Effects of healthy aging on balance
a quantitative analysis of clinical tests

Erika Jonsson
To me, myself and mom.

“Det bästa är att veta vad man letar efter innan man börjar leta efter det”

Nalle Puh

A.A. Milne (1882-1956)
Abstract

Introduction Physical therapists employ various tests to assess balance in elderly individuals, but few of these provide adequate information concerning the aspect of balance or postural control that is being measured or the reason for any decline in performance. Furthermore, tools for assessment of balance should serve as a basis for adequate intervention. In connection with attempts to understand the effects of healthy aging on balance, the first step is to define changes in postural control mechanisms that may result in deficient balance or a fall in elderly individuals.

Objectives The aim of the present thesis work was to apply and evaluate three clinical tests – i.e., functional reach (FR), one-leg stance (OLS) and tandem stance (TS) – to determine the effects of healthy aging on balance. Identification of exactly what is being measured in healthy young and elderly adults by these tests was an additional major goal.

Methods Thirty-three healthy elderly individuals (65-80 years of age) and 30 healthy 25-40-year-old adults performed FR, OLS and TS while standing on force plates, which allowed forces, movement and muscle activity to be monitored in parallel with the clinical parameters. In addition, in Study IV initiation of gait was examined in this same manner.

Results Our findings provide evidence that healthy aging has a deleterious influence on postural control. Age-related differences were observed in the amplitude of postural steadiness during performance of OLS and TS; in the timing of the force peak during the weight transfers involved in OLS, TS and initiation of gait; as well as in the distribution of body weight between the legs prior to performance of the tasks. During FR, healthy elderly subjects exhibited a low degree of correlation between stability limits and reach distance (r=0.38) and a moderate correlation between forward trunk bending and reach distance (r=0.68). In connection with OLS and TS, two phases occurred in both the elderly and younger groups: an initial dynamic phase involving a reduction in force variability during the first 3-5 seconds, followed by a static phase during which a certain constant level of force variability was maintained. Moreover, both groups supported more weight on the rear leg during TS.

Conclusion Altogether, the findings documented here indicate that the aging process in itself has a deleterious influence on balance. Healthy elderly individuals utilize a compensatory strategy in connection with FR, which results in low concurrent validity. In general, this work provides insight concerning not only how elderly and younger individuals perform, but also exactly what is being measured by these balance tests.

Clinical recommendations Balance or postural control is influenced by a number of physiological systems, the functions of which are affected by aging. When applying the FR test, compensatory mechanisms should be taken into consideration. The first 5 seconds of OLS and TS provide the most important information. Tandem stance does not involve equal weight-bearing, so both legs should be tested in both positions. Finally, balance should not only be assessed solely on the basis of task parameters (e.g., number of centimeters or seconds), but also on how the task is performed.

Key words Postural control; Balance; Aging; Clinical balance tests; Functional reach; One-leg stance; Tandem stance; Force; Electromyography; Kinematics
SAMMANFATTNING

Introduktion Sjukgymnaster använder olika tester för att mäta balans och fallrisk hos äldre personer. Många av dagens kliniska balanstester ger otillräcklig information om vilken aspekt av postural kontroll som mäts samt varför man klarar eller inte klarar ett test. Samtidigt skall balanstest ligga till grund för adekvat träning. Att förstå åldrandets inverkan på balans bör vara det första steget för att definiera förändringar i mekanismer, som påverkar postural kontroll och som kan resultera i nedsatt balans hos äldre personer och därmed fall.

Syfte Det övergripande syftet med avhandlingsarbetet var att utvärdera tre kliniska balanstest – Functional reach, enbensstående och tandemstående – för att utröna, hur friskt åldrande påverkar balansen. Syftet var också att definiera, vad testen mäter hos friska yngre och äldre personer.

Metod Stående på kraftplattor utförde trettiotre friska äldre (65-80 år) och trettio friska yngre (25-40 år) personer Functional reach, enbensstående och tandemstående. Krafter, rörelse och muskelaktivitet mättes samtidigt som kliniska mätvärden registrerades. I studie IV utfördes även initiering av gång under kraftregistrering.

Resultat Våra resultat gav stöd för att friskt åldrande påverkar den posturala kontrollen negativt. Åldersrelaterade förändringar var synliga i postural stabilitet under utförandet av enbensstående och tandemstående samt i en tidigare kraftgenerering under tyngdöverföringen till enbensstående, tandemstående och vid initiering av gång. Synliga förändringar fanns också i fördelnings av kroppsvikt mellan benen före utförandet av testen. Korrelationen var låg (r=0,38) för friska äldre mellan stabilitetsgränserna och avståndet för hur långt man sträckte sig under Functional reach, medan korrelationen var moderat (r=0,68) mellan hur långt man sträckte sig och bålens framåtlutning. Två faser identifierades under enbensstående respektive tandemstående. Under de första 3-5 sekunderna var det en dynamisk fas med en snabb minskning i variabiliteten av krafterna och därefter var det en statisk fas där variabiliteten bibehölls på konstant nivå. Under tandemstående fördelade både äldre och yngre mer vikt på det bakre benet.

Konklusion Sammanfattningsvis visar resultaten, att åldrandeprocessen i sig har en negativ inverkan på den posturala kontrollen under balanstesten. Friska äldre använder sig av en kompensatorisk rörelsestrategi under Functional reach, vilket resulterar i en låg kriterierelaterad validitet. Generellt kan detta arbete bidra med kunskap, om vad dessa balanstest mäter.


Nyckelord Postural kontroll; Balans; Åldrande; Kliniska balanstest; Functional reach; Enbensstående; Tandemstående; Kraft; Elektromyografi; Kinematik
LIST OF PUBLICATIONS

This thesis is based on the following publications, which are referred to in the text by their Roman numerals:

I. Jonsson E. Henriksson M. Hirschfeld H. 
Does the Functional reach test reflect stability limits in elderly people? 

II. Jonsson E. Seiger Å. Hirschfeld H. 
One-leg stance in healthy young and elderly adults: a measure of postural steadiness? 

III. Jonsson E. Seiger Å. Hirschfeld H. 
Postural steadiness and weight distribution during tandem stance in healthy young and elderly adults. 

IV. Jonsson E. Henriksson M. Hirschfeld H. 
Age-related differences in postural adjustments during different weight transfer tasks while standing. 
Submitted for publication.

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<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>% BoS</td>
<td>Percentage of the base of support</td>
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<tr>
<td>% BW</td>
<td>Percentage of the body weight</td>
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<td>% WT</td>
<td>Percentage of the weight transfer</td>
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<tr>
<td>A/P</td>
<td>Anterior/posterior</td>
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<td>AMTI</td>
<td>Advanced mechanical technology incorporated</td>
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<tr>
<td>ANOVA</td>
<td>Analysis of variance</td>
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<td>APA</td>
<td>Anticipatory postural adjustments</td>
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<td>ASCII</td>
<td>American standard code for information interchange</td>
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<td>BBS</td>
<td>Berg balance scale</td>
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<tr>
<td>BoS</td>
<td>Base of support</td>
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<tr>
<td>CoM</td>
<td>Center of mass</td>
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<td>CoP</td>
<td>Center of pressure</td>
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<tr>
<td>Elite</td>
<td>Elaboratore di immagini televisive</td>
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<td>EMG</td>
<td>Electromyography</td>
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<td>FR</td>
<td>Functional reach</td>
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<td>GRF</td>
<td>Ground reaction forces</td>
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<td>ICF</td>
<td>International classification of functioning, disability and health</td>
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<tr>
<td>LG</td>
<td>Lateral gastrocnemius</td>
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<tr>
<td>M/L</td>
<td>Medial/lateral</td>
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<tr>
<td>OLS</td>
<td>One-leg stance</td>
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<tr>
<td>RMS</td>
<td>Root mean square</td>
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<td>SENIAM</td>
<td>Surface EMG for non-invasive assessment of muscles</td>
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<td>TA</td>
<td>Tibialis anterior</td>
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<td>TS</td>
<td>Tandem stance</td>
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<tr>
<td>WCPT</td>
<td>World Confederation of Physical Therapy</td>
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<td>WHO</td>
<td>World Health Organization</td>
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DEFINITION OF CONCEPTS

Healthy aging

The biological process of becoming older without being afflicted by disease.

Postural control

Control of the body’s position in space for stability and orientation (Horak and Macpherson, 1996; Shumway-Cook and Woollacott, 2000).

Postural orientation

The ability to maintain an appropriate physical relationship between the body segments, as well as between the body and the environment during performance of a task (Shumway-Cook and Woollacott, 2000).

Postural stability

The ability to maintain the body in spatial equilibrium under both static and dynamic conditions (Shumway-Cook and Woollacott, 2000).

Balance

Clinically, balance is often synonymous with postural control, whereas in connection with research on postural control, balance is identical to postural stability or equilibrium. In the present thesis, the clinical concept of balance is employed.

Center of pressure, CoP

The point at which the reaction force of the ground impacts on the body (Winter et al., 2003).

Center of mass, CoM

A point representing the weighted average of the CoM of all segments of the body (Winter, 1995).
1 INTRODUCTION

Age-related deterioration in balance or postural control exerts a significant negative impact on the ability to perform everyday activities safely and are a major cause of falls (Tinetti et al., 1988; Camicioli et al., 1997; Tang and Woollacott, 2004). Falls are a serious health problem among older adults, often requiring hospital care and more intensive social care, as well as sometimes resulting in impaired function and permanent loss of independence (Zethraeus and Gerdtham, 1998; Randell et al., 2000). Thus, preventing falls is a major goal in the care of the elderly and improving our understanding of the factors and mechanisms involved in postural control in healthy elderly is a first step towards achieving this goal.

Physical therapists and other health-care professionals employ a wide range of tests and scales to evaluate balance and the risk of falls. Functional reach (FR), one-leg stance (OLS) and tandem stance (TS) are well-established tests that measures temporal and distance parameters that allow quantification of balance. Although these tests have been shown to be reliable and valid, they provide little information concerning the specific aspects of postural control that is being measured and the reasons for decline in performance. Moreover, their relevance with respect to postural control abilities in daily life has not been definitively established. The present studies were designed to determine the effects of healthy aging on balance and to identify the specific features of balance that are measured by three clinical tests involving forces, movement and muscle activity in a laboratory setting.

1.1 Aging

The proportion of the population that consists of elderly people is increasing in Western, as well as in many other countries. The average life-span in Sweden continues to rise and it is predicted that individuals over 65 years of age will account for 17 % of the Swedish population in 2006 and 25 % in 2030 (SCB, 2006).

Aging involves gradual, progressive and spontaneous deterioration of most physiological functions (Rundgren, 1991; Beers and Berkow, 2000). Although many studies on the process of aging have revealed declines in numerous sensory and motor functions in elderly individuals, how and why we age remains unclear. The several contemporary theories concerning aging can be divided into two categories: theories that maintain that aging is genetic and predetermined, advancing linearly with increasing age; and hypothesis concerning non-genetic secondary aging, which claim that this process is mainly due to environmental factors such as diseases and catastrophes that result in rapid, non-linear decline of behavioral functions. Obviously, genetic and secondary factors interact in conjunction with the aging process (Rundgren, 1991; Shumway-Cook and Woollacott, 2000; Beers and Berkow, 2000).
1.2 Postural control

Daily living requires that the central nervous system regulates and coordinates multi-joint movements and postural control during one and the same task. Virtually every movement that an individual makes contains both postural components, which stabilize the body, and prime mover components, designed to achieve a particular movement (Shumway-Cook and Woollacott, 2000). Postural control is a complex process requiring the integration of many bodily systems (Massion, 1994; Horak and Macpherson, 1996).

There is still no generally accepted definition of postural control nor is there any clear consensus with respect to the underlying mechanisms. Previously, postural control was often defined as the ability to maintain the body’s center of mass (CoM) within the boundaries of the base of support (BoS). However, this definition is not sufficient, since it is not applicable to all types of daily activities (Tang and Woollacott, 2004). For example, in connection with dynamic tasks such as walking, the CoM passes outside the medial border of the supporting leg rather than passing through the BoS.

The current view is that postural control in an upright standing position involves both static and dynamic components (Horak and Macpherson, 1996). The goals are to maintain the body’s position in space for stability and orientation. Postural orientation is defined as the ability to maintain a relationship between the different segments of the body as well as between the entire body and the environment that is appropriate for performance of the task at hand; while postural stability is the ability to maintain the body in spatial equilibrium under both static and dynamic conditions (Horak and Macpherson, 1996; Shumway-Cook and Woollacott, 2000). From this perspective good postural control means integration of posture and voluntary movements in such a manner that the individual is able to carry out a voluntary task safely (Tang and Woollacott, 2004).

The studies described in this thesis are based on a systems approach to motor control. This systems approach, first proposed by Bernstein in (1967), describes human movements as the result not only of physiological processes alone but also of constraints imposed by biomechanical factors such as internal (inertia and muscle force) and external forces (gravity). According to this theory postural control involves the interaction of multiple physiological subsystems with environmental factors and the performed task. Horak (1991; 1997) characterizes the primary physiological subsystems involved in postural control as sensory organization, perception of orientation, motor coordination, predictive central set, musculoskeletal subsystems and environmental adaptation. However, a somewhat simplified version of the systems approach model is employed here (Figure 1). This model divides the physiological subsystems involved in maintaining postural control into four main categories i.e. sensory, musculoskeletal, neural control and cognitive, but it is clear that the functioning of many subsystems actually requires interactions between these categories.

The task being performed often poses a direct threat to postural control. For instance, climbing up a ladder to reach for the curtains involves a greater threat to postural
control than does quiet maintenance of an upright stance. Similarly, the environment often influences postural control, e.g., it is generally more threatening to walk on ice than to walk on a gravel road. Each of these three constraints-individual, task-related and environmental, influence postural control to different extents.

1.2.1 Aging and postural control

The systems approach helps the clinician to determine the extent to which constraints in the function of specific subsystems contribute to deterioration of postural control in elderly individuals (Woollacott, 2000; Tang and Woollacott, 2004). The individual factors undergo age-related changes. Such changes in the organization of the sensory subsystem may include impaired vision (Gittings and Fozard, 1986), a reduction in somatosensory and vestibular information (Alvarez et al., 2000) and enhanced sway due to inaccurate proprioceptive input in conjunction with either preserved or absent visual input (Camicioli et al., 1997). Age-related changes in the musculoskeletal subsystem include attenuated muscle strength (Aniansson et al., 1986) and a reduced range of motion (Schenkman et al., 1996); while anticipatory postural adjustments (APA) made by the neural control subsystem may decrease or even be abolished (Inglin and Woollacott, 1988; Henriksson and Hirschfeld, 2005) and postural muscle responses may be slower due to prolonged onset latencies (Woollacott et al., 1986). Cognitive changes include a decline in performance that demands attention, e.g., simultaneous execution of dual tasks (Lundin-Olsson et al., 1997; Shumway-Cook et al., 1997; Pettersson, 2005).
1.2.2 Control of upright standing

The fact that the human body is able to maintain stable vertical posture is rather amazing, considering the number of joints present along its vertical axis. In addition, our BoS is relatively small and our CoM is localized high above the BoS, continuous fine tuning of the interaction between the movements of the different joints is thus necessary for the maintenance of postural control in a gravitational field.

Postural stability in a static position reflects equilibrium between all of the forces acting on the body (Horak and Macpherson, 1996). However, maintaining stability in a standing position is not a static task, since the body is never completely motionless, e.g., the CoM moves continuously as the body sways. The continuously varying forces exerted by different groups of muscles are reflected in the ground reaction forces (GRF) (Horak and Macpherson, 1996).

Many researchers employ sway or postural steadiness as an index of postural stability in the upright standing position and the numerous procedures that have been developed to measure this parameter include center of pressure (CoP), CoM, GRF, accelerometry and simply the movement of a single point on the body at a specified level (Gurfinkel, 1973; Murray et al., 1975; Kirby et al., 1987; Goldie et al., 1989; Goldie et al., 1992; Horak et al., 1992; Richardson et al., 1996; Moe-Nilssen and Helbostad, 2002; Winter et al., 2003). The ability to keep the body as motionless as possible can be defined as postural steadiness and measured as the standard deviation (SD) of the GRF (Murray et al., 1975; Goldie et al., 1989). However, although sway or postural steadiness can be measured in various ways, these parameters reflect different aspects of human standing (Karlsson and Frykberg, 2000). The GRF is not a measure of the actual body sway, but the acceleration of the CoM.

Goldie and coworkers (1989) have demonstrated that in healthy young individuals, variations in the GRF provide a more sensitive indicator of the changes in steadiness in different positions than do variations in the CoP; while others (Le Clair and Riach, 1996) have concluded that both the CoP and GRF provide reliable measures. When attempting to determine the optimal sampling time for quantitating SD values for CoP and GRF in connection with quiet standing and TS, Le Clair and Riach (1996) noted a tendency for GRF to decrease and for CoP to increase with standing time.

Postural steadiness has been shown to decrease with age (Murray et al., 1975; Hageman et al., 1995; Gill et al., 2001), although the use of sway as a measure of postural control has been questioned. Horak and colleagues (1996) found that patients with neurological disorders such as Parkinson’s disease exhibit normal sway even though they have poor postural control. On the other hand, recent evidence indicates that recording trunk sway in patients with Parkinson’s disease during the performance of stance and gait tasks can provide useful information concerning balance deficits that can cause such individuals to fall. (Adkin et al., 2005).

During activities performed in a standing position without moving the feet such as reaching and bending, the CoM remains within the limits of stability. If the CoM moves outside of these limits the individual will fall unless adequate postural
adjustments are made (McCollum and Leen, 1989). In response to destabilizing external perturbations, three different compensatory strategies have been described, i.e., the ankle strategy (Nashner, 1977), the hip strategy (Horak and Nashner, 1986) and the stepping strategy (Nashner et al., 1979). Elderly appear to use a hip strategy more often than an ankle strategy, in contrast to younger adults (Horak et al., 1989b).

The current view is that upright posture is regulated by an individual’s subconscious internal model of their body posture and limits of stability, together with their sensory and biomechanical capabilities. This internal model predicts the expected relationship between motor output and the environment, thereby providing a basis for the selection of an appropriate response for optimal control of equilibrium (Kawato, 1999). Thus, these limits of stability can be considered to be perceived, whereas the actual limits of stability are dependent on the support base, the environment and the nature of the task being performed (McCollum and Leen, 1989).

Perceived limits of stability may be defined as the distance a person is willing and/or able to move without losing balance or being forced to take a step and these perceived limits can be narrowed by the presence of musculoskeletal and/or neurological pathology or by the fear of falling. Under optimal conditions the actual and perceived limits of stability are the same. Perceived limits of stability have been found to decline with advancing age (King et al., 1994).

The limits of CoP movement reflect the limits of stability within which the CoM moves (King et al., 1994). One way to explore these limits of stability is to characterize the location of the CoP and the path of its movement in relationship to the BoS during the performance of a task. Thus, these limits have been explored with tasks involving leaning in different directions (Murray et al., 1975; King et al., 1994; Wallmann, 2001) and utilizing a task that requires reaching forward (Duncan et al., 1990).

### 1.2.3 Coordination of posture and movement

During the performance of most voluntary movements, the primary parameters of posture must be maintained in order to accomplish the goal successfully. The preservation of equilibrium and orientation of the body segments with respect to gravity are essential, since these provide a frame of reference for planning the path of movement necessary to reach the goal (Massion et al., 2004).

Voluntary movements such as lifting an arm or taking a step evoke reaction forces that act on the body to destabilize the equilibrium of the CoM. Maintenance of equilibrium under these conditions involve APA together with voluntary, focal movements. First described by Belen’kii (1967) in association with arm movements, APAs are postural adjustments that are invariably proportional to the focal movement, suggesting that such adjustments are an integral part of the planning of the movement (Cordo and Nashner, 1982; Nardone and Schieppati, 1988).

Usually, an APA precedes the voluntary movement, activating muscles which compensate for the coming change in the posture. However, APAs also assist
movement by creating a distance between the CoP and CoM (i.e., a moment arm). For example, posterior displacement of the CoP occurs prior to forward oriented movements such as reaching and initiation of gait (Crenna and Frigo, 1991). This investigation concluded that inhibition of the soleus muscle in conjunction with activation of the tibialis anterior (TA) is responsible for this backward shift in the CoP.

However, the occurrence of APAs is dependent on the task being performed and on the status of the support base (Horak and Macpherson, 1996). Thus, no backward shift of CoP occurs when subjects rise up on their toes while holding on to a support or initiate movement while leaning forward (Clement et al., 1984; Nardone and Schieppati, 1988). APAs that precede and assist voluntary movements decrease in frequency or even disappear in the elderly (Man'kovskii et al., 1980; Inglin and Woollacott, 1988; Henriksson and Hirschfeld, 2005).

In addition to APAs, compensatory postural adjustments may be required during voluntary movements to maintain equilibrium by compensating for miscalculation of the voluntary movement itself or for some unexpected external perturbation. Sensory inputs provide information concerning postural disturbance and trigger appropriate postural adjustments in a feedback manner (Horak and Macpherson, 1996). For example, maintenance of postural control while shifting weight requires APAs and compensatory postural adjustments. Such postural adjustments are reflected in the spatial and temporal characteristics of the GRF generated beneath the leg that is about to be lifted (Rogers and Pai, 1993; Patchay et al., 2002).

In humans, a transfer of body weight from one leg to the other, with a concomitant decrease in the BoS, is a basic aspect of locomotion and other everyday activities. Transfer of the body’s CoM laterally while standing is a prerequisite for lifting one leg off the ground (Rogers and Pai, 1990). Accordingly, the ability to transfer weight from one leg to the other is central for gait initiation, as well as for successful performance of balance tests such as the OLS and TS (Rogers and Pai, 1990; Breniere and Do, 1991). However, unlike OLS and TS, gait is a dynamic forward-oriented task that does not require the CoM to be maintained within the BoS.

Instead, the central nervous system controls CoM during gait by continuously anticipating changes in the BoS. A transfer of weight begins with an increase in the vertical and lateral forces beneath the leg that is about to be lifted and accompanying reduction in the forces beneath the standing leg, with the CoM shifting towards the standing leg (Rogers and Pai, 1990; Patla et al., 1993; Rogers and Pai, 1993). Thereafter, the leg to be lifted is unloaded and the standing leg loaded in connection with which the CoM reaches its final position (Mouchonino et al., 1992). The current view is that gait initiation is performed in a feedforward manner, being programmed prior to the lift of the swing leg from the ground (Breniere and Do, 1991).
1.3 Clinical evaluation of balance

A number of clinical measures of balance have been developed and utilized to assess balance in elderly individuals (Tinetti, 1986; Berg K et al., 1989; Duncan et al., 1990; Podsiadlo and Richardson, 1991; Lundin-Olsson et al., 1997; Thomas et al., 2004). Latash (1993) has divided the information obtained from clinical and laboratory measures into task and performance parameters. Clinical tests measure task parameters associated with what a subject does, i.e. what is required of the subject, but not how the task is performed. For example the OLS test determines the number of seconds a person can stand on one leg, but provides no information concerning how this position is achieved or maintained.

Patla and colleagues (1990) classified the wide range of balance tests into the following categories: static-unperturbed balance tests; static- perturbed balance tests; tests of balance during unperturbed voluntary movements; and tests of balance in connection with perturbed voluntary movement. According to this categorization, OLS and TS are examples of static-unperturbed balance tests, while FR tests balance control during unperturbed voluntary movements. These three balance tests are employed individually or in combination for clinical assessments (Tinetti, 1986; Berg K et al., 1989; Bohannon and Leary, 1995; Thomas et al., 2004). All three are included in the Berg Balance Scale (BBS), a well-known balance assessment that has been validated in elderly adults and stroke patients (Berg et al., 1989; Berg et al., 1992; Berg et al., 1995). An overview of the reliability and validity studies that have been performed on these tests is presented in Table I.

1.3.1 ICF

Outcome measures can be classified with respect to various levels of functioning. The International Classification of Functioning, Disability and Health (ICF), developed by the World Health Organization (WHO), provide a systematic classification of health and health-related domains. According to the ICF, all conditions of health are determined by body functions and body structures, activities and participation, together with interacting environmental factors (WHO, 2001). The ICF is a tool for systematically relating outcome measures to these levels. Thus, FR, OLS and TS can be assigned to the activity and participation level of the mobility domain (i.e., maintaining a standing position, shifting the body’s centre of gravity, and reaching).

1.3.2 Functional reach

FR, a well-established clinical measure of balance, was developed by Duncan and coworkers (1990), and has been tested for both validity and reliability (Duncan et al., 1990; Weiner et al., 1992; Duncan et al., 1992; Weiner et al., 1993). Based on a leaning task, FR is proposed to measure the limits of stability. This test measures the distance between the length of the outstretched arm and a maximal forward reach from a standing position, with maintenance of a fixed BoS. It was developed as a dynamic measure of balance, with no concern for the movement strategy, and elderly people
who are unable to reach more than 15 cm (6 inches) exhibit an increased risk for falls
and frailty (Duncan et al., 1990; Weiner et al., 1992; Duncan et al., 1992).

FR is applied to patients with diagnoses as varied as stroke (Fishman et al., 1997),
Parkinson (Smithson et al., 1998), vestibular hypofunction (Wernick-Robinson et al.,
1999), multiple sclerosis (Frzovic et al., 2000) and hip fractures (Ingemarsson et al.,
2000). However, a number of factors exert a major influence on this evaluation. Earlier
research revealed that the movement strategy and a reduced spinal flexibility both
affect the reach distance (Wernick-Robinson et al., 1999; Cavanaugh et al., 1999;
Schenkman et al., 2000); while others question the ability of FR to differentiate elderly
non-fallers and fallers, as well as its relationship to leaning tasks (Wallmann, 2001).

1.3.3 One-leg stance

OLS is frequently employed for the clinical assessment of patients with various balance
disorders (Fugl-Meyer et al., 1975; Tinetti, 1986; Berg et al., 1989; Goldie et al., 1989;
Bohannon and Leary, 1995; Smithson et al., 1998; Frzovic et al., 2000; Allum et al.,
2001). The task of standing on one leg requires the initially voluntary shift of the CoM
to the leg which will be stood on, followed by maintenance of postural orientation in
space, by controlling weight support, the vertical alignment of the different segments of
the body and equilibrium (Rogers and Pai, 1990; Horak and Macpherson, 1996). This
test has been evaluated for both reliability and validity (see Table I) and is claimed to
be a marker of frailty (Drusini et al., 2002). The OLS outcome is sensitive to aging
(Briggs et al., 1989; Gill et al., 2001) exhibiting a significant decrement as early as the
fifth decade of life (Balogun et al., 1994).

OLS assesses postural steadiness in a static position by a quantitative measurement
being the number of seconds that an individual can maintain a one-leg stance, the
underlying assumption being that the better one’s postural steadiness, the longer such a
position can be held. However, different established scales for assessing balance
require different OLS times to obtain a maximal score (Fugl-Meyer et al., 1975; Tinetti,
1986; Berg et al., 1989; Bohannon and Leary, 1995). Thus to obtain the highest score
possible on the BBS (Berg et al., 1989) and the Fugl-Meyer assessment (Fugl-Meyer et
al., 1975), a subject must to stand unsupported on one leg for at least 10 seconds;
whereas in the case of Bohannon’s ordinal balance scale (Bohannon and Leary, 1995)
30 seconds is the required time, and Tinetti’s Balance Subscale (Tinetti, 1986)
considers a subject as having normal balance if he/she can stand on one leg without
support for 5 seconds. Frändin et al. (1995) noted that most patients who are able to
stand in this manner for approximately 15 seconds also manage to stand for 30 seconds.

1.3.4 Tandem stance

Performance of tandem stance, an accepted clinical measure of standing balance (Berg
et al., 1989; Tesio et al., 1997; Thomas et al., 2004) has been shown to deteriorate with
advancing age (Briggs et al., 1989). The ability to stand in a heel-to-toe position reflects
the degree of postural steadiness when the BoS in the medial/lateral (M/L) direction is
narrow. Also known as the sharpened Romberg test, TS was developed on the basis of the classic Romberg’s sign and originally utilized as a bedside indicator of abnormality in the functioning of the proprioceptive system (Lanska and Goetz, 2000). Today, TS is included in several balance assessments (Berg et al., 1989; Tesio et al., 1997; Thomas et al., 2004).

Table I An overview of reliability and validity studies of the Functional reach (FR), one-leg stance (OLS) and tandem stance (TS) tests on elderly individuals.

<table>
<thead>
<tr>
<th>Test</th>
<th>Study</th>
<th>Measurement properties</th>
</tr>
</thead>
<tbody>
<tr>
<td>FR</td>
<td>Duncan et al., 1990</td>
<td>Parallel validity FR vs. CoP, r= 0.71</td>
</tr>
<tr>
<td></td>
<td>Duncan et al., 1992</td>
<td>Test-retest reliability ICC= 0.81</td>
</tr>
<tr>
<td></td>
<td>Franchignoni et al., 1998</td>
<td>Predictive validity ≤ 6 inches OR= 4.03 frailty</td>
</tr>
<tr>
<td></td>
<td>Giorgetti et al., 1998</td>
<td>Test-retest reliability ICC= 0.86-0.88</td>
</tr>
<tr>
<td></td>
<td>Rockwood et al., 2000</td>
<td>Test-retest reliability ICC= 0.92</td>
</tr>
<tr>
<td></td>
<td>Franchignoni et al., 1998</td>
<td>Inter-rater reliability ICC= 0.96-0.97</td>
</tr>
<tr>
<td></td>
<td>Giorgetti et al., 1998</td>
<td>Inter-rater reliability ICC= 0.73</td>
</tr>
<tr>
<td></td>
<td>Rockwood et al., 2000</td>
<td>Inter-rater reliability ICC= 0.92</td>
</tr>
<tr>
<td></td>
<td>Wallmann, 2001</td>
<td>Content validity, (feasibility)</td>
</tr>
<tr>
<td></td>
<td>Weiner et al., 1992</td>
<td>Concurrent validity, (specificity)</td>
</tr>
<tr>
<td></td>
<td>Weiner et al., 1993</td>
<td>Responsiveness RI= 0.97, r= 0.38</td>
</tr>
<tr>
<td></td>
<td>Wernick-Robinson et al., 1999</td>
<td>Concurrent validity, (specificity)</td>
</tr>
<tr>
<td></td>
<td>OLS 30 s, ICC= 0.99-0.91</td>
<td>FR vs. LOS, r ≤0.17</td>
</tr>
<tr>
<td></td>
<td>OLS 30 s, ICC= 0.99</td>
<td>Fallers vs. non-fallers, P= 0.82</td>
</tr>
<tr>
<td></td>
<td>OLS 30 s, ICC= 0.75</td>
<td>FR vs. gait speed, r= 0.71,</td>
</tr>
<tr>
<td></td>
<td>Hurvitz et al., 2000</td>
<td>OLS &lt; 30 s OR= 108</td>
</tr>
<tr>
<td></td>
<td>Lin et al., 2004</td>
<td>Predictive validity to falls, 0.10, ADL decline, 0.19 and improvement, 0.00.</td>
</tr>
<tr>
<td></td>
<td>Thomas and Lane, 2005</td>
<td>Construct validity, (convergent)</td>
</tr>
<tr>
<td></td>
<td>TS 60 s, ICC= 0.99-0.91</td>
<td>≤ 1.02 s OR= 15.2</td>
</tr>
<tr>
<td></td>
<td>TS 60 s, ICC= 0.99</td>
<td>Fallers vs. non-fallers, P&lt;0.05</td>
</tr>
<tr>
<td></td>
<td>Inter-rater reliability</td>
<td>ICC= 0.69</td>
</tr>
</tbody>
</table>

RI= responsiveness index, CoP= center of pressure, VH=vestibular hypofunction, ADL= activities of daily living, IADL= instrumental activities of daily living, LOS=limits of stability, PD= Parkinson’s disease, MS= multiple sclerosis, ICC= Intra-class correlation, OR= odds ratio.
As in the case of OLS, TS assesses postural steadiness in a static position by measuring the number of seconds that a person can maintain this position, again assuming better postural steadiness allows longer standing in this position. In connection with the use of TS in the BBS, only one foot in front of the other is tested. However, earlier investigations have revealed that younger adults tend to support most of their body weight on the rear leg during TS (Kirby et al., 1987; Nichols et al., 1995). Thus, in the light of the fact that motor and sensory deficits frequently affect the two lower limbs to different extents, TS scores might be dependent on which leg is in the rear position (Mizrahi et al., 1989; Kusoffsky et al., 2001).

1.3. Measurement properties

Measurement properties include reliability, validity, and the ability of an assessment method to detect changes (Finch et al., 2002). Obviously, the value of an assessment depends on its validity and reliability. A reliable measure must not only provide values associated with a small margin of error, but also be capable of differentiating between individuals with different degree of balance (Finch et al., 2002). Although reliability is necessary for validity, it does not validate the meaning behind the measure.

Validity is commonly regarded as the degree to which a procedure for assessment actually does measure what it is designed to measure (Johnstone et al., 1992; Finch et al., 2002). The aspects of validity on which the present work focuses are ‘construct validity’ i.e., the degree to which an instrument measures the theoretical construct it was designed to measure (Johnstone et al., 1992) and ‘criterion validity’ more specifically ‘concurrent validity’, which means the accuracy or validity of the measurement obtained in comparison to a standard measurement (Domholt, 2000). In this context, laboratory measurements can be employed as reference standards.

1.4 Laboratory measures of postural control

Balance performance can be quantitated in a laboratory setting employing everything from simple portable equipment to highly sophisticated technical devices, such as force platforms and systems for analysis of movement. Analysis of movements, forces and muscle activity can provide insight into the manner in which the nervous system contributes to the control of posture and balance. In contrast to clinical balance tests which measure task parameters (i.e., what the subject is doing) laboratory analyses concern performance parameters (i.e., how the subject performs the task) (Latash, 1993).

1.4.1 Measurements of movement

Objective, systematic analysis of human movements was initially undertaken during the Renaissance period (Borelli, 1685) and has since developed into a highly technological endeavor. Contemporary systems for movement analysis involve video-based
movement measurements, electrogoniometry, accelerometry and/or electromagnetic systems. Camera-based systems involve either passively reflective or actively signaling markers placed on body segments in alignment with specific bony landmarks. When viewed from at least two camera angles the instantaneous three-dimensional positions of these markers relative to a fixed laboratory coordinate system can be determined. The kinematic information thus obtained provides information concerning the linear and angular displacement of the bony landmarks and body segments, joint dynamics and displacement of CoM (Kaufman, 2004).

1.4.2 Measurements of force

Kinetic information provides knowledge concerning the forces that cause the movements observed, and can therefore be both descriptive and explanatory information. Force plates are commonly employed to monitor the kinetics of movement. Piezoelectric crystals or strain gauges mounted on the force plates sense forces in three spatial directions, measuring variables such as the magnitude of forces, their rate (the first derivative of the force as a function of time), force impulse (i.e., integration of force over an interval of time) and displacement of the CoP. Numerous researchers have documented the relationship between various clinical tests of balance and measurements with force plates (Tropp and Odenrick, 1988; Goldie et al., 1989; Duncan et al., 1990; Goldie et al., 1992; Frändin et al., 1995; Hageman et al., 1995; Richardson et al., 1996; Camicioli et al., 1997).

When standing quietly, a person exerts a force on the ground, the so-called action force, which is equal to body mass multiplied by the acceleration of gravity. At the same time, the ground exerts an equal, but opposite force on the person, referred to as the GRF (Enoka, 2002).

1.4.3 Measurements of muscle activity

An important area in the investigation of neuromuscular control of posture and movement is concerned with analysis of muscle activity. Electrical events in the muscles can be recorded and analyzed by electromyography (EMG) involving electrodes placed within the muscle or on the surface of the skin lying above the muscle (Basmajian, 1979; Söderberg and Knutson, 2000). Surface EMG is treacherous territory, since this procedure is very easy to apply and the resulting often very difficult to interpret correctly (Deluca, 1997).

The amplitude of the EMG signal is usually expressed as the root mean square average (RMS), an average rectified value or an integrated rectified value (Basmajian, 1979). Although these values respond in a similar fashion to alterations in force, use of the RMS value is generally recommended. Since the size of the amplitude reflects the number of active motor units and their rates of firing, an increase in either of these parameters causes an elevation in the RMS value. In order to adjust for potential variations arising from differences in the placement and spacing of electrodes and from
anatomical factors, as well as to allow comparison between different muscles and subjects, the RMS value should be normalized (Deluca, 1997).

The time of onset of muscle activity is one of the parameters evaluated most commonly. However, no standard procedure for determination of EMG onset has yet been described (Hodges and Bui, 1996; Hermens et al., 1999), and various approaches ranging from visual observation to automatic computer algorithms has been employed (Hodges and Bui, 1996; Abbink et al., 1998). There are often no strict rules with respect to visual determination and the decision concerning onset is left to the discretion of the examiner (Söderberg and Knutson, 2000).

A certain level of objectivity has been made possible through the development of procedures designed to identify the onset of EMG on the basis of computer algorithms. These algorithms are often designed to identify the timepoint at which the mean activity of a specified number of samples exceeds the baseline activity (averaged over a given period of time prior to the onset of movement) by a specified quantity, e.g., SD (Hodges and Bui, 1996). However, the EMG recording must also be examined visually in order to ensure validity (Difabio, 1987; Hodges and Bui, 1996). A recent review of the literature (Jonsson, 2002) concluded that the most frequently employed approach is to define onset as the timepoint at which the EMG activity exceeds the baseline mean by at least two SD’s together with confirmation by visual inspection. Determination of EMG onset is also influenced by earlier signal processing, e.g., the use of low-pass filtering, the width of the sliding window utilized to analyze the EMG and the sampling frequency (Hodges and Bui, 1996).

1.5 The rationale behind this thesis

Movement is the basic concept underlying physical therapy (Hislop, 1975; Tyni-Lenné, 1998; WCPT, 1999) and postural control is a prerequisite for voluntary movements (Shumway-Cook and Woollacott, 2000). Physical therapists generally assess balance in elderly patients with various balance tests. It seems as there is a serious gap between clinical balance testing and the theoretical framework concerning postural control. Many of these tests evaluate what the individual is doing (task parameters), but not how the task is being performed (performance parameters). There is a need to identify clearly what is being measured by clinical balance tests, in order to allow assessment of the subsystems involved, which is a prerequisite for adequate intervention. There is also a need to characterize the role that aging plays in connection with the assessment of balance in order to be able to interpret test results correctly. Investigation of the effects of healthy aging on balance may provide insight into alterations in the mechanisms underlying postural control that may eventually lead to deficient balance and/or a fall.
2 AIMS

The general aims were
- to determine the effects of healthy aging on balance by evaluating three clinical balance tests, FR, OLS and TS, employing measures of force and movement together with muscle activity in a laboratory setting; and
- to identify what is being measured by these tests.

The specific aims were
- to correlate reach distance with displacement of the CoP in order to determine whether this test provides a measure of the limits of stability and, in addition, to investigate how healthy elderly adults perform during FR.  
  (Study I)
- to compare the degree of and changes in postural steadiness in healthy elderly and younger adults during 30 seconds of OLS.  
  (Study II)
- to compare postural steadiness and weight distribution, as well as alterations in these parameters, in healthy elderly and younger adults during 30 seconds of TS.  
  (Study III)
- to characterize age-related differences in postural adjustments occurring prior to and during the performance of three tasks involving lateral transfer of weight, i.e., OLS, TS and gait initiation.  
  (Study IV)
3 METHODS

3.1 The subjects

Thirty-three healthy volunteers 65-80 years of age were recruited from senior citizen organizations in and around Huddinge, Sweden. In addition, 30 healthy young adults 25-40 years of age were recruited from this same region through advertisements. In the different studies the findings on certain subjects were excluded from the analysis due to technical problems or the fact that they did not perform all of the tasks under investigation.

The characteristics of our subjects are documented in Table II. None had any history of neurological illness, musculoskeletal disorders, degenerative conditions or any other disease, and none was taking medication that might interfere with normal postural control and/or gait. Subjects with less than 90 degrees of shoulder flexion were excluded. A clinical examination concerning leg discrepancy and range of motion of the lower extremities revealed normal findings for both groups, with no side differences. All of the subjects walked freely without any sort of aid and were participating actively in several outdoor and indoor activities every week.

<table>
<thead>
<tr>
<th>Study I</th>
<th>Number of subjects</th>
<th>Sex (F/M)</th>
<th>Age (years)</th>
<th>Weight (kg)</th>
<th>Height (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elderly</td>
<td>27</td>
<td>18/9</td>
<td>71.3 (4.0)</td>
<td>72.0 (12.6)</td>
<td>1.67 (0.09)</td>
</tr>
<tr>
<td>Younger</td>
<td>28</td>
<td>20/8</td>
<td>70.5 (3.8)</td>
<td>71.7 (13.1)</td>
<td>1.67 (0.09)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Study II</th>
<th>Number of subjects</th>
<th>Sex (F/M)</th>
<th>Age (years)</th>
<th>Weight (kg)</th>
<th>Height (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elderly</td>
<td>28</td>
<td>20/8</td>
<td>70.5 (3.8)</td>
<td>71.7 (13.1)</td>
<td>1.67 (0.09)</td>
</tr>
<tr>
<td>Younger</td>
<td>28</td>
<td>16/12</td>
<td>29.9 (4.2)</td>
<td>75.0 (16.8)</td>
<td>1.74 (0.10)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Study III</th>
<th>Number of subjects</th>
<th>Sex (F/M)</th>
<th>Age (years)</th>
<th>Weight (kg)</th>
<th>Height (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elderly</td>
<td>26</td>
<td>18/8</td>
<td>70.6 (3.8)</td>
<td>71.5 (13.5)</td>
<td>1.67 (0.09)</td>
</tr>
<tr>
<td>Younger</td>
<td>27</td>
<td>15/12</td>
<td>30.0 (4.1)</td>
<td>76.5 (16.8)</td>
<td>1.75 (0.10)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Study IV</th>
<th>Number of subjects</th>
<th>Sex (F/M)</th>
<th>Age (years)</th>
<th>Weight (kg)</th>
<th>Height (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elderly</td>
<td>24</td>
<td>17/7</td>
<td>70.3 (3.8)</td>
<td>72.5 (12.8)</td>
<td>1.67 (0.08)</td>
</tr>
<tr>
<td>Younger</td>
<td>26</td>
<td>15/11</td>
<td>30.0 (4.2)</td>
<td>75.6 (16.1)</td>
<td>1.74 (0.10)</td>
</tr>
</tbody>
</table>

A questionnaire designed to determine the subjects’ level of regular physical activity (at least 45 minutes in duration; 1 = every day, 2 = 5-6 times per week, 3 = 3-4 times per week, 4 = 1-2 times per week, 5 = none) was administered. The elderly group reported a slightly higher level of regular physical activity (mean score of 1.0, range 1-4) than did the younger group (2.0, range 1-4, $P < 0.01$), (Henriksson and Hirschfeld, 2005). Questions about fear of falling and actual falls revealed that only one elderly individual
had fallen once during the two years preceding this study and that two elderly subjects expressed some fear of falling.

### 3.2 Clinical tests and procedures

The following three tests of balance were performed:

*Functional reach* (Figure 2). The initial position was 90 degrees of shoulder flexion and a straight arm. Subjects then reached horizontally forward with their left hand while keeping their feet on the ground. The instructions they received were similar to those described by Duncan et al. (1990). A 150-cm yardstick was mounted horizontally on the wall, at the height of acromion. In accordance with the procedure employed by Duncan and colleagues (1990) the reaching strategy was not controlled for, in any other way.

![Figure 2](image)

*Figure 2* The initial and final positions associated with functional reach *(Study I)*

*One-leg stance* (Figure 3). The subjects were standing freely on one leg for 30 seconds or as long as they were able. Each subject was instructed to keep his or her arms along the side of the body during this period.
Tandem stance (Figure 4). Two different TS positions were examined: an original position (TS\textsubscript{org}) and, thereafter, a modified one (TS\textsubscript{mod}). The subjects were instructed to stand in the position with one foot in front of the other for 30 seconds. The instructions during TS\textsubscript{mod} were also “stand in the middle, with equal weight on each leg”.

In addition, gait initiation (Study IV) was performed. Here, the subjects were asked to initiate normal walking and cross a custom-designed walkway when a traffic light changed from red to green (Henriksson and Hirschfeld, 2005).

Due to the nature of our experimental set-up and because this task was the least tiring, test sessions always began with gait initiation. Thereafter the remaining three tests were performed in random order by each subject, with breaks between the tasks. The subjects were allowed to practice prior to testing and to choose which leg to lift in connection with OLS and TS, as well as the leading limb during gait initiation. In the following descriptions, the leg on which the subjects stood is referred to as the stance leg and the leg which was lifted as the swing leg. In the case of TS the rear and front legs are also referred to.

The performance of these balance tests were monitored simultaneously by measuring distance (FR) or time (OLS and TS) and employing laboratory measures. Initially, the subjects were asked to stand in a relaxed position with their eyes open and weight distributed evenly on both feet. An auditory (FR, OLS and TS) and visual (gait
initiation) cue indicated when to start the test. The investigator terminated OLS and TS either after 30 seconds, or when the subject was unable to maintain the position any longer. FR was performed 5 times, OLS and TS 3 times and gait initiation a total of 20 times but only the result of the first three trials with the same leg were analyzed.

3.3 **Laboratory measurements**

An overview of the clinical tests and laboratory measurements employed in these studies is presented in Table III.

**Table III** Overview of the clinical tests and laboratory measurements employed in studies I-IV.

<table>
<thead>
<tr>
<th>Clinical test</th>
<th>Laboratory measurements</th>
</tr>
</thead>
<tbody>
<tr>
<td>Study I</td>
<td></td>
</tr>
<tr>
<td>Functional reach</td>
<td>Movement: Two-camera system</td>
</tr>
<tr>
<td></td>
<td>Force: Two force plates and CoP</td>
</tr>
<tr>
<td></td>
<td>Muscle activity: Surface EMG on the ankle muscles bilaterally</td>
</tr>
<tr>
<td>Study II</td>
<td></td>
</tr>
<tr>
<td>One-leg stance</td>
<td>Movement: n/a</td>
</tr>
<tr>
<td></td>
<td>Force: Two force plates</td>
</tr>
<tr>
<td>Study III</td>
<td></td>
</tr>
<tr>
<td>Tandem stance</td>
<td>Movement: n/a</td>
</tr>
<tr>
<td></td>
<td>Force: Four force plates</td>
</tr>
<tr>
<td>Study IV</td>
<td></td>
</tr>
<tr>
<td>One-leg stance</td>
<td>Movement: n/a</td>
</tr>
<tr>
<td>(Gait initiation)*</td>
<td>Force: Two force plates</td>
</tr>
<tr>
<td></td>
<td>Muscle activity: n/a</td>
</tr>
</tbody>
</table>

n/a= not applicable, CoP= center of pressure, *=not a clinical test, only used for comparison

3.3.1 **Measurement of movement**

Movements were recorded by the Elite system (Elaboratore di immagini televisive BTS, Milan, Italy) which consists of two charge-coupled device (CCD) cameras with a sampling rate of 100 Hz. These cameras were placed 4 m from the force plates, at 35° angles to the sagittal plane and facing the right side of the body. The size of the field was 2 m x 2 m, providing an accuracy of 0.8 mm. Spherical reflective markers (1 cm in diameter) were placed on ten anatomical landmarks on the right side of the body (see Figure 5). In addition, two markers were placed on landmarks on the left side of the body (i.e., the lateral humeral epicondyle and the tip of the third phalanx) and for spatial orientation, one marker on the wall and one more on the back corner of the force plate beneath the right foot. The kinematic data obtained were processed utilizing the Elite system, which tracks the markers to allow three-dimensional reconstruction.
3.3.2 Measurement of force

Ground reaction forces were recorded from four equal force plates (size 457 x 203 mm; accuracy 0.25 N; model Mc818-6-1 000; purchased from AMTI, Advanced Mechanical Technology Incorporation Watertown, USA). These force plates were positioned in a square level with the floor and 40 and 75 mm apart in the M/L and anterior/posterior (A/P) directions, respectively. Three orthogonal forces, i.e., the A/P, M/L and vertical forces were determined using a sampling frequency of 100 Hz. Initially, the subjects stood with one foot on each of the first two force plates and prior to the signal to begin the task, 2-3 seconds of baseline data were recorded.

3.3.3 Measurement of muscle activity

Electromyographic recording of muscle activity performed with the Bagnoli-8 system, (type DE-02; size 23 x 17 mm; Boston, USA,) using electrodes placed bilaterally on the ankles. The 4 bipolar surface electrodes with an interelectrode distance of 10 mm were attached with adhesive tape over the belly of the left and right TA and left and right lateral gastrocnemius (LG) muscles. In the case of the TA the electrodes were placed at the proximal 1/3 on the line between the tip of the head of the fibula and the tip of the medial malleolus, and for LG at the proximal 1/4 on the line between the tip of the head of the fibula and the tip of the lateral malleolus (Hermens et al., 1999, SENIAM).

The electrodes employed had a built-in gain amplification of 10, the sampling frequency was 800 Hz and the raw EMG signals were further amplified by a factor of 200 and band-pass filtered between 20-500 Hz. These amplified signals were then processed utilizing the RMS with 1.25-ms sampling interval for purposes of rectification and these traces used for onset analysis. For analysis of the mean amplitude, the traces were treated by RMS with an averaging technique involving 20-ms sliding window and thereafter stored together with the force data in a SC/ZOOM flexible laboratory computer system (Department of Physiology, Umeå University, Sweden) for additional analysis.

3.4 Analysis of the data

Kinematic, force plate and electromyographic signals were recorded simultaneously. These data were subsequently transformed into ASCII files and analyzed with Axograph (Axon Instruments, Union City, USA), a Macintosh-based software package, with the exceptions of the determination of EMG onset on the basis of EMG recordings which was performed by SC/ZOOM. In order to make comparisons between different subjects possible, the amplitudes of the force signals were normalized for and expressed as a percentage of body weight (%BW). More detailed information concerning the data analysis is presented in the individual papers.
3.4.1 Study I

The netCoP displacement in the A/P (netCoPx) and M/L (netCoPy) directions was calculated as follows:

\[
\text{netCoPx} = \frac{(\text{CoPx1} \times \text{Fz1}) + (\text{CoPx2} \times \text{Fz2})}{\text{Fz1} + \text{Fz2}}; \\
\text{netCoPy} = \frac{(\text{CoPy1-223} \times \text{Fz1}) + (\text{CoPy2+ 20} \times \text{Fz2})}{\text{Fz1} + \text{Fz2}}
\]

Where 1 refers to the force plate beneath the left and 2 beneath the right foot, see Figure 5B. In the A/P direction the displacement of the CoP was normalized for foot length and expressed as a percentage of the base of support (%BoS).

**Figure 5** A) Schematic drawing of the experimental set-up employed in Study 1. The placement of the markers 1-14 is indicated on the stick figure. The reference coordinates for the Elite system are indicated. B) Force plate coordinates and the netCoP, are also shown. C) Definition of the trunk segment angle (trunk segment versus vertical axis Y). D) Definition of the ankle joint angle as the projection between the foot and the shank segment along the sagittal plane. Angular displacement increases in the direction of the arrow.
The peak amplitudes of the trunk and ankle angles (defined in Figure 5) were computed and changes in these angles analyzed in relationship to the baseline values associated with initial standing. An increase in the trunk segment angle reflects bending of the trunk forward (forward rotation), while increase in the ankle joint angle reflects plantar flexion (extension).

The onset of the plateau of the finger marker displacement was defined from cursor read-outs and defined as time zero and all other amplitudes under consideration were measured at this same timepoint. More detailed analysis (unpublished data, Kvimark and Jonsson, 2003) of the finger marker in relationship to the peak amplitude of the A/P CoP as well as analysis of the CoP in the M/L direction was carried out on the values obtained from 15 of the elderly subjects. Onset latency in the EMG burst; defined as an activity that was more than 2 SD greater than the mean baseline activity and lasted longer than 30 milliseconds was measured relative to the onset of the A/P force.

3.4.2 Studies II and III

Postural steadiness during OLS and TS was examined by dividing the 30-second period into shorter intervals and characterizing the variability in force between these intervals. In Study II the variabilities in the vertical and M/L forces beneath the stance leg were analyzed in terms of the mean SD of the force signals observed during five separate intervals (Figure 6A). These intervals were chosen to cover the initial period following lift-off (intervals 1 and 2) and 5, 10 and 30 seconds (intervals 3-5) of the OLS assessment.

In Study III the M/L force was monitored during ten separate intervals (Figure 6B). Vertical forces and muscle activity during a static phase (10 seconds) in the middle of the 30-second trial periods for both TS positions (the grey area in Figure 6B) were analyzed. The muscle activity was expressed relative to the mean amplitude of two 5-seconds period of normal upright standing. Furthermore, in study II, the impulse of the vertical and M/L force signal prior to lifting the swing leg was measured as the area of the force-time curve lying above the baseline (i.e., the grey area in Figure 7).
A quantitative analysis of clinical tests

Figure 6 A) An individual time trace of the medial/lateral (M/L) forces exerted by one elderly subject performing one-leg stance (OLS). The amplitude of the force is expressed in Newtons (N). The variability in the M/L force beneath the stance leg was analyzed in terms of the mean standard deviation during the following five time intervals: 1 = 0-0.49 s; 2 = 0.5-0.99 s; 3 = 1-4.99 s; 4 = 5-9.99 s; and 5 = 10-30 s. B) Individual time traces of the M/L forces exerted by the rear (R) and front legs (F) of one elderly subject performing tandem stance (TS). The traces begin when this subject is standing with his feet side-by-side. Thereafter, the changes in force associated with placement of the front foot on the force plate in front of the force plate under the rear foot are shown. Time interval 1 extended from the onset of change in the vertical force on the plate beneath the front leg until the front foot had been completely placed and this vertical force attained a plateau. The remaining time was then divided into five 1-s intervals (intervals 2-6) followed by four 5-s intervals (interval 7-10). The grey area represents analysis of the 10-s static phase.

3.4.3 Study IV

In this case, analysis was limited to what together with lift-off is characterized as the first phase of gait initiation (Jian et al., 1993). The period of weight transfer was defined as the interval between the first continuous increase in the vertical force signal relative to the baseline (line A, Figure 7) and the instant at which the foot lost contact with the force plate (line C) and defined as 100% weight transfer time. The peak amplitudes of forces and force rates of changes in the force \( \frac{dF}{dt} \), as well as the period required to reach maximal force during the weight transfer were analyzed as percentages of the weight transfer (% WT). The weight transfer was divided at the
timepoint of maximal lateral force (line B) into a thrust phase and an unloading phase. In addition to the descriptive analysis, qualitative evaluation of the shape of the peak vertical force was performed.

Figure 7 Time traces of a single one-leg stance performed by an elderly subject, showing the vertical and medial/lateral (M/L) ground forces (%BW) exerted beneath the leg to be lifted (swing leg) during weight transfer. In addition, the rate of change (%BW/s) in the M/L force is presented to illustrating the graphic relationship between force and its derivative. The period of weight transfer was defined as the interval between the first continuous increase in the vertical force relative to the baseline (line A) and the instant at which the foot lost contact with the force plate (line C). The weight transfer was divided by the peak of the M/L force (line B) into a thrust phase occurring prior to the maximal force (force rate pulse is zero) and a subsequent unloading (take-off) phase. The grey areas represent the force impulses of the M/L and vertical forces.

3.5 Statistical analysis

An overview of the statistical procedures employed in this thesis is presented in Table IV. In all cases the STATISTICA program for Windows, versions 6 and 7 (Statsoft Inc., Tulsa USA) was used for the actual analysis. The level of statistical significance was set at $P < 0.01$ (Study I) or $P < 0.05$ (Studies II-IV).

Parametric statistical analyses were utilized in all studies. Correlations were tested for by means of the Pearson product moment correlation (study I-II) and the strength of the correlation coefficients thus obtained classified according to Munro (2001):

<table>
<thead>
<tr>
<th>Correlation Coefficient</th>
<th>Description</th>
</tr>
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<tbody>
<tr>
<td>.00-.25</td>
<td>Little if any correlation</td>
</tr>
<tr>
<td>.26-.49</td>
<td>Low correlation</td>
</tr>
<tr>
<td>.50-.69</td>
<td>Moderate correlation</td>
</tr>
<tr>
<td>.70-.89</td>
<td>High correlation</td>
</tr>
<tr>
<td>.90-1.00</td>
<td>Very high correlation</td>
</tr>
</tbody>
</table>
The variability in force was transformed logarithmically to allow comparison between the younger and elderly subjects, who exhibited different degree of variance and skewed distribution. In Study II a multivariate approach to repeated analysis of variance (ANOVA) was used to compare the variabilities in force during the different time intervals and between the groups, as well as the relationship of the force impulse to the standing time. This same approach was employed in Study III to compare variabilities in force and data for different positions the front and rear legs and the younger and elderly groups. Wherever a significant interaction was observed planned comparison was subsequently performed. Differences in force impulse and temporal aspects between the younger and elderly individuals were analysed with $t$-test for independent groups (study II).

In Study IV the ANOVA was used to compare the values for different tasks, the stance and swing legs and the two different groups. If the sphericity was $P > 0.05$ the Fisher LSD post-hoc test was employed to analyze for significant differences between tasks; and otherwise planned comparisons were performed. The possible correlations between the baseline and other values were looked for with categorized scatter plots and the Pearson product moment correlation.

### 3.6 Ethical approval

All subjects were fully informed both verbally and in writing concerning the experimental procedure and were free to withdraw from the studies at any time.

The design of these studies was approved by the Human Ethics Committee at Karolinska University Hospital at Huddinge (Dnr: 374/00 and 500/00).
4 RESULTS

4.1 Functional reach (Study I)

The mean values of the reach distance obtained in connection with FR were 29.4 ± 5.4 cm and 27.9 ± 5.6 cm in the clinical and experimental measurements, respectively, i.e., the experimentally measured value was significantly smaller. As illustrated in Figure 8, the initial mean position of the CoP in the A/P direction was 36.6 ± 7.8 % of the BoS, starting with 0% at the edge of the heel and progressing to 100% at the toe, during FR, the CoP was initially displaced posteriorly, (by 8.1 ± 3.7 %) and thereafter anteriorly, (37.4 ± 10.6 %). The overall anterior displacement from the initial position (i.e., the difference between these two values = 29 ± 8.3 %) was used in the correlation analysis, which revealed that the correlation between the FR and this anterior displacement was low (r=0.38). More detailed analysis (n=15) revealed that during initial standing the position of the CoP in the M/L direction was located towards the side to be involved in reaching, by an average of 2.3 ± 2.2 cm left of the midline (see Figure 8). During reaching, the M/L CoP was displaced towards the reaching side, to an even greater extent (3.9 ± 2.5 cm) from midline. This analysis also revealed that even after the plateau of the CoP had been reached, the position of the finger continued to move forward by a total value of 7.3 ± 4.1 cm, which accounted for 30 % of the total movement during the FR (Figure 9).

The movement during FR was characterized by extensive bending of the trunk forward (by a mean of 32 ± 10º) and a small plantar flexion in the ankle (7 ± 3º). The FR distance and forward bending of the trunk demonstrated a moderate degree of correlation (r=0.67) whereas the correlation between FR distance and the angle of the ankle joint was low (r=0.39).
The onset of activation in connection with the right and left TA occurred after 57 ± 15 milliseconds and 56 ± 15 milliseconds prior to the onset of A/P forces in 16% and 13% of the trials performed by 13 and 12 subjects, respectively. In these trials the onset of activation of LG occurred significantly later than the onset of TA.

4.2 One-leg stance (Study II)

All of the younger subjects succeeded in performing OLS for 30 seconds in all three of the trials. In contrast, even though they were all healthy, only seven of the elderly individuals managed OLS for 30 seconds in all three trials.

Figure 10 illustrates the variability in the pattern of force observed with the younger and elderly individuals. The variabilities in the vertical and M/L forces decreased during the first 5 seconds (intervals 1-3) and remained unchanged thereafter, with the exception of an increase for the elderly group during interval 5. Although the variability in the level of force at the beginning of OLS was similar for both groups, this variability subsequently decreased to a greater extent in the younger subjects. Prior to lifting, all of the subjects exerted both vertical and lateral forces beneath the leg to be lifted in order to achieve the shift in weight. However, there was no significant difference in either the magnitudes of the vertical and lateral force impulses or in the total impulse time between the groups.
4.3 Tandem stance (Study III)

As seen from Figure 11, both groups displayed similar varabilities in the pattern of M/L force beneath the rear and front legs. A decrease in this variability beneath both legs occurred during the first 3-4 seconds of the task (intervals 1-5 and 1-4) in both groups, following which this parameter did not change during the remainder of the 30-second period of TS, with the exception of an increase in force variability between intervals 6 to 7 in the case of the younger group. Although the variability in the level of force beneath the rear leg was similar for both groups at the start of performance this parameter was reduced to a greater extent for the younger group during the first 4 time intervals after foot placement. In regard to the front leg, the younger individuals began with a lower variability in force level and during the first 5 time intervals this variability decreased to a more pronounced extent than for the elderly group, resulting in a larger overall force variability for the younger individuals.

With respect to the two different TS positions, the only evident difference between the rear and front legs, was a lower level of force variability in the front leg during TS\textsubscript{org} than during TS\textsubscript{mod}. Furthermore, the TA activity for both legs in connection with both TS positions was higher in the elderly group. Both groups supported twice as much weight on the rear than on the front leg during TS\textsubscript{org}. During TS\textsubscript{mod}, this difference was less pronounced, although, more weight was still supported by the rear leg.
4.4 Weight transfers (Study IV)

Comparison of the weight transfers that occurred in connection with OLS, TS and gait initiation in the younger and elderly subjects revealed that the onset of change in the vertical and lateral forces occurred simultaneously, but that the maximal vertical force was generally obtained more rapidly than the maximal lateral force. This difference was less pronounced in the elderly subjects than in the younger adults. The earlier force peak for elderly individuals in connection with all GRF indicated that this group had a shorter thrust and longer unloading phase than the younger subjects. For both groups, the timing of the force peak during OLS and TS was similar, whereas in connection with gait initiation this peak occurred later.

Multiple peaks of the force occurred during the unloading phase of OLS in the case of 11 elderly (18 trials in all) and 4 younger (6 trials) subjects and in 14 elderly (22 trials) and 4 younger (6 trials) subjects during TS. No such multiple peaks of force were evident in connection with gait initiation.

In connection with all three tasks, during the thrust phase there was no significant difference between the timepoint of maximal rate of change in the lateral force in younger and elderly subjects. However, during the unloading phase the maximal rate of change in the lateral force was observed at a significantly earlier timepoint in the elderly than in the younger subjects (Figure 12).
There were no age-related differences in the generation of forces, the rates of change in the force or their duration in connection with the three tasks involving weight transfer. The generation of vertical and posterior forces, the rate of change in the forces and their duration in connection with OLS and TS differed from the corresponding parameters for gait initiation, but the lateral force generated in connection with three of these tasks was similar.

An increase in the lateral force beneath the swing leg and a concomitant decrease beneath the stance leg occurred simultaneously and at the same phase of the tasks for both groups. The change in lateral force beneath the swing leg during OLS and TS was approximately three times larger than the corresponding change beneath the stance leg, while this relationship in the case of gait initiation was 5:4.

**Figure 12** The rate of change in the lateral force (see Figure 7) exerted by one elderly and one younger subject during tandem stance, expressed as a percentage of the weight transfer. Note that the first peak during the thrust phase appears at a similar timepoint with respect to % WT for both subjects, whereas the second peak occurs earlier in the elderly subject.
4.5 Weight distribution during initial standing (Studies I, IV)

Even prior to performance of the balance test and gait initiation, both the younger and elderly subjects prepared by distributing their weight asymmetrically between the two legs (see Figure 13). Prior to FR more weight was supported by the left leg, ipsilateral to the reaching arm. Before beginning OLS, TS and gait initiation, the subjects placed more weight on the anticipated stance leg than on the swing leg. In general, the elderly group displayed a more unequal distribution of weight than did the younger group, although the extent of inequality in weight distribution differed in connection with the various tasks.

Figure 13 Distribution of the weight (vertical force) between the swing (SW) and stance (ST) legs during initial standing prior to one-leg stance (OLS), tandem stance (TS) and gait initiation (Gait) and between the left (L) and right (R) legs during Functional reach (FR) in elderly and younger subjects. The values presented are mean ± SD.
5 DISCUSSION

The results presented here indicate that even healthy aging involves changes in postural control. Thus, age-related differences were observed with respect to the amplitude of postural steadiness during performance of OLS and TS; the timing of the force peak during the weight transfers involved in OLS and TS and initiation of gait; and the distribution of body weight between the legs prior to performance of the tasks. Healthy elderly individuals employ a compensatory reaching strategy during FR which results in low concurrent validity. In general, this thesis provides insight concerning what exactly is being measured by the balance tests examined here.

5.1 Effects of aging on balance

Although all of our elderly subjects were healthy older adults, age-related reduction in postural control was nonetheless evident. Performance of one-leg and tandem stance was seen to involve two phases: an initial dynamic phase during the first 3-5 seconds characterized by a rapid decrease in force variability, followed by a static phase during which a certain level of force variability is maintained. The elderly group exhibited a less pronounced decrease in force variability after the transition to the OLS or TS position (dynamic phase), indicating attenuated postural steadiness in this group. Moreover, the variability level in force associated with the static phase is influenced by the extent of reduction in force variability during the dynamic phase. A more pronounced decrease during the dynamic phase may improve the ability to maintain stability during the subsequent static phase.

In the case of TS, this lower initial decrease in the elderly group appeared to be compensated for by an increase in TA muscle activity. Winter and coworkers (1996) reported that the ankle invertors/evertors are the primary factors in the M/L balance during TS. The inability to compensate with increased muscle activity might thus shorten the period of time for which the standing position can be maintained.

Difficulties standing in either the OLS or TS position for a prolonged period of time might reflect problems with muscle strength and endurance and/or an inability to reduce the force variability during the dynamic phase. Decreases in muscle strength and endurance may result from aging of the muscles and/or the nervous, vascular and/or endocrine systems (Aniansson et al., 1980). Iverson and colleagues (1990) demonstrated that OLS duration and muscle strength in the hip flexors, extensors and abductors are positively correlated ($r=0.34 - 0.40$), i.e., as muscle strength increases, so does OLS standing time.

The ability of our elderly subjects to transfer weight from one leg to the other which is fundamental to many activities of daily living was also diminished. The age-related changes observed suggest that even active, healthy elderly adults may have some difficulties in connection with weight transfer, not because of any inability to generate the forces required but because of a disruption in the timing of these forces and their rates of change. In comparison with the younger adults, the elderly individuals
exhibited a consistently longer unloading phase, a larger temporal delay between attainment of the maximal vertical and lateral forces, earlier occurrence of the maximal rates of change in the forces and an elevated number of peaks of force during the unloading phase. However, it is noteworthy that adults 65-77 years of age can generate forces similar in magnitude to those generated by younger adults (24-40 years), i.e., age is not associated with physical decline in all respects.

The enhanced duration of the unloading phase observed in elderly, as compared to younger adults appears to be related to the release of forces as reflected in the multiple peaks in the force profile. Such multiple peaks suggest the occurrence of postural adjustments elicited by feedback control mechanisms (Brooks, 1986). In the present investigation, it appears likely that the elderly individuals began to unload too soon with respect to the displacement of CoM and that they therefore needed to make postural adjustments in order to be able to control the CoM. Mouchino et al. (1992) also found that the unloading phase is longer for normal individuals than for dancers, although these investigators examined the net CoP. Their explanation for the prolonged unloading involved inaccuracy in the internal representation of the position of the CoM.

In connection with our study, this would imply that the larger number of peaks exhibited by the elderly subjects during OLS and TS might be due to a lack of congruency between the internal representation and the actual equilibrium value, i.e., the new CoM position. The occurrence of corrections in the elderly individuals suggests that somatosensory components (probably involving plantar afferents and proprioception from the stance leg) participate in feedback control designed to achieve the new CoM position. The corrections made during weight transfer during OLS and TS can be regarded as successful, since postural steadiness thereafter was similar in both groups. Previously, more prolonged deceleration profiles in connection with arm movements have been observed in elderly subjects (Cooke et al., 1989). Since the GRF reflects the acceleration of the CoM, the prolonged unloading phase observed in our case could reflect a slower deceleration of the CoM.

The rate of the change of the GRF has been utilized to evaluate movements such as standing from a sitting position (Hirschfeld et al., 1999) and initiation of gait (Malouin and Richards, 2000). However, the timing of the maximal rate of change in force has not been fully characterized and the importance of an earlier maximal change during unloading from the perspective of aging remains somewhat obscure. The timing of the maximal rate of change in the force may be related to the velocity of the swing leg at the time it is lifted since gait initiation generally exhibited a later maximum. Unfortunately, we did not record any kinematics during OLS and TS which might have allowed us to verify this relationship.

As also shown previously (Blaszczyk et al., 2000; Henriksson and Hirschfeld, 2005) our elderly subjects displayed a more pronounced asymmetry in weight distribution during initial standing prior to performance of the tasks. This phenomenon may serve to compensate for an age-related elevation in the time required to achieve weight transfer (Horak et al., 1989a). In addition, our results reveal that the extent of this initial asymmetry is influenced by the nature of the task to come. In connection with the
balance tasks this asymmetry was larger than that associated with initiation of gait. The question that arises is, what happens if the nature of the task suddenly changes? It has been shown that elderly individuals demonstrate reduced ability to adapt to threats to their balance (Woollacott et al., 1986).

It is noteworthy that such age-related differences were rather limited in connection with the clinical measure of the balance tests. The moderate age-related changes documented here may have even greater significance in combination with other factors such as cognitive demands and adaptation to the environment. For instance, in connection with a dual-task paradigm, attentional demands were found to affect balance control in older adults (Shumway-Cook et al., 1997; Pettersson, 2005).

With regard to the methodology possible, selection bias yielding subjects who differ in some manner from the general population, may jeopardize the general validity of the findings made (Domholt, 2000). Domholt (2000) claims that individuals who are willing to act as research subjects may differ from the general population. In our present work, healthy elderly subjects were recruited from senior citizen organizations and those who were interested in participating were screened by telephone employing well-defined exclusion criteria’s prior to being summoned for an examination and testing. If we had recruited subjects via other organizations or groups such as exercise groups for senior citizens, the outcomes may have been different.

Most researchers in this field agree that when examining the effects of the aging process on human functions, these effects should be distinguished from those due to pathological conditions associated with diseases, even though chronic diseases are more common among the elderly. However, it has also been argued that it is impossible to draw a definitive boundary between what is a part of normal healthy aging and what constitutes disease or dysfunction (Rundgren, 1991). Thus, the findings from studies on aging can vary as a consequence of the inclusion and/or exclusion criteria employed.

The exclusion criteria (see Study I) employed here may not describe our elderly subjects in an entirely appropriate manner. This elderly group was active and healthy, which the exclusion criteria may not clarify. Thus, one possible threat to the external validity of our results may be that the study groups were not fully defined. Additional tests concerning level of activity, balance and gait would have provided a more complete description of our subjects. The questionnaire concerning physical activity did not elicit information concerning the intensity of this activity. It seems probable that, most of the elderly subjects participated in low-intensity activities such, as walking and golf, whereas high-intensity activities were more common among the younger adults.

Moreover, gender differences have not been examined in the present work, primarily because the groups of subjects were too small to allow division into subgroups. In connection with OLS differences between elderly men and women have been reported (Frändin et al., 1995).

Normalization of the EMG amplitude is essential to compensate for potential variations in anatomical factors and in the spacing and placement of electrodes, as well as to allow
comparison between different muscles and subjects (Lehman and McGill, 1999). Although, the RMS value is commonly expressed relative to the maximal voluntary contraction, this approach to normalization has been questioned (Marras and Davis, 2001). Thus, Yang and Winter (1984) found that normalization of muscle activity to sub-maximal exertions is more reliable. EMG activity has also been normalized as a ratio or percentage of the normal activity observed during walking (Hirschfeld and Forssberg, 1991; Tang et al., 1998) or standing (Lehman, 2002) which was the approach applied here. It is, of course, important to choose the appropriate technique for normalization since this choice may influence outcomes and clinical interpretation (Benoit et al., 2003).

Certain limitations in the kind of studies reported on here should be mentioned. This type of laboratory research always involves a compromise between measuring all of the parameters that might be necessary to answer the question being asked and practical considerations concerning resources, time and fatigue. It would have been desirable here to obtain additional information concerning the CoM and muscle activity in other muscles of the lower extremities by exploring the performance parameters associated with the clinical balance tests more thoroughly. Our experimental set-up was designed to give the elderly subjects more breaks between tasks and a smaller number of trials (only one leg was tested with respect to OLS and TS). Also an investigator stood close to the elderly subject throughout the experimental session to prevent falls or injuries.

Two different types of standardization were utilized here. In the case of FR the subjects reach forward with the left arm as the experimental set-up required that. In connection with OLS, TS and initiation of gait, the subjects were free to choose which leg to lift or move forward. This self-selection of the swing leg may constitute a limitation, since the choice of leg varied between the tasks. However, since the selections by the younger and elderly groups were similar, the major conclusions should not be affected by this factor. Moreover, the applicability of laboratory findings to a clinical setting can be questioned. However, our experimental setup was designed to resemble a clinical environment and the balance tests were performed in accordance with the same instructions as are used in the clinic.

I am also aware that medical screening in this study was limited. The use of prescribed medications was registered, but medications can affect elderly adults differently and the effects on our subjects were not evaluated by a specialist in geriatric medicine.

This thesis documents age-related changes in 65-80-year-old individuals and it would also be of interest to determine whether these changes persist or become more pronounced in the oldest old. Further work is also required to examine whether the phasing in elderly adults can be influenced by exercise designed specifically to train weight transfer. Healthy elderly adults can improve their balance (Ledin et al., 1990), but to our knowledge the effects of balance training on temporal aspects of postural adjustments have not been evaluated. Furthermore, if changes in phasing represent the first impact of aging, perhaps we should also look for such early changes in motor function in connection with mild cognitive decline. Earlier studies of persons with mild cognitive decline or very mild dementia have detected little motor impairment related
to functional activities common in daily life (Goldman et al., 1999; Pettersson et al., 2005)

Most of the age-related changes observed here in connection with the clinical tests were not apparent to an observer. These subtle changes may be the beginning of a gradual process of aging that may eventually limit the safe performance of everyday activities and possibly promote falls.

Hopefully, by attempting to distinguish between pathological and age-related changes, this thesis can improve our understanding of balance in elderly individuals. It is crucial for health-care professionals to understand that balance may be compromised even in elderly person without obvious signs of disease. At the same time, it should be remembered that the findings presented here are not necessarily applicable to individuals with impairments. However, knowledge concerning the performance of clinical balance tests by healthy young and elderly persons should provide the physical therapist with valuable tools and insights for solving the problems associated with balance disorders.

5.2 The validity of these clinical balance tests

5.2.1 Functional reach

Concurrent validity is an issue when comparing a measurement with a measurement standard (Domholt, 2000). This was performed here by correlating the reach distance during FR with the A/P displacement of the CoP, which suggested that FR does not reflect the limits of stability in healthy elderly individuals. Only 15% of the variation in CoP displacement could be explained by how far a subject could reach during FR, leaving 85% of the variation to be explained by other factors. Our findings also reveal that this reaching task is influenced more strongly by other factors, such as the movement of the trunk.

The mean FR distance of healthy elderly subjects observed here was similar to those reported by other investigators (Duncan et al., 1990; Wernick-Robinson et al., 1999). Notably, a similar anterior displacement of the CoP has been seen in connection with performance of a leaning task by elderly subjects (King et al., 1994), indicating that CoP displacement during reaching is not larger than during leaning. The moderate correlation found by Duncan et al. (1990) may be explained by the M/L displacement associated with CoP, since these investigators compared the sum of CoP displacements in both the A/P and M/L directions with the reach distance. FR is an asymmetric task and, therefore, a small lateral displacement of the CoP towards the reaching arm will occur together with the shift in weight. However, the additional results presented here demonstrate that the M/L displacement towards the reach side is relatively small. Another possible explanation for the findings by Duncan and coworkers (1) may be that there was a much larger age span (21–87 years) among their subjects. Our more detailed analysis of CoP and reach distance confirmed the existence of only a low correlation
between these parameters since the finger continued to move after the CoP had reached its maximum.

It appears that bending of the trunk forward was associated with a backward shift of the pelvis and ankle plantar flexion. This pattern of movement has been regarded as a compensatory postural adjustment during bending of the trunk with the hip and knee moving in a direction opposite to that of the upper trunk (Crenna et al., 1987). This is reminiscent of the hip strategy described by Horak and Nashner (1986).

The posterior displacement of CoP seen prior to the reaching movement is an APA designed to create a distance between the location of the CoP and CoM, i.e., a moment arm (Crenna and Frigo, 1991). Crenna and Frigo (1991) described the occurrence of APAs in the TA and soleus prior to initiation of a forward-oriented arm movement. In the present work, however, monitoring of the TA and LG by EMG revealed no clear pattern of muscle activation prior to the reaching movement, although, APAs in the CoP could be seen in the form of a posterior displacement prior to the onset of forward reach in all subjects. This suggests that suppression of the activity of the soleus or other muscles activity contributes to the posterior displacement of CoP.

Computer-based determination of the onset of movement has been well-characterized with respect to both inter- and intra-test reliability. However, the validity of this type of measurement is a different issue. Determination of validity requires a judgment concerning whether a meaningful physiological parameter is being measured. Therefore, EMG onsets and the EMG signal itself must also be examined visually by an experienced investigator in order to ensure validity (Difabio, 1987; Hodges and Bui, 1996). Based on a review of the literature concerning posture and the signal processing, a computerized algorithm involved two SD was employed in combination with visual inspection here.

If FR is to be used as a clinical test of balance, possible interference by mechanisms that compensate for decreased flexibility and strength must be considered. In our opinion a task involving leaning in both the A/P and M/L directions may provide a more valuable measure of the limits of stability. How best to measure stability limits in different directions remains, however, unclear (Murray et al., 1975; King et al., 1994; Wallmann, 2001) and further research in this area is required. Theoretically, it should be possible to determine these limits by monitoring the extent of ankle dorsiflexion, but reliable determination of these limits in a clinical setting with controlled trunk bending is a challenge for the future. One possibility could be to measure the angle of the ankle in connection with performance of a leaning task, when the body behaves like an inverted pendulum and joint motion is limited to the ankle joint.

Efforts have been made to develop the FR test further. In this connection Bauer and coworkers (1999) have explored the reliability and validity of a lateral reach test and Newton (1996) has investigated a test involving reaching in four directions for use with elderly adults. The NeuroCom (Smart) Balance Master System is designed to measure the limits of stability, but this equipment is expensive and not routinely available in geriatric clinics in Sweden.
5.2.2 One-leg and tandem stance

The concept of construct validity is based on the degree to which a test measures the parameters it was designed to measure and this is the most complex and difficult, and at the same time the most valuable type of validity (Johnstone et al., 1992). Construct validity concerns whether the abstract concept under investigation is being adequately measured (Domholt, 2000). As mentioned above, postural control is influenced by numerous physiological subsystems, as well as the nature of the task and environmental conditions (Figure 1). In order to maximize the construct validity of a balance test, awareness of exactly what the test measures is necessary. The systems theory is helpful in identifying the subsystems being measured by clinical balance tests.

Construct validity was not assessed fully in connection with this thesis work but an attempt was made to define what the tests are measuring and to distinguish the different subsystems that influence postural control in connection with these balance tests. Accordingly, we addressed the concept of postural steadiness in connection with OLS and TS. Assessment of postural steadiness on the basis of the time during which a static position can be maintained, assuming that better postural steadiness is associated with a longer duration, implies that postural unsteadiness should decrease linearly with the number of seconds that the position is maintained. However, our findings indicate that during the first 3-5 seconds after assuming the OLS or TS position, postural unsteadiness (i.e., force variability) decreased (the dynamic phase), and thereafter this unsteadiness did not change (static phase). This observation questions the theoretical basis for measuring OLS and TS for more than 5 seconds.

Although measurement of GRF has been shown to be valid and reliable (Goldie et al., 1989; Hanke and Rogers, 1992), assessment of postural steadiness as the SD of the GRF may be biased if the signal is composed of different frequencies (Tang and Woollacott, 1996). Thus, if the signal consists of superimposed low- and high-frequency components, the low-frequency components are not always detected in a short test but are detected and influence the SD during a test of longer duration. In this manner the SD would increase as the interval during which the force variability is measured is prolonged. In the present study, force variability was measured during the shortest intervals at the beginning and the longest intervals at the end. Any eventual bias would thus presumably mean that the differences observed are even more pronounced. However, no frequency analysis was performed here.

Thirty seconds of OLS and TS appear to involve postural adjustments, including a necessary decrease in postural unsteadiness followed by adjustments in muscle strength and endurance designed to maintain the position. Lichtenstein et al. (1990) suggested that the OLS is the most important position for measuring sway, since 20% to 40% of walking is performed on one foot. However, this claim can be questioned, since the single-support phase associated with gait is a dynamic task designed to re-locate the CoM ahead of the support base provided by the stance leg.
Moreover, the ability to perform OLS does not have to be measured for a long time. Assessment of an individuals ability to make the weight transfer to OLS or TS and hold this position for just a few seconds may be a more appropriate and functional measure of balance than determining the number of seconds that a person can maintain the position. Such assessment of dynamic weight shifts without substantial influence from muscle strength and endurance would provide clinicians with a better understanding of the problems that cause instability and falls in the elderly. Although muscle strength and endurance also affect balance (Adlerton et al., 2003) these parameters can be assessed more reliably and with higher validity using other tests. Moreover, another problem with using OLS to evaluate postural control is that many elderly subjects are unable to perform OLS for a long period (Thapa et al., 1994). Since the present studies only involved healthy subjects, additional investigation is required in order to more clearly delineate the pattern of force variability in subjects with difficulties standing on one leg.

The asymmetry associated with TS was reflected in the differences between the legs with respect to postural steadiness, vertical forces and TA muscle activity. Asymmetry with respect to limb loading during TS was evident in both the younger and elderly groups, with loading of the rear leg being almost twice as large during TS_{org}. Earlier studies have revealed tendencies towards such asymmetry in young subjects, but provided no actual values (Kirby et al., 1987; Nichols et al., 1995). Interestingly, both the young and elderly had difficulties in distributing their weight equally between the rear and front legs after being asked to do so, confirming that TS is not a task that involves equal weight bearing.

This asymmetry associated with TS might be due to anatomical limitations, since the position involved leads to functionally different leg lengths. The fact that the rear leg serves initially as the stance leg and the front as the swing leg may also contribute to maintaining more weight on the rear leg and simply using the front leg as a support leg. Another important factor in this context might be that alignment of the body segments and thus the CoM over one leg helps stabilize balance by decreasing the degree of torque around the joints of the leg. Vision may also be a contributing factor, since by placing more weight on the rear leg, the front leg can be seen and this may help stabilize the body in space (Patla et al., 1991). The difficulty in achieving equal weight bearing during TS appears to be not only a strategy for postural control but also due to anatomical limitations.

Since TS has been shown here to involve unequal weight bearing, in the clinical setting, this task should be performed with the right and left legs in both the rear and front positions. Furthermore, when assessing the capacity for unequal weight-bearing by subjects with motor disorders that affect the two legs to different extents, comparison of the TS parameters with the different legs in the rear position might be informative. The influence of hemiparesis and other unilateral disorders on weight-bearing in connection with TS needs to be elucidated. The tendency to stand with more weight on the rear leg may help us to understand activities in connection with which subjects prefer, or are required, to stand with one foot in front of the other.
These simple clinical tests may be useful as rough indications of the type of more precise diagnostic tests that should be performed. However, if the purpose of the clinical assessment of balance is to determine the underlying cause of a deficit in order to treat it effectively, evaluation of the numerous subsystems involved in postural control is essential. Moreover, if the goal is to determine whether a balance problem exists and whether treatment is needed, the clinician needs to characterize the functional consequences of the disorder (Horak, 1997). In this context, FR, a reaching task without a goal, and OLS and TS, which involve static standing, may not resemble functional everyday activities sufficiently to reflect limitations in a real-life environment.

As mentioned above, the ICF can be employed to classify various measures of outcome into different levels of function. However, postural control is not covered by the ICF. Balance is described in terms of the vestibular function (i.e., sensory functions of the inner ear related to determining the balance of the body) and involuntary reaction movements (i.e., righting reactions, body adjustment reactions and balance reactions) and categorized as on the body function level. This resembles a hierarchical model of motor control.

On the other hand, perhaps the ICF does not regard balance or postural control as a function, but rather as a system involving several functions, including neuromusculoskeletal functions related to movement, sensory functions and mental functions as in the systems approach. In order to encompass all aspects of health, it is necessary to consider measures of outcome within all areas of the ICF. Nonetheless, most clinical balance tests involve performance of an activity or task and thus appear to relate to the activity and participation level (under the mobility domain) and few of these tests provide information concerning a specific bodily structure or function.

Improvement and further development of balance tests designed to determine the underlying causes of balance deficits, thereby providing a basis for effective treatments is essential. Due the complexity of postural control, no single test can assess all aspects of this process (Thapa et al., 1996). Development of balance tests or, rather, a battery of tests that assesses and differentiates between all the different aspects of postural control is a major challenge for the future. The ideal balance assessment should also take into account the sensory environment, postural adaptation, the effects of cognitive distractions and predictive, as well as compensatory postural adjustments.
6 Clinical recommendations

- Aging in itself is associated with less effective balance (postural control).

- Balance or postural control is influenced by several physiological systems. Be aware of what you are measuring.

- When using the Functional reach test compensatory mechanisms should be taken into consideration.

- The first five seconds of one-leg and tandem stance provide the most valuable information.

- Tandem stance is not a task involving equal weight-bearing. Both legs should be tested in the rear position.

- In summary, do not assess balance solely on the basis of task parameters (number of centimeters or seconds). Also assess how the task is performed.
7 CONCLUSIONS

- The Functional reach test provides only a poor measure of the limits of stability in healthy elderly individuals. Movement of the trunk influences the reach distance to a greater extent than does displacement of the CoP. When employing the FR for assessment of balance, compensatory mechanisms should be taken into consideration. (Study I)

- The one-leg and tandem stance tests involve two phases: an initial dynamic phase (i.e., a rapid decrease in force variability during the first 3-5 seconds), followed by a static phase, associated with maintenance of a certain level of force variability. Age-related changes were seen in the decrease in force variability and ankle muscle activity. The present findings suggest that the first 5 seconds of OLS and TS are critical in assessing balance. (Studies II and III)

- At all ages, TS does not involve equal weight-bearing. Therefore, this test should be performed both with the right and left leg in the rear position. (Study III)

- Our findings indicate the presence of age-related differences in the timing of the changes in the forces and their rates of change beneath the feet for all three tasks involving weight transfer. In contrast, the magnitudes of the forces and their rates of change are independent of age. (Study IV)

- Even prior to initiation of the balance tests, preparatory changes in weight distribution occur. The nature of these changes is influenced by the task to be performed and they are more pronounced in elderly individuals. (Studies I and IV)
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