BODY MOVEMENTS DURING POSTURAL STABILIZATION

Measurements with a motion analysis system

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Abstract

Good postural stability is needed during most activities in daily life. Balance can be improved with specific training programmes in physiotherapy. The goals of treatment differ, depending on the subject's age and disorders. In clinical practice, postural stability is commonly evaluated by scaled functional tests or by recording body sway on a platform. On the basis of therapeutic modalities, information of separate movements of body segments would be important. The data might be useful in developing balance evaluation and training programmes in physiotherapy.

The aim of this project was to present one method and to estimate its reliability and validity in studying the movements of separate body segments in postural control. In addition, the association between age, gender and anthropometric factors and the movements of separate body segments during quiet stance with the eyes open and closed were studied in a group of healthy subjects.

A method was developed to measure the body movements during standing with a motion analysis system, and the reliability of body movement measurements was evaluated. The validity of the motion analysis measurements was evaluated to compare the balancing body measurements during stance on two legs and on one leg obtained with a motion analysis system and a platform. In cross-sectional studies, 100 healthy randomly selected subjects were stratified into ten groups (by age and gender). The body movements of all subjects standing on two legs with the eyes open and closed were measured using a motion analysis system and calculated as maximal and total movements. The movement velocities and accelerations were analysed and compared between the eyes open and eyes closed conditions. The associations between movement values and age and gender were analysed. In addition, the body anthropometrics of the subjects were measured and the relations between the body characteristics and the body balancing movements were calculated using regression analysis.

The results showed that motion analysis can be used in measuring body movements in postural stability. Better reproducible balance measurement results are obtained with the total movement values than with the maximal amplitude values. In a comparison of the parameters used in a validity study, motion analysis and platform seemed to reflect the same aspect of balance, although the views of measurement were different. During standing on two legs with the eyes open, there was a statistically significant difference in the maximal anterior-posterior head movement and in the vertical navel movement between the age groups, but the results did not show other statistically significant differences between the balancing movements of separate body segments of the groups or between the balance measurement values of men and women in standing on two legs with the eyes open and closed. It seems that healthy female and male subjects control their stance with quite similar ranges of body adjustment. Body characteristics had slight but considerable effects on the variations of body balancing movements in standing on two legs with the eyes open, but almost none in the eyes-closed conditions. There were differences in the results between the male and female groups. In standing on two legs with the eyes closed, all the measured body parts except the ankles had significantly higher maximal velocity and acceleration values than in standing with the eyes open. The effect of visual information on balancing the body seems to be essential. The results indicated that the motion analysis system is also a useful tool in further balance studies, but the methods of analysis need to be developed. Postural stability should be evaluated and practised even in more demanding balance performances. Movement speed and the special role of each body part in maintaining balance should be paid attention.

Keywords: acceleration, age, anthropometrics, Reliability, validity, velocity

To my family and parents

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Abbreviations

A/D	Analog/digital
Ant-post	Anterior-posterior
ANOVA	One-way analysis of variance
BMI	Body mass index
CNS	Central nervous system
COG	Centre of gravity
COP	Centre of pressure
d	Distance
EMG	Electromyogram
Fig.	Figure
Hz	Herz
ICC	Intraclass correlation coefficient of reliability
n	Number of subjects
NS	Not statistically significant
p	p-value = observed significance level
r	Correlation coefficient
R_p	Pearson's correlation coefficient
R_s	Spearman's correlation coefficient
R^2	Squared correlation coefficient
RMS	Root mean square
s	Second
S1	Sacral (I) vertebra
SD	Standard deviation
SEM	Standard errors of measurement
SPSS	Statistical package for social science
x	Lateral dimension
y	Anterior-posterior dimension
z	Vertical dimension
3D	Three-dimensional

List of original publications

This thesis is based on the following articles referred to in the text by their Roman numerals.

I Kejonen P, Kauranen K & Vanharanta H. (1998) Body movements in postural balance with motion analysis. Eur J Phys Med Rehabil 2:8; 39-43.

II Kejonen P & Kauranen K (2002) Reliability and validity of standing balance measurements with a motion analysis system. Physiotherapy 88:1; 25-32.

III Kejonen P, Kauranen K, Ahasan R & Vanharanta H. Motion analysis measurements of body movements during standing. Association with age and gender. Int J Rehabil Res, in press.

IV Kejonen P, Kauranen K & Vanharanta H. The relationship between anthropometric factors and body balancing movements in postural balance. Arch Phys Med Rehabil, in press.

V Kejonen P, Kauranen K & Vanharanta H. Velocities and accelerations of body parts during standing. Association with visual information. Submitted.

Contents

1 Introduction

Proper balancing movements of separate body segments are essential in maintaining upright stance, firstly, to stabilize the body segments, and secondly to enable the necessary dynamic changes of the body. The balancing of the human body is a complicated process, which demands coordination of the sensory, skeletal, muscle and central nervous systems. First, stimuli from visual, vestibular and somatosensory sources are required to contribute information about the body's position in space. Soon thereafter, coordinated responses to stimuli must be transmitted to the appropriate muscles to produce corrective movements in certain joints to balance a standing position.

Good postural stability is needed in most physical functions and activities of daily life. Even reaching out for a piece of paper during quiet stance requires that the system to control not only the raising of the arm but also postural stability. Influx from proprioceptors becomes more important in a darkened environment, and body position is thereby controlled more by small and rapid balancing movements around the ankles. On a slippery surface, even the help of sensory cues is negligible and the need to control the body position with appropriate corrective movements all over the body increases. It has been pointed out that a young person can manage this kind of postural control better than older subjects, who therefore run a bigger risk for falls. With the increase of our ageing population, the importance of maintaining balance and mobility is becoming even more critical.

In physiotherapy, balance performance can be improved with specific training programmes. For example, postural stability can be improved by including interrelated postural adjustments and limb movements in various tasks. Besides, it is recommended that muscle and coordination exercises be included in training programs. It has been reported that practising a symmetrical upright posture has positive effects on balance control in some cases. In aged populations and patient groups, the goals of treatment obviously differ. To develop therapeutic modalities, information of the physical characteristics of body segments, e.g. amplitudes, frequencies, velocities and accelerations, during standing would be needed.

In clinical practice, postural stability is commonly evaluated by scored functional tests, which give information of balance performance in common daily tasks. In laboratory circumstances, postural stability has also been assessed by recording the spontaneous postural sway on a platform, to gain information about postural control mechanism as shifts of the centre point of pressure. As an attempt to enhance knowledge of the balance control mechanism, this thesis presents one possible method to analyse postural stability and tests for its reliability and validity by studying movements of separate body parts during standing with motion analysis. In addition, the body movements of healthy subjects and the factors that contribute to them are examined during standing on two legs with the eyes open and closed.

2 Review of the literature

2.1 Terminology

The terminology used in balance and gait research depends on the perspective through which this phenomenon is approached. According to the International Dictionary (Becker *et al.* 1986), the term “postural” is defined as “relating to posture or position”, whereas the term “posture” is specified as “the physical disposition of the body” and “position” an “an arrangement of bodily parts “.

‘Balance’ can be described as the ability to maintain the body’s position over its base of support (Berg *et al.* 1989, Spirduso 1995). Balance, or “a state of equilibrium” (Becker *et al.* 1986), has been divided into static balance and dynamic balance, depending on whether the base is stationary or moving (Spirduso 1995). Since the human body is never absolutely stable, a control system is required to stabilize the body. Such words as “postural control” and “balance control” have been used parallel to refer to the act of returning or keeping the body close to the equilibrium point (Karlsson & Fryktberg 2000). In detail, maintaining postural control means keeping the body’s centre of gravity (COG) within the borders of the base of support (Nashner 1985).

Postural stability can be defined as the maintenance of an upright posture during quiet stance (Pyykkö *et al.* 1988, Toppila & Pyykkö 2000). Hence, the supporting portions of the body must be stabilized during the segmental oscillation (Rothwell 1994). In the present study, postural stability is referred to as balanced stance, which can be achieved by coordinated movements, and these movements of separate body segments are examined with a motion analysis system during standing on a stationary base.

2.2 Postural control

During quiet stance, healthy subjects control their upright posture with small movements made in different segments of the body (Nashner 1985, Carr & Shepherd 1990). The optimal position during balanced stance requires that the centre of the body mass is maintained within the support frames of the soles. In the lateral direction of body sway, keeping the feet apart gives the best base of support, i.e. introduces a diagonal force against the ground. The shoulders should be directly above the hips and the head and trunk erect (Carr & Shepherd 1982). Balanced stance also requires an ability to move one's position while standing and to move out of the standing position, all without using the arms for support. This includes an ability to shift weight in the lateral and anterior-posterior directions (Carr & Shephard 1987) and to make flexible movements in the vertical direction (Woollacott & Shumway-Cook 1990). Postural activity is specific to the balance tasks, and during quiet stance, no conscious activation of muscles by the nervous system is required (Enoka 1994).

Gaining postural stability after body perturbation is controlled by three motor systems, as shown summarised in Table 1 (Diener & Dichens 1986, Schmidt 1991, Nashner 2001, Schmidt & Lee 1999). The first motor response to outside perturbation is a spinally mediated reflex (Nashner 2001). The role of this "stretch reflex" is to regain postural stability (Rothwell 1994) by a rapid muscle response. Any movement threatening the body balance is detected by an afferent input via muscle and tendon proprioception, which initiates the first muscle action by contracting selected muscles all over the body. The reflexes do not contribute directly to the recovery of balance (Nashner 2001). The first response against falling is an automatic reaction, seen in EMG, that occurs as medium-latency muscle responses. These reactions are coordinated and conveyed through vestibulospinal reflexes and affect all muscles of the legs, trunk and neck (Allum & Keshner 1986, Nashner 2001). In addition to the medium-latency responses, long-latency responses have been found to co-occur with them in the antagonist muscle (Diener & Dichgans 1986). Automatic responses can be thought of as overlearned, "long-loop" reflexes that rapidly respond by resisting disturbances (Nashner & McCollum 1985, Diener & Dichgans 1986). Automatic reactions are context-dependent and adaptable to the specific balance demands. For example, coordination patterns can be changed, depending on the reliability of the support surface and recent experience (Nashner 2001).

Table 1. Properties of the three motor systems in balance movement control (Adapted from Nashner 2001).

System property	Motor System		
	Reflex	Automatic	Voluntary
Pathways	Spinal	Brainstem/ subcortical	Cortical
Activation	External stimulus	External stimulus	External stimulus Self-generator
Response	Local to point of stimulus and stereotyped	Coordinated and stereotyped	Unlimited variety
Role in balance	Muscle force regulation	Resist disturbances	Purposeful movements
Latency (in legs)	Fixed 35-45 ms	Fixed, medium-latency (mean 95 ms) or long-latency (mean 120 ms)	Variable 150+ ms

In contrast to reflexive and automatic reactions, voluntary movements are based on our conscious attention and may vary (Nashner 2001). Voluntary postural adjustments displace the position of the centre of gravity. For example, abduction of the left arm causes the centre of gravity to shift towards the left. In self-paced movements, both postural adjustments and voluntary limb movements appear to be parts of the same motor program (Lee et al 1987, Zattara & Bouisset 1988). The main idea of postural adjustments has been illustrated in Figure 1 (Gahery and Massion 1981, Rothwell 1994). In this illustration, the control system is defective. However, this description gives good specifications about the feed-forward and feedback postural adjustments produced by voluntary movements.

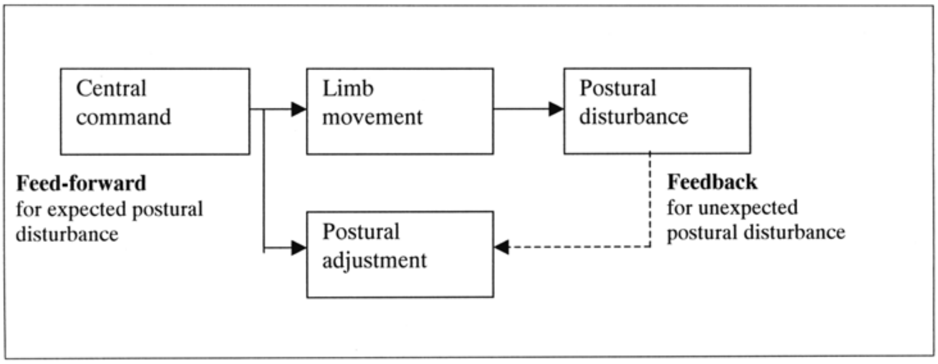


Fig. 1. Feed-forward and feedback postural adjustment (Adapted from Gahery and Massion 1981 by Rothwell 1994).

Three preferred movement strategy models have been used by healthy adults to control their postural stability. In the ankle strategy, the body can be regarded as a stiff pendulum, and balance adjustments are mainly made in the ankle joint, with the person swaying like an inverted pendulum (Nashner 1985). In the hip strategy, the resulting motion is primarily focused about the hip joints (Horak & Nashner 1986). The third way to achieve a balanced standing position in more difficult conditions is to take steps (Horak & Nashner 1986, Carr & Shepherd 1987). It has been proved that subjects can synthesise different postural movements by combining strategies of different magnitudes and temporal relations that are influenced by the subject's recent experience (Horak & Nashner 1986). Thereby, it is also possible for subjects to adjust their postural program during measurements (Woollacott *et al.* 1988).

2.3 Postural control systems

Corrective movements are needed to keep the centre of gravity within the base of support. To achieve this goal, co-ordination of the sensory, skeletal muscle and central nervous systems is needed. The parts of the postural control system are presented in Table 2.

Table 2. Postural control systems (Adapted from Era 1997).

Sensory system	Skeletal muscle system	CNS
Vestibular system located in the inner ear (semicircular canals, otholiths, maculaes)	Muscles of the upper and lower extremities	Stretch reflex
Vision (retina)	Trunk muscles	Long-loop reflexes
Proprioceptive system (muscle spindle-type I and II, Golgi tendon organ, joint receptors)	Neck muscles	Preprogrammed reactions (Learned skills) Synergistic action
Cutaneous receptors		

2.3.1 Sensory systems

The basic idea of sensory systems is to provide information to the system concerning its own state and that of its surroundings. The information is transferred from the sensory receptors to the CNS via afferent pathways. Sensory receptors convert energy of various forms, such as light, pressure, temperature, and sound (Enoka 1994). The main types of sensory receptors needed in postural control are presented in Table 2.

2.3.1.1 Visual system

Visual information is delivered from the retina to at least two different locations in the brain, and these pathways of information have been assumed to be specialised for different purposes; the focal system for object identification and the ambient system for movement control (Trevarthen 1968, Schmidt 1991). The latter has also been shown to affect strongly both stability and balance (Lee & Aronson 1974).

Vision is important for postural control, but it can be compensated for by other information sources (Brandt *et al.* 1986). Vision seems to influence balance by reacting to motion as a relative image shift on the retina (Brandt *et al.* 1986), and it also triggers the muscle activation required for postural corrections. The efficiency of vision in postural control depends on visual acuity (Paulus *et al.* 1984), visual contrast (Leibowitz *et al.* 1979), object distances (Brandt *et al.* 1986) and room illumination. The visual system control of balance is best when the visual distance is less than 2 m (Brandt *et al.* 1986). It has been reported that when the horizon is manipulated so that the vestibular and visual cues are mutually contradictory, elderly persons place more reliance on their visual cues than younger people (Pyykkö *et al.* 1988)

2.3.1.2 Vestibular system

The semicircular canals respond sensitively to velocity changes of movement at frequencies from 0.2 to 10 Hz, and they have hence been found to be active at the beginning and end of movement, whereas the otholiths operate at low frequencies of less than 5 Hz and provide information of linear acceleration, e.g. gravity (Markham 1987, Toppila & Pyykkö 2000). The information from the otholiths and semicircular canals is conveyed to the vestibular nuclei in the brainstem, which also receives information from other sensory sources. The vestibulo-ocular reflex stabilizes vision by producing the eye movements into opposite direction upon turning of the head (Baloh *et al.* 1993), and the main goal of the vestibulo-spinal reflex is to stabilize the head and the body. Although it is known that the vestibular system may contribute to the perception of body orientation and thereby be involved in postural control, some studies have shown that the vestibular system does not play an important role in the perception of sway during normal quiet stance (Fitzpatrick & McCloskey 1994).

2.3.1.3 Proprioceptive and exteroceptive systems

The somatosensory system provides information related to body position by proprioceptors and exteroceptive receptors. The proprioceptive receptors are located in muscles, tendons and joints (Enbom 1990, Jäntti 1993), and they give information about the position of the limbs and the body and the distension of the respective muscles. Proprioceptors include muscle spindles (type Ia and II), Golgi tendon organs (Ib) and joint receptors (McComas 1996). Exteroceptive information is derived from different type of pressoreceptors on the foot sole. Exteroceptive receptors are located in the cutaneous and subcutaneous tissue (Johansson & Vallbo 1980). The major types of cutaneous receptors are Meissner corpuscles and Merkel disks, located closest to the skin surface, and Ruffini ending and Pacinian corpuscles, located deeper in the skin (Latash 1998)

While the receptors in joint capsules give information about the movements and positions of the body parts relative to each other, their role in postural control has not been fully defined yet. The muscle spindles give information about the changes in muscle length and tension (dynamic stretch), and they can also be activated by passively stretching the entire muscle. In addition to an afferent system, the intrafusal fibers in the muscle spindles also receive an efferent input via γ-motoneuron (Enoka 1994). The pressoreceptors detect the body sway, whereas the mechanoreceptors can determine both the site and velocity of an indentation of the skin, as well as acceleration and pressure changes (Johansson & Vallbo 1980, Magnusson *et al.* 1990).

There are some essential inputs for postural control during stance produced by proprioception. First, the information from ankle joints should be recognized, as it is effected by the movement of the centre of gravity, resulting in torque around the ankle joint. Second, the information from the neck muscles gives important references concerning head movement in relation to the trunk. And third, it has been suggested that the eye muscles reflect the eye position in relation to the head (Spirduso 1995).

2.3.2 Skeletal and muscle system

Although the calf musculature is activated first to provide postural control during body movements (Nashner 1983), the co-activation of certain "prime postural muscles", such as the neck muscles, the hamstring musculature, the soleus and supraspinalis muscles, occurs in this order (Nashner 1983, Johansson & Magnusson 1991). Apart from these, however, several muscles participate in producing both reflective movements with different latency times (Nashner 1983) and voluntary movements to balance the body position. Whenever the muscles are stretched, the proprioceptive receptors within the muscle and tendon signal the change in muscle length to the central mechanism of the postural control system (Prochazka & Wand 1980, Spirduso 1995).

Postural control requires coordinated muscle action (Johansson & Magnusson 1991), for example, to produce adequate muscular contractions (Era *et al.* 1996). As the muscles act about the joints in balancing the body, especially the role of the ankle, knee and hip joints is essential (e.g. movement strategies page 14). According to the passive stiffness control model, ankle stiffness, as a result of the CNS being limited to the selection of appropriate muscle tonus, stabilizes the unstable mechanical system in quiet stance (Winter *et al.* 1998). However, other researchers have pointed out the active mechanism of postural stabilization in balanced stance (Johansson *et al.* 1988, Morasso & Schieppati 1999), where the muscle and foot skin receptors play an essential role (Morasso & Schieppati 1999).

2.3.3 Central nervous system

Several parts of the central nervous system (CNS), which consists of the spinal cord and the brain, take part in controlling posture. Input signals to the cortical neurons come mainly from the thalamic nuclei, which transmit information from the spinal cord, basal ganglia and cerebellum and from the parietal and frontal areas of the cortex. The first and fastest response to a change in stance is triggered by spinal reflexes (e.g. Allum & Keshner 1986). The excitation within the CNS is mediated through the synapses consisting of afferent fibers on neurons, whereas the inhibitory synapses use special mediators, interneurons. For example, reciprocal Ia inhibition is evoked by the low-threshold afferents in antagonist muscles (e.g. Tanaka 1983), and recurrent inhibition of motoneurons is meditated by interneurons called Renshaw cells (e.g. Pierrot-Deseilligny *et al.* 1983).

The voluntary movements needed for balancing posture are planned within the brain. The output commands are sent to the muscles via the pyramidal and extrapyramidal systems. The pyramidal cells, with their connections to the premotor and parietal cortex, transmit information to the spinal motoneurons and interneurons, which control voluntary movements and the segmental reflexes needed for balancing posture (Jäntti 1993) The output of the cortical motor areas also includes projections to the basal ganglia, cerebellum and red nucleus. The basal ganglia, which constitute the major component of the extrapyramidal system, consist of the substantia nigra and subthalamic nuclei. They are connected to three nuclear groups (caudate nucleus, putamen and globus pallidus).

The basal ganglia and the nuclear groups take part in the facilitation and planning of both voluntary and reflex movement during postural control. The cerebellum and its connections are responsible for the co-ordination and smoothing of the reflex movements and the regulation of voluntary movement.

2.3.4 Integration of the different components of the postural control system

To ensure proper postural control, the sensory influx must be integrated in the CNS to provide adequate motor output. The sensory information from the visual, vestibular and proprioceptive and exteroceptive systems is used as input. However, it is suggested that only one of the three main sources of afferent input is necessary to ensure balance in non-complicated circumstances (Rothwell 1994). Forget and Lamarre (1990) demonstrated that even in the absence of peripheral feedback, postural adjustments may be adequate.

At the spinal cord, afferent impulses trigger stretch reflexes, whereas at the higher levels in the CNS, neural connections mediate more complicated motor responses. On the effector side, one important precondition for balancing is the ability to choose appropriate responses, to modify these responses on the basis of sensory input and, finally, to produce the needed muscular contractions to maintain posture (Era *et al.* 1996). The context-dependent responses that utilise all available sensory input and lead to preprogrammed motor responses are based on past experience.

Although it is impossible to differentiate reflective postural adjustments from voluntary movements in any reliable way, the reflexes can be differentiated by the time series (Allum & Keshner 1986). The role of reflex movements is to regulate muscle force, whereas the stretch reflex-based movements (automatic reactions) resist possible disturbances. Purposeful voluntary movements affect balance directly or indirectly, for example, when we move our COG with the intent of rising from a chair or opening a door (Nashner 2001).

2.4 Balance assessment

Many different balance tests (Mathias *et al.* 1986, Berg *et al.* 1989) and measurements (Juntunen *et al.* 1987, Briggs *et al.* 1989, Wing *et al.* 1995, Era *et al.* 1996, Chang & Krebs 1999) have been developed and presented to obtain appropriate information of balance capabilities during standing. The selection of a suitable method generally depends on the goals and results aimed at. There is no single assessment technique that could be used as a true indicator of the overall integrity of the balance control system (Winter *et al.* 1990). For example, functional balance scales are easy to perform and suitable for daily clinical use, but not always accurate enough. Therefore, laboratory systems with new technologies may give more detailed information about postural balance (Alexander 1996).

Balanced stance can be achieved by coordinated movements of body segments, taking advantage of the interaction of internal and external forces, and accomplished through the action of the neuro-musculo-skeletal system (Medved 2001). This is the reason for the three distinct subsets of physical variables to be viewed when measuring standing balance in laboratory circumstances: 1. kinematics is concerned with details of the movement itself rather than forces (e.g. motion analysis) (Winter 1990), 2. kinetic data (e.g. platform measurements) present the forces and the moments of forces that are developed during movements, and 3. bioelectric changes are associated with skeletal muscle activity (e.g. electromyography) (Medved 2001).

These modern methods can be used separately or together (Gatev *et al.* 1999) in balance measurements, depending on the aim of the study. Enough attention should be paid to the time synchronization of the data (Winter 1990). Besides, in all laboratory measurements, the selection of measuring time and stance conditions is essential. Typically in platform measurements, for example, the most frequently used duration is from 20 to 30 seconds. But longer (e.g. Scott & Dzendolet 1972) and shorter (Era & Heikkinen 1985) measuring times have also been used. The measuring time should be long enough to provide a relevant result, but short enough to avoid fatigue with the measurements. For example, Iverson *et al.* (1990) found a clear decrement in balance times in the Romberg test among subjects aged 60 to 90 years.

Several variations in the technique of measuring posture during stance have been used, such as the width of the foot position, e.g. standing with the feet together, touching the heel to the toe (tandem stance) or standing on one leg (e.g. Konradsen *et al.* 1993, Harrison *et al.* 1994). Standing on two legs is naturally the easiest position.

2.4.1 Motion analysis measurements

Fischer (1861-1917) and Braune (1830-1892) were the first investigators to perform an analysis of the three-dimensional (3D) movements of the segments of the human body (1898) by using four cameras, and the work of Bernstain (1896-1966) in developing 3D analysis was also of great importance. Since those days, the measurement systems and the possibilities for movement analysis have developed quickly, being characterised by an ever greater influence of new technologies. The systems could be classified, for example, into ones with active or passive cameras or active or passive markers. Several optoelectric methods for measuring the kinematics of human movement are currently in use (Furnee 1991, Medved 2001).

In balance studies, optical systems have been utilised, first, to track the position of the body segments (Benvenuti et al 1999) and, then, to calculate the COG position. This can be done by measuring the positions of the light-emitting markers. Then, by knowing the positions of the centres of the body segments, the COG value can be calculated. This method requires modelling of the body, and although the control of balance in human upright standing is particularly well suited for modelling (Kuo 1995), it is also a popular experimental paradigm (Nashner & McCollum 1985). However, any given analysis is able to use only a small fraction of the available kinematic variables (Winter 1990), and the model should correlate with the natural system anatomically and physiologically

(Johansson & Magnusson 1991). For example, a complex biomechanical model of the whole body consisting of 17 segments (Hatze 1980) was designed anthropomorphically and linked by joint connections of various complexity. Another attempt at modelling was the precise representation of an anatomical model for surgical purposes (Delp *et al.* 1990). There are limitations that should be considered when using these models based on the measurements of body movements, such as the positioning of the markers relative to the joints, the adequate number of markers and the quality and quantity of the anthropometcic data (Cappello *et al.* 1995). Besides, the calculation of even a simple mechanical system is quite demanding (Medved 2001).

Another possibility is to record the movements of body segments and their directions on all cardinal axes with optoelectric imaging systems. Motion occurs in space and time, and typically, human movement analysis is accomplished by the determination of position, velocity and acceleration (Enoka 1994). The distance of movement can vary in magnitude and in direction. Average velocity is described as the rate of change in position with respect to time;

$$v_a = \frac{\text{change in position}}{\text{change in time}} = \frac{\Delta d}{\Delta t}$$

(Lerner 1996)

Instantaneous velocity at a certain point of time is calculated as follows:

$$\text{Instantaneous velocity} = v(\Delta t) = \lim_{\Delta t \to 0} \Delta d / \Delta t$$

(Lerner 1996)

Average acceleration is described as the rate of change in velocity over time;

$$a_a = \frac{\text{change in velocity}}{\text{change in time}} = \frac{\Delta v}{\Delta t}$$

(Lerner 1996)

$$\text{Instantaneous acceleration} = a(\Delta t) = \lim_{\Delta t \to 0} \Delta v / \Delta t$$

(Lerner 1996)

Velocity and acceleration values can be either positive or negative. When the sign of the movement velocity changes, the movement changes its direction. The sign of the movement acceleration shows whether the movement is accelerated or decelerated (Ohanian 1989, Enoka 1994).

Motion analysis has been used to analyse movement parameters during walking (You *et al.* 2000), and it can also be used to assess problems in human locomotion (Selfe 2000). Motion analysis makes it possible to analyse single joint angles, angle velocities and angle accelerations in balance performances (Gauffin *et al.* 1993, Aramaki *et al.* 2001). The sensitivity and precision of the motion analysis system has been shown to be adequate for studying body movements (Levy & Smith 1995), but some studies have suggested the values to be device-specific (Mannion & Troke 1999). The accuracy of marker centre estimation can be increased by a suitable calculation system (Jóbbagy *et al.* 1995) and, for example, the most frequently used method, geometric centroid calculation (Taylor *et al.* 1982), has been reported to produce moderate accuracy. Accuracy can also be increased by increasing the resolution of digitisation while converting the analogue video output signal of a camera (Jóbbagy *et al.* 1995), and the technological advances in digital filtering provided a more promising solution to noise reduction (Winter 1990). It has been reported that, under carefully controlled conditions, 3 D motion measurement can produce an acceptable accuracy of measurements for clinical purposes (Wilson *et al.* 1999, Paul 1995). The target points should be selected according to the experimental requirements. For example, the reflective markers (at least two) should be oriented towards a sufficient number of available cameras. Besides, mounting the markers on the subject should be a fast and easy to perform (Cappello *et al.* 1995).

2.4.2 Body sway measurements

The body movements performed by humans are usually called "body sway". Typically, the term "body sway" is used to describe the extent of the centre point of pressure (COP) or the centre of gravity (COG) excursions. It is possible to measure body sway with a simple technology, for example, by using a "swaymeter" (Lord *et al.* 1991a) or Wright's ataxiameter (Overstall *et al.* 1977, Brocklehurst *et al.* 1982, Nayak 1987). The swaymeter measures displacements of the body at the waist level, whereas the ataxiameter can be used to define sway as an angular movement of the body around the ankle joint. Also, an inclinometry-based method has been developed to provide information about body sway (Viitasalo *et al.* 2002).

One of the most popular computerised laboratory systems for evaluating human postural stability is to measure spontaneous postural sway with the subject standing on a force platform (Era *et al.* 1996, Kinney LaPier *et al.* 1997) or the subject's response to an applied postural perturbation (Era & Heikkinen 1985). Respectively, these measurements can be defined as static and dynamic posturography. The basic principle of the force platform test is to measure the movements of the COP that reflect both the horizontal location of the COG and the reaction forces due to muscular activity (Era *et al.* 1996). The aim of data processing is to compute selected parameters of total body sway from the time series of COP positions. Typical parameters in platform measurements are the mean COP position (as a reference point), anterior-posterior and lateral sway (Kinney LaPier 1997), the length of the sway path (Juntunen *et al.* 1987) as well as sway velocity (Figura *et al.* 1991, Hytönen *et al.* 1993) and sway area (Ekhdahl *et al.* 1989).

Body sway can be measured with a platform under variable visual and surface conditions, and measures of postural sway have been reported rather to capture sensorimotor deficits than to differentiate between functional performance abilities (Hughes *et al.* 1996). Typically, sway is assessed during stance on a stable platform both with the eyes open and with the eyes closed (e.g. Baloh *et al.* 1994, Baloh *et al.* 1998). Less commonly, sway has been analysed during standing on a foam plastic covered surface (Hytönen *et al.* 1993). A similar sway-referenced condition can be caused in dynamic posturography. Postural stability can also be measured with a platform using visual control (Juntunen *et al.* 1987) or in other visual conditions, such as blurred vision (Geurts *et al.* 1993) or peripheral or central vision (Brandt *et al.* 1985).

Computerized dynamic posturography has been reported to detect the causes underlying functional limitations by quantifying impairments in the sensory input and the automatic motor response systems (Nashner 2001). In the latest dynamic posturography studies, the sensory organization test (SOT) has been used for these purposes (Stewart *et al.* 1999, Gianoli *et al.* 2000). The complete test consists of all combinations of fixed (normal), eyes closed, and sway-referenced visual and support surface conditions (Nashner 2001).

The reliability of platform measurements has been found to be rather high, as in testing healthy subjects' (n= 24) balance velocity and sway (Ishizaki *et al.* 1991), or moderate (intraclass correlation coefficients of reliability (ICC) values from 0.58 to 0.92), as in measuring hemiplegic patients' (n=20) total balance sway (COP in the mediolateral and anteroposterior directions) (Levine *et al.* 1996). However, Goldie *et al.* (1989) reported about the poor reliability of platform measurements (n= 28) and Hill *et al.* (1995) about the low reliability coefficients of COP measurements between test occasions (n= 17). Benvenuti *et al.* (1999) used 3 D motion analysis and a force platform to evaluate the quiet standing of aged subjects with balance problems (n= 36). ICC values indicated a high level of retest reliability in measurements of body sway, joint alignment and body position. The explanation of the different reliability values could lie in differences in the methods and populations used.

2.4.3 Other balance assessment methods

There are varying possibilities to assess balance during stance, and it is impossible to present them completely in this context. Apart from the body movement and body sway measurements, the main attention has been focused on EMG measurements and functional balance measurements.

2.4.3.1 Electromyographic measurements

Electromyography assesses the electrical signal associated with muscle contraction, and the outcome of the measurement is called an electromyogram (EMG) (Winter 1990). EMG is utilised to analyse the electrical signals produced by the activation patterns of the muscles. In detail, this activation process includes the generation and propagation of action potentials in nerve and muscle cells. EMG measurements are often used together with other methods, such as platforms, to gather information about motor programmes to balance posture (Macpherson *et al.* 1989). The advantage of this method is the possibility to identify the muscles or muscle groups involved in the movements.

It has been reported that, in stance, the trunk and leg muscles are activated in a similar sequence during all types of rapid arm movements, but even more variable postural adjustments have been found during slower arm movements (Horak *et al.* 1984). In experimental conditions, the subjects have typically been exposed to translations of the support surface (Diener *et al.* 1988, Macpherson *et al.* 1989), after which EMG activities

(Macpherson *et al.* 1989) or EMG latencies (Diener *et al.* 1988) have been studied. The EMG results have shown that the postural response patterns may change with the conditions of support (Cordo & Nashner 1982, Macpherson *et al.* 1989), whereas the latency times apparently give some basic information about the involvement of the CNS in postural control (Diener *et al.* 1988).

2.4.3.2 Functional balance measurements

In clinical use, functional balance scores, such as "Get up and go" (Mathias *et al.* 1986), Berg's Balance Scale (Berg *et al.* 1989, Woollacott & Shumway-Cook 1996), "Functional Reach" (Duncan *et al.* 1990) and other performance-oriented scales (e.g. Tinetti 1986), have been used in assessing balance. These tests have been designed to assess balance capabilities in various tasks related to everyday life. The advantage of functional scales is that they are focused on receiving information of typical balance problems. For example, in the functional reach test, the ability to move the COG towards the boundary of stability could be assessed, whereas the Get-up and Go test reflects postural stability in situations where elderly people often fall. A specific test of dynamic standing, called the Step Test, has also been developed to evaluate dynamic single-limb stance (Hill *et al.* 1996).

The limitation of functional tests is that the assessment is usually based on scores (e.g. from 1 to 5 in the Get-up and Go) that only give a single value to describe the balance abilities of a subject. Thus, these tests do not provide detailed information of impairments in the postural control system or specific individual needs in rehabilitation therapies (Horak 1997). However, tests are suggested to be utilised in follow-up during rehabilitation courses (Bohannon & Leary 1995). The reproducibility of the scores and the internal consistency of the scale have found to be good (Berg. *et al.* 1989, Hill *et al.* 1996), and the correlation between force platform measurements and functional balance tests has turned out to be significant, especially among healthy subjects with their eyes open (Ekdahl *et al.* 1989). It has been reported that performance-orientated assessments can predict falls among the elderly (Berg *et al.* 1992, DiFabio & Seay 1997).

2.5 Effects of anthropometric factors on postural stability

Many of the earlier studies concerning body and limb measurements ignored biomechanics, but based on more recent knowledge, body characteristics affect postural stability. A major impetus for anthropometric measurements has come from the needs for technological development (Winter 1990). The most basic body dimensions are the distances between joints (Winter 1990), and an average set of distances expressed as proportions of body height gives a good approximation of a body model. Anthropometric factors should be considered in biomechanical modelling of the body (Winter 1990), in planning a measurement and in assessing the results of measurements. Data based on some new balance measurement systems can be normalized and related according to body height (e.g. Kinney LaPier *et al.* 1997).

The biomechanical model can be used in, for example, segmental analysis (Benvenuti *et al.* 1999), but the information of body characteristics may also directly affect the measurement values. In the inverted pendulum model (Nashner 1985), a longer lever arm, e.g. longer height, would cause a greater amplitude of movement than a shorter height. Besides, the support surface size (foot size) is related to height (e.g. men–female). Body characteristics should, therefore, also be considered in measurement settings, e.g. as far as marker placement in measurements with motion analysis is concerned. The differences in body characteristics have been assumed to influence the boundaries of individual postural stability, and this variability may affect the selection of motor strategies to maintain postural balance control (Woollacott & Shumway-Cook 1990).

Short body height and knee height were shown to be reasons for an increased risk for falls among older Japanese women (n=705), but weight and BMI were not found to be associated with falls. The examination procedures included the “Get and Go” test, a full tandem balance test and measurements of walking speed, but no laboratory balance measurements (Davis *et al.* 1999). Small body mass was found to associate with a poorer posture control in a large cross- national comparative study carried out on the force platform (Era *et al.* 1996). Differences in balance values were analysed between elderly subjects living in different geographical areas.

The reports on the effect of leg length on postural stability are conflicting; in a study on 45 subjects, the results indicated no differences in postural sway values between the subjects with and without the leg length discrepancy (Murrell *et al.* 1991), whereas another study (n=14) revealed a significant increase in mediolateral sway among the subjects with a small (1cm) leg length discrepancy (Mahar & Kirby 1985). The role of foot anthropometry and foot problems, such as foot deformities, contribute to functional impairment and, hence, to postural instability (Menz 1998, Menz & Lord 1999b). Yet, there were no studies dealing with the association between normal foot anthropometry and postural stability. Conversely, there seemed to be more studies concerning foot problems affecting postural instability (e.g. Lord *et al.* 1991b, Cavanagh *et al.* 1992). The prevalence of foot problems tends to increase in older age groups (White & Mulley 1989), which may be due to the cumulative effect of chronic systemic diseases that affect the anatomic structures of the foot. The effects of inappropriate footwear, e.g. high heels and a narrow toe box (e.g. Frey *et al.* 1993), on postural stability should also be noted (Gabell *et al.* 1985, Herman & Bottomly 1992, Menz & Lord 1999a). For practical measurements, a footwear assessment form was presented to enable determination of the contribution of footwear characteristics to instability (Menz & Sherrington 2000).

2.6 Effects of gender and age on postural stability

2.6.1 Gender relations

The different body heights of men and women have been assumed to contribute to the poorer postural stability of men compared to women (Kinney LaPier *et al.* 1997), and it is possible that the balance differences between men and women are mainly due to their different anthropometrics. Typically, the postural control of both genders has been assessed with platforms (e.g. Era *et al.* 1996), but there are also motion analysis results showing differences in movement strategies between men and women (Yoshida *et al.* 1983).

The presence of gender differences in sway amplitudes or velocities differs between studies (Ekhdahl *et al.* 1989, Suomi & Kojeca 1994). Ekhdahl *et al.* (1989) studied the balance performance of 78 women and 74 men with some traditional functional balance tests and a force platform and found women to be more stable than men. This finding was also supported by Ojala *et al.* (1989) and Juntunen *et al.* (1987). In the latter study, however, the balance of more aged adults was not measured. Overstall *et al.* (1977), however, found in their study women to be less stable than men. Further, there are many studies that failed to find significant relationships with gender (Black *et al.* 1982, Brocklehurst *et al.* 1982, Kinney La Pier *et al.* 1997). Any comparison of results concerning the effects of gender should be made with caution, as methods and populations vary. For example, in the study by Black *et al.* (1982), postural stability was measured in adults aged 20-49, whereas in the study by Brocklehurst *et al.* (1982) covered a sample of elderly people.

2.6.2 Age relations

2.6.2.1 Effects of age on the postural control systems

The physiology of systems for postural control declines with age. Visual, vestibular and somatosensory functions have been reported to diminish with normal ageing (Woollacott *et al.* 1986, Wolfson *et al.* 1986, Teasdale *et al.* 1991). The impaired balance ability can often be traced to a decreased sensory input, slowing down of motor responses and weakness of support (Baloh *et al.* 1994). Typically, diseases, medications and musculoskeletal limitations also contribute to balance problems (Winter 1995, Tinetti *et al.* 1988). Besides, the slower rate of central processing (Stelmach & Worringham 1985, Woollacott *et al.* 1986, Teasdale *et al.* 1991) and the integrated sensory inputs within the central nervous system (Manchester *et al.* 1989, Stelmach *et al.* 1989) have been reported as reasons for difficulties in the coordination of reflexes and voluntary movements in postural control at older age. Both induced postural reactions (Quoniam *et al.* 1995) and perceptual responses (Hay 1995) seem to decline with age. The slower anticipatory postural responses of aged people may also contribute to the onset of voluntary responses (Man'kovskii *et al.* 1980, Woollacott & Manchester 1993), and differences have been found between ages in postural responses during rapid self-paced and reaction time arm movements (Rogers *et al.* 1992).

In conflicting visual and somatosensory conditions, elderly subjects show impaired balance and follow visual cues (Woollacott *et al.* 1986, Baloh *et al.* 1993, Hay *et al.* 1996). Loss of visual acuity, depth perception, peripheral vision and contrast sensitivity seem to occur with ageing (Spirduso 1995). Proprioceptive loss increases the threshold to movement detection and decreases the accuracy of reproducing or matching joint angles (Stelmach & Sirica 1986, Simoneau *et al.* 1992, Ferrell *et al.* 1992), and an impaired sense of vibration in the legs has been shown to increase postural sway (Brocklehurst *et al.* 1982). The findings of vestibular function over age are contradictory (e.g. DiZio & Lackner 1990, Peterka *et al.* 1990a, Peterka *et al.* 1990b, Paige 1991), and some studies have shown older subjects to be even more sensitive to vestibular stimuli than young ones. People with severely reduced vestibular function are not necessarily aware of a vestibular deficit when falling (Black & Nashner 1985).

It has been suggested that the declining ability to provide information about the mechanical displacement of muscles and joints by muscle proprioceptors (e.g. Stelmach *et al.* 1989), the loss of muscle strength (Whipple *et al.* 1987) and the more dominant proportion of slow-twitch fibres among the elderly lead to slower force production and, thus, to poorer postural control (Studenski *et al.* 1991, Kuo & Zajac 1993), which may be predispose the person to falling (Aniasson *et al.* 1984, Lord *et al.* 1991a). The tibialis anterior muscle and the muscles working synergistic with it that produce backward body sway seem to contract somewhat more slowly in older than younger subjects (Spirduso 1995). The postural responses of foot flexors were found to be slower than those of foot extensors in aged subjects (n=15 in aged group, n=15 in younger group) (Inglin and Woollacott 1988). This may result in old people bending their knees more than young ones to balance their body. On the other hand, many experts believe that the fear of falling contributes to the "stiffening" of the body, which produces inappropriate balancing movements of body segments (Baloh *et al.* 1994, Spirduso 1995).

2.6.2.2 Associations between age and balance

An analysis of postural control kinematics (Alexander *et al.* 1992) in young (n=24) and elderly (n=15) adults during perturbation showed small but consistent differences between the groups, particularly in more challenging conditions, whereas the kinematic analysis of quiet standing (n=24) in adult subjects produced no evidence of postural instability (Panzer *et al.* 1995). Kinematics was also used to assess head movement in the elderly during base translations (Wu 2001), and this head movement was found to be significantly increased in the elderly group (n=10). This supports the results of earlier balance studies during voluntary activities, in which head rotation in the sagittal plane was shown to be significantly larger in elderly than younger adults (Hirasaki *et al.* 1993, Di Fabio & Emasithi 1997).

Different results concerning the associations between balance and age with a platform have been reported (Era & Heikkinen 1985, Ekhdahl *et al.* 1989, Ojala *et al.* 1989, Hytönen 1993), and most of them indicate that older people are less stable than younger people (Brocklehurst *et al.* 1982, Era & Heikkinen 1985, Ekhdahl *et al.* 1989) because of their larger amplitude and frequency of postural sway during stance. Some studies that have failed to show any significant age differences in standing balance (Black *et al.* 1982, Juntunen *et al.* 1987). Other balance aspects, such as sway velocity, have also been shown to be changed among older subjects (Hytönen *et al.* 1993). The notable large increase of sway velocity in standing with the eyes closed has been taken to indicate the importance of vision in balance control among the elderly (Hytönen *et al.* 1993), whereas visual cues are not so effectively utilised in standing at a young age (Pyykkö *et al.* 1988).

The differences between the study results can be caused of many factors, e.g. different numbers or ages of subjects and differences in measurement settings. For example, in the study by Era & Heikkinen (1985), the study sample (n=318) consisted of men of different ages (age groups: 31-35 yr. 51-55 yr. and 71-75 yr.), whereas Ekhdahl *et al.* (1989) measured male and female subjects aged 20–64. In the study of Brocklehurst *et al.* (n= 151) (1982), the reference sample consisted of only five young subjects, while Black *et al.* (1982) did not measure the postural balance of the most aged people. Besides, it should be pointed out that the real effects of age on balance need to be studied longitudinally. Although the influence of age on balance measurements during stance is controversial, the results show that elderly people have difficulties to stand on one leg (Potvin *et al.* 1980, Era & Heikkinen 1985), and it is suggested that a one-leg balance performance with the eyes open may be an age-sensitive way to measure balance capabilities (Stones & Kozma 1987).

Despite the many age-related changes in the postural control system that may cause an increased risk for falls (Wootton *et al.* 1982, Ryynänen 1993), balance performances can be improved with exercise. Muscle strength can be improved (Era 1988, Sauvage *et al.* 1992, Lord *et al.* 1993), but its effects on balance performance are limited. Specific balance exercises (Woollacott *et al.* 1993, Magnusson *et al.* 1996), such as *tai chi* movements, walking and coordination exercises, dancing, etc., have been found to improve balance and posture. For example, the ability to stand on a one leg (Johansson & Jarnlo 1991, Judge *et al.* 1993) has been improved. Balance training by using sensory input manipulation has improved the balance of older adults, as shown by the measurements made with force platform (Hu & Woollacott 1994a), kinematics and EMG (Hu & Woollacott 1995b), and it is suggested that the use of sensory information for postural control might also be assessed for purposes of therapy (Shumway-Cook & Horak 1986). Facilitation of the proprioceptive system may also affect the outcome in postural stability (Maki *et al.* 1999). Studies of habitual (Shephard *et al.* 1993) and vestibular rehabilitation therapies have further given good results on the improvement of the general physical condition, fitness and balance (Shephard & Telian 1995, Black *et al.* 2000) among patients with vestibular deficits. The encouraging and motivational effect of balance training cannot be ignored (Era *et al.* 1991).

3 Purpose of the study

Information on standing balance is commonly based on performance in functional tests or measurements of body sway or shifts of COG/ COP. Little information is available on the movements of separate body segments in postural control. This information could be useful in developing balance evaluation and balance training programmes in physiotherapy.

The purpose of this study was to estimate motion analysis and to test its reliability and validity for studying the movements of separate body segments during stance. In addition, the association between age, gender and anthropometric factors with the movements of separate body segments during stance in a healthy population were studied.

The detailed aims of this study were:

1. To evaluate body movements with motion analysis during stance on two legs and on one leg (I).
2. To evaluate the test-retest repeatability of motion analysis and the validity of measurements for analysing human stance (II).
3. To study the associations between age, gender and anthropometric factors with body movements during stance using motion analysis (III-IV).
4. To study the velocities and accelerations of separate body segments during stance with the eyes open and closed (V).

4 Subjects and methods

4.1 Subjects

Study I: The participants in study I were 10 healthy (6 females and 4 males) volunteers (physiotherapists) at the Oulu University Hospital. The subjects had no history of problems of postural instability, and the main criterion for inclusion was normal balance in performing the balance tests. The characteristics of the subjects are presented in Table 3.

Study II: The repeatability evaluation was made in a group of 10 healthy subjects (2 men and 8 women), who were volunteers aged 30 to 43 years, and the validity evaluation consisted of 16 healthy female volunteers working at the Oulu University Hospital. The characteristics of the subjects are presented in Table 3. The study subjects had no history of neurological diseases or balance problems and were able to perform the standing balance tests.

Studies III-V: An age- and gender-specified sample consisting of healthy subjects were randomly selected from the local population of Oulu (on Jan 1st 1998: 113 567 inhabitants). 156 letters were sent in the order of the names on the lists to recruit 100 healthy volunteers of appropriate age and gender (the response rate was 64.1%). The final sample comprised 50 women and 50 men aged 31 to 80 years categorized by gender and age decade into 10 groups as follows: women aged 31-40 yr., men aged 31-40 yr., women aged 41-50 yr., men aged 41-50 yr., women aged 51-60 yr., men aged 51-60 yr., women aged 61-70 yr., men aged 61-70 yr., women aged 71-80 yr., men aged 71-80 yr. Each group consisted of ten subjects.

The subjects were of various socioeconomic and educational backgrounds, and the main reasons for exclusions were an incapacity to stand for 30 seconds without any standing aid (n= 18, mostly in the age group of 71-80 yr.) and a failure to come for the appointment without any prior information (n=23, mostly in the age group of 31-40 yr.). The main criterion for inclusion was the ability to stand on two legs with the eyes open and the eyes closed for 30 seconds without any standing aid. The study subjects had no history of neurological diseases, previous serious injuries or diseases of the lower extremities or abnormal vision (eyeglasses for better visual acuity were allowed) or standing balance problems. Informed consent was obtained from each subject.

Table 3. Characteristics of subjects in studies I-V.

Characteristic		Study I	Study II		Studies III-V		
			Repeatability	Validity	All	Female	Male
Number of subjects		10	10	16	100	50	50
Gender: female/male		6/4	8/2	16/-	50/50	50/-	-/50
Age (yr)	Mean	33.3	37.4	41.4	55.2	54.6	55.8
	SD	5.9	4.1	7.1	13.8	13.9	13.9
	Range	26-44	30-43	21-52	31-80	31-77	31-80
Height: (cm)	Mean	169.9	166.9	163.1	167.6	161.2	174.0
	SD	7.8	8.9	5.1	9.2	5.9	7.3
Weight: (kg)	Mean	67.9	63.6	62.0	73.9	68.7	79.2
	SD	11.5	7.2	5.2	12.4	11.1	11.4

4.2 Study designs

The research design and the testing protocols were approved by the ethical committee of the Oulu University Medical Faculty. All studies were designed and carried out in line with the principles for human research (Currier 1981) and the Helsinki Agreement (1964, revised 2001).

Study I was a methodological study in which the movements of separate body segments were measured once during 10 seconds. The standardised instructions and explanations of the testing procedure were given to the subjects, and each subject was given an opportunity to practise standing on one leg before the measurements. The measurements of body movement were then analysed.

In study II, the repeatability of body movement measurements during stance with the eyes open was evaluated. The test was made three times: on two consecutive days and one week after the first measurement. The measurements were carried out at the same time of the day, and the final results were then compared. The validity of the motion analysis measurements was evaluated by comparing the balance measurements with a motion analysis system and a platform at the same time. The validity test battery consisted of three tests: standing on two legs with the eyes open and the eyes closed and standing on the right leg with the eyes open. The subjects were given an opportunity to practise standing on one leg. Recording time in each test was 10 seconds, and the results obtained with the two systems were then compared. All the repeatability and validity measurements were done by the same therapist.

In the studies III-V, the movements of separate body segments of all subjects were measured during standing on two legs with the eyes open and with the eyes closed, using a motion analysis system. The subjects were measured once during the day (between 8.00 and 18.00 o'clock) after they had been given the test instructions and opportunities for practising. In addition, the subjects' anthropometric factors were measured. The results of the motion analysis measurements were then compared between the different age and gender groups and between the eyes-open and the eyes-closed conditions (III, V) and with the anthropometrc factors (IV).

4.3 Methods

4.3.1 Balance measurements with a motion analysis system (studies I-V)

The Mac Reflex 3D motion analysis system (Qualisys AB, Partille, Sweden) was used to measure the movements of separate body segments during stance on two legs with the eyes open and closed and on one leg with the eyes open. The system contains two infra-red flashlight cameras (sampling rate 50 Hz) with video processors, a computer and light-reflective markers. Six light-reflective markers were placed on different parts of the subject's body (on the forehead, the navel, and the anterior surfaces of both the knees and the ankles) (Fig. 2). In study I, an additional marker was placed on the chest. The markers were positioned in this way to define the movement parameters at these points during standing.

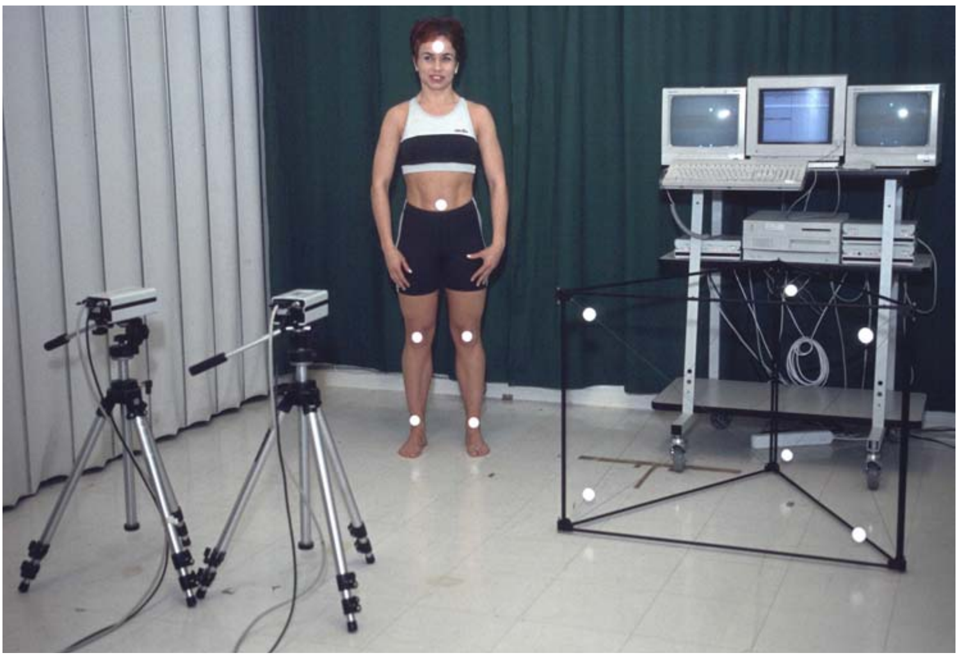


Fig. 2. Mac Reflex motion analysis system. The markers were placed on the subject's forehead, navel, both knees and ankles.

Calibration was made before each subject's measurements. The purpose of calibration was to define a three-dimensional coordinate system and to find the optimal camera orientation for the markers. The cameras sampled data from the subject's frontal plane. The data was fed into a video recorder from the cameras, while the sorting editor performed the sorting and tracking of the video data. The movements of the marked body segments were saved and evaluated with a Macintosh computer and calculated along the three cardinal axes. The WingZ statistical package (Informix Software Inc. London, UK) was used to process the data. The ranges of body movement values (maximal amplitudes) were calculated as the difference between the maximum and minimum movement values during the measurement period (studies I-IV). The maximal velocity and acceleration of the movements of the body segments during stance were computed for movements in all cardinal axes (studies I, V). In addition, the movement magnitudes of separate body segments were calculated and quantified as total movements, including movement excursions along all the three cardinal axes during 10 seconds (studies II-IV). The following formula was used:

$$d = sqrt[(x_{n+1}-x_n)^2+(y_{n+1}-y_n)^2+(z_{n+1}-z_n)^2],$$

where $x_{n,}\ y_{n,}\ z_n$ are the coordinate positions in the lateral (x), anterior-posterior (y) and vertical (z) directions.

The same therapist measured all subjects, and standardised testing instructions were supplied to the subjects. The movements of separate body segments were measured in three different tests as follows:

In standing on two legs with the eyes open (studies I-V). The subjects were standing barefooted with their heels 12 cm apart and their arms down their sides. They were instructed to look forward at a given point on the wall 5.5 m ahead. The standing time was 30 seconds, which included a 20-second stabilization period and a 10-second recording time.

In standing on two legs with the eyes closed (studies II-V). The subjects performed the test as in the eyes-open conditions. During the measurements, an assistant stood behind the subjects to ensure their safety. In addition, the subjects were instructed to open their eyes immediately if they felt dizzy.

In standing on right leg with the eyes open (studies I-II). On right-leg stance, the subjects were asked to raise their left foot off the ground in the direction they liked best. The subjects were not, however, allowed to hold their left leg behind the right leg because of the markers. As with the two-leg standing, the subjects were told to look forward. The measurement time was 10 seconds after the subject had reached the one-leg standing position. The subjects were given an opportunity to practise right-leg standing.

The system error was analysed first by measuring the "noise" of a stable marker on the wall during 10 seconds. As the system is able to process data relative to the desired marker, the error caused by the system was also analysed by measuring the "noise" of a stable marker at the position where the subject was standing with the same protocol as described in Figure 2 during 10 seconds. The stable marker was placed in the field of measurement against a motionless object, 10 cm left of the subject's left foot and three centimetres above the ground. First, the maximal movement amplitudes of the markers in the lateral, anterior-posterior and vertical directions were calculated based on the difference between the maximum and minimum values. Second, the same marker movements were analysed relative to the stable marker (reference marker).

4.3.2 Repeatability and validity measurements (study II)

Repeatability: The repeatability of body movement measurements during standing were made with a motion analysis system. Six markers were placed in different parts of the subjects' body: on the forehead, the navel and the anterior surfaces of both the knees and the ankles (Fig 2). During the measurement, the subjects were instructed to stand as normally as they could, with their arms down their sides, and to look towards a given point on the wall 5.5 m ahead. The recording time was 10 seconds. The test was repeated three times: on two consecutive days and a week after the first measurement. The magnitude of the movements of body segments was calculated as the difference between the maximum and minimum values during the measurement period (maximal amplitude). In addition, the magnitude of the movements of body segments was calculated as total movement values. All the three measurements were carried out at the same time of the day and by the same therapist.

Validity: The validity of the motion analysis measurements was evaluated to compare the body measurements with the Mac Reflex 3D motion analysis system and the balance with the In Good Balance platform (Metitur Ltd, Jyväskylä, Finland) (sampling rate 50 Hz) at the same time. This platform was selected because system support was available in Finland. Besides, the possibility of graphical analysis made it easier to compare the results with the motion analysis results. The initiation of the measurement was synchronised with an interface cable between the computers. The study subjects' total body sway was measured with the triangular In Good Balance platform connected to a computer through a three-channel amplifier and an A/D converter. The force transducers were situated at the corners of the triangular platform, and vertical forces were recorded. Commercial software (Metitur Ltd., Jyväskylä, Finland) was used to process the data.

The three-dimensional (lateral, sagittal, vertical) movements of the head and the hip were measured with the Mac Reflex motion analysis system. One reflective marker was placed on the occiput and another on the S1 sacral vertebra between the hip "dimples". The maximal amplitudes of the body segments in motion analysis were calculated as in the repeatability study. The validity of the motion analysis was evaluated by comparing the maximal x (lateral) movement amplitudes of the head and the S1 sacral vertebra to the lateral-distance values (maximum amplitudes in lateral direction) measured with a platform in three different tests. In addition, the maximal anterior-posterior amplitudes of the head and the S1 sacral vertebra were compared with the anterior-posterior distance values (maximum amplitude in the anterior-posterior direction) obtained in the same tests.

In the validity study, the subjects stood on the platform with their backs towards to the video cameras, with their arms down their sides and looking forward. The test battery consisted of three tests: standing on two legs with the eyes open and the eyes closed and standing on the right leg with the eyes open. The recording time in each test was 10 seconds. The subjects were given an opportunity to practise standing on one leg. All the validity measurements were done by the same therapist.

4.3.3 Anthropometric measurements (study IV)

The subjects stood barefooted with their heels 12 cm apart and their arms down their sides. They were told to distribute their body weight symmetrically on both feet. Each subject's body height and the distances of the hip and knee joints to the ground were measured using two fixed tape measures on the wall and a digital level in the horizontal direction. The result was recorded when the level showed 0° between the anatomic landmark and the tape measure scale. The anthropometric measurements were made according to the anatomical landmarks (spaces of the knee and hip joints) identified by palpation and marked with a dermatographic pencil.

The lengths (distance between the heel and the tip of the great toe) (cm) and the widest widths of the forefeet and the heels (cm) of each study subject were determined utilizing footprints on paper. The subjects were weighed using a digital weighing machine (precision 0.2 kg), and their Body Mass Index (BMI) values were calculated (mass/ height^2, kg/m^2). All the measurements were made by the same person. Both sides of the extremities were measured, but only the right-side values were presented together with the height and BMI values.

4.3.4 Statistical methods

The mean values and the standard deviations were calculated and presented. The repeatability of the motion analysis balance measurements (study II) was tested using both the maximal and the total values of the measured movements. Wilcoxon's Signed Ranks Test was used to evaluate the possible systematic changes between the three measurement sessions. Non-parametric statistics was used in testing. The intraclass correlation coefficient (= ICC) and the standard error of measurement (=SEM) were used in reporting repeatability results (Fleiss 1986). As the ICC values are not always appropriate for the assessment of reliability (Altman &Bland 1983, Bland & Altman 1986, Bruton *et al.* 2000), the repeatability results of total navel movement were presented schematically. In the validity study (study II), Spearman's correlation coefficient was used to compare the balance results of the motion analysis system and the platform.

In the studies III and V, Student's t-test for paired samples was used to evaluate the differences in the movement values between the tests with the eyes open and the eyes closed. Student's t-test for independent samples and one-way analysis of variance (ANOVA) were used to compare the values between women (n=50) and men (n=50). In comparing the different groups (groups 1-10) by age and gender, the non-parametric Kruskall-Wallis test for independent samples was used.

In study IV, Pearson's correlation coefficients were used to compare both the maximal amplitudes and the total movements with the subjects' anthropometric characteristics in the whole population (n=100) and separately in the male (n=50) and female (n=50) groups and in the different age groups. The independent effects of the anthropometric characteristics, correlating at the level of 0.05 with the balancing movements, were assessed by linear regression analysis. To illustrate the associations between the anthropometric values and the body movements in stance on two legs more accurately, the correlating independent variables were entered into the model one after another, following the squared multiple correlations, adjusted r square values and p-values. The model was built by entering the variables whose addition increased the r square most at each step. For all statistical tests, the 0.05 level of probability was accepted as a criterion of statistical significance.

5 Results

5.1 Body movements in postural stability with motion analysis (Study I)

In standing on two legs, the greatest maximal amplitude of the body segments was observed in the anterior-posterior direction. The largest maximal amplitude in the anterior-posterior direction was measured in the head (mean 16 mm) and the smallest in the ankles (mean 1 mm). The maximal movement amplitudes in the lateral direction also decreased from the head (mean 4.7 mm) to the ankles (0.6 mm). Both maximal velocity (mean 38.2 mm/s) and maximal acceleration (mean 582 mm/s^2) were highest in the head. On two-leg stance, the different segments of the body moved in the same direction at almost the same time. The detailed movements of the head (marker number 1), navel (3), right knee (6) and ankle (8) of one healthy subject are presented in Figure 3.

In standing on the right leg, the greatest maximal amplitudes were observed in the lateral direction. The largest maximal amplitude in the lateral direction was measured in the head (mean 45.7 mm) and the smallest in the ankles (mean 9.5 mm). In the lateral direction, the maximal movement amplitudes of the different body segments were over 10-fold on one-leg compared to two-leg stance. The maximal anterior-posterior movement amplitude was also remarkably greater on right-leg stance. The magnitudes of movement amplitudes did not differ between the different body segments as much as on two-leg stance. Vertical body movements were small, but noticeable in both standing positions. Maximal velocities were greater in all segments of the body on one-leg stance than on two-leg stance, and maximal velocity was greatest in the knee (mean 96 mm/s). The maximal accelerations of movements were much higher on one-leg than two-leg stance, being especially prominent in the knee (mean 1055 mm/s^2) and the ankle (mean 980 mm/s^2).

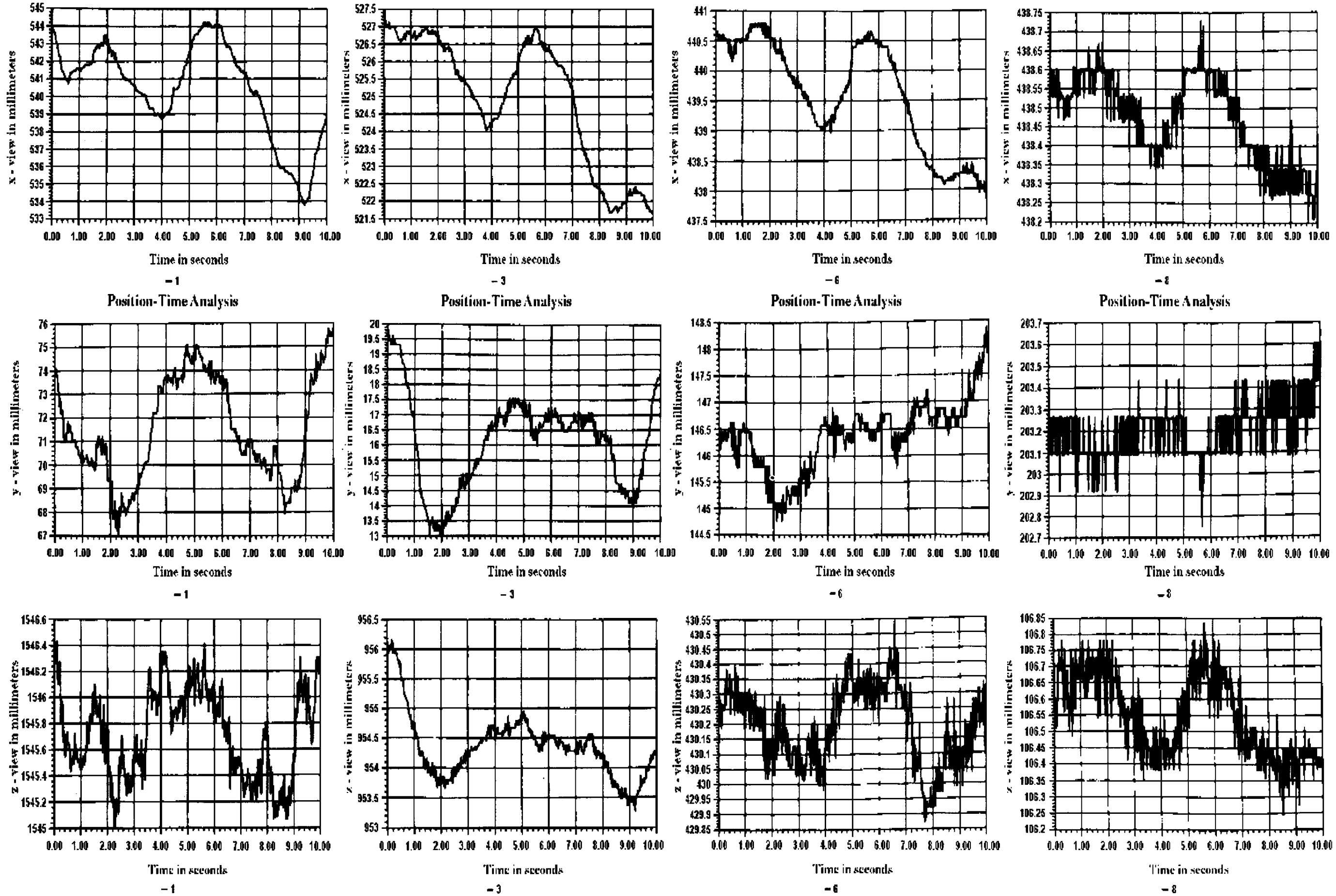


Fig. 3. The movements of the head (-1), chest (-3), right knee (-6) and right ankle (-8) in the lateral (x), anterior-posterior (y), and vertical (z) directions of a healthy subject on two-leg stance during a measurement time of 10 seconds.

The maximal errors of movement values in the lateral (x) (= 0.4 mm), anterior-posterior (y)(=3.43 mm) and vertical (z) (=0.48 mm) directions were calculated based on the difference between the maximum and minimum values during the 10-second measurement period (stable marker). The following maximal amplitudes of movements with and without a reference marker were obtained, respectively: *in the lateral direction*; head (8.5 mm and, 8.7 mm), navel (4.8 mm and 4.8 mm), right knee (2.3 mm and 2.4 mm) and right ankle (0.4 mm and 0.5 mm), *in the anterior-posterior direction*; head (14.0 mm and 14.4 mm), navel (10.2 mm and 10.8 mm), right knee (5.5 mm and 5.8 mm) and right ankle (1.5 mm and 2.0 mm), *in the vertical direction*; head (3.2 mm and 3.4 mm), navel (3.1 mm and 3.1 mm), right knee (0.7 mm and 0.9 mm) and right ankle (0.3 mm and 0.8 mm). A comparison of the movement values with and without a reference marker showed that the error caused by the system did not have significant effects on the maximal amplitude values. The accuracy of motion analysis to detect a movement of 0.5-0.7 mm in the different axial directions is demonstrated in Figure 4.

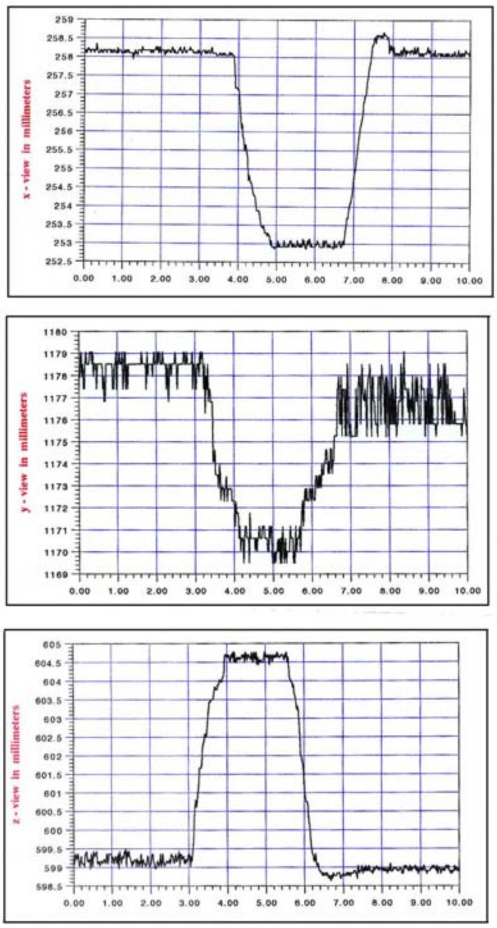


Fig. 4. The accuracy of the motion analysis system to detect the movement in the lateral (x), anterior-posterior (y) and vertical (z) directions during a measurement time of 10 seconds.

5.2 Repeatability and validity of body movement measurements (Study II)

5.2.1 Repeatability of the measurements

In a comparison of balance measurement values between the measurement sessions, no statistically significant differences between the sessions in body movements in standing on two legs with the eyes open were observed. Using the total movement values, repeatability was better (ICC values ≥ 0.79) than with maximal movement amplitudes in all the measured body segments. The standard errors of the measurements (=SEM) seemed to be highest in the repeatability values of the head and the ankles, but repeatability was almost equally good in the measurements of these body segments as in the others. The details of the results are presented in the tables 4 and 5. The schematic representation of the reproducibility of navel movement was good (Fig. 5).

Table 4. Intraclass correlation coefficients (ICC) and standard errors of the measurements (SEM) of the three balance measurements using values of total movements (n=10).

Body part	ICC	SEM (mm)
Head	0.82	16.12
Navel	0.91	7.84
Knee, right	0.90	8.88
Knee, left	0.92	8.57
Ankle, right	0.79	19.23
Ankle, left	0.87	26.25

Table 5. Intraclass Correlation Coefficients (ICC) of the three balance measurements using values of maximal amplitudes in the three cardinal directions (n=10).

Movement direction	ICC (mean)	ICC (range of values)
Lateral (x) movement amplitude	0.62	0.44-0.70
Ant-post (y) movement amplitude	0.61	0.33-0.86
Up-down (z) movement amplitude	0.55	0.27-0.79

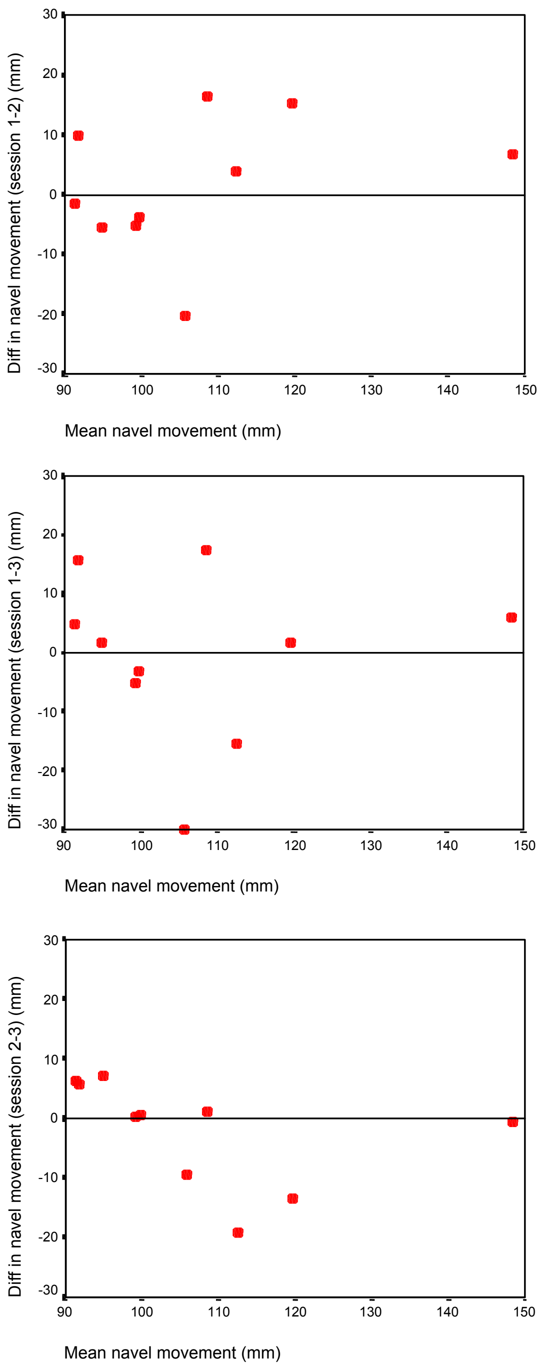


Fig. 5. Differences (mm) in repeated total movement values of the navel (n=10) based on the motion analysis system between the measurement sessions (1-2, 1-3, 2-3).

5.2.2 Validity of measurements

In comparing the maximal lateral head and S1 sacral vertebra movements (Mac Reflex motion analysis) to the lateral distance values (In Good Balance platform) measured with the subject standing on two legs with the eyes open, the correlation coefficients were 0.70 ($p<0.01$) for the head and 0.72 ($p<0.01$) (Spearman's correlation test) for the S1 sacral vertebra movement. The respective values measured for standing on two legs with the eyes closed were 0.61 ($p<0.05$) for the head and 0.79 ($p<0.01$) for the S1 sacral vertebra movement and those for standing on the right leg 0.70 ($p<0.01$) for the head and 0.90 ($p<0.01$) for the S1 sacral vertebra movement. In comparing the maximal anterior-posterior head and S1 sacral vertebra movement to the anterior-posterior distance values measured with the subject standing on two legs with the eyes open, the correlation coefficient was 0.75 ($p<0.01$) (Spearman) for the head and 0.92 ($p<0.01$) for the S1 sacral vertebra movement. The respective values measured when standing on two legs with the eyes closed were 0.90 ($p<0.01$) for the head and 0.87 ($p<0.01$) for the S1 sacral vertebra movement and those when standing on the right leg 0.60 ($p<0.05$) for the head and 0.94 ($p<0.01$) for the S1 sacral vertebra movement. For two-leg stance, the maximal distances (maximal amplitudes) measured with the In Good Balance platform were smaller than the maximal head and hip amplitudes measured with the Mac Reflex motion analysis system (Fig. 6).

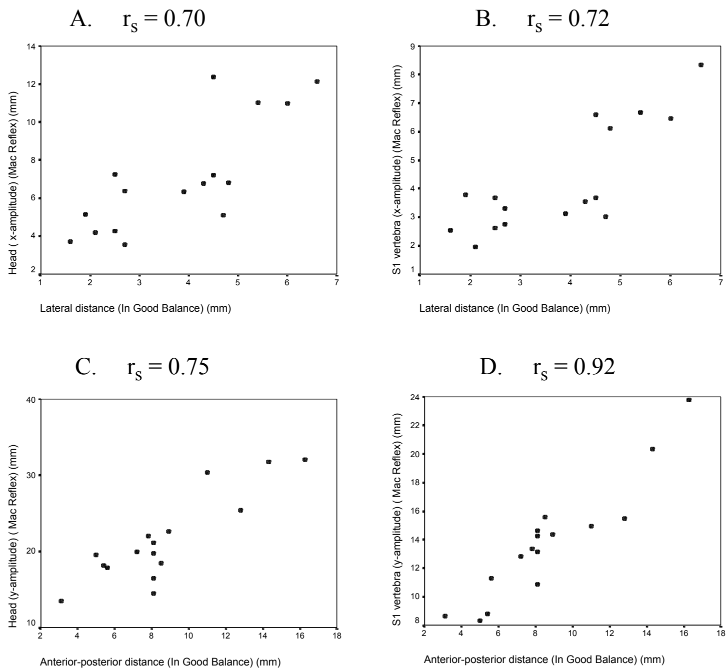


Fig. 6. Relationship (r_S = Spearman's correlation coefficient) between the Mac Reflex motion analysis and the In Good Balance platform measurements during stance on two legs with the eyes open using A. maximal lateral head movement amplitude (Mac Reflex) and lateral distance value (In Good Balance), B. maximal lateral S1 movement amplitude (Mac Reflex) and lateral distance value (In Good Balance), C. maximal anterior-posterior head movement amplitude (Mac Reflex) and anterior-posterior distance value (In Good Balance) and D. maximal anterior-posterior S1 movement amplitude (Mac Reflex) and anterior-posterior distance value (In Good Balance).

5.3 Associations between age, gender and body movements (Study III)

The absolute numerical values of maximal movement amplitudes in the whole population during standing with the eyes open and closed are presented in Table 6 and the absolute numerical values of total movements in Table 7. The detailed maximal movement amplitudes in the groups categorised by age and gender are presented in the Tables 1-6 in the Appendix. The total movements in the eyes-open and eyes-closed conditions were stratified by age and gender group and are illustrated in Figure 7.

Table 6. The maximal amplitudes of the body segments in the lateral, anterior-posterior and vertical directions in the whole population (n=100) during standing with the eyes open and closed.

Body part	Movement direction	Maximal amplitudes (mm)			
		Eyes open		Eyes closed	
		Mean	SD	Mean	SD
Head	Lateral	10.6	4.6	12.2	6.4
	Ant-post	19.4	7.5	27.3	12.0
	Vertical	3.4	1.5	4.3	1.8
Navel	Lateral	6.8	3.7	8.5	5.3
	Ant-post	12.0	5.3	15.4	6.6
	Vertical	4.1	1.8	4.7	2.0
Knee right	Lateral	4.0	2.2	4.7	3.0
	Ant-post	5.6	2.8	6.6	3.0
	Vertical	1.8	1.3	2.5	2.0
Knee left	Lateral	3.7	2.0	4.7	3.0
	Ant-post	5.6	2.4	6.6	3.0
	Vertical	1.7	1.3	2.2	1.9
Ankle right	Lateral	0.9	0.5	1.2	0.6
	Ant-post	1.7	0.7	2.1	1.0
	Vertical	0.8	0.4	1.0	0.5
Ankle left	Lateral	0.9	0.5	1.2	0.9
	Ant-post	1.6	0.7	2.0	1.0
	Vertical	0.8	0.4	1.0	0.6

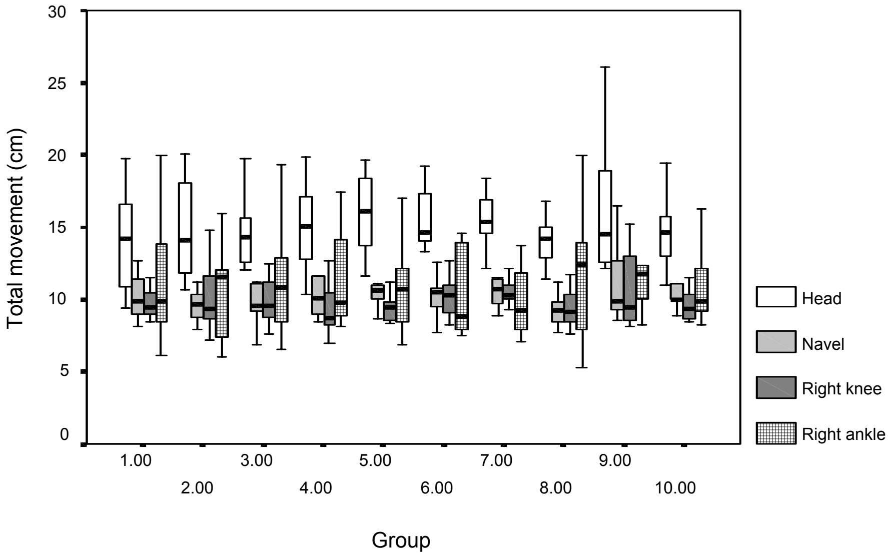
Maximal amplitudes: In two-leg stance with the eyes open, there was a statistically significant difference between the age groups in the anterior-posterior head movement ($p< 0.05$). During two-leg stance with the eyes closed, all the measured body parts moved more both in the lateral and the anterior-posterior directions ($p< 0.001$) than in the eyes-closed conditions. During standing on two legs with the eyes open, the “largest” maximal vertical amplitudes were recorded for the navel and the head, and there was a statistically significant difference between the age groups in the vertical navel movement ($p<0.05$). The subjects aged 71-80 years presented higher navel movement values than the younger subjects.

Table 7. The total movements of the body segments in the whole population (n=100) during standing with the eyes open and closed.

Body part	Total movements (cm)			
	Eyes open		Eyes closed	
	Mean	SD	Mean	SD
Head	15.0	2.9	17.0	4.0
Navel	10.3	1.8	11.5	3.1
Knee right	9.8	1.5	10.4	2.3
Knee left	10.1	1.9	11.0	4.0
Ankle right	11.1	3.3	12.0	3.7
Ankle left	11.1	3.4	11.2	2.8

Total movements: During two-leg stance with the eyes open, the head moved most, while the quantities of the other measured body movements did not differ remarkably in the whole population. During two-leg stance with the eyes closed, all the above body parts moved more ($p < 0.001$) than in two-leg stance with the eyes open. The results did not show any statistically significant differences between the balancing movements of the separate body segments of the groups in either standing position, and there were no statistically significant differences in the balance measurement values between the men and the women in standing on two legs with the eyes open and closed.

In standing on two legs with the eyes open



In standing on two legs with the eyes closed

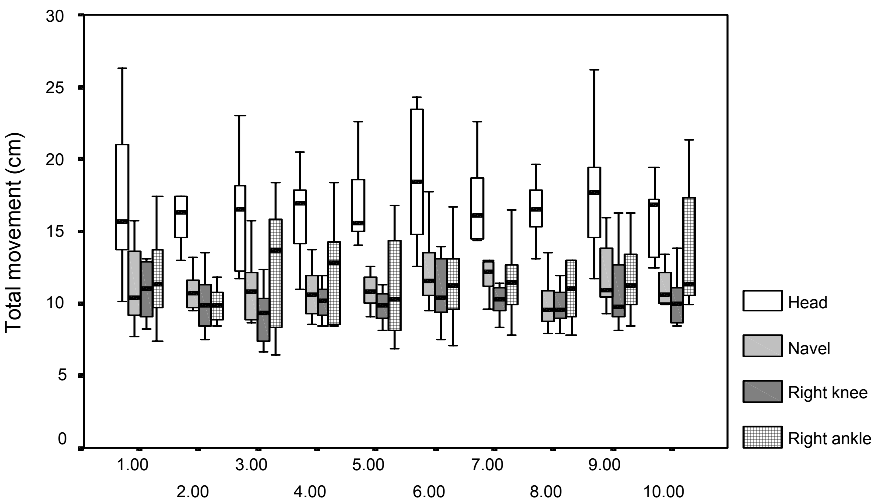


Fig. 7. Total movements of the head, navel, knee and ankle in the groups. In Fig. 7 a boxplot summarising the median (a line across the box), quartile (interquartile range = 50% of values) and extreme values (whiskers extending from the box to the highest and lowest values) defined by the group number (1=women aged 31-40 yr., 2=men aged 31-40 yr., 3=women aged 41-50 yr., 4=men aged 41-50 yr., 5=women aged 51-60 yr., 6=men aged 51-60 yr., 7=women aged 61-70 yr., 8=men aged 61-70 yr., 9=women aged 71-80 yr., 10=men aged 71-80 yr.)

5.4 Associations between anthropometric factors and body movements (Study IV)

The correlation coefficients between balancing movements and anthropometric factors are presented in the Appendix (Table 7). In a linear regression analysis based on squared multiple correlations in the whole study population (n=100) *in the eyes-open conditions*, 6.8 % ($p<.05$) of the variation in the anterior-posterior head movement and 8.3 % ($p<.05$) of the variation in the anterior-posterior navel movement were explained by heel width. Body height explained 5.7 % ($p<.05$) and body height and foot length together 6.5 % ($p<.05$) of the variation in lateral knee movement. Further, 8.5 % ($p<.05$) of the variation in anterior-posterior knee movement was explained by foot length and heel and foot width and 10.0 % ($p<.05$) by foot length, heel width and knee-ground distance. BMI explained 7.4 % ($p<.05$) of the variation in anterior-posterior ankle movement and 6.3 % ($p<.05$) of the variation in vertical ankle movement. *In the eyes-closed conditions*, BMI explained 5.9 % ($p<.05$) of the variation in vertical navel movement and body height 4.1 % ($p<.05$) of the variation in anterior-posterior knee movement.

Gender differences. a. Males. *In the eyes-open conditions,* 10.7 % ($p<.05$) of the variation in anterior-posterior head movement was explained by heel width. Body height explained 19.8% ($p<.01$) and body height, knee-ground distance, hip-ground distance and foot length together 23.5% ($p<.05$) of the variation in lateral navel movement. Body height also explained 9.2 % ($p<.05$) of the variation in lateral knee movement and foot length 9.6% of the variation in anterior-posterior knee movement. Further, body height explained 14.5% ($p<.01$) and body height, foot length and knee-ground distance 15.5% ($p<.05$) of the variation in total navel movement. There were not statistically significant correlations *in the eyes-closed conditions.*

b. Females. *In the eyes-open conditions*, foot length explained 10.7% of the variation in lateral navel movement and 11.3 % ($p<.05$) of the variation in anterior-posterior knee movement. In addition, foot length explained 18.8 % and foot length, body height, knee-ground distance and hip-ground distance together 23.8 % ($p<.05$) of the variation in lateral knee movement. 10.3% ($p<.05$) of the variation in anterior-posterior ankle movement and 13.9 % ($p<.01$) of the variation in vertical ankle movement was explained by the BMI value.

In the eyes-closed conditions, BMI explained 7.9 % ($p<.05$) of the variation in vertical navel movement and body height 10.2 % ($p<.05$) of the variation in anterior-posterior knee movement. In addition, 13.7% of the variation in total navel movement was explained by foot width ($p<.01$).

5.5 Velocities and accelerations of body segments during stance (Study V)

During stance on two legs with the eyes open, the highest maximal velocity and acceleration values were recorded for the head, while the movement velocities and accelerations of the navel, the knees and the ankles did not differ remarkably from each other in the whole study population (n=100). During stance on two legs with the eyes closed, all the above body segments except the balancing movement of the left ankle had statistically significantly higher maximal movement velocity and acceleration values than those recorded during standing on two legs with the eyes open ($p < 0.05$). In comparing the results between the age groups, no statistically significant differences appeared in the velocities or accelerations of the balancing body movements in either standing position. The body movement measurement results are presented in the Tables 8 and 9.

Table 8. Maximal velocity of the movements of separate body segments in stance on two legs with the eyes open and closed (n=100).

Body part	Maximal velocity of body movements in stance on two legs					
	Eyes open (1)		Eyes closed (2)		*p* between the tests 2-1 (*)	Change (%) between tests 2-1 (**)
	Mean (mm/s)	SD	Mean (mm/s)	SD		
Head	43	11	50	15	< 0.01	+ 17
Navel	28	7	32	11	< 0.01	+ 15
Right knee	26	8	30	11	< 0.01	+ 16
Left knee	26	8	30	14	< 0.01	+ 17
Right ankle	25	9	28	11	< 0.05	+ 11
Left ankle	24	9	25	8	NS	+ 1

* , ** p-values and percentage (%) changes in maximal velocity values between the eyes closed and the eyes open conditions

Table 9. Maximal acceleration of the movements of separate body segments in stance on two legs with the eyes open and closed (n=100).

Body part	Maximal acceleration of body movements in stance on two legs					
	Eyes open (1)		Eyes closed (2)		*p*–between the tests 2-1 (*)	Change (%) between tests 2-1 (**)
	Mean (mm/s^2)	SD	Mean (mm/s^2)	SD		
Head	850	247	952	276	< 0.01	+ 12
Navel	544	156	601	186	< 0.01	+ 11
Right knee	511	170	576	194	< 0.01	+ 13
Left knee	504	153	596	267	< 0.01	+ 18
Right ankle	487	197	542	217	< 0.05	+ 11
Left ankle	482	227	492	171	NS	+ 2

* , ** p-values and percentage (%) changes in maximal acceleration values between the eyes closed and eyes open conditions

6 Discussion

6.1 Subjects

In the studies I and II, the subjects were healthy staff members, which somewhat restricts the generalisation of the results. However, study I was mainly a methodological study, and subject selection was hence probably not the most critical point. In the reliability and validity studies, the populations were rather small but separate. It must be admitted that the results might have been more conclusive if larger samples of subjects had been used. In previous studies concerning the methodological aspects of balance measurements with motion analysis, the study populations have commonly been rather small (e.g. Woollacott *et al.* 1988; n= 10, Aramaki *et al.* 2000; n=6). One reason for the small populations may be the fact that movement analyses are laborious to make. Still, in a reliability and validity study that also included movement analysis, the study population was bigger than in the present study (e.g. Benvenuti *et al.* 1999; n= 36).

In the studies III-V, a hundred healthy subjects aged 31 to 80 years from various socioeconomic categories were randomly selected from the population of Oulu. This composition of the present study sample can be presumed to represent the normal adult population of this area. The results can be used as a tentative basis for generalization, but some unintended selection bias may be present. In previous movement analyses, the relationship between aging and the movements of individual body segments during stance have been examined using even smaller study samples, e.g. 24 subjects aged 21 to 78 years (Panzer *et al.* 1995). More detailed information about the effects of age on the postural control mechanism could have been obtained if the age groups had been bigger, as in previous studies with platforms (e.g. Era & Heikkinen 1985). The sizes of the separate gender populations were adequate, but the age range was possibly too large to indicate the exclusive effects of gender. For example, Yoshida *et al.* (1983) assessed the differences in body movements during standing up from and sitting down on a chair between young males (n=10) and females (n= 10) and between older males (n=10) and females (n=10).

6.2 Comments on the methods

6.2.1 Measurements with a motion analysis system

The body movements during stance were measured using the Mac Reflex motion analysis system, previously documented in 3 D coordinate predictions (Levy & Smith 1995). To my knowledge, there were no previous studies using this measurement system in balance studies. As information about separate body segments during stance is important for developing balance training programmes, this study was considered necessary. The variability was minimised as far as possible by having the same physiotherapist plan the measurement setting and procedure and measure all the subjects. Having more physiotherapists involved in the measurements might have had negative effects, especially on the repeatability of marker placement and subject instructions.

Although the measurement procedures with motion analysis were short and constant, the analyses were time-consuming. Technically, all the measurements worked well, but the interpolation for a disappearing marker, for example, had to be done mechanically and manually. The marker disappeared in some measurements during standing on one leg, when the subject raised his/her left leg behind the right leg. There would have been many possibilities for marker placing, including certain joint positions (e.g. Cappello *et al.* 1995). This would have required different camera placing or measurements with 4 cameras (e.g. Benvenuti *et al.* 1999). The placement of ankle markers on the anterior angular surface may not have been an optimal choice in view of the camera creating and recording the image of the centre of the marker. This may have produced some slight errors in the ankle movement values. Mounting the markers on the lateral malleoli of the ankles (Benvenuti *et al.* 1999) could have been given more accurate results. However, this would have demanded camera placement on the lateral sides of subject.

Separate scripts were made in the Mac Reflex module to calculate the movement values. The method of calculating the maximal amplitude values needs only the highest amplitudes of movement. The reason for using these values was that the mean values processed by the measurement system would have been useless. The total movement values were calculated to obtain a better description of the summarised movement during the period of measurement, as suggested by platform studies (e.g. Ekhdahl *et al.* 1989, Ojala *et al.* 1989). The choice of using a modified Pythagoras' theorem in the calculation of total movements values may have produced some bias, because this theorem is basically used to calculate the sides of a triangle (Enoka 1988). In the present study, the movement measurements were three-dimensional. In the studies I and V, velocity and acceleration values were used to analyse the movements of separate body segments, as suggested by Pyykkö *et al.* (2000). It could have been more informative to use, for example, mean velocity and acceleration values (Korpelainen *et al.* 2000, Moe-Nilssen 1998) instead of maximal values. However, this was not possible in WingZ.

6.2.2 Repeatability and validity measurements

In the repeatability study, the measurements with a Mac Reflex motion analysis system were made three times, and their reliability was tested using two values: maximal amplitudes and total movements. Three measurement sessions were done, because the test-retest repeatability was assessed during a short (one day) and a longer (one week) interval. The repeatability of using maximal amplitude values was assessed concerning their value in the evaluation of momentary balancing movements, whereas the total movements (resultant movements) could be used in the evaluation of stance during the measurement time.

The validity study was made by using two different balance measurement systems, motion analysis and platform, which were synchronised with an interface cable. To get appropriate data for the motion analysis measurements, two body parts, head, on the top of the inverted pendulum, and the S1 sacral vertebra, near the centre of gravity, were selected to describe postural control. The validity of the motion analysis system was assessed by comparing the results with the platform results, because platforms are widely used in balance measurements (e.g. Kinney LaPier *et al.* 1997, Era *et al.* 1996). Validity was evaluated by comparing both the maximal lateral and the anterior-posterior amplitude values of the head and the S1 sacral vertebra to the corresponding distance values measured with a platform in three different tests. These values were used, as they were thought to reflect the same aspects of balancing stance. Horak and Nashner (1986) pointed out that the possible impairment of the balance system may both increase body sway and also alter the movement strategy used to control balance. In a study with a platform and a video –system, however, no significant correlations were found between force plate and movement strategy measures (Karlsson & Frykberg 2000).

6.2.3 Anthropometric measurements

The aim was to evaluate the basic body characteristics that may affect the movements of separate body segments. For example, a longer lever arm would possibly cause a greater amplitude of movements. In previous balance studies, short body height has been found to be one reason for the increased risk for falls (Davis *et al.* 1999), and a small body mass has also been reported to be associated with poorer posture control (Era *et al.* 1996). Therefore, the subject's body height, weight and the distances from the hip and knee joints to the ground were measured. Body part distances were measured according to the anatomical landmarks (spaces of the knee and hip joints) identified by palpation and marked with a dermatographic pencil (e.g. Benvenuti *et al.* 1999) and then by using two fixed tape measures on the wall and a digital level in the horizontal direction. The same person made all the anthropometric measurements. However, the possible errors in space palpation may have decreased the measurement accuracy to some extent. Only the right side values were reported (Table 7 in Appendix), because the results of the left side were similar.

Because foot size structure and disorders, such as foot deformities, contribute to functional impairment and, thereby, postural stability (Menz & Lord 1999a,b), the size of the foot (the lengths and widest widths of the forefeet and heels) of each subject were determined utilizing footprints on paper. A simple footprint analysis was done using the width and length values of the feet, because the aim in the present study was to evaluate the basic body characteristics. However, it would have been useful to analyze footprint indices (Cavanagh & Rodgers 1987, Hawes *et al.* 1992) more accurately, to obtain more information on weight distribution (e.g. Haas & Burden 2000) and support size (e.g. Pyykkö *et al.* 2000) affecting postural stability.

6.2.4 Limitations of the study design

During the motion analysis, the subjects stood on two legs with their heels 12 cm apart and their eyes open or closed. The subjects were told to distribute their body weight symmetrically on both feet. It would have been more accurate if the distribution of the base of support could have been measured, as was done in some other balance studies (Haas & Burden 2000). In force plate studies, the extent of medio-lateral body sway has been found to be larger when the feet are moved apart and more muscles are involved in the movement (Rothwell 1994, Aramaki *et al.* 2001). This possibility was eliminated as a source of variability, although individually based stance strategies (e.g. Karlsson & Frykberg 2000) could otherwise have been more appropriate in order for the subject to be able to cope with environmental changes in stance.

The visual distance for the target was selected to be 5.5 m, and the subjects were instructed to look forward at the marked point on the wall. Brandt *et al.* (1986) reported that the accuracy of the visual system to control postural stability is relevant when the looking distances are rather small (less than 2 m). This selected visual distance may not have been adequate to show the visual control on posture. Nevertheless, the subjects were standing in structural visual surroundings that did not cause disturbances in postural control. Besides, the organisation of the system of measurement, especially the placement of cameras, had to be considered.

The standing time in the tests with the eyes open and closed was 30 seconds, which included a 20-second pre-sampling time and a 10-second recording time. In the previous balance studies, a measurement time longer than 10 seconds was recommended (LeClair & Riach 1992). Therefore, a longer measurement time could also have been used. The reason for the rather short time was the magnitude of the data available for motion analysis. On the other hand, the data processing required a lot of accuracy and work, and a longer measurement would have been even more laborious. However, the study should have been carried out in 20-30 seconds, as recommended by LeClair & Riach (1992), to obtain a more detailed description of the postural control mechanism in adults.

6.3 Comments on the results

6.3.1 Body movements in postural stability with motion analysis

In the present study, the motion analysis system identified the differences in both the movements of separate body segments and the movement directions. Thus, motion analysis introduces a new view to balance measurements, as the system can analyse balance from a different perspective. Studies with platforms yield information on the amount and direction of total body sway, but motion analysis gives versatile information on the movements of separate body segments. This provides possibilities to analyse segmental movement variability in postural control. For example, vertical movements of body segments have seldom been reported, because they are more difficult to measure with posturography. Nevertheless, old people flex their knees and hips more for balancing their body position (Woollacott & Shumway-Cook 1990). The results of the small sample showed that the method is both sensitive and precise in studying body movements. However, it is possible that the smallest body movements were not distinguishable from the measurement disturbances. The present values were calculated from the difference between the minimum and maximum values. This method may involve bias, because the method processes only the highest amplitudes of sway. The reason for using these values was that the mean values did not give real information about movements. This calculation system, which has previously also been used in platform studies (e.g. Kinney LaPier 1996), is only one method for analysing measured data.

During standing on two legs and one leg, the results showed that the measured body segments moved in all three directions: anterior-posterior, lateral and vertical directions. During standing on two legs, the different body parts had the greatest maximal amplitude in the anterior-posterior direction. Several researchers using different methods have previously reported similar results about total body sway on normal stance (Hasselkus & Shambes 1975, Nashner 1985, Era & Heikkinen 1985, Palovaara *et al.* 1992). It was noticeable that, in standing on two legs, the measured body segments moved almost at the same time and in the same direction as in the pendulum type of movement (see Fig.3).

During standing on one leg, the main direction of body movements was lateral. Nashner *et al.* (1985) explained that the increased lateral movement on one-leg stance is due to the decreased area of weight bearing and the narrow base of support. In the present study, in standing on one leg, the maximal amplitude of different body segments oscillated approximately 10 times more in the lateral direction than during standing on two legs. Besides, the magnitudes of movement of individual body segments did not differ from each other as much as in two-leg standing. In standing on one leg, the maximal amplitudes of the movements of individual body segments were also somewhat greater in the anterior-posterior and vertical directions. The results of previous studies support these findings. Era & Heikkinen (1985) found in their study with a platform that total body sway was eight times greater during standing on one leg than in standing on two legs. They also noticed the increase of lateral sway in standing on one leg.

In standing on one leg, both the maximal velocity and the acceleration values of the knee and the maximal acceleration value of the ankle also increased significantly. According to these results, it is essential to be able to move the knee and the ankle very quickly to balance one's position in one-leg standing. However, this may cause difficulties for old people, who often have muscle, joint and nerve impairments. For example, Lord *et al.* (1991a) noticed that the decreased muscle strength of elderly people contributes to their partly impaired postural control. Besides, Whipple *et al.* (1987) found knee and ankle weakness to be risk factors for falling.

6.3.2 Repeatability and validity of measurements

The repeatability values were examined both numerically (ICC, SEM) and schematically, because graphical techniques have been suggested for the assessment of repeatability (Altman & Bland 1983, Bland & Altman 1986). The range of variation in ICC values (repeatability) using maximal movement values was from 0.27 to 0.79. The system was accurate enough to distinguish the differences during stance between the measurement sessions, and the results showed that the momentary movements may vary between the sessions. The repeatability of the total body movement values seemed to be higher (ICC= 0.82-0.92). Although the measured body segments moved maximally in somewhat different ways, they still moved equally much during the ten second measurement time in each test. It is thus recommended that, in further studies, and especially in longitudinal studies, total movement values should be used instead of maximal amplitude values. The maximal amplitudes may have value in the evaluation of momentary balancing movements. In the present study, the correlations were high when the results of the maximal head and S1 sacral vertebra movements measured with motion analysis were compared to the results of the lateral (x) and the anterior-posterior (y) distances measured with a platform. It seems that these parameters reflect the same aspect of balance, although the view of measurement was different in measuring the movements of separate body segment and total body sway.

No previous studies dealing with this phenomenon were found, but there were studies concerning the comparison of other balance measurements and tests. Benvenuti *et al.* (1999) used a three-dimensional motion analysis and a force platform to evaluate quiet standing of aged subjects with balance problems. ICC values indicated a high level of retest reliability. The reliability of the "Step-balance" test was also found to be good (Hill *et al.* 1996), but the accuracy of functional balance tests is lower than that of laboratory tests. The correlation between force platform measurements and functional balance tests has also been found to be significant (Ekdahl *et al.* 1989), and in future balance rehabilitation follow-ups, it would be interesting to compare the results of motion analysis and functional balance tests. The results of reliability and validity studies using a force platform have been controversial; some researchers have reported poor reliability (e.g. Goldie *et al.* 1989), while some others have found the reliability to be rather good (Taguchi *et al.* 1978, Ishizaki *et al.* 1991, Levine *et al.* 1996). Karlsson and Fryktberg (2000) analysed and compared five types of measurement values obtained with a force platform and two video cameras: the results revealed significant correlations, but the movement strategy measures did not correlate in any of the other measurements. Any examination of repeatability and validity should be done carefully, because of the variation of the measurement parameters, methods and population volumes. Geurts *et al.*

(1993), for example, found in their study with a force platform that root mean square velocity had maximal intrasubject consistency.

The motion analysis system seems to be reliable and valid in balance research of healthy subjects of different ages and subjects with balance impairments both when used on its own and when combined with other measurement systems (e.g. Benvenuti *et al.* 1999). As many of the therapeutic methods used in the rehabilitation of balance disorders are based on assumptions of normal movements and reflex function (Woollacott & Shumway-Cook 1990), the information of separate body movements could be used in planning physiotherapeutic balance training and to assess the need for development of balance training programs.

6.3.3 Associations between age, gender and body movements

An ability to effectively treat patients with dysequilibrium requires better understanding of the balance control of healthy subjects of different ages. An evaluation of postural stability in stance on two legs with the eyes open and with the eyes closed showed that the biggest maximal amplitudes were recorded for the head, and the values decreased from the head to the ankles. As suggested by the balance studies done with other measurement systems (e.g. Brandt *et al.* 1986, Palovaara *et al.* 1992), vision seemed to have an increasing influence on both momentary balancing body movements (maximal amplitudes) and total movement excursions.

Although the total movement excursions during the period of 10 seconds were most marked in the head, the movements of the other body parts did not differ remarkably and the total movement excursions of the knees and ankles were almost equally large as the total movement excursion of the navel, although the ankles and knees are closer to the ground. A more detailed graphical evaluation shows, for example, that during the measurement period the amplitude of head movement is large, but the movement frequency is rather low, whereas the amplitude of ankle movements is low, but the movement frequency high. Both the amplitude and the frequency of navel movement are moderate compared to the corresponding head and ankle movements. This may imply that the nature of movement is different in different segments of the body. It could also be that the movement velocities and accelerations of the ankles and knees differ from the velocities and accelerations of the head and navel movements. However, this phenomenon warrants further examinations. It is also possible that the present measurement system with six reflective markers may not have distinguished absolutely between the movements of the different body segments, and the movements of the body segments may have been autocorrelated to some extent.

Di Fabio & Emasithi (1997) found in their study that older adults may use a "backward" head stabilization strategy to maintain balance during more difficult balance tasks. In the present study, the older adults (aged 71-80) had more extensive anterior-posterior head movement in standing with their eyes open than the younger adults. In non-visual conditions, there were no differences in the maximal anterior-posterior head amplitudes or the total head movement excursions between the groups. Yet, it is obvious that the significance of head movements and positions should be given more attention in balance training and evaluation, because the overall control of posture is supposed to be executed by the central vestibular mechanism that receives information about the positions of body segments, especially the velocity of head movement (Toppila & Pyykkö 2000). In other studies, the role of coordinating head movements in balance maintenance has been found to be important (Shupert *et al.* 1988), and measurements of head sway have been utilised in an analysis of postural stability (Koncek *et al.* 1993).

Apart from the difference in maximal anterior-posterior head amplitude, the comparison of age groups revealed no other differences in maximal amplitudes or total movement excursions, with the exception of the maximal vertical navel amplitude, in either standing test. The difference in vertical navel movement should be assessed critically, because it might been caused partly by the movement of the diaphragm. Thus, the possible age-related changes in balancing systems were not projected into the body movement amplitudes in a normal standing position with the present analysis method. Previous platform studies support our results (Black *et al.* 1982, Ojala *et al.* 1989), but opposite results have also been reported (e.g. Hytönen *et al.* 1993, Matheson *et al.* 1999). A more obvious effect of age has appeared in balance studies with more difficult balance performances, such as standing on one leg (Bohannon *et al.* 1984, Briggs *et al.* 1989)

Gender seemed to have no effect on the balancing movements of the measured body segments. The gender samples (n=50 in both groups) should have been sufficient to indicate the possible differences between the female and male subjects. Similar results were reported in a platform study with subjects aged 20-49 (n=132) (Black *et al.* 1982). However, in some balance studies with a platform, male subjects have been found to sway more than females (Ekhdahl *et al*. 1989, Palovaara *et al.* 1992), while some other studies have yielded opposite results (e.g. Hinchcliffe 1983). Any evaluation of these results should be made with caution, because of the variation in the measurement systems and the variables used and the number of subjects measured. In addition, physical characteristics, such as the differences in height between men and women, may account for at least part of the differences in the above studies.

This information may be used in clinical work and research. The healthy female and male subjects of different ages seemed to control their stance with quite similar quantities of body adjustment. The normal position of standing on two legs is adequate and "easy" enough for practising balance by using variable information sources at all ages, but postural control should also be evaluated and practised in more difficult circumstances, because potential changes are not necessarily seen in easier balance performances. Therefore, balance assessments using different sensory inputs, e.g. with a dynamic platform (Nashner 2001) are recommended. They can be used to determine possible limitations in the afferent system and the automatic motor system that may cause unbalance in stance.

6.3.4 Associations between anthropometric factors and body movements

As the body characteristics have been assumed to affect even the selection or combination between the ankle, hip or suspensory strategies (Woollacott & Shumway-Cook 1990), the relations between these factors and body movements during standing were of interest. The comparison to anthropometric values was performed using two different balancing movement values: total movements and maximal amplitudes. No correlations between the total body movement values and the measured anthropometric body characteristics were found in evaluating the whole population in both measurement conditions, although it was assumed that body height and the dimensions of the extremities could have an effect on at least total head movement, which occurs at the top of the “inverted pendulum” pivoted around the ankles in a standing position (e.g. Kinney LaPier *et al.* 1997, Woollacott & Shumway-Cook 1990). It is likely that the postural adjustments measured with the motion analysis system were made more evenly in the different segments of the body than has been described in a rigid one-link model (Barin & Stockwell 1985). On the other hand, the system for calculating the total movement values included all the movements along the three cardinal axes during 10 s rather than the movement directions separately. Therefore, the effects of single body characteristic could have become covered.

A number of significant correlations were found between the maximal movement amplitudes, which reflect momentary balancing adjustments in the different body parts and the body characteristics in the eyes-open conditions, but only some single significant correlations in the eyes-closed conditions. When the whole study group was evaluated during standing with the eyes open, greater body height slightly increased the maximal lateral movement amplitude of the knee. The maximal lateral navel and knee amplitudes were mainly explained by body height in the male group and by foot length in the female group. To my knowledge, there are no previous studies that would have reported similar gender differences, but it is known that height values and foot size values usually correlate (Nashner 2001).

The variation in maximal anterior-posterior movement amplitudes was explained by foot characteristics rather than by body height and extremity measurement values in both gender groups. A bigger foot size resulted in somewhat greater maximal amplitudes of the head, navel and knee movements in the anterior-posterior direction in standing on two legs. By using footprint indices (Cavanagh & Rodgers 1987, Hawes *et al.* 1992), we might have obtained more information of weight distribution. For example, the lateral displacement of the hallux has been reported to lead to a decreased ability to actively propel the body forwards (Hutton & Dhanendran 1981). This is easy to understand, as the foot provides the only source of direct contact during standing and walking. Special footwear insoles have been found to reduce instability and the risk of falling (Maki *et al.* 1999) by improving weight distribution. The results of this study suggest that the “outside” design of footwear should also be given attention.

Although it seems that there are a number of correlations between especially the maximal amplitudes and the anthropometric factors, there is no single anthropometric factor that would explain the variations in balancing adjustments during stance. The BMI value was the only body characteristic that correlated significantly with the maximal anterior-posterior ankle amplitude and maximal vertical ankle amplitude in the whole study population. These correlations were also significant in the female group. Contrary to these results, Era *et al.* (1996) found inversely relation between BMI and postural stability.

6.3.5 Velocities and accelerations of body segments during stance

Information of the behaviour of different body segments is required to develop better insight into balance control and to give more attention to training concerning the movement coordination, balance reactions and muscular action needed in postural balance. Commonly, positional data (e.g. body movement amplitude or body sway path) have been presented in balance studies as a way to describe postural stability. In addition to this data, there are also recommendations to evaluate other parameters, such as body movement velocities and accelerations, that may give more information of postural control mechanism, e.g. the function and responds of the proprioceptive and vestibular systems (Toppila & Pyykkö 2000).

In the present study, during two-leg stance with the eyes open, the head was shown to have the highest maximal velocity and acceleration, while the velocities and accelerations of the other measured body segments did not differ significantly from each other. It seems that the different segments of the human body are continually "working" to stabilize the body with its movements instead of merely controlling the static body position against external forces. In a chaotic model of postural stability, sway velocity is assumed to arise from dynamic muscle forces used to correct body sway oscillation (e.g. Toppila & Pyykkö 2000).

As in balance studies with a platform (Juntunen *et al.* 1987, Matheson *et al.* 1999, Hughes *et al.* 1996), the effect of visual stabilization on balance control was also seen in this study. Both the maximal velocity and the acceleration values of separate body segments increased when the study subjects stood on a bare surface without visual control. Hytönen *et al.* (1993) also found a remarkable increase in body sway velocity in eyes-closed conditions during stance, especially among their subjects aged 76-90 years. In the present study, the head turned out to have the highest maximal velocity, but the percentage change between the velocities in visual and non-visual conditions was equally remarkable in the knees and the navel. There was also a marked difference in the maximal acceleration of the knees between the test conditions. This could mean that, although "the stability" of the head, knees and trunk decreases in non-visual conditions, their role in controlling body balance increases. Especially the rapid movements of the knees may become more important in controlling the standing position with the eyes closed. In previous studies, elderly subjects have been found unable to produce fast movement corrections (Lord *et al.* 1991a), and their knee and ankle weakness appears to be a risk factor for falling (Whipple *et al.* 1987). The results of this study suggest that more training in the control of these body movements should be provided in various conditions (= using different information sources) to help the elderly to balance their standing position more effectively.

The assumption that an eyes-closed test performance will always yield values higher than those obtained with the eyes open was not fully verified, and a detailed evaluation of values showed that some individuals (20-30) had a lower maximal velocity or acceleration value in some body segments in non-visual conditions. For example, the mean maximal head acceleration in group 10 was somewhat lower in standing with the eyes closed than with the eyes open. This finding is surprising, because the visual information in postural control was eliminated. However, these differences may be explained by the assumption that individuals use different information sources in balancing their body position. For example, in a platform study using the Romberg test (Black *et al.* 1982), the stabilising influence of vision was not very marked in all subjects. It is also possible that subjects' abilities to utilise different somatosensory information had some effects on the ankle velocities and accelerations in the present study. The results indicate that, presumably, the role of ankle movements was similar in controlling standing balance when the eyes were open and closed. This can be partly explained by the observation that the functional stretch reflex (Nashner 1976) was similarly stimulated in both test conditions. It is also possible that the system noise present in this study may have obscured the small ankle movements to some extent. However, the effect of system noise was equal on all markers. Nevertheless, it has been previously reported that, on two-leg stance and in stable conditions, proprioceptive cues may be more important than other perceptual information (Nashner 1989), especially in the absence of visual information (Koceja *et al.* 1999). Yet, it must be admitted that the postural control mechanism is highly context-dependent, and vestibular influx, for example, cannot be ignored (Allum & Keshner 1986).

No differences between the groups in either balance test were found, and the velocities and accelerations of the measured body movements seemed to be similar at all ages from 31 to 80 years during stable stance. Although each group consisted of only ten subjects, possible distinct differences between the groups would have appeared. Hytönen *et al.* (1993) found in their study with a platform that equilibrium with the eyes open was most stable at the ages of 30 to 60 years, and sway velocity was represented by a U-shape curve with children and old people swaying most. There have also been other studies with similar results (Sihvonen *et al.* 1998, Matheson *et al.* 1999). Any comparison of the results of this study with those obtained with a platform must be made with caution, because the system used here analyses balance from a different perspective. Information gathered by both systems is related to functional balance abilities, but postural sway measurements have been reported to be more likely to identify sensorimotor deficits (Hughes *et al.* 1996), whereas motion analysis introduces the values of separate movements in general or the values of joints or joint angles (Aramaki *et al.* 2001) in a standing position. In the present study, it would have been better if it had been possible to use mean values (e.g. mean velocity or RMS acceleration) instead of maximal values, as reported in the other studies with accelerometry-based methods (e.g. Moe-Nilssen 1998, Korpelainen *et al.* 2000). In the future studies, more attention should be paid to the methods of analysis and calculation.

Still, different opinions exist concerning the mechanism by which the postural system is controlled. Some authors believe that subjects who have weak muscle power improve their stability rather by limiting their sway than increasing their sway, and slow response times would therefore give a better description of instability in standing (Hughes *et al.* 1996). Generally, increased postural sway values, such as high sway velocity, have been thought to indicate poor postural control (e.g. Hytönen *et al.* 1993). In the present study, the effect of visual information on velocities and accelerations of balancing movements was found to be significant. However, it seems that parameters used in human posture evaluation can indicate both the "instability" and the required correction forces. Therefore, based on the results of this study, it is suggested that exercises recognizing movement speed and the special role of each body part in balance should be developed and used in balance training. Besides, balance training should be provided in various conditions, to develop a better capacity of balanced standing.

7 Summary

1. The results indicate that the motion analysis system can detail both differences in the movements of separate body segments and differences in movement directions. Thus, motion analysis introduces a new approach to balance measurements. More reproducible measurement results are obtained with total movement values than with maximal amplitude values. Total movement values should be applied to longitudinal balance studies, whereas maximal amplitudes may have value in the evaluation of momentary movements. In a comparison of the parameters used, motion analysis and platform seemed to reflect the same aspect of balance, although the methods of measurement are different. Movement analyses of separate body segments during stance provide additional information about the postural control mechanism. However, because the procedures are laborious, the methods of analysis and calculation should be developed further.
2. During standing on two legs with the eyes open, there was a significant difference in the maximal anterior-posterior head amplitude and maximal vertical navel amplitude between the age groups, but the results did not show any other statistically significant differences between the balancing movements of the separate body segments of the groups or between the balance measurement values of men and women in standing on two legs with the eyes open and closed. In standing on two legs with the eyes closed, all the measured body parts except the ankles had significantly higher maximal velocity and acceleration values than in standing with the eyes open. Body characteristics seemed to have slight but significant effects on the variations of body balancing movement amplitudes in standing on two legs with the eyes open, but almost none in the eyes-closed conditions. Healthy female and male subjects controlled their steady standing position with quite similar ranges of body adjustment. The effect of visual information in balancing the body movements is essential.
3. The results of this study showed that the normal position of standing on two legs is adequate and “easy” enough for assessing and practising postural balance at all ages. To obtain more versatile information on postural control mechanisms, motion analysis could be used combined with other methods. The potential effect of body characteristics, e.g. body part distance values, foot and footwear characteristics, should be considered in both balance assessment and practice. Standing balance and

posture should also be evaluated and practised in more difficult circumstances and by using variable information sources, because the potential changes are not necessarily detected in easier balance performances. Exercises that also recognise different movement speeds and the special roles of each body part in balance should be developed.

8 Conclusions

The following main conclusions can be presented:

1. Motion analysis of separate body segments during stance provides additional information of the postural control mechanism.
2. To obtain valid results, different values should be used in longitudinal balance studies and in the evaluation of momentary balancing movements.
3. In a sample of subjects aged 31 to 80 years, age seemed to have no clear effect on balancing movements during standing on two legs with the eyes open and closed, except on the maximal anterior-posterior head amplitude and the maximal vertical navel amplitude in the eyes-open conditions. There were no significant differences between the genders in balance measurement values during standing on two legs with the eyes open and closed.
   Body characteristics had slight but significant effects on the variations of body balancing movements in stance on two legs with the eyes open. In the male group the maximal movement amplitudes in the lateral direction were mainly explained by body height and in the female group by foot length. Bigger foot size resulted in somewhat greater maximal movement amplitudes in the anterior-posterior direction in both gender groups.
4. The effect of visual information on the velocities and accelerations of balancing body movements was significant.
   In further studies, balance should be evaluated in more challenging circumstances and motion analysis could be used combined with other methods to achieve more versatile information of postural control mechanisms. The methods of analysing and calculating motion analysis data need to be further developed.

9 References

Alexander NB (1996) Using technology-based techniques to assess postural control and gait in older adults. Clin Geriatr Med 12: 725-44.

Alexander NB, Shephard N, Gu MJ & Schultz A (1992) Postural control in young and elderly adults when stance is perturbed: kinematics. J Gerontol 47: M79-87.

Allum JHJ & Keshner EA (1986) Vestibular and proprioceptive control of sway stabilization. In: Bles W, Brandt T (eds.). Disorders of posture and gait. Amsterdam. Elsevier. pp.19-39.

Altman DG & Bland JM (1983) Measurement in medicine: the analysis of method comparison studies. The Statistician 32: 307-317.

Aniasson A, Zetterberg C, Hedberg M & Henriksson KG (1984) Impaired muscle function with aging. A background factor in the incidence of fractures of the proximal end of the femur. Clin Ortop Rel Res 191: 193-201.

Aramaki Y, Nozaki D, Masani K, Sato T, Nakazawa K & Yano H (2001) Reciprocal angular acceleration of the ankle and hip joints during quit standing humans. Exp Brain Res 136: 463-473.

Baloh RW, Jacobson KM & Socotch TM (1993) The effect of ageing on visual-vestibulo-ocular responses. Exp Brain Res 95: 509-516.

Baloh RW, Fife TD, Zwerling L, Socotch T, Jacobson K, Bell T & Beykirch K (1994) Comparison of static and dynamic posturography in young and older normal people. J Am Geriatr Soc 42: 405-412.

Baloh RW, Jacobson KM, Beykirch K & Honrubia V (1998) Static and dynamic posturography in patients with vestibular and cerebellar lesions. Arch Neurol 55: 649-654.

Barin K & Stockwell CW (1985) A mathematical model of human postural control in sagittal plane, In: Igarashi, Black (eds.): Vestibular and visual control on posture and locomotor equilibrium. Basel, Karger. pp. 29-33.

Becker EL, Butterfield WJH, Harwey MC, Gehee A, Heptinstall RH & Thomas L (eds) (1986) International dictionary of medicine and biology. A Wiley Medical Publication. John Wiley & Sons. New York. USA.

Benvenuti F, Mecacci R, Gineprari I, Bandinelli S, Benvenuti E, Ferrucci L, Baroni A, Rabuffetti M, Hallett M, Dambrosia JM & Stanhope SJ (1999) Kinematic characteristics of standing disequilibrium: reliability and validity of a posturographic protocol. Arch Phys Med Rehabil 80: 278-87.

Berg KO, Wood-Dauphinee S, Williams JI & Gayton D (1989) Measuring balance in the elderly: preliminary development of an instrument. Physiotherapy Canada 41: 304- 311.

Berg KO, Wood-Dauphinee SL, Williams JI & Maki BE (1992) Measuring balance in the elderly: validation of an instrument. Can J Public Health 83: S7-11.

Black, FO, Wall C, Rockette H & Kitch R (1982) Normal subject postural sway during the Romberg Test. Am J Otolaryngol 3: 309–318.

Black FO & Nashner LM (1985) Postural control in four classes of vestibular abnormalities. In: Igarashi, Black (eds.): Vestibular and visual control on posture and locomotor equilibrium. Basel, Karger pp. 271-277.

Black FO, Angel CR, Pesznecker SC & Gianna C (2000) Outcome analysis of individualized vestibular rehabilitation protocols. Am J Otol 21: 543-551.

Bland JM & Altman DG (1986) Statistical methods for assessing agreement between two methods of clinical measurement. Lancet 1: 307-310.

Bohannon RW & Leary KM (1995) Standing balance and function over the course of acute rehabilitation. Arch Phys Med Rehabil 76: 994-996.

Bohannon RW, Larkin PA, Cook AC, Gear J & Singer J (1984) Decrease in timed balance test scores with aging. Phys Ther 64: 1067–1070.

Brandt T, Paulus WM & Straube A (1985) Visual acuity, visual field and visual scene characteristics affect postural balance. In: Igarashi M, Black FO (eds.):Vestibular and visual control in posture and locomotor equilibrium. Basel, New York: Karger. pp. 93-98.

Brandt T, Paulus W & Straube A. Vision and posture. (1986) In: Bles W, Brandt T (eds.) Disorders of posture and gait. Amsterdam. Elsevier. pp.157-175.

Briggs R, Gossman M, Birch R, Drews J & Shaddeau S (1989) Balance performance among noninstitutionalized elderly women. Phys Ther 69: 748-756.

Brocklehurst JC, Robertson D & James-Groom P (1982) Clinical correlates of sway in old age-sensory modalities. Age Ageing 11: 1-10.

Bruton A, Convay JH & Holgate ST (2000) Reliability: What is it, and how is it measured? Physiotherapy 86: 94-99.

Cappello A, Cappozzo A, Leo T & Paul JP (organizators)(1995) 3-D reconstruction of human motion. Theoretical and practical aspects. Notes of a tutorial held in Jyväskylä (Finland) prior to the XV Congress of the international society of biomechanics. July 2.

Carr JH & Shepherd RB (1982) Motor relearning programme for stroke.1st edition. London: William Heineman Medical Books Ltd.

Carr JH & Shepherd RB (1987) Movement science foundation for physical therapy in rehabilitation. Aspen Publishers, Inc. Maryland. pp. 31-49.

Carr JH & Shepherd RB (1990) A motor learning model for rehabilitation of the movement disabled. In L. Ada and C. Canning (eds.) Key Issues in Neurological Physiotherapy. Melksham: Redwood Press Ltd. pp. 1–24.

Cavanagh PR & Rodgers MM (1987) The arch index: a useful measure from footprints. J Biomech 20: 547-551.

Cavanagh P, Derr J & Ulbrecht J (1992) Problems with gait and posture in neuropathic patients with insulin-dependent diabetes mellitus. Diab Med 9: 469.

Chang H & Krebs DE (1999) Dynamic balance control in elders: Gait initiation assessment as a screening tool. Arch Phys Med Rehabil 80: 490-494.

Cordo PJ & Nashner LM (1982) Properties of postural adjustments associated with rapid arm movements. J Neurophysiol 47: 287-302.

Currier DP (1981) Elements of research in physical therapy. Williams & Williams. Baltimore.

Davis JW, Ross PD, Nevitt MC & Wasnich RD (1999) Risk factors for falls and for serious injuries on falling among older Japanese women in Havaii. J Am Geriatr Soc 47: 792-798.

Delp SL, Loan P, Hoy MG, Zajac FE, Topp EL & Rosen JM (1990) An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. IEEE Trans Biomed Eng 37: 757-767.

Diener HC & Dichgans J (1986) Long loop reflexes and posture. In: Bles W, Brandt T (eds.). Disorders of posture and gait. Amsterdam. Elsevier. pp.41-44.

Diener HC, Horak FB & Nashner LM (1988) Influence of stimulus parameters on human postural responses. J Neurophysiol 59:1888-1905.

Di Fabio RP & Emasithi A (1997) Aging and the mechanisms underlying head and postural control during voluntary motion. Phys Ther 77: 458–475.

Di Fabio RP & Seay R (1997) Use of the "Fast evaluation of mobility, balance and fear" in elderly community dwellers: validity and reliability. Phys Ther 77: 904-917.

DiZio P & Lackner JR (1990) Age differences in oculomotor responses to step changes in body velocity and visual surround velocity. J Gerontol 45: M89-94.
Duncan P, Weiner D, Chandler J & Studenski S (1990) Functional Reach: A new clinical measure of balance. J Gerontol 45: M192-197.
Ekhdahl C, Jarnlo G & Andersson S (1989) Standing balance in healthy subjects. Scand J Rehabil Med 21:187-195
Enbom H (1990) Vestibular and somatosensory contribution to postural control. Thesis, University of Lund. Lund.
Enoka RM (1994) Neuromechanical basis of kinesiology. Second edition. Human kinetics. Champaign.IL.
Era P & Heikkinen E (1985) Postural sway during standing and unexpected disturbance of balance in random samples of men of different ages. J Gerontol 40: 287-295.
Era P (1988) Posture control in the elderly. Int J Technol Aging 1:166-179.
Era P, Lahtinen U, Harri-Lehtonen O & Konttinen A (1991) The effect of conventional movement training and trampoline training on balance and gait in chronic hemiplegic patients. Physiotherapy Theory and Practise 7: 223-230.
Era P, Schroll M, Ytting H, Gause-Nilsson I, Heikkinen E & Steen B (1996) Postural balance and its sensory – motor correlates in 75-year-old men and women: A cross-national comparative study. J Gerontol: Medical Sciences 51A: M53-M63.
Era P (1997) Havaintomotoriikan ja kehon asennon hallintakyvyn muutokset vanhetessa ja liikunta In: Era P (eds.) Ikääntyminen ja liikunta. Liikunnan ja kansanterveyden julkaisuja 108. Koipijyvä Oy. Jyväskylä. pp. 49-61.
Ferrell WR, Crighton A & Sturrock RD (1992) Age-dependent changes in position sense in human proximal interphalangeal joints. Neuro Rep 3: 259-261.
Figura F, Cama G, Capranica L, Guidetti L & Pulejo C (1991) Assessment of static balance in children. J Sport Med Phys Fit 31: 235-242.
Fitzpatrick R & McCloskey DI (1994) Proprioceptive, visual and vestibular thresholds for the perception of sway during standing in humans. J Physiol 478.1:173-186.
Fleiss JL (1986) The design and analysis of clinical experiments. John Wiley & Sons. New York. pp 2-12.
Forget R & Lamarre Y (1990) Anticipatory postural adjustment in the absence of normal peripheral feedback. Brain Res 508: 176-179.
Frey CC, Thompson F, Smith J, Sanders M & Horstman H (1993) American orthopedic foot and ankle society women's shoe survey. Foot Ankle 14: 78-81.
Furnee EH (1991) Opto-electric movement measurement systems: aspects of data acquisition, signal processing and performance. In: Boenick U, Nader M, Mainka C (eds.) Gangbildanalyse-Stand der Messtechnik und Bedeutung fur die Orthopädie-Technik. Duderstadt: Mecke Druck. pp. 112-129.
Gabell A, Simons MA & Nayak U (1985) Falls in the healthy elderly: predisposing causes. Ergonomics 28: 965-975.
Gahery Y & Massion J (1981) Coordination between posture and movement, TINS 4: 199-202.
Gatev P, Thomas S, KeppleT & Hallett M (1999) Feedforward ankle strategy of balance during quiet stance in adults. J Physiol 514: 915-928.
Gauffin H, Areblad M & Tropp H. (1993) Three-dimensional analysis of the talocrural and subtalar joints in single-limb stance. Clin Biomech 8: 307-314.
Geurts A, Nienhuis B & Mulder T, (1993) Intrasubject variability of selected force-platform parameters in the quantification of postural control. Arch Phys Med Rehabil 74: 1144-1150.
Gianoli G, McWilliams S, Soileau J & Belafsky P (2000) Posturographic performance in patients with potential for secondary gain. Otolaryngol Head Neck Surg 122: 11-18.
Goldie PA, Bach TM & Evans OM (1989) Force platform measures for evaluating postural control: reliability and validity. Arch Phys Med Rehabil 70: 510-517.
Haas BM & Burden AM (2000) Validity of weight distribution and sway measurements of the balance performance monitor. Physiotherapy Research International 5: 19-32.
Hasselkus B & Shambes G (1975) Ageing and postural sway in women. J Gerontol. 30: 661-667.

Harrison EL, Duenkel N, Dunlop R & Russell G (1994) Evaluation of single-leg standing following anterior cruciate ligament surgery and rehabilitation. Phys Ther 74: 245-252.

Hatze H (1980) Neuromusculoskeletal control systems modeling – a critical survey of recent developments. IEEE Trans Autom. Control. 25: 375-385.

Hawes MR, Nachbauer W, Sovak D & Nigg B (1992) Footprint parameters as a measure of arch height Foot Ankle 13: 22-26.

Hay L (1995) Posture control and muscle proprioception in the elderly. In: Ferrandez AM, Teasdale N (eds) Changes in sensory-motor behaviour in aging. (Advances in Psychology) Elsevier Science. Amsterdam.

Hay L, Bard C, Fleury M & Teasdale N (1996) Availability of visual and proprioceptive afferent messages and postural control in elderly subjects. Exp Brain Res 108: 129-139.

Herman H & Bottomly J (1992) Anatomical and biomechanical considerations of the elder foot. Top Geriatr Rehab 7: 1.

Hinchcliffe R (1983) Epidemiology of balance disorders in the elderly. In: Hinchcliffe R: Hearing and balance in the elderly.1st edition. Singapore.

Hill K, Carroll S, Kalogeropoulos A & Schwarz J (1995) Retest reliability of centre of pressure measures of standing balance in healthy older women. Australian Journal on Ageing 14: 76-80.

Hill K, Bernhardt J, McGann A, Maltese D & Berkovits D (1996) A new test of dynamic standing balance for stroke patients: Reliability, validity and comparison with healthy elderly. Physiotherapy Canada I Fall 257-262.

Hirasaki E, Kubo T, Nozawa S, Matano S & Matsunaga T (1993) Analysis of head and body movements of elderly people during locomotion. Acta Otolaryngol (Stockh) Suppl. 501: 25-30.

Horak FB, Esselman P, Anderson ME & Lynch MK (1984) The effects of movement velocity, mass displaced and task certainty on associated postural adjustments made by normal and hemiplegic individuals. J Neurol, Neurosurg, Psychiatr 47: 1020-1028.

Horak FB & Nashner LM (1986) Central programming of postural movements: Adaptation to altered support-surface configurations. J Neurophysiol 55: 1369-1381.

Horak FB (1997) Clinical assessment of balance disorders. Gait Posture 6: 76-84.

Hu M & Woollacott M (1994a) Multisensory training of standing balance in older adults: I. Postural stability and one-leg stance balance. J Gerontol 49: M 52-M61.

Hu M & Woollacott M (1994b) Multisensory training of standing balance in older adults: II. Kinematic and electromyographic postural responses. J Gerontol 49: M62-M71.

Hughes MA, Duncan PW, Rose, DK, Chandler JM & Studenski SA (1996) The relationship of postural sway to sensorimotor function, functional performance, and disability in the elderly. Arch Phys Med Rehabil 77: 567-572.

Hutton WC & Dhanendran M (1981) The mechanics of normal and hallux valgus feet: a quantitative study. Clin Orthopedics 157: 7-13.

Hytönen M, Pyykkö I, Aalto H & Starck J (1993) Postural control and age. Acta Otolaryngol (Stockh) 113: 119–122.

Inglin B & Woollacott M (1988) Age-related changes in anticipatory postural adjustments associated with arm movements. J Gerontol 43: M105-113.

Ishizaki H, Pyykkö I, Aalto H & Starck J (1991) Repeatability and effect of instruction of body sway. Acta Otolaryngol (Stockh). Suppl. 481: 598-592.

Iverson BD, Gossman MR, Shaddeau SA & Turner Jr ME (1990) Balance performance, force production, and activity levels in noninstitutionalized men 60 to 90 years of age. Phys Ther 70: 348-355.

Jóbbagy A, Gyöngy L, Martin F & Abraham G (1995) Video signal processing with high resolution in passive marker based motion analysis. In: Häkkinen K, Keskinen KL, Komi PV, Mero A (eds.) Book of Abstracts. XV th Congress of International Society of Biomechanics. July 2-6. University of Jyväskylä. Finland. pp. 436-437.

Johansson JO & Jarnlo GB (1991) Balance training in 70-year–old women. Physiotherapy Theory and Practise 7: 121-125.

Johansson R & Vallbo Å (1980) Spatial properties of the population of mechanoreceptive units in the glabrous skin of the human hand. Brain Res 184: 353- 366.

Johansson R, Magnusson M. & Åkesson M (1988) Identification of Postural Dynamics. IEEE Trans Biomed Eng 35: 858-869.

Johansson R & Magnusson M (1991) Human postural dynamics. Biomed Eng 18: 413-437.

Judge J, Lindsey C, Underwood M & Winsemius D (1993) Balance improvements in older women: Effects of exercise training. Phys Ther 73: 254-261.

Juntunen J, Ylikoski J, Ojala M, Matikainen E, Ylikoski M & Vaheri E (1987) Postural body sway and exposure to high-energy impulse noise. Lancet 11: 261-264.

Jäntti P (1993) Falls in elderly. Acta Universitatis Tamperensis. University of Tampere. Ser A Vol 365.

Karlsson A & Frykberg G (2000) Correlations between force plate measurements for assessment of balance. Clin Biomech 15: 365-369.

Kinney LaPier TL, Liddle S & Bain C (1997) A comparison of static and dynamic standing balance in older men versus women. Physiotherapy Canada 49: 207–213.

Koceja DM, Allway D & Earles DR (1999) Age differences in postural sway during volitional head movement. Arch Phys Med Rehabil 80:1537– 1541.

Koncek V, Pollak VA & Yu AM (1993) Analysis of stability of human upright posture based upon the measurement of the head sway. Med Prog Technol 19: 55-60.

Konradsen L, Ravn J & Sorensen A (1993) Proprioception at the ankle: the effect of anaesthetic blockade of ligament receptors. J Bone Joint Surg 75-B: 433-436.

Korpelainen J, Myllylä V, Kampman V & Leppävuori S (2000) A novel accelerometry-based method for the quantification of balance and postural sway. J Am Geriatr Soc 48: 709-710.

Kuo AD (1995) An optimal control model for analyzing human postural balance. IEEE Trans Biomed Eng 42: 87-100.

Kuo AD & Zajac FE (1993) A biomechanical analysis of muscle strength as a limiting factor in standing posture. J Biomech 26(Suppl.1): 137-150.

Latash ML (1998) Postural Control. Vision. In Human Kinetics (eds.) Neurophysiological Basis of Movement. Champaign. Illinois. USA. pp.163-194.

LeClair K & Riach C (1992) Postural stability measurement: Test duration and outcome parameters. In: Woollacott M, Horak F (eds): Posture and Gait: Control mechanism, Volume 1. University of Oregon books. pp. 400-403.

Lee DN & Aronson E (1974) Visual proprioceptive control of standing in human infants. Percep Psychophys 15: 529-532.

Lee WA, Buchanan TS & Rogers MW (1987) Effects of arm acceleration and behavioral conditions on the organization of postural adjustments during arm flexion. Exp Brain Res 66: 257-270.

Leibowitz HW, Rodemer CS & Dichgans J (1979) The independence of dynamic spatial orientation from luminance and refractive error. Percep Psychophys 25: 75-79.

Lerner LS (1996) Physics for scientists and engineers. Jones and Bartlett Publishers Inc., Sudbury. Massachusetts. USA.

Levine D, Whittle M, Beach J & Ollard P (1996) Test-retest reliability of Chattecx Balance System in the patient with hemiplegia. J Rehab Res Dev 33: 36-44.

Levy M & Smith GA (1995) Validation of MacReflex motion analysis system for three -dimensional coordinate prediction. In: Häkkinen K, Keskinen LK, Komi PV, Mero A (eds.) Book of Abstracts. XVth Congress of International Society of Biomechanics. July 2-6. University of Jyväskylä. Finland. pp.552-553.

Lord S, Clark R & Webster I (1991a) Postural stability and associated physiological factors in a population of aged persons. J Gerontol 46: M69-76.

Lord S, Clark R & Webster I (1991b) Physiological factors associated with falls in an elderly population. J Am Geriatr Soc 39:1194.

Lord S, Caplan G & Ward J (1993) Balance, reaction time and muscle strength in exercising and none-exercising older women: a pilot study. Arch Phys Med Rehabil 74: 837-839.

McComas AJ (1996) Skeletal muscle. Human Kinetics. Champaign, IL 61825-5075.

Macpherson JM, Horak FB, Dunbar DC & Dow RS (1989) Stance dependence of automatic postural adjustments in humans. Exp Brain Res 78: 557-566.

Magnusson ML, Aleksiev A, Wilder DG, Pope MH, Spratt K, Lee SH, Goel VK & Weinstein JN (1996) European Spine Society-the AcroMed Prize foe Spinal Research 1995. Unexpected load and asymmetric posture as etiologic factors in low back pain. Eur Spine J 5: 23-35.

Magnusson M, Enbom H, Johansson R & Pyykkö I (1990) Significance of pressor input from the human feet in anterior-posterior postural control. Acta Otolaryngol 110: 182-188.

Mahar RK, Kirby RL & MacLeod DA (1985) Simulated leg-length discrepancy: Its effect on mean center –of-pressure position and postural sway. Arch Phys Med Rehabil 66: 822-824.

Maki BE, Perry SD, Norrie RG & McIlroy WE (1999) Effect of facilitation of sensation from plantar foot-surface boundaries on postural stabilization in young and older adults. J Gerontol Seriaes A, Biological Sciences & Medical Sciences 54A: M 281-287.

Manchester D, Woollacott M, Zederbauer- Hylton N & Marin O (1989) Visual, vestibular and somatosensory contributions to balance control in the older adults. J Gerontol 44: M118-M127.

Man'kovskii NB, Mints AY & Lysenyuk VP (1980) Regulation of the preparatory period for complex voluntary movement in old and extreme old age. Human Physiology 6:46-50.

Mannion A & Troke M (1999) A comparison of two motion analysis devices used in the measurement of lumbar spinal mobility. Clin Biomech 14: 612-619.

Markham CH (1987) Vestibular control of muscular tone and posture. Can J Neurol Sci 14: 493-496.

Matheson AJ, Darlington CL & Smith PF (1999) Further evidence for age-related deficits in human postural function. J Vestibular Res 9: 261-264.

Mathias S, Nayak US & Isaacs B (1986) Balance in elderly patients: The "Get-up and Go" test. Arch Phys Med Rehabil 67: 387-389.

Medved V (2001) Measurement of Human Locomotion. CRC Press LLC, 2000 N.W. Corporate Blvd, Boca Raton, Florida 33431.

Menz HB (1998) Alternative techniques for the clinical assessment of foot pronation. JAPMA 88:119-129.

Menz HB & Lord SR (1999a) Footwear and postural stability in older people. JAPMA 89: 346-357.

Menz HB & Lord SR (1999b) Foot problems, functional impairment and falls in older people. JAPMA 89: 458-467.

Menz HB & Sherrington C (2000) The footwear assessment form: a reliable clinical tool to assess footwear characteristics of relevance to postural stability in older adults. Clin Rehabil 14: 657-664.

Moe-Nilssen R (1998) Test-retest reliability of trunk accelerometry during standing and walking. Arch Phys Med Rehabil 79: 1377-85.

Morasso PG & Schieppati M (1999) Can muscle stiffness alone stabilize upright standing? J Neurophysiol 83:1622-1626.

Murrell P, Cornwall MW & Doucet SK (1991) Leg-length discrepancy: effect on the amplitude of postural sway. Arch Phys Med Rehabil 72: 646-648.

Nashner LM (1976) Adapting reflexes controlling the human posture. Exp Brain Res 26:59-72.

Nashner LM (1983) Analysis of movement control in man using the movable platform. In: Motor Control Mechanism in Health and Disease, Ed. JE Desmedt, Raven Press, New York. pp. 607-619.

Nashner LM (1985) Strategies for organization of human posture. In: Igarashi, Black (eds.): Vestibular and visual control on posture and locomotor equilibrium. Basel, New York: Karger pp.1-8.

Nashner LM & McCollum G (1985) The organization of human postural movements: A formal basis and experimental synthesis. Behav Brain Sci. 8: 135-172.

Nashner LM (1989) Sensory, neuromuscular and biomechanical contributions to human balance. In: Proceedings of the American Physical Therapy Association Forum. Nashville. TN. pp. 5-12.

Nashner LM (2001) Computerized dynamic posturography. In: Joel A Goebel: Practical Management of Dizzy Patient. Lippicott Williams & Wilkins. Philadelphia pp.143-170.

Nayak USL (1987) Comparison of the Wright ataxiameter and the Kistler force platform in the measurement of sway. J Biomed Eng 10: 302-304.

Ohanian HC (1989) Physics. Second edition, expanded. Rensselear Polytechnic institute. WW Norton & Company Inc., New York.

Ojala M, Matikainen E & Juntunen J (1989) Posturography and the dizzy patient: a neurological study of 133 patients. Acta Neurol Scand 80:118–122.

Overstall PW, Exton-Smith AN, Imms FJ & Johnson AL (1977) Falls in the elderly related to postural imbalance. BMJ 1: 261-264.
Paige GD (1991) The aging vestibulo-ocular reflex (VOR) and adaptive plasticity. Acta Otolaryngol (Suppl) 481: 297-300.
Palovaara T, Sipponen E & Era P (1992) Tasapaino ja eräitä siihen liittyviä tekijöitä 75-vuotiailla miehillä ja naisilla. Gerontologia 6: 185-195.
Panzer VP, Bandinelli S & Hallett M (1995) Biomechanical assessment of quiet standing and changes associated with aging. Arch Phys Med Rehabil 76: 151-157.
Paul JP (1995) Motion analysis laboratory architecture. In: Cappello A, Cappozzo A, Leo T, Paul JP. 3-D reconstruction of human motion. Theoretical and practical aspects. Notes of a tutorial held in Jyväskylä (Finland) prior to the XV Congress of the international society of biomechanics. July 2. Chapter 5.
Paulus WM, Straube A & Brandt T (1984) Visual stabilisation of posture. Brain 107: 1143-1163.
Peterka RJ, Black FO & Schoenhoff MB (1990a) Age-related changes in human vestibulo-ocular reflexes: sinusoidal rotation and caloric tests. J Vestibular Res 1: 49-59.
Peterka RJ, Black FO & Schoenhoff MB (1990b). Age-related changes in human vestibulo-ocular reflexes: Pseudorandom rotation tests. J Vestibular Res 1: 61-71.
Pierrot-Deseilligny E, Katz R & Hultborn H (1983) Functional organization of recurrent inhibition in man: changes preceding and accompanying voluntary movements. In Desmedt JE (eds) Advances in neurology 39; Motor control mechanism in health and disease. pp.443-457.
Potvin AR, Syndulko K, Tourtellotte WW, Lemmon JA & Potvin JH (1980) Human neurologic function and ageing process. J Am Geriatr Soc 28: 1-9.
Prochazka A & Wand P (1980) Tendon organ discharge during voluntary movements in cats. J Physiol (Lond) 303: 385-390.
Pyykkö I, Aalto H, Hytönen M, Starck J, Jäntti P & Ramsay H (1988) Effect of age on postural control. In Amblard B, Berthoz A & Clarac F (eds): Posture and Gait: development, adaption and modulation. Proc 9th Int symp Postural and Gait Research. Marseille. pp 95-104.
Pyykkö I, Toppila E, Aalto H, Ishizaki H, Kentala E, Hirvonen T & Honkavaara P (2000) Determination of parameters for computing postural stability. Automedica 19: 39-62.
Quoniam C, Hay L, Roll JP & Harlay F (1995) Age effects on reflex and postural responses to proprioceptive inputs generated by tendon vibration. J Gerontol 50A: B155-B165.
Rogers MW, Kukulka CG & Soderberg GL (1992) Age-related changes in postural responses preceding rapid self-paced and reaction time arm movements. J Gerontol 47: M159- M165.
Rothwell John (1994) Control of Human Voluntary Movement. Published by Chapman & Hall. 2-6 Boundary Row. London. Pages 252-290.
Ryynänen O-P (1993) Incidence and risk factors for falling injuries among the elderly. Acta Universitatis Ouluensis Medica. D265.
Sauvage LR, Myklebust BM, Crow-Pan J, Novak S, Millington P, Hoffman MD, Hartz AJ & Rudman D (1992) A Clinical trial of strengthening and aerobic exercise to improve gait and balance in elderly male nursing home residents. Am J Phys Med Rehabil 71: 333-342.
Schmidt RA (1991) Motor learning and performance. Human Kinetics, Champaign, IL.
Schmidt RA & Lee TD (1999) Motor control and learning. A behavioral emphasis. 3rd ed. Human Kinetics, Champaign, IL.
Scott DE & Dzendolet E (1972) Quantification of sway in standing humans. Aggressologie 13B: 35-40.
Selfe J (2000) Motion analysis of an eccentric step test performed by 100 healthy subjects. Physiotherapy 86: 241-247.
Shephard NT, Telian SA, Smith-Wheelock M & Raj A (1993) Vestibular and balance rehabilitation therapy. Ann Oto Rhin Laryn 102: 198-205.
Shephard NT & Telian SA (1995) Programmatic vestibular rehabilitation. Otolaryngol Head Neck Surg 112: 173-182.
Shumway-Cook A & Horak FB (1986) Assessing the influence of sensory interaction on balance. Phys Ther 66: 1548-1550.

Shupert CL, Black FO, Horak FB & Nashner LM (1988) Coordination of head and body in response to support surface translations in normals and patients with bilaterally reduced vestibular function. In Amblard B, Berthoz A, Clarac F (eds) Posture and gait: Development, Adaptation and Modulation. New York: Elsevier Science Publishers.

Sihvonen S, Hofmann C & Era P (1998) Development of postural abilities across the life span parallels changes in muscle strength. In Conference Book: Häkkinen K (eds.) International Conference On Weightlifting And Strength Training. Nov. 10-12. Lahti. Finland. pp. 215-216.

Simoneue GG, Ulbrecht JS, Becker MB, Derr JA & Cavanagh PR (1992) Postural assessment of subjects with and without diabetic distal peripheral sensory myopathy. In: Woollacott M, Horak F (eds.) Posture and gait: control mechanisms, Volume 1. University of Oreganon Books, Portland, pp. 143-146.

Spirduso WW (1995) Physical Dimensions of Aging. Human Kinetics. Champaign. Illinois. USA.

Stelmach GE & Worringham CJ (1985) Sensimotor deficits related to postural stability. Implications for falling in the elderly. Clin Geriatr Med 1: 679-694.

Stelmach GE & Sirica A (1986) Aging and proprioception. Age 9: 99-103.

Stelmach GE, Teasdale N, Di Fabio RP & Phillips J (1989) Age-related decline in postural control mechanisms. Int J Aging Hum Dev 29: 205-223.

Stewart MG, Chen AY, Wyatt JR, Favrot S, Beinart S, Coker NJ & Jenkins HA (1999) Cost-effectiveness of diagnostic evaluation of vertigo. Laryngoscope 109: 600-605.

Stones MJ & Kozma A (1987) Balance and age in the sighted and blind. Arch Phys Med Rehabil 68: 85-89.

Studenski S, Duncan PW & Chandler J (1991) Postural responses and effector factors in persons with unexplained falls: results and methodological issues. J Am Geriatr Soc 39: 229-234.

Suomi R & Kojeca DM (1994) Postural sway patterns of normal men and women and men with mental retardation during a two-legged stance test. Arch Phys Med Rehabil 75: 205–209.

Taguchi K, Iijima M & Suzuki T (1978) Computer calculation of movement of body's center of gravity. Acta Otolaryngol 85: 420-425.

Tanaka R (1983) Reciprocal Ia inhibitory pathway in normal man and in patients with motor disorders. In: Desmedt JE (eds) Advances in neurology 39. Motor control mechanism in health and disease pp.433-441.

Taylor KD, Moltier FM, Simmons DW, Cohen W, Pavla R, Cornell DP & Hankins B (1982) An automated motion measurement system for clinical gait analysis. J Biomech 15: 505-516.

Teasdale N, Stelmach G, Breunig A & Meeuwsen H (1991) Age differences in visual sensory integration. Exp Brain Res 85: 691-696.

Tinetti ME (1986) Performance-oriented assessment of mobility problems in elderly patients. J Am Geriatr Soc 34:119-126.

Tinetti ME, Speechley M & Ginter SF (1988) Risk factors for falls among elderly persons living in the community. N Engl J Med 319: 1701-1707.

Toppila E & Pyykkö I (2000) Chaotic model of postural stability - A position and velocity dependent system? Automedica 19: 115-134.

Trevarthen CB (1968) Two mechanisms of vision in primates. Psychologische Forschung 31: 299-337.

Viitasalo M, Kampman V, Sotaniemi K, Leppavuori T, Myllylä V & Korpelainen J (2002) Analysis of sway in Parkinson's disease using a new inclinometry-based method. Movement Disorders, 17: 663-669.

Whipple RH, Wolfson LI & Amerman PM (1987) The relationship of knee and ankle weakness to falls in nursing home residents. An isokinetic study. J Am Geriatr Soc 35: 13-20.

White E & Mulley G (1989) Footcare for very elderly people: a community survey. Age Ageing 18: 275.

Wilson DJ, Smith BK, Gibson KJ, Choe BK, Gaba BC & Voelz JT (1999) Accuracy of digitization using automated and manual methods. Phys Ther 79: 558-566.

Wing AM, Clapp S & Burgess-Limerick R (1995) Standing stability in the frontal plane determined by lateral forces applied to the hip. Gait Posture 3: 38-42.

Winter DA (1990) Biomechanics and motor control of human movement. Second Edition. Wiley-Interscience. John Wiley & Sons Inc. New York. USA.

Winter DA (1995) Human balance and posture control during standing and walking. Gait Posture 3: 193-214.

Winter DA, Patla AE & Frank JS (1990) Assessment of balance control in humans. Medical Progress through Technology 16: 31-51.

Winter DA, Patla AE, Prince F, Ishac M & Gielo-Perczak K (1998) Stiffness control of balance in quiet standing. J Neurophysiol 80:1211-1221.

Wolfson L, Whipple R & Amerman P (1986) Stressing the postural response. J Am Geriatr Soc 34: 845-850.

Woollacott M, Shumway-Cook A & Nashner L (1986) Aging and postural control: Changes in sensory organization in muscular coordination. Int J Aging Hum Dev 23: 97-114.

Woollacott MH, von Hosten C & Rosblad B (1988) Relation between muscle response onset and body segmental movements during perturbations in humans. Exp Brain Res 72: 593-604.

Woollacott MH.& Shumway-Cook A (1990) Changes in posture control across the life span- A systems approach. Phys Ther 70: 799-807.

Woollacott & Manchester (1993) Anticipatory postural adjustments in older adults: are changes in response characteristics due to changes in strategy? J Gerontol 48: M64-70.

Woollacott MH, Moore S & Hu MH (1993) Improvements in balance in the elderly through training sensory organization abilities. In book: Stelmach, GE, Homberg V. Sensimotor impairment in elderly. Kluwer Academic Publishers. Dordrecht. pp. 377-392.

Woollacott M & Shumway-Cook A (1996) Concepts and methods for assessing postural instability. J Aging and Physical Activity 4: 214-233.

Wootton R, Bryson E, Elsasser U, Freeman H, Green JR, Hesp R, Hudson EA, Klenerman L, Smith T & Zanelli J (1982) Risk factors for fractured neck of femur in the elderly. Age and Ageing 11:160-168.

Wu G (2001) Head movement during sudden base translations as a measure of risks for falls in the elderly. Clin Biomech 16: 199-206.

Yoshida K, Iwakura H & Inoue F (1983) Motion analysis in the movement of standing up from and sitting down on a chair. Scand J Rehab Med 15: 133-140.

You J, Chou Y-L, Lin C-J & Su F-C (2000) Effect of slip on movement of body center of mass relative to base of support. Clin Biomech 16: 167-173.

Zattara M & Bouisset S (1988) Posturo-kinetic organisation during the early phase of voluntary upper limb movement. 1 Normal subjects. J Neurol Neurosurg Psychiatr 51: 956-965.

Appendix

Table 1. Maximal movement amplitudes of the body segments in the lateral direction by age and gender during standing with the eyes open.

Body part	Gender	31 – 40 yr.		41 – 50 yr.		51 – 60 yr.		61 – 70 yr.		71 – 80 yr.	
		Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)
Forehead	Women	10.5	5.9	9.5	3.8	13.5	4.5	10.6	5.0	10.9	3.7
	Men	9.9	5.3	9.5	4.1	11.1	3.6	10.7	6.2	10.2	3.0
Navel	Women	6.1	4.4	5.9	2.3	8.6	3.9	7.8	3.8	7.3	3.2
	Men	7.2	4.6	5.9	3.8	8.0	4.2	5.1	2.3	6.5	3.3
Knee	Women	4.1	3.0	3.2	1.5	4.7	2.2	4.0	1.8	4.0	1.8
right	Men	5.0	3.3	3.1	2.1	4.5	2.4	3.4	2.0	3.9	1.5
Knee	Women	3.3	2.5	2.9	1.5	4.7	2.3	3.9	1.6	4.0	1.9
left	Men	3.9	2.4	3.3	2.0	4.4	2.2	2.8	1.6	3.8	1.4
Ankle	Women	0.8	0.4	0.9	0.2	1.2	0.3	0.9	0.5	0.9	0.3
right	Men	1.2	1.0	0.9	0.3	0.9	0.6	0.8	0.4	0.9	0.3
Ankle	Women	0.8	0.5	0.9	0.4	1.1	0.3	1.1	0.4	0.9	0.4
left	Men	0.9	0.4	0.9	0.4	1.0	0.5	0.6	0.2	1.0	0.5

Table 2. Maximal movement amplitudes of the body segments in the lateral direction by age and gender during standing with the eyes closed.

Body part	Gender	31 – 40 yr.		41 – 50 yr.		51 – 60 yr.		61 – 70 yr.		71 – 80 yr.	
		Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)
Forehead	Women	12.5	7.1	9.8	5.3	14.2	11.3	15.2	5.4	11.5	3.1
	Men	11.8	3.6	8.8	3.6	16.3	8.8	11.6	5.7	10.3	4.0
Navel	Women	9.0	6.4	7.1	4.2	9.1	8.1	10.4	4.7	7.5	3.7
	Men	8.3	1.3	5.9	3.8	12.8	8.0	7.5	3.7	7.1	2.7
Knee	Women	5.8	3.3	3.9	2.3	5.1	5.1	5.5	2.4	4.2	1.6
right	Men	4.5	1.3	3.3	2.0	6.6	4.1	4.3	2.8	4.0	2.3
Knee	Women	5.3	3.3	4.0	2.3	5.3	5.4	5.4	2.2	4.0	1.7
left	Men	4.5	1.2	3.0	1.8	6.7	4.4	4.2	2.4	4.1	2.2
Ankle	Women	1.5	0.7	1.1	0.7	1.1	0.6	1.1	0.3	1.0	0.3
right	Men	1.1	0.6	1.0	0.4	1.6	1.1	1.0	0.4	1.1	0.5
Ankle	Women	1.0	0.6	1.1	0.7	1.4	1.0	1.1	0.4	1.2	0.6
left	Men	1.0	0.4	0.8	0.3	2.0	1.9	1.1	0.7	1.2	0.7

Table 3. Maximal movement amplitudes of the body segments in the ant-post-direction by age and gender during standing with the eyes open.

Body part	Gender	31 – 40 yr.		41 – 50 yr.		51 – 60 yr.		61 – 70 yr.		71 – 80 yr.	
		Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)
Forehead	Women	15.8	4.4	14.6	5.8	20.5	4.2	19.4	6.7	24.1	10.7
	Men	16.5	5.5	16.1	7.5	20.9	6.9	20.8	7.8	25.7	7.6
Navel	Women	8.8	3.3	8.7	3.0	12.7	2.9	13.2	2.8	14.6	8.3
	Men	11.8	5.7	12.5	7.7	11.6	4.6	11.4	4.7	14.8	5.0
Knee right	Women	5.8	5.2	4.2	1.3	5.7	1.8	5.5	1.9	6.1	3.3
	Men	5.3	2.3	5.8	3.4	5.5	1.9	6.0	3.4	6.2	2.6
Knee left	Women	4.7	2.7	4.5	1.8	5.7	1.4	5.4	2.0	6.5	2.8
	Men	5.3	2.2	6.1	4.1	5.4	2.0	5.3	2.5	6.5	1.8
Ankle right	Women	1.3	0.5	1.5	0.6	1.9	1.2	1.9	0.8	2.1	0.8
	Men	1.6	0.8	1.7	0.3	1.9	0.8	1.5	0.6	1.8	0.7
Ankle left	Women	1.2	0.2	2.0	1.0	1.7	0.6	1.8	0.8	1.9	1.1
	Men	2.0	1.0	1.7	0.7	1.8	0.8	1.2	0.5	1.7	0.6

Table 4. Maximal movement amplitudes of the body segments in the ant-post-direction by age and gender during standing with the eyes closed.

Body part	Gender	31 – 40 yr.		41 – 50 yr.		51 – 60 yr.		61 – 70 yr.		71 – 80 yr.	
		Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)
Forehead	Women	20.4	6.6	20.8	7.2	34.3	20.4	26.6	11.7	31.8	14.9
	Men	25.3	7.9	25.1	9.1	32.3	12.9	27.3	9.5	28.9	8.9
Navel	Women	12.4	5.4	12.9	5.3	16.8	9.8	15.6	7.4	17.2	9.4
	Men	15.8	5.3	15.1	6.0	18.0	5.5	15.5	6.4	14.0	3.6
Knee right	Women	6.8	3.4	5.1	2.1	7.2	4.3	6.2	2.4	6.6	2.8
	Men	6.7	2.5	6.4	2.7	8.0	3.5	7.7	3.6	5.4	1.8
Knee left	Women	6.7	2.9	5.6	2.9	7.4	4.0	5.8	2.4	6.8	3.5
	Men	6.9	2.4	6.4	2.2	7.7	3.5	7.4	2.9	5.3	1.7
Ankle right	Women	1.9	0.6	2.0	1.5	1.9	1.1	2.2	0.7	2.3	0.9
	Men	1.8	0.7	2.0	0.6	2.6	1.8	1.7	0.8	2.2	0.9
Ankle left	Women	1.6	0.8	2.1	1.1	2.0	0.5	2.2	1.0	2.5	1.8
	Men	1.7	1.1	1.8	0.5	2.5	1.6	1.7	0.7	1.6	0.6

Table 5. Maximal movement amplitudes of the body segments in the vertical direction by age and gender group during standing with the eyes open.

Body part	Gender	31 – 40 yr.		41 – 50 yr.		51 – 60 yr.		61 – 70 yr.		71 – 80 yr.	
		Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)
Fore-	Women	3.5	2.0	3.0	1.3	3.5	1.4	2.9	0.7	3.8	2.2
head	Men	3.5	1.1	3.5	1.7	3.2	1.2	3.7	1.4	4.1	2.0
Navel	Women	4.1	2.5	3.3	1.9	3.4	1.0	4.8	1.1	5.6	2.0
	Men	2.8	1.4	3.8	1.7	4.0	1.5	4.2	2.2	5.1	1.4
Knee	Women	1.6	1.3	1.5	1.2	2.0	1.5	2.1	0.9	1.8	1.0
right	Men	1.6	0.7	1.3	0.5	2.6	2.2	1.9	1.7	2.0	1.2
Knee	Women	1.1	0.6	1.3	0.6	1.4	0.5	1.5	0.6	3.0	2.8
left	Men	1.9	1.3	1.7	0.9	1.8	0.9	1.6	0.8	1.5	0.7
Ankle	Women	0.7	0.3	0.8	0.3	1.0	0.5	0.9	0.2	0.7	0.2
right	Men	1.0	0.8	0.8	0.3	0.8	0.3	0.7	0.2	0.8	0.3
Ankle	Women	0.7	0.4	0.9	0.4	0.9	0.4	0.8	0.3	0.8	0.4
left	Men	1.2	1.0	0.7	0.2	0.8	0.2	0.7	0.2	0.8	0.4

Table 6. Maximal movement amplitudes of the body segments in the vertical direction by age and gender during standing with the eyes closed.

Body part	Gender	31 – 40 yr.		41 – 50 yr.		51 – 60 yr.		61 – 70 yr.		71 – 80 yr.	
		Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)	Mean (mm)	SD (mm)
Fore-	Women	4.2	1.6	3.4	1.2	5.0	2.7	3.7	1.4	4.7	2.4
head	Men	4.3	2.2	4.0	1.1	4.4	1.6	3.6	0.7	5.6	1.8
Navel	Women	4.8	2.5	4.0	1.6	4.6	2.8	5.5	1.2	5.2	2.4
	Men	3.8	1.4	4.1	2.1	5.9	2.3	4.4	1.6	5.3	1.6
Knee	Women	2.1	1.6	1.6	1.0	2.8	2.2	3.0	2.7	2.8	1.7
right	Men	3.3	2.7	1.6	0.5	3.5	2.4	2.9	2.8	1.8	0.9
Knee	Women	2.4	2.0	2.2	2.2	1.8	0.7	2.3	1.4	2.2	0.9
left	Men	2.8	2.8	1.9	1.6	1.9	0.9	2.7	2.5	1.7	1.5
Ankle	Women	0.9	0.5	1.0	0.8	1.0	0.4	1.0	0.3	1.0	0.5
right	Men	1.0	0.6	0.9	0.4	1.1	0.7	0.8	0.4	0.9	0.4
Ankle	Women	0.9	0.7	1.0	0.7	0.9	0.5	1.1	0.5	1.3	1.1
left	Men	0.9	0.6	0.8	0.5	1.2	0.8	0.7	0.2	0.8	0.6

Table 7. Correlations (r_P) between anthropometric factors and maximal balancing movements (x, y, z) in a standing position with the eyes open. in the whole population (n=100) and in the gender groups (n=50 in both).

	Maximal balancing movements											
Anthropometric	Head			Navel			Knee			Ankle		
measurements	Lateral	Ant-post	Vertical	Lateral	Ant-post	Vertical	Lateral	Ant-post	Vertical	Lateral	Ant-post	Vertical
BMI (mass/ height^2)												
All	0.04	0.10	-0.05	0.02	-0.04	0.14	0.01	0.08	0.07	0.15	0.27**	0.25*
Women	-0.06	0.08	0.02	-0.02	-0.02	0.21	-0.01	0.09	0.07	0.16	0.32*	0.37**
Men	0.18	0.14	-0.14	0.06	0.02	0.02	0.03	0.06	0.07	0.16	0.17	0.16
Height												
All	0.10	-0.03	0.12	0.19	0.09	-0.17	0.24*	0.22*	0.06	0.17	-0.11	0.06
Women	0.24	-0.13	0.12	0.23	-0.10	-0.21	0.40**	0.26	0.07	0.19	-0.15	0.03
Men	0.20	-0.09	0.08	0.45**	0.17	-0.14	0.30**	0.26	0.06	0.28	0.02	0.14
Hip-ground distance												
All	0.13	-0.02	0.11	0.18	0.10	-0.21*	0.20*	0.24*	0.04	0.12	-0.08	0.05
Women	0.30*	-0.09	0.02	0.23	-0.10	-0.24	0.35	0.27	0.07	0.15	-0.18	-0.05
Men	0.15	-0.12	0.08	0.30*	0.13	-0.19	0.19	0.20	- 0.01	0.16	0.05	0.13
Knee-ground distance												
All	0.09	0.01	0.13	0.16	0.18	-0.11	0.17	0.26**	0.06	0.10	-0.04	0.01
Women	0.25	-0.11	0.13	0.20	-0.10	-0.17	0.32*	.24	0.06	0.12	-0.11	-0.01
Men	0.13	-0.06	0.05	0.35*	0.27	-0.03	0.18	0.31*	0.03	0.16	0.06	0.01
Foot width												
All	-0.02	0.10	- 0.03	0.03	0.14	- 0.11	0.10	0.21*	0.19	0.11	0.06	0.06
Women	-0.19	0.16	- 0.05	0.09	0.20	0.20	0.08	0.29*	0.31*	0.19	0.13	0.20
Men	0.19	-0.07	- 0.19	0.10	0.01	0.01	0.16	0.12	0.13	0.13	0.09	0.01
Heel width												
All	0.03	0.24*	0.08	0.02	0.23*	-0.02	0.03	0.23*	0.10	0.07	0.09	0.02
Women	-0.22	0.07	- 0.08	- 0.02	0.13	-0.06	0.05	0.18	0.08	0.20	0.26	0.25
Men	0.13	0.33*	0.12	0.12	0.24*	0.07	0.03	0.25	0.10	0.01	-0.02	-0.10
Foot length												
All	0.13	0.02	0.04	0.18	0.11	-0.10	0.21*	0.28**	0.15	0.12	0.04	-0.01
Women	0.25	-0.18	-0.04	0.33*	-0.09	-0.06	0.45**	0.34*	0.12	0.32*	0.14	0.09
Men	0.25	-0.01	-0.06	0.32*	0.12	-0.07	0.20	0.30*	0.20	0.10	0.04	-0.07

Pearson correlation coefficients significant * at 0.05 level, ** at 0.01 level.