodes place on the skin.
Feedback System:	A control system which senses the difference between actual and desired states, and produces counteractive effects to minimize the difference.
Non parametric test (statistics):	A statistical test dealing with variables without making assumptions about the form or the parameters of their distribution.
Perturbation:	A disturbance forcing the body to deviate from its current position or movement.
Pitch:	Segment rotation in the frontal plane.
Proprioception:	The awareness of the load, position and movement of one part of one's own body compared to another from sensory information provided by receptors known as proprioceptors.
Postural control:	The management of balance of the body's position.
Sensory reweighting:	A process that changes the relative importance of an individual sensory system with respect to an other sensory system for maintaining postural control.
Roll:	Segment rotation in the frontal, sagittal, transverse plane.
Sensory cue:	A signal that can be extracted from the sensory input by a perceiver, that indicates the state of some property of the world that the perceiver is interested in perceiving.
Somatosensory:	Means related to sensory activity having its origin not in the special sense organs such as eye, ears and vestibular organ providing information about the state of the proper body and its immediate environment.

Strategy:	A plan, method, or series of manoeuvres for obtaining a specific goal or result.
Tandem stance:	Standing with one foot ahead the other.
Unimodal sample:	A sample drawn from some continuous distribution having a single mode/maximum.
Vestibulo-ochlear reflex(VOR):	A reflex in which eye position compensates for movement of the head. It is induced by excitation of the vestibular apparatus.
Vestibulo-spinal reflexes(VSR):	Coordinates the head and neck movement with the trunk and body, with the goal of maintaining the head in an upright position and can act in synergy or opposition to neck muscle stretch reflexes which work to keep the neck upright
Wavelet:	A wavelet is a mathematical function with zero mean and compact extent useful in digital signal processing and image compression. The principles are similar to those of Fourier analysis, which was first developed in the early part of the 19th century.
Yaw:	Segment rotation in the transverse plane.

Appendix

Appendix

Overview

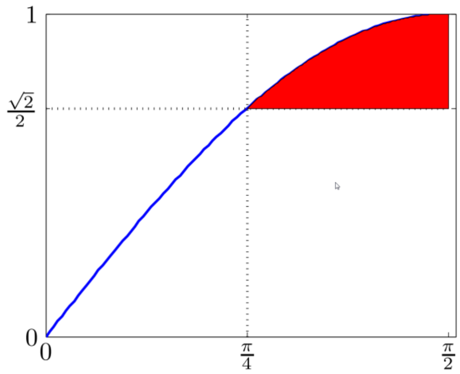
The PSD figures presented in chapters 2 to 5 displayed either estimates of the thresholds of the semicircular canals and the otoliths or the noise spectrum of the sensors or both. The following two sections summarize the calculations of the threshold estimates. The third section is briefly dedicated to sensor selection and comparison. An image gallery presenting the equipment used in the studies rounds completes the last section of this supplement.

Angular velocity power spectrum characteristics at sinusoidal angular threshold stimulation

Problem outline

The calculation is guided by the following ideas and assumptions:

- The overall concept of vestibular perception has its limitations. For the results of aeronautics studies that mostly focus on perception thresholds the concept of Mulder's rule (AvMed, 2013) is applied, that states basically states, that an acceleration is only perceived if the integral taken over a specific time interval is above a specific threshold (Jones and Young, 1978, Allison et al., 2006). It is assumed, that for the physiological thresholds the same must be true, but this is not stated directly in the literature. Having this in mind and to keep our estimate conservative, we calculated the threshold characteristic by considering a stimulation for which the average acceleration over 1/8 of the cycle i.e. over the interval [π/4,π/2] is equal to the thresholds.

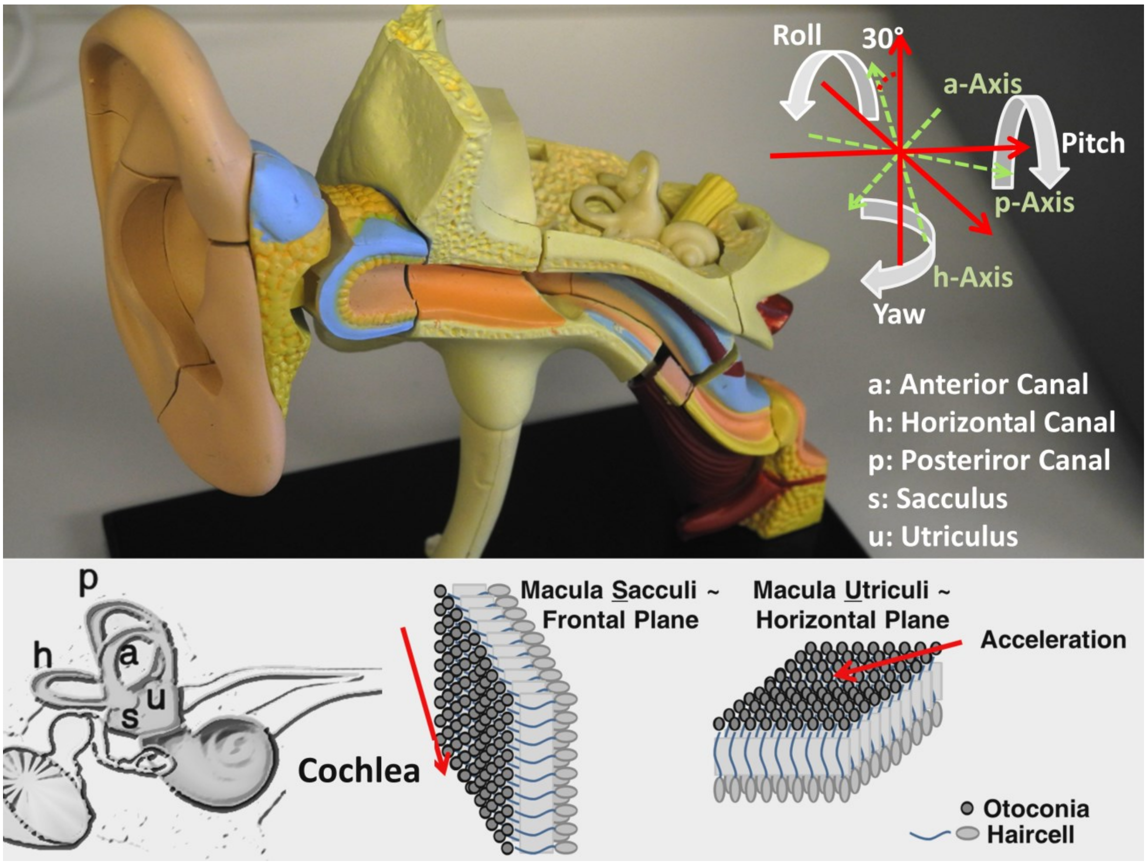


- The body height was assumed to be 1.7 m. This corresponds approximately 1.57 m (h) from the sole of the foot to the vertical location of the vestibular organ.

- The body was assumed to move as an inverted oscillating with an angle $\phi(t) = \phi_0 \cdot \sin(2\pi \cdot f \cdot t)$ where $\phi(t)$ is the deviation angle from the vertical and ϕ_0 the amplitude or maximal deviation.

- The thresholds are constant over the frequency range considered.

- The geometrical alignment of the canals and otoliths is shown on the following annotated private photo taken from a «4D Human Anatomy Model Ear» puzzle.



Threshold found in the literature

The following tables show the mean physiological and perceptual ankle proprioception and vestibular thresholds in different planes for balance control. No references were found for physiological vertical linear acceleration threshold values.

Ankle Proprioception		
units	Physiological[1]	Perceptual[2]
Deg	0.628 (0.25-1.25)	0.60 (0.41-0.99)

1 (Fitzpatrick and McCloskey, 1994, Simoneau et al., 1996, Winter et al., 1998)

2 (Thelen et al., 1998, Refshauge et al., 2000, Deshpande et al., 2003, Kavounoudias et al., 2005)

Angular Acceleration				
	Physiological		Perceptual	
units	Pitch[3]	Roll[4]	Pitch[5]	Roll[6]
deg/s^2	0.05	0.51	6.75 (5.3-8.2)	0.87 (0.5-1.18)

3 (Meiry, 1965)

4 (Seemungal et al., 2004)

5 (Sadoff et al., 1955)

6 (Clark, 1967, Guedry, 1974, Seemungal et al., 2004)

Appendix

Linear Accelerations						
	Physiological			Perceptual		
units	AP[7]	Lateral[8]	Vertical	AP[9]	Lateral[10]	Vertical[11]
cm/s²	4.75 (1.7-8.5)	3.6 (1.1-6.5)	-	5.57 (2,4-8)	5.4 (3,4-8)	9.77 (6-15,4)

7 (Gundry, 1978, Winter et al., 1998, Kingma, 2005)

8 (Gundry, 1978, Winter et al., 1998, Hamann et al., 2001, Kingma, 2005)

9,11 (Guedry, 1974, Benson et al., 1986a, Benson et al., 1986b)

10 (Guedry, 1974, Benson et al., 1986a, Benson et al., 1986b, Gianna et al., 1996, Hamann et al., 2001)

Threshold used in calculations

For lateral linear accelerations we used a weighted average of physiological and perceptual values. For vertical we used the only perceptual value because no physiological where found.

Linear accelerations thresholds

	Count	Average	Sum
Lateral	5	4.75	23.75
	4	3.6	14.4
	3	5.57	16.71
	5	5.4	27
Sum	17		81.86
Average			4.82
Vertical			9.77

Angular accelerations thresholds

Roll	0.87
Pitch	6.75

Calculation for vertical receptors

Only the sacculus is considered in the following calculation.

Linear acceleration

The acceleration deviation from g in radial direction at tilt angle ϕ is given by $\Delta a = a_c + g \cdot (1 - \cos(\phi))$, $a_c = h \cdot \omega^2$, where a_c is the centrifugal force, and g earth acceleration.

Let us modulate the inverted pendulum with a sinusoidal of frequency f so that $\phi(t) = \phi_0 \cdot \sin(2\pi \cdot f \cdot t)$. The angular velocity in rad/s is then $\omega(t) = \phi'(t) = \phi_0 \cdot 2\pi \cdot f \cdot \cos(2\pi \cdot f \cdot t)$ or if written in deg/s $\alpha'(t) = \frac{180}{\pi} \cdot \phi'(t) = 360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t)$.

For $\Delta a(t)$, the acceleration relative to the zero position, we then get

$$\Delta a(t) = h \cdot (\phi_0 \cdot 2\pi \cdot f \cdot \cos(2\pi \cdot f \cdot t))^2 + g \cdot (1 - \cos(\phi_0 \cdot \sin(2\pi \cdot f \cdot t)))$$

$$= 4\pi^2 \cdot f^2 \cdot \phi_0^2 \cdot h \cdot \cos(2\pi \cdot f \cdot t)^2 + g \cdot (1 - \cos(\phi_0 \cdot \sin(2\pi \cdot f \cdot t))).$$

Angle

The integral of $\Delta a(t)$over 1/8 cycle for a given frequency f represents a nonlinear equation for ϕ_0. The equation can be solved by numerically finding the roots $\phi_0(f)$ of the function, that is that the threshold (see section «Thresholds used in calculation») is just reached.

$$F(\phi_0(f), f) = 8 \cdot f \cdot \int_{\frac{1}{8f}}^{\frac{1}{4f}} \Delta a(t) \cdot dt - c_{thresh} = 0.$$

PSD

Recall that the angular velocity is given by $\alpha'(t) = 360 \cdot \phi_0 f \cdot \cos(2\pi \cdot f \cdot t)$ and that the *PSD* is the fourier transform of the auto-correlation function (Wiener–Khinchin theorem). The Fourier transform $F$of $\alpha'(t)$ is given by

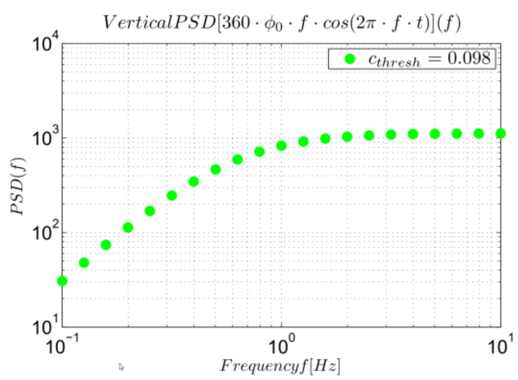
$$F_t[360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t)](k) = \int_{-\infty}^{\infty} e^{-2\pi ikt} 360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t) \cdot dt = 360 \cdot \phi_0 \cdot f \int_{-\infty}^{\infty} e^{-2\pi ikt} \cos(2\pi \cdot f \cdot t) \cdot dt = 360 \cdot \phi_0 \cdot f \cdot \frac{1}{2}[\delta(k-f) + \delta(k+f)],$$

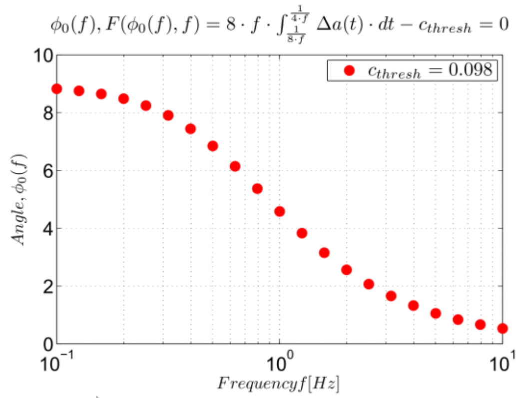
where $\delta$is the Dirac delta function. Therefore the *PSD* of $\alpha'(t) = 360 \cdot \phi_0 f \cdot \cos(2\pi \cdot f \cdot t)$is given by

$$PSD[360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t)](f) = F_t[C(360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t))](k) = (360 \cdot \phi_0 \cdot f)^2 \cdot \frac{1}{2}[\delta(k-f) + \delta(k+f)].$$

Plots

The plots show the PSD solution for each frequency f.





Calculation for horizontal receptors

Lateral linear accelerations (roll) are detected by both utriculus and sacculus. In anterior posterior direction (Pitch) the utriculus is involved only.

Linear acceleration

The net acceleration for a given tilt angle $\phi(t)$of a angular modulation $\phi(t) = \phi_0 \cdot \sin(2\pi \cdot f \cdot t)$is given by

$$\Delta a = g \cdot \sin(\phi_0 \cdot \sin(2\pi \cdot f \cdot t)) - h \cdot \phi_0 \cdot (2\pi \cdot f)^2 \cdot \sin(2\pi \cdot f \cdot t).$$

Angle

Again we assume that the average acceleration$\Delta a(t)$ moving from $\frac{\phi_0}{2}$ to ϕ_0- is equal to the thresholds Threshold used in calculations.

Appendix

$$c_{thresh} = \frac{\int_{\frac{1}{8f}}^{\frac{1}{4f}} \Delta a(t) \cdot dt}{\int_{\frac{1}{8f}}^{\frac{1}{4f}} dt}$$

$$= 8 \cdot f \cdot \int_{\frac{1}{8f}}^{\frac{1}{4f}} (g \cdot \sin(\phi_0 \cdot \sin(2\pi \cdot f \cdot t)) - h \cdot \phi_0 \cdot (2\pi \cdot f)^2 \cdot \sin(2\pi \cdot f \cdot t)) \cdot dt$$

For a given frequency f this integral represents a non linear equation for ϕ_0. The problem is solved by numerically finding the roots $\phi_0(f)$of the function

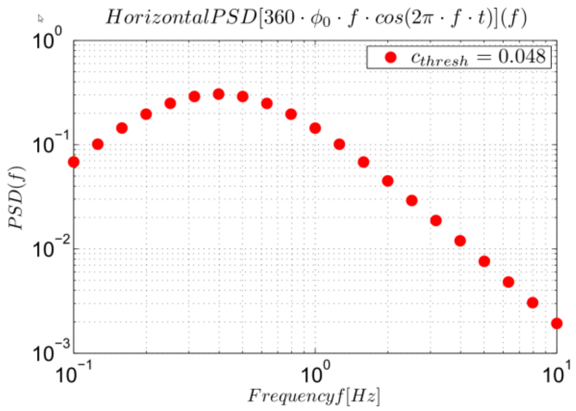
$$F(\phi_0(f), f) = 8 \cdot f \cdot \int_{\frac{1}{8f}}^{\frac{1}{4f}} \Delta a(t) \cdot dt - c_{thresh} = 0.$$

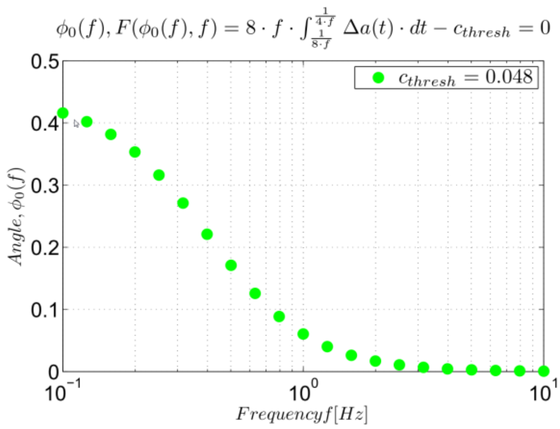
PSD

The procedure to calculate is the same as for the vertical acceleration.

Plots

The plots show the PSD solution for each frequency f.





Calculation for canal stimulation

Angular acceleration

The net angular acceleration for a given tilt angle $\phi(t)$ of a angular modulation

$$\phi(t) = \phi_0 \cdot \sin(2\pi \cdot f \cdot t)$$

is given by the second derivative

$$\phi''(t) = \phi_0 \cdot \sin''(2\pi \cdot f \cdot t) = -\phi_0 \cdot (2\pi \cdot f)^2 \cdot \sin(2\pi \cdot f \cdot t).$$

Angle

Again as with the linear acceleration, we assume that the average of $\phi''(t)$ moving from $\frac{\phi_0}{2}$ to ϕ_0, is equal to the thresholds listed in the section «Thresholds used in calculation» above.

$$c_{thresh} = \left| \frac{\int_{\frac{1}{8f}}^{\frac{1}{4f}} \phi''(t) \cdot dt}{\int_{\frac{1}{8f}}^{\frac{1}{4f}} dt} \right| = \left| 8 \cdot f \cdot [\phi'(t)]_{\frac{1}{8f}}^{\frac{1}{4f}} \right| = \left| 8 \cdot f \cdot [\phi_0 \cdot 2\pi \cdot f \cdot \cos(2\pi \cdot f \cdot t)]_{\frac{1}{8f}}^{\frac{1}{4f}} \right|$$

$$c_{thresh} = \left| 16\pi \cdot f^2 \cdot \phi_0 [\cos(\frac{\pi}{2}) - \cos(\frac{\pi}{4})] \right| = 8\pi \cdot \sqrt{2} \cdot \phi_0 \cdot f^2$$

$$\phi_0 = \frac{c_{tresh}}{8\sqrt{2}\pi \cdot f^2}$$

PSD

Recall that the angular velocity is given by

$$\alpha'(t) = \frac{180}{\pi} \cdot \phi'(t) = \frac{180}{\pi} \cdot \phi_0 \cdot 2\pi \cdot f \cdot \cos(2\pi \cdot f \cdot t) = 360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t)$$

and that the *PSD* is the Fourier transform of the auto-correlation function (Wiener–Khinchin theorem) . The Fourier transform $F$of $\alpha'(t)$is given by

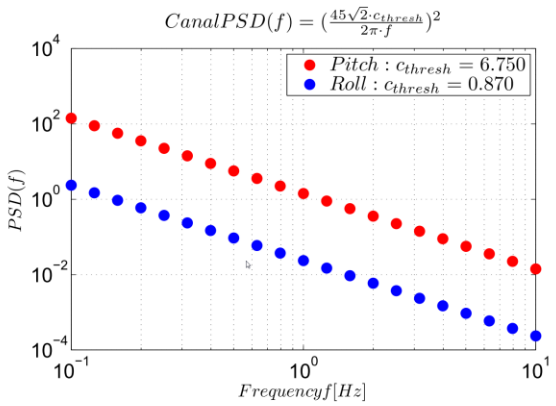
$$F_t[360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t)](k) = \int_{-\infty}^{\infty} e^{-2\pi ikt} 360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t) dt =$$
$$360 \cdot \phi_0 \cdot f \int_{-\infty}^{\infty} e^{-2\pi ikt} \cos(2\pi \cdot f \cdot t) = 360 \cdot \phi_0 \cdot f \cdot \frac{1}{2}[\delta(k-f) + \delta(k+f)] \quad ,$$

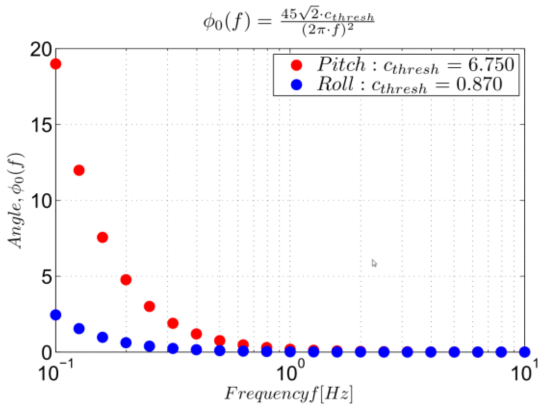
where $\delta$is the Dirac delta function. Therefore the *PSD* of $\alpha'(t) = 360 \cdot \phi_0 f \cdot \cos(2\pi \cdot f \cdot t)$ is given by

$$PSD[360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t)](f) = F_t[C(360 \cdot \phi_0 \cdot f \cdot \cos(2\pi \cdot f \cdot t))](k) =$$
$$(360 \cdot \phi_0 \cdot f)^2 \cdot \frac{1}{2}[\delta(k-f) + \delta(k+f)]$$

Plots

Plotting the *PSD* as a function of $f$gives the following characteristic.





Appendix

Sensor Selection and Comparison

In mixed sensor experiment as presented in Chapters 2-5, where the data is used interchangeable in the data analysis it is important to verify the properties and restrictions of the individual sensors to recognize the limits of interpretation. Where as It was clear for an engineer in the days of analog equipment to feed the signal through an antialiasing filter before digitizing the analog signal and that the biggest problem was drift, currently most inertial sensor have a digital read out and data can be obtained nearly at arbitrary sampling rates. But it is not clear, that the data stream can be reliably digital processed without some pre processing.

Cheaper systems use several micro mechanical sensors. The output of such system is the result of combining the information of the individual sensors using optimal sensor fusion techniques based on noise and error reducing filters like variants of Kalman or particle filters to name just two (Xu et al., 2011, Khaleghi et al., 2013). Algorithmic instabilities to spike input can lead to an erroneous output, noise might be self correlated and slower systems may output a step function, i.e. holding the output constant over several samples, forcing a data smoothing by interpolation.

The integration of sensors in cellular phones and music players makes these devices attractive to be used directly as intelligent sensors. Tuned for gestures and geo localization, the usage in posture motion analysis must be carefully validated, particularly because, as is the case for the iPhone (Lee et al., 2012) specifications for the specially branded and proprietary sensors are not available, and potential specifications can only be derived by forensic industrial analysis (Teardown, 2010).

Sensor specification sheets are not standardized and often tuned to the most used application, making it hard to compare different sensors directly from quality control sheets. Therefore it is best to protocol sensor feature by measuring and identify bias, drift, sample to sample correlation the shape of the power spectrum and making sure that the movements to be track do not result in sensor overload. Information from saturated accelerometers is not useful for any analytical purposes. Further if a sensor is not properly designed for the target measuring its properties may all have an influence on the interpretation of the data. Depending on the outcome of data processing it may be necessary to identify and remove spikes and other glitches in the data stream.

Because of all the above circumstances and the requirement that even angular rates as low as 0.5 deg/s must be tracked, the easiest approach to record reliable data is to use drift free high quality sensors, that stabilize quickly after power on, have Gaussian linear noise (Fig. 1) and that show no sample correlation (Fig. 2). A symmetric noise distribution like the Gaussian distribution is preferred, because the bias in this case is well defined as the average of noise. Angular rate measurements of posture over a longer time period result in a quasi stationary signal with zero mean. Therefore the bias can be basically removed after integration by simply subtracting the mean.

The SwayStar units used host two Litef µ-Fors-6 fiber optic gyros. These sensors have a bias of less the $5\ deg/h$ ($0.0014\ deg/s$) and a standard deviation of $1\ deg/\sqrt{h}$. This means, that for the steady on floor measurements in idle state depicted in Figs.1+2 these sensors already sense environmental influences. By positioning the system with the plane of the roll sensor approximately parallel to the floor this sensor could sense earth rotation, because of the non vanishing projection of the angular velocity vector onto the sensor plane. The pitch sensor being mounted perpendicular to the roll sensor was aligned to north. The projection of angular velocity is much smaller and nearly zero. The different standard deviations of noise above specification observed, indicate that vibrational noise originating from the environment is sensed differently by the pitch and roll sensors. By turning the sensor box one could observe that these value changed. With the specifications for the given µ-Fors-6 sensor, it becomes clear that earth rotation ($15\,deg/h$ resp. $0.0042\,deg/s$) rather then noise distribution limits the lower resolution. Consequently these sensors are excellently suited for tracking body motion in tasks like standing normally on a firm surface with eyes open.

Due to its large size and weight, head motion cannot be measured with a SwayStar device. When the selection for the head sensor was taken, the Xsens MTi sensor was the best MEMS based sensor system that fulfilled all requirements. In contrast to the µ-Fors-6, which represents a single axis angular rate sensor, the Xsens MTi is an attitude and heading reference system operating by mean of sensor fusion. Nonetheless it was used as a two axis gyroscope in the experiments. As it can be seen from Figs.1+2 the noise profile of the MTi has a standard deviation, that is ten times larger then that of the SwayStar, implying that the power spectrum is shifted up by 20dB. The limiting factor for these sensors is the noise itself. This can be clearly seen in Fig.3, that displays a 20s recording of lower trunk, while standing eyes closed on foam support (The measurements are similar in amplitude to these of the head). Above 10Hz, Xsens noise and trunk signals completely overlap. As a consequence in all the studies covered by this theses, the analysis was restricted to the frequency range below 10Hz.

Appendix

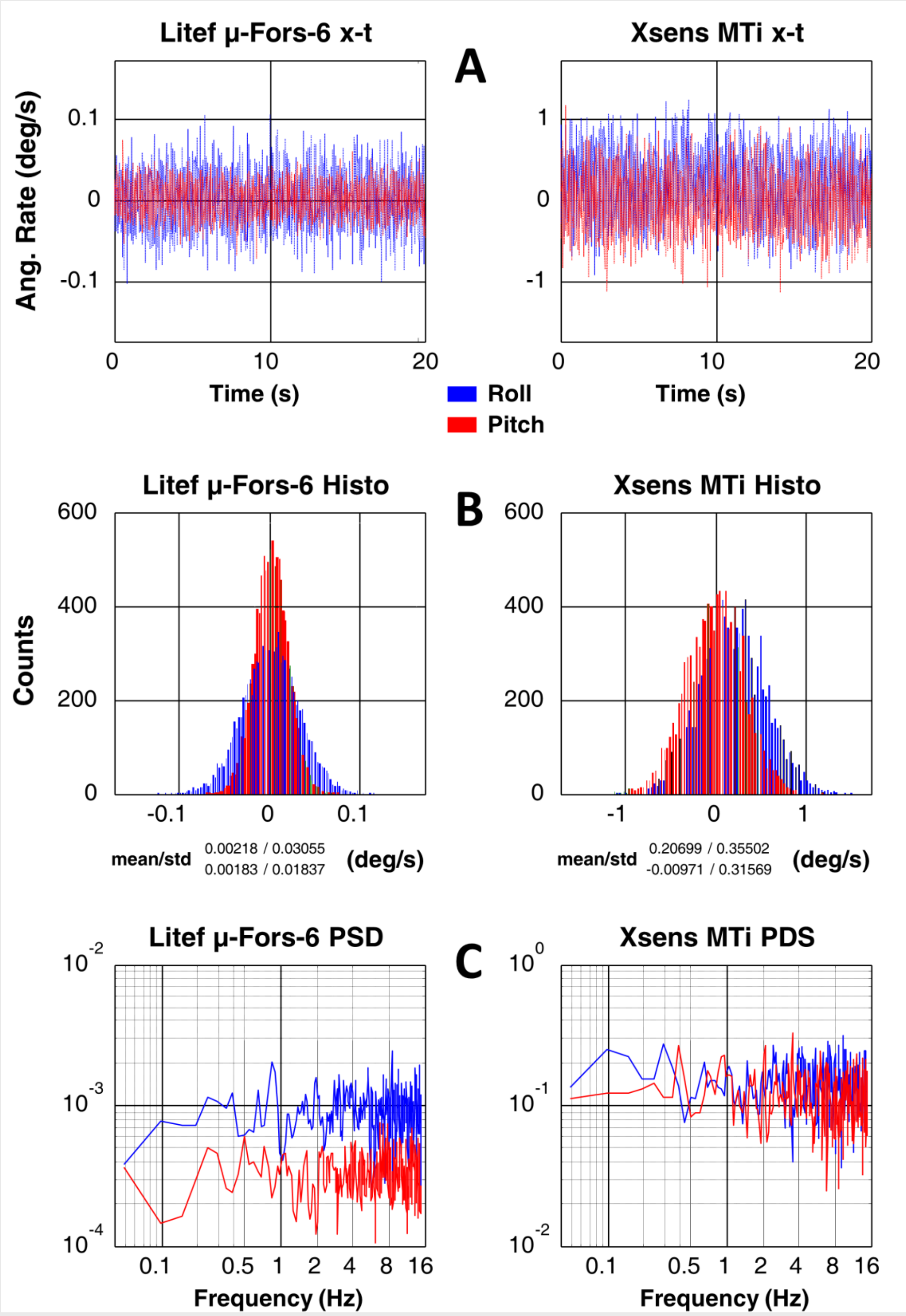


Fig. 1. In sub-figure A the first 20s of a 100s recording sampled with 100Hz are plotted as an x-t plot. The sensors were carefully aligned to each other and placed on floor. The roll plane was parallel to the floor and the pitch plane parallel to the earth rotation axis. In sub-figure B the histograms of the angular rate data of the individual channels are shown. In sub-figure C finally the PSD estimates after Welch's method with a window size of 2048 and an overlap of 1024 samples were displayed. Note that the scales of the Xsens sensor in A and B are a factor of 10 and in C a factor of 100 greater.

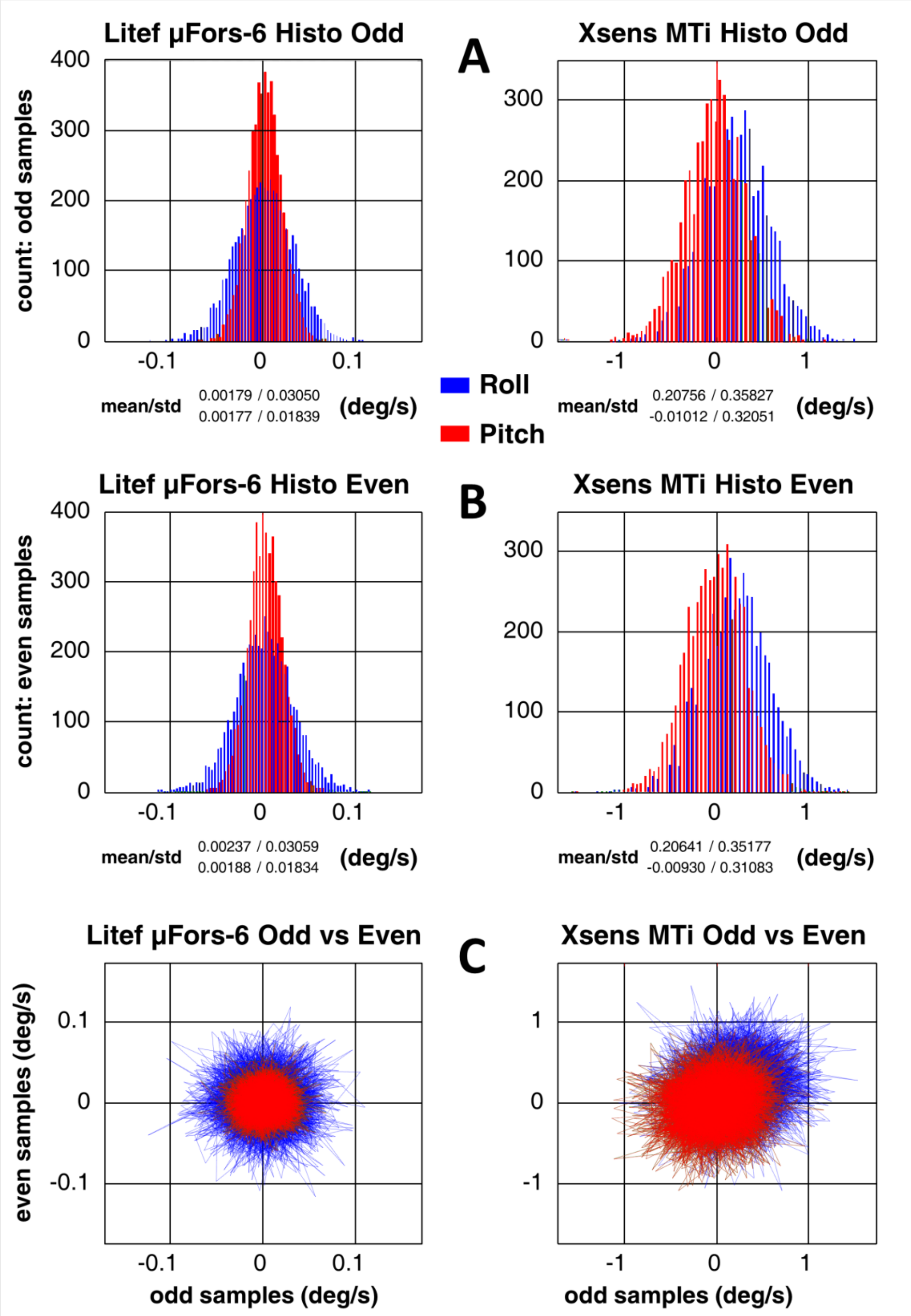


Fig. 2. The data of Fig.1 is split into to different sets, those with even numbered samples and those with odd. Section A displays the histograms for the even data sets and section B the odd sets. In section C odd samples are plotted against the even sample. The data indicates that odd and even samples are uncorrelated.

Appendix

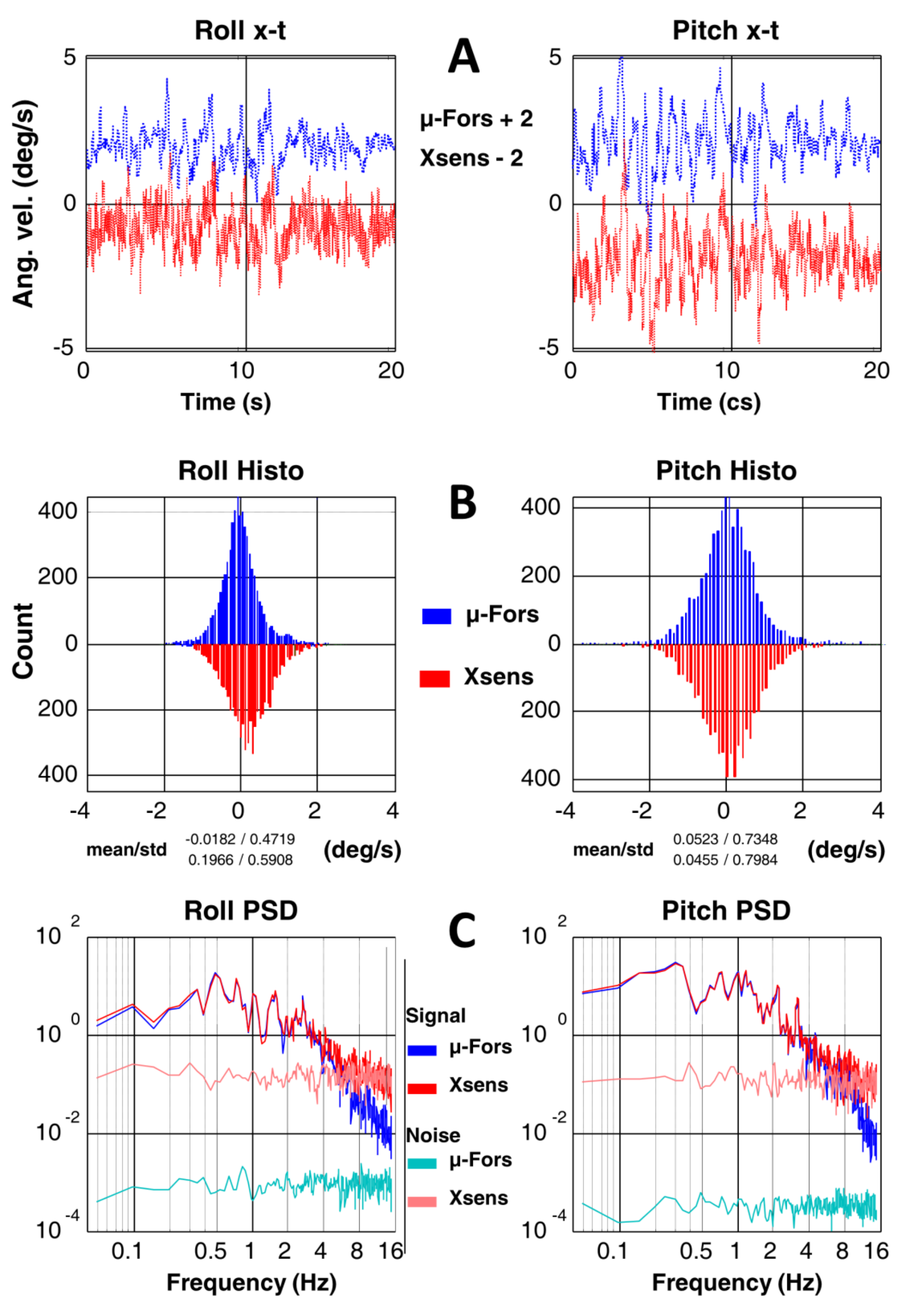


Fig. 3. In section A 20s recording of lower trunk data is shown while standing eyes closed on foam support. In sub-figure B the histograms of the angular rate data of the individual channels are shown. Sub-figure C shows the PSDs together with the «noise levels» of Fig.1.

Image Gallery

Images of the Equipment

The image gallery that follows complements the compact equipment descriptions found in chapters 2 to 5 and gives an pictorial overview of the materials used. As it can be clearly seen except for the EMG system that was stationary, the «lab equipment» is completely portable. The gallery is also intended to also give an impression of reliability of the materials and devices used and to provide a source of inspiration for the solution seeking experimenter.

SwayStar™ unit on converted motor-cycle kidney-belt

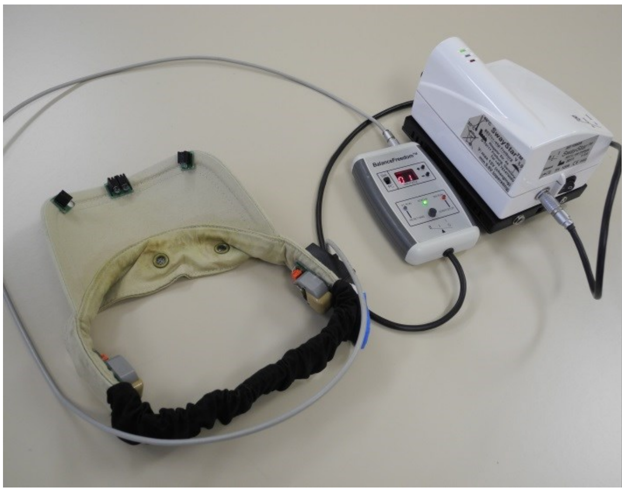
SwayStar™ unit on Shoulder harness

Kidney-Belt Rear View

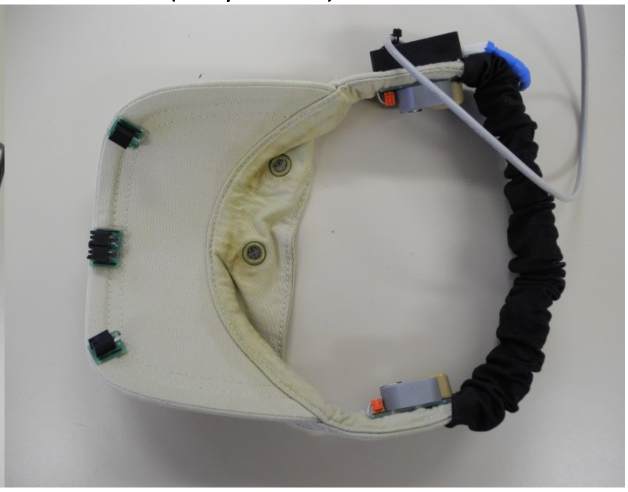
Shoulder Harness Rear View

Appendix

SwayStar™ with Balancefreedom™
Feedback unit and Cap



Feedback Cap with Actuators:
LEDs, Vibrators (Rings), Bone Conduction Transducers (Gray Blocks)



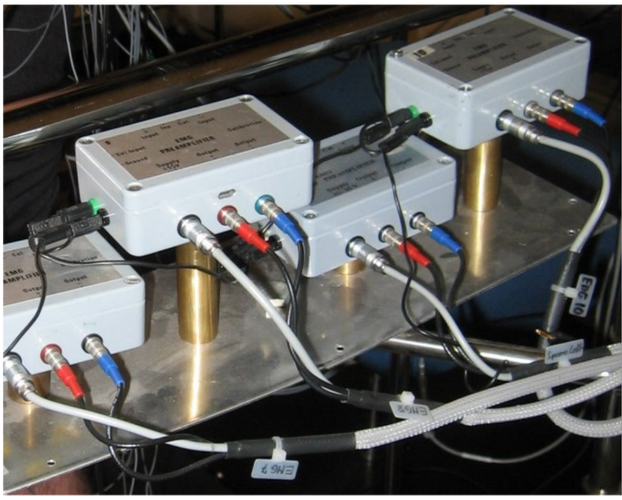
MTI Xsens Sensor Unit on
Headband Mounting

MTI Xsens Sensor Unit on
Shank Belt

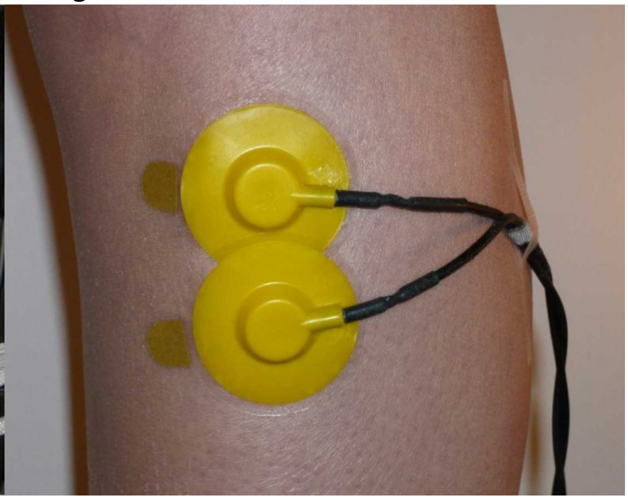
Trigger Unit based on
FTDI RS232 to USB Adapter

Analog Digital Convertor Unit
Data Translation® DT9804

EMG Preamplifier



EMG Electrodes in Place
for Tight M. Tibials Anterior



Foam pad
DIN 53577 Compression Hardness: 5-6 kPa

Tight rope
Under Load

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About the Author

About the Author

CURRICULUM VITAE

Flurin Honegger was born January 5, 1959 in Chur (Switzerland) as the fourth child of Elisabeth and Alfred Honegger. From 1965 to 1971 he attended the primary school in Chur where he also was in the secondary school until 1973. He subsequently attended the «Bündner Kantonschule» in Chur and successfully passed his Matura in 1978. Before starting his studies in physics at the Swiss Federal Institute of Technology (ETH) in Zurich he was recruited in to the Swiss army where he trained as a tank and off-road driver.

In the autumn of 1981 he suspended his studies at the ETH for half a year because he had the opportunity to work as a full-time substitute teacher in mathematics at «Bündner Kantonschule» in Chur. In 1983 he graduated in experimental solid-state physics. His diploma thesis «Neutronen spectroscopic examination on non-stoichiometric Cerium Phosphide» was dedicated to the examination of magnetic structure and phase transitions at low temperatures.

In spring 1984 he joined the group of Prof. Dr. Walter Lukosz at the Lab of Physical Optics of ETHZ were he worked as research and teaching assistant. For two years he focused on laser spectroscopy, single photon counting, observing narrow gap wetting transitions and dendritic grow and writing automated data acquisition and analysis software that was inexistent before he joined the group.

In 1986 he left the «narrow gap» and changed to commuting to Basel and started working with Prof. John H.J. Allum of the ORL department of Kantonsspital Basel, who was looking for a person skilled in programming and interested in posturography and nystagmography. He was then engaged in several research projects funded by the Swiss National Research fund but also in industrial supported projects that were focused on the development of algorithms and software to automate and ease running clinical vestibular test-batteries.

In 1993 Flurin Honegger became fully employed by the University Hospital Basel and had more clinical involvement. He is the information computer specialist of the ORL department with a focus on integrating administrative and medical equipment and as well a research associate with responsibility in coaching PhD and MD students in data acquisition, data analysis, programming, statistics and computer skills besides technical aspects of standard clinical test performed in audiology and neuro-otology. He continuously influenced and contributed to peer reviewed articles. Between 2001 and 2010 he dedicated some of his spare time in the development of the software for clinical balance control measurement equipment.

In 2010 he decided to fulfill his goal of completing a PhD thesis in biomedical engineering. This project was completed in Spring 2013 and was supervised by Prof. John H.J. Allum and by Prof. Bert Müller of the Biomaterials Science Center of the University Basel. He is married since 1990 and has three children born 1990, 1992, 1994, the youngest being a boy. He can be reached by e-mail at the permanent address flurin.honegger@alumni.ethz.ch.

Flurin Honegger used to train in endurance sports several times a week. Due to time restrictions, he had to reduce this activity during his PhD studies. He now longs run and swim, again catching up with exercise, and to burn away the ballast so gained during work on his doctorate.

PUBLICATION TRACK

Conference Abstracts related to PhD studies

1. Honegger F, Hillebrandt IMA, van der Elzen NGA, Tang KS, Allum JHJ (2012a) Strategies and synergies underlying replacement of vestibular function with prosthetic feedback. In: 27th Bárány Society Meeting Uppsala, Sweden: Bárány Society.
2. Honegger F, Tielkens RJM, Allum JHJ (2012b) Movement strategies in tandem stance: Differences between trained tightrope walkers and untrained subjects. In: 20th conference of the International Society of Posture and Gait Research (IPSGR) Trondheim, Norway: IPSGR.
3. Honegger F, van Spijker GJ, Allum JHJ (2012c) Coordination of the head with respect to the trunk and pelvis in the roll and pitch planes during quiet stance. In: 20th conference of the International Society of Posture and Gait Research (IPSGR) Trondheim, Norway: IPSGR.
4. Hubertus JW, Honegger F, Allum JHJ (2012) Coordination of the head with respect to the trunk, pelvis, and lower leg during quiet stance after vestibular loss. In: 27th Bárány Society Meeting Uppsala, Sweden: Bárány Society.

Journal Publications submitted for PhD

5. Honegger F, Hillebrandt IMA, van den Elzen N, G.. A., Tang KS, Allum JH (2013a) The effect of prosthetic feedback on the strategies and synergies used by vestibular loss subjects to control stance. submitted 2012-11-07 to Journal of NeuroEngineering and Rehabilitation (in review).
6. Honegger F, Tielkens RJM, Allum JH (2013b) Movement strategies in tandem stance: Differences between trained tightrope walkers and untrained subjects. submitted 2014-04-28 to Neuroscience.

Journal Publications for PhD

7. Honegger F, Hubertus JW, Allum JH (2012d) Coordination of the head with respect to the trunk, pelvis, and lower leg during quiet stance after vestibular loss. Neuroscience.
8. Honegger F, van Spijker GJ, Allum JH (2012e) Coordination of the head with respect to the trunk and pelvis in the roll and pitch planes during quiet stance. Neuroscience 213:62-71.
10. Honegger F, Hillebrandt IM, van der Elzen NG, Tang KS, Allum JH (2012f) Strategies and synergies underlying replacement of vestibular function with prosthetic feedback. Conf Proc IEEE Eng Med Biol Soc 2012:6132-6136.

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1. Tang KS, Honegger F, Allum JH (2012) Movement patterns underlying first trial responses in human balance corrections. Neuroscience 225:140-151.
2. Allum JH, Carpenter MG, Horslen BC, Davis JR, Honegger F, Tang KS, Kessler P (2011) Improving impaired balance function: real-time versus carry-over effects of prosthetic feedback. ConfProcIEEE Eng Med BiolSoc 2011:1314-1318.

About the Author

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1. Kung UM, Horlings CG, Honegger F, Allum JH (2010) The effect of voluntary lateral trunk bending on balance recovery following multi-directional stance perturbations. ExpBrain Res 202:851-865.
2. Reinert J, Honegger F, Gurtler N (2010) High homogeneity in auditory outcome of pediatric CI-patients with mutations in Gap-Junction-Protein Beta2. IntJPediatrOtorhinolaryngol 74:791-795.
3. Storck C, Gehrer R, Fischer C, Wolfensberger M, Honegger F, Friedrich G, Gugatschka M (2010a) The Role of the Cricothyroid Joint Anatomy in Cricothyroid Approximation Surgery. JVoice.
4. Storck C, Juergens P, Fischer C, Wolfensberger M, Honegger F, Sorantin E, Friedrich G, Gugatschka M (2010b) Biomechanics of the Cricoarytenoid Joint: Three-Dimensional Imaging and Vector Analysis. JVoice.
5. Horlings CG, Carpenter MG, Honegger F, Allum JH (2009a) Vestibular and proprioceptive contributions to human balance corrections: aiding these with prosthetic feedback. AnnNYAcadSci 1164:1-12.
6. Horlings CG, Carpenter MG, Kung UM, Honegger F, Wiederhold B, Allum JH (2009b) Influence of virtual reality on postural stability during movements of quiet stance. NeurosciLett 451:227-231.
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9. Kung UM, Horlings CG, Honegger F, Duysens JE, Allum JH (2009b) Control of roll and pitch motion during multi-directional balance perturbations. ExpBrain Res 194:631-645.
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12. Horlings CG, Kung UM, Bloem BR, Honegger F, Van AN, Van Engelen BG, Allum JH (2008) Identifying deficits in balance control following vestibular or proprioceptive loss using posturographic analysis of stance tasks. ClinNeurophysiol 119:2338-2346.
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18. Carpenter MG, Allum JH, Honegger F, Adkin AL, Bloem BR (2004) Postural abnormalities to multidirectional stance perturbations in Parkinson's disease. JNeurolNeurosurgPsychiatry 75:1245-1254.

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40. Keshner EA, Allum JH, Honegger F (1993) Predictors of less stable postural responses to support surface rotations in healthy human elderly. JVestibRes 3:419-429.
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51. Fattinger C, Honegger F, Lukosz W (1986a) Dendritic solidification in narrow gaps and surface-pressure-induced wetting transition. PhysRevLett 57:2536-2539.
52. Fattinger C, Honegger F, Lukosz W (1986b) Observation of dendritic crystal growth in narrow gaps by microsegregation of lumiscent tracers. HelvPhysActa 59:967-971.
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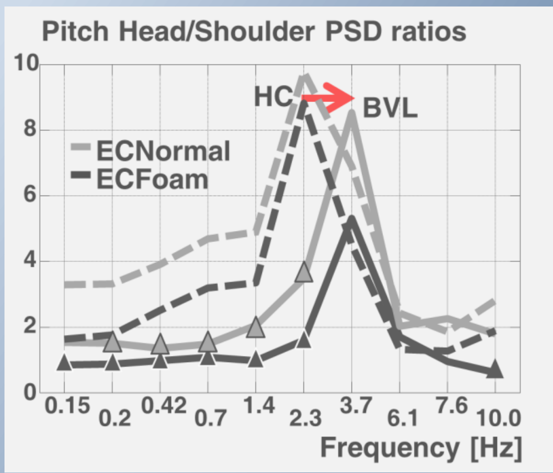
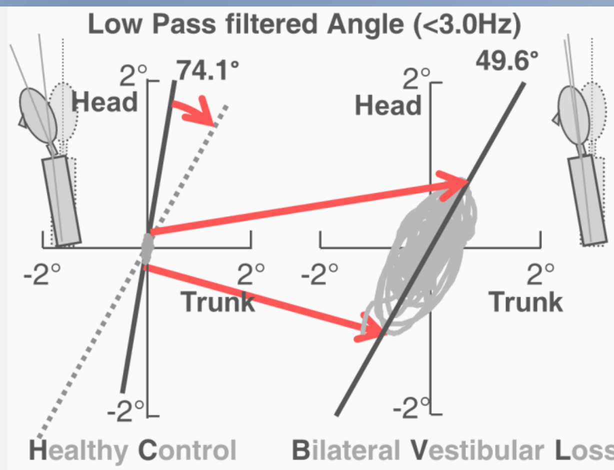
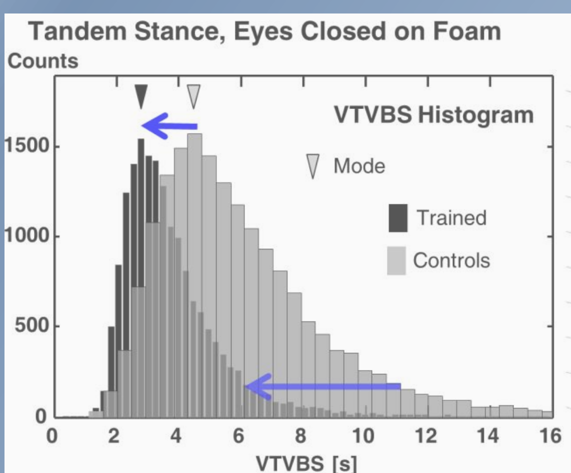
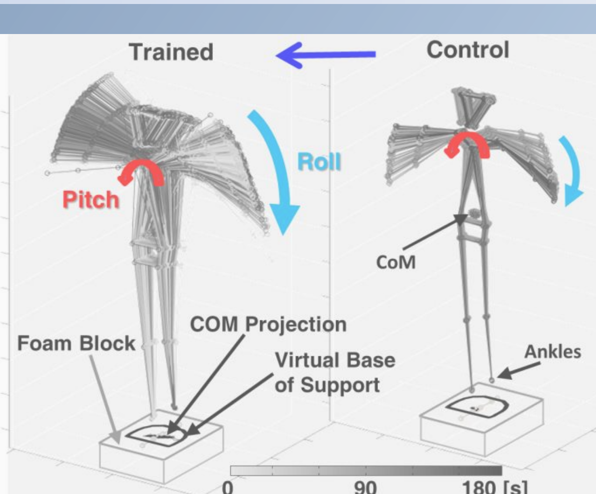


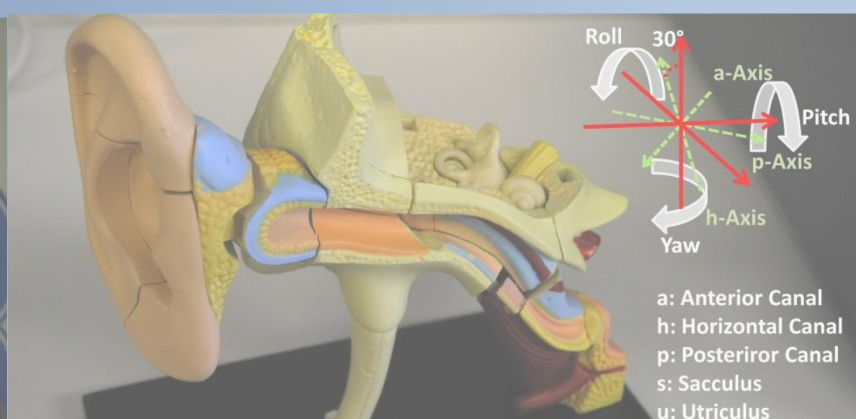


Head and trunk movement strategies in quiet stance

From the deficit of vestibular loss to the expertise of tightrope walkers via prosthetic feedback







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