Postural control and adaptation to threats to balance stability

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The defence of this Doctors Thesis in Medical Sciences will be on Saturday the 16th May, 2009 at Belfragesalen (D15), The Biomedical Centre, Lund University at 9:00am. All are welcome to attend.

Opponent: Professor Floris Wuyts
Dept of Otorhinolaryngology, University Hospital Antwerp, Belgium
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POSTURAL CONTROL AND ADAPTATION TO THREATS TO BALANCE STABILITY

Abstract

Postural control is the ability to maintain equilibrium and orientation in a gravitational environment. The control mechanism is dependent on feedback and feedforward mechanisms, involving visual, vestibular, and proprioceptive sensory systems, that generate appropriate corrective movements evidenced as body-sway. Since information from the various senses is not always accurate (e.g., diseased states) or available (e.g., with eyes closed), the postural control system must adapt to maintain a stable stance. This thesis aimed to investigate postural control and adaptation to threats to balance stability.

Effective approaches for the clinical measurement of postural control still remains elusive. In the past, it has been common to investigate patients' balance by having them stand upon compliant foam blocks with eyes open and closed since standing on foam is believed to affect the accuracy of information from cutaneous mechanoreceptors on the soles of the feet. However, when assessing balance on foam blocks with different compliances and mechanical properties, it was found that postural sway was larger on firmer compliant surfaces, which also increased the importance of visual information.

Postural adaptation was also investigated by repeatedly perturbing balance using muscle vibration. In healthy young persons, adaptation was observed over this time. This adaptation involved decreased costs of standing in terms of decreased energy, body movement and muscle activity and changes to the relationship between muscle activity and movement. The characteristics of the adaptation responses also depended on the availability of visual information. The elderly had poor postural control with and without being perturbed but were able to adapt to improve their poor balance. However, decreased mechanoreceptive sensation in the elderly prevented them from adapting their balance to the level of younger test subjects. Sleep deprivation decreased attention and alertness and resulted in decreased postural control and adaptation.

The findings in this thesis extend what is known about motor learning. The adaptive learning capability of the postural control system, and hence the accurate reconstruction of the kinematics and kinetics of movement, was dependent on one's own mechanoreceptive somatosensation and availability of visual information. Decreasing attention and alertness through sleep deprivation decreased adaptive capabilities, suggesting an important role for sleep in learning a new motor skill.

Key words: Postural Control, Adaptation, Visual, Vestibular, Somatosensory, Proprioception, Posturography, Compliant surface, Ageing, Mechanoreceptive sensation, Sleep deprivation, Muscle activity

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Front cover picture drawn by me. The drawing shows a tumble in progress. Incorporated in this drawing is the concept that the body is multi-segmented. Additionally, the illustration presents a basic overview of the balance sensors.

“The most beautiful thing we can experience is the mysterious. It is the source of all true art and all science. To whom this emotion is a stranger, who can no longer pause to wonder and stand rapt in awe, is as good as dead” – **Albert Einstein**

“Give me a lever long enough and a fulcrum on which to place it, and I shall move the world” – **Archimedes**

To my family

Contents

Definitions

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

Alertness: Ones state of arousal/awakeness i.e., being highly alert is to be watchful and prompt to meet danger or emergency, or being quick to perceive and act.

Angular position: Angular position values provide information about the forward body leaning using the ankle joint as a reference. Body posture can be quantified when the angular positions from the joints along the side of the body are considered together.

Attention: The cognitive process of selectively concentrating on one aspect of the environment.

Consolidation: The process by which recent memories (short-term memories) are converted into long-term memory.

Electromyography (EMG): A technique for detecting the electrical potential generated by muscle cells to make them contract, and when they are at rest.

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

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

Linear movement: Linear movement values provide information about how much the body has moved in each plane e.g., anteroposterior, lateral and vertical.

Postural control: The management of balance.

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

Proprioception: The awareness of the load, position and movement of one part of one's own body compared to another from sensory receptors known as proprioceptors.

Root Mean Square: The square root of the arithmetic mean of the squares of the numbers in a given set of numbers.

Strategy: A plan, method, or series of manoeuvres for obtaining a specific goal or result.

Torque variance: In this thesis, the parameter represents the energy used towards the support surface to maintain stability.

Variance: The square of the standard deviation in a given set of numbers, showing the variation from the arithmetic mean.

List of Publications

The following articles form the basis from which this thesis is written

I. The effect of foam surface properties on postural stability assessment while standing. Patel M, Fransson PA, Lush D, Gomez S. Gait & Posture 2008; 28 (4): 649-656.

II. The contribution of mechanoreceptive sensation on stability and adaptation in the young and elderly. Patel M, Magnusson M, Kristinsdottir E, Fransson PA. European Journal of Applied Physiology 2009; 105 (2): 167-173.

III. The effects of ageing on adaptation during vibratory stimulation of the calf and neck muscles. Patel M, Magnusson M, Fransson PA. Gerontology 2009; 55 (1): 82-91.

IV. Effects of 24-h and 36-h sleep deprivation on human postural control and adaptation. Patel M. Gomez S, Berg S, Almbladh P, Lindbladh J, Petersen H, Magnusson M, Johansson R, Fransson PA. Experimental Brain Research 2008; 185(2): 165-173.

V. Effects of proprioceptive vibratory stimulation on body movement at 24-h and 36-h of sleep deprivation. Gomez S, Patel M, Berg S, Magnusson M, Johansson R, Fransson PA. Clinical Neurophysiology 2008; 119(3):617-625.

VI. Adaptation and vision change the relationship between muscle activity of the lower limbs and body movement during human balance perturbations. Patel M, Gomez S, Lush D, Fransson PA. Clinical Neurophysiology 2009; 120 (3): 601-609.

Other Publications

VII. Changes in multi-segmented body movements and EMG activity while standing on firm and foam support surfaces. Fransson PA, Gomez S, Patel M, Johansson L. European Journal of Applied Physiology 2007;101(1): 81-89.

VIII. The effects of foam surface properties on standing body movement. Patel M, Fransson PA, Lush D, Petersen H, Magnusson M, Johansson R, Gomez S Acta Otolaryngologica 2008; 128(9): 952-960.

IX. Effects of 24-hour and 36-hour sleep deprivation on smooth pursuit and saccadic eye movements. Fransson PA, Patel M, Magnusson M, Berg S, Almbladh P, Gomez S. Journal of Vestibular Research 2008; 18(4): 209-222.

X. Differences in postural adaptation to calf and neck muscle vibratory stimulation. Gomez S, Patel M, Magnusson M, Johansson L, Einarsson EJ, Fransson PA. In Press Gait and Posture.

XI. Effects of dyslexia on human postural control. Patel M, Magnusson M, Lush D, Gomez S, Fransson PA. Submitted for publication.

XII. Change of body movement coordination with cervical proprioceptive disturbances with increased age. Patel M, Fransson PA, Karlberg M, Malmstrom EM, Magnusson M. Submitted for publication.

XIII. Tiredness decreases vibration perception on the soles of the feet. Patel M, Fransson PA. Submitted for publication

Thesis at a glance

Study	Question	Methods	Results	Conclusions
I	What is the effect of changing the compliance and mechanical properties of the supporting surface on human postural control?	Torque variance between the feet and support surface was measured on three different foam blocks with different mechanical properties.	Torque variance increased with firmer compliant surfaces.	The foam surface properties affect recorded balance.
II	How does mechanoreceptive information contribute to postural stability during balance perturbations?	Tactile perception and vibration sensitivity thresholds and torque variance during calf vibration were measured in young and elderly.	There was strong correlation between mechanoreceptive sensitivity and torque variance during vibration.	Mechanoreceptive sensitivity contributes to how well one can adapt to balance perturbations.
III	How does ageing affect human postural adaptation?	Torque variance during calf or neck vibration was measured in elderly and younger subjects.	The elderly showed evidence of adaptation in both the anterior-posterior and lateral directions.	Ageing does not compromise adaptive mechanisms, though the elderly are not able to adapt their balance to the same levels as the young.
IV	Do 24 or 36 hours of sleep deprivation affect human postural control and adaptation?	Torque variance during calf vibration was measured after a normal night of sleep and at 24 and 36 hours of sleep deprivation.	Sleep deprivation increased torque variance during perturbations and decreased adaptation.	Sleep deprivation increases the chances of falling and decreases learning associated with postural control.
V	How do 24 or 36 hours of sleep deprivation affect body movement and adaptation?	Body movement variance during calf vibration was measured after a normal night of sleep and at 24 and 36 hours of sleep deprivation.	Sleep deprivation decreased adaptation and promoted a breakdown of the postural strategy.	Sleep deprivation increases body movement and prevents the selection of the appropriate strategy to repeated balance perturbations.
VI	How are the tibialis anterior and gastrocnemius postural muscles involved in the control of body movement during postural adaptation?	Electromyography (EMG), body movement and torque variance were recorded during calf vibration. The correlation between EMG and movement parameters were then determined.	Vision and repeated exposure to perturbation changed the correlation between postural muscle activity and movement.	The relationship between postural muscle activity and movement is changed through adaptation.

Introduction

Upright standing is one of humankind's most important evolutionary accomplishments. A steady upright posture provides a reliable platform from which movements can be launched such as walking, running or turning. It also allows us to reach elevated levels, see over tall grasses and objects, free the upper limbs for carrying and throwing and permit physical dominance over other species to name a few advantages.

The basic physiological explanation of how humans are able to maintain standing stability has changed little for nearly a century [1]. At the beginning of the 1900s, Sherrington established that corrective postural movements were initiated by reflexive mechanisms in the spinal cord and brain stem [2]. Sherrington also suggested that the reflexes were tonic and adjusted the postural configuration by attitudinal reflexes (i.e., reflexes originating at different segments of the body) and restored disturbances of normal posture by righting reflexes [2]. However, despite this seemingly simple mechanism, falls are common and the outcome can sometimes be severe injury or even death. Therefore, research is essential to help us better understand the act of standing. This can in turn assist in the development of techniques to limit the number of falls, especially in those who are particularly susceptible, such as those with musculoskeletal or neurological disorders or simply from ageing. One of the main features of standing is being able to adapt to different environments. By being able to adapt, one can compensate for balance threats or neurophysiological or musculoskeletal disorder affecting balance. This thesis aimed to explore postural control and adaptation to threats of balance i.e., when the standing surface is compliant, through ageing and when sleep deprived, and explore some of the underlying physiological changes accompanying adaptive motor behaviour. These balance threatening conditions were selected as each is of general public interest. Falls can be costly in terms of health, lost earnings and health care costs and the possibility of slipping on compliant surfaces such as grass, sand or soil, due to age or when sleepy affects numerous people every day.

Postural Control

During standing, humans are not still but are constantly in a sway-like motion. This is because we are long structures balanced on a small base at the feet. In addition, we have nearly two thirds of our body weight above the waist, locating the whole body centre of mass a short distance in front of the ankle joints when we stand [3]. Gravity acts on the unevenly loaded body, attempting to topple the standing person. However, in healthy people, falling does not occur due to a number of control mechanisms continuously in operation.

The muscles in the calf, the gastrocnemius and the soleus, actively oppose the forward toppling effect of gravity. When these muscles are activated, the forces they generate rotate the body against the pull of gravity upwards and backwards, i.e., towards the upright. However, producing the right amount of force to perfectly balance the body at the right time is near enough impossible, which causes a continuous cycle of corrections collectively termed postural sway, maintained through a homeostatic system known as postural control.

The postural control process (figure 1) begins by sensing postural sway using continually integrated information from the visual, vestibular and somatosensory receptors i.e., the sensory receptors. The central nervous system (CNS) is then able to determine body position, sites of instability and movement [4, 5]. The corrective responses are regulated through complex feedback and feedforward mechanisms, which are expressed through the musculoskeletal system and evidenced as tonic changes to postural muscles such as the soleus, gastrocnemius and tibialis anterior. These responses usually keep the body within a small zone of movement when standing [6].

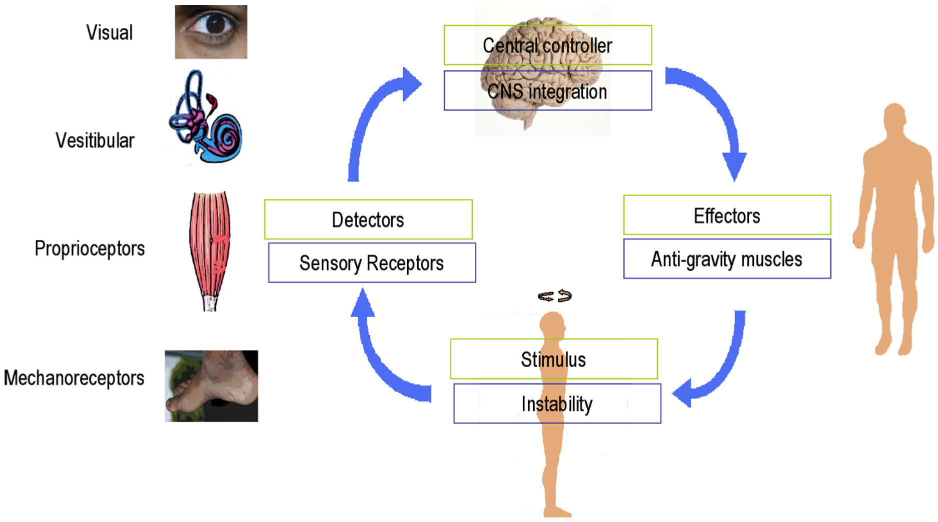


Figure 1: Illustration of the feedback cycle for human standing postural control.

The Sensory Receptors

The Somatosensory System

The somatosensory system plays a significant role in postural control and comprises two different types of receptors: the mechanoreceptors involved in the sensation of touch, pressure and vibration, and proprioceptors which sense the positions of the muscles, tendons, ligaments and joints during static postures and movement. The mechanoreceptors and proprioceptors therefore contribute, with the other sensory receptors, to the precise detection of body position.

The mechanoreceptors are located at various depths of skin and respond to physical stimuli, such as mechanical pressure or movement. Those at the surface, known as cutaneous mechanoreceptors, have small accurate receptive fields, and are located at areas requiring accurate sensations such as the soles of the feet and ankles which are both involved in the maintenance of postural control. There are four different types of mechanoreceptors on the soles of the feet. Merkel's Disks and Ruffini-like Endings are slowly adapting, and detect touch, pressure and stretch of movements below a frequency of 5Hz whereas the Meissner Corpuscles and Pacinian Corpuscles are rapidly adapting and detect the rapid changes of movement necessary for the control of standing [7, 8].

Proprioceptors are found in muscles, tendons and joints and include muscle spindles, Golgi tendon organs (GTOs) and joint receptors. Muscle spindles are located within skeletal muscles and continually sense muscle length. Their density is greater in muscles requiring precise movements, such as the finger muscles and the deep cervical muscles. The firing frequency of the neurones linking the muscle to the CNS changes with muscle length, therefore muscle spindles can dynamically transmit any stretch or contraction of the muscle and its velocity. GTOs are located within tendons and are sensitive to stretch. GTOs are only responsive to large changes in length and are mainly concerned with signalling to the CNS how hard the connecting muscle is contracting. Joint receptors are located around the connective tissue of the joints and function like low threshold mechanoreceptors, responding to physical stimuli, such as mechanical pressure or movement, created by joint movements. Most proprioceptors are capable of detecting changes up to a frequency of 200 Hz. It has been suggested that proprioception is the primary sense responsible for maintaining static standing balance and for triggering automatic balance responses during perturbations, such as a sudden horizontal surface displacement [7, 9].

The Visual System

The visual system comprises light sensitive photoreceptors which detect stationary objects or movement in the visual field. The role of visual information in postural control can be assessed experimentally in numerous ways including closing the eyes, standing in complete darkness, altering visual acuity, changing the size of the visual field and moving the visual surrounds [10]. These methods have revealed that vision is important for accurate postural control as the processed information allows us to anticipate any oncoming obstacles or hazards so that posture can change accordingly (i.e., feedforward control). Additionally, precise visual information is important for the awareness of body position and the perception of self-motion [11], which aids in the fine-tuning of postural responses [12, 13]. Vision is also thought to play a significant role in the detection and correction of head movement, especially slow movement (i.e., between 0.01-0.1Hz) [14], which may trigger pre-programmed postural strategies to reduce postural sway [15]. Visual information can provide a frame of reference for movement which reduces the amount of fast, high frequency motion (e.g., > 0.1Hz) [13].

The Vestibular System

The vestibular system in the inner ear includes the otoliths which are gravitational sensors that detect the position of the head during linear movement, and three semicircular canals (anterior, posterior and lateral) known as the labyrinths, which detect the three dimensional position of the head during angular movement [16]. These receptors are ideally situated to detect changes to the position of the head. Vestibular information has an important role in the maintenance of postural control [17, 18].

The Balance Reflexes

The maintenance of standing balance requires the postural muscles of the body to contract in response to sensations of movement. In part, human postural control is maintained by several kinds of reflexes.

When stretched, a muscle can respond with a rapid contraction and this is known as a stretch reflex. The muscle stretch is signalled to the CNS by muscle spindles and in response the CNS initiates contraction of the skeletal muscles opposite to the muscle stretch. This reflex aims to keep the muscle length constant and is important when an external balance perturbation threatens postural control.

The reticulo-spinal tracts form a direct pathway between the reticular formation in the brain and the spinal motor neurones. Many of the reflexes transmitted by the reticulo-spinal tracts are important for the maintenance of postural control.

The vestibular nuclei in the CNS have an important role in processing motion information and the balance reflexes. These reflexes are grouped into three main categories: vestibulo-ocular reflexes for controlling eye movements, vestibulo-collic reflexes which control head and neck position and movement, and vestibulo-spinal reflexes which control the position and movement of the limbs and trunk. To compensate for movement of the head, the vestibulo-ocular reflex (VOR) is involved in controlling the muscles of the eyes to either contract or relax in just the right way so that the eyes move in exactly the opposite direction to the head movement [19-21]. This helps to keep the object of interest on the fovea of the eyes and focussed while the head is in motion. The vestibulo-collic reflex keeps the head and body aligned and is a two-stage process. Initially, there is contraction or relaxation of the neck muscles to oppose gravitational forces and keep the head level, steady and upright on the shoulders. Secondly, there is a reflexive change of the body position relative to the head. Hence, the vestibulo-collic reflex ensures that when the head position is equilibrated, the rest of the body will follow [19, 20]. The vestibulo-spinal reflexes help to keep the body upright and prevent falls when the body is unexpectedly knock off balance and there is a sudden head movement. This reflex tends to relax groups of muscles on one side of the body and contract similar groups on the other side [19, 20, 22].

Sensory Integration and the CNS

The integration of information from the sensory receptors produces an internal representation (i.e., an internal model) of the position and movement of one's own body (kinaesthesia). This occurs in several parts of the brain important for motor reflexes. The integration process requires unconscious attention [23-25] especially when information from any of the sensory systems is unreliable [26, 27]. When such instances arise (e.g., through disease or novel conditions such as turning off a light in the dark), the CNS relies on information from the other receptor systems to compensate. This is possible because the sensory systems have partially overlapping functional ranges [28]. In fact, the sensory systems converge anatomically, physiologically, and functionally in regions such as the vestibular nuclei, cerebellum, several areas of the cerebral cortex, thalamus, brain stem and spinal cord, which allows opportunities for many interactions [29]. In the CNS, a higher level sensory organisation or integration system is used to adjust the weighting on the different sensory inputs, termed re-weighting [30]. However, when the information from two or more systems is unreliable causing conflicts between information provided by the different senses, the CNS often has difficulties in compensating for the sensory deficit and the result is often an increase in postural sway, dizziness and nausea amongst others.

The cerebellum is one destination where the sensory information is integrated and the internal representation of the body mechanics computed [31-34]. The cerebellum is central to the coordination and smoothness of the balance response [35] possibly by modulating the timing and amplitude of muscle activity [36]. These responses are learnt and stored through repetition. The basal ganglia are another destination for sensory integration. They are important in the planning, control and initiation of motor programs [36].

Body Mechanics

The standing position is often described in terms of an inverted pendulum where the feet are fixed in position and the head is free to move. However, the movement of the human body cannot always simply be described in terms of one single-link between the ankle and head (as in a pendulum). The body is multi-segmented with a number of joints where rotation can occur [37]. The movement of each segment is restricted by the flexibility in muscles, joints and tendons [38]. However, due to the high coordination between the body segments, the CNS only has to alter the movement of one segment to achieve a global change of posture [39].

Problems with Human Stance

There are a number of medical conditions that can affect the postural control system leading to reduced postural competence and an increased risk of falling. Problems with human standing can arise from a conflict between the various sensory inputs or from decreased sensory information, interrupted sensory integration or damage to brain structures or neuromuscular or musculoskeletal systems associated with motor control. The CNS expects a certain relationship between the sensory receptors so that if there is a sensory mismatch, it takes some time before the CNS is updated with the most accurate sensory information [19, 32]. This is likely to depend on specific training of the sensory systems [19]. Therefore, by experimentally affecting the accuracy of sensory information in healthy people, it is possible to further our understanding of the conditions that affect balance in a controlled way.

Balance performance is significantly poorer in individuals with profound visual deficits [40]. The information from the visual receptors is of such importance that postural sway is larger when standing is perturbed in complete darkness compared with no visual information with eyes closed [41]. However, patients with visual impairment can perform equally well as healthy persons with eyes closed [42], or when there is a surface perturbation using compliant foam [43].

Unilateral destruction of the vestibular organ in humans generates a variety of disturbing symptoms dominated by dizziness, nausea and a loss of balance [44].

However, the symptoms generally vanish out with time through compensation, mainly due to central excitability changes that level out the unequal inputs from the two sides [19] i.e., there is a shift in vestibular dependence to the unaffected side. Patients with bilateral vestibular loss (BVL) are highly dependent on visual information [45]. Furthermore, when they have their eyes closed, BVL patients are able to balance but derive their information about self-orientation primarily from the high threshold mechanoreceptors on the soles of the feet in contact with the support surface [46].

Ageing results in a general decline of proprioception [47] and vibration sensation in the lower limbs [48, 49]. Furthermore, when the somatosensory information from the lower legs is decreased by neuropathy, static and dynamic balance is impaired and under dynamic conditions patients can easily lose balance with their eyes closed [50].

Balance Assessment Methods

Posturography is a clinical test commonly used for assessing postural control. The method is non-invasive. A subject stands upon a force platform comprising pressure or force sensors that detect force variations which can be converted into parameters such as centre of pressure (CoP), sway velocity, sway path or torque variance. Posturography is mainly used clinically to assess the balance of patients susceptible to falls, such as those with vestibular deficits [51, 52], elderly or those with poor somatosensory function e.g., diabetics. Posturography differs from most other sensory function assessments because it assess the actual outcome i.e., the balance, rather than attempting to assess peripheral or central function directly [53].

However, posturographic testing of normal, quiet standing patients often lacks the sensitivity for distinguishing between normal and abnormal patterns [38] due to the ability of a person to re-weight unreliable sensory information. Therefore, different balance challenges involving various disturbing techniques have been introduced to help increase the sensitivity of the tests. Furthermore, the balance disturbance can be selected to target the function of a specific sensory receptor to help in the diagnosis of a particular disorder. Some have also modified the perturbation paradigms by affecting more than one sensory receptor at a time to examine the abilities of subjects to make use of the available sensory information and the effectiveness of sensory re-weighting to maintain balance [54]. Thus, the ability to adapt to sensory mismatch is assessed. Also, by changing the sensory stimulus or the location of stimulus, different balance responses and adaptations may be expected.

Some existing posturographic techniques include Computerised Dynamic Posturography (CDP) which is the foundation for EquiTest® platform and the Sensory Organization Test (SOT). In the case of CDP, the force platform is mechanised so that it can either move in a linear horizontal plane, or rotate forwards and backwards [53]. The software allows for two types of test: (1) recording of responses to small brief movements of the support surface, either translations or rotations; and (2) recording of postural sway during different combinations of sensory inputs. These two tests are known as the motor control test and the SOT respectively [51]. Another technique is to decrease the certainty of the standing surface by having patients stand on a compliant foam block [55]. The somatosensory system can also be perturbed by vibration of the postural muscles such as calf or neck muscles.

Movement at different body locations can be recorded to study balance. This approach can be used to analyse movement of the body segments, which due to the various joints allowing rotation (e.g., ankles, knees, hip, shoulders and neck), can differ substantially when a balance perturbing technique is used. This information might be important in the evaluation of the type, severity, or rehabilitation status of a disorder [56]. Movement information can also be combined with other techniques such as electromyography (EMG) which records the activity from selected muscles. In many postural studies, EMG activity of the lower legs is recorded from the muscles responsible for the control of the ankle joints i.e., the tibialis anterior, soleus and gastrocnemius.

The effect of vibratory proprioceptive stimulation on postural control

In experiments where the somatosensory system has been disturbed, the characteristics of the human standing posture have been altered and the amplitude of movement has increased markedly, indicating the vital role that proprioception plays in human postural control [57-59]. One way of disturbing the proprioceptive system is by vibrating skeletal muscles or tendons, including those of the calf or neck [60-62] which increases anteroposterior [57, 63, 64] (forward and backward) and often lateral (side to side) postural sway. Vibration of skeletal muscles or tendons increases activity in muscle spindle afferents. The increased activity is sent to the CNS, creating the illusion of movement [65, 66]. In response, spinally mediated, tonic stretch reflexes are initiated [67] to resist the apparent movement [68]. Due to the interconnected nature of the various segments of the human body, muscle vibration results not only in local postural changes to the vibration site [69], but also in a widespread alteration of segmental and joint orientations remote from the vibrated site [70, 71].

Adaptation of postural control is evidenced when muscle vibrations are repeated, as the movement responses usually decrease significantly [18, 63] and the segmental body movements and body posture change to enhance stability [15, 72, 73].

Adaptation and postural control

Balance correcting responses are estimated to occur about 370 milliseconds after a balance perturbation [1]. This means that the CNS has time to modulate the amplitude and timing of the corrective muscle activity. The learning network comprising brain areas such as the cerebellum, basal ganglia [74] and hippocampus, basolateral amygdale and striatum [75, 76] is involved when standing balance is disturbed, allowing adaptation to perturbations. However, damage to the cerebellum nearly always impairs motor adaptation [77].

The adaptive control of balance is an important feature [78] as it decreases instability and the likelihood of falling [79, 80] by making the postural challenge more manageable [81]. Within the first few cycles or seconds of perturbation in experimental set-ups, a person predicts the characteristics of perturbations and their destabilising effects, and sets their balance control system to minimise these effects [82]. Additionally, when a perturbation is repeated, the postural movements are often optimised through the fine-tuning of motor responses [80] and sensory reorganisation [83]. Therefore, postural responses may change over time to a repeated perturbation. During adaptation, the brain alters the movement pattern to minimise the costs associated with a repeated perturbation [33, 84]. Costs might include energy demands, forces, fatigue, inaccuracy, jerkiness etc [77]. For any situation, the CNS will decide which costs are the most important to reduce in order to achieve the goal of the task [77].

Adaptation of balance is heavily dependent on an accurate internal model of body position and prediction of a loss of balance [32]. Accurate internal models decrease the reliance on unreliable sensory information [77, 85] and are thus vital in restoring appropriate function during balance disorders. In addition, it means that when certain exercises are repeated on a daily or weekly basis that compromise balance but does not induce falling, balance control can be improved through learning and the formation of a new, hard-wired, motor strategy [86]. It is possible to see significant rehabilitative effects from after-effects following adaptation training exercises [87]. Even when the after-effects from a single adaptation training exercise do not yield long lasting effects, the process is still extremely important for rehabilitation, because identifying adaptation allows us to determine whether the postural control system is still capable of an improved movement pattern [77].

Conditions under Investigation

The effect of standing on a foam block on postural control

One way of perturbing balance in a clinical environment is to have patients stand on a compliant surface such as foam [55]. Standing on foam is believed to affect the accuracy of information from cutaneous mechanoreceptors on the soles of the feet and ankle joints [88]. These mechanoreceptors sense changes to weight distribution [64, 89] and body orientation [90, 91] through the detection of forces on the skin [92, 93]. The input to the mechanoreceptors on the soles of the feet is affected more by standing on soft foam surfaces than on firm foam surfaces [88]. However, it is difficult to determine the exact degree mechanoreceptive information is affected [94]. In addition, one effect of standing on foam surfaces not normally considered is that it involves a viscoelastic surface, and this reduces the effectiveness of ankle torque for postural stabilisation [28, 95].

The effect of ageing on postural control

It is well accepted that one of the characteristics of ageing includes postural instability [96]. As age increases, the incidence of falls also increases [97], and it has been estimated that total fall related costs could exceed $32 billion in the US alone by the year 2020 [98]. In the UK, around 30% of people aged 65 years or older living in the community and 50% of those living in residential care facilities or nursing homes fall every year, and about half of those who do fall, do so repeatedly [99]. It is therefore essential that predisposing factors associated with falls be captured early in order to begin administering simple, low-cost therapies. One way to do this is to determine the mechanisms related to poor balance control in the elderly. Studies have showed that the deficits in postural control in the elderly can been attributed to a number of causes including deterioration of the sensory and motor systems. For example, ageing causes deterioration of the visual system [100, 101], proprioception from the lower legs [102], somatosensory information from the soles of the feet [103, 104], vestibular system [105] and reduces muscle strength of the lower legs [106]. Ageing has also been showed to affect postural control in cognitive or attention-demanding tasks. Hence, the age-related changes to the sensory and motor systems appear to increase the requirement of cognitive or attention resources for sensory-motor activity [107].

The effect of sleep deprivation on postural control

Sleep provides the only time that the cerebral cortex can relax and recover [108]. Other organs in the body can do this when the body is in a state of relaxed wakefulness [108]. Without the necessary amount of sleep, normal brain functioning is decreased and tiredness, lethargy and irritability ensue. These characteristics of sleep loss may account for a large number of accidents. Studies have suggested that sleepiness causes about 20% of the serious motor accidents in the UK [109], and that staying awake for 18 hours equates to the same level of impaired performance as driving while intoxicated legal alcohol limits in the UK and US of 0.08% Blood Alcohol Concentration (BAC) (0.8 g/L) [109, 110].

Functional magnetic resonance imaging has showed that sleep deprived individuals express greater levels of brain activity in the prefrontal cortex during cognitive tasks which suggests that it takes more 'brain power' for simple tasks. Also, the temporal lobe (important for language processing) and parietal lobe (responsible for memory) show less activity in sleep-deprived individuals.

Over a hundred years ago, Patrick and Gilbert first reported that sleep deprivation decreases levels of attention and alertness [111]. These attention and alertness deficits following sleep deprivation have been attributed to a decreased activity in certain cerebral areas such as the pre-frontal cortex, thalamus and reticular activating system [112] and as the duration of sleep deprivation extends from 24 to 48 hours, these areas undergo further deactivation [113].

Postural control is an attention-demanding task. As previous postural studies have shown that the attentional requirement increases when the information from at least one sensory source becomes unreliable, postural control may be significantly affected by sleep deprivation when balance is perturbed [23, 114]. Furthermore, postural control is also thought to follow a circadian cycle associated with levels of alertness [115, 116].

Aims of this Thesis

The overall aim of this thesis was to explore adaptation and postural control to threats to balance stability: when the surface is compliant (i.e., on foam); when there is multiple system deterioration through ageing; and when attention and alertness levels are low following a lack of sleep. It was also important to give a detailed view of the changes during adaptation.

The aims of each study were to:

- Investigate how foam surfaces of different compliances affect postural control and hence whether the foam surface properties can affect that clinical test findings **(Study I)**.

- Determine the effects of poor mechanoreceptive sensation on the soles of the feet from ageing on postural control and adaptation **(Study II)**.

- Determine the effects of ageing on postural control and adaptation **(Study III)**.

- Determine the effects of 24 hours and 36 hours of sleep deprivation of human postural control and adaptation **(Studies IV and V)**.

- Investigate the relationship between muscle activity from the lower leg postural muscles and movement during adaptation in young healthy human **(Study VI)**.

Materials and Methods

Measuring Torque (Studies I, II, III, IV and VI)

Torque was measured using a customised force platform containing 6 pressure sensors at 50Hz with an accuracy of 0.5N. Torques are forces produced by a rotational movement around a central rotational axis. In the presented studies, the rotational forces are actuated from the feet and body towards the support surface. The anteroposterior and lateral axis are level with the force platform surface (see figure 3).

Torque τ can be calculated from the formula $\tau = CoP \cdot Fz$; where CoP is centre of pressure (in metres) with distances in relation to the rotational axes and $Fz \approx m \cdot g$; where m = the assessed subjects mass (in kg) and g = gravitational constant 9.81 (in metre/s^2). Fz (vertical forces) will fluctuate slightly due to body leaning or when the subject applies additional force to the surface to accelerate/decelerate a movement. The advantage of using torque instead of CoP to represent stability is that the variance of torque directly corresponds to the amount of energy used to maintain standing [38, 117].

Torque variance

The variance of the anteroposterior and lateral torques, Mx and My respectively (see figure 3), were calculated using the formula below, in this example calculated for the anteroposterior torque

$$\tau_{AP} = \sum_{i=1}^{n} \frac{\tau_{AP}(i)}{n} \qquad (1)$$

$$\tau(AP)_{\mathrm{var}} = \frac{1}{n-1} \sum_{i=1}^{n} \left(\tau_{AP}(i) - \tau_{AP}\right)^2 \qquad (2)$$

where $\tau(AP)_{\mathrm{var}}$ represents the variance of the anteroposterior torque and τ_{AP} represents the marker's sampled anteroposterior torque under the timed period analysed.

Spectral separation of torque variance (Study I)

The variance of measured torque was divided into three spectral categories: variance of all measured torque denoted “Total”; variance of torque contents below 0.1 Hz “low frequency torque variance”; and variance of torque contents above 0.1 Hz “high frequency torque variance”. A fifth-order digital Finite duration Impulse Response (FIR) filter [118], with filter components selected to avoid aliasing, was used for spectral separation. The torque contents above 0.1 Hz primarily reflect the fast corrective movements used to maintain balance, whereas torque contents below 0.1 Hz can be described as smooth corrective changes to stance. The cut-off frequency was based on the cut off frequencies of the vestibular and visual sensory systems which is about 0.1Hz [48].

Basis for anthropometrical normalisation of torque variance

The torque variance values were normalised to account for the differences in subject anthropometric variations (i.e., height and weight). Due to the biomechanical differences of having more weight and a longer frame, those who are heavier and taller have larger recorded torques and therefore these differences must be corrected for [119]. This is explained by the inverted pendulum model of human postural dynamics (figure 2).

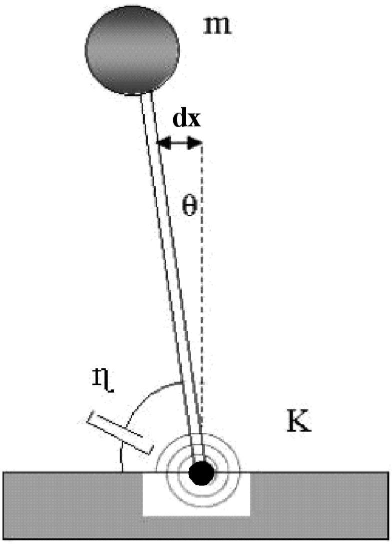


Figure 2: Inverted pendulum model of human postural dynamics with the balancing torque T_{bal} similar to that achievable with a spring (K) and a dashplot (η).

If we assume that the inverted pendulum model is valid, the torques τ recorded by the force platform can be described as

$$\tau(t) = J\frac{d^2\theta}{dt^2} + mgl\sin\theta(t) + T_d(t) \qquad (3)$$

where $J=ml^2$= moment of inertia, m=subjects mass (kg), l = distance to the body's centre of mass (CoM) (metre) which is located at about 55% of the way up the normal standing human, g= gravitational constant (approx 9.81m/s^2), θ = ankle joint angle and $T_d(t)$ is some disturbance torque from the environment or measurement noise.

In formula (3), the first factor explains the torques in dynamic events such as body acceleration and deceleration. The second factor explains the torques in the static standing human. These torques are caused by the leaning position of the body, which is approximately 4 to 4.5 degrees between the ankle and head in normal upright standing.

When formula (3) is re-written, formula (4) shows that the recorded torque is strongly dependent on individuals' mass (m) and height (l). However, as the formula also shows, differences caused by height cannot be completely eliminated through normalisation but are reduced.

$$\tau(t) = ml(l\frac{d^2\theta}{dt^2} + g\sin\theta(t) + \frac{1}{ml}Td(t)) \qquad (4)$$

Since the calculation for variance of torque contains a square element, see formula (2), the square of the height and weight is required to achieve unit agreement when normalising torque variance values.

Measuring Body Movement (Studies V and VI)

Body movement at five anatomical positions were tracked in three dimensions at 50Hz, using an ultrasound motion analysis system (Zebris™ CMS-HS Measuring System for 3D motion analysis) with an accuracy of 0.4mm. The markers were positioned at the cheek bone (zygomaticum), shoulder (tuberculum major), hip (spino-anterior of crista iliaca), knee (lateral epicondyle of the femur) and the anklebone on the right side of the body, facing the Zebris™ system.

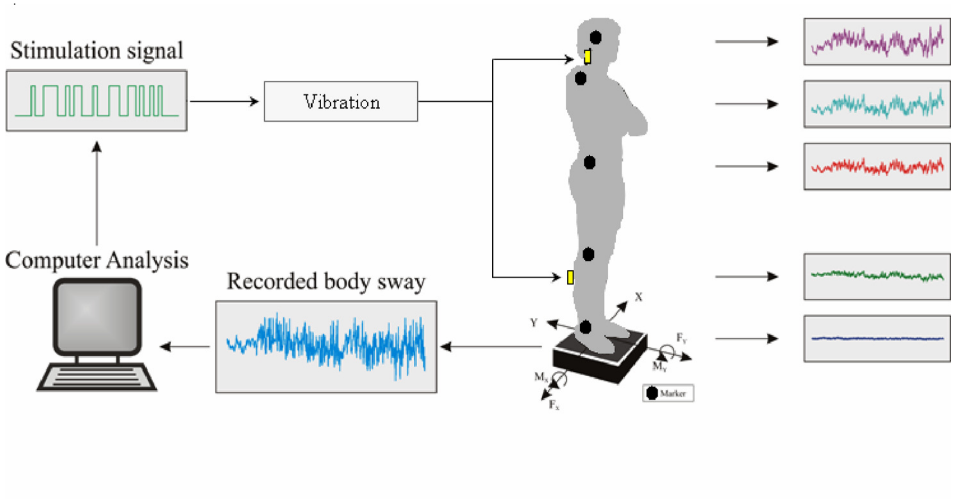


Figure 3: Illustration of the posturographic measurement system. The illustration shows the position of the movement markers in black and examples of the data captured simultaneously by both Zebris and the force platform.

Linear body movement variance

The linear body movement (the amount of movement at the recorded body position), was quantified in terms of movement variance at the head, shoulder, hip and knee, as recorded with the Zebris™ system using the formula below, in this example calculated for the head marker:

$$\bar{x}_{Head} = \sum_{i=1}^{n} \frac{x_{Head}(i)}{n} \tag{5}$$

$$x(Head)_{\mathrm{var}} = \frac{1}{n-1} \sum_{i=1}^{n} \left(x_{Head}(i) - \bar{x}_{Head} \right)^2 \tag{6}$$

where $x(\mathrm{Head})_{\mathrm{var}}$ represents the variance of the linear anteroposterior head movements and $x_{\mathrm{Head}}(i)$ represents the marker's sampled linear anteroposterior position under the time period analysed.

Basis for anthropometrical normalisation of linear movement variance

If the body moves like a single link pendulum, which is the most common pattern of movement, the size of the linear movement will be gradually larger the higher up the body. This is expressed by the formula below:

$$dx = l \sin \theta(t) \qquad (7)$$

where the distance dx will be directly dependent on l (distance to the position) and θ (joint angle) (see figure 2). The human body is fairly proportional. Short people have smaller feet than the tall ones. This allows taller humans to lean as much in degrees as shorter ones. Thus, although the angular movement in degrees with respect to the ankle joint is the same, the size of the linear anteroposterior and lateral movements in millimetres will be affected by subjects' height.

Since the calculation for variance of linear movement contains a square element, see formula (6), the square of the height is required to achieve unit agreement when normalising linear movement variance values.

Eyes open/eyes closed quotients (Study V)

From the Zebris™ linear movement recordings, quotients between eyes closed and eyes open tests (EC/EO) were calculated for each marker position at 24 and 36 hours of sleep deprivation and after a normal night of sleep to show the relative difference in multi-segmented movement with respect to each marker.

Average angular position (Study VI)

Average angular position was calculated from Zebris™ recordings using the ankle marker as the reference point and the linear distance of the marker from the perpendicular with an error <1.5%. The values were used to show the amount of forward leaning i.e., body posture. Average angular position was calculated using the formula below, presented here for the head marker:

$$\overline{x}_{Head} = \sum_{i=1}^{n} \frac{x_{Head}(i)}{n} \qquad \overline{x}_{Ankle} = \sum_{i=1}^{n} \frac{x_{Ankle}(i)}{n} \qquad (8)$$

$$\overline{z}_{Head} = \sum_{i=1}^{n} \frac{z_{Head}(i)}{n} \qquad \overline{z}_{Ankle} = \sum_{i=1}^{n} \frac{z_{Ankle}(i)}{n} \qquad (9)$$

$$x(head)_{ang} = \arcsin\left(\frac{\overline{x}_{Head} - \overline{x}_{Ankle}}{\overline{z}_{Head} - \overline{z}_{Ankle}}\right) \qquad (10)$$

where x(head)ang represents the mean angular position of the head, x represents the marker's sampled anteroposterior position and z the marker's sampled vertical position. The ankle position change in any direction was on average less than 1 mm, with a maximum change of less than 3.5 mm. Change in average ankle position was compensated for in the analysis of angular position, by using the mean position of the ankle for the selected time period.

Measuring Muscle Activity (Study VI)

In addition to recordings of body movement, the Zebris Measuring System also simultaneously recorded Electromyographic (EMG) activity at 1500Hz, from the tibialis anterior and gastrocnemius muscles of both legs using surface electrodes. The recorded EMG activity from the tibialis anterior and gastrocnemius muscles of both legs was band-pass filtered using 20 and 200Hz respectively as cut off frequencies. Gastrocnemius EMG was further notch filtered between 100 and 130Hz to remove any distortion caused by vibratory stimulations of the calf muscles. The Root Mean Square (RMS) value for the filtered muscle activity was calculated according to the formula below

$$x_{RMS} = \sqrt{\frac{1}{n}\sum_{i=1}^{n}(x(i))^2} \qquad (11)$$

where xRMS represents the RMS value for the EMG muscle activity and x(i) represent the recorded muscle activity. In the final step, the averages of the tibialis anterior and gastrocnemius RMS EMG values from both legs were calculated and used in the analysis.

Other Measurments

Measuring the Properties of the Compliant Foam Surfaces (Study I)

In Study I, three foam blocks of different properties placed on the force platform. These were categorised: soft foam (SF); medium foam (MF); firm foam (FF) by their elastic modulus. The density (ρ) of the foams was calculated by dividing the mass of the foam (kg) by the volume of the foam block (metre^3). The volume (V) was calculated from V= foam length (metre) · foam width (metre) · foam height (metre).

The elastic modulus (E), which is the mathematical description of an object or substance's tendency to be deformed elastically (i.e., non-permanently) when a force is applied to it, was calculated using the formula:

$$E = \frac{k \cdot H}{A} \tag{12}$$

where *k* denotes the axial stiffness (N/metres), H (metres) the height of the foam to which the force is applied on top, and A (metres^2) is the cross-sectional area over which the force is applied, in this case A= foam length (metres) · foam width (metre).

The axial stiffness of the foams was calculated after the completion of the study by tracking each foams displacement (metres) against force applied (N) using a designated force-compression measurement system which yielded a displacement versus applied force graph for each foam block. The axial stiffness is the tendency of an object's volume to deform when under pressure. The inverse of axial stiffness is compliance. Therefore, measuring the elastic modulus provides a measurement of the compliance of the foam surfaces used. Force was applied with steps of 2N increments, and care was taken to measure the response to a statically applied force. The load was applied over the entire surface area using a plate that was of the same size as the foam block. Using the resulting force versus displacement graphs, the average axial stiffness value *k* within the force range from 550N to 800N was calculated using the formula:

$$\kappa = \frac{F}{dx} \tag{13}$$

where *k* denotes the axial stiffness (N/metres), F denotes vertical force (N) and *dx* denotes compression of the foam (metres). The force range was selected to cover the range of vertical forces within the test group caused by the subjects weight (N) while standing on the foam.

Foam name	Density (kg/m^3)	Elastic Modulus (N/m^2)
Firm	82.0	49200
Medium	21.3	20900
Soft	21.9	4200

Table 1: The density and elastic modulus of the foam surfaces

Measuring Sensitivity of the Mechanoreceptors on the Soles of the Feet (Study II)

Vibration perception of the plantar surface was measured using a biothesiometer electronic device (Model EG electronic BioThesiometer, Newbury, Ohio, USA) that generated a 120Hz vibration of varying amplitude (in µm). The vibration was applied to the plantar surface of the first distal phalanx (big toe), the fifth distal phalanx (little toe), the first proximal phalanx (base of big toe), the fifth proximal phalanx (base of little toe) and the tuberosity of calcaneous (heel). Subjects were asked to indicate to the examiner whether they were able to feel the vibration “Yes” or “No”. Three readings in ascending intensity and 3 readings in descending intensity were made until the subject could no longer feel the vibration, and an average of these 6 measurements was recorded as the vibration threshold [120].

Tactile sensitivity was measured with a Semmes-Weinstein pressure aesthesiometer (Semmes-Weinstein Monofilaments, San Jose, USA). The aesthesiometer comprised 20 nylon filaments of equal length, with varying diameter. The filaments were applied to the plantar surface of first distal phalanx (big toe), the fifth distal phalanx (little toe) and the tuberosity of calcaneous (heel). Subjects were instructed that when the filament was placed on any of the positions above, the examiner would say “big toe”, “little toe” or “heel”, and if they felt the filament in contact with the skin, they must report to the examiner whether they felt it on the “big toe”, “little toe” or “heel.” Tactile threshold was determined by presenting suprathreshold filaments initially, then applying thinner and thinner filaments until the subject could no longer detect them [120]. Thicker filaments were applied until the filament was detected. The threshold was determined from 3 ascending and descending steps [120] and is presented in the table as monofilament diameter size (mm).

Measuring Signs of Sleep (Studies IV and V)

EEG data were recorded over a period of 24 hours (from 12 hours into testing to 36 hours) using an Embletta® device to ensure that subjects had stayed awake. The Embletta® comprised three electrodes; an active electrode (positioned at the left temple), a reference electrode (forehead) and a ground electrode (right occiput) and recorded alpha wave activity for evidence of alertness. The device was removed prior to posturography testing at 24 SDep and 36 SDep and data stored for analysis.

Measuring Subjective Alertness (Studies IV and V)

Prior to the tests at 24 SDep and 36 SDep, subjects were instructed to give a subjective score of alertness using the Visuo-Analogue sleepiness Scale (VAS) ranging from “completely alert” to “exhausted and near sleep”. The subjects’ analogue scores were converted into numbers ranging from 1 to 10, where 1 = “completely alert” and 10 = “exhausted to near sleep”.

Adaptation Analysis

When using vibratory stimulation (Studies II, III, IV, V and VI) the recordings were split into five periods: Quiet stance (0-30s), Period 1(30-80s), Period 2 (80-130s), Period 3 (130-180s) and Period 4 (180-230s) and values were calculated for each period. In Study I, one value was calculated for the entire 120 second period recorded.

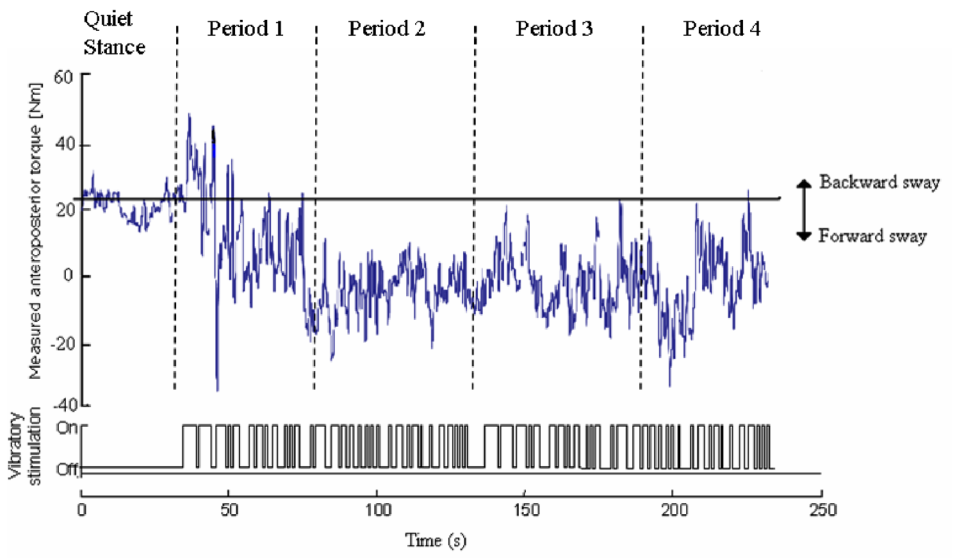


Figure 4: Illustration of an example vibration test. The vibration pulse sequence is pictured at the bottom, and example vertical lines have been put in to show each of the periods used in the calculations.

Measuring adaptation

Two quotients were calculated to quantify adaptive changes in Studies III, IV, V and VI. The first quotient (dividing vibration Period 1 values by Quiet stance values) shows how the assessed parameters were initially affected by the balance perturbations. The second quotient (dividing vibration Period 4 values by vibration Period

1 values) shows how the assessed parameters were maximally affected by repeated balance perturbations.

In Study VI, to investigate whether time-varying changes in EMG activity and body movement parameters were associated with one another for each test condition, correlation analysis was carried out between EMG activity and body movement parameters using 4 quotients. One describing the initial effects of balance perturbations (QS/P1), and three others describing the subsequent adaptive responses (P2/P1, P3/P1, P4/P1).

Testing procedure

Quiet stance posturography procedure (Study I)

Subjects were asked to stand relaxed, with arms folded and feet open at 30 degrees and heels 3cm apart on a force platform or on foam placed upon the force platform. Subjects were instructed to focus on a target at eye level at a distance of 1.5 m, or closed their eyes when instructed. Subjects listened to calm music through headphones to avoid distraction from external noises. All test sequences were randomised. The Quiet stance activity was recorded for 120 seconds.

Vibratory stimulation posturography procedure (Studies II, III, IV, V and VI)

Subjects were asked to stand relaxed, with arms folded and feet open at 30 degrees and heels 3cm apart on a force platform while being perturbed by calf (Studies II, IV, V and VI) or calf or neck (Study III) muscle vibratory stimulation. In all studies subjects focused on a target at eye level at a distance of 1.5 m, or closed their eyes when instructed. Subjects listened to calm music through headphones to avoid distraction from external noises. All test sequences were randomised.

The pulses of stimulation were from 0.8 seconds to 6.4 seconds and were controlled using a Pseudo-Random Binary Sequence (PRBS) schedule lasting 200 seconds. Before the vibratory pulses commenced, a 30 second period of Quiet stance was recorded, making the tests 230 seconds long. The vibrators produced a vibration of 1.0mm amplitude and frequency of 85 Hz generated through a revolving DC motor (Escap, Switzerland). The vibrators were 6cm long and 1cm in diameter and were either strapped bilaterally over the belly of the calf muscles or bilaterally over the paravertebral neck muscles.

Statistical Analysis

In all studies, non-parameteric tests were used because Shapiro-Wilk and Komogorov–Smirnov tests showed that data were not normally distributed. Wilcoxon statistical tests were used to compare between tests (Studies I and V) and investigate any adaptation between Quiet stance and Period 1, and between Period 1 and Period 4 (Studies III, IV, V and VI), where p-values after the Bonferroni correction of < 0.01 were considered statistically significant [121]. A Mann-Whitney non-parametric test was used to determine whether there was a significant difference between groups (Studies II and III).

A GLM Univariate ANOVA (General Linear Model univariate Analysis of Variance) statistical test was used to determine whether the independent variables investigated significantly affected measurements of movement or whether there was an interaction (All Studies). The GLM model accuracy was evaluated by testing the model residual for normal distribution. In the GLM analysis, p-values < 0.05 were considered significant.

The Studies

Study I: The effect of foam surface properties on postural stability assessment while standing

Subjects: Thirty young healthy subjects (nineteen males and eleven female); mean age 22.5 years, range 19-43 years) participated in the study.

Study Design: Eight randomised tests were performed each lasting 120 seconds. The tests were:

- Standing on the solid force platform surface (Solid Surface) with eyes open (EO) and eyes closed (EC)
- Standing on a low compliance foam (Firm Foam) surface with EO and EC
- Standing on a medium compliance foam (Medium Foam) surface with EO and EC
- Standing on a high compliance foam (Soft Foam) surface with EO and EC

Results: Generally, standing on a compliant foam surface significantly increased postural sway, especially with eyes closed (often $p<0.001$). However, postural sway was significantly affected by the properties of the foam surface used, especially rapid movements in the lateral direction ($p<0.05$). Postural sway was generally largest when subjects stood on the Firm Foam and smaller when standing on the other foam surfaces (generally $p<0.001$).

Vision decreased postural sway on all surfaces (generally $p\leq0.001$). However, when standing on foam, the visual contribution was affected by the properties of the surface with a larger decrease in anteroposterior postural sway with eyes open when subjects' stability was increasingly challenged by the support surface ($p<0.05$).

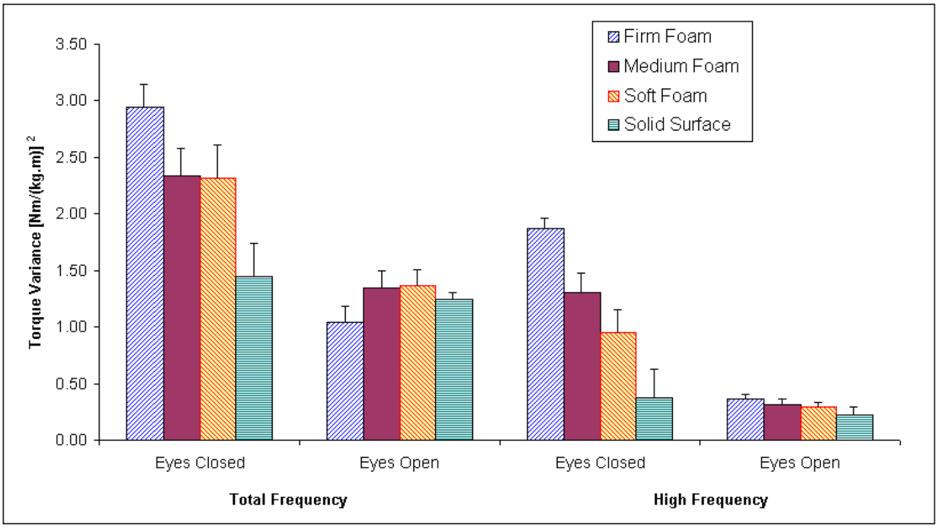


Figure 5: Total and high frequency anteroposterior mean and (standard error of mean (SEM)) torque variance on the different selected foam surfaces and on a solid surface.

Conclusions: Postural control assessment was affected by the properties of the foam surface. The foam surface properties affected the ability to generate sufficient torques for standing stability. A careful choice of foam surface is therefore required for use in clinical balance assessment or rehabilitation training as what might seem like evidence of disorder might be an appropriate response for that foam.

Study II: The contribution of mechanoreceptive sensation on stability and adaptation in the young and elderly

Subjects: Twenty-five young healthy subjects (twelve males and thirteen female; mean age 25.1 years, range 19-41 years) and sixteen Elderly subjects (eleven male & five female; mean age 71.5 years, range 64-79 years) participated in the study.
Study design: Mechanoreceptive sensitivity was measured using vibration threshold and tactile sensitivity tests. In addition, two randomised perturbed balance tests were performed, each lasting 230 seconds. The tests were:

- Calf muscle vibration with eyes closed (Eyes Closed) eyes and open (Eyes Open)

Results: The younger age group displayed better mechanoreceptive sensitivity compared to the elderly ($p<0.001$) except for tactile sensitivity at the heel. Anteroposterior postural sway was larger for the elderly in all vibration periods ($p<0.001$) but not in Quiet Stance. Decreased mechanoreceptive sensation (at most positions and Periods $p<0.05$) and standing without visual information ($p\leq0.011$) significantly increased postural sway in the vibration periods. However, mechanoreceptive sensation significantly affected postural sway more in Periods 2 (80-130s), 3 (130-180s) and 4 (180-230s) than in Period 1.

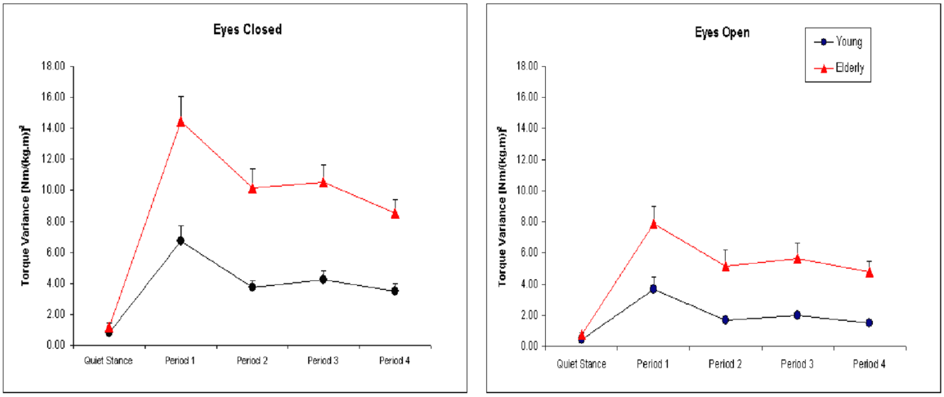


Figure 6: Eyes Closed and Eyes Open mean and SEM torque variance for the elderly and young.

Conclusions: The elderly had poorer mechanoreceptive sensitivity when compared to young healthy adults. Poor mechanoreceptive sensitivity directly contributed to poor balance and poor postural adaptation. Therefore, one major risk factor for falls in the elderly is decreased sensation on the soles of the feet.

Study III: The effects of ageing on adaptation during vibratory stimulation of the calf and neck muscles

Subjects: Eighteen young healthy subjects (nine males and nine female; mean age 29.1 years, range 18-49 years) and sixteen Elderly subjects (eleven male & five female; mean age 71.5 years, range 64-79 years) participated in the study.

Study design: Four randomised tests of perturbed balance were performed on young healthy adults and the elderly, each lasting 230 seconds. The tests were:

- Neck muscle vibration with eyes open (EO-Neck) and eyes closed (EC-Neck)
- Calf muscle vibration with eyes open (EO-Calf) and eyes closed (EC-Calf)

Results: Anteroposterior and lateral postural sway was larger with increasing age during quiet standing ($p \leq 0.001$) and when balance was perturbed by calf or neck muscle vibration ($p \leq 0.002$). However, the elderly were able to adapt their balance to decrease anteroposterior ($p<0.01$) and lateral postural sway ($p<0.01$), except with EC-Neck. The adaptive changes in the young and elderly were proportionally about the same in the anteroposterior direction. In the lateral direction, adaptive adjustments in the elderly were generally proportionally larger compared to the young. However, the elderly were unable to reach the same low levels of postural sway as the young through adaptation in either direction. Vision had a prominent effect on postural sway and adaptation in the young and elderly when balance was perturbed ($p \leq 0.001$).

Conclusions: When balance was repeatedly threatened by perturbations, the elderly were capable of initiating adaptation to the stimulus. Adaptation training therefore offers one technique for balance rehabilitation in the elderly. However, the elderly were unable to adapt their balance to the same levels as the younger age group.

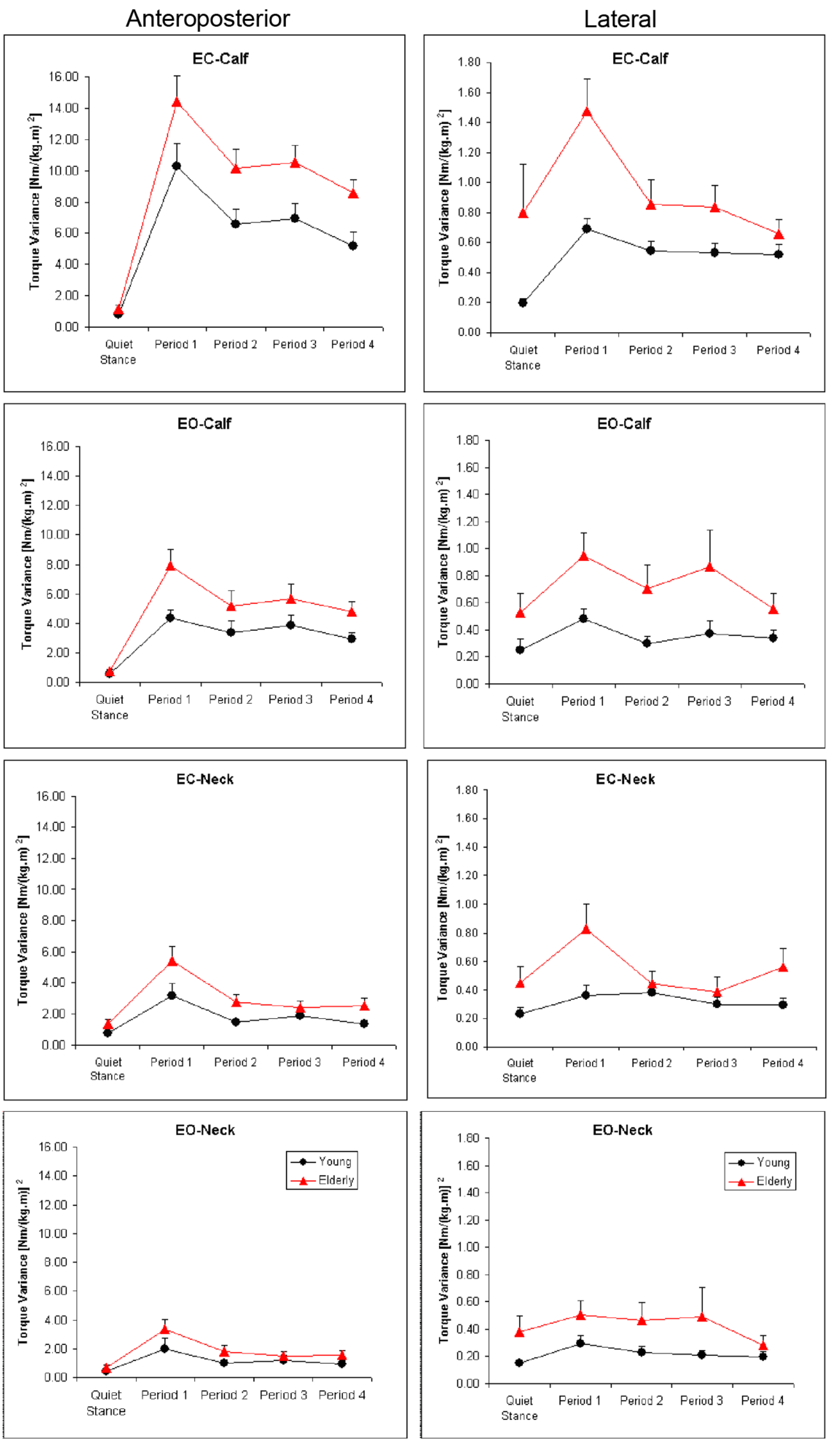


Figure 7: Anteroposterior (left) and lateral (right) mean and SEM torque variance for the elderly and young in all tests.

Study IV: Effects of 24-h and 36-h sleep deprivation on human postural control and adaptation

Subjects: Eighteen healthy subjects (ten males and eight female; mean age 23.8 years, range 16-38 years) participated in the study.
Study design: Six perturbed balance tests were performed by each subject, each lasting 230 seconds. The test following a normal night of sleep (Control) was randomly performed before or after sleep deprivation tests. The tests were:

- Normal night of sleep with eyes open and eyes closed (Control)
- 24 hours of sleep deprivation (24 SDep) with eyes open and eyes closed
- 36 hours of sleep deprivation (36 SDep) with eyes open and eyes closed

Results: EEG recordings ensured that all subjects remained awake for the entire 36 hours. Subjective sleepiness increased from 5.2 at 24 SDep to 6.8 at 36SDep on a VAS scale ranging between 1 and 10 (1 being highly alert and 10 being exhausted).

Sleep deprivation increased postural sway when balance was perturbed (in most Periods, $p<0.05$) and decreased adaptive capabilities. Whereas subjects were able to adapt their balance after a normal night of sleep ($p<0.01$), there was no significant adaptation following sleep deprivation. This was the case both when standing with eyes open or closed. The increase in postural sway was clearest mid-way through testing, between 100 and 150s (Period 3) into vibration. Furthermore, the increased postural sway was clearer at 24 SDep compared with 36 SDep.

Conclusions: Sleep deprivation compromised the central nervous systems ability to compensate for balance threats, i.e., there was decreased postural control and adaptation. Sleep deprivation therefore increases the likelihood of falling and accidents in demanding tasks.

The effects of sleep deprivation were worse at 24 SDep, and thus might follow a circadian rhythm instead of being affected by the duration of sleep deprivation and one's own subjective estimation of sleepiness.

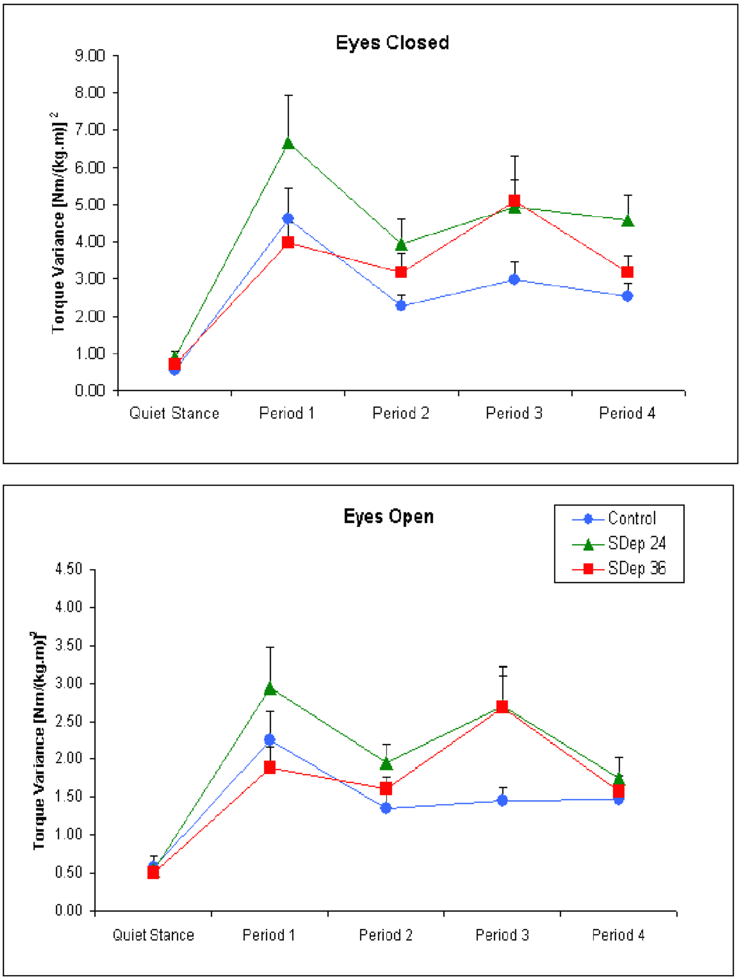


Figure 8: Anteroposterior mean and SEM torque variance for a normal night of sleep (Control), 24 hours of sleep deprivation and 36 hours of sleep deprivation for all periods with Eyes Closed and Eyes Open.

Study V: Effects of proprioceptive vibratory stimulation on body movement at 24-h and 36-h of sleep deprivation

Subjects: Eighteen healthy subjects (ten males and eight female; mean age 23.8 years, range 16-38 years) participated in the study.

Study design: Six tests were performed by each subject, each lasting 230 seconds. The test following a normal night of sleep (Control) was randomly performed before or after sleep deprivation tests. The tests were:

- Normal night of sleep with eyes open and eyes closed (Control)
- 24 hours of sleep deprivation (24 SDep) with eyes open and eyes closed
- 36 hours of sleep deprivation (36 SDep) with eyes open and eyes closed

Results: Sleep deprivation increased anteroposterior whole body movement from 100 to 150s into vibration tests ($p<0.05$). Furthermore, the increased body movement was most apparent at 24 SDep compared with 36 SDep. However, the clearest indication that sleep loss affected postural control was the decreased adaptation to repeated balance perturbations (generally non-significant) compared with a normal night of sleep (generally $p<0.01$), both with eyes open and eyes closed.

Additionally, there were indications that the normal movement pattern was altered by sleep deprivation. At 24 SDep, subjects adopted a more precautious movement pattern. The movement at hip was proportionally less than at the other body segments during Quiet stance ($p<0.05$) and from 50 to 150s (Periods 2 and 3) of balance perturbations ($p<0.05$). There were also indications of this movement pattern between 0-50s of balance perturbations (Period 1), but this was not significant.

Conclusions: Sleep deprivation decreased postural control and adaptive capabilities and at 24 SDep slowed down the ability to select the appropriate movement strategy.

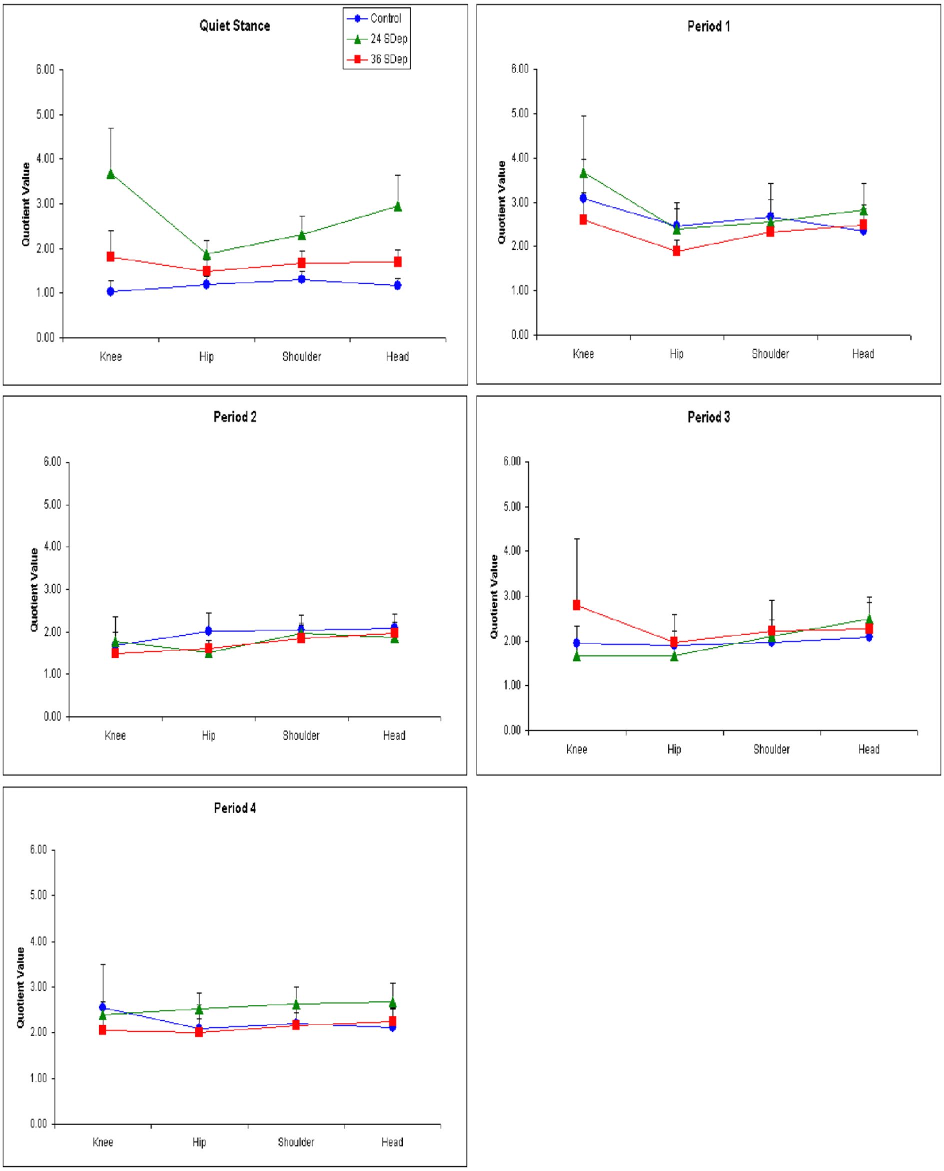


Figure 9: Eyes Closed/Eyes Open mean and SEM quotient values showing the proportional changes of movements at the knee, hip, shoulder and head in Quiet stance and the vibration periods following a normal night of sleep and at 24 (24 SDep) and 36 hours (36 SDep) of sleep deprivation. The figure shows less hip movement relative to the other body positions at 24 SDep in Quiet stance and in Period 1 to Period 3.

Study VI: Adaptation and vision change the relationship between muscle activity of the lower limbs and body movement during human balance perturbations.

Subjects: Eighteen healthy subjects (nine males and nine female; mean age 29.1 years, range 18-49 years) participated in the study.
Study design: Two randomised tests were performed, each lasting 230 seconds. The tests were:

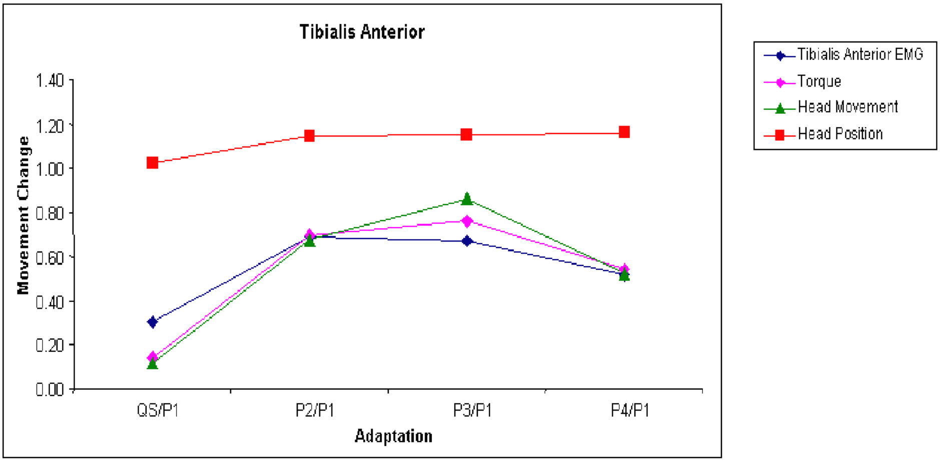
- Calf muscle vibration with eyes open (Calf-EO) and eyes closed (Calf-EC)

EMG muscle activity from the tibialis anterior and gastrocnemius muscles and body movement parameters were recorded. The relationship between EMG muscle activity and the movement measurements were investigated by correlation.

Results: Various costs of standing including energy ($p<0.01$), body movement ($p<0.01$) and tibialis anterior EMG muscle activity ($p<0.01$) were decreased by adaptation. Additionally, body posture changed to a further forward leaning position in all tests (generally $p\leq 0.002$ at most recorded positions).

Analysis showed a relationship between EMG muscle activity and movement. At the onset of perturbation, there was a significant relationship between tibialis anterior and gastrocnemius EMG muscle activity and movement. In the latter parts of the test, there was a significant relationship between tibialis anterior EMG muscle activity and body movement and body posture. There was also evidence of a significant relationship between gastrocnemius EMG muscle activity and body posture. These significant relationships differed when visual information was available compared to when not available.

Conclusions: Visual information and adaptation changed the standing strategy by modifying the relationship between muscle activity and movement when repeatedly perturbed by calf muscle vibration. This highlights the importance of vision and the optimisation of muscle activity and movement through adaptation. Furthermore, muscle activity costs might be decreased by adaptation through leaning further forward.



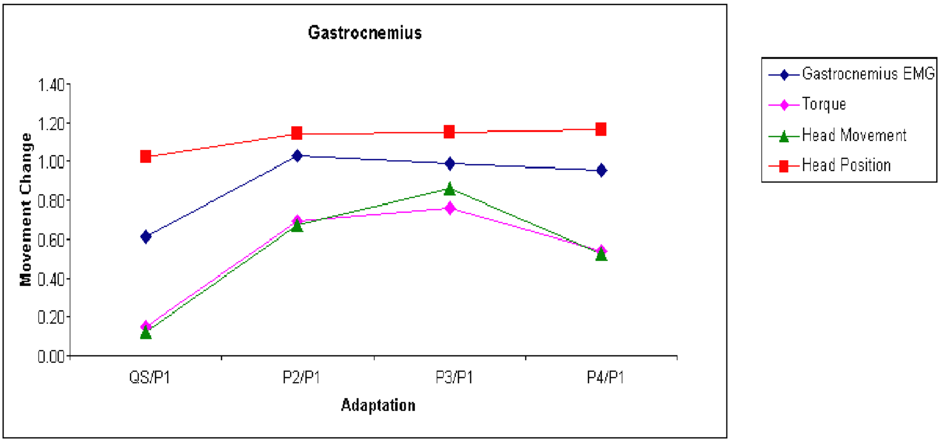


Figure 10: Relationship between EMG muscle activity and movement with eyes closed. In the axis, 1 represents no change between Periods, whereas 0 shows a large change. The figures show a relationship between tibialis anterior muscle activity and head movement and torque variance.

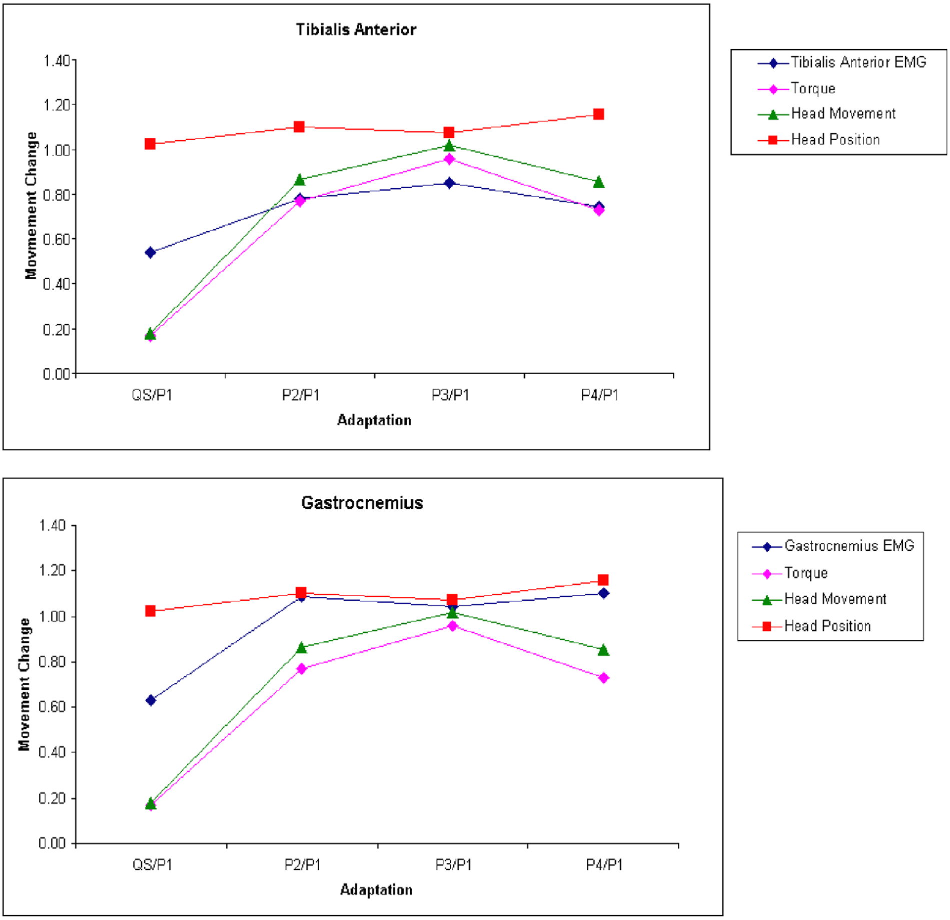


Figure 11: Relationship between EMG muscle activity and movement with eyes open. The figures show a relationship between tibialis anterior muscle activity and torque variance. Also, with eyes open, there was a relationship between gastrocnemius muscle activity and head angular position at P2/P1.

Discussion

This thesis illustrates that assessments of postural adaptation are highly useful in determining whether the postural control system is able to reach improved levels through repeated training exercises. One of the main objectives of adaptive behaviour was to minimise the costs involved in standing [72] i.e., reduce energy, body movement and muscle activity.

Influences on Postural Control and Adaptation

Effects of decreased plantar somatosensation on postural control and adaptation

It is generally accepted that when standing on a compliant foam surface, the accuracy of information from the mechanoreceptors on the soles of the feet decreases [28, 122]. Hence due to the close interactions between the vestibular and somatosensory systems, compliant foam surfaces are often employed to assess balance in patients with vestibular deficits [55, 123, 124]. Similar to others, the findings in **Study I** confirmed that standing on a foam surface perturbed balance [95, 125, 126].

Previously, it was assumed that when the surface increases in compliance, the information from the soles of the feet decrease [125, 126]. One would therefore expect balance to be much worse when the standing surface is more compliant. However, **Study I** showed that the stability recorded was directly related to the foam properties, not a major effect of decreased sensory information from the soles of the feet. When the foam surface was denser and had a higher elastic modulus, postural sway was larger. The results can be explained as so. The softer foam surfaces allowed a larger compression, resulting in a shorter distance between the feet and the hard surface beneath the foam. This ultimately increased the effectiveness of ankle torque for postural control compared with the firmest foam surfaces which permitted less compression.

Vision decreased the energy needed to remain standing more on surfaces with higher density or elasticity i.e., when subjects were increasingly perturbed by the support surface. This confirms that vision is highly important in postural control and also markedly decreases movement on foam surfaces [127]. The increased contribution of vision when the surface conditions increasingly challenged postural control could have provided a more stable frame of reference.

There are a few points that need to be considered when using foam surfaces. Since the destabilising effects on foam are complex and vary from one foam surface to another, one could question the use of this technique. The foam block used in balance assessments is likely to vary between clinics, and hence the results are not entirely comparable. One solution would be to use a standard foam surface for balance assessment. Since the firmest foam surface probably absorbed the most toque forces, the energy exerted when standing on this surface might well have been larger than captured. Similarly, the other foam surfaces used would have absorbed torque forces, but because the distance between the feet and the surface were small, there might not have been such an effect. One must also consider variations in weight between individuals. When standing on a foam block, heavier persons will apply a greater mass over the same area, and compress the foam material more than lighter persons. As a result, the distance between the feet and the surface is less for a heavier person.

The finding that firmer foams always disturb balance more than softer foams might not be true for all foam surfaces. For example, if the foam surface is very firm, it may produce results similar to standing on a solid surface. However, the range of foam surfaces used in **Study I** was selected on the basis of the densities used in other studies of balance using foam surfaces i.e., from low density (25 kg/m^3) [128] to high density (54.53 kg/m^3) [56].

Ultimately, there is still much uncertainty with how the mechanoreceptive information from the soles of the feet is affected by standing on foam surfaces and how the associated sensations affect postural control. However, when compared to standing with feet desensitised through ice-cooling, the torque variance when standing on foam was between 60-105% larger in the anteroposterior direction with eyes closed [129]. Thus, an alternative method for disturbing the accuracy of information on the soles of the feet should be considered to standing on foam blocks.

Effects of ageing on postural control and adaptation

Ageing decreases mechanoreceptive sensation on the soles of the feet [103, 104]. Accordingly, it has been found that clinical tests of plantar cutaneous sensation such as vibration perception and tactile sensitivity are affected by ageing [130]. The results in **Study II** were consistent with this. The results revealed that somatosensation was particularly important for postural control in the last 150 seconds of repeated calf muscle vibration with and without visual feedback, but less during quiet standing or initially to balance perturbations, where other sensory feedback may have had a greater contribution. Therefore, the level of mechanoreceptive sensation can affect the adaptive capabilities of postural control. Those with poorer sensation were unable to reach the same performance level as those with good sensation and this could not be compensated for by visual information.

Study III showed that balance was decreased with increasing age during quiet standing and when perturbed by vibration. The deficits were clearer in the lateral direction compared with the anteroposterior which is consistent with other reports [131-133]. It is well-known that the elderly experience a deterioration of the sensory, motor [134-136] and cognitive [107] systems. Any, or more likely an accumulation of several of these deficits could account for the instability observed in the elderly [136].

Similar to previous findings, the elderly subjects were able to adapt their balance to repeated balance perturbations [80, 137, 138]. Despite the adaptive capabilities of the elderly, they were unable to reach the same level of balance as healthy young adults. **Study II** showed that this is partly related to poor mechanoreceptive sensation. Hence, even after repeated perturbation training, the elderly might still be vulnerable to falls. The adaptive response was stronger in the lateral direction in the elderly compared to the young. Since a large threat to stability is required for adaptation [139], this finding possibly shows an increased threat to stability in the elderly in the lateral direction during balance perturbations. Balance training using repeated balance perturbations could however offer the elderly one way of improving their chances of avoiding a fall.

Vision was also a highly influential factor for stability and adaption in **Studies II and III.** It is well known that there is a deterioration of vision (i.e. acuity as well as motion perception) with age. For example, previous reports have showed that the number of ganglion cells of the retina decrease from the age of 42 resulting in a parallel decrease in the optokinetic nystagmus gain [100, 140, 141]. Thus, if the elderly rely on a failing visual system in the control of postural control, this could cause decreased postural control and adaptation, and in some cases even instigate falls.

Effects of sleep deprivation on postural control and adaptation

A lack of sleep is common in society. Causes include health problems and stress [142]. One should also consider that many elderly have sleeping problems [143]. The features of sleep deprivation include fatigue, a decrease in sustained attention [23] and decreased alertness [144]. Additionally, many therapeutic drugs taken by all ages induce sleepy effects.

Sleepiness is controlled by our endogenous circadian timing system, and in most people it aligns with the external 24 hour environment [145], peaking between 2-7am and 2-5pm, whether we have slept or not [146]. In **Studies IV and V**, balance and adaptation were assessed at 7 or 8am, 24 hours into continued wakefulness, and at 7 or 8pm, 36 hours into continued wakefulness. These tests revealed that balance was affected by a loss of sleep but only when perturbed. The integration of information from the visual, vestibular and somatosensory receptors and motor coordination are processes known to require attention [23, 25], especially when information from any of the sensory systems is not reliable [26]. As sleep loss decreases sustained attention, this finding is consistent with Zils et al. suggesting that there is a high risk of accidents following sleep loss when higher levels of attention are required [147]. This could account for the decreased postural control and adaptation following sleep deprivation.

To speculate, sleep deprivation might make subjects so tired, or inattentive, that they are unable to "identify" a balance threat and the CNS does not initiate the adaptive response. The effects of postural learning have not been considered following sleep deprivation, though evidence has suggested an association between sleep and cognition [148] and working memory [149]. Sleep, particularly paradoxical sleep, seems to have permissive effects on the consolidation of previously acquired information [150, 151] and this was showed by decreased performance at 36 hours of sleep deprivation, despite having experienced the perturbing stimuli at 24 hours of sleep deprivation. **Studies IV and V** therefore suggested that sleep deprivation not only affects the consolidation of previously acquired information (i.e., long-term learning) but also the adaptation to repeated stimuli (i.e., short-term learning), which ultimately has a number of repercussions for both education and safety.

Additionally, as sleepiness increases, the probability of micro-sleeps also increases [152], and these micro-sleeps are associated with brief but large cortical lapses of attention. Micro-sleeps could explain the large increases in instability midway (100-150s of vibration) through testing where the continuation of a boring task might have exaggerated balance deficits. Prior studies have also showed that sleep deprivation has a more deleterious effect on performance in long, monotonous, boring tasks than on short, interesting ones [108, 153, 154].

When standing unperturbed, we often assume an 'ankle strategy' [155, 156] characterised by rotation about the ankles. However, under some circumstances during a difficult task, the knees and hips can assume the position at which pivoting occurs with the trunk musculature involved, and this is termed a 'hip strategy' [155]. One change caused by a lack of sleep in **Study V** was a hip strategy at 24 hours of sleep deprivation. Here, a hip strategy suggests that subjects felt unsteady during sleep deprivation. Furthermore, since the largest effects of sleep deprivation were present at 24 SDep and not 36 SDep when participants felt the sleepiest, balance appears to be partly associated with the time-of-day [115, 116], not the duration of continued wakefulness or one's own perception of sleepiness.

Adaptation and the optimisation of postural control

Studies II, III, IV and V all showed that the assessment of postural adaptation is sensitive to balance deficits and provides information on whether the individual can learn to adapt their standing balance and adapt their balance sufficiently well to appropriate patterns. **Study VI** detailed these adaptive responses. Findings revealed that the adaptive responses were continual and altered the association between postural muscle activity and movement. These changes decreased various costs involved in standing i.e., the energy, movement and muscle activity, and thus show optimisation of postural control.

Study VI revealed that the associations between EMG muscle activity and postural sway are complex. The correlation coefficients between gastrocnemius muscle activity and body posture were larger during the initial increase in movement in Period 1 and the initial adaptation in Period 2 whereas there was a significant relationship between changes in tibialis anterior muscle activity and movement variance in Periods 1, 3 and 4 of vibration. The optimisation of postural control was apparent through decreased tibialis anterior EMG muscle activity which correlated with decreased movement, and further with a forward movement of body posture.

The importance of visual information during standing is well known [13, 157]. Visual information is important in postural control for detecting movement in the visual surrounds. This enables humans to recognise upcoming hazards and trigger pre-programmed postural strategies i.e., pattern of muscle activation from prior experiences. The associations between EMG muscle activity and movement were affected by the availability of visual information which corroborates **Studies II and III**, showing that vision can affect adaptive responses. Hence, in learning to compensate for balance perturbations, visual information not only improves stability but helps in formulating different motor strategies. This is consistent with Hafstrom et al. that closing the eyes changes the postural standing strategy when compared to standing with eyes open in complete darkness [41].

There was no adaptation of the gastrocnemius EMG muscle activity whereas there was adaptation of the tibialis anterior EMG muscle activity. This response suggests that the CNS placed a high demand on the gastrocnemius muscles for the regulation of the standing position. Gastrocnemius EMG activity was not decreased over time despite a concomitant reduction in body movement variance. Hence, the standing forward leaning position was largely determined by the gastrocnemius muscles, whereas some of the tibialis anterior muscle activity was replaced by leaning further forward.

Postural control is regulated by skeletal muscles that form an uninterrupted muscular chain that extends from the head to the feet [158] as confirmed by the induced balance perturbations caused by proprioceptive vibration at various locations [159]. Thus, one possibility is that during some periods of time some other postural muscles along the muscular chain may influence the body movements more than the gastrocnemius and tibialis anterior muscles.

Methodological Considerations

Choice of stimulus

In **Studies II-VI**, vibration was selected as the method to perturb standing balance. Recordings of vibration-induced postural sway have been used to distinguish the postural responses to sensory deficits such as visual impairment [41], vestibular impairment [160] and plantar mechanoreceptive impairment [129] and to evaluate postural control in the elderly [48, 161]. Hence, a significant sensory re-weighting would be required to compensate for these effects. An alternative method of perturbation would be to use Computerised Dynamic Posturography (CDP). However, CDP operates using a mechanical moving platform and hence directly alters orientation along with vestibular information. Furthermore, CDP is less sensitive to balance deficits than perturbing balance to the limits of stability [162].

In studies where balance has been perturbed using vibration, the muscle vibrated significantly affected the direction of instability [163, 164] and the subsequent postural movements [165, 166]. In **Study III**, both neck and calf muscle vibration were considered whereas only calf vibration was considered in **Studies II, IV, V and VI**. In **Study III**, differences in postural responses between calf and neck vibration were evident. Whereas calf vibration directly elicited movement mainly in a posterior direction, neck vibration directly elicited movement mainly in an anterior direction. Postural sway was larger with calf vibration than with neck. The neck and calf muscles have different roles in the maintenance of postural control and thus their vibration resulted in different imbalances. Neck muscle afferents are mainly involved in the regulation of body orientation [167, 168] whereas the calf muscles are mainly involved in the maintenance of equilibrium [168]. Hence, consideration should be given to the location of disturbance and the proprioceptive information that is being affected in studies of postural control since the nature of the imbalance (i.e., its amplitude and location) directly affects the movement [166] and adaptive responses.

Ageing

Ageing results in deterioration of the sensory and motor systems [136]. Before balance assessment, the elderly in **Studies II and III** were checked for visuo-vestibular deficits and plantar sensation loss. This revealed an age-related deterioration of the sensory systems in line with expectations. Hence, the data from all elderly participants were used.

General Discussion

The ability to select and dynamically re-weight [169, 170] alternative orientation references adaptively in conflicting and demanding situations is considered one of the main issues for postural control [171]. Recent models of postural control [29, 85, 172] have proposed that the controls of self-motion and self-orientation largely depend on how the CNS internally reconstructs, from sensory information, the kinematics (geometry of the support surface as a reference frame) and kinetics (gravito-inertial pull on the body and joints) of our own movements and those of the environment with which we interact [83]. In this way, motor learning can be conceived as the establishment of an internal model which represents the exact matching between sensory and motor information [38, 173].

This thesis extends what is known about motor learning. The postural control system was capable of dynamically re-weighting orientation references to improve postural control. The adaptive learning capability of the postural control system, and hence the accurate reconstruction of the kinematics and kinetics of movement to repeated balance perturbations was dependent on one's own level of mechanoreceptive somatosensation and availability of visual information. By decreasing attention and alertness through sleep deprivation [23], adaptive capabilities were decreased, suggesting an important role for sleep in memory and consolidation of a new motor skill.

Using vibratory stimulation, errors in the sensory information from the proprioceptors occurred rapidly, and instigated rapid adaptive changes. However, the persistence of these errors also instigated slow adaptive changes. This finding is consistent with Kording et al [174]. Hence, adaptation to repeated balance perturbations probably occurs on different time scales. Therefore, when the nervous system observed the error in performance by repeated balance perturbations, it faced a credit assignment problem given that there were different timescales of error. The solution to this could have been that the nervous system kept a measure of uncertainty about its current parameter estimates to allow an optimal combination of new information with current knowledge [175]. This could also include a mechanism mediated by neuro-modulators for allowing fast changes at catastrophic moments [174], such as before a fall.

Clinical Implications

- In the past, the methodology used to study the role of the plantar cutaneous messages in postural control consisted of changing the supporting surface on which the subject is standing, such as having them stand on a foam surface [64]. In addition, foam surfaces are often used in vestibular rehabilitation [123, 124] due to the close interactions between the vestibular and somatosensory systems for postural orientation [176]. Hence, the significantly different results on each of the foam surfaces and the change in contribution of vision on different foam surfaces suggests that to improve the assessment and rehabilitation of the vestibular patient, care should be taken in selecting the foam surface used to test postural control **(Study I)**.

- Clinicians should be aware of the large lateral impairments with increasing age. However, repeated balance perturbation training can be used to improve balance in the elderly in both the anteroposterior and lateral directions. Visual and mechanoreceptive deterioration with age contributed greatly to decreased postural control and adaptation. Clinicians might therefore be advised to check both visual and mechanoreceptive functions prior to balance assessments in all patients **(Studies II and III)**.

- Sleep deprivation increased the risk for falls, even when an individual had experience of the perturbation. The risk of accidents is higher in tasks that require attention, which would include driving. Clinical, chronic, disorders of sleep, such as obstructive sleep apnea or even by taking drowsiness inducing drugs, may decrease postural control, the learning mechanisms involved in motor tasks and the consolidation of motor skills **(Studies IV and V)**.

- Vision changed the standing strategy by modifying the relationship between muscle activity and movement and thus, in the evaluation of balance disorders, posturography must be assessed with eyes open and eyes closed. As the relationship between EMG muscle activity and body movement appears to be optimised over time i.e., less tibialis anterior EMG muscle activity with an increased body leaning forward and less linear movement and torque variance, adaptation training through proprioceptive vibration could benefit those susceptible to falls. Furthermore, this stresses that additional information about an individual's rehabilitation status following musculoskeletal injury or visual deficit would be gained when assessing the relationship between muscle activity and movement **(Study VI)**.

Conclusions

- Different compliant foam surfaces affect postural control depending on their properties. Thus, the choice of foam surface used in balance assessment can significantly determine the measured outcome. Therefore, foam surfaces should be used with caution, as what may seem like excessive postural sway might be within normal ranges for that kind of surface. An alternative method for disturbing the accuracy of information on the soles of the feet should be considered (**Study I**).

- Accurate mechanoreceptive information from the cutaneous receptors on the soles of the feet is vitally important for postural control and adaptation. When the sensitivity of these receptors is decreased by age, adaptive capabilities are reduced **(Study II)**.

- Ageing decreases postural control but the adaptive capabilities are still intact in both the anteroposterior and lateral directions. However, the elderly are unable to reach the levels of the young from this adaptation and are therefore susceptible to falls even after balance training **(Study III)**.

- Sleep deprivation decreases postural control and the ability to learn from previous perturbations. Sleep is therefore required for the learning of a motor skill. The largest effects of sleep deprivation were present at 24 SDep and not 36 SDep where participants felt the sleepiest. Therefore, the effects of sleep loss on balance control appear to be partly associated with the time-of-day. This shows that one's own perception of sleepiness is not a reliable indicator of motor control **(Studies IV and V)**.

- Postural adaptation to repeated balance perturbations was characterised by a decrease in the costs of standing. The relationship between muscle activity and movement is complex, but optimised through repeated training. Furthermore, vision helps in formulating different motor strategies **(Study VI)**.

Summary in Swedish

Människans förmåga att kunna anpassa sin balanskontroll till nya företeelser och omständigheter är en viktig egenskap som är beroende av reflexutlösta och anticipatoriska mekanismer som baseras på information från synen, innerörats balansorgan (vestibularisorganen), led- och muskelsinnet (proprioception) och känselreceptorer i huden (mekanoreceptorer). Det centrala nervsystemet (CNS) sammanställer och omarbetar kontinuerligt sensorernas information till en inre bild av kroppens aktuella position och rörelser. Förmåga till anpassning (adaption) är viktig för balanskontrollen både under normala förhållanden och när man behöver kompensera för tillstånd och skador som orsakar balanssvårigheter. Den sensoriska informationen från syn, vestibularisorgan, proprioception och känselreceptorer i huden är delvis överlappande för balanskontrollen. Om ett sinnesorgan skadas kan därför de andra fungerande organen ta över och hjärnan vikta om informationsinflödet till bästa möjliga informationskälla.

Syftet med denna avhandling var att undersöka människans balanskontroll och förmågan till adaption och inlärning under några vanliga vardagliga situationer; när informationen från fotsulorna inte är tillförlitlig; när sensoriska och motoriska system är påverkade av åldersrelaterade förändringar, samt när uppmärksamhet och vakenhet påverkas av sömnbrist. Dessutom undersöktes på vilka sätt aktiviteten i postural vadmuskulatur och kroppens rörelsemönster förändras till följd av adaption vid balansstörningar. Sex experimentella studier på friska försökspersoner ingår i avhandlingen.

Balanskontrollen och adaptionsförmågan undersöktes på tre olika sätt; genom att med posturografi (ståplatta) mäta energin som användes mot underlaget för att behålla kroppsstabiliteten; genom att med ett ultraljudsbaserat rörelseanalyssystem (Zebris™) mäta tredimensionella rörelser av knän, midja, axlar och huvud, och genom att med EMG mäta muskelaktiviteten i underbensmuskulatur (tibialis anterior och gastrocnemius).

I delarbete II, III, IV, V och VI användes vibrationsstimulering mot vader och nackmuskler för att ge kontrollerade balansstörningar. Vibrationsstimulering ger upphov till sträckreflexer i musklerna och om man stör proprioceptionen i vad- och nackmuskulaturen med vibrationsstimulering ökar kroppssvajet framför allt i anteroposterior (framåt-bakåt) riktning. Balansstörningar från vibrationsstimulering ger oftast hos friska personer upphov till en inlärningsprocess som försöker minska effekterna av störningen, dvs ju mer erfarenhet en försöksperson har av vibrationsstimuleringen desto mindre effekt har stimuleringen.

Visuell information spelar en viktig roll för människans balanskontroll. Om man inte får tillförlitlig information från synen blir balanskontrollen mer beroende av att andra organ ger korrekt information. För att undersöka synens roll för balanskontrollen gjordes därför samtliga balanstester både med öppna och slutna ögon. Ytliga receptorer under fotsulorna är en annan viktig informationskälla för balanskontrollen i stående. Kroppsstabiliteten blir t ex sämre om man står på mjukt underlag, och man har tidigare antagit att detta beror på att känselreceptorerna i fotsulorna inte lika tydligt känner av kroppens rörelser och krafterna mot underlaget. För att enklare kunna diagnostisera balansproblemens orsaker har man därför ofta undersökt stabiliteten hos patienter med misstänkta vestibulära skador när de står på mjuka skumgummiplattor.

I **delarbete I** undersöktes kroppssvajet hos 30 friska försökspersoner när de med öppna eller slutna ögon fick stå på skumgummiplattor med 3 olika mjukhetsgrader. Försökspersonerna använde betydligt mer energi för att behålla kroppsstabiliteten på det ”minst elastiska” skumgummit, dvs. på underlaget som komprimerades minst av vikten och krafterna. Betydelsen av visuell information för kroppsstabiliteten var också relaterad till underlagets egenskaper. Visuell information förbättrade stabiliteten som mest när personerna svajade som mest, dvs. när man stod på det ”minst elastiska” underlaget. Resultaten tyder på att när man står på mjukt skumgummiunderlag så påverkas friska försökspersoners stabilitet framför allt av underlagets kraftabsorberande egenskaper. Det är således viktigt att ta hänsyn till underlagets egenskaper när man gör balanstester eftersom uppmätta resultat markant kan påverkas av underlagets egna mekaniska egenskaper.

Åldersrelaterade degenerativa förändringar påverkar balanskontrollen både genom att samtliga sensoriska inflöden blir försämrade och genom att den neuromuskulära kontrollen blir försämrad. I **delarbete II** jämfördes hur viktig informationen från mekanoreceptorerna i fotsulorna var för stabiliteten hos 16 äldre försökspersoner och 25 yngre personer under balanstester med vibrationsstimulering mot vadmusklerna. Mekanoreceptorernas känslighet undersöktes genom att mäta minsta tryckkraft som kunde uppfatta mot huden (taktil perception) och minsta vibrationsamplitud man kunde uppfatta (vibrationskänsel). De äldre försökspersonerna presterade sämre än de yngre i båda dessa tester. Dessutom fanns ett tydligt samband mellan mekanoreceptorernas funktion och förmågan till adaption och stabilitet i stående. Äldre personer och patienter med dålig sensibilitet i fötterna torde därför löpa större risk att drabbas av fallolyckor.

I **delarbete III** jämfördes balanskontrollen hos 16 äldre med 18 yngre försökspersoner i stående under balanstester med vibrationsstimulering mot vad- och nackmuskler. De äldre var ostadigare både i sidled (lateralt) och i framåt-bakåt riktning (anteroposteriort) jämfört med de yngre försökspersonerna. De äldre kunde adaptera sig till balansstörningarna i båda riktningarna, men de var generellt betydligt mer instabila än de yngre och nådde trots tydliga förbättringar aldrig upp till samma stabilitetsnivå. Visuell information hade stor betydelse för stabiliteten hos både de äldre och yngre försökspersonerna. Fynden visar att äldre markant kan förbättra sin balans och stabilitet genom balansträning, vilket tillsammans med säkerställande av god synfunktion bör ge dem bättre balanskontroll och förebygga fallolyckor.

CNS integration av information från syn, vestibularisorgan och somatosensoriska organ samt muskelkoordinationen förbättras vid höjd uppmärksamhet och hög vakenhetsgrad. Förbättringarna är speciellt tydliga när informationen från något av sinnesorganen inte är tillförlitlig. Sömnbrist med försämrad vakenhet skulle därför kunna orsaka en försämrad balanskontroll i stående. I **delarbete IV** och **delarbete V** undersöktes hur nedsatt vakenhet påverkar balanskontroll och adaption till balansstörningar. I studierna mättes hur mycket energi försökspersonerna använde mot underlaget för att behålla stabiliteten (**IV**) och hur kroppsrörelserna såg ut (**V**) under balanstester med upprepade vibrationer mot vadmusklerna. Detta undersöktes efter en normal natts sömn och efter 24 och 36 timmars vakenhet. Försökspersonerna behövde använda signifikant mer energi mot underlaget, för att hålla balansen vid sömnbrist, både med öppna och slutna ögon. Adaptionen till balansstörningarna, dvs anpassning och inlärning av nya rörelsemönster, var också betydligt sämre vid sömnbrist. Sömnbrist påverkade också kroppens rörelsemönster och ökade rörelserna av huvud, axlar, midja och knän vid vibrationsstimulering.

Postural muskulatur är viktig för att upprätthålla balanskontrollen och här spelar underbenens muskulatur en stor roll. Det är dock till stora delar okänt på vilket sätt relationen mellan musklernas arbete och kroppsrörelserna förändras till följd av adaption vid olika balansstörningar. I **delarbete VI** undersöktes hur adaption till följd av vadvibration påverkar relationen mellan muskelaktivitet, mätt med EMG, och kroppsrörelserna hos 18 försökspersoner. Resultaten visade att relationen mellan EMG aktiviteten i underbenens muskler och kroppsrörelserna påverkades både av adaption och av tillgången till visuell information. Resultaten stödjer de tidigare slutsatserna i avhandlingen; om man genomför balansövningar som initierar en inlärningsprocess så kan inlärningen förbättra funktionen mellan musklernas arbete och kroppens rörelser. Undersökningsmetoden kan också ge en bättre applicerbar utredningsmetodik för att diagnostisera patienter med visuellt relaterade balansproblem.

Slutsatser

Avhandlingen ger ökad förståelse för hur komplexa motoriska processer som människans balanskontroll kan förbättras och anpassas om förhållandena medger möjlighet till inlärning och adaption. Resultaten visar att det finns god kapacitet både hos äldre och yngre att omvärdera och prioritera olika sensorisk information för att uppnå så god stabilitet som det är möjligt. Balanssystemet kan modifiera och optimera relationerna mellan muskelaktivitet och kroppsrörelser vid störningar av balanskontrollen och vid minskat sensoriskt inflöde. När underlagets egenskaper gradvis ökar utmaningarna för balanskontrollen, så ökar också betydelsen av att ha tillgång till god visuell information. Förmågan till adaptiv inlärning är dock till viss grad beroende av korrekt information från mekanoreceptorerna i fotsulorna och av hög kvalité på visuell information. Effekterna av sömnbrist visar att uppmärksamhet och förmågan till fokusering spelar en betydelsefull roll för att man kan förbättra och anpassa sin balanskontroll när man har balansproblem.

Fynden i avhandlingen är betydelsefulla för vår förståelse av människans balanssystem och för utvecklingen av välfungerande rehabiliteringsprogram, vilka ofta kan behöva skräddarsys till individuella behov för att uppnå optimal effekt. Avhandlingen visar också betydelsen av att utvärdera förmågan till adaptation och inlärning bland patienter som har balanssvårigheter.

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Articles

Article I

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The effect of foam surface properties on postural stability assessment while standing

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Abstract

A common assessment of postural control often involves subjects standing on a compliant surface, such as a foam block, to make balance tests more challenging. However, the physical properties of the foam block used by different researchers can vary considerably. The objective of this study was to provide an initial approach for investigating whether two of the foam properties, i.e. density and elastic modulus, influenced recorded anteroposterior and lateral torque variance with eyes open and eyes closed. Thirty healthy adults (mean age 22.5 years) were assessed with posturography using three different types of foam block placed on a force platform. These blocks were categorised: firm foam, medium foam and soft foam by their elastic modulus. To investigate the spectral characteristics of recorded body movements, variance values were calculated for total movements, movements <0.1 Hz and movements >0.1 Hz. Results showed that anteroposterior and lateral torque variances >0.1 Hz were larger when standing on the firm foam compared with medium and soft foam and in turn were larger on the medium foam compared with the soft foam with eyes closed. Moreover, GLM and correlation analysis demonstrated that the properties of the foam blocks affected anteroposterior torque variance >0.1 Hz and lateral torque variance in all frequency ranges. In addition, the stabilising effect of vision in the anteroposterior direction had a greater influence when the subjects' stability was increasingly challenged by the support surface, as illustrated by the higher torque variance values. In conclusion, caution should be taken when analysing balance deficits with foam test setups, because the foam properties may influence the recorded body movements.

Keywords: Postural control; Foam; Surface properties; Vision

1. Introduction

During quiet stance on a firm surface, postural control is characterised by continuous, small corrective movements [1,2] known as postural sway. These movements are initiated through feedback and feedforward mechanisms and are coordinated by the sensory and motor systems [3]. An important source of afferent information required to regulate postural stability is believed to come from specialised cutaneous mechanoreceptors on the soles of the feet [4]. These mechanoreceptors are able to provide information about surface contact pressures [5], and are thus important for sensing the small, continuous changes of posture.

However, clinical assessment of quiet stance on a firm surface sometimes lacks the sensitivity for distinguishing healthy patients from those with balance disorders [6], and therefore a number of balance perturbing methods have been devised to place an increased demand on the postural control mechanisms so that any balance disorder becomes apparent. One common method that has been used to peturb balance in such a way is to have patients standing on a compliant surface such as a foam block [7]. When standing on a foam block, the ability to sense pressure distribution and body orientation decreases [8,9]. Furthermore, standing on foam also causes a mechanical perturbation as compression of the compliant visco-elastic surface reduces the effectiveness of

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ankle torque required for postural stabilisation [10,11]. Previously, various studies have used foam blocks to investigate postural stability during standing [12–14] or during gait [7,9,10]. However, the thickness and density of the foam surfaces in different studies have varied from one another considerably. For example, the foam block can range from being 2-cm thick [15] to 12-cm thick [10] and from low density (25 kg/m^3) [16] to high density (54.53 kg/m^3) [17]. Although it is difficult to determine the degree to which mechanoreceptive information is affected by compliant surfaces [18], research has suggested that the input to the mechanoreceptors on the soles of the feet is increasingly affected by adding layers of foam of the same material [8]. Additionally, it is reasonable to assume that postural stability will also be affected by the mechanical properties of the compliant surface [10,11]. Thus, when using foam blocks to perturb standing, we hypothesise that the density and elastic modulus properties of the foam block base may affect the level of postural instability. Also, since vision plays an important role in the maintenance of postural stability, especially when information from another sensory receptor is unreliable [12], the availability of visual information may affect stability while standing on foam.

Our study aimed to compare human postural control in terms of torque variance on three foam blocks of different properties and on a control solid surface. We also wished to assess the contribution of vision to postural control on each surface.

2. Materials and methods

2.1. Subjects

All 30 subjects (19 male and 11 female) were healthy adults aged between 19 and 43 years (mean 22.5 years with a standard deviation (S.D.) of 5.5 years). Their average mass was 67.3 kg (S.D.: 10.6 kg) and average height 172 cm (S.D.: 9 cm). No subject had previously experienced balance problems, neurological disorder or significant injury of the legs, nor were any taking medication. All were asked to refrain from alcohol at least 48 h prior to testing. Full, informed consent was obtained before any tests were performed, all of which were carried out according to the Helsinki Declaration of 1975.

2.2. Equipment

Postural stability while standing is commonly analysed using force platforms and the movements of the centre of pressure (CoP), i.e. the point of application of the ground reaction force. We analysed the variance of the torque values, where the torque τ is calculated from the formula $\tau = \text{CoP}\, mg$; where m = the assessed subjects mass (kg) and g = gravitational constant 9.81 (m/s^2). Hence, changes in recorded torque are equivalent to changes in CoP [19]. One benefit with presenting torque variance values is that the calculated value corresponds directly with the energy used towards the support surface to maintain stability [20].

Torque variance was recorded using a force platform containing six force sensors beneath its surface which measured the ground reaction forces and shear forces with an accuracy of 0.5 N. The data from the force platform were fed at 50 Hz into a customised computer program which stored all recordings. Torque variance was recorded in the anteroposterior and lateral directions on three foam blocks of different properties placed on the force platform. These were categorised: soft foam (SF), medium foam (MF), firm foam (FF) by their elastic modulus. We also recorded torque variance while standing directly on the solid surface of the force platform (SS; dimensions 423 mm length × 420 mm width × 117 mm height above ground) as a control response. The foam blocks were commercially obtained (Foams4Homes, Bristol, UK) and were selected on the basis of two criteria. The first being that they were within the density range of the foams used in other posturographic studies [16,17]. The second, that they differed in their compliance (i.e. their ability to compress under applied force). The foams blocks used were slightly larger than the force platform. However, in the measurement setup the foams were placed on top of the force platform, without allowing any contact with the surroundings or ground.

The dimensions, densities and elastic modulus of the foam blocks are listed in Table 1. The density (ρ) of the foams was calculated from the formula:

$$\rho = \frac{m}{V}$$

where m denotes mass of the foam (kg) and V denotes the volume of the foam block (m^3). The volume (V) was calculated from V = foam length (m) × foam width (m) × foam height (m). The elastic modulus (E) was calculated from the formula:

$$E = \frac{kH}{A}$$

where k denotes the axial stiffness (N/m), H (m) is the height of the foam to which the force is applied on top off, and A (m^2) is the cross-sectional area over which the force is applied, in this case A = foam length (m) × foam width (m). Since foam can have non-linear axial stiffness depending on applied force and compression of the foam, following the completion of the study, the axial stiffness of the foams was calculated by tracking each foams displacement (m) against force applied (N) using a designated force–compression measurement system at the University of the West of England, UK which yielded a displacement versus applied force graph for each foam block. The force was applied with steps

Table 1
The dimensions, density and elastic modulus of the foam blocks used in this study

Foam material	Dimensions, length × width × height (mm)	Density (kg/m^3)	Elastic modulus (N/m^2)
Firm	467 × 473 × 136	82.0	49,200
Medium	466 × 467 × 134	21.3	20,900
Soft	464 × 468 × 132	21.9	4,200

of 2 N increments, and care was taken to measure the response to a statically applied force. The load was applied over the entire surface area using a plate that was of the same size as the foam block. Using the resulting force versus displacement graphs, the average axial stiffness value k within the force range from 550 to 800 N was calculated using the formula:

$$\kappa = \frac{F}{\mathrm{d}x}$$

where k denotes the axial stiffness (N/m), F denotes vertical force (N) and dx denotes compression of the foam (m). The force range was selected to cover the range of vertical forces within the test group caused by the subjects weight (N) while standing on the foam. Of note, all three foams used had fairly linear elastic modulus properties within this force range.

2.3. Procedure

Each subject was asked to stand barefoot on the force platform or on a foam block placed on top of the force platform, in a relaxed posture, with arms folded. The subject's heels were 3 cm apart and feet positioned at an angle of 30° apart open to the front using guidelines on the platform. Subjects either focused on a visual target (6 cm × 4 cm high-quality picture of a sea-side landscape) positioned at eye level and mounted squarely, flush, on the wall at a distance of about 1.5 m, or had their eyes closed when instructed. During the tests with eyes open, stabilising visual references were provided by the visual target straight ahead and floor-to-wall edges, wall-to-ceiling edges and corner edges within the field of view.

Each subject was asked to stand on the four surfaces twice, once with eyes open (EO) and once with eyes closed (EC), with each of the eight tests lasting for 2 min. The orders of surface and vision conditions were randomised using a Latin square design to reduce any potential order effect. To avoid fatigue, there was 3 min rest between tests.

2.4. Data analysis

Anteroposterior and lateral movements were analysed in terms of variance of torque (denoted M_x and M_y in Fig. 1) from the force platform recordings and divided into three spectral categories: total, movements below 0.1 Hz (<0.1 Hz, low frequency), and movements above 0.1 Hz (>0.1 Hz, high frequency). These separations were used to distinguish between smooth corrective changes of posture (i.e. <0.1 Hz) and fast corrective movements to maintain balance (i.e. >0.1 Hz) [21]. A fifth-order digital finite duration impulse response (FIR) filter, with filter components selected to avoid aliasing [22] was used for spectral separation of the raw data into recorded torque below and above 0.1 Hz, and from these data low and high frequency torque variance were calculated. The frequency cut-off level of 0.1 Hz was based upon previous studies showing that vision effectively reduces the torque activity above 0.1 Hz on a firm surface [21,23]. Torque variance was normalised using the subjects' squared height and mass before the statistical analysis in order to compensate for the individual differences in height and mass [19,24,25].

2.5. Statistical analysis

Non-parametric statistical tests were used as the Shapiro–Wilk test revealed that the values were not normally distributed before or after logarithmic transformation. Wilcoxon, matched-pairs tests were used to investigate the differences in torque variance between the surfaces. In the Wilcoxon statistical analysis, Bonferroni correction for multiple comparisons was used and p-values < 0.01 were considered statistically significant. However, we present the p-values < 0.05 in the figures for consistency.

The Spearman's rank correlation statistical test was used to investigate the relationship between surface density and torque variance, and surface elastic modulus and torque variance. In the Spearman's statistical analysis, p-values < 0.05 were considered statistically significant.

In addition, a GLM univariate ANOVA (general linear model univariate analysis of variance) [26] statistical test on log-transformed values was used to determine whether vision, the type of foam or their interaction significantly affected torque activity between all compliant surfaces. The GLM model accuracy was evaluated by testing the model residual for normal distribution. In the GLM analysis, p-values < 0.05 were considered significant.

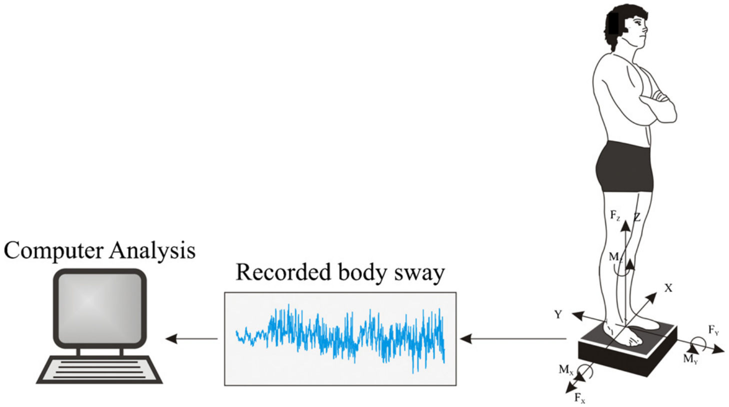

Fig. 1. Schematic of the measurement system and force platform, with recorded forces and toques marked.

3. Results

3.1. Anteroposterior torque variance

With eyes closed, anteroposterior total torque variance on FF was about 25% larger than on MF and SF ($p < 0.01$) and about 105% larger than on SS ($p < 0.001$), see Fig. 1. Torque variance was also about 60% larger on MF and SF compared with SS ($p < 0.001$).

Low frequency torque variance was about 35% larger on SF than on MF ($p < 0.01$).

High frequency torque variance on FF was about 45% larger than on MF ($p < 0.001$), 95% larger than on SF ($p < 0.001$) and 400% larger than on SS ($p < 0.001$). High frequency torque variance on MF was also about 35% larger than on SF ($p < 0.001$) and about 250% larger than on SS ($p < 0.001$). Additionally, torque variance on SF was 155% larger than on SS ($p < 0.001$).

With eyes open, statistical analysis revealed no significant differences below the Bonferroni corrected level between the surfaces in the total variance values, see Fig. 1.

Low frequency torque variance was about 45% smaller on FF than on MF ($p < 0.01$).

High frequency torque variance on FF was about 60% larger than on SS ($p < 0.001$). High frequency torque variance on MF and SF was also about 40% larger than on SS ($p < 0.001$).

3.2. Lateral torque variance

With eyes closed, total torque variance on FF was about 35% larger than on MF ($p < 0.001$), about 80% larger than on SF ($p < 0.001$) and about 300% larger than on SS ($p < 0.001$), see Fig. 2. In addition, total torque variance was about 195% larger on MF ($p < 0.001$) and about 120% larger on SF ($p < 0.001$) compared with on SS.

Low frequency torque variance was about 65% larger on the FF than on SF ($p < 0.001$) and about 175% larger than on SS ($p < 0.001$). Low frequency torque variance was also about 95% larger on MF than on SS ($p < 0.001$).

High frequency torque variance on FF was about 35% larger than on MF ($p < 0.001$), 90% larger than on SF ($p < 0.001$) and 460% larger than on SS ($p < 0.001$). High frequency torque variance on MF was also about 40% larger than on SF ($p < 0.001$) and about 310% larger than on SS ($p < 0.001$). Additionally, high frequency torque variance was 195% larger on SF than on SS ($p < 0.001$).

With eyes open, lateral total torque variance was about 110% larger on FF ($p < 0.001$), about 75% larger on MF ($p < 0.001$) and about 65% larger on SF ($p < 0.01$) than on SS, see Fig. 2.

Low frequency torque variance was only about 95% larger on FF than on SS ($p < 0.01$).

High frequency torque variance was about 35% larger on FF than on MF ($p < 0.01$), 40% larger than on SF ($p < 0.001$) and about 125% larger than on SS ($p < 0.001$). High frequency torque variance on MF and SF were also about 70% larger than on SS ($p < 0.001$).

3.3. GLM univariate ANOVA of torque values

Table 2 shows that the type of foam block significantly affected anteroposterior high frequency torque variance ($p < 0.001$) and lateral torque variance in the total and high frequency ranges ($p < 0.001$). The type of foam block also affected lateral low frequency torque variance, but only at $p < 0.05$. High frequency anteroposterior torque variance and lateral torque variance were all clearly larger on FF compared with on MF and SF. Furthermore, there were some

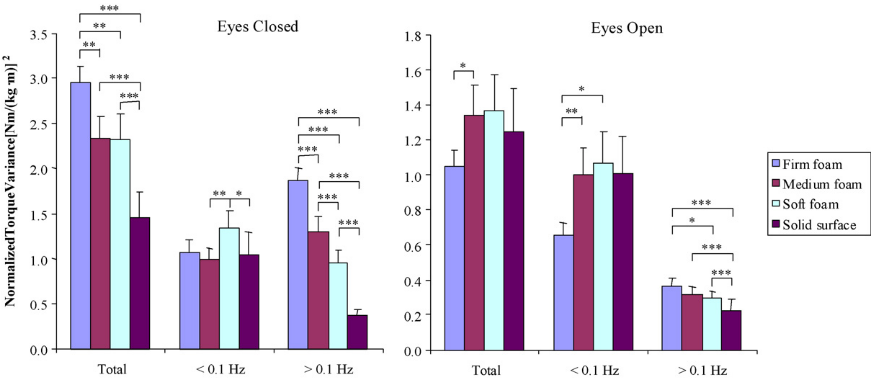

Fig. 2. Anteroposterior torque values with eyes closed and eyes open (mean and S.E.M.) for variance of total torque, variance of torque <0.1 Hz and variance of torque >0.1 Hz. The variance values have been normalised with the subject's mass and height. The statistical differences found between the surfaces are marked with asterisks, where $*p < 0.05$, $**p < 0.01$ and $***p < 0.001$. Of note, after Bonferroni correction for multiple comparisons p-values < 0.01 were considered statistically significant. However, we present the p-values < 0.05 in the figures for consistency.

Table 2
Statistical evaluation of the torque variance values in anteroposterior and lateral directions using the GLM univariate ANOVA method showing the effect of different types of foam on torque variance

Torque variance	p-Value		
	Foam	Visual influence	Foam × visual influence
Anteroposterior			
Total	ns	<0.001	0.029
LF < 0.1 Hz	ns	0.015	ns
HF > 0.1 Hz[a]	<0.001	<0.001	0.034
Lateral			
Total	<0.001	<0.001	ns
LF < 0.1 Hz	0.043	0.001	ns
HF > 0.1 Hz[a]	<0.001	<0.001	ns

The notation "<0.001" means that the p-value is smaller than 0.001, and "ns" signifies no significant difference.
[a] The GLM model residual was not normally distributed. These statistical values may therefore be somewhat less accurate.

Table 3
Correlation values between foam density, foam elastic modulus and recorded torque values with A, eyes closed and B, eyes open

	Density		Elastic modulus	
	R-value	p-Value	R-value	p-Value
(A) Eyes closed				
Anteroposterior				
Total	0.287	0.005	ns	ns
LF < 0.1 Hz	ns	ns	ns	ns
HF > 0.1 Hz	0.307	0.003	0.431	<0.001
Lateral				
Total	0.337	0.001	0.265	0.010
LF < 0.1 Hz	0.240	0.020	0.231	0.026
HF > 0.1 Hz	0.318	0.002	0.256	0.013
(B) Eyes open				
Anteroposterior				
Total	ns	ns	ns	ns
LF < 0.1 Hz	ns	ns	−0.0225	0.030
HF > 0.1 Hz	ns	ns	ns	ns
Lateral				
Total	ns	ns	ns	ns
LF < 0.1 Hz	ns	ns	ns	ns
HF > 0.1 Hz	0.227	0.029	0.255	0.014

indications that torque variance on MF was larger than on SF (Fig. 3).

Vision significantly reduced anteroposterior and lateral torque variance on all compliant surfaces and in all spectral separations. One important result to note was the interaction effect of the type of foam and vision significantly affected total and high frequency anteroposterior torque variance ($p < 0.05$). This shows that vision was more important for postural control when the foam properties evoked the largest torque variance.

3.4. Relationship between torque variance, foam density and foam elastic modulus

Surface density and the elastic modulus correlated similarly with torque variance. The significant correlation results shown in Table 3 indicate that in some cases, both surface density and elastic modulus were positively related to torque variance, i.e. the larger the density or elastic modulus of the compliant surface, the larger torque variance.

4. Discussion

4.1. Foam properties and stability

The importance of cutaneous mechanoreceptive information in accurately maintaining postural stability is well-known [23,27], and the typical reason for using foam in posturography studies is to cause a challenging disruption of sensory information at the point of contact with the

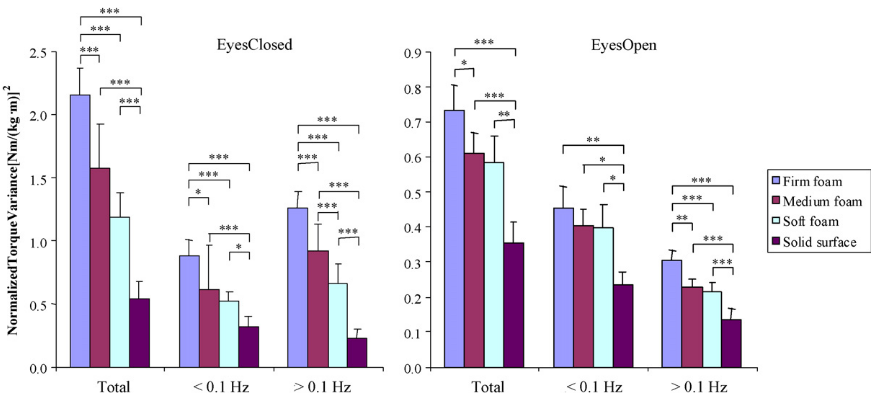

Fig. 3. Lateral torque values with eyes closed and eyes open (mean and S.E.M.).

supporting surface. However, some observations suggest that disruption of sensory information might not be the only cause for decreased postural stability when standing on a compliant foam surfaces. For example, in this study the recorded torque variance at the support surface in the anteroposterior plane was between 60% and 105% larger while standing on foam then while standing on a solid surface with eyes closed. In comparison, Stal et al. found that the torque variance was 30% lower when the mechanoreceptors of the feet were desensitised by hypothermia than during normal quiet stance on a solid surface [23]. Other studies have shown that the torque variance is only about 20% higher in elderly subjects with substantially decreased sensation of the receptors on the soles of the feet compared with normal subjects [21]. Additionally, in this study and in line with our hypothesis, we found a significant correlation of the foam density and elastic modulus (i.e. how much the foam compresses under force) with the level of recorded postural instability. The recorded torque variance was greater when the subjects stood on a foam block with a higher elastic modulus (i.e. which compressed less under force) than on foam blocks with a low elastic modulus. These findings support previous suggested notions that standing on foam causes two different effects: a decreased ability to accurately detect pressure distribution and body orientation [8,9] and a decreased ability to exert accurate, corrective responses due to the visco-elastic nature of foam blocks [10,11].

Firm foam offers more mechanical resistance to a downward force than soft foam, shown by the high elastic modulus value in Table 1. Thus, it is reasonable to assume that firm foam absorbs more of the corrective torque forces applied on it than soft foam. Additionally, there would be a greater compression of the block when standing on soft foam compared with firm foam, which might have enabled subjects to come into closer contact with the rigid surface beneath the foam. This was confirmed by the smaller elasticity modulus of the foams and the subjective testimonies of the test subjects. Subjects might therefore gain an advantage on softer foam surfaces through an opposing resistance, from which accurate corrective movements could be produced at a lower force cost. Larger compression of the foam surface, allowing the subjects to partly sense the hard surface beneath the foam, might also make sensory information from the mechanoreceptors on the soles of the feet more accurate. Hence, it might be easier to both sense body orientation and exert corrective body movements while standing on soft foam than on firm foam. These observations would explain many of our results, including why we detected the largest differences in the high frequency range. Similarly, other reports have shown that foam surfaces increase high frequency postural sway [19,28]. Although we observed increased instability while standing on foam blocks with a higher elastic modulus than on a lower elastic modulus, this may not generally hold true for all types of foam. For example, when standing on a foam block with an extremely high elastic modulus, its high rigidity might improve the effectiveness of ankle torque thereby stabilising posture.

Numerous researchers have used foam blocks to investigate the role of mechanoreceptive information in human postural stability [8,14,19,29]. However, as the nature and properties of the foam material tends to differ between these studies, there is a possibility that the results obtained are partly influenced by the properties of the foam used, and thus, not entirely by the physiological factors under investigation. Hence, it may not be possible to accurately compare results between studies. Compliant surfaces such as foam are also used clinically to determine balance deficits. Therefore, precaution should be taken when analysing these results, because what may initially seem to be due to a balance deficit might instead be a response for that type of foam material.

4.2. *Effect of surface properties in the lateral direction*

Our findings showed that the foam properties affected the lateral torque variance more than anteroposterior torque variance. The findings in the present study showed that even though movement is more restricted in the lateral direction, a seemingly unthreatening perturbation to normal subjects, such as standing on foam, can induce prolific changes of stability in the lateral direction. One possible reason for the larger effects of the foam properties in the lateral plane could be that subjects were unable to distribute their CoP forces evenly between their feet on the compliant surfaces, and therefore they tended to be more unbalanced in the lateral direction causing a 'to-and-fro' like movement. Another reason could be that the inherent shape of the feet, i.e. larger foot length than foot width, and the additional support given by the toes, may have ensured that postural stability was regulated more accurately on foam in the anteroposterior direction than in the lateral direction.

4.3. *Surface properties and vision*

Vision is known to help maintain postural stability especially when another sensory system is compromised [28,30]. In line with this, we found that vision decreased torque variance, especially in the high frequency range on all foam blocks, which shows that imbalances caused by compliant surfaces can be quickly detected and prevented when the eyes are open [19,31]. Our findings also showed a combined effect of vision and the compliant foam properties, where the increase in total and high frequency anteroposterior torque variance caused by the compliant surface properties was increased further by the absence of visual information. Hence, the stabilising effect of vision in the anteroposterior direction was of more importance when the subjects' stability was increasingly challenged by the support surface, as illustrated by the higher torque variance values. Clinically, this would imply that when evaluating balance deficits using compliant surfaces, the absence or presence of visual information might affect recorded torque variance differently depending on the properties of the foam block.

4.4. Torque variance, foam density and foam elastic modulus

In the present study, the medium foam was less dense than the soft foam, though its elastic modulus value was markedly larger than for the soft foam (Table 1). This may explain why we did not find identical relationships between torque variance and density and between torque variance and elastic modulus (Table 3). Instead, our findings showed that both factors should be considered when using different kinds of foam to investigate postural stability.

5. Conclusions

Our findings showed that torque variance above 0.1 Hz is markedly larger on firmer foam surfaces compared with on softer foam surfaces, and this appears to be partly related to the decreased ability to exert corrective movements on compliant surfaces with a high density or high elastic modulus value. Also, the stabilising effect of vision in the anteroposterior direction was of more importance when the subjects' stability was increasingly challenged by the support surface, as illustrated by the higher torque variance values. Therefore we suggest that precaution should be taken when evaluating postural control stability using foam surfaces with different compliance properties.

Conflict of interest

There was no conflict of interest for any of the authors in this study.

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Article II

The contribution of mechanoreceptive sensation on stability and adaptation in the young and elderly

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ORIGINAL ARTICLE

The contribution of mechanoreceptive sensation on stability and adaptation in the young and elderly

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Abstract The aim was to determine the contributions of foot mechanoreceptive sensation, vision and their interaction on postural stability during quiet stance, balance perturbations and adaptive adjustments. Postural stability was measured as anteroposterior torque variance in Young ($n = 25$, average age = 25.1 years) and Elderly subjects ($n = 16$, average age = 71.5 years) during repeated calf vibrations while standing with eyes open and closed. Sensation, recorded using vibration perception and tactile sensitivity, was poorer in elderly than young subjects. Sensation was of low importance for stability during quiet stance and the first 50 s of repeated vibrations, but was associated with stability during the last three 50 s periods of balance perturbations, suggesting that the mechanoreceptive sensation affected how well postural control could adapt to repeated balance perturbations. The findings suggest that clinicians should investigate whether patients with balance problems and poor adaptation have mechanoreceptive sensation deficits.

Keywords Postural stability · Balance · Somatosensation · Elderly · Vision

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Introduction

The maintenance of human postural control involves sensory information from the visual, vestibular and proprioceptive receptors, and includes somatosensory information from cutaneous mechanoreceptors located on the soles of the feet. These mechanoreceptors can provide detailed spatial and temporal information about contact pressures on the foot (Vallbo and Johansson 1984), and are thus vital for sensing changes to body orientation (MacLellan and Patla 2006).

Ageing is known to increase the likelihood of falling and various studies have shown that the manifestations include postural instability determined using posturography recordings. As age increases, the incidence of falls also increases (Prudham and Evans 1981), and it has been estimated that total fall related costs could exceed $32 billion in the US alone by the year 2020 (Englander et al. 1996). It is therefore essential that risk factors associated with falls is captured early in order to begin administering simple, low-cost therapies before a fall occur. One of the major risk factors for falling in the elderly is decreased somatosensory function in the feet (Magnusson et al. 1990a) which can be tested clinically by determining sensation from plantar cutaneous mechanoreceptors on the soles of the feet. Various reports have demonstrated that ageing decreases the function of the plantar mechanoreceptors (Cauna and Mannan 1958; Verrillo et al. 2002). Accordingly, it has been found that clinical tests of plantar cutaneous sensation such as vibration perception and tactile sensitivity are affected by ageing (Perry et al. 2000). Previous studies have shown a negative correlation between sensation and postural sway during quiet stance in the elderly (Hughes et al. 1996; Prince et al. 1997; Menz et al. 2005). However, since unexpected, externally induced, balance perturbing forces often cause falls in the elderly (Fransson et al. 2004), investigating the association between sensation and perturbed balance might be more relevant and might better show whether sensation can be used as a marker of postural performance.

One commonly used method for externally inducing balance perturbations involves vibrating skeletal muscles or muscle tendons of the calf. This result in an increased activation of muscle spindle afferents signaling to the central nervous system that the vibrated muscle is being stretched (Matthews 1986). The increased activity induced from muscle spindles results in a proprioceptive illusion of movement. Shortly after muscle vibration, tonic stretch reflexes are elicited that result in increased anteroposterior body movements (Patel et al. 2008). Though when repeated, muscle vibration can evoke postural adaptation in both young (Fransson et al. 2007b) and elderly (Fransson et al. 2004) subjects which markedly reduces the likelihood of imbalance and hence falls. From the perspective of equilibrium training to prevent falls, evaluating postural control adaptability in the elderly is very important (Fujiwara et al. 2007). The aim of this study was to investigate the contributions of mechanoreceptive sensation, vision and their interactions on postural performance and to investigate whether these factors affected postural control adaptation.

The hypotheses for this study were that the mechanoreceptive sensation in the feet would be associated with the postural performance and that the mechanoreceptive sensation would influence the degree to which subjects could adapt to repeated balance perturbations.

Methods

Subjects

Two groups of subjects were used in this study, a Young group and an Elderly group. The Young group comprised 25 (13 female and 12 male) healthy volunteers aged between 19 and 41 years [mean 25.1 years, standard deviation (SD) 4.6 years; mean mass 68.8 kg, SD 13.3 kg; and mean height 175 cm, SD 9 cm]. The Elderly Group comprised 16 healthy volunteers (5 female and 11 male) aged between 64 and 79 years [mean 71.5 years, SD 3.9 years; mean mass 79.8 kg, SD 12.1 kg; and mean height 166 cm, SD 8 cm]. No subject had previously experienced neurological disease or injury to the legs, nor were any taking medication and all were asked to refrain from alcohol at least 48 h prior to testing. At the time of experimentation, no subject was taking any form of medication and signed consent was obtained. The experiments were performed in accordance to the Helsinki declaration of 1975 and approved by the local ethical committee.

Equipment

A custom-built force platform recorded the forces with six degrees of freedom and with an accuracy of 0.5 N. A customized computer program controlled the vibratory stimulation and sampled the force platform data at 50 Hz (see Fig. 1).

The vibrators had vibration amplitude of 1.0 mm and frequency of 85 Hz (Fransson et al. 2004), were 6 cm long and 1 cm in diameter and were placed over the gastrocnemius muscles and secured by elastic straps.

Procedure

Sensitivity assessment

Vibration perception of the plantar surface was measured using a biothesiometer electronic device (Model EG electronic BioThesiometer, Newbury, OH, USA) that generated a 200 Hz vibration of varying amplitude (in µm). The vibration was applied to the plantar surface of the first distal phalanx (big toe), the fifth distal phalanx (little toe), the first proximal phalanx (base of big toe), the fifth proximal phalanx (base of little toe) and the tuberosity of calcaneous (heel). Subjects were asked to indicate to the examiner whether they were able to feel the vibration "Yes" or "No". Three readings in ascending intensity and three readings in descending intensity were made until the subject could no longer feel the vibration, and an average of these six measurements was recorded as the vibration threshold (Lord et al. 2003).

Tactile sensitivity was measured with a Semmes–Weinstein pressure aesthesiometer (Semmes–Weinstein Monofilaments, San Jose, USA). The aesthesiometer comprised eight nylon filaments of equal length, with varying diameter. The filaments were applied to the plantar surface of first distal phalanx (big toe), the fifth distal phalanx (little toe) and the tuberosity of calcaneous (heel). Subjects were instructed that when the filament was placed on any of the positions above, the examiner would say "big toe", "little toe" or "heel", and if they felt the filament in contact with

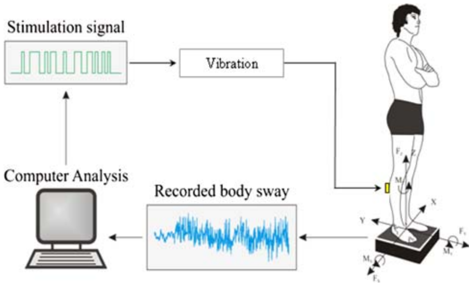

Fig. 1 Schematic illustration of the test setup

the skin, they must report to the examiner whether they felt it on the "big toe", "little toe" or "heel." Tactile threshold was determined by presenting suprathreshold filaments initially, then applying thinner and thinner filaments until the subject could no longer detect them (Lord et al. 2003). The examiner then applied thicker filaments until the filament was detected. The touch threshold was determined from 3 ascending and descending steps (Lord et al. 2003) and is presented in the table as monofilament diameter size (mm).

Posturography assessment

Each subject stood barefoot on the force platform in a relaxed posture with arms folded across the chest (see Fig. 1). The subject's heels were 3 cm apart and feet positioned at an angle of 30° along guidelines on the platform. Subjects were instructed to focus on a target 1.5 m in front of them at eye level or keep their eyes closed depending on the test condition. The subjects listened to music through headphones in order to reduce possible movement references from external noise sources and to avoid extraneous sound distractions (Fransson et al. 2007a).

The following two tests were performed in a randomized order, using a Latin Square design, by all subjects.

- Vibration of the calf muscles with eyes closed (EC Calf).
- Vibration of the calf muscles with eyes open (EO Calf).

Before the vibration commenced, a 30-s period of Quiet Stance was recorded. The vibratory stimulation pulses were applied according to a pseudorandom binary sequence (PRBS) (Johansson 1993) during a period of 205 s making each test 235 s long, where both pulses and inter-pulses had random durations from 0.8 to 6.4 s, yielding an effect bandwidth of the vibratory stimulation within 0.1–2.5 Hz. A 5-min rest period was given to the subjects between each test.

Analysis

Vibratory calf muscle stimulation induces body movement primarily in the anteroposterior direction. Therefore, only responses in this direction are considered here. Postural stability while standing is commonly analyzed using force platforms and the movements of the centre of pressure (CoP), i.e., the point of application of the ground reaction force. We analyzed the variance of the anteroposterior torque values, where the torque τ is calculated from the formula $\tau = \mathrm{CoP} \times \mathrm{F}_z$; where $\mathrm{F}_z \approx m \times g$; where m = the assessed subjects mass (in kg) and g = gravitational constant 9.81 (in m/s^2), see Fig. 1. F_z will fluctuate slightly due to present body leaning, and when the subject applies additional forces to the surface to accelerate or decelerate a movement. Hence, changes in recorded torque are equivalent to changes in CoP (Patel et al. 2008). One benefit with presenting torque variance values is that the calculated value corresponds directly with the energy used towards the support surface to maintain stability (Magnusson et al. 1990b).

Each test was divided into five periods: Quiet Stance (0–30 s), and four 50 s stimulation periods (period 1: 30–80 s; period 2: 80–130 s; period 3: 130–180 s; period 4: 180–230 s). Of note, the vibration sequence is randomized but each 50 s period contain similar amount and kind of stimulation (similar amount of long and short vibration pulses) validated by FFT-analysis of spectral contents in the stimulation. Torque variance values were normalized for individual anthropometrical factors using the subject's squared height and squared mass (Patel et al. 2008), thus providing inter-individual compensation for individual variations in height and mass. The squared nature of the variance algorithm required normalization using squared parameters for unit agreement (Patel et al. 2008). Vibration perception and tactile sensitivity was recorded from both feet and then averaged.

Statistics

Normality of distribution was tested with the Shapiro–Wilk test. Non-parametric statistics were used since some values were not normally distributed. The Mann–Whitney (Exact sig. 2-tailed) (Altman 1991) was used for the statistical comparison of anteroposterior torque variance between the two groups. In addition, the effects of sensation, vision and their interaction on anteroposterior torque variance during the Quiet Stance period and in each of the vibration Periods was analyzed using a GLM univariate ANOVA (General Linear Model univariate Analysis of Variance) test on log-transformed values (Altman 1991). In addition, a GLM univariate ANOVA was used to investigate the effects of mechanoreceptive sensation, vision, the period of vibration and their interactions on anteroposterior torque variance during vibration. In the GLM ANOVA tests, the data from the elderly and young groups were combined to increase the range of data. The accuracy of the GLM model was evaluated by testing whether the model residuals were distributed normally. The statistical analysis was carried out with Bonferroni correction for multiple comparisons and the statistical significance levels were set to $P < 0.01$ for Mann–Whitney tests and $P < 0.05$ for GLM ANOVA tests (Altman 1991).

Results

Age and mechanoreceptive sensation

Recordings of sensitivity showed that the vibration threshold for vibration perception was significantly better for the Young Group compared with the Elderly Group at every position (see Table 1). The tactile threshold for tactile sensitivity was also better for the Young Group compared with the Elderly Group, though there was no difference at the heel.

Torque variance results

In the Quiet Stance period, there was no significant difference in torque between the Elderly Group and the Young Group with EC Calf or EO Calf. Anteroposterior torque variance was larger for the Elderly Group in all vibration Periods with EC Calf and EO Calf (Table 2).

GLM ANOVA for vision and mechanoreceptive sensation in Quiet Stance

The analysis in Table 3 showed that in the young and elderly, mechanoreceptive sensation did not affect Quiet Stance torque variance. Vision significantly affected torque variance. With eyes closed, Quiet Stance torque variance was larger than with eyes open. Moreover, there was no interaction between vision and mechanoreceptive sensation in the young and elderly.

GLM ANOVA for vision, mechanoreceptive sensation and period during vibration

GLM analysis in Table 4 revealed that the availability of visual information, the threshold level of mechanoreceptive sensation in the vibration perception and tactile sensitivity tests, and the Period of vibration, all significantly affected torque variance in the young and elderly. Torque variance was larger with eyes closed compared with eyes open, larger when the threshold level of mechanoreceptive sensation was larger and was also larger during the initial periods of vibration, an indication that torque variance decreased adaptively over time. However, we found no interactions between these factors.

Table 1 Mean and standard deviation (SD) of mechanoreceptive sensation for the Elderly and Young groups and *P* values indicating whether there was a difference between the groups

	Mechanoreceptive sensation		
	Elderly mean (SD)	Young mean (SD)	*P* value
Vibration perception (μm)			
Big toe	4.78 (3.84)	0.68 (0.65)	<0.001
Little toe	4.78 (3.31)	0.62 (0.73)	<0.001
Base of big toe	6.01 (5.71)	0.51 (0.43)	<0.001
Base of little toe	6.45 (3.86)	0.47 (0.45)	<0.001
Heel	7.85 (7.68)	0.39 (0.26)	<0.001
Tactile sensitivity (mm)			
Big toe	4.14 (0.42)	3.39 (0.38)	<0.001
Little toe	4.08 (0.40)	3.27 (0.36)	<0.001
Heel	3.89 (0.56)	3.75 (0.37)	NS

Table 2 Mean and standard error (SEM) of normalized torque variance for the Elderly and Young groups

	Elderly	Young	*P* value
	Mean (SEM) $[Nm/(Kg\ m)]^2$	Mean (SEM) $[Nm/(Kg\ m)]^2$	
Eyes closed (EC) calf vibration			
Quiet stance	1.18 (0.23)	0.79 (0.10)	NS
Period 1	14.43 (1.65)	6.74 (0.97)	<0.001
Period 2	10.14 (1.26)	3.72 (0.47)	<0.001
Period 3	1.51 (1.11)	4.24 (0.56)	<0.001
Period 4	8.56 (0.88)	3.48 (0.49)	<0.001
Eyes open (EO) calf vibration			
Quiet stance	0.76 (0.55)	0.46 (0.07)	NS
Period 1	7.90 (4.48)	3.67 (0.82)	<0.001
Period 2	5.17 (4.23)	1.65 (0.18)	<0.001
Period 3	5.68 (3.95)	1.96 (0.24)	<0.001
Period 4	4.80 (2.66)	1.51 (0.13)	<0.001

The *P* values indicate where there was a significant difference between the groups. The periods were: Quiet Stance (0–30 s), and four 50 s stimulation periods (period 1: 30–80 s; period 2: 80–130 s; period 3: 130–180 s; period 4: 180–230 s

Table 3 Statistical evaluation of the Quiet stance torque variance values using the GLM univariate ANOVA method showing the effect of vision, sensation and their interaction for vibration perception and tactile sensitivity tests (*P* values)

Torque variance	*P* value		
	Vision	Sensation	Vision × sensation
Vibration perception (μm)			
Big toe	0.003	NS	NS
Little toe	0.007	NS	NS
Base of big toe	0.005	NS	NS
Base of little toe	0.004	NS	NS
Heel	0.005	NS	NS
Tactile sensitivity (mm)			
Big toe	0.035	NS	NS
Little toe	NS	0.006	NS
Heel	NS	NS	NS

The notation “<0.001” means that the *P* value is smaller than 0.001, *NS* no significant difference

GLM ANOVA for vision and mechanoreceptive sensation for each period

GLM analysis in Table 5 shows that the influence of mechanoreceptive sensation changed over time. Mechanoreceptive

Table 4 Statistical evaluation of the torque variance values using the GLM univariate ANOVA method showing the effect of vision, mechanoreceptive sensation, period and their interactions for vibration perception and tactile sensitivity tests (*P* values)

Torque variance	*P* value						
	Vision	Sensation	Period	Vision x sensation	Vision × period	Sensation × period	Vision × sensation × period
Vibration perception (µm)							
Big toe	0.001	<0.001	<0.001	NS	NS	NS	NS
Little toe	0.001	<0.001	<0.001	NS	NS	NS	NS
Base of big toe	0.001	<0.001	<0.001	NS	NS	NS	NS
Base of little toe	0.001	<0.001	<0.001	NS	NS	NS	NS
Heel	<0.001	<0.001	<0.001	NS	NS	NS	NS
Tactile Sensitivity (mm)							
Big toe	<0.001	<0.001	0.001	NS	NS	NS	NS
Little toe	<0.001	<0.001	<0.001	NS	NS	NS	NS
Heel	<0.001	<0.001	<0.001	NS	NS	NS	NS

The notation "<0.001" means that the *P* value is smaller than 0.001, *NS* no significant difference

sensation were associated more with torque variance in Periods 2, 3 and 4, compared with Period 1 as shown by the exact *P* values. In the young and elderly, the contribution of vision was important in all Periods. However, there was no interaction between the two.

Discussion

One factor associated with an increased instability in the elderly is decreased mechanoreceptive sensation on the soles of the feet (Lord and Ward 1994). In line with this, we found that torque variance was markedly larger in the elderly and that both tactile sensitivity and vibration perception thresholds were considerably poorer. As expected, both mechanoreceptive tests results were significantly affected by age, except at the heel in the tactile discrimination test, and were significant determinants of torque variance during balance perturbations. Our sensitivity findings are in line with those in the literature. The tactile sensitivity values in the present study are similar to those previously also found by Perry et al. (2006). Perry et al. obtained average detection thresholds at 3.8 for the young adults and 4.7 for the older adults, whereas we obtained between 3.27 and 3.75 for the young adults and between 3.89 and 4.14 for the elderly. However, in contrast to Perry et al. (2006), who found the largest difference in sensation between the young and elderly adults at the heel, there was no difference at the heel in the present study. Our values for sensation detection threshold in the elderly are slightly lower than the ones by Perry especially at the heel (5.2 vs. 3.89 mm). The elderly in the present study had quite large inter-individual variability in sensation, with several of our elderly subjects scoring very well at the heel. One possible reason for this could be the foot care practice commonly performed by the elderly in this country which may increase sensitivity and account for the found difference. Thus, the effects of foot care prior to sensation assessments and posturography might be an aspect worth monitoring or controlled in further studies. Inter-individual differences could also be related to activity levels, since continuous impact at the heel often causes a toughening, which is likely to reduce sensitivity.

In the quiet stance period, sensation threshold scores were not as strongly associated with the recorded torque variance as vision. However, during balance perturbations, the sensation threshold, the vibration Period and vision had all significant influence on the recorded variance of torque values (Table 4). Postural performance was considerably better when subjects had lower vibration perception and tactile sensitivity thresholds, and the performance also improved over time through adaptation, which supports our previous findings showing that elderly (Fransson et al. 2004) and young subjects are able to enhance their stability when repeatedly perturbed (Fransson et al. 2007b). The observed adaptation effect is consistent with Corna et al. (1999) showing that within the first few cycles of perturbation, participants predict the characteristics of perturbations and their destabilizing effects, and set their balance control system to minimize these effects. Torque variance was also always larger with eyes closed than with eyes open, concurring with the previous literature regarding the importance of vision when sensory information from at least one other sensory receptor is unreliable (Fransson et al. 2004). However, there was no interaction between vision, sensation and the vibration period, suggesting that these factors act independently on postural control. Thus, for example, increased visual contribution seems not to be able to compensate for poor mechanoreceptive sensation.

When studying in detail, the effects of mechanoreceptive sensation and vision in each perturbation period, we sur-

Table 5 Statistical evaluation of the torque variance values using the GLM univariate ANOVA method showing the effect of vision, sensation and their interaction for each period (*P* values)

Torque variance	*P* value		
	Vision	Sensation	Vision × Sensation
Period 1			
Vibration perception (μm)			
Big toe	0.004	NS	NS
Little toe	0.001	0.009	NS
Base of big toe	0.011	NS	NS
Base of little toe	0.008	NS	NS
Heel	0.003	0.049	NS
Tactile sensitivity (mm)			
Big toe	0.011	NS	NS
Little toe	0.001	0.011	NS
Heel	0.006	0.009	NS
Period 2			
Vibration perception (μm)			
Big toe	0.001	0.041	NS
Little toe	0.001	0.011	NS
Base of big toe	0.001	0.010	NS
Base of little toe	0.003	NS	NS
Heel	0.000	0.003	NS
Tactile sensitivity (mm)			
Big toe	0.003	0.030	NS
Little toe	0.001	0.002	NS
Heel	0.006	0.015	NS
Period 3			
Vibration perception (μm)			
Big toe	0.001	0.041	NS
Little toe	0.001	0.013	NS
Base of big toe	0.001	0.002	NS
Base of little toe	0.004	NS	NS
Heel	0.001	0.012	NS
Tactile sensitivity (mm)			
Big toe	0.001	0.001	NS
Little toe	0.001	0.006	NS
Heel	0.003	0.001	NS
Period 4			
Vibration perception (μm)			
Big toe	0.001	0.016	NS
Little toe	0.003	NS	NS
Base of big toe	0.001	0.002	NS
Base of little toe	0.001	0.037	NS
Heel	0.001	0.001	NS
Tactile sensitivity (mm)			
Big toe	0.002	0.039	NS
Little toe	0.001	0.004	NS
Heel	0.007	0.001	NS

prisingly found that the mechanoreceptive sensation levels were more associated with the stability in vibration Periods 2, 3 and 4 than in vibration Period 1 (Table 5). This novel finding suggests that the initial responses to the balance perturbations were only partly influenced by the mechanoreceptive sensation, but the sensation level had a clear effect on how well postural control could adapt to the repeated balance perturbations in Periods 2, 3 and 4. In the present study, calf vibration provided a stimulus giving a false perception of movement. As such, the sensory information from some proprioceptors becomes 'unreliable' in terms of accurately portraying the actual movement of the body. Various findings suggest that when information from any of the balance receptors, i.e., the visual, vestibular and somatosensory, are temporarily disturbed or provide erroneous information due to lesions or disorder, the erroneous information can be overridden by the information from the more reliable receptors (Oie et al. 2002). During continuous balance perturbations using a randomized pseudorandom binary sequence of vibration pulses of varying durations, the perturbations became more predictable over time, and the postural challenge, although still threatening, became much more manageable. It has been suggested that within the first few perturbations, participants can predict the characteristics of the perturbations and their destabilizing effects, and set their balance control system to minimize these effects (Corna et al. 1999; Akram et al. 2008), and this is consistent with our findings. Furthermore, continued perturbed balance using repeated vibrations, makes the central nervous system respond with adaptive adjustments characterized by an alteration of the importance of the information from the different receptors, along with a change of the motor control strategy to reduce the likelihood of balance loss (Fransson et al. 2007b), and this has previously been found in both young and elderly subjects (Fransson et al. 2004). In the present study, this re-weighting of afferent information was evidenced in the elderly and the young as an increased postural stability after vibration Period 1 (Table 5). Since the limits of stability are set using sensory information, it is reasonable to assume that the mechanoreceptive sensation might influence the postural stability and set limits for how effective the adaption might get due to restricted sensory information. In line with our findings of poorer mechanoreceptive sensation with increasing age, it is well known that ageing decreases the function of the plantar mechanoreceptors (Cauna and Mannan 1958; Verrillo et al. 2002).

Based on the presented results, it is recommended that clinicians should investigate whether patients with balance problems and poor postural control adaptation have mechanoreceptive sensation deficits, and train these patients to effectively use the remaining sensory resources as effectively as possible. Previous reports have showed a strong correlation between muscle strength and postural stability in elderly and disabled subjects (Prince et al. 1997; Menz et al. 2005). Therefore, the relationship between muscle strength and torque variance might be stronger than the one we found between mechanoreceptive sensation and torque variance. However, the finding of a significant relationship between vibration perception and tactile sensitivity and torque variance during balance perturbations suggests that in constructing rehabilitation protocols for elderly subjects, a combined practice of developing musculoskeletal strength and plantar sensitivity, particularly vibration tests, should be employed. Maki et al. (2007) have previously assessed different techniques for improving postural stability in elderly subjects . They found that balance training using perturbations from a moving platform, specially designed insoles, handrails and walking aids improved postural stability. This was supported by Mansfield et al. (2007), who suggested that a possible method for improving postural stability in the elderly is to initiate a perturbation-based balance rehabilitation program using platform perturbations.

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Article III

The effects of ageing on adaptation during vibratory stimulation of the calf and neck muscles

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Effects of Ageing on Adaptation during Vibratory Stimulation of the Calf and Neck Muscles

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Key Words
Ageing · Attention · Adaptation · Balance · Vision

Abstract
Background: The ability to adapt and habituate based on prior experiences is important for human movement control, fall prevention and for the ability to enhance performance during various human activities. However, little is known about the ability for the elderly to adapt to balance perturbations in the lateral direction. ***Objective:*** To determine whether adaptation, i.e., the ability to adjust postural control to handle balance perturbations better over time, differed in the elderly subjects compared with young subjects in the anteroposterior and lateral directions, and whether the site of the balance perturbation or the presence or absence of vision affected the response. ***Methods:*** Postural stability was measured as anteroposterior and lateral torque variance in a young group (n = 18 (9 female and 9 male), average age = 29.1 years) and an elderly group (n = 16 (5 female and 11 male), average age = 71.5 years) with eyes open and closed during balance perturbations from calf and neck vibrations. After a 30-s period of quiet stance, these vibrations were repeated over a period of 200 s, so the adaptive responses could be analyzed by splitting the data into 50-s periods. ***Results:*** The adaptive responses in the anteroposterior and lateral directions were different. Adaptation in the anteroposterior direction occurred to an almost equal extent in the elderly and young, whereas adaptation in the lateral direction was markedly larger in the elderly in all tests except for neck vibration with eyes closed. Age, vision and vibration site were all influential factors for recorded body movements, but no significant combined effects were found. ***Conclusion:*** Balance perturbation instigates an adaptive response in the elderly in both the anteroposterior and lateral directions. However, during perturbation, age and vision are both very influential factors for the stability, thus associating the previously documented age-related decline in visual functioning with a higher risk of falls in this age range.

Introduction

Ageing is known to increase the likelihood of falling [1] and various studies have shown that one of the characteristics of ageing includes an increased postural sway during unperturbed standing [2, 3]. Approximately one-third of elderly individuals report a fall each year [3] and the result is often injuries such as wrist and hip fractures [4]. Estimates suggest that the accumulative cost of fall-related injuries in the elderly could exceed USD 32 billion by the year 2020 in the USA alone [5]. The deficits in postural stability in the elderly have been attributed to a number of causes including a deterioration of the sensory

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and motor systems. For example, ageing causes deterioration of the visual system [6], proprioception in the lower legs [7, 8], somatosensory information from the soles of the feet [9], vestibular dysfunction [10, 11], and reduced muscle strength [12].

As it has been reported that unexpected, externally-induced, balance-perturbing forces are often the cause of falls in the elderly [13], assessing the ability of the elderly to regain balance once perturbed might be of greater relevance to fall prevention than assessments of quiet standing. One method commonly used to perturb balance through the somatosensory system is vibration of skeletal muscles or tendons, such as those at the calf or neck [14]. This increases the afferent signals from the muscle spindles [15] and creates a proprioceptive illusion that the vibrated muscle is being stretched [16]. This perceived stretch induces tonic stretch reflexes intended to return the vibrated muscle to its perceived original length [17]. Vibration of the neck or calf muscles often induces body movements primarily in an anterior-posterior direction [18], though there are often increases in the lateral direction also [19]. When these vibrations are repeated, the body movements are often accompanied, over time, by a gradual adaptive alteration of the average centre of pressure position more posterior than the original position, suggesting that the subjects adopt a slightly more forward-leaning posture [19, 20]. These adaptive responses decrease the likelihood of falling and the reactive responses required to maintain stability while exposed to balance perturbations [21]. Adaptation can be defined as a series of changes initiated to adjust a process to function more adequately to a new or changed environment, in this case an ability to adjust postural control to handle repeated balance perturbations better over time. It is widely acknowledged that vibration of different muscle groups puts subjects into a new postural condition [22, 23] and thus requires different controlling strategies. Moreover, the neck and calf muscles have different proprioceptive roles in the maintenance of postural stability. Neck muscle afferents are mainly involved in the regulation of body orientation [24], whereas the calf muscles are mainly involved in the maintenance of equilibrium [24]. Therefore, one might expect different results on postural control when vibrating different muscle groups.

Previous studies of postural stability in the elderly have perturbed standing using calf or neck muscle vibration and have found significant differences in postural stability [20, 25]. In one of these studies, Fransson et al. [20] demonstrated that elderly subjects were able to show significant levels of postural adaptation in the anteroposterior direction when balance perturbations from calf muscle vibration were repeated. However, as some reports have shown that lateral instability is mostly affected in the elderly [26–28], the effects of ageing on postural adaptation might be different in the lateral direction compared with the anteroposterior direction.

The aim of this study was to investigate the effect of ageing on the adaptive postural control changes to handle neck and calf muscle vibration better. We also wished to investigate whether ageing changed the effect of vibration applied to calf or neck muscles, the use of vision and their interaction. We hypothesized that the stability of the elderly would be affected more by proprioceptive vibratory stimulation of the calf and neck muscles in the lateral direction compared with young subjects, as lateral stability appears to be particularly affected by age. We hypothesized further that the elderly would have a greater reliance on visual information for postural control, since it is believed that older adults rely more heavily on visual input when somatosensory information is altered compared with young subjects [29].

Methodology

Subjects

Two groups of subjects were recruited in this study, a young group and an elderly group. The young group comprised 18 healthy subjects (9 men and 9 women) aged between 18 and 49 years (mean age 29.1 ± 7.8; mean height 1.74 ± 0.1 m; mean mass 73.4 ± 10.8 kg). The elderly group comprised 16 healthy volunteers (5 female and 11 male) aged between 64 and 79 years (mean 71.5 ± 3.9; mean mass 79.8 ± 12.1 kg; mean height 166 ± 8 cm). No subject had previously experienced balance problems, neurological disease or a significant injury to the legs, nor were any taking medication and all were asked to refrain from alcohol at least 48 h prior to testing. At the time of experimentation no subject was taking any form of medication and signed consent was obtained before any testing began. The experiments were all performed in accordance to the Helsinki Declaration of 1975 and approved by the local ethical committee.

Equipment

The vibrators had a vibratory amplitude of 1.0 mm and a vibration frequency of 85 Hz. The vibration was produced using a revolving DC-motor (Escap, Geneva, Switzerland) equipped with a 3.5-g mass attachment contained within a cylindrical plastic coating with dimensions of 6 cm in length and 1 cm in diameter. The vibrators were placed either over the belly of the gastrocnemius muscles of both legs and secured by elastic straps around the legs or placed over the paravertebral neck muscles on both sides of the neck and held in place by a custom-made collar, depending on the test condition.

A force platform, developed in cooperation with the Department of Solid Mechanics, Lund Institute of Technology, recorded

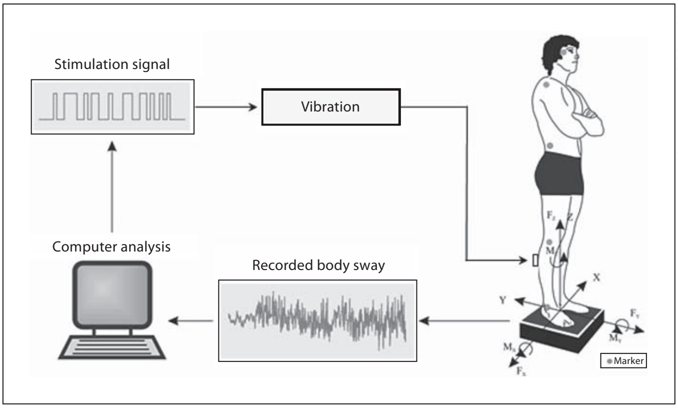

Fig. 1. Schematic illustration of the test setup.

the forces actuated at the feet with 6 degrees of freedom with an accuracy of 0.5 N. A customized computer program controlled the vibratory stimulation, and sampled the force platform data at 50 Hz.

Procedure

Each subject stood barefoot on the force platform in a relaxed posture with arms folded across the chest (fig. 1). The subject's heels were 3 cm apart and feet positioned at an angle of 30° open to the front along guidelines on the platform. Subjects were instructed to focus on a target (6 × 4 cm image) 1.5 m in front of them at eye level or keep their eyes closed depending on the test condition. The subjects listened to music through headphones in order to reduce possible movement references from external noise sources and to avoid extraneous sound distractions [30].

The following four tests were performed once by all subjects in a randomized order, the randomization was ensured by using a Latin square design: (1) vibration of the calf muscles with eyes closed (EC-Calf); (2) vibration of the calf muscles with eyes open (EO-Calf); (3) vibration of the neck muscles with eyes closed (EC-Neck), and (4) vibration of the neck muscles with eyes open (EO-Neck).

Before the vibration commenced, a 30-s period of quiet stance was recorded. The vibratory stimulation pulses were applied according to a pseudorandom binary sequence (PRBS) [31] during a period of 205 s making each test 235 s long, where both pulses and inter-pulses had random durations from 0.8 to 6.4 s, yielding an effect bandwidth of the vibratory stimulation within 0.1–2.5 Hz. A 5-min rest period was given to the subjects between each test.

Analysis

Postural stability while standing is commonly analyzed using force platforms and the movements of the centre of pressure (CoP), i.e., the point of application of the ground reaction force. We analysed the variance of the torque values, where the torque x is calculated from the formula x = CoP · Fz; where Fz ≈ m · g; where m = the assessed subject's mass (in kg) and g = gravitational constant 9.81 (in meter/s^2), see figure 1. Fz will fluctuate slightly due to present body leaning and when the subject applies additional forces to the surface to accelerate/decelerate a movement. Hence, changes in recorded torque are equivalent to changes in CoP [30]. The formula for variance is given by:

$$\bar{x} = \sum_{i=1}^{n} \frac{x(i)}{n}$$

$$\text{var } x = \frac{1}{n-1} \sum_{i=1}^{n} \left(x(i) - \bar{x} \right)^2$$

One benefit of presenting torque variance values is that the calculated value corresponds directly to the energy exerted against the support surface to maintain stability [32].

Anteroposterior and lateral movements were analyzed in terms of variance of torque from the force platform recordings [19]. Each test was divided into five periods: quiet stance (0–30 s), and four 50-s stimulation periods (period 1: 30–80 s; period 2: 80–130 s; period 3: 130–180 s; period 4: 180–230 s). Torque variance values were normalized for individual anthropometrical factors using the subject's squared height and squared mass [30].

Statistical Analysis

Normality of distribution was tested with the Shapiro-Wilk test. Non-parametric statistics were used since some values were not normally distributed. The Mann-Whitney (Exact sig. 2-tailed) [33] was used for the statistical comparison between the two groups. In the Mann-Whitney analysis, p values <0.05 were considered statistically significant. The Wilcoxon matched-pairs signed-rank test (Exact sig. 2-tailed) [33] was used for the analysis of variations between quiet stance and period 1 and between period 1 and period 4 in the two groups [19, 34, 35]. The torque variance changes between quiet stance and period 1 were evaluated to determine how the assessed parameters were initially affected by the balance perturbations evoked by vibratory proprioceptive stimulation compared to the activity during quiet stance. The

Table 1. Anteroposterior torque variance differences between elderly and young

	Eyes closed (EC)			Eyes open (EO)		
	elderly	young	p value	elderly	young	p value
A. Calf stimulation						
Quiet stance	1.18 (0.23)	0.80 (0.22)	p = 0.046	0.76 (0.14)	0.57 (0.10)	NS
Period 1	14.43 (1.65)	10.25 (1.46)	NS	7.90 (1.12)	4.34 (0.55)	p = 0.030
Period 2	10.14 (1.26)	6.54 (1.02)	p = 0.017	5.17 (1.06)	3.38 (0.78)	p = 0.039
Period 3	10.51 (1.11)	6.96 (0.95)	p = 0.015	5.68 (0.99)	3.84 (0.68)	NS
Period 4	8.56 (0.88)	5.18 (0.88)	p = 0.006	4.80 (0.67)	2.90 (0.44)	p = 0.033
B. Neck stimulation						
Quiet stance	1.38 (0.24)	0.78 (0.14)	p = 0.045	0.69 (0.16)	0.41 (0.05)	NS
Period 1	5.41 (0.94)	3.18 (0.74)	p = 0.023	3.32 (0.70)	2.00 (0.44)	NS
Period 2	2.78 (0.48)	1.45 (0.22)	p = 0.025	1.80 (0.44)	1.01 (0.20)	p = 0.037
Period 3	2.40 (0.44)	1.86 (0.37)	NS	1.51 (0.27)	1.18 (0.27)	NS
Period 4	2.55 (0.43)	1.36 (0.22)	p = 0.041	1.54 (0.35)	0.93 (0.31)	p = 0.017

Statistical evaluation of the torque variance differences in anteroposterior direction between elderly and young subjects while submitted to calf stimulation (A) or neck stimulation (B) with eyes closed (EC) or eyes open (EO). In the tables, the torque variance means and SEM for the elderly group and young group are presented as well as significant differences between the elderly and young groups (NS = non-significant difference). The torque variance values for the elderly group was larger than for the young group in all significant cases.

torque variance changes between period 1 and period 4 were evaluated to determine how the assessed parameters were affected by repeated vibratory stimulation, quantifying possible effects of adaptation to vibratory proprioceptive stimulation [19, 34, 35].

The Wilcoxon statistical analysis was carried out with Bonferroni correction for multiple comparisons. Of note, no more than two matched pairwise comparisons were performed for each single dataset. In the Wilcoxon analysis, p values <0.01 were considered statistically significant, though we present p values <0.05 in the figures and tables for consistency. In addition, the effects of the site of vibration, vision, age and their interactions on recorded torque variance were analyzed using a GLM univariate ANOVA (general linear model univariate analysis of variance) test on log-transformed values [33]. The accuracy of the GLM model was evaluated by testing whether the model residuals were distributed normally. In the GLM analysis, p values <0.05 were considered statistically significant.

Results

Anteroposterior Torque Variance

Elderly and Young Comparison

Table 1 shows that anteroposterior torque variances were generally larger for the elderly group, though the differences were not statistically significant in some periods.

GLM ANOVA of the Site of Vibration, Vision and Age

GLM analysis showed that vibration site, vision and age all significantly affected the recorded torque variance during all stimulation periods (table 2). Calf vibration evoked larger anteroposterior torque variance than neck vibration in all vibration periods ($p < 0.001$). Vision decreased anteroposterior torque variance in all periods (QS, $p < 0.001$; periods 1–4 $p < 0.001$). Age increased anteroposterior torque variance in all periods (QS, $p = 0.002$; periods 1–4 $p \leq 0.002$). However, there was no interaction between the vibration site, vision and age during any period. The interactions show whether the effect of two or more variables is not simply additive, but an additional effect might be produced by certain variable combinations.

Adaptation of Anteroposterior Torque Variance

In all tests, during the first period of vibratory stimulation (period 1), torque variance was significantly larger compared to the quiet stance torque variance for the elderly group and the young group ($p < 0.001$) (fig. 2). The percentage increases in torque variance for the elderly group were: 1,123% (EC-Calf), 939% (EO-Calf), 292% (EC-Neck), and 381% (EO-Neck). The percentage increases in torque variance for the young group were almost the same compared with the elderly group in each

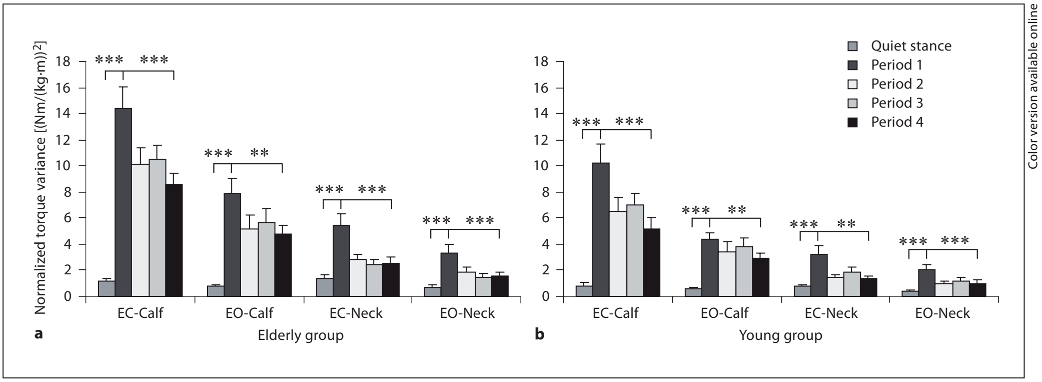

Fig. 2. Anteroposterior torque variance for the elderly group (**a**) and young group (**b**) for calf vibration with eyes closed (EC-Calf), calf vibration with eyes open (EO-Calf), neck vibration with eyes closed (EC-Neck) and neck vibration with eyes open (EO-Neck). ** $p < 0.01$; *** $p < 0.001$.

Table 2. General linear model analysis of torque variance in the anteroposterior direction

	Vibration site	Vision	Age	Vibration site × vision	Vibration site × age	Vision × age	Vibration site × vision × age
Quiet stance	NS (0.3)	p < 0.001 (12.2)	p = 0.002 (10.3)	NS (1.2)	NS (0.1)	NS (0.5)	NS (0.1)
Period 1	p < 0.001 (89.8)	p < 0.001 (26.2)	p < 0.001 (22.9)	NS (1.2)	NS (0.2)	NS (0.0)	NS (0.5)
Period 2	p < 0.001 (118.7)	p < 0.001 (25.0)	p < 0.001 (18.0)	NS (1.9)	NS (0.1)	NS (0.0)	NS (0.1)
Period 3	p < 0.001 (131.1)	p < 0.001 (20.1)	p = 0.002 (9.8)	NS (1.3)	NS (0.2)	NS (0.0)	NS (0.0)
Period 4	p < 0.001 (118.0)	p < 0.001 (21.9)	p < 0.001 (24.2)	NS (0.0)	NS (0.1)	NS (0.0)	NS (0.0)

Effects of the site of vibration, vision, age and their interactions on recorded anteroposterior torque variance analyzed using a GLM univariate ANOVA (general linear model univariate analysis of variance) test on log-transformed values. Mean and SEM values for the torque variance data analyzed are presented in table 1. F values are presented in parentheses.

test, the increases were: 1,181% (EC-Calf), 661% (EO-Calf), 308% (EC-Neck), and 388% (EO-Neck).

After the initial rise in torque variance in period 1, torque variance decreased and stayed low between periods 2–4. The extent of adaptation to the vibratory stimulation is presented as percentage decrease in torque variance between period 1 and period 4. The decreases in torque variance for the elderly group were: 41% (EC-Calf, $p < 0.001$), 39% (EO-Calf, $p = 0.001$), 53% (EC-Neck, $p < 0.001$), and 54% (EO-Neck, $p < 0.001$). The percentage decreases for the young group were again almost the same compared with the elderly group in each test: 49% (EC-Calf, $p < 0.001$), 33% (EO-Calf, $p < 0.01$), 57% (EC-Neck, $p < 0.01$), and 54% (EO-Neck, $p < 0.001$).

To summarize, stability in the anteroposterior direction in elderly subjects was clearly challenged more by vibratory proprioceptive calf and neck stimulation than it was in younger subjects. However, we found no significant interactions between the factors (vibration site, vision and age) which suggest that these factors independently affect postural control. Additionally, both elderly and young subjects became progressively better at handling the vibratory balance perturbations over time, suggesting an ability to adapt to these perturbations.

Lateral Torque Variance

Elderly and Young Comparison

Table 3 shows that lateral torque variance was significantly larger for the elderly group compared with the

Table 3. Lateral torque variance differences between elderly and young

	Eyes closed (EC)			Eyes open (EO)		
	elderly	young	p value	elderly	young	p value
A. Calf stimulation						
Quiet stance	0.80 (0.32)	0.20 (0.03)	p = 0.008	0.53 (0.14)	0.25 (0.08)	p = 0.049
Period 1	1.47 (0.22)	0.69 (0.07)	p = 0.001	0.95 (0.17)	0.48 (0.07)	p = 0.004
Period 2	0.86 (0.16)	0.54 (0.06)	NS	0.70 (0.18)	0.30 (0.05)	p = 0.013
Period 3	0.84 (0.15)	0.53 (0.06)	NS	0.87 (0.27)	0.37 (0.09)	p = 0.013
Period 4	0.66 (0.10)	0.52 (0.07)	NS	0.55 (0.12)	0.34 (0.06)	NS
B. Neck stimulation						
Quiet stance	0.45 (0.11)	0.23 (0.04)	NS	0.38 (0.12)	0.15 (0.02)	p = 0.019
Period 1	0.83 (0.17)	0.36 (0.07)	p = 0.005	0.50 (0.11)	0.29 (0.06)	NS
Period 2	0.44 (0.09)	0.38 (0.07)	NS	0.47 (0.13)	0.23 (0.05)	NS
Period 3	0.39 (0.11)	0.30 (0.05)	NS	0.49 (0.22)	0.21 (0.03)	NS
Period 4	0.56 (0.13)	0.29 (0.05)	NS	0.28 (0.07)	0.20 (0.04)	NS

Statistical evaluation of the torque variance differences in lateral direction between elderly and young subjects while submitted to calf stimulation (A) or neck stimulation (B) with eyes closed (EC) or eyes open (EO). In the tables, the torque variance means and SEM for the elderly group and young group are presented as well as significant differences between the elderly and young groups (NS = non-significant difference). As in the anteroposterior plane, torque variance for the elderly group was larger than for the young group in all significant cases.

young group during quiet stance except for EC-Neck, and in period 1 in all tests except for EO-Neck. In periods 2 and 3, lateral torque variance for the elderly was only significantly larger than the young for EO-Calf.

GLM ANOVA of the Site of Vibration, Vision and Age

GLM analysis showed that vibration site, vision and age all significantly affected torque variance during all stimulation periods in the lateral direction also (table 4). Calf vibration evoked larger lateral torque variance than neck vibration in all vibration periods ($p < 0.001$). Vision decreased lateral torque variance in all stimulation periods (periods 1 and 4, $p \leq 0.001$; period 2, $p < 0.01$; period 3, $p < 0.05$). Age increased lateral torque variance in all periods (QS, $p < 0.001$; periods 1 and 2, $p \leq 0.001$; periods 3 and 4, $p < 0.01$). However, there was no interaction between the vibration site, vision and age during any period.

Adaptation of Lateral Torque Variance

In the lateral direction, torque variance only significantly increased at the onset of vibration for the elderly group with EC-Calf ($p < 0.01$) and for the young group with EC-Calf ($p < 0.001$) and EO-Calf ($p < 0.01$). The percentage increases in torque variance for the elderly group were: 84% (EC-Calf), 79% (EO-Calf), 84% (EC-Neck), and 32% (EO-Neck). The percentage increases in torque variance for the young group were larger compared with the elderly group in all tests except for EC-Neck, though from noteworthily much lower quiet stance values than in the elderly (fig. 3). The increases were: 245% (EC-Calf), 92% (EO-Calf), 57% (EC-Neck), and 93% (EO-Neck).

Torque variance decreased significantly between period 1 and period 4 for the elderly group with EC-Calf ($p < 0.001$), EO-Calf ($p = 0.002$) and EO-Neck ($p < 0.001$). However, there was no evidence of adaptation with EC-Neck. In contrast, torque variance did not significantly decrease in any test for the young group. The decreases in torque variance for the elderly group were: 55% (EC-Calf), 42% (EO-Calf), 33% (EC-Neck), and 44% (EO-Neck). The percentage decreases for the young group were markedly lower: 25% (EC-Calf), 29% (EO-Calf), 19% (EC-Neck), and 31% (EO-Neck).

To summarize, stability in the lateral direction in elderly subjects was clearly challenged more by vibratory proprioceptive calf and neck stimulation than in younger subjects, as shown by the quantitative levels of torque variance (fig. 3). Again, we found no significant interactions between the factors (vibration site, vision and age), which

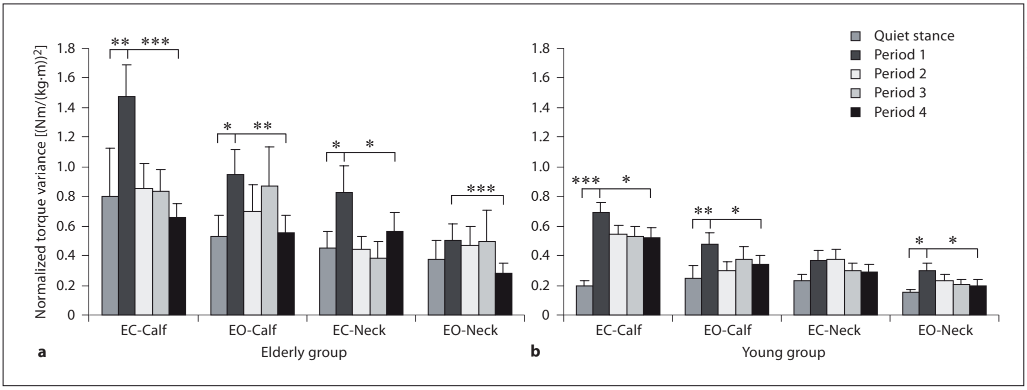

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Fig. 3. Lateral torque variance for the elderly group (**a**) and young group (**b**) for calf vibration with eyes closed (EC-Calf), calf vibration with eyes open (EO-Calf), neck vibration with eyes closed (EC-Neck) and neck vibration with eyes open (EO-Neck). * $p < 0.05$; ** $p < 0.01$; *** $p < 0.001$.

Table 4. General linear model analysis of torque variance in the lateral direction

	Vibration site	Vision	Age	Vibration site × vision	Vibration site × Age	Vision × age	Vibration site × vision × age
Quiet stance	NS (0.8)	NS (2.5)	p < 0.001 (22.9)	NS (0.1)	NS (0.3)	NS (0.1)	NS (0.2)
Period 1	p < 0.001 (37.5)	p < 0.001 (13.1)	p < 0.001 (34.4)	NS (0.1)	NS (0.0)	NS (0.5)	NS (0.5)
Period 2	p < 0.001 (17.8)	p = 0.003 (9.4)	p = 0.001 (12.1)	NS (0.7)	NS (0.5)	NS (2.5)	NS (0.2)
Period 3	p < 0.001 (30.9)	p = 0.039 (4.4)	p = 0.006 (7.9)	NS (0.3)	NS (0.9)	NS (0.8)	NS (0.0)
Period 4	p < 0.001 (21.7)	p = 0.001 (10.9)	p = 0.004 (8.4)	NS (0.0)	NS (0.0)	NS (0.1)	NS (0.3)

Effects of the site of vibration, vision, age and their interactions on recorded lateral torque variance analyzed using a GLM univariate ANOVA (general linear model univariate analysis of variance) test on log-transformed values. Mean and SEM values for the torque variance data analyzed are presented in table 3. F values are presented in parentheses.

suggests that these factors independently affect postural control in the lateral direction. Additionally, only the elderly showed signs of adaptation to the vibratory balance perturbations in the lateral direction over time, whereas the young subjects only made minor improvements.

Discussion

Postural Control and Ageing

It is generally accepted that ageing can affect anteroposterior and lateral quiet standing postural stability and our findings are generally consistent with this. We observed that quiet stance anteroposterior stability was significantly poorer in the elderly than in young subjects with eyes closed and in the lateral direction in all tests except EC-Neck (tables 1, 3).

One important finding was that the initial perturbations (i.e., in period 1) did not always exacerbate the difference between the elderly and young groups. Due to the sensory/motor deficiencies that are evident in the elderly, one would have expected vibratory stimulation to increase the difference between the elderly group and the young group. It is well known that the integrity of the sensory systems during sensory processing is important for postural control [36], and the present findings suggest that postural control in both young and elderly subjects are influenced by the integrity of sensory information. However, the observed adaptation of postural control suggests that both young and elderly are capable of learning more effective ways to use the remaining reliable afferent information to compensate for the proprioceptive/visual deficits induced by the vibratory proprioceptive stimulation to the calf or neck muscles and by standing with eyes closed. This supports ideas of training benefits in the elderly, as our findings showed that through training the elderly could compensate and learn over time to handle repeated balance perturbations better and thereby enhance their stability in both the anteroposterior and lateral directions.

Effect of Ageing on Postural Adaptation

In the anteroposterior direction, the percentage of adaptation between periods 1 and 4 was similar between the elderly and the young groups in all tests and is in line with our previous findings showing that the elderly have an effective anteroposterior postural control adaptation, which is similar to the middle-aged [20]. Pavol et al. [37] have also showed that older adults rapidly learn to avoid falling from sit-to-stand balance perturbations at a similar rate to that of young healthy individuals. However, there were marked differences in adaptation between the young and elderly groups in the lateral direction. The human standing posture usually needs to be challenged by a sufficiently large postural disturbance for postural adaptation to occur [35, 38], and our findings suggest that the elderly found their stability in lateral direction sufficiently challenged to initiate such adaptation whereas young subjects did not. Thus, the quantitatively higher levels of torque variance in the lateral direction may signify that the elderly were closer to their limits of stability in lateral direction during the trials than young subjects. This is consistent with our hypothesis that the elderly would be affected more by proprioceptive vibratory stimulation of the calf and neck muscles in the lateral direction compared with young subjects, as lateral stability appears to be particularly affected by age. This might also explain why we observed marked differences in postural adaptive performance between the elderly group and the young group in the lateral direction and not in the anteroposterior direction (fig. 2, 3). The elderly may have had to place attention on postural stability in both anteroposterior and lateral directions since the stability was regarded as 'threatened' in both directions. The young subjects might only have perceived a noteworthy 'threat' to the stability in the anteroposterior direction and subsequently initiated an adaptation only in this direction. One implication of our findings is that the attentional demands during postural tasks could increase with age, and this is consistent with previous findings from dual-task experiments [39]. However, ageing seems not to affect some of the postural control mechanisms associated with adaptation such as recognition of balance perturbation effects and the ability to find and effectively use reliable afferent information. Thus, a possible method for postural control adaptation training and rehabilitation could be to administer the elderly with courses of vibratory proprioceptive stimulation, and this concept has been suggested previously using other types of repeated perturbation [40].

Effect of the Site of Vibration

We found only some evidence that the site of vibration significantly affected the anteroposterior and lateral torque variance difference between the elderly and young groups during the vibration periods (shown in tables 1 and 3), which suggests that the elderly are capable of recognizing and handling balance perturbations of various origin. We also found that the EC-Neck test caused a different challenge compared to the other test conditions for the elderly shown by a lack of adaptation and a larger increase of torque variance at the onset of vibration in the lateral direction (for details, see fig. 3). One possibility is that neck cervical input is of far more importance to the elderly compared with the young, and as such neck vibration might have a different effect in the elderly, especially without a visual feedback of movements. Additionally, previous studies have reported that ageing results in structural changes to muscle spindles when comparing the brachii muscle nuclear fibres of a group of 69- to 83-year-olds with a group of 19- to 48-year-olds [41]. However, when combining these results with the ones from the present study, it appears that age-related structural changes of the muscle spindles might be of more importance when the structural changes affect certain muscles, such as the neck muscles. Thus, more studies are required to investigate the effects on postural control of local age-related structural changes in the muscles.

Effect of Ageing on Vision

Several of our observations support the hypothesis that vision can provide information to assist postural control [42, 43] in elderly and young groups (tables 2, 4). Previous studies have similarly shown that during quiet standing, elderly subjects can use visual information to reduce their level of instability in anteroposterior direction [44]. The positive effect of vision was however not as apparent for quiet standing in the elderly and young groups in lateral direction (table 4), which might be explained by generally quite good inherent stability to begin with in the lateral direction due to biomechanical reasons (for details, see tables 1, 3). However, during repeated balance perturbations, vision clearly provided important information for postural stability in the anteroposterior and lateral directions in both the elderly and young groups. This suggests that, in both groups, vision is of vital importance for postural control. It is well known that there is a deterioration of vision with age. For example, previous reports have showed that the number of ganglion cells of the retina decrease from the age of 42 resulting in a parallel decrease in the optokinetic nystagmus gain [45–47]. Thus, if the elderly rely on a failing visual system in the control of postural stability, this could cause some of the well-known balance problems the elderly experience, in some cases even instigating falls. Nonetheless, our findings do not support our hypothesis that the elderly per se have a greater reliance on visual information for postural control (i.e., we found no interaction between vision and age in the GLM analysis).

Statistical analysis showed no combined effect of age, the site of vibration and vision on postural control, which suggests that these factors act independently on postural control. Our previous research has similarly shown that the distortion of sensory information when standing on foam acts independently to the effect of vision [30]. Therefore, the integration of sensory information by the central nervous system could be more complex than is proposed by the sensory re-weighting theory, e.g., if some sensory information is disrupted, the central nervous system can quickly learn to use other more reliable sources of afferent information [48].

One finding in the present study was that the stability of the elderly was more variable across subjects compared with the young subjects, both during quiet stance and during balance perturbations, as expressed by the torque variance values. This observation suggests that the stability of individual elderly subjects might need to be evaluated on an individual basis given the present status of age-related deficits. Large data variability could potentially affect the reliability of the statistics and accordingly we used non-parametric statistics since the obtained data were not normally distributed. Furthermore, the accuracy of the statistical GLM ANOVA was proven by testing the GLM model residuals for normal distribution in all statistical analyses.

Conclusions

Adaptation in the anteroposterior direction occurred to an almost equal extent in the elderly and young, whereas adaptation in the lateral direction was markedly larger in the elderly in all tests except for neck vibration with eyes closed. Thus, the ability to adapt seemed not to decline with age per se, although age has been associated with decline in both sensory and motor systems used for postural control. Age, vision and vibration site were all influential factors for recorded torque variance, but no significant interaction effects were found. Hence, during perturbation, age and vision were both very influential factors for the stability, thus associating the previously documented age-related decline in visual functioning with a higher risk of falls in this age range.

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Article IV

Effects of 24-h and 36-h sleep deprivation on human postural control and adaptation

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RESEARCH ARTICLE

Effects of 24-h and 36-h sleep deprivation on human postural control and adaptation

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Abstract This study investigated whether human postural stability and adaptation were affected by sleep deprivation and the relationship between motor performance and subjective scores of sleepiness (visuo-anlogue sleepiness scores, VAS). Postural stability and subjective sleepiness were examined in 18 healthy subjects (mean age 23.8 years) following 24 and 36 h of continued wakefulness, ensured by portable EEG recordings, and compared to a control test where the assessments were made after a normal night of sleep. The responses were assessed using posturography with eyes open and closed, and vibratory proprioceptive stimulations were used to challenge postural control. Postural control was significantly affected after 24 h of sleep deprivation both in anteroposterior and in lateral directions, but less so after 36 h. Subjective VAS scores showed poor correlation with indicators of postural control performance. The clearest evidence that sleep deprivation decreased postural control was the reduction of adaptation. Also several near falls after 2–3 min during the posturographic tests showed that sleep deprivation might affect stability through momentary lapses of attention. Access to vision, somewhat, but not entirely reduced the effect of sleep deprivation. In conclusion, sleep deprivation can be a contributing factor to decreased postural control and falls.

Keywords Postural control · Sleep deprivation · Adaptation · Subjective scores · Attention

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Introduction

Chronic sleep restrictions are an endemic in modern society, with large fractions of the population reporting daily sleep below the recommended 8 h per night (National_Sleep_Foundation 2005). This problem is associated with long working hours, commuting, and family responsibilities especially in occupations such as health-care, the military and industrial manufacturing where the potential for sleep-related accidents is also high (Balkin et al. 2004). The frequency of this problem was highlighted in a survey by the National Sleep Foundation in 2005, showing that 60% of drivers in the United States admit to driving while feeling drowsy (National_Sleep_Foundation 2005). Horne and Reyner estimated that 20% of serious motorway collisions in the United Kingdom were due to sleepiness on the basis of surveys from police officers and police collision reports (Horne and Reyner 1995). Sleep deprivation may produce effects normally associated with drunkenness such as a lack of coordination, judgment and reaction time (Williamson and Feyer 2000; Wilson et al. 2006). The similarities between tiredness and drunkenness have been evidenced in driving simulation studies. Williamson and Feyer (2000) found that people who drove after being awake for 17–19 h performed worse than those with a blood alcohol level of 0.05%, which is the legal limit for driving in most western European countries. Although sleep-related accidents are well documented in tasks requiring appropriate cognition and decision making (e.g. driving), tiredness may also be dangerous in daily life activities and could increase the likelihood of falling accidents.

Despite the prevalence of sleep deprivation, few studies have evaluated the postural control performance effects and sleep-loss-mediated deficits in performance capacity caused by sleepiness (Schlesinger et al. 1998; Gribble and Hertel 2004; Avni et al. 2006). Although levels of sleepiness and fatigue can be subjectively assessed, such evaluations may not reflect the objective physiological status of the tired person, mainly because subjective scores can be biased by motivation, personal factors, experience, training, etc. (Avni et al. 2006). Therefore, there is a need to find practical non-invasive objective methods to measure the effects of sleepiness, especially when reaching critical levels involving higher risks for accidents. One possibility is to assess postural stability (Avni et al. 2006; Karita et al. 2006).

The effects of balance perturbations on postural control can be studied experimentally by inducing inappropriate sensory information. There are numerous ways of achieving this and one such method involves disturbing somatosensory afferents by means of vibrating the calf muscles (Ivanenko et al. 1999; Fransson et al. 2002; Vuillerme et al. 2002; Ledin et al. 2004). This action results in an increased activation of muscle spindle afferents (Eklund 1973) signalling to the central nervous system that the vibrated muscle is being stretched (Matthews 1986). The increased activity induced from muscle spindles results in a proprioceptive illusion of movement. Shortly following muscle vibration, tonic stretch reflexes are elicited that result in increased body movements (Goodwin et al. 1972). Repeated muscle vibrations over a short period of time result in an increased ability to handle the induced perturbations through postural adaptation (Fransson et al. 2000; Tjernstrom et al. 2005).

Motor control (Haslam 1984; Frey et al. 2004) and other functions such as cognitive ability, attention and alertness, first scientifically investigated by Patrick and Gilbert (1896), have been found to be affected by sleep deprivation. In support of these findings, analysis of brain activity has shown that the areas of the cerebral cortex that regulate aspects of attention, alertness and cognitive ability, such as thalamus and regions within the pre-frontal cortex, undergo deactivation following 24 h of sleep deprivation (Thomas et al. 2000) and decrease activity further as the period of sleep deprivation increases (Thomas et al. 2003). Previously, human postural stability has been shown to decrease following 24 h of continued wakefulness (Liu et al. 2001; Gribble and Hertel 2004; Avni et al. 2006; Fabbri et al. 2006) or with the addition of an attention-demanding task (Schlesinger et al. 1998). To our knowledge, no previous research has been carried out in which postural stability and postural adaptation have been assessed following both 24 (24SDep) and 36 h of sleep deprivation (36SDep) using vibratory proprioceptive stimulation, and in which these posturographic assessments have been compared against subjective visuo-analogue sleepiness scale (VAS) scores which has been an accepted way of monitoring subject sleepiness.

The first aim of this study was to investigate whether postural stability and postural adaptation, recorded as anteroposterior and lateral torque variance, differed between the control tests, 24SDep and 36Sdep, while subjected to proprioceptive vibration with eyes open (EO) and eyes closed (EC). The second aim was to determine whether postural control performance correlated with VAS scores. The final aim was to investigate whether postural stability and adaptation in anteroposterior and lateral torque variance differed with eyes open and closed between the control tests 24SDep and 36SDep.

Methods and materials

Subjects

Eighteen (10 males and 8 females) healthy volunteers, aged between 16 and 38 years (mean 23.8 years, SD 4.8 years; mean weight 78.5 kg, SD 18.8 kg and mean height 1.77 m, SD 0.09 m) with no previous history of central nervous diseases, performed postural stability tests involving sleep deprivation. Subjects were instructed not to consume any alcohol, sleepiness-inducing or revitalizing products, such as caffeine, 48 h before and during testing. At the time of experimentation, no subject was taking any form of medication and signed consent was obtained before any testing began. The experiments were all performed in accordance to the Helsinki declaration of 1975 and approved by the local ethical committee.

Equipment

A custom built force platform recorded the forces with six degrees of freedom and with an accuracy of 0.5 N. A customized computer program controlled the vibratory stimulation and sampled the force platform data at 50 Hz.

The vibrators, had a vibration amplitude of 1.0 mm and a frequency of 85 Hz, were 6 cm long and 1 cm in diameter. The vibrators were placed over the gastrocnemius muscles and secured by elastic straps.

Procedure

Sleep deprivation testing was performed on 2 consecutive days, days 1 and 2. On day 1, subjects were asked to wake up at 7 am or 8 am (depending on the recording schedule) to begin their sleep deprivation session. The subjects were instructed to stay awake, without using stimulants, and go about their daily routines as normal. Subjects came to the laboratory at 7 pm or 8 pm on day 1 (12 h into their sleep deprived state) to be attached with a portable EEG recoding device (Embletta™). The EEG recordings were made to ensure that the subjects had not fallen asleep during the sleep deprivation period. Signs of sleep were monitored with a unilateral electrode measurement set-up using routine electrodes. The EEG device comprised three electrodes; an active electrode positioned on the upper temple; a reference electrode positioned on the upper mastoid bone on the opposite side to the active electrode; and a ground electrode positioned on the mid-forehead. Subjects returned on day 2 at 7 am or 8 am, 24 h into sleep deprivation, then again that evening at 7 or 8 pm, 36 h into sleep deprivation for posturographic assessment. The EEG equipment was removed prior to testing to avoid any possible interference from tactile information from EEG electrodes, the recording device and EEG cables. Additionally, it was appraised that correct EEG recordings could not be assured during the tests due to the electrical noise produced by the posturographic equipment. The EEG data were stored for off-line analysis before re-attachment. Scoring of wakefulness/sleep was done according to Rechtschaffen and Kales (1968). Uninterrupted sleep stage II for more than 2 min was considered to be sleep.

Subjects were also instructed to provide a subjective score using visuo-analogue sleepiness scale (VAS) of alertness ranging from "completely alert" to "exhausted, to near sleep". Subjects analogue scores were converted into numbers ranging from 1 to 10, where 1 = "completely alert" and 10 = "exhausted to near sleep".

A control posturographic test (i.e. following a normal night of sleep) was performed either 1 week before or 1 week after the sleep deprivation tests. The test order was decided by randomization.

Posturographic assessment

Each subject stood barefoot on the force platform in a relaxed posture with arms folded across the chest (see Fig. 1). The subject's heels were 3 cm apart and feet positioned at an angle of 30° along guidelines on the platform. Subjects were instructed to focus on a target 1.5 m in front of them at eye level or keep their eyes closed depending on the test condition. The subjects listened to music through headphones in order to reduce possible movement references from external noise sources and to avoid extraneous sound distractions.

Tests were performed during three different test conditions: (i) after a normal night of sleep, (ii) after 24 h of sleep deprivation (24SDep) and (iii) after 36 h of sleep deprivation (36SDep). During each of these test conditions, the following two posturographic tests were performed by all subjects in a randomized order using a Latin Square design.

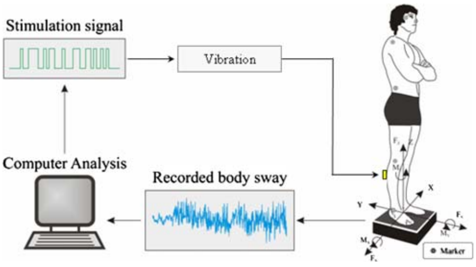

Fig. 1 Schematic picture of the measurement setup

- Vibration of the calf muscles with eyes closed (EC).
- Vibration of the calf muscles with eyes open (EO).

Before the vibration commenced, a 30-s period of quiet stance was recorded. The vibratory stimulation pulses were applied according to a pseudorandom binary sequence (PRBS, Johansson 1993) during a period of 205 s making each test 235 s long, where each pulse had random durations from 0.8 to 6.4 s, yielding an effect bandwidth of the vibratory stimulation within 0.1–2.5 Hz. A 5-min rest period was given to the subjects between each test.

Analysis

Stability while standing is commonly analyzed using force platforms and the movements of the center of pressure (CoP), i.e., the point of application of the ground reaction force. We analyzed the variance of the torque values, where torque = CoP $\times$ m $\times$ g; m = the assessed subjects weight and g = gravitational constant 9.81. Hence, changes in recorded torque are equivalent to changes in CoP. One benefit with presenting torque variance values is that the calculated value directly corresponds to the energy used toward the support surface to maintain stability (Magnusson et al. 1990).

Anteroposterior and lateral movements were analyzed in terms of variance of torque from the force platform recordings. Each test was divided into five periods: quiet stance (0–30 s), and four 50 s stimulation periods (period 1: 30–80 s; period 2: 80–130 s; period 3: 130–180 s; period 4: 180–230 s).

Statistical analysis

Torque variance values were normalized to account for anthropometric differences between the subjects, using the subject's squared height and squared weight (Johansson et al. 1988). The squared nature of the variance algorithm made it necessary to use normalization with squared parameters to achieve unit agreement with the standardization.

The Wilcoxon matched-pairs signed-rank test (exact signed two-tailed, Altman 1991) was used for the statistical comparison between test conditions and for the analysis of variations over time. The torque variance changes between quiet stance and period 1 were evaluated to determine how the assessed parameters were initially affected by the balance perturbations evoked by vibratory proprioceptive stimulation compared to the activity during quiet stance. The torque variance changes between periods 1 and 4 were evaluated to determine how the assessed parameters were affected by repeated vibratory stimulation, quantifying possible effects of adaptation to vibratory proprioceptive stimulation. The effects of sleep deprivation, vision and the combined effect of vision and sleep deprivation were analyzed using a GLM univariate ANOVA (general linear model univariate analysis of variance) test on log-transformed values (Altman 1991). The accuracy of the GLM model was evaluated by testing the model residual for normal distribution. Correlation analysis was performed between subjective VAS scores and torque variance using Spearman's correlation test (Altman 1991).

Normality of distribution was tested with the Shapiro–Wilk test. Non-parametric statistics were used since some values were not normally distributed. The statistical analysis was carried out with Bonferroni correction for multiple comparisons where no more than two matched pair comparisons were performed for each test. In the analysis, P values < 0.01 were considered to be statistically significant, except in the GLM analysis, where P values < 0.05 were considered to be significant (Altman 1991). However, we present P values < 0.05 in the figures for consistency. The statistical analysis was performed using SPSS version 14.0 (SPSS) and Matlab version 7.2 (The MathWorks).

Results

The subjective VAS score increased from a level of 5.2 at 24SDep to a level of 6.8 at 36SDep ($P < 0.001$), where the VAS ranges were defined as 1 = "completely alert" and 10 = "exhausted to near sleep".

Anteroposterior torque variance

Adaptation of anteroposterior torque variance

Anteroposterior torque variances showed similar trends during quiet stance and during the vibration periods, both with eyes open (EO) and eyes closed (EC), although the magnitude of the variance values was markedly larger with EC (Fig. 2). In all tests, during the first period of vibratory stimulation (period 1) the torque variances were significantly larger compared to the quiet stance torque variance ($P < 0.001$). The increase in torque variance in percentage with the 10 and 90% confidence intervals are presented within the square brackets, between quiet stance and period 1 were for the different tests: 1,175% [320, 2,030%] (control-EC); 965% [370, 1,560%] (24SDep-EC); 611% [370, 850%] (36SDep-EC); 469% [290, 640%] (Control-EO); 713% [330, 1,100%] (24SDep-EO); and 341% [240, 440%] (36SDep-EO).

In the control tests, after the initial rise in torque variance at period 1, torque variance decreased and stayed low between periods 2 and 4. The extent of adaptation to the vibratory stimulation are presented as percentage decrease in torque variance between periods 1 and 4 with the 10 and 90% confidence intervals presented within the square

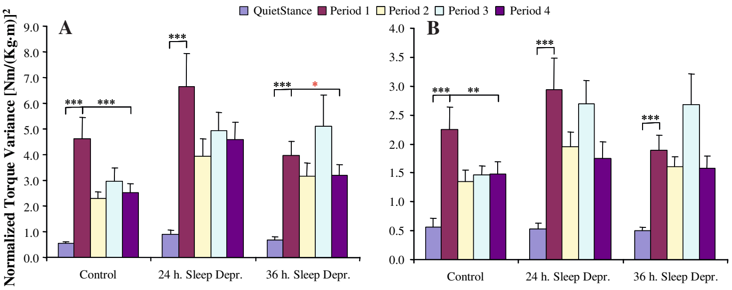

Fig. 2 Normalized torque variance values for anteroposterior torque [mean and standard error of mean (SEM)] during tests with (**a**) eyes closed, and (**b**) eyes open. The variance values have been normalized with the subject's weight and height. Note the differences in scale of the y-axis of the two graphs. The statistical differences found between quiet stance and period 1, and between periods 1 and 4 are marked with *asterisk*, where $*P < 0.05$, $**P < 0.01$ and $***P < 0.001$. For period 1, there was a clear increase in variance values in all tests compared with the quiet stance period. Adaptation was only evidenced in the control tests

brackets. In the control tests, the torque variances were reduced by about 29% [14, 44%] ($P < 0.001$) with EC and 21% [0, 42%] ($P < 0.01$) with EO between periods 1 and 4. However, during 24SDep and 36SDep there was no significant difference between the Periods 1 and 4 torque variance values with EO or EC, and up to about 30% of subjects increased their torque variance between periods 1 and 4. Moreover, the torque variance values were markedly high in period 3 during several sleep deprivation tests.

GLM analysis of anteroposterior torque variance

The anteroposterior torque variances were not significantly different between the control tests, 24SDep and 36SDep during quiet stance with EO and EC, although there were trends of larger torque variances during 24SDep with EC (Table 1). In post hoc tests, none of these changes reached Bonferroni compensated significant levels of $P < 0.01$. Sleep deprivation significantly increased anteroposterior torque variance in periods 1, 2 and 3 ($P < 0.05$). In these periods, torque variance was on average about 57% greater at 24SDep with EO and EC compared to the control test and on average about 30% greater at 36SDep with EO and EC.

Table 1 Statistical evaluation of the anteroposterior torque variance values using the GLM univariate ANOVA method

Torque variance	*P* value		
	Sleep deprivation	Visual influence	Sleep deprivation × visual influence
Quiet stance	ns	0.012	ns
Period 1	0.039	<0.001	ns
Period 2	0.015	<0.001	ns
Period 3	0.004	<0.001	ns
Period 4	ns	<0.001	ns

"ns" signifies no significant difference. The notation "<0.001" means that *P* value is smaller than 0.001. Sleep deprivation had a clear effect on the torque variance values, especially during periods 2 and 3. However, vision had a marked influence during all periods, whereas the combined effect was non-significant

Vision clearly reduced torque variance during quiet stance ($P < 0.05$) and in periods 1–4 ($P < 0.001$). Torque variance values were on average about 49% smaller with eyes open than with eyes closed. However, there was no evidence of a combined effect between sleep deprivation and vision, which shows that the effects of sleep deprivation were not influenced by whether the subjects had their eyes open or closed during the tests.

Lateral torque variance

Adaptation of lateral torque variance

Similar to anteroposterior results, lateral torque variances were also affected by sleep deprivation (Fig. 3). During the first period of vibratory stimulation (period 1), the torque variance values were significantly larger compared to the quiet stance torque variance by about 170% [80, 260%] in the control test with EC ($P < 0.01$), and by about 154% [50, 250%] with EO during 24SDep ($P < 0.001$), (the 10 and 90% confidence intervals for the changes are presented within the square brackets).

Moreover, the lateral adaptation was different with and without sleep deprivation. In the control test with EO, the torque variance was reduced by about 20% [3, 36%] ($P < 0.01$) between periods 1 and 4 but no statistical decrease was found during 24SDep or 36SDep, (the 10 and 90% confidence intervals for the changes are presented within the square brackets)

GLM analysis of lateral torque variance

The lateral torque variances were not significantly different between the control tests, 24SDep and 36Sdep, during quiet stance with EO and EC, although there were clear trends of larger torque variances during 24SDep with EC (Table 1). In post hoc tests, none of these changes reached Bonferroni compensated significant levels of $P < 0.01$. Sleep deprivation significantly increased lateral torque variance values in periods 2, 3 and 4 ($P < 0.05$, Table 2). Torque variance

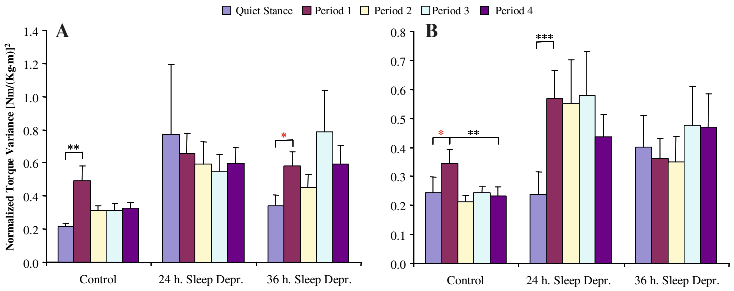

Fig. 3 Normalized torque variance values for lateral torque [mean and standard error of mean (SEM)] during tests with (**a**) eyes closed, and (**b**) eyes open. For period 1, there was a clear increase in variance values in the control test with eyes closed and after 24SDep with eyes open compared with the quiet stance period. Adaptation was only evidenced in the control test with eyes open

increased on average by about 100% at 24SDep compared to the control, both with EO and EC and on average by about 90% at 36SDep.

Vision also clearly reduced torque variance in periods 1, 2 and 4 ($P < 0.05$). The torque variance values were on average 24% less with EO than with EC. However, there was no significant combined effect of sleep deprivation and vision.

Correlation between VAS and torque variance parameters

We found no significant correlation between the VAS scores and the anteroposterior and lateral torque variance values during 24SDep or 36SDep. The highest correlation between subjective sleepiness and recorded body movements was found in lateral direction during quiet stance at 24SDep while standing with eyes closed ($P = 0.052$, $R = 0.478$).

Discussion

Nearly everyone has had first hand experience of sleepiness at numerous situations in their life, but it is probably less well known that sleepiness may affect different aspects of motor performance. This study has allowed us to compare the subjective feelings of sleepiness under objective experimental conditions, particularly related to motor responses and reflexes that are used in one of our basic postures, i.e, standing.

Postural control and sleep deprivation

Our study showed an unexpected result, in that the initial deterioration of postural stability after 24 h of sleep deprivation was not followed by a further deterioration in performance after 36 h of sleep deprivation. Instead, the torque variance was in most cases similar and in several cases even clearly smaller at 36SDep (Figs. 2, 3). A possible explanation for these improvements in stability might be that postural control is affected by circadian rhythm, which is in agreement with other studies that has monitored motor performance during sleep deprivation (Nakano et al. 2001; Gribble and Hertel 2004; Avni et al. 2006). Hence, the deterioration of motor performance during sleep deprivation may not be affected in a linear manner by the length of continued wakefulness. Another possible explanation is that the subjects may have benefited from prior experience when performing the tests at 36SDep from previously had conducted the same tests at 24SDep (Tjernstrom et al. 2002). The precaution we have made in our study protocol was to perform the control tests prior to the sleep deprivation tests in half of the subjects. Hence, half of our subjects had prior experience of testing at 24 h of sleep deprivation, yet there was a marked increase in torque variance between the control and 24SDep tests. Moreover, the novel experience of performing tests for the first time could possibly also suppress the effects of sleep deprivation because of increased attention. Additionally, it should also be noted that the quiet stance performance was changed in a circadian rhythm fashion, although with lower magnitude than that was observed during balance perturbation.

Although our observations suggest improvements in stability at 36SDep from 24SDep, there were several notable signs of that postural control were compromised at 36 h of sleep deprivation, such as markedly high torque variance values in periods 3 and 4 in about 30% of subjects.

A major difference between the control tests and both sleep deprivation states (24SDep and 36SDep) was the way in which subjects responded to perturbations in periods 2–4. In the control tests, torque variance remained at a consistent, low level between periods 2 and 4, whereas when sleep deprived, torque variance was markedly larger, especially in period 3. Moreover, in period 3, three of our 18 subjects briefly, during a couple of seconds, almost toppled over as their knees buckled and arms unfolded, but all managed to regain sufficient control to remain on the force platform. Notably, these near-fall events occurred both with eyes open and closed. It is possible that these near-falls were caused by lapses of attention or 'micro-sleeps' which would be in line with the findings presented in other studies (Broadbent 1958; Williams et al. 1974). Previous reports have shown that lack of attention can affect the ability to maintain postural stability (Maki and McIlroy 1996; Schlesinger et al. 1998; Teasdale and Simoneau 2001; Gribble and Hertel 2004) and could be a contributing factor to the increases in torque variance in the latter periods seen after sleep deprivation in the present study. Prior studies have also shown that sleep deprivation has a more deleterious

Table 2 Statistical evaluation of the lateral torque variance values using the GLM univariate ANOVA method

Torque variance	P value		
	Sleep deprivation	Visual influence	Sleep deprivation × visual influence
Quiet stance	ns	ns	ns
Period 1	ns	0.010	ns
Period 2[a]	0.002	0.016	ns
Period 3	0.022	ns	ns
Period 4	0.007	0.014	ns

[a] The GLM model residual for these data was not normally distributed. These statistical values may therefore be somewhat less accurate. Sleep deprivation had a clear effect on the lateral torque variance values, primarily during periods 2, 3 and 4. Vision influenced the torque variance values during periods 1, 2 and 4, whereas the combined effect was non-significant

effect on performance in long, monotonous, boring tasks than on short, interesting ones (Wilkinson 1965; Dinges and Kribbs 1991; Horne 2000). Therefore, our long-duration tests may have provided an appropriate, objective approach for determining the effects of sleep deprivation. In addition, vibration per se may temporarily increase alertness in both humans and animals (Magnusson 1986). This may explain why the effect of sleep deprivation on vibratory perturbations was less pronounced in period 1.

Although it was previously known that sleep deprivation has an effect on postural control, the precise nature of the change is disputed. Some authors report that postural stability is affected in unperturbed standing (Nakano et al. 2001; Gribble and Hertel 2004; Avni et al. 2006; Fabbri et al. 2006; Karita et al. 2006). Likewise, we found a small increase in quiet stance torque variance in the anteroposterior and lateral directions following sleep deprivation. However, we only found a marked decrease in postural stability due to sleep deprivation when we perturbed stance using calf vibration. In contrast, Uimonen et al. (1994) also using posturography with vibratory proprioceptive stimulation to the calf, found no effects of sleep deprivation on postural control. However, Uimonen et al. assessed the postural stability during much shorter duration (55 s) and only exposed the subjects to five vibratory perturbations compared with 235 s of recordings and 64 vibratory perturbations in each of our tests.

It should be noted that total sleep deprivation, as investigated in our study, and chronic sleep restrictions may not have identical effects on motor control and postural stability (Haslam 1984). However, decreased postural stability has also been reported among subjects with chronic-restricted sleep hours (Karita et al. 2006). Therefore, it is reasonable to assume that our findings may also apply for patients with severe sleep disorders, such as obstructive sleep apnea syndrome (OSAS), which is characterized by a stoppage or decrease in breathing, resulting in inadequate sleep.

Adaptation and sleep deprivation

The ability to adapt and habituate on the basis of prior experiences is important for human movement control, fall prevention (Eccles 1986; Fransson et al. 2003) and for the ability to enhance the performance during various human activities (Horak and Nashner 1986; Keshner et al. 1987). In our study, the changes in adaptation presented the clearest evidence that sleep deprivation significantly affects postural control. We found that anteroposterior and lateral adaptation clearly differed under sleep deprivation compared to the control tests. The investigated subjects were clearly able to adapt to the balance perturbations evoked by the vibratory stimulation in the control tests, but could not do so significantly when sleep deprived. Some subjects even increased their torque variance between periods 1 and 4. These findings showed that sleep deprivation may affect some subject's ability to learn and respond appropriately to balance perturbations. Integration of information from the visual, vestibular and somatosensory receptors and motor coordination are processes known to require attention (Schlesinger et al. 1998; Fabbri et al. 2006), especially when information from any of the sensory systems is not reliable (Redfern et al. 2001). One possible reason for the lack of adaptation during sleep deprivation might be that the accompanying decrease in attention led to slower, or inappropriate sensory integration, which also affected the adaptation processes, i.e., the ability to choose an appropriate motor response to enhance balance stability.

Subjective sleepiness and postural control

In this study, we found no correlation between subjective sleepiness and motor performance. Subjective sleepiness, as assessed by VAS scores, increased markedly from 24SDep to 36SDep, although in some instances torque variance decreased. Thus, assessments of subjective sleepiness may not be a reliable indicator of actual postural control (Fabbri et al. 2006). The findings presented therefore highlight the value of implementing appropriate regulations for work durations, particularly in attention-demanding occupations, such as long distance driving, because the subjective feeling of sleepiness may not reflect the actual performance decline. One possible reason for why we were unable to find a significant correlation between torque variance and subjective VAS scores might be that the mere act of performing a test might momentarily enhance attention and motivation (Avni et al. 2006). Another explanation could be that in motor tests that require a high level of attention such as aerobic tasks, the association between progressive sleep deprivation, subjective sleepiness and decreased performance might be stronger than that was detected in our study (Souissi et al. 2003).

Vision and sleep deprivation

Several of our findings support evidence from Edwards showing that vision can provide information to assist postural control (Edwards 1946) as sleep deprivation had less effect on postural stability with eyes open than with eyes closed both in anteroposterior and lateral direction. Moreover, the statistical analyses showed no combined effect of sleep deprivation and vision on postural control, which suggests that these two factors possibly act independently. Similar results were found by Karita et al. (2006), in an investigation of postural control in subjects with chronic sleep deprivation. Still, the near falls and decreased adaptation occurred both in tests with eyes open and eyes closed.

Under normal conditions, vision provides an orientational frame of reference, where any imbalance can be quickly captured and an appropriate response can be initiated to maintain precise postural control. However, the findings presented in this study suggest that although visual information provides enhanced stability, this additional information may not at times be sufficient to compensate for the deterioration of performance caused by sleep deprivation.

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Article V

Effects of proprioceptive vibratory stimulation on body movement at 24-h and 36-h of sleep deprivation

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Effects of proprioceptive vibratory stimulation on body movement at 24 and 36 h of sleep deprivation ☆

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Abstract

Objective: To investigate whether postural stability and adaptation differed after a normal night of sleep, after 24 h (24 SDep) and 36 h (36 SDep) of sleep deprivation while subjected to repeated balance perturbations. Also, to determine whether there was any correlation between subjective alertness scores and objective posturographic measurements. Lastly, to investigate the effects of vision on the stability during sleep deprivation.
Methods: Body movements at five locations were recorded in 18 subjects (mean age 23.8 years) using a 3D movement measurement system while subjected with eyes open and closed to vibratory proprioceptive calf stimulation after a normal night of sleep, 24 and 36 SDep.
Results: The clearest sleep deprivation effect was reduced ability to adapt head, shoulder and hip movements, both with eyes open and eyes closed. Additionally, several near falls occurred after being subjected to balance perturbations for 2–3 min while sleep deprived. Unexpectedly, postural performance did not continue to deteriorate between 24 and 36 h of sleep deprivation, but showed some signs of improvement. Subjective scores of sleepiness correlated poorly with actual changes in postural control performance.
Conclusions: Sleep deprivation might affect postural stability through reduced adaptation ability and lapses in attention. Subjective alertness might not be an accurate indicator of the physiological effects of sleep deprivation.
Significance: Sleep deprivation could increase the risk of accidents in attention demanding tasks. There is a need for objective evaluation methods to determine actual performance capacity during sleep deprivation.

Keywords: Postural control; Proprioceptive vibratory stimulation; Attention; Sleep deprivation

1. Introduction

In the modern society, large fractions of the population report daily sleep below the recommended 8 h per night (National Sleep Foundation, 2005). This can induce tiredness and affect daily activities such as driving (Cummings et al., 2001). Sleep deprivation can drastically increase the risk of accidents and has been revealed as one of the main causes of some high-profile catastrophic disasters (Mitler et al., 1988). The features of sleep deprivation include fatigue, a decrease in sustained attention and reduced alertness. Sleep loss may therefore result in a higher risk for accidents and errors particularly where high levels of attention are required (Zils et al., 2005). Recent findings have also indicated that postural stability (Liu et al., 2001; Nakano et al., 2001; Avni et al., 2006) and motor control (Frey et al., 2004) are affected by sleep deprivation, though the mechanisms have yet to be determined. Some consider that the motor deficits are caused by alterations in the attentional state of the brain (Schlesinger et al., 1998; Fabbri et al., 2006). Other authors have proposed that detrimental postural effects are the result of daily circadian changes involving alertness (Nakano et al., 2001; Gribble and Hertel, 2004). Although levels of sleepiness can be measured subjectively, such assessment may not reflect the objective

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performance of the tired person, due to motivation, personal factors, experience, training, etc., (Avni et al., 2006).

In order to explore the effects of sleep deprivation on postural control, we have used the archetypal quiet standing posture as an experimental model. The standing position is often described in terms of an inverted pendulum where the feet act as the point of 'anchorage' and the 'free, movable' end is the head with both extremes joined by a 'single rod' provided by the rest of the body. However, the pendulum model of standing might, in some cases, be oversimplified because the human body is multi-segmented with a number of points where pivoting can occur (i.e., neck, hip, knees and ankles). In order to quantify the standing position of the body, positional markers were placed at the main points of pivot in order to record the more subtle movement changes along the body.

An extension to the methodology was to investigate the standing position during balance perturbations. One method of perturbing the body is vibratory proprioceptive stimulation of postural muscles or tendons. Vibratory stimulation increases the afferent signals from the muscle spindles (Eklund, 1973) and creates a proprioceptive illusion that the vibrated muscle is being stretched (Matthews, 1986). Tonic stretch reflexes are subsequently induced which return the vibrated muscle to its perceived original length (Goodwin et al., 1972) resulting in a change of posture and increased postural sway (Fransson et al., 2000). When repeated, muscle vibration can evoke postural adaptation which enhances postural performance (Horak and Nashner, 1986; Keshner et al., 1987; Fransson et al., 2002) and markedly reduce the likelihood of imbalance and prevent falls (Pai and Iqbal, 1999; Pavol and Pai, 2002).

In one of the few hitherto studies of postural control stability while subjected to vibratory proprioceptive calf stimulation during sleep deprivation, Uimonen et al. found no effects of sleep deprivation on stability (Uimonen et al., 1994). However, Uimonen et al. assessed postural stability for only 55 s and only exposed the subjects to five vibratory perturbations, whereas we intend to record for 235 s and expose subjects to 64 vibratory perturbations in each of our tests. Furthermore, Uimonen et al., did not investigate whether postural adaptation or the actual body movements were changed by sleep deprivation.

The first aim of this study was to investigate whether postural stability and postural adaptation differed between tests after a normal night of sleep (Control), after 24 h and after 36 h of sleep deprivation while subjected to proprioceptive vibratory stimulation with eyes open and eyes closed. The second aim was to determine whether there was any correlation between subjective alertness scores (VAS) and objective posturographic measures. The third aim was to investigate whether postural control and adaptation differed with eyes open and eyes closed during sleep deprivation.

Our main hypothesis was that sleep deprivation would increase body movement both with eyes open and eyes closed. However, because the maintenance of postural stability is regulated by visual, vestibular and somatosensory information, the destabilizing effects of sleep deprivation might be larger with eyes closed. In addition, as the duration of sleep deprivation has been shown to increase cerebral deactivation (Thomas et al., 2003), another possibility was that body movement would be larger at 36 hours of sleep deprivation compared with 24 h.

2. Methods and materials

2.1. Subjects

Tests were performed on eighteen (ten male and eight female) healthy subjects (mean age 23.8 years, range 17–38 years; mean height 1.77 m, range 1.55–1.90 m; mean weight 78 kg, range 54–117 kg) with no history of balance problems, central nervous disease or injury to the musculoskeletal system. The participants were instructed not to consume any alcohol, medication, drowsiness-inducing or revitalizing products, such as caffeine, for a period of 48 h before and during testing, and all signed consent forms. The experiments were performed in accordance with the Helsinki Declaration of 1975 and approved by the local Ethics Committee.

2.2. Equipment

The proprioceptive stimulators had a vibration amplitude of 1.0 mm and frequency of 85 Hz and were 6 cm long and 1 cm in diameter. The vibrators were placed over the gastrocnemius muscles and secured by elastic straps.

An ultrasonic 3D-Motion Analysis system (Zebris™) measured the linear movements of five markers positioned at anatomical landmarks. The first marker ('Head') was attached to the subject's cheekbone (os zygomaticum), the second ('Shoulder') to tuberculum majus, the third ('Hip') to the crista iliaca, the fourth ('Knee') to the lateral epicondyle of femur, and the fifth ('Ankle') to the lateral distal fibula head, see Fig. 1. Each marker was tracked in three directions, i.e., anteroposterior, lateral and vertical. The measurement resolution in all dimensions was 0.4 mm. A customized computer program controlled the vibratory stimulation, and sampled the kinematic data at 50 Hz.

2.3. Procedure

To investigate the effects of sleep deprivation, testing was performed on two consecutive days. On day 1, subjects were asked to wake up at 7am or 8am (depending on the recording schedule) to begin their sleep deprivation session. The subjects were instructed to stay awake, without using stimulants, and go about their daily routines as usual. Subjects came to the laboratory 12 h later when they were fitted with a portable EEG recording device (Embletta™) to monitor their alertness during the experimental period.

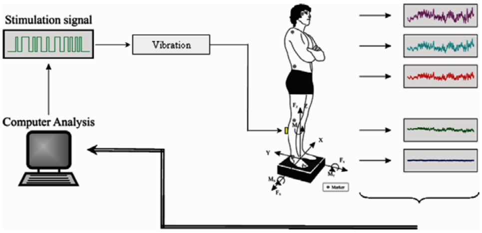

Fig. 1. Schematic picture of the measurement setup and placement of the five Zebris markers attached on a subject. The marker locations are shown by small circles.

Signs of sleep were monitored with an unilateral electrode measurement setup using routine electrodes. The EEG equipment comprised three electrodes; an active electrode positioned on the upper temple; a reference electrode positioned on the upper mastoid bone on the opposite side to the active electrode; and a ground electrode positioned on the mid-forehead. During the night of sleep deprivation, subjects reported that they remained awake by, for example, reading, watching television and taking long walks. The subjects returned on day 2 at 7am, 24 h into sleep deprivation (denoted 24 SDep), then again that evening at 7pm, 36 h into sleep deprivation (denoted 36 SDep) for their final posturographic assessment. The EEG equipment was removed prior to testing in order to avoid any possible interference from tactile information from EEG electrodes, the recording device and EEG cables. Additionally, it was appraised that correct EEG recordings could not be assured during the tests due to the electrical noise produced by the posturography equipment. The EEG data were stored for off-line analysis before re-attachment. Scoring of wakefulness/sleep was carried out according to Rechtschaffen and Kales (1968). Uninterrupted sleep stage II for more than 2 min was considered sleep.

Before the posturographic measurements at 24 SDep and before the measurements at 36 SDep, the subjects provided a subjective score of alertness ranging from "completely alert" to "exhausted and near sleep" using a Visuo-Analogue sleepiness Scale (VAS). Each subject's analogue score was converted into a number ranging from 1 to 10, where 1 = "completely alert" and 10 = "exhausted to near sleep". The subjective VAS score was collected before the posturographic measurements in order to avoid, for example, experiences of poor performance during the posturographic measurements influencing the VAS score given.

A Control posturography test following a normal night of sleep was performed either 1 week prior to sleep deprivation tests or 1 week after, in a randomized order. No VAS scores were obtained prior to the Control posturography tests as this was regarded as the normal state.

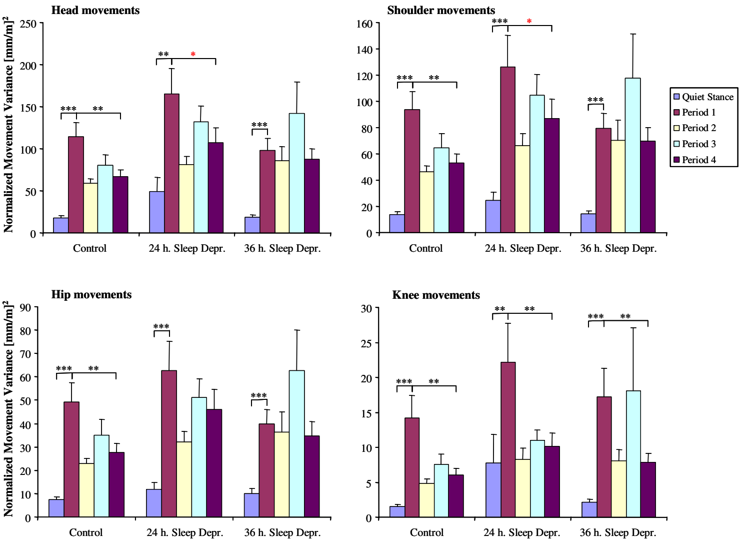

Fig. 2. Variance values for anteroposterior linear head, shoulder, hip and knee movements with eyes closed (mean and standard error of mean (SEM)). The presented values have been normalized using the subject's height. The statistical differences found between Quiet Stance and Period 1 of vibration, and between Periods 1 and 4 are marked with asterisks, where $^{*}p < 0.05$, $^{**}p < 0.01$ and $^{***}p < 0.001$. At Period 1, there was a clear increase in variance values in all tests and at all recording sites compared with Quiet Stance. Indications of adaptation were only apparent in all body segments in the Control test, whereas adaptation was poor in the upper body segments during sleep deprivation tests, particularly after 36 SDep. Note the differences in scale of Movement Variance axes indicating the different extents of sway for each segment of the body.

2.4. Posturography assessment

The five Zebris markers were attached on the right side of the subject facing the Zebris transmitter. Each subject was then instructed to stand barefoot on a force platform, relaxed and with arms folded across the chest. Subjects focused on a target 1.5 m in front of them at eye level or closed their eyes, depending on the test condition.

Tests were performed during three different test conditions: (I) after a normal night of sleep, (II) after 24 h of sleep deprivation (24 SDep) and (III) after 36 h of sleep deprivation (36 SDep). During each of these test conditions the following two posturography tests were performed by all subjects in a randomized order using a Latin Square design.

- Vibration of the calf muscles with eyes closed (EC).
- Vibration of the calf muscles with eyes open (EO).

The subjects were allowed to rest for five minutes between tests. Before vibration, a 30-s period of Quiet Stance was recorded. The vibratory stimulation pulses were activated using a pseudorandom binary sequence (PRBS) schedule (Johansson, 1993) over 205 s making each test 235 s long. Each pulse had random durations from 0.8 to 6.4 s, yielding an effect bandwidth of the vibratory stimulation within 0.1–2.5 Hz.

2.5. Analysis

Vibratory calf muscle stimulation induces body movement mainly in the anteroposterior direction, therefore, only linear movement in this plane is considered here (Fransson et al., 2000). Postural sway was analyzed in terms of the variance of the head, shoulder, hip and knee movements recorded by the Zebris™ system (Fransson et al., 2007). Furthermore, EC/EO quotient values showing the proportional differences in body movements between eyes open and eyes closed tests for each marker position were calculated for all three test conditions.

Each test was divided into five periods: Quiet Stance (QS) (0–30 s), and four 50-s stimulation periods (Period 1: 30–80 s; Period 2: 80–130 s; Period 3: 130–180 s; Period 4: 180–230 s).

2.6. Statistical analysis

Anteroposterior linear movement variance values were normalized using the subject's squared height before the statistical analysis to account for anthropometric differences between the subjects (Johansson et al., 1988). The squared nature of the variance algorithm made it necessary to use normalization with squared parameters to achieve unit agreement with the standardization.

The Wilcoxon matched-pairs signed-rank test (Exact sig. 2-tailed) (Altman, 1991) was used to statistically compare results between the test conditions and for the analysis of quotients. The movement variance changes between Quiet Stance and Period 1 were evaluated to determine how the assessed parameters were initially affected by the balance perturbations evoked by vibratory proprioceptive stimulation compared to the activity during Quiet Stance. The movement variance changes between Periods 1 and 4 were evaluated to determine how the assessed parameters were affected by repeated vibratory stimulation, quantifying possible effects of adaptation to vibratory proprioceptive stimulation. The EC/EO quotient changes were analyzed between all periods in order to monitor periodic changes in body movement pattern. In addition, a GLM univariate ANOVA (General Linear Model univariate Analysis of Variance) statistical test on log-transformed values (Altman, 1991) was used to determine whether vision or sleep deprivation significantly affected results and whether there was an interaction between the two factors on measured linear body movement. The GLM model accuracy was evaluated by testing the model residual for normal distribution. Correlation analysis was performed between subjective VAS scores and movement variance using Spearman correlation test.

Non-parametric statistics were used because the values were not normally distributed. The statistical analysis was carried out with Bonferroni correction for multiple comparisons with no more than four matched-pair analyses performed on each single data set. In the analysis, p-values <0.01 were considered statistically significant, except for the GLM analysis and Spearman's correlation analysis where p-values <0.05 were considered significant (Altman, 1991). However, we present the p-values <0.05 in the figures (in red) for consistency.

3. Results

Average subjective VAS sleepiness scores increased from 5.2 at 24 SDep to 6.8 at 36 SDep ($p < 0.001$), where the VAS range was defined as 1 = "completely alert" and 10 = "exhausted to near sleep".

3.1. Linear head, shoulder, hip and knee movements

3.1.1. Effect of 24 and 36 h of sleep deprivation on Quiet Stance (QS)

During QS with eyes open (EO), there was a progressive increase in variance of movement from the knee towards the head in the Control test, see Fig. 3. This is characteristic of the movement associated with the single-link, pendulum model. This pattern of body movements was largely retained in the 24 and 36 SDep tests, only the amplitudes were larger than in the Control test. With eyes closed (EC), QS during the Control test and sleep deprivation tests showed a similar pattern to that of EO, though there were indications that the hip movements increase was proportionally smaller compared with the other body movements at 24 SDep, see Fig. 2.

3.1.2. Effect of balance perturbations with eyes closed

During the first vibratory stimulation period (Period 1) with eyes closed there was a significant increase in body movement variance at all marker positions compared to Quiet Stance in the Control test ($p < 0.001$), see Fig. 2. Head, shoulder and hip movements increased by 560% and knee movement by about 800%. Between Periods 1 and 4, there was a decrease in head, shoulder, and hip movement variances by about 35% ($p < 0.01$) and 60% at the knee ($p < 0.01$). In Period 1 of 24 SDep there was a rise in movement variance from Quiet Stance by about 240% at the head ($p < 0.01$), 420% at the shoulder ($p < 0.001$) and hip ($p < 0.001$) and 190% at the knee ($p < 0.01$). Between Periods 1 and 4, there was a reduction and leveling off in the body movements, though the reduction only reached a significant level at the knee. The decrease in knee movement variance was about 50% ($p < 0.01$).

In Period 1 of 36 SDep, the body movement variance increased similarly to Control test values, and the movement variance changes were smaller than the movement variance changes found at 24 SDep. Head and shoulder movement variances increased by about 440% ($p < 0.001$), hip by 300% ($p < 0.001$), and knee movements by about 700% ($p < 0.001$). Like at 24 SDep, only the knee movement variance reduced significantly between Periods 1 and 4, by about 55% ($p < 0.01$).

3.1.3. Effect of balance perturbations with eyes open

In the Control test, body movement variance increased during the first vibratory stimulation period at the shoulder, hip and knee by about 320% ($p < 0.001$) compared to Quiet Stance, whereas the head movement increased by about 200% ($p < 0.001$) with eyes open, see Fig. 3. With repeated vibration, there was a significant reduction in head and shoulder movement variance by about 35% ($p < 0.01$) between Periods 1 and 4. Knee and hip movement variance also decreased, but the changes did not reach the appropriate level of significance.

In Period 1 of 24 SDep, body movement variance increased by 355% at the head, 465% at the shoulder, 390% at the hip and 290% at the knee ($p < 0.001$) compared to Quiet Stance. However, during the repeated vibrations movement variance only decreased at the knee by about 55% ($p < 0.01$) between Periods 1 and 4.

Similarly, in Period 1 of 36 SDep, body movement variance increased by 240% at the head ($p < 0.001$), by 165% at the shoulder ($p < 0.001$), by 100% at the hip ($p < 0.001$) and by 50% at the knee ($p < 0.01$) compared to Quiet Stance. However, there was no significant reduction of movement variance at any marker position between vibration Periods 1 and 4.

3.2. GLM analysis of linear body movements

Sleep deprivation significantly affected all body movement variances in Period 3 ($p < 0.05$), see Table 1. The analysis also showed that vision influenced almost all body movements, the variances being significantly lower with EO in all periods, particularly at the head and shoulder. In addition, we found that there was no interaction of vision and sleep deprivation.

3.3. Analysis of EC/EO quotient values

The EC/EO quotient values at all recorded body positions were proportionally the same in all periods in the Control test, see Fig. 4. However, the EC/EO quotient values for 24 SDep showed differing movement amplitudes at the body levels in QS and in Periods 2 and 3. During QS, there were proportionally larger knee and head movement variance differences compared to the hip and shoulder between EC and EO, though some changes were only determined at significant level ($p < 0.05$). In Periods 2 and 3, there were proportionally larger head and shoulder movement variance differences compared to the knee and hip ($p < 0.05$). In Period 3, there were proportionally larger head movement variance differences compared to the knee ($p < 0.01$) between EC and EO tests. Like the Control test, the QS EC/EO quotient values for 36 SDep showed an equal reduction in movement with EO and EC. However, in Period 2, head movement variance differences were proportionally larger than hip and shoulder movement variance differences between EC and EO ($p < 0.05$). Also, during Period 3, shoulder movement variance differences were proportionally larger than hip movement variance differences ($p < 0.01$).

3.4. Correlation between subjective sleepiness and anteroposterior body movement

There was an indication of a negative correlation between subjective sleepiness VAS scores and hip movement only at 24 SDep in Period 3 with eyes closed ($p < 0.05$, $R = -0.547$) and in Period 1 with eyes open ($p < 0.05$, $R = -0.494$).

4. Discussion

Most people have had first hand experience of sleepiness. However, by using recordings of movement from multiple articulation points, this study has provided some new insights into how balance control and the movement strategies are affected by 24 and 36 h of sleep deprivation. In previous sleep deprivation studies on postural control, the investigations have been limited to center of pressure measurements using force platforms. However, the findings presented in the present study suggest that body movement variance and the movement pattern are also affected in sleep deprived subjects. Although some evidence showed that sleep deprivation affected unperturbed standing, the most prominent effects were found when sleep deprived subjects were exposed to proprioceptive vibratory stimulation, illustrated for example, by a decreased ability to adapt to balance perturbation. Furthermore, our findings suggested that postural performance might be partially

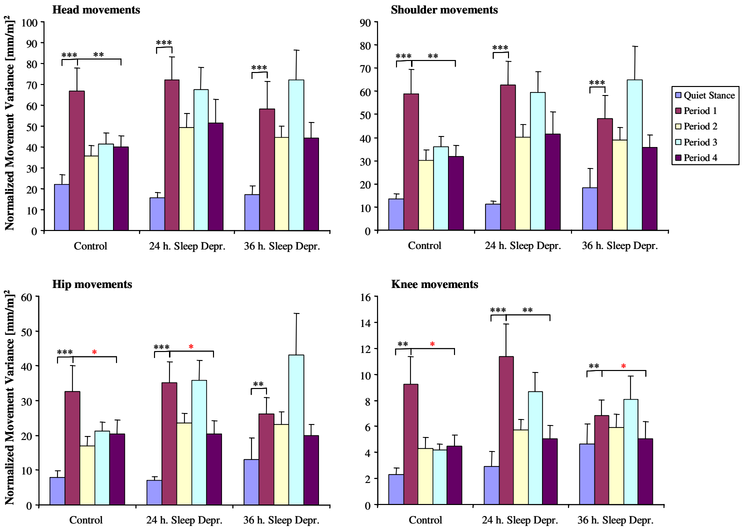

Fig. 3. Variance values for anteroposterior linear head, shoulder, hip and knee movements with eyes open (mean and standard error of mean (SEM)). At Period 1, there was a clear increase in variance values in all tests and at all recording sites compared with Quiet Stance. Indications of adaptation were only apparent in the Control test in the upper body segments, whereas adaptation was poor during sleep deprivation tests.

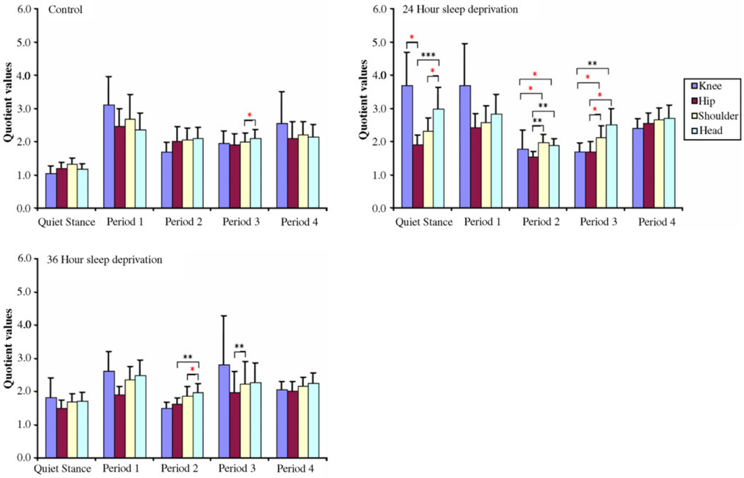

Fig. 4. EC/EO quotients (mean and (SEM)) showing the average body movement values for each test.

affected by circadian rhythm effects and not only by the length of sleep deprivation.

4.1. Postural control and sleep deprivation

Although it is generally accepted that sleep deprivation has a destabilizing effect on posture (Schlesinger et al., 1998; Liu et al., 2001; Nakano et al., 2001; Avni et al., 2006; Fabbri et al., 2006), our study showed that body movement only markedly increased when balance was perturbed by calf vibration. This might be related to the level of attention of sleep deprived subjects. Previous findings have shown that sleep loss results in a higher risk of accidents and errors when high levels of attention are required

Table 1
Statistical evaluation of the linear body movement values using the GLM univariate ANOVA method

	Position	*p*-Value		
		Sleep deprivation	Visual influence	Sleep deprivation × visual influence
Linear body movements				
Quiet Stance	Head	ns	0.041	ns
	Shoulder[a]	ns	0.040	ns
	Hip[a]	ns	ns	ns
	Knee	ns	ns	ns
Period 1	Head	ns	<0.001	ns
	Shoulder	ns	<0.001	ns
	Hip	ns	<0.001	ns
	Knee	ns	0.002	ns
Period 2	Head	ns	<0.001	ns
	Shoulder	ns	<0.001	ns
	Hip	ns	0.007	ns
	Knee	ns	ns	ns
Period 3	Head	0.013	<0.001	ns
	Shoulder	0.015	0.001	ns
	Hip	0.048	0.047	ns
	Knee[a]	0.024	ns	ns
Period 4	Head	ns	<0.001	ns
	Shoulder	ns	<0.001	ns
	Hip	ns	<0.001	ns
	Knee	ns	<0.001	ns

The notation "<0.001" means that the *p*-value is smaller than 0.001, and "ns" signifies no significant difference. Vision had a clear effect on all body movement, especially at the head and shoulder, during all periods. Sleep deprivation only had significant effect on the body movements in Period 3. The combined effect was non-significant. [a]The GLM model residual was not normally distributed. These statistical values may therefore be somewhat less accurate.

(Zils et al., 2005; Fabbri et al., 2006). Attention has long been thought to play a vital role in processing sensory information to maintain postural stability (Woollacott and Shumway-Cook, 2002; Fabbri et al., 2006), especially when sensory information from at least one source is unreliable (Redfern et al., 2001). In this study, calf vibration provided the stimulus to give a false perception of movement. As such, the sensory information from the proprioceptors was 'unreliable' in terms of accurately portraying the actual movement of the body. Under normal conditions, the attentional state of the subject is sufficiently high to re-weight the different sensory inputs and place greater importance on the more reliable receptors (Schlesinger et al., 1998; Woollacott and Shumway-Cook, 2002). This ability to prioritize sensory input may be lost during calf vibration in sleep deprived individuals as evidenced by the larger body movement variances.

Another finding was that the most prominent effects of sleep deprivation were found in Period 3 of stimulation, i.e., during the 100- to 150-s period of vibratory stimulation. During this period, several subjects exhibited a sudden and severe movement so that they almost fell, despite having experienced the effects of vibratory stimulation over the previous 100 s. One explanation could be that these near-falls were caused by lapses of attention as the length of the tests increased, which would be in line with the "Lapse hypothesis" (Wilkinson, 1965). Of note, evaluations of the real-time recordings showed no visible large changes of the capacity to handle the balance perturbations prior to or after the near-fall event, so our conclusion is that the large values in Period 3 are not a sign of poor adaptation but rather another representation of how sleep deprivation may momentarily affect postural control.

These findings might have significant implications for tired workers, especially in the transport industry, as a lapse in attention could lead to an increased risk of an accident. It is possible that these marked events of near-falls, represented by substantially increased body movements during Period 3, may have made the subjects more aware of the instability hazard caused by their sleep deviation and thereby to become more stable for the remainder of the test.

It should be noted that total sleep deprivation, as investigated in this study, and chronic sleep restrictions may not have identical effects on motor control and postural stability (Haslam, 1984). However, decreased postural stability has also been reported among subjects after chronic restricted sleep (Karita et al., 2006). Therefore, it is reasonable to assume that our findings might also be relevant for patients with severe sleep disorders such as obstructive sleep apnea syndrome (OSAS) which is characterized by a stoppage or decrease in breathing, resulting in inadequate sleep.

The present study also showed an unexpected result in that the initial deterioration of postural stability at 24 h of sleep deprivation was not followed by a further deterioration in performance at 36 h of sleep deprivation. Instead, in most cases, the body movement variances were similar, and in several cases even clearly smaller at 36SDep (Figs. 2 and 3). One explanation for this could be that performance follows a circadian rhythm (Nakano et al., 2001; Gribble and Hertel, 2004; Avni et al., 2006). A way of confirming this would be to re-assess subjects after a further 24-h period of sleep deprivation.

4.2. Adaptation and sleep deprivation

Adaptation is an important function of postural control which results in decreased body movement and a reduced risk of falling (Eccles, 1986; Pai and Iqbal, 1999; Pavol and Pai, 2002; Fransson et al., 2003). As expected, after a normal night of sleep, subjects responded to the balance perturbations with an initial increase in body movement variance followed by a gradual reduction in movement variance when repeatedly perturbed by calf vibration. However, sleep deprivation seemed to compromise this mechanism, and in fact the clearest effect of sleep deprivation was found to be a lack of adaptation of the body movement variances at the head, shoulder and hip. One possible reason for this might be that the initiation and maintenance of an adaptive response may require a certain amount of attention. The integration of information from the visual, vestibular and somatosensory receptors and motor coordination are processes known to require attention (Schlesinger et al., 1998; Fabbri et al., 2006), especially when information from any of the sensory systems is not reliable (Redfern et al., 2001). Hence, sleep deprivation and the accompanying decrease in attention may lead to slower or inappropriate sensory integration, which also affected the ability to choose the most appropriate motor response to enhance balance stability.

4.3. Subjective VAS scores and postural control

Previous research has shown that sleep deprivation can decrease subjective alertness (Harma et al., 1998; Liu et al., 2001). However, in the present study, high subjective sleepiness scores did not correlate with increased body movement variances. Instead, in the only two comparisons in which we found a significant correlation, body movement was actually lower among the subjects that subjectively regarded themselves as the sleepiest. This suggests that subjective sleepiness may not be a reliable indicator of actual postural control performance (Fabbri et al., 2006). These findings therefore highlight the value of implementing regulations for work durations, particularly in attention demanding occupations, such as long distance driving, because subjective sleepiness may not reflect actual performance decrease, which potentially could cause safety hazards and traffic accidents. Although sleepiness and fatigue can be subjectively assessed, such evaluations may not reflect the objective physiological status of the tired person, because subjective scores can be biased by motivation, personal factors, experience, training, etc (Avni et al., 2006). Additionally, the mere act of performing a test might momentarily enhance attention and motivation (Avni et al., 2006). Therefore, there is a need for objective evaluation methods to determine actual performance capacity during sleep deprivation.

4.4. Vision and sleep deprivation

Our findings support evidence from Edwards showing that vision can provide information to assist postural control (Edwards, 1946), as sleep deprivation had less effect on body movement variance with eyes open compared with eyes closed in the anteroposterior direction. In most cases, head and shoulder movement variances were significantly larger in tests conducted with eyes closed compared with eyes open. However, although the EC/EO quotients suggested that the body movement pattern was different while sleep deprived, we were unable to find statistical evidence showing that vision was more important while sleep deprived. Additionally, the near-falls and decreased adaptation occurred both in tests with eyes open and eyes closed. Hence, although visual information provided enhanced stability, this additional information seemed not to be sufficient to compensate fully for the deterioration of performance caused by sleep deprivation.

4.5. EC/EO quotients and sleep deprivation

In normal conditions, unperturbed stance induces continual body movements that resemble an inverted pendulum, with proportionally larger movements at the head and shoulder than at positions closer to the support surface. Consistent with this body movement strategy, we found that the proportional changes of the EC/EO quotients were the same in all body segments after a normal night of sleep both during Quiet Stance and during balance perturbations.

However, at 24 h of sleep deprivation, the Quiet Stance body movement pattern was different, reflected by the finding that knee, shoulder and head movements were proportionally changed more between eyes closed and eyes open tests than hip movements. This finding suggests that the subjects used a more precautious hip movement strategy during Quiet Stance while standing with eyes closed. Some data values suggest that this movement pattern was maintained during the first stimulation period, though the latter observation could not be statistically confirmed. After about 150s of balance perturbations the subjects appeared to have returned to using a single-link pendulum movement strategy though with larger body movements. This finding suggests that while sleep deprived, the segmental movement pattern during Quiet Stance and in response to repeated balance perturbations is changed, and that the ability to select an appropriate movement pattern appears to be slower than in normal conditions (Maki and Whitelaw, 1992; Chong et al., 1999).

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Article VI

Adaptation and vision change the relationship between muscle activity of the lower limbs and body movement during human balance perturbations

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Adaptation and vision change the relationship between muscle activity of the lower limbs and body movement during human balance perturbations

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ABSTRACT

Objective: Investigate the relationship between changes in lower limb EMG root mean square (RMS) activity and changes in body movement during perturbed standing. Specifically, linear movement variance, torque variance and body posture were correlated against tibialis anterior and gastrocnemius RMS EMG activity during perturbed standing by vibration of the calf muscles.
Methods: Eighteen healthy participants (mean age 29.1 years) stood quietly for 30 s before vibration pulses were randomly applied to the calf muscles over a period of 200 s with eyes open or closed. Movement variance, torque variance and RMS EMG activity were separated into five periods, thereby allowing us to explore any time-varying changes of the relationships.
Results: Changes of tibialis anterior muscles EMG activity were positively correlated with changes in linear movement variance and torque variance throughout most of the trials, and negatively correlated with some mean angular position changes during the last 2 min of the trials. Moreover, the initial changes in Gastrocnemius EMG activity were associated with initial changes of mean angular position. Additionally, both tibialis anterior and gastrocnemius muscle activities were more involved in the initial control of stability with eyes closed than with eyes open.
Conclusions: Visual information and adaptation change the association between muscle activity and movement when standing is perturbed by calf muscle vibration.
Significance: Access to visual information changes the standing strategy to calf muscle vibrations. Training evoking adaptation could benefit those susceptible to falls by optimising the association between muscle activities and stabilising body movement.

1. Introduction

Everyday experience suggests that we are able to improve multiple motor skills through practice and this is more commonly termed adaptation. In current approaches to motor learning, adaptation is viewed as a process in which prediction errors result in proportional changes in parameter estimates (Krakauer et al., 2006). More recently, studies have indicated that adaptation through motor learning can be applied to the human standing posture during balance perturbations (Fransson et al., 2007b; Fujiwara et al., 2007). As such, repeatedly perturbing balance could be one way of decreasing the number of falls in those at risk. The maintenance of the human standing posture depends on the availability and accuracy of somatosensory (muscle, joint, skin and pressure receptors), visual and vestibular sensory inputs and descending commands from the central nervous system (CNS) (Akram et al., 2008). When the information from one or more of the sensory inputs becomes unreliable, a re-weighting occurs as the CNS places an increased demand on the reliable system or systems (Oie et al., 2002). Hence, one way of perturbing balance is by altering the information from the somatosensory receptors through vibration of the calf muscles. By vibrating the calf muscles, body sway and torque at the ankles increases (Fransson et al., 2000; Ivanenko et al., 1999; Kavounoudias et al., 1998) and when the vibrations are repeated, the adaptations are evidenced as a decrease in movement variability and ankle torque (Fransson et al., 2007b).

In patients with suspected balance disorders, it is often useful to assess the muscle activity using electromyography (EMG), along with other recordings of balance such as exerted forces and body movements, to determine the severity of disorder or rehabilitation status. However, additional important information might also be gained by analysing the way EMG activity relates to body movement. Some authors have suggested that since angular acceleration is proportional to joint torque in single joint movements, there should be a clear relationship between kinematics and EMG activ-

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ity (Gabriel, 2002; Soechting and Flanders, 1991). St-Onge and Feldman have also suggested that EMG activity of multiple muscles should correlate with the direction, magnitude and velocity of movement (St-Onge and Feldman, 2004). However, the relationship between muscle activity and exerted body movements might change through adaptation (Buchanan and Horak, 2003). In addition, the availability of key information from the visual system might also influence this relationship (Buchanan and Horak, 1999).

During balance perturbations, postural muscles in the lower extremities such as the tibialis anterior and gastrocnemius are responsible for counteractive movements (Gollhofer et al., 1989). Two key goals of the present study, then, were to: (1) identify whether adaptation through balance training (i.e., repeated calf muscle vibration) affects the relationship between tibialis anterior and gastrocnemius EMG activity and body movement, recorded as segmental body movements, body posture and exerted torque to the support surface and (2) whether access to vision can affect these relationships. These goals would provide clinicians with information about the importance of the visual system and usefulness of repeated balance perturbations for rehabilitation. We hypothesize that there would be a correlation between tibialis anterior and gastrocnemius muscle activity and body movement and that this relationship might be affected by adaptation due to the online updating of motor performance (Pavol and Pai, 2002; Pavol et al., 2002) but should not be affected by the availability of visual information.

2. Methods and materials

2.1. Subjects

Posturographic tests were performed on 18 healthy subjects (9 men and 9 women; mean age 29.1 years, range 18–49 years; mean height 1.74 m, range 1.50–1.85 m; mean mass 73.4 kg, range 58.1–95.0 kg). Subjects had no previous history of balance deficits, neurological disease or injury to the musculoskeletal system of the lower extremities, nor were they taking any medication and were instructed to refrain from alcohol for at least 24 h prior to testing. Full, informed consent was obtained from all subjects before any experiments were performed in accordance to the Helsinki declaration of 1975.

2.2. Equipment

Vibration of amplitude 1.0 mm and frequency of 85 Hz was produced using a DC-motor (Escap, Geneva, Switzerland) with a 3.5 g mass attached eccentrically to the spindle within a cylindrical plastic coating of dimensions 6 cm length and 1 cm diameter. The vibrators were attached over the middle of the gastrocnemius muscles on both legs using elastic straps.

A force platform, developed in co-operation with the Department of Solid Mechanics, Lund Institute of Technology, recorded forces actuated at the feet with six degrees of freedom and with an accuracy of 0.5 N. A customized computer program controlled the vibratory stimulation, and sampled the force platform data at 50 Hz.

An ultrasound 3D-Motion Tracking system (Zebris™ CMS-HS Measuring System) recorded linear body movements at five anatomical landmarks. The first marker ('Head') was attached to the subject's cheekbone (os zygomaticum), the second marker ('Shoulder') to tuberculum majus, the third ('Hip') to the crista iliaca, the fourth ('Knee') to the lateral epicondyle of femur, and the fifth ('Ankle') to the lateral distal fibula head, see Fig. 1. Each marker registered its position in three directions, i.e., its anteroposterior, lateral and vertical position with a resolution of 0.4 mm. The same Zebris™ system simultaneously recorded EMG activity of the tibialis anterior and gastrocnemius medialis muscle of both legs using eight active surface electrodes. The computer sampled the marker position data simultaneously at 50 Hz and EMG activity at 1500 Hz.

The recorded data from the force platform and Zebris™ measurement systems were later synchronised by off-line time matching of a reference signal, which had been simultaneously sampled by both measurement systems.

2.3. Procedure

The five position markers were attached on the right side of the subject, directly facing the transmitter of the Zebris™ system. EMG surface electrodes were fixed on the skin over the middle and upper end of the tibialis anterior and the gastrocnemius medialis muscles on both legs and the vibrators strapped in place on both legs. Each subject was then instructed to stand barefoot on the force platform in a relaxed posture with arms folded across the chest. The subject's heels were 3 cm apart with feet at an angle of 30° open to the front using guidelines on the force platform. Subjects were instructed to focus on a target 1.5 m in front at eye level or stand with their eyes closed depending on the test condition.

The following 2 tests were performed by all subjects in a randomized order using a Latin Square design:

- Vibration of the calf muscles with eyes closed (EC-Calf) and eyes open (EO-Calf).

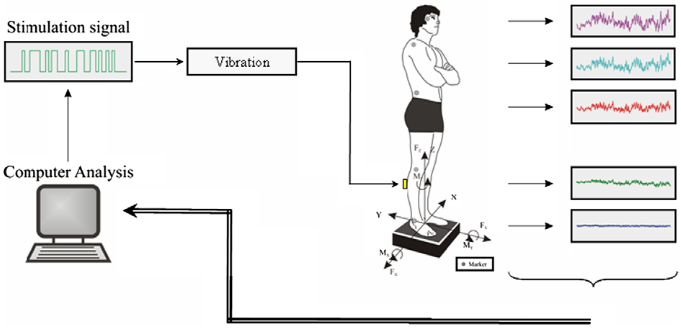

Fig. 1. Schematic picture of the measurement setup and placement of the five Zebris markers attached to a subject standing on a force platform. The marker locations are shown as small circles.

Before vibration, a 30-s control period of quiet stance was recorded. The vibratory stimulation pulses were applied according to a pseudorandom binary sequence (PRBS) schedule (Tjernstrom et al., 2002) during a period of 200 s making the combined test 230 s long. Each pulse had random time durations between 0.8 and 6.4 s. The PRBS stimulation sequence was selected because the randomised vibratory stimulation is difficult to predict and therefore lessened the likelihood of pre-emptive responses and is known to result in postural adaptation (Fransson et al., 2007b). Additionally, the PRBS stimulation sequence has a broad effective bandwidth in the region of 0.1–2.5 Hz. There was a 5-min rest period between the two tests.

2.4. Analysis

Vibratory calf muscle stimulation induces body movement primarily in the anteroposterior direction. Therefore only responses in this direction are considered here. The anteroposterior linear body movements were quantified in terms of movement variance at the head, shoulder, hip and knee, as recorded with the Zebris^{TM} system using the formula below, in this example calculated for the head marker:

$$\bar{x}_{Head} = \sum_{i=1}^{n} \frac{x_{Head}(i)}{n}$$

$$x(head)_{\mathrm{var}} = \frac{1}{n-1} \sum_{i=1}^{n} (x_{Head}(i) - \bar{x}_{Head})^2$$

where $x(\mathrm{head})_{\mathrm{var}}$ represents the variance of the linear anteroposterior head movements and x represents the marker's sampled anteroposterior position under the period analysed. The anteroposterior torque variance was calculated using the same formula as above, where the x = torque exerted in anteroposterior direction to the surface recorded by a force platform (Fransson et al., 2007a). The mean angular position with respect to the ankle joint was calculated for each marker position using the ankle marker as the zero-position reference point and using the vertical and anteroposterior linear perpendicular distances to the marker (Fransson et al., 2007a) according to the formula below, in this example calculated for the head marker:

$$\bar{x}_{Head} = \sum_{i=1}^{n} \frac{x_{Head}(i)}{n} \quad \bar{x}_{Ankle} = \sum_{i=1}^{n} \frac{x_{Ankle}(i)}{n}$$

$$\bar{z}_{Head} = \sum_{i=1}^{n} \frac{z_{Head}(i)}{n} \quad \bar{z}_{Ankle} = \sum_{i=1}^{n} \frac{z_{Ankle}(i)}{n}$$

$$x(head)_{ang} = \arcsin\left(\frac{\bar{x}_{Head} - \bar{x}_{Ankle}}{\bar{z}_{Head} - \bar{z}_{Ankle}}\right)$$

where $x(\mathrm{head})_{\mathrm{ang}}$ represents the mean angular position of the head, x represents the marker's sampled anteroposterior position and z the marker's sampled vertical position under the period analyzed. The Zebris measurement accuracy allowed the marker angular values to be calculated with an error of less than 1.5%. If the mean angular positions of all individual markers are viewed upon together in a simple stick model, a view also of the entire body posture is obtained (Fransson et al., 2007a). Body posture and body movement amplitude commonly change independently of one another, and were therefore analysed separately as mean angular position and linear body movement variance, respectively (Fransson et al., 2007b).

When recording the EMG activity from the tibialis anterior and gastrocnemius muscles, a significant effort was made to determine that the crosstalk from muscles near the muscle of interest did not contaminate the recorded signal. EMG data from the tibialis anterior and gastrocnemius muscles of both legs were band-pass filtered, using 20 and 200 Hz, respectively, as frequency cut-off limits, and the root mean square (RMS) value was calculated (Fransson et al., 2007a). Gastrocnemius EMG signals were further notch filtered between 100 and 130 Hz to remove the distortion effects caused by vibratory stimulations of the calf muscles. The distortion in the EMG recordings due to the vibratory stimulation was at about 115 Hz. Notably, the distortion frequency was different from the mechanical vibration, which was at about 85 Hz. Hence, the most likely source of the distortions is the electrical device producing the vibration, not the mechanical vibration itself. No notch filtering was required for tibialis anterior EMG signals, as they were not distorted by vibration of the calf muscles. A fifth-order digital Finite duration Impulse Response (FIR) filter, selected to avoid aliasing, was used for spectral separation. Quiet stance EMG activity assessed during calf vibration with eyes open served as the reference (Fransson et al., 2007a). Hence, the EMG results presented for each test were normalised for each subject.

Each test was divided into five periods: Quiet Stance (0–30 s), and four 50 s stimulation periods (period 1: 30–80 s; period 2: 80–130 s; period 3: 130–180 s; period 4: 180–230 s). The selection of 50 s analysis periods were based on prior studies on how postural control are gradually affected by prolonged randomised vibratory proprioceptive stimulation (Tjernstrom et al., 2002).

2.5. Data analysis

Torque variance values were normalised to account for anthropometric differences between the subjects, using the subject's squared height and squared mass (Fransson et al., 2007b; Johansson et al., 1988). Similarly, the anteroposterior linear movement

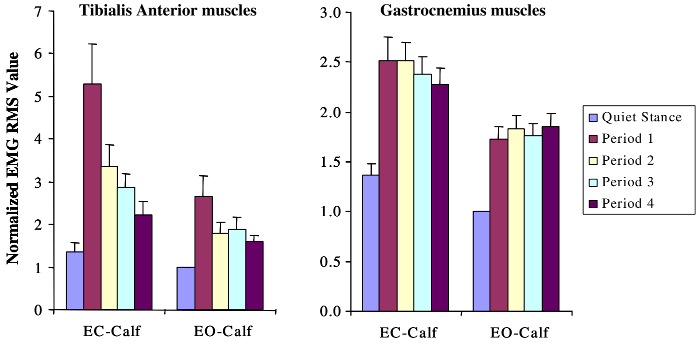

Fig. 2. Tibialis anterior and gastrocnemius EMG RMS values (mean and standard error of mean (SEM)).

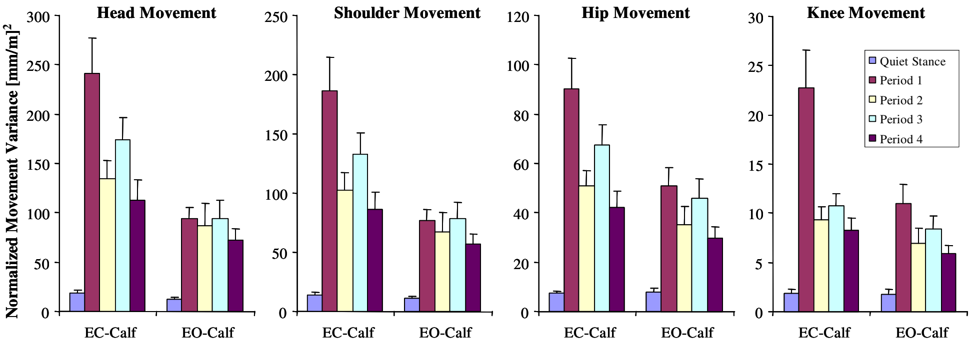

Fig. 3. Normalised movement variance values for anteroposterior linear head, shoulder, hip and knee movements (mean and standard error of mean (SEM)).

variance values were normalised using the subject's squared height before the statistical analysis. The averaged RMS values from the right and left tibialis anterior and gastrocnemius muscles were calculated and used in the analysis.

Four quotients were calculated to assess individual changes over time in RMS EMG activity; segmental body movement variance; body posture; and torque variance. The data on which the quotient calculations were done are presented in Figs. 2–5. The first quotient (dividing quiet stance values by stimulation period 1 values) shows how the assessed parameters were initially affected by the balance perturbations evoked by vibratory proprioceptive stimulation compared to the activity during quiet stance. The three other quotients (dividing stimulation periods 2, 3 and 4 values by stimulation period 1 values) show how the assessed parameters were gradually affected by repeated vibratory stimulation, quantifying possible effects of adaptation to vibratory proprioceptive stimulation.

2.6. Statistical analysis

The Wilcoxon non-parametric matched-pairs signed-rank test (Exact sig. 2-tailed) (Altman, 1991) was used for analysis of variations over time for each test condition. The changes between Quiet Stance and Period 1 in EMG RMS activity, body movement variance, mean angular position and torque variance were evaluated to determine how the assessed parameters were initially affected by vibratory proprioceptive stimulation under the test condition compared to the activity during quiet stance (Patel et al., 2008). The changes in these parameters between Period 1 and Period 4 were also evaluated to determine the totally gained improvement under the entire trial, quantifying possible effects of adaptation to vibratory proprioceptive stimulation (Patel et al., 2008). The Spearman two-tailed rank correlation coefficient test was used to analyze the correlation between the RMS EMG quotient values and the quotient values of linear movement variance, mean angular position and torque variance. Non-parametric statistics were used for the Spearman's correlations because values were not normally distributed using the Shapiro–Wilk test. In the analysis $p < 0.01$ were considered statistically significant (Altman, 1991). However, we present p-values <0.05 in the correlation figures for consistency.

3. Results

3.1. Recorded RMS EMG activity, linear body movement, mean angular position and torque variance

The tibialis anterior EMG RMS activity increased significantly during all test conditions in period 1 compared with quiet stance ($p < 0.001$), see Fig. 2. The EMG increases were about 300% for EC-Calf and 165% for EO-Calf. The tibialis anterior EMG RMS activ-

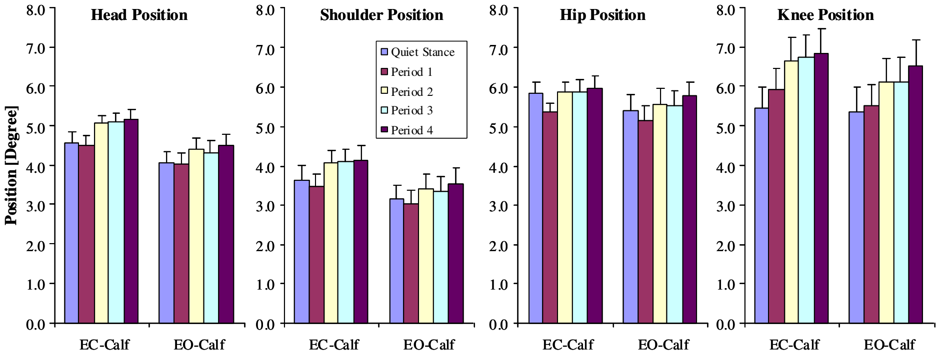

Fig. 4. Angular values for anteroposterior head, shoulder, hip and knee positions (mean and standard error of mean (SEM)).

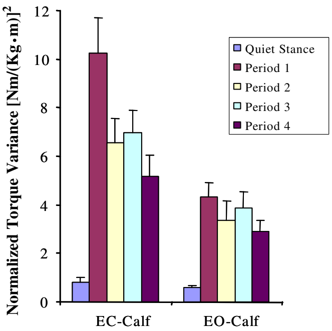

Fig. 5. Normalised torque variance values for anteroposterior linear head, shoulder, hip and knee movements (mean and standard error of mean (SEM)).

ity was significantly smaller in period 4 compared with period 1 during both test conditions ($p < 0.01$). The EMG RMS activity decreases were about 60% for EC-Calf and 40% for EO-Calf.

The gastrocnemius EMG RMS activity increased significantly during both test conditions in period 1 compared with quiet stance ($p < 0.001$). The EMG RMS activity increases were about 80% for EC-Calf and 70% for EO-Calf. In contrast to the tibialis anterior EMG RMS activity, the gastrocnemius EMG RMS activity was not significantly different in period 4 compared with period 1 during any of the test conditions.

The stimulus onset significantly increased body movement variances at all measured sites ($p < 0.001$), see Fig. 3. The significant movement variance increases for the tests was approximately 1180% with EC-Calf and 570% with EO-Calf at all positions.

Analysis of the variance values showed that with EC-Calf, there was an equal reduction of the movement variances at all measured sites by about 55% ($p < 0.001$) in period 4 compared with period 1. However, with EO-Calf another pattern was observed. With EO-Calf only the movement variances at the lower segments decreased significantly in period 4 compared with period 1, the knee movement variance by about 45% and the hip by about 40% ($p < 0.01$).

During calf vibration, the mean angular position did not significantly change between quiet stance and vibration period 1, see Fig. 4. Instead, with EC-Calf, the angular positions increased by approximately 15% at all measured sites in period 4 compared with period 1 (head ($p = 0.002$); shoulder ($p = 0.002$); hip ($p = 0.001$); knee ($p = 0.008$)), i.e., the subjects increased their leaning forward. Similarly, with EO-Calf, the angular positions increased in period 4 compared with period 1, (head ($p = 0.014$); shoulder ($p = 0.014$); hip ($p < 0.001$); knee ($p < 0.001$)).

The anteroposterior torque variance increased significantly during all tests in period 1 compared with quiet stance ($p < 0.001$), see Fig. 5. The increase in torque variance was about 1180% for EC-Calf and 665% EO-Calf. Moreover, the torque variance values were significantly smaller in period 4 compared with period 1 in both test conditions, EC-Calf ($p < 0.001$) and EO-Calf ($p < 0.01$). The decrease in torque variance for the tests was about 50% for EC-Calf and 35% for EO-Calf.

3.2. Correlations between alteration of RMS EMG activity and alterations of linear body movement, mean angular position and torque variance

3.2.1. Linear body movement variance and RMS EMG activity

When studying the initial effects of balance perturbations (see QS/P1 quotient correlation) we found that the increase in muscle activity of the tibialis anterior muscles correlated to the increases in linear body movement variance at all positions (head, shoulder and hip, $p < 0.01$; knee $p < 0.001$) during the EC-Calf test (Figs. 3 and 6). In the following period, we observed a sharp decrease in linear body movement variance in all tests, which levelled off in Periods 3 and 4 (Fig. 3). However, the initial decrease in the tibialis anterior muscle activity did not reflect the adaptive decrease in linear movement variance at any position during EC-Calf, see P2/P1 quotient correlation. Additionally, when reaching Period 3 of EC-Calf test (see P3/P1 quotient), there was a significant correlation between the decrease in tibialis anterior muscle activity and the decrease in linear body movement variance at all positions except

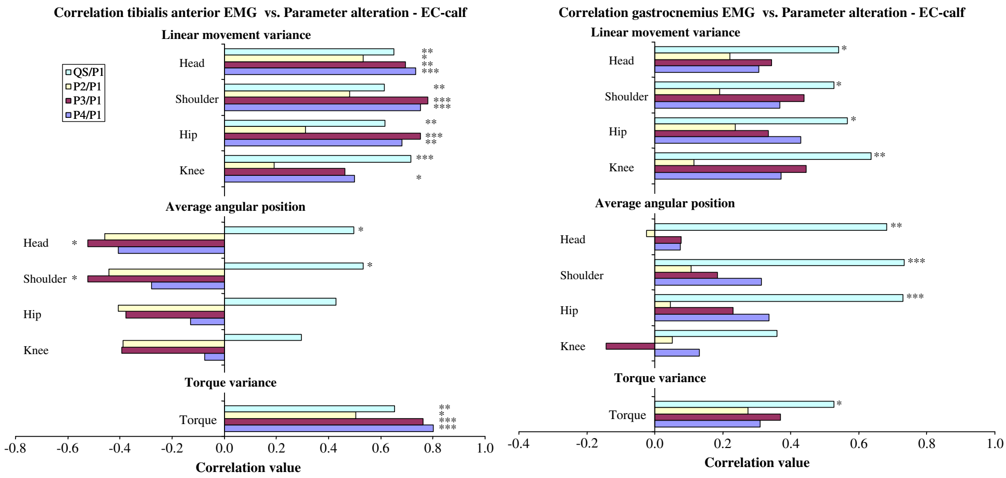

Fig. 6. Correlation values between alterations in tibialis anterior and gastrocnemius RMS EMG activity and alterations of linear body movement, mean angular position and torque variance with EC Calf. The statistical differences found are marked with asterisk, where $^{*}p < 0.05$, $^{**}p < 0.01$ and $^{***}p < 0.001$.

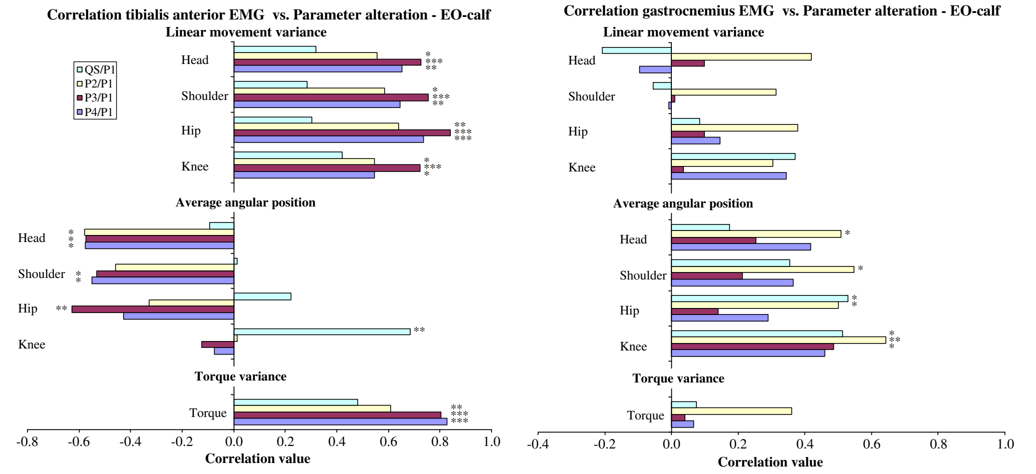

Fig. 7. Correlation values between alterations in tibialis anterior and gastrocnemius RMS EMG activity and alterations of linear body movement, mean angular position and torque variance with EO Calf.

at the knee (head, $p < 0.01$; shoulder and hip, $p < 0.001$), and these correlations were almost the same in Period 4 (head and shoulder, $p < 0.001$; hip, $p < 0.01$), see P4/P1 quotient.

In the EO-Calf test (see Fig. 7), there was no significant relationship between the initial changes (QS/P1 and P2/P1 quotient correlations) in tibialis anterior muscle activity and recorded changes in body movement variance (Fig. 3). However, during the second half of the test (P3/P1 quotient; at all positions, $p < 0.001$) and (P4/P1 quotient; head and shoulder, $p < 0.01$; hip, $p < 0.001$), the correlation values shows that the decrease in tibialis anterior muscle activity reflected the decrease in body movement variance.

At the onset of vibration (QS/P1 quotient), the increase in gastrocnemius muscle activity correlated with the increase in linear body movement variance at the knee (knee $p < 0.01$) in the EC-Calf test (see Figs. 3 and 6). In the EO-Calf test, there was no significant relationship between the changes in gastrocnemius muscle activity and changes linear body movement variance, i.e., whereas movement variances decreased markedly over time in these tests, the gastrocnemius muscle activity changes were not associated with these changes (see Fig. 7).

3.2.2. Mean angular position and RMS EMG activity

At the onset of vibration (QS/P1 quotient), the initial increase in tibialis anterior muscle activity correlated with a small increase in mean angular position at the knee ($p < 0.01$) in the EO-Calf test, see Fig. 6. Although there were some clear trends between decreased tibialis anterior activity and increased mean angular position in a number of comparisons in both EC-Calf and EO-Calf tests, the adaptive decrease in tibialis anterior muscle activity only significantly correlated negatively with the increase in the mean angular position at the hip between Period 1 and Period 3 ($p < 0.01$) during the EO-Calf test, see P3/P1 quotient. Hence, decreased tibialis anterior muscle activity was associated with increased leaning forward of the hip.

At the onset of vibration, the increase in gastrocnemius muscle activity correlated with the increase in mean angular position at the head ($p < 0.01$), shoulder ($p < 0.001$) and hip ($p < 0.001$) in EC-Calf, see QS/P1 quotients Figs. 4 and 6. In contrast, there was no indication of a relationship between gastrocnemius muscle activity and mean angular position with EC-Calf at any position during the reminder of the test. During the remaining test periods, the small change in gastrocnemius muscle activity only correlated with a slight change in mean angular position forward at the knee in EO-Calf ($p < 0.01$), see P2/P1 quotient Figs. 4 and 7.

3.2.3. Torque variance and RMS EMG activity

When studying the initial effects of balance perturbations (see QS/P1 quotient, Fig. 6) we found that the large increase in tibialis anterior muscle activity correlated with the initial increase in torque variance with EC-Calf ($p < 0.01$). However, the decrease in the tibialis anterior muscle activity from Period 1 to Period 2 (see P2/P1 quotient) did not significantly correlate with the decrease in torque variance. Though, when reaching Period 3 and Period 4 of EC-Calf test (see P3/P1 and P4/P1 quotients), there was a significant correlation between the decrease in tibialis anterior muscle activity and the decrease in torque variance ($p < 0.001$).

In the EO-Calf test, there was no significant relationship between the initial changes (QS/P1 quotient) in tibialis anterior muscle activity and recorded toque variance, see Fig. 7. However, in the last three periods of the test (P2/P1 quotient, $p < 0.01$; P3/P1 and P4/P1 quotients, $p < 0.001$), the correlation values shows that the decrease in tibialis anterior muscle activity reflected the adaptive decrease in torque variance.

The initial increase in gastrocnemius muscle activity did not significantly reflect any of the changes in torque variance during any of the tests. Additionally, we found no evidence in any of the tests that the changes in gastrocnemius muscle activity were significantly related to the large decrease in torque variance.

4. Discussion

4.1. Relationship between EMG activity and movement

Both adaptation and somewhat surprisingly the availability of visual information affected the relationship between tibialis anterior and gastrocnemius muscle activity and body movement. During the continuous perturbations using a pseudorandom sequence of vibration pulses the postural challenge, although still threatening, became more controllable, evidenced by a reduction in the body movement variance and torque variance, see Figs. 3 and 5. On the whole, the postural stability as assessed by body movement and torques towards the surface improved rapidly until vibration Period 3, followed by no real change between Periods 3 and 4. These patterns are similar to the ones we have found previously, and we have deemed that at this plateau phase the adaptation has subsided (Fransson et al., 2002). Tibialis anterior muscle activity and body posture also showed adaptive responses; tibialis anterior muscle activity decreased (Fig. 2) and body posture leaned further forward (Fig. 4), though there was no adaptive behaviour in the gastrocnemius muscle activity (Fig. 2). These findings are consistent with the findings by Nardone et al., 2000, showing that within the first few cycles of a balance perturbation, participants predict the characteristics of perturbations and their destabilizing effects, and set their balance control system to minimize these effects (Akram et al., 2008).

When considering the initial changes in response to the balance perturbations, we found that tibialis anterior muscle activity changes correlated well with torque variance and body movement variance changes during eyes closed only, and gastrocnemius muscle activity changes correlated with head, shoulder and hip angular position changes also during eyes closed only. This finding shows that vision has a significant influence on the relationship between muscle activity and recorded body movements in the initial phase of exposure to balance perturbations. Therefore, the presence or absence of vision dramatically changes the strategy employed for the maintenance of postural stability (De Nunzio et al., 2005). This finding is similar to Buchanan and Horak (1999) that without visual information, EMG activity of muscles including the tibialis anterior and gastrocnemius were associated with slow drift of the head and Centre of Mass (CoM) motion, suggesting that either otolith or somatosensory information trigger the muscle responses. Additionally, the initial control of linear movement correlated to both tibialis anterior and gastrocnemius muscle activity. Another implication from the presented findings is therefore that without visual information, initial postural stability might be enhanced through co-contraction of the tibialis anterior and gastrocnemius muscles. However, this initial relationship changed over time, as tibialis anterior activity decreased between periods 1 and 4, which is consistent with reports suggesting that co-contraction can only be maintained for a short period of time before muscular fatigue occurs (Hogan, 1984). One may question whether the EMG activity in tibialis anterior and gastrocnemius was changed due to adaptation or merely as an effect of increased body leaning forward. Some findings in this study suggest that the body leaning forward was of importance for the tibialis anterior and gastrocnemius EMG activity but findings also suggest that body leaning might not be the only factor influencing the EMG activity. Most of the tibialis anterior EMG activity reduction occurred during period 1 to period 3 under the same periods the body leaning was most notably changed forward. Additionally, we found several correlations at $p < 0.05$ between tibialis anterior, gastrocnemius EMG activity changes and mean angular position changes, see Fig. 7, so a relationship between increased body leaning forward and muscle activity reduction can not be excluded. However, the body leaning changes in degrees were about the same with eyes open and eyes closed. Nonetheless, the size and changes of the tibialis anterior and gastrocnemius EMG activity were clearly larger with eyes closed than with eyes open which suggests that vision influenced the muscle EMG activity and the adaptation of the EMG activity independently of body leaning.

Vibratory stimulus of the gastrocnemius muscle can directly influence the fusimotor activity in the muscle and thereby the EMG activity recorded. However, the gastrocnemius EMG activity had almost identical properties when the subjects were exposed to neck stimulation as when exposed to calf stimulation (unpublished observations), which suggests that the vibratory stimulation of the gastrocnemius muscle do not have a major detectable influence on the recorded EMG activity after the precautious filtering procedures used in the study.

4.2. Muscle activity and adaptation

The correlations between tibialis anterior and gastrocnemius muscle activities and the recorded body movement parameters were prone to adaptation, see Figs. 6 and 7. The correlation coefficients between gastrocnemius muscle activity and body posture were larger during the initial increase in Period 1 and the initial adaptation in Period 2, whereas there was a significant relationship between changes in tibialis anterior muscle activity and movement variance in Periods 1, 3 and 4 of vibration. Hence, the relationship between muscle activity and body movement is complex, and cannot simply be through a parallel change of EMG activity and movement variance. The control of postural stability is regulated by postural muscles that form an uninterrupted muscular chain that extends from the head to the feet (Roll et al., 1989) as confirmed by the induced balance perturbations caused by proprioceptive vibration at various locations (Courtine et al., 2007). Thus, one possibility is that during some periods of time some other postural muscles along the muscular chain may influence the body movements more than the gastrocnemius and tibialis anterior muscles. Alternatively, the reduced correlation between gastrocnemius muscle activity changes and body posture changes after 100 s of vibration suggests that once the level of movement variance had decreased sufficiently through adaptive mechanisms, gastrocnemius may not have the same role in postural control. The muscles shifting role in postural control is also illustrated by the lack of correlation between linear body movement and tibialis anterior EMG activity between Period 1 and Period 2, possibly because the muscle activity could not be suppressed to the same extent as body movement, particularly with eyes closed, since the muscles during this phase also had an important role for postural stability.

A surprising finding was that the positive correlations between tibialis anterior muscle activity and linear movement variance and the negative correlation between tibialis anterior muscle activity and an increased mean angular position (i.e., forward leaning) generally increased in the P3/P1 and P4/P1 quotients. This latter period of the test represents a settling period where subjects have adapted to use less energy to maintain postural stability (Fransson et al., 2002), and our findings show that this is by an increased control of movement through forward leaning, and evidenced also by lower tibialis anterior muscular RMS activity (Fig. 2) and decreased movement variance (Fig. 3). Benefits of this adaptation is decreased risk for muscular fatigue and an enhanced standing postural strategy (Mihelj et al., 2000), since by forward leaning the reliance on sensory feedback is reduced (Madigan et al., 2006) and the muscle spindles in the plantar flexors gain improved ability to sense changes in muscle length and velocity due to the increased gamma motor neurone drive (Madigan et al., 2006).

As somewhat expected by the negligible change in gastrocnemius muscle activity compared with the changes in body movement and mean angular position, there was almost no relationship between these variables at our Bonferroni-corrected level of significance ($p < 0.01$). This furthers and corroborates the findings by Loram et al. (2005) showing little relationship between gastrocnemius EMG and CoM movements under quiet stance. Furthermore, this implies that, although the gastrocnemius muscles are important in the regulation of the upright standing posture, particularly with sudden balance perturbations, the gastrocnemius muscles might not be fully associated with the tonic maintenance of postural control.

4.3. Muscle activity and movement

Several previous investigations of the relationship between muscle activity and body movement are largely based on theoretical presumptions (Riccio and Stoffregen, 1988; Soechting and Flanders, 1991; St-Onge and Feldman, 2004) or based on studies of arm movements (Darling and Cooke, 1987a; Darling and Cooke, 1987b; Gabriel, 2002). Furthermore, although a strong link between a single muscle and a single joint may be established for some tasks, this is most probably not the case for multi-joint tasks. For example, work employing a two joint arm system (Kelso et al., 1991) demonstrated that prime movers drop out when inertia can accomplish the same action. These models might not therefore be adequate enough to illustrate the complex relationship between local muscle activity and recorded body movements in upright standing posture. Additionally, several findings in this study suggests that several partly independent factors such as vision, body leaning and adaptation may change the relationship between muscle activity and recorded movements. Moreover, complex relationships and adaptive changes might be more obvious when studying the effects over a long period of time, such as the 50 s periods used in the present study, rather than studying short periods of EMG activity directly associated with a particular movement.

4.4. Clinical significance of findings

It is well-known that to maintain upright stance, the central nervous system (CNS) must coordinate motion across many joints and muscles using sensory information provided by the visual, somatosensory and vestibular systems (Akram et al., 2008). The multiple segments of the body are inter-connected (Ivanenko et al., 2000), and as evidenced in this study, a local change in proprioceptive information led to a widespread alteration in posture remote from the vibration site, thus complimenting the findings by Ivanenko et al. (2000) and Thompson et al. (2007). In other words, the CNS must use different strategies for appropriate balance control when the information from one of the sensory receptors is unreliable. However, in some commonly used posturography tests there is sometimes no detectable change in balance, even in patients with sensory disorders, as sensory reweighting shifts the reliance of afferent information from unreliable sources to other more reliable receptors (Oie et al., 2002). Therefore, the finding in the present study that the change in strategy is detectable when assessing the correlation between muscle activity and body movements during balance perturbation, might warrant a new balance testing approach to asses rehabilitation effects. For example, the used approach might be beneficial assessing patients recovering from surgical procedures performed on the neuromuscular or musculoskeletal system affecting postural control, or to check whether the appropriate control of posture and balance control is gained from vision. Furthermore, postural control's remarkable ability to learn how to handle vibration-induced balance perturbations as illustrated in this study supports the idea that proprioceptive vibration training could be used as a rehabilitation technique. This is particularly true for the elderly because while the elderly fall frequently when surface somatosensory information is altered, they become capable of maintaining normal steadiness after repetitive experience (Woollacott et al., 1986).

5. Conclusions

Both adaptation and the availability of visual information affected the relationship between tibialis anterior and gastrocnemius muscle activity and body movement. Without visual information, initial postural support might be enhanced through the use of co-contraction of the tibialis anterior and gastrocnemius muscles. However, these initial relationships changed over time as an effect of adaptation. Thus, adaptation training using vibratory proprioceptive stimulation could benefit those susceptible to falls by changing the association between muscle activity and movement.

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