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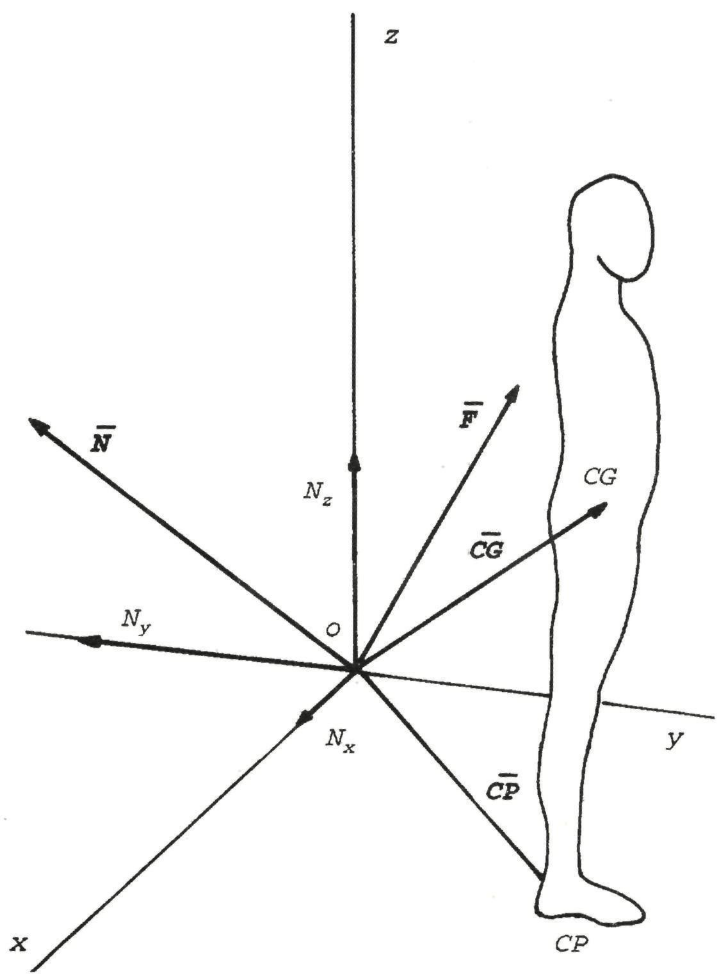
# CENTRAL ADAPTATION OF POSTURAL ORGANIZATION TO PERIPHERAL SENSORIMOTOR IMPAIRMENTS

## Clinical Experiments in Persons with Lower Limb Amputation and in Persons with Hereditary Motor and Sensory Neuropathy

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Alexander C.H. Geurts

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# TO PERIPHERAL SENSORIMOTOR IMPAIRMENTS

## Clinical Experiments in Persons with Lower Limb Amputation and in Persons with Hereditary Motor and Sensory Neuropathy

Een wetenschappelijke proeve op het gebied van de
Sociale Wetenschappen,
in het bijzonder de Psychologie

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ter verkrijging van de graad van doctor
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[1]psychologische funktieleer

[2]revalidatiegeneeskunde

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To my parents

For Heleen, Malou and Wouter

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## CONTENTS

# PROLOGUE

## THE SCOPE OF THIS THESIS

## BACKGROUND

Although there is a growing body of literature dealing with the mechanical and metabolic characteristics of standing and walking with a lower limb prosthesis, hardly any study has focused on the *restoration* of gross-motor control following lower limb amputation in terms of sensorimotor *coordination*. For instance, the literature about the control of vertical posture during standing and walking with a lower limb prosthesis is surprisingly scarce although safe balance is a relevant skill to be relearned by almost every one with a prosthesis, even by those who are merely trained to make (high) body transfers. One can only speculate about the reasons why so few clinical studies have dealt with this subject. An important reason could be that the sensorimotor recovery following peripheral lesions is often taken for granted both by the patient and by the clinical practitioner. This idea is supported by the fact that the bulk of the research in this area is focused either on prosthesis-related topics or on the improvement of surgical techniques and stump care, even though the functional outcome of the rehabilitation process is for a large part determined by the extent to which the patient is able to adapt to the gross alterations of mechanical constraints and sensory feedback caused by the amputation.

Superficially, it seems that clinical experience provides us with sufficient arguments to expect a fast and uncomplicated adaptation process in most persons with an amputation whenever proper peripheral conditions have been created; an expectation which is based on the integrity of the central nervous system (CNS). Motor recovery is often considered simply a matter of time. Due to the seemingly triviality of motor learning, little is known about the characteristics of the central processes that are responsible for (un)complicated sensorimotor reorganization following lower limb amputation, or about the neural mechanisms that underlie these processes. As a consequence, the literature leaves us with a lacuna regarding the most critical sensorimotor determinants of gross-motor recovery after lower limb amputation or, more generally, after irreversible damage to the neuromuscular system.

Yet, the restoration and preservation of gross-motor control in patients with peripheral neuromuscular deficits becomes an interesting subject of study if one realizes the complexity of the processes through which the CNS is ultimately capable of mastering the new situation. Besides its theoretical impact, the practical relevance of such investigation becomes evident as soon as we intend to develop valid procedures for functional assessment in order to evaluate the effects of different treatment policies on the restoration of gross-motor skills. At the level of the individual patient, fundamental knowledge about motor recovery after peripheral lesions may also be important in the case of a decreased learning potential, such as in the elderly, as well as in subjects with concomitant pathologies affecting the nervous system. The present thesis must be viewed as an attempt to provide some insight into the *central* processes that determine the recovery after structural damage to the *peripheral* sensorimotor system with emphasis on the implications for the development of process-oriented assessment procedures in clinical rehabilitation.

## AIM AND STRUCTURE

Because the (re-)acquisition of basic equilibrium reactions is of primary concern to reduce locomotor as well as body disposition disabilities (categories D-4 and D-5 of the International Classification of Impairments, Disabilities and Handicaps or ICIDH respectively) (WHO, 1980), this thesis is focused on the (re-)organization of *postural* control during normal upright standing in patients with peripheral sensorimotor impairments of the lower limbs. The basic idea is that such lesions will lead to an efferent-afferent postural disorganization by impeding the adequacy of prestructured muscular synergies as well as by distorting part of the somatosensory input from the legs. This sensorimotor disorganization requires the development of novel postural strategies by linking altered input patterns to new forms of muscular coordination. Consequently, a breakdown of well-established input-output patterns, which are for a large part subserved by highly automated "lower level" control mechanisms, induces a temporary need for "higher level" attention-invested control processes as well as for a re-weighting of sensory cues in postural control. When new postural strategies have been formed, a (partial) return to the premorbidly existing situation becomes possible.

This study is primarily conducted to elucidate the role of central processes in the adaptation to structural alterations of peripheral constraints. Because only part of this central adaptation is concerned with reprogramming of the (observable) motor output per se, sensory and cognitive processes are additionally examined as equally important (but less visible) factors in the restoration of postural control; these factors include the (changing) role of attention-demanding processes as well as of visual information in gross-motor control. Chapter 1 deals with the general importance of task manipulation in the clinical assessment of gross-motor skills to detect the changing contribution of such output-organizing processes to the observed motor behaviour.

The main emphasis of this study is on balance restoration following an acquired unilateral lower limb amputation above the ankle and below the hip joint. The situation after such an amputation offers the possibility to study the effects of a sudden and irreversible disruption of both somatosensory input and motor mechanisms on the control of posture, whereas the integrity of the CNS remains relatively unaffected. Chapter 2 is a theoretical overview on this subject starting with a discussion of the major peripheral sensory and motor impairments related to postural control, together with some of the possible compensatory mechanisms. In particular, a unilateral loss of ankle mechanisms and of somatosensory input from the leg are put forward as crucial deficits determining the need for a central adaptation. The latter part of this chapter deals with the main characteristics of the short- and long-term central adaptation following amputation as well as with some preliminary ideas on the underlying neural mechanisms contributing to this central reorganization. This chapter must be viewed as a prelude to the chapters 4 and 5.

Because in all the reported studies quiet two-legged standing is repeatedly assessed by means of a force platform, data on the intrasubject variability of several selected force-platform parameters in healthy adult subjects are reported in chapter 3. Thereafter, chapters 4 and 5 present the results of a clinical experiment directed at the changing role of cognitive and sensory processes, respectively, in the postural organization following lower limb amputation. Chapter 5 also discusses some reorganizational aspects of both static (weight distribution) and dynamic (control activity) asymmetry characteristics.

In addition to recovery from an *acute* disruption of the efferent and afferent organization of posture (amputation), this study is extended to the adaptation of postural organization in patients with a *chronic* degenerative condition, viz. type I or type II hereditary motor and sensory neuropathy (HMSN). Similar to the situation after an amputation, HMSN patients suffer from a combination of efferent and afferent impairments with the added complication of secondary foot-ankle deformities. Consequently, as in the amputation group, ankle mechanisms and somatosensory input from the legs are impaired, but now bilaterally and developed over years instead of days. Thus, the study of HMSN patients provides an opportunity to assess the influence of long-term central adaptation to slowly developing peripheral impairments on postural control and to compare these results with the data derived from the amputation group. Therefore, the investigated variables in chapters 4 and 5 are again examined in chapters 6 and 7, but now concerning the postural organization in persons suffering from HMSN type I or type II. Whereas chapter 6 deals with some basic (long-term) aspects of the postural organization in HMSN, chapter 7 is focused on the effect of a sudden change of peripheral constraints, i.e. the (short-term) influence of new orthopedic footwear, on balance automaticity.

Chapter 8 is an attempt to integrate some of the ideas derived from the work described in the previous chapters and to (re-)test these ideas by confronting the posture-control system of *healthy* subjects with different support-surface configurations, thus, "artificially" reducing the efficacy of well-developed postural strategies. This chapter has a predominantly theoretical scope in order to corroborate or falsify some of the earlier conclusions based on clinical data. Chapter 9 presents a more elaborate study dealing with the attention demands in balance recovery after lower limb amputation. Like chapter 8, the study of chapter 9 is primarily undertaken from a theoretical perspective. Chapter 10 must be viewed as an integrative and final contribution to this thesis. In contrast with the chapters 8 and 9, it is focused on the practical implications for clinical rehabilitation, in particular on the development of process-oriented procedures for the "routine" clinical assessment of gross-motor performance and recovery. Chapter 10 is concluded by a reference to the ICIDH and by indicating some promising directions for future research.

With the exception of chapter 10, which has the status of an epilogue, the entire thesis is a compilation of nine "independent" articles, either published, accepted or submitted. Chapter 2 is a slightly adapted version of a published article. As a result, some overlap exists between the different chapters; this, however, mainly concerns the "Methods" sections. Keeping in mind this inevitable consequence of the need for (early and timely) publications over an extended period of PhD research, the reader is kindly requested to exercise patience and consideration while working through this thesis.

## REFERENCE

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# CHAPTER 1

## FROM THE ANALYSIS OF MOVEMENTS TO THE ANALYSIS OF SKILLS:

### Bridging the Gap between Laboratory and Clinic

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**ABSTRACT**

Clinical assessment of balance and gait is discussed on the basis of an information processing approach, in which motor behaviour is seen as the result of a fine-tuned interaction between perceptual, cognitive and motor processes. From such a process-oriented viewpoint, stance and gait are regarded as complex skills that cannot be adequately assessed when the motor output is taken as an isolated object of study. It is argued that the assessment of a basic motor act under more complex environmental conditions will reveal essential information about the motor behaviour as a skill and, in the case of pathology, may give insight into essential aspects of the disability. Several suggestions are made with respect to creating complex task conditions. As an example, an estimation of the automaticity of balance control by means of a dual task procedure is discussed. The value of a process-oriented approach to disability assessment is stressed as well as the need for a skills laboratory in rehabilitation medicine.

## INTRODUCTION

Movement analysis refers to a broad field of activities, ranging from the study of highly coordinated fine motor skills, such as writing and drawing, to the study of gross motor abilities, such as standing and walking. This paper is focused on the assessment of balance and gait, which forms an important part of daily clinical practice e.g. in neurology, orthopaedics and rehabilitation medicine. Generally, a marked discrepancy exists between the availability of many new and advanced procedures for stance and gait analysis, and the application of these techniques on a routine basis in a clinical context. It is unlikely that a lack of sophistication of the available devices is responsible for this situation. In addition, practical consequences, in terms of time, space, money and expertise, can only account for a part of the lack of clinical application of instrumental stance and gait analysis (Messenger and Bowker, 1987).

However, a more important reason for the gap between the laboratory and the clinic could be the fact that stance and gait analysis procedures are generally aimed at the understanding of causes and mechanisms at a rather basic level, whereas many clinical questions require an overall assessment of motor behaviour in terms of skills instead of functions (Rozendal, 1989; Mulder, in print).

Particularly in the field of rehabilitation medicine, the need for assessment of balance and ambulation skills (e.g. for therapy evaluation) is greater than a detailed analysis of forces, joint angles and muscle activations.

In a rehabilitation context, general questions on standing and walking abilities predominate. Such questions include:

- Is the patient able to perform basic locomotor activities in his or her daily life (what distance, at what speed)?
- Is the patient also able to perform these activities under various environmental conditions (e.g. downhill, uphill, irregular surfaces, stairs)?
- Is the patient able to react to or anticipate sudden disturbances in the environment (e.g. avoiding an obstacle on the pavement)?
- Is the patient able to perform both locomotor activities and other tasks at the same time (e.g. walking and memorizing)?
- Does the patient suffer from a disproportional dependency on one particular source of sensory information (e.g. vision)?
- Is the patient able to walk without extreme energy costs in relation to his/her physical capacity?
- Is the patient able to take advantage of walking aids?

In terms of ICIDH (WHO,1980) it can be said that in rehabilitation medicine the development of reliable procedures for the assessment of disabilities is more important than the development of methods directed at the detailed analysis of impairments.

In clinical practice, assessments of motor dysfunctions are generally made by observation, often quite unsystematically. The limitations of such unstandardized subjective assessments, in terms of a relatively low reliability and sensitivity, have been discussed by several authors (Krebs et al., 1985; Sheikh, 1986; Geurts et al., 1988). Yet, until now, little attempt has been made by researchers to develop more reliable, valid and sensitive procedures for a standardized assessment of standing and walking skills. It must be realized, however, that such work would require a theoretical viewpoint which differs in essential aspects from that of the conventional medical approach. In the present text, some possible ideas on the objective assessment of gross motor skills will be presented, which may be helpful in bridging the gap between laboratory and clinic. They may provide some tools for the development of meaningful procedures for stance and gait analysis in a rehabilitation context. The ideas are strongly influenced by an information processing approach to human motor behaviour. Against this background, a disability can be regarded as the breakdown of behaviour at the level of skills instead of movements.

## THE ASSESSMENT OF GROSS MOTOR SKILLS

In a process-oriented view, movements are regarded as the end result of a fine-tuned interaction between perceptual, cognitive and motor processes (see for instance Prinz and Sanders, 1984; Schmidt, 1988; Colley and Beech, 1989). From such a perspective, assessment is of limited value if the motor output is taken as an isolated object of study, because in this way no insight is gained into the underlying response organizing processes leading to the observed movements or movement dysfunctions (Hulstijn and Mulder, 1986). Therefore, with respect to the assessment of gross motor dysfunctions in rehabilitation, it is not only important to record the kinematics of an act under basic conditions, but also to see how the same act is performed under more complex conditions. Indeed, it could be the case that essential information on functional progress during the rehabilitation of individual patients is not gained by studying standing and walking under relatively simple and predictable conditions, but in the study of the task performance under more variable and difficult environmental circumstances.

To meet these requirements, several task conditions should be added to the basic condition. The following kinds of task manipulations are suggested:

1. *Perceptual* manipulations: changing the perceptual input during a basic task (e.g. reducing or distorting the visual information may reveal an abnormally high

visual dependency within the multi-sensory control of balance and gait).

2. *Cognitive* manipulations: performing more than one task at the same time (e.g. walking while simultaneously making some calculations will give information on the level of automaticity of the locomotor act).

3. *Motor* manipulations: increasing the dynamic complexity of the basic task (e.g. walking at various speeds, starting, stopping, turning and avoiding provide information on the intrinsic flexibility of the performance and the ability to anticipate).

4. *Mechanical* manipulations: changing the environmental constraints (e.g. making the ground surface sandy or irregular provides information on the adaptability to external disturbances).

Each separate manipulation adds a specific factor to the basic task, and together these manipulations may represent a fairly complete picture of the skill under evaluation. It is assumed that results so obtained will be of more value in explaining and predicting daily life performance. To support this assumption, the cognitive type of manipulation will be further elaborated, because it is highly illustrative of the general theme of this paper.

One of the crucial characteristics of daily life performance is its multivarious nature, which means that very often more than one task has to be performed at the same time. Generally, for healthy adult subjects this is not a problem, because most of the motor skills (standing, walking, grasping) are more or less automated. As a result, these skills require a minimum of attention.

However, for those subjects with gross motor-skill deficiencies, the performance of routine acts (such as walking) may only be possible at the cost of considerable information processing capacity, resulting in the inability to perform these acts simultaneously with other, even non-motor, attention-demanding tasks (such as memorizing). The concurrent task interferes with the adequate performance of the main task. If the similarity between the tasks is minimized, the degree of dual task interference will reveal information on the level of automaticity of a skill (Wickens, 1989). A well-trained skill will not suffer from the simultaneous performance of a concurrent task, whereas the performance of a novel task will be very vulnerable to interference.

Rehabilitation is usually directed at the successful participation of the patient in the activities of daily life. Because there is a theoretical relationship between the degree of automaticity of gross motor skills and the ability to perform safely and flexibly in more or less complex daily situations, the degree of dual task interference could be a relevant parameter of the rehabilitation process. Therefore, the use of a dual task procedure will now be discussed by regarding the automaticity of balance control in lower limb amputees.

## DUAL TASK INTERFERENCE IN BALANCE ASSESSMENT

In healthy adult subjects, balancing and walking are sometimes regarded as actions requiring little or no cognitive processing (see for instance Turvey and Kugler, 1984; Warren et al.,1986). However, most pathological conditions differ from the normal situation in essential aspects. For instance, in cases of lower limb amputation, balance control must be reorganized by learning new strategies that make adequate use of stump muscles and compensatory hip and trunk mechanisms (Moncur, 1969; Murdoch, 1969). Such a learning process requires a substantial amount of information processing capacity. Fortunately, most lower limb amputees have a relatively intact central nervous system to meet these requirements.

However, during the first stages of learning, the capacity to perform tasks simultaneously will be severely reduced (Mulder, in print). Hence, if the above-mentioned notions on automaticity are valid, interference of a concurrent attention-demanding task should be clear at the start of the rehabilitation therapy, whereas this interference should be reduced at the end of a successful learning process. This hypothesis was tested in a study involving eight lower limb amputees, who received their first prosthesis after a recent amputation and who were individually trained until they had reached an acceptable level of independent ambulation (Geurts et al., 1991). None of the patients suffered from a marked perceptual or cognitive deficiency. Besides, balance control was tested in a group of healthy subjects, who had been matched for age and sex.

During a total registration period of thirty seconds, subjects were instructed to stand as still and as symmetrically as possible on a dual plate force platform connected to a micro-processor, which determined the virtual centre of the vertical ground reaction forces in a transverse plane ("centre of pressure" or CP). After a fifteen second period, a concurrent task was added to the balance task, while emphasis was given to maintain the same balance strategy (see for an example Figure 1). A modified version of a well-known psychological test (Stroop test) was selected as the concurrent attention-demanding task. Twentyfive coloured words (5 x 5) representing colour names were projected onto a white screen 1.5 metres in front of the subject. However, the colour names did not correspond with the colours in which the words were printed, e.g. the word "blue" was printed in red, and so on. The subjects were instructed to name the colours of the printed inks as fast as possible during the remaining 15 seconds, while suppressing a strong tendency to read the words. With three amputees, an arithmetical task was used as the concurrent task because they could not differentiate between all the colours of the Stroop test. These subjects were asked to subtract in threes as quickly as possible from a starting number between 50 and 100. The starting number was given to them immediately after the 15th second. They had to continue subtracting until the end of the procedure (30th second). Before every balance test procedure, the concurrent task was first practiced and then recorded in a sitting position as a single task performance.

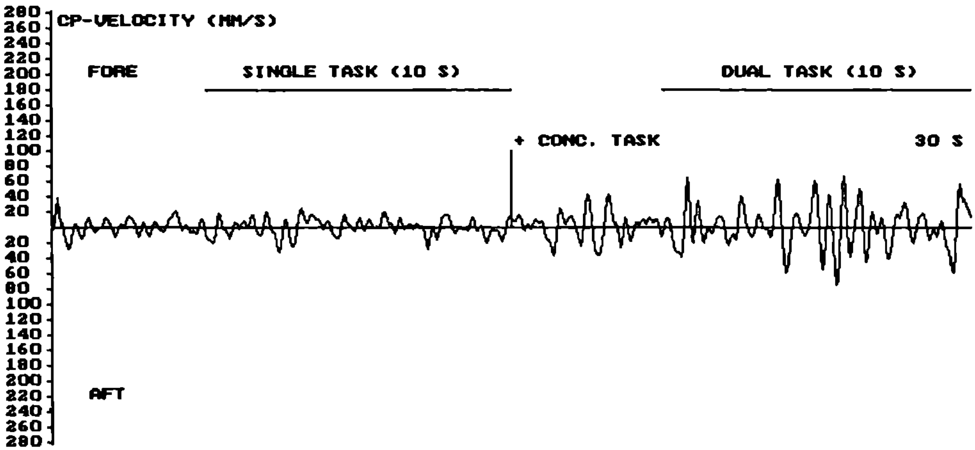


**Figure 1. A balance registration (fore-after sway) of one of the amputees at the start of rehabilitation.** Note the difference between the single and dual task performance.

The first five seconds of both halves of the balance registration were discarded from the analysis, because undesired visual or starting effects might have influenced body sway. The last ten seconds of the registration represented balancing in a dual task condition, while the ten seconds before the introduction of the secondary task represented the level of balance control without further demands (single task condition) (see also Figure 1). The efficiency of balance control was expressed as the root mean square of the CP-velocities derived from a ten second registration period. The number of items and mistakes scored on the Stroop (colours) or arithmetical task (subtractions) served as a measure of the concurrent task performance.

Although all amputees made clear functional progress during their rehabilitation, the results from this study showed that there was no significant difference in balance control in the single task condition between the start (one or two days after the first training with the definitive type of prosthesis) and the end of the rehabilitation process (just before the completion of the rehabilitation therapy). However, the dual task condition revealed a significant improvement in balance control over this period in both directions of sway. In comparison with the control group, the amputees showed significantly more dual task interference at the start of rehabilitation, again in both directions of sway. At the end of the rehabilitation process, this difference was no longer significant with respect to the fore-after sway. It must be mentioned that the performances of the concurrent task showed no significant difference between any set of paired data. The balance registrations (fore-after sway) of one of the amputees are presented, respectively recorded at the start of rehabilitation (Figure 1) and at the end of rehabilitation (Figure 2). They are fairly representative of the group results.

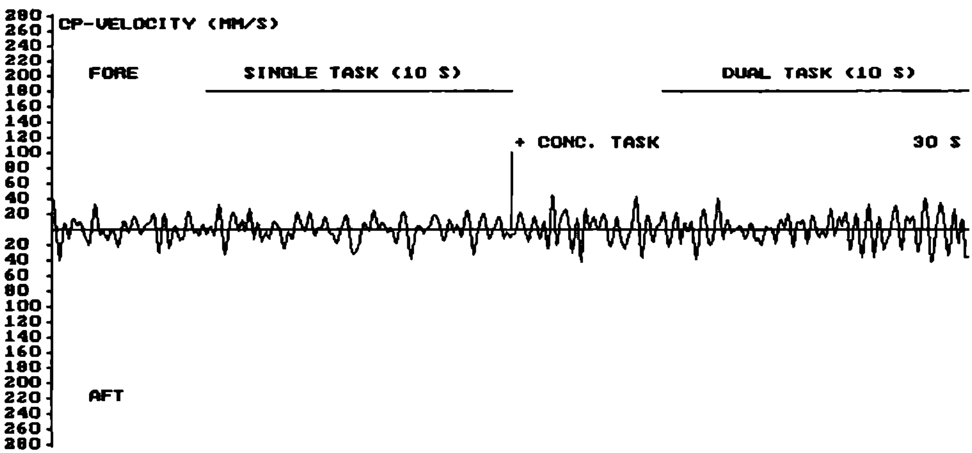


**Figure 2. A balance registration (fore-after sway) of the same amputee as in Figure 1 at the end of rehabilitation. Note the improvement of the dual task performance.**

## TOWARDS A SKILLS LABORATORY

The results of the summarized study are in accordance with the above-mentioned ideas. The results support the process-oriented view that assessments which take the integration of motor processes and cognitive processes as a starting point provide essentially different information compared with assessments which are solely focused at the motor processes In fact, in the present study the basic condition data give no information at all about the rehabilitation process. Indeed, the results indicate that at the start of rehabilitation the amputees maintained a reasonably upright posture as well as at the end of rehabilitation. This finding could force the investigator to conclude that the rehabilitation process had no effect on balance control. This conclusion, however, would be illegitimate, because the data show that the dual task performance improved significantly. In other words, maintaining an upright posture changed from an act requiring considerable attention to a more automated skill. The data demonstrate convincingly that measuring motor output isolated from the underlying information transforming processes may leave important aspects of the motor behaviour undiscovered. It is argued that these aspects are particularly relevant for rehabilitation medicine.

Although, in this paper, the main emphasis has been placed on the estimation of automaticity of gross motor skills by means of a dual task procedure, the reader will remember that aspects other than cognitive manipulations were suggested (motor, perceptual and mechanical). It can be expected that these other types of manipulation will lead to similar results, which means that they will provide additional and sometimes indispensable information on motor behaviour as a skill and on rehabilitation as a learning process It is therefore concluded that for rehabilitation medicine, a laboratory for the analysis of movements under simple and context-independent conditions seems of limited importance, whereas facilities to analyse motor behaviour at the level of skills and disabilities are urgently needed. This could be realized in a "skills laboratory," in which different types of complex conditions can be simulated, standardized and validated as relevant aspects of daily life activities.

Such a skills laboratory may contribute considerably to the bridging of the gap between the laboratory and the clinic. It can provide information for a meaningful assessment of therapy results. It may also help in making a prognosis on daily life performance. Furthermore, a process-oriented analysis of motor dysfunctions will lead to additional diagnostic information, as disorders at a perceptual, cognitive or motor level can be differentiated with clear implications for therapy choice. In addition, dual task procedures may give insight into the stage of a motor learning process in individual patients. In this respect, they can support the decision regarding benefit of (continuation of ) therapy.

Whether assessments must be made by means of objective or subjective procedures will depend on the characteristics of the desired parameters and on the required sensitivity. Any procedure may be potentially useful, as long as its reliability and validity have been assured.

The authors of this paper realize that there is still a long way to go before such a task taxonomy has been developed, by means of which ecologically valid statements and predictions can be made. It is argued that a process-oriented theoretical framework can be a useful instrument in attaining this goal.

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# CHAPTER 2

## REORGANIZATION OF POSTURAL CONTROL FOLLOWING LOWER LIMB AMPUTATION:

### Theoretical Considerations

This chapter is an adapted version of an article which will be published in Physiotherapy Theory and Practice (accepted for publication).

**ABSTRACT**

The reorganization of standing balance after a lower limb amputation is considered with emphasis on persons with an acquired unilateral amputation above the ankle and below the hip joint. In the first section, three major peripheral motor and sensory impairments are discussed: (a) a lack of ankle torque generation to restore equilibrium in the sagittal plane, (b) a lack of weight-shifting capacity to control posture in the frontal plane, (c) a distorted somatosensory input from the side of amputation. In the second part of the paper, it is argued that a lower limb amputation, as any other serious peripheral lesion, also affects the highest levels of the sensorimotor system, because the functional recovery after amputation requires a central adaptation to the alterations of peripheral motor and sensory conditions. A reduction in the cognitive regulation of posture as well as a decrease in visual dependency are proposed as two of the most critical parameters of the long-term central adaptation process and as relevant indicators of the restoration of (the safe performance of) gross-motor skills.

I had indeed to be made to rise, and stand, and walk--but how could I do so, and what indeed might happen, in a case such as mine, where to all the usual fears, inhibitions, hesitations, was superadded a fundamental disruption and "dissolution" of the leg, a disruption and dissolution at once physiological and existential?
Oliver Sacks, *A Leg To Stand On*

## INTRODUCTION

The reorganization of postural control in rehabilitation can be studied in many different types of patients. Persons with a lower limb amputation represent an interesting group because they suffer from peripheral neuromuscular and skeletal damage, leading to a sudden and irreversible disruption of both the efferent and afferent organization of posture. On the other hand, the learning capacity of most persons with amputation is relatively unimpaired because the integrity of the central nervous system (CNS) is largely unaffected.

Patients with lower limb amputation experience serious balance problems, particularly during the early phases of rehabilitation (Moncur, 1969; Murdoch, 1969). Yet, the literature on the *restoration* of balance control following lower limb amputation is scarce and fragmentary (Stolov et al, 1971; Gauthier-Gagnon et al, 1986; Geurts et al, 1990, 1991). Although in daily life postural control is nearly always related to movement (Talbott and Humphrey, 1979; Brooks, 1983), this paper will primarily focus on "static" balance because normal upright standing is an essential and basic skill to be re-learned by many patients. Emphasis will be laid on persons with an acquired unilateral amputation above the ankle and below the hip joint.

It should be mentioned beforehand that the balance problems in elderly persons with lower limb amputation are complicated by more diffuse degenerative impairments related to ageing. Since it is beyond the scope of this paper to discuss these impairments in detail, the reader is referred to studies which demonstrate clear effects of ageing on balance control, i.e. increases in onset latencies and disruptions in the temporal organization of muscle responses (Woollacott and Shumway-Cook, 1990), poorer coordination between postural reflexes and voluntary movement (Stelmach et al, 1989), increased vulnerability to deprivation of peripheral vision and ankle somatosensation (Manchester et al, 1989), a decline in sensitivity to optical flow (Warren et al, 1989), decreased capacity to stabilize the body in a dual-task condition (Stelmach et al, 1990), an increased dependency on external events for selective motor behaviour (Rabbitt and Vyas, 1980), as well as various additional risk factors for falls associated with old age (e.g. decreased activity level, psychotropic drugs) (Campbell et al, 1989).

The balance problems specifically related to lower limb amputation are largely based on the following *peripheral* motor and sensory deficits, which must be regarded as interdependent: (a) a lack of active ankle torque generation to restore equilibrium in the sagittal plane, (b) a lack of weight-shifting capacity to control posture in the frontal plane, (c) a distorted somatosensory input from the side of amputation. In the first part of the text, these motor and sensory consequences of lower limb amputation are further elaborated to better understand the basic balance *impairments* and some of the possible compensatory mechanisms.

In the second part of the text, it is argued that an amputation is not merely a peripheral disorder, but that it also affects the highest levels of the sensorimotor system. That is, a *central* reorganization process is required to adapt to the sudden alterations of peripheral constraints, thus, forming the basis for the restoration of gross-motor *skills*. Against this background, the strict distinction between peripheral and central processes will be defied as far as motor recovery is concerned. Furthermore, the distinction between efference and afference, which may hold for the peripheral sensory and motor deficits, is considered less adequate with respect to the central sensorimotor adaptation.

## PERIPHERAL MOTOR AND SENSORY DEFICITS: THE "IMPAIRMENT" LEVEL

### Balance Reactions in the Sagittal Plane

The important role of ankle torque generation about (a "frontal axis" through) the talocrural joint for restoring equilibrium in the sagittal plane is well-known. There is evidence for a distal-to-proximal sequence of muscle activation starting in the ankle joint muscles when the body is externally perturbed in the sagittal plane, at least when standing upright on a firm support surface (Nashner, 1976, 1977). Ankle torques are made effective mainly through vertical ground reaction forces acting at the feet, thus, counteracting the angular acceleration of the body caused by gravity. The *selection* of an ankle synergy, which is sometimes referred to as the "ankle strategy" (Nashner, 1985; Horak and Nashner, 1986), makes the body rotate primarily about the ankle joints.

Mathematically, postural sway about the ankle joints has been modelled as a simple inverted-pendulum system (Gurfinkel, 1973). The inverted-pendulum model has, however, been criticized because it ignores relative motion between various adjacent body segments and because it fails to predict correlations that should emerge from the model between body sway and physique variables, such as body height (Soames and Atha, 1980; Mizrahi et al, 1989). Still, the selection of an ankle synergy to control posture in the sagittal plane is a preferred strategy as long as the body is supported by a rigid floor and perturbations remain small. This notion is underlined by the close relationship between kinematic and force-platform data during tasks that evoke relatively limited amounts of body sway (Lestienne et al, 1977; Nayak, 1987; Van Asten et al, 1988a). In general, the efficacy of an ankle strategy is apparent because, by minimizing the corrective motions at the more proximal articulations, movements of the trunk and the upper extremities can be optimally controlled to perform other, e.g. manipulative tasks.

Unilateral damage to the ankle joint and its muscular structures is an inevitable consequence of *any amputation through or above ankle level*, leading to a considerable loss of stabilizing ankle torque in the case of an anteroposterior body perturbation. To some extent, this loss can be compensated through the contralateral limb. However, a unilateral ankle synergy will lead to a rotation torque in the transverse plane at the level of the pelvis. Consequently, pelvic and trunk rotators must be activated to counterbalance a torsional body moment about the vertical axis.

Compensatory activity through the non-amputated limb may be insufficient when the support surface does not provide adequate resistance, when the neuromusculo-skeletal condition of the intact limb is seriously impaired or during severe body perturbations (McCollum and Leen, 1989). Such compensation is even impossible in the case of bilateral amputation, as well as during single-support on the prosthetic limb in gait. Indeed, Nashner and Forssberg (1986) have found that anteroposterior postural strategies are similar during in-place standing and during the support phase of locomotion. In all these situations, the body will show a reversed muscular activation when it is perturbed in the sagittal plane starting at trunk level and descending towards the thigh muscles. This so-called "hip strategy" (Nashner, 1985; Nashner and McCollum, 1985) evokes primarily movements about the hip joints in a more direct attempt to bring back the centre of body mass within safe boundaries over the base of support. This strategy leads to an increase in the reactive shear forces acting at the feet. Hence, sufficient friction is required between the shoes and the support surface. The activated muscles are antagonistic to those used in the ankle strategy, although combinations of ankle and hip strategies may occur in different temporal relations (Horak and Nashner, 1986).

In addition, all persons with a *through- or above-knee amputation* are unable to make "suspensory reactions" (Nashner and McCollum, 1985; Nashner et al, 1979) through which the centre of gravity is displaced in the vertical direction by active control of knee flexion. In those subjects with an *unlocked* prosthetic knee, there is also a continuous hazard of destabilizing hip flexor activation at the amputated side. Particularly in the case of backward perturbation, when the vertical projection of the centre of gravity tends to fall behind the rotation axis of the knee, bilateral hip flexor activity will greatly facilitate destabilization of the prosthetic knee. Therefore, the sensorimotor system must suppress such inappropriate responses.

As for passive stability, most prosthetic foot components are stabilized in the anterior direction by a dorsiflexion stop, which substitutes for the loss of natural passive stiffness which is normally provided by the tarsus (Rozendal, 1986) and the ankle joint muscles (Grillner, 1972; Gurfinkel' et al, 1974).

**Balance Reactions in the Frontal Plane**

The organization of posture in the frontal plane basically differs from balance control in the sagittal plane. Normally, in the frontal plane, the limbs and pelvis together with the support surface and parts of the feet constitute a four-bar closed kinematic chain, which is controlled by the abductors and adductors of the hips and by the lower leg muscles (Rozendal, 1986). There is a clear influence of stance width and toeing-out angle on body stability in the frontal plane, both of which factors may show considerable variability from moment to moment. In contrast, there seems to be no long-term influence of leg-length discrepancy on postural sway (Murrell et al, 1991). Toeing-out of the feet increases the contribution of ankle torque generated about the talocrural joint to lateral sway control. Even with toed-out feet, no single joint exists in the tarsocrural joint complex with a sagittal axis (Van Langelaan, 1983), which is one of the reasons that the passive stiffness of this joint complex also contributes to lateral body stability (Rozendal, 1986). With greater distance between the feet, the lateral velocity of the different body segments decreases, which is most apparent up to an intermalleolar distance of approximately 8 cm (Day, 1990). With greater stance width, lateral sway control seems to be increasingly controlled at hip level. Indeed, by alternating activation of the hip abductors or adductors, the centre of body mass can be transferred from one leg to the other making use of the intrinsic mechanical stability of biped stance.

*With a lower limb prosthesis*, it is still possible to use the intrinsic stability inherent in double-limb support, provided one is able to bear substantial weight on the prosthesis spontaneously as well as voluntarily (Lord and Smith, 1984; Summers et al, 1987, 1988). If this ability is not achieved, lateral sway control will significantly depend on ankle torques generated mainly about (a "sagittal axis" through) the subtalar joint, which normally occurs only when standing with the feet together, in a tandem position, or during single-limb support.

However, the efficacy of such compensatory control of lateral body sway through the non-amputated limb is limited by the fact that ankle torques must be made effective through the relatively short lever formed by the foot in this direction. As a result, active trunk and arm displacements may become inevitable in the case of severe body perturbations. Hence, substantial spontaneous weight-bearing on the prosthesis as well as the ability to make fast and accurate voluntary weight shifts towards the prosthesis are prerequisites for safe balance reactions in the frontal plane, as well as for safe ambulation (Gauthier-Gagnon et al, 1986).

Normally, if the body is exposed to a reciprocal vertical displacement of the feet causing lateral body sway, it reacts with a flexion synergy at the side of upward tilting and with an extension synergy at the side of downward tilting (Nashner et al, 1979; Perham et al, 1987). In the case of a *through- or above- knee amputation*, the possibility of making such "lateral-tilting reactions" is seriously reduced, particularly if the direction of upward displacement is towards the prosthesis. In such instances, the disturbance must be counteracted by a tilt of the pelvis compensated by a lateroflexion of the trunk in the direction of the tilt.

**Afferent Control of Balance**

In healthy subjects, equilibrium reactions are under the control of three different sensory modalities, i.e the visual (Lee and Lishman, 1975, Brandt et al, 1986; Lestienne et al, 1977), vestibular (Allum and Pfaltz, 1985, Keshner et al, 1987) and somatosensory (Nashner, 1977; Diener et al, 1984; Magnusson et al, 1990) systems. Muscle, joint and cutaneous inputs from the legs are responsive to the contact forces and motions of the legs with respect to the support surface. These sensory inputs are regarded as particularly adapted to compensate for postural disturbances through fast stereotyped reactions of the leg muscles (Dietz, 1986; Amblard et al, 1990) characterized by a high degree of right-left coordination on a spinal level (Dietz and Berger, 1982). It should be realized that the orientation information provided by the support surface can be used to maintain posture only when its reference position is fixed or when its motions can be predicted in advance (Nashner et al, 1982). Nonetheless, there is a considerable task-dependent modulation of timing and gain of the above-mentioned postural "reflexes" (Nashner, 1976; Diener et al, 1983), which probably indicates that they are also mediated by rapid contributions from vestibular (Keshner et al, 1987) and visual (Nashner and Berthoz, 1978) inputs. Woollacott et al (1984) have reported that even advance (directional) information can modulate the long-latency components of calf muscle responses during the preparatory period for pushing or pulling arm movements through a supraspinal influence.

The sensory contributions of vestibular and visual information to postural control also comprise slower components. Against this background, there is a growing body of evidence for the idea that the vestibulum serves as the most important internal reference system to which other sensory inputs are spatially tuned in order to internally represent the three-dimensional properties of the body and the outside physical world (Nashner et al, 1982; Berthoz, 1991) On the other hand, the visual system has the lowest threshold for (head) motion detection and is, therefore, the most accurate source of sensory information (Brandt et al, 1986), provided the visual conditions are good and the eye-target distance is small (Paulus et al, 1989). However, whereas vestibular and proprioceptive afferents exclusively reflect self-motion, the visual system must decompose retinal image motion or "optical flow" into either self (body) motion or object (world) motion (Brandt et al, 1986). One mechanism contributing to this differentiation is the fact that the sensitivity for detecting self motion is limited to the frequency range below 0.3-0.4 Hz (Van Asten et al, 1988a, 1988b), whereas object motion can be detected at much higher frequencies. In addition, visual input is weighted with other sensory information in order to discriminate between body and object motion (Nashner, 1982).

*After a lower limb amputation*, a considerable amount of proprioceptive (lower leg muscles, ankle joint) and cutaneous (footsole) sources of input have been cut off at one side of the body. Furthermore, a probably endless stream of inappropriate stimuli from the stump tissues bombard the CNS, which is partly due to abnormal depolarizations of sectioned peripheral nerves (Wall, 1981; Ribbers et al, 1989). As a result, the contribution of other sensory modalities to balance control will be enhanced. Particularly the visual system has been shown to increase its contribution to postural sway reduction and prevention of falls in the case of somatosensory underspecification (Bronstein, 1986; Paulus et al, 1987), sensory conflict (Bronstein, 1986; Di Fabio and Badke, 1991), as well as during more dynamic postural tasks (Soechting and Berthoz, 1979). It is, therefore, not surprising that an increased visual control of posture is a longlasting consequence of both below- and above-knee amputation (Dornan et al, 1978; Fernie and Holliday, 1978; Holliday et al, 1978).

An integration of somatosensory input from the stump into the multi-sensory control of posture would lead to a decrease in visual dependency which would greatly facilitate the reacquisition of standing balance after lower limb amputation (Geurts et al, 1990). Although not a prerequisite, the integration of somatosensation in the control of posture seems to be strongly related to the safety of balance performance during daily activities (Shumway-Cook and Horak, 1986). Pew (1966) already indicated that at the start of a learning process motor performance is jerky, temporally unstructured and heavily dependent on visual feedback. In this line of thought, Adams (1971) suggested that a gradual shift from visual control towards proprioceptive control of movement might be a general characteristic of motor learning. Recently, the importance of continuous somatosensory input, in particular proprioception, from the feet to the extra-ocular eye muscles for building up a representational (topological) link between body space and extra-personal space has been stressed (Roll et al, 1991) as well as the essential role of a well-adapted kinaesthetic body representation or "body scheme" in postural control (Gurfinkel and Levick, 1991). Usually, the postural body scheme is thought of as a modality-unrelated sensorimotor (topological) representation of the body which would be partly genetically determined and partly acquired through learning. It includes a representation of the verticality as well as of the body geometry and its dynamics (Massion, 1992). It would also comprise a structural organization of a set of basic movement strategies preserved in memory. The existence of a central programme of postural regulation is strongly supported by observations of human postural behaviour under conditions of minimal gravitation (microgravity), during which postural strategies remain quite stable, even though they are based on the presence of a terrestrial gravitation factor. In contrast, the way in which these strategies are implemented by muscle synergies may change dramatically (Clément et al, 1984).

## CENTRAL ADAPTATION: THE "SKILL" LEVEL

On the basis of the far-reaching consequences of the above-mentioned anatomical and physiological impairments, it must be assumed that an extensive sensorimotor reorganization must take place to re-acquire safe balance control following lower limb amputation. Such a reorganization will only for a small part depend on peripheral healing processes in the stump tissues. Instead, a central adaptation process is needed to integrate the altered somatosensory information and to learn the appropriate postural responses (Geurts et al, 1991). Before we expand on the most important characteristics of the relearning of balance skills as well as on the possible neural mechanism underlying the long-term central adaptation following lower limb amputation, we will first discuss the normal adaptability of the postural organization in healthy subjects in order to understand some of the short-term adaptive mechanisms that contribute to postural control during the learning process.

### Normal Adaptability of Postural Organization

Nashner (1982) has argued that the temporal and spatial organization of muscular coordination (efference) varies with the mechanical configuration of the support base and that the relative weighting of different inputs in postural control (afference) varies with the availability and reliability of sensory information from the environment. He stated that adaptation to alterations of environmental context is accomplished by activating motor patterns that are to a large extent *preplanned* within the organizational structure for movement control; their activation being based upon assumptions about how the environment is structured.

Such type of short-term central adaptation goes beyond the variability of the postural "reflexes" mentioned in the previous section, because it refers to the *selection* of (a combination of) distinct postural synergies. Previous experience is used, in addition to sensory cues, to select the most appropriate response pattern from a limited repertoire of synergies in a particular situation (strategy) (Nashner, 1982; Nashner and McCollum, 1985). The fact that prior knowledge or "central set" modulates the magnitudes of postural responses to body perturbations indicates the "penetrability" of cognitively mediated feedforward processes even into these basic equilibrium reactions (Horak et al, 1989). Nonetheless, the postural responses themselves remain fast and more or less "automatic" also in unpredictable situations, because *learned* strategies are used in combination instead of new strategies (Horak and Nashner, 1986; Brown and Frank, 1987). Yet, when healthy subjects must adapt to a major environmental change, such as to microgravity conditions, a short-term *learning* process may come into play to carry out an operational recalibration (Clément et al, 1984).

The interplay between cognition and (the adaptability of) postural control is further indicated by studies on the interaction of posture and movement. When making rapid arm movements, preprogrammed postural adjustments precede any displacement or even any contraction of the prime movers, which indicates that postural activity can be part of the voluntary response and can be controlled via feedforward (besides feedback) mechanisms (Belen'kii et al, 1967; Pal'tsev and El'ner, 1967; Lee et al, 1980; Bouisset and Zattara, 1981; Friedli et al, 1984). Similar observations have been reported for fast voluntary movements of the upper trunk which appear to be consistently accompanied or preceded by postural leg muscle activity (Crenna et al, 1987). Cordo and Nashner (1982) have proposed a common motor organization of postural adjustments associated with focal movements and postural reactions elicited by e.g. support-surface perturbations. However, rather than feedback controlled reactions, feedforward controlled anticipatory adjustments are dependent on centrally stored information (cognition) about the task and the body in terms of its geometry and dynamics. The importance of this internal model of body dynamics (Frank and Earl, 1990) or postural body scheme (Gurfinkel and Levick, 1991) has been convincingly demonstrated. Hence, anticipatory postural activity must for a substantial part be acquired through learning (Massion, 1992).

The influence of prior knowledge and expectation on the timing of anticipatory postural adjustments as well as on the postural-focal onset latency has been extensively reported for rapid arm flexion movements (Horak et al, 1984; Zattara and Bouisset, 1986; Lee et al, 1987). These experiments show that, although posture and movement are well-tuned to each other, they are not tightly coupled. Instead, they seem to be controlled by relatively independent parallel processes and their relationship is strongly influenced both by mechanical demands and by behavioural conditions of the task (Brown and Frank, 1987, Lee et al, 1987).

Dietz and Noth (1978) nicely demonstrated that the pre-innervation of the triceps brachii muscles to decelerate the body after a self-initiated forward fall shows a similar timing with respect to the moment of landing whether vision is available or not, provided the falling height is known to the subject Only when the falling height is unknown, activation of the triceps brachii occurs at the start of the fall Similar preprogrammed innervation has been shown for the triceps surae muscles in hopping movements (Melvill Jones and Watt, 1971a) and in falls on the feet (Melvill Jones and Watt, 1971b, Greenwood and Hopkins, 1976)

Nashner and Cordo (1981) have reported that in conditions of postural stability (reaction-time) voluntary body sways could be executed as rapid as long-latency postural reactions to support-surface perturbations, although in a reversed proximal-to-distal order However, during postural instability, automatic postural reactions consistently preceded triggered voluntary movements. Although their data suggest an organizational distinction between automatic and voluntary movements, they defied the classical dichotomous classification in either voluntary or reflexive activities, because the earliest stabilizing muscle activations were found to be functionally related to the balance perturbations and did not conform to either of these extremes. The distinction between postural "reflexes" and voluntary movements seems even harder to make with regard to highly skilled balance performances (Nashner and Cordo, 1981).

When considering again the postural organization *following lower limb amputation*, it is a logical assumption that, during the early stages of learning, amputees employ the same adaptive mechanisms to voluntarily influence postural control as described above in healthy subjects. In other words, central adaptation after lower limb amputation is initially based on an intensified use of existing neural pathways that are normally responsible for a considerable degree of gross-motor flexibility and adaptability in healthy subjects in a continuously changing environment. Nonetheless, there is a major difference from the physiological situation. Under most circumstances, equilibrium reactions in healthy subjects are made fast and more or less automatically on the basis of a well-adapted internal model of body dynamics (Frank and Earl, 1990). As was mentioned above, postural control is normally subserved by selecting the appropriate responses from a limited set of prestructured synergies only requiring a (context-dependent) parametrization of timing and gain The cognitive regulation of posture is discontinuous because it merely concerns the selection of (a combination of) appropriate synergies (Massion, 1992) It is involved in a fast feedforward mode of control during a short period preceding a change of action or during a sudden alteration of mechanical or behavioural conditions (Droulez et al, 1985). Hence, few if any noticeable demands on attentional mechanisms are made. Furthermore, the regulation of posture will mainly concern spatial processing (Kerr et al, 1985).

However, in case of a *recent lower limb amputation*, the postural body scheme is no longer adapted to the actual physical state of the body, whereas some of the prestructured synergies have become inappropriate for the novel situation. As a result of maladapted sensorimotor representations, too little centrally stored information is available to adequately specify the motor output in advance. Hence, a cognitive mode of balance control will be more continuously active reflecting a shift towards an attention-invested feedback (or closed-loop) mode of control. As a result, postural control may impose considerable attention demands on a general-purpose central processor, reflecting a considerable loss of automaticity of postural control. It is apparent that some degree of balance automaticity should be restored to warrant functional recovery. Therefore, the re-acquisition of automatic balance reactions will be addressed in the next section using theoretical concepts derived from the literature on motor skill acquisition.

**The Re-acquisition of Balance Skills: Long-term Central Adaptation**

Motor learning in rehabilitation concerns the (re-)acquisition of skills and depends on a fine-tuned interaction between perceptual, cognitive and motor processes (Holding, 1989, Mulder, 1991a) "Cognition" in its most elementary sense refers to the basic ability of the CNS to store and retrieve information (Stillings et al, 1987) It is important to realize that an essential role of cognitive systems in skill acquisition does not always imply a marked influence of intentional processes on the central storage of knowledge (Masson, 1990) Yet, interaction with awareness and intention may occur dependent on the type of knowledge representation. Although the contribution of cognitive processes to motor control has been criticized by "action" theorists (see for a review of "the" motor-action controversy Meijer and Roth, 1988), a cognitive or computational motor approach offers a fruitful theoretical background for understanding the characteristics of human motor *learning*, particularly in the complex field of clinical rehabilitation (Mulder, 1985, 1991a, 1991b). It is also in good conformity with modern psychobiological concepts of the organization of learning in animals (Gallistel, 1990).

From a cognitive viewpoint, motor learning depends on central information processing (Schmidt, 1988) Because in most cognitive theories the available processing resources are assumed to be limited, resource competition may occur during the performance of more than one attention-demanding task leading to task interference (Neumann, 1984, Wickens, 1989) As for balance *after a lower limb* *amputation*, lack of automatic postural responses may result in clear interference effects with an attention-demanding, non-motor task (Geurts et al, 1991). It can be argued that the slowness of the balance reactions and their vulnerability to secondary task performance in amputees are related to the ongoing reorganization processes within the CNS as a reaction to the alterations of peripheral motor and sensory conditions. They are temporary manifestations of an attention-invested closed-loop control mode as a compensation for poorly adapted postural synergies and strategies. In this line of thought, the re-acquisition of standing balance with a lower limb prosthesis resembles the learning of any other novel motor skill and, therefore, obeys the "laws" of learning theory. In the first ("cognitive") stage of learning, the need for attentional resources to guide the motor performance is considerable (Adams, 1971; Johnson, 1984). There is a strong dependency on continuous sensory feedback concerning errors or success (Holding, 1965; Bilodeau, 1969; Salmoni et al, 1984). This early stage is characterized by error detection and correction of inappropriate postural responses in order to re-establish adequate input-output patterns. The necessary computations are initially responsible for a marked slowness of information processing. Through repetitions and training, the actual output patterns correspond more closely with the desired output patterns, so that gradually a faster and more automatic control of posture and movement develops on the basis of a long-term adaptation within central neural networks (Brooks, 1983; Melvill Jones, 1985).

In the amputee, there will be a gradually increasing ability to use the altered sensory input to execute novel patterns of coordinative muscular activity by the acquisition of *new* postural synergies and strategies. Indeed, as was discussed in the previous sections, leg and trunk muscle activity must be "re-tuned" to each other, inappropriate muscle activation must be suppressed, altered somatosensory input from the stump tissues must be integrated, and the boundaries of safe displacements of body mass must be explored in a three-dimensional space. The newly developed sensorimotor patterns are probably based on the reformation of a topological representation or spatial map (Gallistel, 1990, p103) that captures the invariant components of the altered relationships between input and output patterns (Masson, 1990). In close interaction with these sensorimotor recalibrations, the internal model of the upright body in space is adapted (Gurfinkel and Levick, 1991). At the level of these sensorimotor representations, which reside in the higher associative brain areas (Brooks, 1986; Mori, 1990), a clear distinction between afference and efference is virtually impossible, because there is no longer a strict causality chain between sensory and motor processes (Droulez and Berthoz, 1986).

Hence, on the basis of long-term adaptive changes within these sensorimotor representations, an adapted repertoire of preprogrammed postural reactions and adjustments is formed which, together with the (current) sensory feedback, can adequately specify the motor output in a particular situation. In this way, essential information about the altered mechanical and sensory constraints is internalized in a task-specific manner. Consequently, postural responses become faster and less dependent on immediate sensory feedback. As a result, a gradual shift from feedback to feedforward control becomes possible through a better specification of the motor output in advance based on previous experience (Brooks, 1983). The most striking feature of this central reorganization is the restitution of a considerable degree of balance automaticity leading to a decreased need of attentional resources. Consequently, activities of daily life that are performed in a vertical position are accomplished with greater flexibility and less chance of task interference due to a limited attention capacity. Moreover, the integration of altered somatosensory input from the stump tissues into the multi-sensory control of posture will lead to a significant decrease in visual dependency.

So far, we have only discussed the sensorimotor reorganization after lower limb amputation from a *behavioural* viewpoint. It is interesting to question what neural mechanisms might underlie the central adaptation after amputation of a body part. Against this background, some work has recently been reported addressing the reorganization of the motor cortex in subjects with an acquired or congenital upper limb amputation. It was found that motor evoked potentials to transcranial magnetic stimulation can be elicited from a larger cortical area and with a lower threshold in the muscles on the amputated than in those on the contralateral side (Hall et al, 1990; Cohen et al, 1991). It appears that there remains a substantial degree of central plasticity in the adult nervous system such that parts of the brain that premorbidly corresponded with distal muscles now control the muscles above the level of amputation. It has been suggested that this plasticity might be decreased in elderly subjects (Hall et al, 1990). Similar observations have been reported with respect to the somatosensory cortex which shows an immediate reorganization following sectioning of peripheral nerves or amputation of a body part (Merzenich and Kaas, 1982; Kaas et al, 1983). Here too, there appears to be a functional reorganization so that brain structures that have been deprived of their normal activating input become responsive to the stimulation of new parts of the body surface. Both the enlargement of motor representation areas and the somatosensory reactivation following peripheral lesions can just as well be explained by a cortical reorganization as by a subcortical reorganization within descending or ascending pathways. Moreover, both a (rapid) activation of previously inactive connections (unmasking) and a (slow) development of new synaptic contacts (sprouting) could be the underlying mechanisms (Kaas et al, 1983; Cohen et al, 1991). As for the reorganization of somatosensory cortex, it has been argued that this may be related to a self-organizing capacity within the CNS (Ritter and Schulten, 1986).

It is, nonetheless, unlikely that the abovementioned neural reorganization processes are fully responsible for the functional (balance) recovery following lower limb amputation, because they reflect the changing activity within modality-related neural pathways or cortical representations. Hence, they do not account for the formation of new linkages between sensory information and muscular coordination. Such a process of sensorimotor reorganization takes place on the basis of motor *learning* through active experience with a task (Masson, 1990). Although the neural mechanisms underlying learning processes are largely unknown (Brooks, 1983), it is quite certain that such long-term central adaptation is dependent on cortical activity (Ioffé et al, 1988). It requires changes within sensorimotor representations that form a critical link between the somatosensory and motor systems. It has recently been proposed by Kalaska (1991) that parts of the parietal cortex posterior to the primary somatosensory cortex contain such sensorimotor association areas. These areas of the brain would be particularly capable of representing movement in a coordinate system encoding kinematic (spatio-temporal) parameters.

In summary, we argue that a long-term central adaptation process is necessary to re-acquire balance skills after lower limb amputation. It has been proposed that a decrease in cognitive regulation of posture may be a critical parameter reflecting this sensorimotor reorganization. In addition, a re-integration of somatosensation from the amputated body side into the afferent control of posture should lead to a reduction in the dependency on visual information. We believe that either determinant of recovery is strongly related to the *safety* of balance performance. Indeed, attention demands in postural control should be minimal to prevent hazardous interference effects with almost every other (attention-demanding) task. The dependency on visual information in the control of posture should also be limited for several reasons. First of all, unlike other sensory inputs, vision is not always available and, if it is available, its stabilizing effect on posture may vary considerably dependent on environmental factors such as the structure and illumination of the visual scene (Brandt et al, 1985; Paulus et al, 1989; Van Asten 1988a). Secondly, the visual system has an inherent difficulty in the differentiation between self motion and object motion which, in the case of somatosensory deficits, may lead to inappropriate postural (over)reactions to motions of visual scenes. Lastly, a marked visual dependency in maintaining a vertical posture may lead to structural interference effects with many other tasks that require visual processing. From the theoretical considerations in this paper it can be concluded that any substantial peripheral lesion will eventually require a central adaptation to preserve an optimal functional performance. From such a viewpoint, the strict distinction between peripheral and central processes is artificial as far as it concerns motor control and recovery. In other words, central factors play a critical role in the functional recovery after peripheral damage to the sensorimotor system which has clear implications for the rehabilitation of patients with peripheral disorders.

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# CHAPTER 3

# INTRASUBJECT VARIABILITY OF SELECTED FORCE-PLATFORM PARAMETERS IN THE QUANTIFICATION OF POSTURAL CONTROL

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submitted for publication.

**ABSTRACT**

Platform stabilometry is increasingly applied to monitor or re-educate standing balance in clinical rehabilitation. Consequently, insight is needed into the validity, reliability and sensitivity of different force-platform parameters. This study is focused on the intrasubject variability being the major source of variance (unreliability) in the study of human motor skills. The intrasubject variability of several, commonly applied force-platform parameters was determined across ten repeated tests of quiet two-legged standing in healthy subjects to identify the most consistent and stable parameters in the quantification of postural control. The variability of the root mean square (RMS) amplitude, peak-to-peak amplitude, mean frequency and RMS velocity of the fore-aft and lateral components of the center-of-pressure fluctuations was investigated under varying (visual and cognitive) task conditions. The results indicate that all selected parameters show considerable intrasubject variability irrespective of the task context. Nonetheless, both the RMS amplitude and RMS velocity in either direction of sway do not demonstrate a significant trend across repeated tests. Among the selected parameters, the RMS velocity in the fore-aft direction shows the greatest intrasubject consistency, as well as a high sensitivity to e.g. visual deprivation. These findings support the reliability and validity of this parameter in the clinical quantification of postural control.

## INTRODUCTION

In the clinical field of rehabilitation, platform stabilometry is increasingly applied to monitor or re-educate standing balance, e.g. in patients with hemiplegia,[1,2] traumatic brain injury,[3] lower limb amputation,[4,5] or neuromuscular disease.[6,7] However, in order to compare the data of different studies, there is a need for standardization of the methodology.[8] In particular, information must be obtained about the validity, reliability and sensitivity of different force-platform parameters.[9]

This study mainly focuses on the problem of reliability reflecting the degree to which a parameter is free of error variance. Because the major source of "error" in biomechanical studies on human posture and locomotion is associated with the variability of the subject's performance,[10] we investigated the *intrasubject* variability, i.e. the variance of force-platform parameters across *repeated* tests of quiet two-legged standing in healthy adult subjects. Previous studies addressing this matter have yielded ambiguous results, varying from poor,[4,9,11] to moderate,[12,13] to good reliability.[3,14,15,16]

Our reasons for studying intrasubject variability are twofold. The primary reason is based on the idea that the least variable (or most consistent) parameters are probably well related to the neural control processes and, thus, may also be the parameters with the greatest validity in the clinical quantification of postural control. Second, it is important to know whether healthy subjects tend toward improved balance performance across repeated testing; a trend for which repeated clinical assessments should be corrected. To justify the selection of parameters under investigation, we need to refer to some theoretical aspects of stabilometry.

Usually, in stabilometry, the point of application of the resultant of the ground reaction forces is calculated from the force signals to study the motions of the center of force or "center of pressure" (CP) under the feet.[11-21] CP recordings can provide accurate information about postural control requiring relatively little effort. However, CP displacements reflect, besides the movements of the body center of gravity (CG) in the transverse plane, also the torques that are applied at the support surface to control the accelerations of body mass.[8,17,18] The application of ankle torques is particularly essential in the control of fore-aft equilibrium when standing quietly on a rigid floor,[22,23] which accounts for the close relationship between kinematic and force-platform data during tasks that evoke relatively small amounts of body sway.[15,18,24-27]

If, as in quiet standing, shear forces are sufficiently small compared to the vertical supporting force, the formal relationship between the CG and CP coordinates along the x-axis (frontal plane) and y-axis (sagittal plane) can be represented by

$$CP_x = CG_x - z_G \cdot F_x/F_z - \dot{H}_y/F_z \tag{1}$$
$$CP_y = CG_y - z_G \cdot F_y/F_z + \dot{H}_x/F_z \tag{2}$$

where $z_G$ refers to the vertical position of CG with respect to the support base, $\dot{H}_y$ and $\dot{H}_x$ refer to the time derivative of the moment of angular momentum of the body (which is equal to the total moment of force) about the y- and x-axis through the CG respectively, and $F_z$, $F_x$ and $F_y$ refer to the vertical and two horizontal components of the resultant of the supporting forces respectively[28] (see Figure 1). The CG terms in equations (1) and (2) represent the displacements of body mass and relate to the relatively large low-frequency components of the CP fluctuations (< 0.5 Hz). In contrast, the $\dot{H}_y/F_z$ and $\dot{H}_x/F_z$ terms comprise the accelerations of body mass (resulting in torque application at the support surface) and relate to the smaller high-frequency components of the CP fluctuations (0.5 - 5 Hz).[29] The $z_G \cdot F_x/F_z$ and $z_G \cdot F_y/F_z$ terms comprise the (limited) contribution of shear forces to the CP fluctuations during quiet two-legged standing which are also caused by the accelerations of body mass.[28]

It has been reported that the higher frequency components of the CP fluctuations show greater regularity than the lower frequency components both with regard to cycle duration and amplitude, whereas frequencies above 5 Hz are associated with tremor.[29] This consideration could imply that, in the control of quiet standing, not so much the *absolute* amplitude of the CG displacements is minimized, but the *relative* CG displacement *in time*, which integrates both amplitude and frequency. Whereas low frequency body sway may be irregular in amplitude (e.g. due to the need for changes of pressure between body segments, or for muscular activity to facilitate venous blood return from the lower limbs), the higher frequency CP fluctuations may show greater regularity in an attempt to minimize the CG velocity and acceleration.

If this hypothesis is true, it can be expected that the root mean square (RMS) amplitude of the CP displacements is more variable than the RMS CP velocity. Indeed, the RMS CP amplitude is frequency independent and is, therefore, mainly determined by the large low-frequency components of the CP fluctuations reflecting the relatively irregular, absolute CG displacements. In contrast, the RMS CP velocity is frequency dependent and reflects, besides the relative CG displacement in time, the time rate of change of torque application at the support surface. The latter factor relates to the power of the higher frequency components controlling the accelerations of body mass. Recently, Goldie and colleagues[9] made a case to use orthogonal force measures instead of CP measures to quantify postural control based on a higher test-retest reliability and sensitivity to changes in stance difficulty. However, this conclusion is somewhat premature because they did not include frequency dependent CP measures in their analysis. Moreover, reliability was based on just two consecutive recordings.

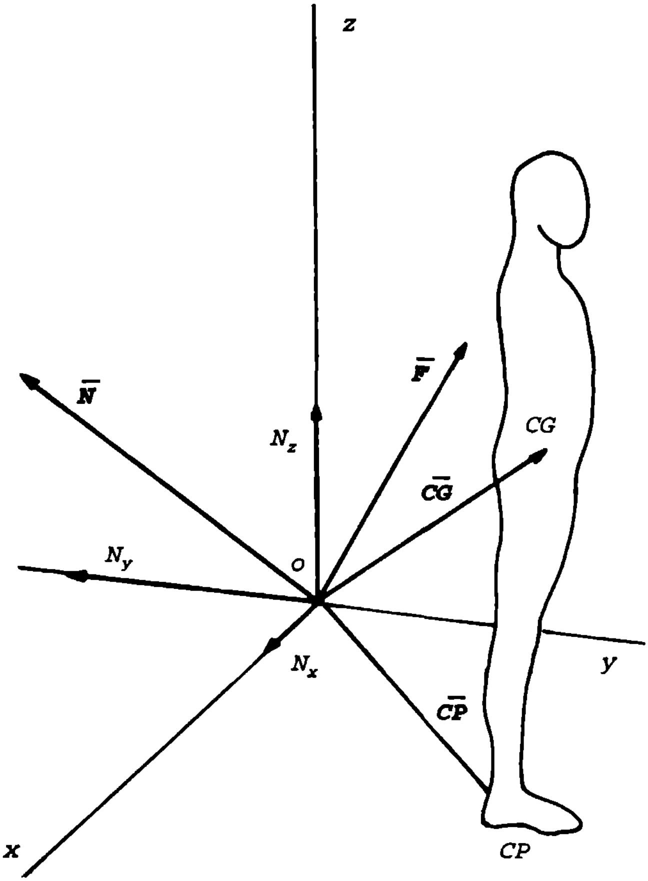


**Figure 1. A schematic representation of an upright body on a force platform (platform not drawn) with reference frame (O-x,y,z).** The forces acting on the body are reduced to a single resultant force ($\bar{F}$) and a single resultant torque ($\bar{N}$) acting at the origin O. $\overline{CP}$ and $\overline{CG}$ represent the position vectors of the point of application of the resultant of the ground reaction forces (CP) and of the body centre of gravity (CG) respectively.

In this study, we tested the intrasubject variability of the RMS CP amplitude and velocity across ten repeated tests of quiet two-legged standing. In addition, we investigated the variability of the mean frequency of the CP fluctuations, representing the proportion of RMS velocity to amplitude, as well as of the peak-to-peak amplitude, which is sometimes used to indicate the outer limits of the statokinesigram.[19] The variability of these parameters was determined separately for both directions of sway because substantial differences exist between the organization of fore-aft and lateral equilibrium control.[30] Two groups of subjects were assessed during two separate time periods employing different task conditions to examine whether the variability of the parameters is affected by the task context.

## METHODS

### Subjects

Two different groups of eight healthy adult subjects, four females and four males, participated in the study The first group was aged 24 to 65 years (mean age 44.3 ± 19.7), whereas the second group was aged 22 to 30 years (mean age 24.9 ± 2.4). All persons were active in daily life, however, none of them had any special balance skills. Informed consent was obtained from all subjects

### Equipment

Balance registrations were made with a firmly secured force platform consisting of two separate aluminum plates, each placed on three force transducers[a] (hysteresis and nonlinearity < 1%) recording the vertical ground reaction forces. Signals were processed by six DC-amplifiers[b] (nonlinearity < 0.1%). In the case of group one, the signals were led through six first-order low-pass filters with a cut-off frequency of 100 Hz and stored into a 8086 microprocessor (10 MHz) after a 14-bit AD-conversion at a sampling rate of 20 Hz. In the case of group two, the signals were led through six first-order low-pass filters with a cut-off frequency of 30 Hz and stored into a COMPAQ 386SX (16 MHz) after a 12-bit AD-conversion at a sampling rate of 60 Hz.

By means of moment-of-force calculations, the point of application of the resultant of the ground reaction forces in a two-dimensional transverse plane was determined for each sample. Because the horizontal components of the single

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[a]Load cells, type LM-100KA, Kyowa Electronic Instruments CO , LTD, Chofu-Higashiguchi Building 2F, 45-6, Fuda 1-chome, Chofu, Tokyo 182, Japan [b]RMP DC-amplifier, type MBP 6218, Elan Schaltelemente GmbH, Holzheimer Weg 50, D-4040 Neuss 1, Germany

resultant force could be assumed to be sufficiently small compared with the vertical component, the CP coordinates ($CP_x$ and $CP_y$) were calculated by means of the following equations

$$CP_x = -N_y / F_z \tag{3}$$
$$CP_y = N_x / F_z \tag{4}$$

where $N_x$ and $N_y$ refer to the single resultant torque about the x- and y-axis in the plane of force application respectively, and $F_z$ refers to the vertical component of the single resultant force[28] (see Figure 1). In this way, a maximum error of ± 1 mm in both directions was obtained. The CP coordinates were then passed either through a digital low-pass 5-Hz (moving average) filter (group one) or through a digital low-pass 6-Hz (Fourier) filter (group two) to eliminate high-frequency components due to noise or tremor.

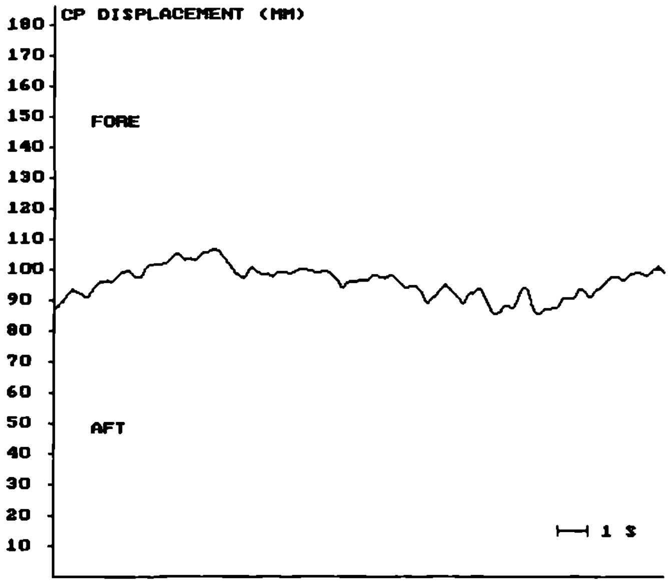
**Procedure**

During all balance tests, subjects stood on the force platform with their feet against a foot frame (medial sides of the heels 8.4 cm apart; toeing-out angle 9°). They were requested to clasp their hands lightly behind their back and to stand as still and symmetrically as possible.

Group one performed blocks of three 20-second balance tests in a fixed sequence: quiet standing (a) with the eyes open (EO), (b) with blurred vision (BV) (wearing milky-white spectacles preventing visual anchoring), and (c) with the eyes closed (EC) (reinforced by closed dark spectacles). Group two performed blocks consisting of two 30-second tests starting with (a) quiet standing without further demands (single task) (ST), followed by (b) quiet standing while simultaneously performing an arithmetic task (dual task) (DT). In the latter condition, subjects were verbally presented with a set of eight single-digit additions, equally timed over the 30-second period. The additions were selected at random and were followed by a sum that could be either correct or incorrect. Subjects were requested to verbally indicate the correctness of the addition problems immediately after each sum.

For both groups, *one* balance assessment consisted of *two* consecutive blocks. A one minute's rest was given after each test, whereas a longer pause was allowed between the two blocks. All subjects were assessed *five* times at two-weekly intervals, so that eventually *ten* identical blocks were obtained for each individual.

A

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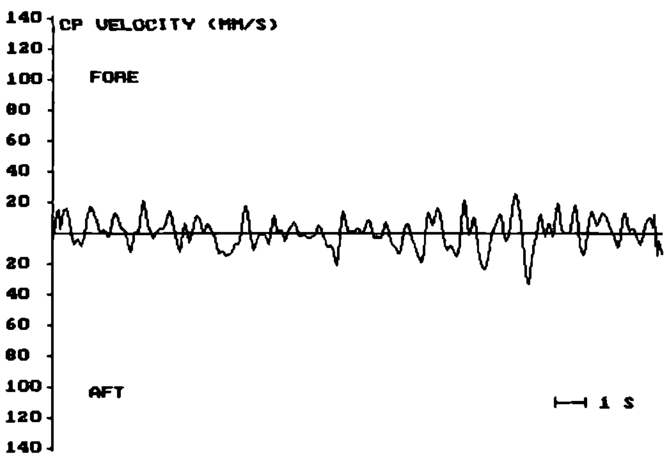


Figure 2. A typical 20-second registration of the fluctuations of the point of application of the resultant of the ground reaction forces (CP) for one of the subjects standing with blurred vision (see text). The CP displacements (Figure 2A) and CP velocities (Figure 2B) of the same registration are shown for the fore-aft direction of body sway. The zero point of the displacement curve corresponds to the rear of the support base.

**Data Analysis**

For every balance test, also the first time derivative was calculated from the CP displacements in the fore-aft (FA) and lateral (LAT) directions separately (see for an example Figure 2). Then, the first and last seconds of every data array were discarded from further analysis to eliminate possible artefacts due to the data processing. All balance parameters were calculated for the FA and LAT sway components separately. The root mean square was derived from both the CP displacement curves (RMS amplitude or Acp) and the CP velocity curves (RMS velocity or Vcp). The mean frequency (Fcp) was determined by means of the following approximation[11]

$$Fcp = Vcp \,/\, (Acp \times 4 \times \sqrt{2}) \qquad (5)$$

In addition, the peak-to-peak amplitude (PPcp) was determined by subtracting the minimum from the maximum CP coordinate of the displacement curves.

Of all selected parameters, only the Vcp is substantially influenced by higher frequency components and, thus, by the noise produced by the measuring system. Even after filtering, system noise adds a constant factor to the Vcp and may erroneously reduce its variability. Therefore, we corrected all Vcp values by subtracting the average Vcp that was obtained from 12 recordings with a bag of gypsum (40 kg) placed in the middle of the force platform. This procedure was done for both experimental configurations separately.

## RESULTS

**Variability**

In order to test differences in intrasubject variability among the selected parameters for each condition in either group, a *coefficient of variation* (CV) was calculated using

$$CV = s \,/\, m \times 100 \qquad (6)$$

where m and s represent the mean and standard deviation of each parameter over ten repeated blocks respectively.[11] The CVs of groups one and two, averaged over eight subjects, are presented in Tables 1 and 2 respectively.

First, we tested a possible effect of condition on the variability of the individual parameters by means of a one-way analysis of variance with repeated measures for condition (EO vs. BV vs. EC in group one; ST vs. DT in group two). In this way, no significant condition effect was found for any parameter in either group.

Table 1: Mean Coefficients of Variation from Ten Repeated Blocks in Group One ($n$ = 8) (%).

| Parameter | Condition | | | |
|---|---|---|---|---|
| | Eyes Open | Blurred Vision | Eyes Closed | All |
| *Lateral direction* | | | | |
| Acp[a] | 39 | 36 | 36 | 37 |
| Fcp[b] | 31 | 26 | 30 | 29 |
| Vcp[c] | 35 | 35 | 35 | 35 |
| PPcp[d] | 35 | 35 | 32 | 34 |
| *Fore-aft direction* | | | | |
| Acp[a] | 37 | 33 | 33 | 34 |
| Fcp[b] | 36 | 30 | 32 | 33 |
| Vcp[c] | 24 | 23 | 20 | 22 |
| PPcp[d] | 29 | 27 | 27 | 28 |

[a]RMS amplitude of the CP fluctuations. [b]Mean frequency of the CP fluctuations. [c]RMS velocity of the CP fluctuations. [d]Peak-to-peak amplitude of the CP fluctuations.

Table 2: Mean Coefficients of Variation from Ten Repeated Blocks in Group Two ($n$=8) (%).

| Parameter | Condition | | |
|---|---|---|---|
| | Single Task | Dual Task | All |
| *Lateral direction* | | | |
| Acp[a] | 27 | 26 | 27 |
| Fcp[b] | 22 | 24 | 23 |
| Vcp[c] | 29 | 29 | 29 |
| PPcp[d] | 26 | 24 | 25 |
| *Fore-aft direction* | | | |
| Acp[a] | 30 | 28 | 29 |
| Fcp[b] | 29 | 25 | 27 |
| Vcp[c] | 21 | 21 | 21 |
| PPcp[d] | 25 | 25 | 25 |

[a]RMS amplitude of the CP fluctuations. [b]Mean frequency of the CP fluctuations. [c]RMS velocity of the CP fluctuations. [d]Peak-to-peak amplitude of the CP fluctuations.

Consequently, we calculated the differences in variability among parameters over conditions. The CVs averaged over subjects and conditions are also presented in Tables 1 and 2 for groups one and two respectively (All). The data of both groups clearly indicate that, in the FA direction of body sway, the

Vcp appears to be the least variable parameter (CV = 22% and 21%), whereas the Fcp shows the greatest consistency in the LAT direction of sway (CV = 29% and 23%). The significance of these results was statistically tested by means of paired *t*-tests.

The data of group one revealed that, with regard to the FA sway, the Vcp had a lower variability than the Acp, Fcp and PPcp, $t(23) = 6.04, 4.42, 3.37$ respectively, $p < .005$. In addition, the PPcp was less variable than the Acp and Fcp, $t(23) = 5.69, 2.43$ respectively, $p < .05$. As for the LAT sway, the Fcp had a lower variability than the Acp, Vcp and PPcp, $t(23) = 4.15, 2.53, 2.44$ respectively, $p < .05$, whereas the PPcp was less variable than the Acp, $t(23) = 3.23$, $p < .005$.

The CVs of group two generally showed a somewhat lower variability, although the pattern was essentially the same as that of group one. With regard to the FA sway, the Vcp was less variable than the Acp and Fcp, $t(23) = 3.14, 2.38$ respectively, $p < .05$, whereas the Vcp marginally differed in variability from the PPcp, $t(23) = 1.97$, $p < .07$. In addition, the PPcp was less variable than the Acp, $t(23) = 3.03$, $p < .01$. As for the LAT sway, the Fcp merely showed a lower variability compared to the Vcp, $t(23) = 2.35$, $p < .05$.

The statistics indicate that, for sagittal plane balance, the Vcp is the most consistent of all parameters under investigation irrespective of the task condition. As for balance in the frontal plane, the statistics of group two are less discriminative than those of group one, although the Fcp yields the lowest CV in both groups.

**Trend**

If the intrasubject variability of a parameter is sufficiently low across repeated tests (consistency), its *actual* values must show an overall stability, i.e. there should be no significant trend in time. Preferably, a consistent and stable parameter should also be sensitive to task manipulations that are known to affect postural control. To investigate these aspects of stability and sensitivity, the Acp and Vcp values were further tested for an effect of repeated testing as well as of task condition. Indeed, both amplitude and velocity information are generally accepted and recommended for quantifying postural control.[8]

First, the corresponding data of the two blocks within one balance assessment were averaged for each condition in either group. In this way, data structures were obtained of five repeated assessments at two-weekly intervals, which are presented in Tables 3 and 4 for groups one and two respectively. For reasons of simplicity, the data of the BV condition in group one were discarded

**Table 3: Means and SDs of Sway Data from Five Repeated Assessments in Group One ($n$ = 8).**

| | Weeks | | | | |
|---|---|---|---|---|---|
| Parameter | 1 | 2 | 3 | 4 | 5 |
| *Eyes open, lateral direction* | | | | | |
| Acp[a] | 1.6±0.7 | 1.4±0.2 | 1.5±0.4 | 2.0±0.5 | 2.0±0.4 |
| Vcp[b] | 3.7±1.1 | 3.4±0.6 | 3.2±1.0 | 3.8±1.9 | 3.9±0.7 |
| *Eyes closed, lateral direction* | | | | | |
| Acp[a] | 1.7±0.6 | 1.5±0.6 | 1.4±0.5 | 1.6±0.5 | 1.8±0.6 |
| Vcp[b] | 3.9±1.8 | 3.1±1.2 | 2.7±0.8 | 3.8±1.6 | 3.6±0.9 |
| *Eyes open, fore-aft direction* | | | | | |
| Acp[a] | 2.6±0.8 | 3.3±1.0 | 2.7±1.0 | 3.4±0.9 | 3.5±1.0 |
| Vcp[b] | 6.1±1.6 | 6.3±1.7 | 5.9±1.4 | 6.6±1.8 | 6.7±1.0 |
| *Eyes closed, fore-aft direction* | | | | | |
| Acp[a] | 3.4±0.8 | 3.7±1.7 | 3.6±1.0 | 4.6±2.8 | 3.5±1.2 |
| Vcp[b] | 7.3±1.2 | 7.5±1.4 | 7.0±1.3 | 8.0±1.8 | 8.3±1.1 |

[a]RMS amplitude of the CP fluctuations (mm). [b]RMS velocity of the CP fluctuations (mm/s).

**Table 4: Means and SDs of Sway Data from Five Repeated Assessments In Group Two ($n$ = 8).**

| | Weeks | | | | |
|---|---|---|---|---|---|
| Parameter | 1 | 2 | 3 | 4 | 5 |
| *Single task, lateral direction* | | | | | |
| Acp[a] | 1.9±0.6 | 2.3±0.6 | 1.9±0.6 | 2.0±0.7 | 2.0±0.5 |
| Vcp[b] | 4.6±1.7 | 5.0±2.1 | 4.4±1.8 | 4.5±1.9 | 4.7±2.4 |
| *Dual task, lateral direction* | | | | | |
| Acp[a] | 1.6±0.5 | 1.6±0.5 | 1.7±0.5 | 1.7±0.5 | 1.8±0.5 |
| Vcp[b] | 4.6±2.2 | 4.9±2.2 | 4.3±1.4 | 4.4±1.5 | 4.3±1.4 |
| *Single task, fore-aft direction* | | | | | |
| Acp[a] | 2.9±1.1 | 3.5±1.8 | 2.9±1.1 | 2.8±1.2 | 2.7±0.8 |
| Vcp[b] | 5.6±0.7 | 6.4±1.3 | 6.1±1.8 | 5.8±1.5 | 5.7±1.7 |
| *Dual task, fore-aft direction* | | | | | |
| Acp[a] | 3.7±1.3 | 2.8±0.9 | 2.8±0.9 | 2.9±1.1 | 3.2±1.0 |
| Vcp[b] | 5.7±1.1 | 6.0±1.2 | 5.6±1.1 | 5.7±0.9 | 6.1±1.5 |

[a]RMS amplitude of the CP fluctuations (mm). [b]RMS velocity of the CP fluctuations (mm/s).

from this analysis because they were almost identical to the data of the EC condition. Each parameter was tested in a two-way analysis of variance of Time (five assessments) x Condition (EO vs. EC in group one; ST vs. DT in group two) with repeated measures on each factor.

The Vcp data of group one did not show significant main or interaction effects of time or condition on the LAT sway. The FA sway results showed a main effect of condition, $F(1,7) = 7.08$, $p < .05$, with greater Vcp values in the absence of vision. However, the main effect of time and the Time x Condition interaction were not significant. The Acp data of group one did not reveal any significant main or interaction effect, although the Acp values on the FA sway were generally higher with the eyes closed. It appeared that, in the FA direction of body sway, the Vcp was more sensitive to visual deprivation than the Acp.

The Vcp data of group two yielded no significant results. Hence, no trend in the Vcp values was found, nor an effect of concurrent mental activity. As for the Acp data, there was merely a significant main effect of condition on the LAT sway, $F(1,7) = 37.01$, $p < .001$, indicating less sway in the dual-task condition.

## DISCUSSION

The aim of this study was to determine the intrasubject variability of several, commonly applied force-platform parameters to identify the most consistent and stable parameters in the quantification of postural control during quiet two-legged standing. We selected only a few of the possible parameters, since many of them show considerable overlap. For instance, both RMS amplitude and mean rectified amplitude give a measure of the average absolute displacement from the mean position, whereas RMS velocity, mean rectified velocity and sway path are all measures of the average absolute displacement in time.[3,8,11,12,20] The sway area, which is a combination of sway amplitude and sway path,[11,19] was not considered because it confounds directional information and does not provide new independent information.[3,11,12]

In the present study, we examined the variability of the RMS amplitude (Acp) and RMS velocity (Vcp) in addition to the variability of the mean frequency (Fcp) and peak-to-peak amplitude (PPcp) of the CP fluctuations. We split these measures into their FA and LAT components because of the considerable differences between equilibrium control in the frontal and sagittal planes both from a mechanical viewpoint[31] and from a neural control perspective.[26,27,30] We did not analyze the power within specific bandwidths of the frequency spectrum because of the problems encountered in the identification and interpretation of distinct frequency patterns in healthy subjects.[32,33] Furthermore, power spectral analysis in stabilometry is mostly applied in a diagnostic sense to identify specific characteristics associated with nervous diseases,[30,34] whereas in clinical rehabilitation one is often interested in the functional assessment of postural control as a basic ability.

Theoretically, we argued that the Vcp would be a reliable (and valid) parameter to quantify postural control. In contrast to the Acp, it integrates frequency and amplitude information into a single measure of regulatory activity which is strongly influenced by the higher frequency components of the CP fluctuations. It was predicted that particularly the power of these higher frequency components would be accurately regulated in order to minimize the relative displacement of body mass in time The results of this study corroborate the prediction that the Vcp is a relatively consistent parameter, however solely with respect to FA sway control They are also in conformity with previous studies indicating a good reproducibility of parameters related to the average absolute CP or CG velocity.[3,12 16] Our data point in the same direction as those of Goldie et al.[9] suggesting that the posture-control system primarily reduces the velocity and acceleration of body mass rather than its absolute displacement.

In contrast, the LAT sway results suggest that the Fcp is the most consistent parameter. This directional difference is probably due to the fact that the application of stabilizing ankle torques is most important for maintaining equilibrium in the sagittal plane Indeed, in this study subjects stood with ample space between their feet which provided a considerable degree of intrinsic mechanical stability in the frontal plane Such intrinsic stability reduces the need for active ankle torque application. As a result, the body may tend toward a preferred frequency of weight shifting between the feet.

In an earlier study, Hufschmidt and colleagues[11] also used variation coefficients to determine the intrasubject variability of the mean amplitude, mean frequency and sway path of the CP fluctuations They were, however, unable to identify one parameter as more consistent than others. This discrepancy from the present results may be explained by the fact that we separately tested the variability of the LAT and FA components of the selected parameters. In accordance with earlier work,[11,35] we found a fairly high coefficient of variation ($>$ 20%) for all selected parameters irrespective of the task condition, which indicates that none of the parameters is actually invariant over repeated tests. With regard to clinical balance assessments, it is therefore recommendable to increase the reliability of force-platform recordings by averaging the outcome of several consecutive tests.

It must be mentioned that there may have been an influence of a relatively short registration time ($<$ 1 min) on the variability of the Acp and PPcp. Particularly these parameters are determined by the low-frequency components, which might have demonstrated greater regularity if balance had been registered for several minutes. Unfortunately, such prolonged measurements would cause serious inconvenience or could even be impossible in patients with severe balance problems, especially under less favourable conditions (e.g. with the eyes closed).[9]

None of the selected parameters in this study revealed a significant trend across repeated assessments at two-weekly intervals. This result is in conformity with the work by Ekdahl et al.,[12] but contrasts with reports of a sway-reducing effect of repeated testing on the CP sway path[35] as well as on the absolute velocity of the pelvic movements[36] in healthy subjects. However, the latter studies dealt with learning effects over a series of tests performed on the same or on consecutive days, whereas in this study and in the study by Ekdahl et al.[12] the interval between balance assessments was a two-week and one-week period respectively. It appears that when repeated clinical assessments are directed at quiet two-legged standing and as long as they are spread over weeks instead of days, no correction is necessary for learning effects that are unrelated to rehabilitation.

The Vcp in the FA direction of sway proved to be more sensitive than the corresponding Acp in discriminating between the EO and EC conditions, whereas both parameters remained unaffected by simultaneous mental activity. The latter result reflects the automaticity of balance control in healthy adult subjects. The former result is in conformity with earlier studies,[11,20] which reported a stronger effect of eye closure on the CP sway path than on the absolute CP displacement from its mean position. Apparently, loss of visual information causes, besides an increase in Acp, a shift of Fcp toward higher values. Together, these factors lead to a marked increment in Vcp reflecting the role of visual information in postural control.[25-27] The stabilizing effect of the concurrent arithmetic task on the LAT CP amplitude cannot be readily explained. It shows that simultaneous attention demands may stabilize rather than destabilize posture in healthy subjects.

## CONCLUSION

This study demonstrates that the FA and LAT components of the RMS amplitude, peak-to-peak amplitude, mean frequency and RMS velocity of the CP fluctuations during quiet two-legged standing all show considerable variability across repeated tests irrespective of the task context. Nonetheless, both the RMS amplitude and RMS velocity in either direction of sway are (on average) sufficiently stable in time. Among the selected parameters, the RMS CP velocity in the FA direction shows the greatest intrasubject consistency, as well as a high sensitivity to e.g. visual deprivation. These findings support the reliability and validity of this parameter in the clinical quantification of postural control.

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# CHAPTER 4

## DUAL-TASK ASSESSMENT OF REORGANIZATION OF POSTURAL CONTROL IN PERSONS WITH LOWER LIMB AMPUTATION

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**ABSTRACT**

Postural control in persons with lower limb amputation was studied using a cognitive approach to motor learning. The aim of this study was to show that an important characteristic of the central reorganization process after a lower limb amputation is the gradually decreasing need of attentional resources to perform a motor task. A dual-task procedure was developed to estimate the level of automaticity of a quiet, upright standing task. The effect of a concurrent attention-demanding task (Stroop task) on the efficiency of balance control was determined using force-platform measurements at the start and the end of the rehabilitation process. In contrast with a control group, the amputation group showed interference effects on body sway caused by the concurrent task both at the start ($p < .05$) and, less severe, at the end of rehabilitation ($p < .05$). Improvement of balance control was significant only for the dual-task condition ($p < .05$). The results corroborated the hypothesis that dual tasks give information about the restoration of automaticity of postural control as an essential characteristic of the central reorganization process in persons with lower limb amputation. The role of dual-task procedures as a useful approach to skill assessment is discussed.

## INTRODUCTION

It has been recognized that persons with lower limb amputation have to regain adequate postural equilibrium reactions by linking altered sensory input to new forms of coordinated muscular activity.[1] If the level of amputation is above or through the ankle joint, the normal functions of the calf and other leg muscles are eliminated, causing loss of the "ankle strategy," an important mechanism for controlling the fore-aft body sway. Consequently, the body compensates for this loss through the unaffected limb and by shifting toward a less efficient "hip strategy."[2] Moreover, persons with through-knee or above-knee amputation must learn to stabilize an unlocked artificial knee joint by using the stump muscles in accordance with their changed functional anatomy. In all cases of lower limb amputation, the body must adapt to a severe asymmetry of mass and muscle power; it must also reform its internal representation or body scheme and learn to bear weight on the artificial limb to reduce asymmetric loading and lateral instability.

Although an amputation is primarily a peripheral disorder, a central reorganization within the sensorimotor system must take place as a reaction to the altered peripheral constraints. Until now, rehabilitation medicine has done little to evaluate the characteristics and underlying mechanisms of central reorganization processes.[3] As a result, knowledge about clinical procedures to monitor or stimulate such processes is limited. The theme of this paper is focused on the reorganization of postural control after a lower limb amputation. It is an attempt to integrate rehabilitation practice with aspects of cognitive neuroscience. Although recently the cognitive motor approach has been criticized by action theorists,[4,5] it offers a fruitful perspective for the elucidation of motor learning, particularly in the field of clinical rehabilitation.[6] Conversely, until now the action approach falls short in accounting for motor learning processes.[7]

From a cognitive viewpoint, the restoration of motor control is regarded as a learning process requiring substantial central information processing.[8,9] Because the processing demand is directly related to the novelty and difficulty of the motor task, the lack of task automaticity is indicated by the induced attentional load. In most cognitive theories the available processing resources are assumed to be limited. As a result, resource competition may occur during the performance of more than one attention-demanding task, leading to task interference.[10,11] The relearning of balance control after a lower limb amputation may well depend on central information processing. Accordingly, the central reorganization should be accompanied by a decrease in interference of a concurrent cognitive task with a primary balance task.

In this study, the effect of a concurrent attention-demanding task on the performance of a quiet, upright standing task was investigated in a group of persons with lower limb amputation at the start and at the end of the rehabilitation process. Assessments were made by means of a force platform, which permitted unrestrained balance recording. A modified version of the Stroop test[12,13] was selected as the concurrent task for several reasons.

The Stroop test demands a considerable amount of focused attention, it needs few instructions and shows relatively small long-term learning effects. Furthermore, it does not address memory, which is often impaired in elderly people. It requires only verbal responses, so that interference due to structural similarity or outcome conflict among tasks can be excluded.

It was hypothesized that the simultaneous performance of the Stroop test and the balance task would provide information that was essentially different from that obtained from simple, upright standing. It was believed that this information could be interpreted in terms of automaticity of the acquired balance behavior as an essential characteristic of the central reorganization process after a lower limb amputation. It was, therefore, predicted that at the start of a rehabilitation process, a clear task interference would occur in comparison with control subjects; at the end of a successful rehabilitation process, interference effects would be reduced or completely absent.

## METHODS

### Subjects

A group of ten persons with unilateral lower limb amputation, who received their first prostheses after a recent amputation and who did not show signs of serious cognitive dysfunctions, participated in the study. The patients ranged in age from 25 to 84 years (mean = 67.7 ± 18.1 years). The seven men and three women had either a below-, above-, or through-knee amputation, due to a vascular or nonvascular cause, which led to an acute disorder of balance and gait. A group of healthy subjects, matched for age (mean age = 65.6 ± 16.5 years) and gender, were also tested. Informed consent was obtained from all participating subjects.

### Equipment

Balance measurements were made with a firmly secured force platform consisting of two separate aluminum plates, each placed on three force transducers[a] (hysteresis and nonlinearity < 1%) recording the vertical ground reaction forces.

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[a]Load cells, type LM-100KA, Kyowa Electronic Instruments CO., LTD, Chofu-Higashiguchi Building 2F, 45-6, Fuda 1-chome, Chofu, Tokyo 182, Japan. [b]RMP DC-amplifier, type MBP 6218, Elan Schaltelemente GmbH, Holzheimer Weg 50, D-4040 Neuss 1, Germany.

Signals were processed by six DC-amplifiers[b] (nonlinearity <0.1%) and first-order low-pass filters (cut-off frequency 100Hz); then they were stored into a microprocessor (IBM XT-compatible) after a 14-bit AD-conversion at a sampling rate of 20Hz.

By means of digital moment-of-force calculations, the virtual center of the ground reaction forces in a two-dimensional transverse plane was determined for each sample, with a maximum error of ±1mm in both directions (center of pressure [CP]). The coordinates of the CP were passed through a digital low-pass 5-Hz filter, and the smoothed fluctuations of the CP were further processed by a first-order differentiation of the displacements.

Thus, possible high-frequency components (>5Hz), which are often associated with physiologic or pathologic (particularly in elderly people) tremor, did not influence the registrations. The force platform was placed 1.5m in front of a white projection screen (120cm x 110cm). A remote-controlled slide projector was used to project samples of the Stroop test onto the screen.

**Procedure**

Balance was registered for 30 seconds. Subjects stood erect on the force platform with the medial sides of their heels 8.4cm apart and with each foot toeing-out at a 9° angle from the sagittal midline. They were instructed to stand as still and symmetric as possible, with their hands folded on the back, without further instructions concerning visual attention.

During the first 15 seconds, balance was measured as a single task. After 15 seconds, a sample of the Stroop test was added to the primary task. Subjects were instructed to start this concurrent task as soon as possible, with emphasis on maintaining the same balance strategy.

The first five seconds of both halves of the registration were discarded from the analysis because undesired visual or starting effects might have been an influence. Consequently, the last 10 seconds of the registration represented balancing in a dual-task condition; whereas, the 10 seconds before the introduction of the secondary task represented the balance behavior in a simple single-task condition.

The modified Stroop test was projected onto the screen at eye level with a size of approximately 70cm (width) x 35cm (height). It consisted of a presentation of 25 colored words (five lines of five words, word size approximately 8cm x 3cm), representing color names that were always incongruent with the printed colors. For instance, the word "BLUE" was printed in the color red. In every sample, the same four colors for words and inks were used--yellow, red, blue, and green. The subjects were instructed to name the colors of the inks, as quickly as possible, from left to right and from the top downward until the end of the test procedure (30th second). At the same time, they had to suppress the strong tendency to read the words.

With three individuals in the amputation group, who could not perceive all colors of the Stroop test well enough, an arithmetic task was used as the concurrent task. They were instructed to subtract, as quickly as possible, the number 3 from an arbitrary starting number between 50 and 100, which was given to them verbally after the 15th second. They had to continue the subtractions until the end of the test procedure.

Before every balance test procedure, the cognitive task was first practiced three times in a sitting position. After practicing, it was recorded in this position to obtain a reference measure of the single-task performance. Speed was registered on the Stroop task (number of colors named in 15 seconds) as well as on the arithmetic task (number of subtractions made in 15 seconds). Also, the number of mistakes was noted.

Each balance assessment was based on two complete test procedures. With every person in the amputation group, balance skills were assessed one or two days after the first training with the definitive type of prosthesis (start of rehabilitation) and just before the completion of the rehabilitation therapy (end of rehabilitation). To obtain a more global measure of the rehabilitation outcome, the functional skills of every patient were assessed at the end of the rehabilitation process by means of an activity list for persons with amputation, as described and validated by Day.[14] The items concerned wearing and handling a prosthesis, indoor and outdoor ambulation and wheelchair use, walking aids, stair climbing, employment, and household activities. Because the automaticity of standing balance was considered to be optimal in healthy individuals, the matched controls were measured only once to obtain reference values.

**Data Analysis**

Balance performance was expressed as the root mean square (RMS) value of the CP velocities (Vcp) in the fore-aft (FA) and lateral (LAT) direction, so that frequency shifts and changes of amplitude were integrated into a single measure of (in)efficiency. Repeated measurements of a standing task with healthy adult subjects in different conditions (eyes opened and closed, single and dual task) proved that the Vcp derived from a 10- or 20-second registration showed a good reproducibility, indicating that subjects attempted to keep the balance activity at a constant low level in all conditions. In this experiment the Vcp was always derived from a 10-second registration. For each assessment, the comparable results on the two test procedures were averaged into single scores to compensate for intrasubject variability. Thus, three groups of data were obtained--the results of the amputation group at the start of rehabilitation, the results of the amputation group at the end of rehabilitation, and the results of the control group.

**Statistical Analysis**

Differences among the three groups of data were analyzed by means of a distribution-free Wilcoxon matched-pairs, signed-ranks test, taking a two-tailed probability of 5 percent as the level of significance.

Table 1: Characteristics of Amputation Group ($n = 8$).

| Subject | Sex | Age (yrs) | Type of amputation | Knee | Cause | Activity score[a] |
|---|---|---|---|---|---|---|
| 1 | M | 81 | through-knee | locked | Vascular | High |
| 2 | M | 25 | above-knee | unlocked | Trauma | High |
| 3 | F | 48 | above-knee | unlocked | Infection | Average |
| 4 | M | 76 | through-knee | locked | Infection | Average |
| 5 | F | 80 | below-knee | -- | Vasc/DM[b] | Restricted |
| 6 | M | 74 | below-knee | -- | Vasc/DM[b] | Restricted |
| 7 | F | 84 | below-knee | -- | Vasc/DM[b] | Average |
| 8 | M | 72 | below-knee | -- | Vasc/DM[b] | Average |

[a]Assessed by means of the amputee activity list[14] at the end of the rehabilitation process. [b]Vasc = vascular; DM = diabetes mellitus.

## RESULTS

From the start of their rehabilitation, all subjects with amputation made considerable clinical progress and reached an acceptable level of ambulation at the end of rehabilitation. Nevertheless, two patients were excluded from further analysis because of confounding factors. One patient developed a slowly progressive vascular deficiency of the nonamputated foot during his rehabilitation period. A second patient revealed a consistent tendency to fall at unexpected instances during the course of his training, which was probably related to a mild hemiparesis of the amputated side. Hence, the group results of eight persons with amputation were analyzed and compared to their matched controls.

Assessments made with the amputation activity list showed that two patients reached "high," four patients "average," and two patients "restricted" activity levels (as defined by Day[14]) at the end of rehabilitation. As a consequence of the criteria applied in the activity assessment, all patients were "inactive" at the start of the rehabilitation process. Table 1 presents some relevant characteristics as well as the individual activity scores of the eight persons in the amputation group.

As for the results on the concurrent task, there was no significant difference in the single- or dual-task performance among the three groups of data, nor between the single- and dual-task performance within any group. Apparently, the cognitive

**Table 2: Results of Fore-Aft Sway[a] (*n*=8).**

| | Amputation group | | | | Control group | |
|---|---|---|---|---|---|---|
| | Start of rehabilitation | | End of rehabilitation | | | |
| Subject | Single | Dual | Single | Dual | Single | Dual |
| 1 | 8.3 | 13.9 | 12.7 | 10.0 | 8.2 | 9.5 |
| 2 | 6.7 | 13.1 | 9.4 | 12.8 | 4.6 | 5.3 |
| 3 | 8.3 | 14.0 | 6.4 | 7.2 | 4.6 | 4.7 |
| 4 | 18.5 | 23.0 | 15.8 | 20.4 | 5.3 | 5.7 |
| 5 | 14.8 | 17.7 | 10.1 | 11.7 | 9.8 | 7.7 |
| 6 | 33.5 | 46.2 | 33.4 | 36.8 | 10.6 | 9.8 |
| 7 | 34.7 | 40.9 | 28.4 | 34.4 | 10.7 | 14.6 |
| 8 | 25.9 | 36.7 | 22.6 | 34.7 | 5.7 | 4.7 |
| Mean | 18.8 | 25.7 | 17.4 | 21.0 | 7.4 | 7.8 |
| SD | 11.4 | 13.5 | 9.8 | 12.4 | 2.7 | 3.4 |

[a]Fore-aft sway in the single- and dual-task conditions (both 10 seconds) expressed as the RMS CP velocity (mm/sec) (mean value of two registrations).

**Table 3: Results of Lateral Sway[a] (*n*=8).**

| | Amputation group | | | | Control group | |
|---|---|---|---|---|---|---|
| | Start of rehabilitation | | End of rehabilitation | | | |
| Subject | Single | Dual | Single | Dual | Single | Dual |
| 1 | 4.7 | 7.6 | 6.2 | 6.5 | 5.6 | 5.8 |
| 2 | 5.4 | 10.3 | 5.5 | 8.4 | 3.6 | 2.5 |
| 3 | 5.1 | 12.7 | 3.5 | 3.9 | 3.1 | 2.8 |
| 4 | 13.3 | 18.1 | 5.7 | 9.3 | 2.0 | 2.3 |
| 5 | 9.2 | 12.5 | 6.4 | 7.9 | 4.9 | 3.3 |
| 6 | 17.7 | 22.8 | 18.9 | 24.0 | 6.0 | 6.3 |
| 7 | 18.1 | 19.2 | 13.7 | 15.1 | 7.6 | 6.4 |
| 8 | 12.0 | 20.7 | 6.7 | 13.2 | 1.9 | 1.9 |
| Mean | 10.7 | 15.5 | 8.3 | 11.0 | 4.3 | 3.9 |
| SD | 5.5 | 5.4 | 5.2 | 6.3 | 2.0 | 1.9 |

[a]Lateral sway in the single- and dual-task conditions (both 10 seconds) expressed as the RMS CP-velocity (mm/sec) (mean value of two registrations).

task was performed rather consistently, which made it possible to estimate interference effects on the basis of the balance results alone Table 2 presents the FA sway results in the single- and dual-task condition for all three groups of data. The comparable results on the LAT sway are reported in Table 3 In contrast with the control group, the balance behavior of the amputation group was less efficient in the dual-task condition than in the single-task condition, both at the start (FA and LAT sway, $p<.05$) and at the end (FA and LAT sway, $p<.05$) of rehabilitation.

Figures 1 and 2 show the mean and standard deviation of the group results, respectively, in the single-task and dual-task conditions In comparison with the controls, the persons with amputation showed at all moments less postural efficiency in the single-task condition as well as in the dual-task condition (FA and LAT sway, $p<$ 05). Comparing the balance control of the amputation group between the start and the end of rehabilitation, the single-task condition showed no significant differences, although there was a slight tendency toward improvement Conversely, the dual-task condition showed a better efficiency of balance at the end of rehabilitation than at the start on the FA sway ($p<$ 05) as well as on the LAT sway ($p<.05$).

The interference of the concurrent task with the balance task is reflected in the difference between the dual- and single-task performance To represent the interference effect, both an absolute value (differential score) and a measure relative to the single-task performance (quotient) were calculated The comparable differential scores, as well as the quotients derived from both test procedures, were averaged for each assessment The mean and standard deviation of the differential scores of each group are graphically presented in Figure 3

The mean interference effect on the FA sway averaged for eight persons with amputation decreased from 6.9mm/sec at the start to 3 7mm/sec at the end of rehabilitation. This reduction proved to be marginally significant ($p<$ 06) The mean interference effect on the LAT sway averaged for the eight subjects with amputation decreased from 4.8mm/sec at the start to 2.7mm/sec at the end of rehabilitation. This reduction was clearly significant ($p<$ 05). Figure 4 demonstrates that the same significance could not be derived from the quotients, although the mean quotient averaged for eight persons in the amputation group decreased from 1 47 to 1 22 on the FA sway and from 1.60 to 1 39 on the LAT sway across rehabilitation

It is important to note that at the start of rehabilitation, the quotients as well as the differential scores of the subjects with amputation always revealed greater interference effects than the reference values of the control group ($p<.05$), which was not true for the FA sway at the end of rehabilitation These findings suggest a clear tendency also for relative differences toward improvement, at least on the FA sway.

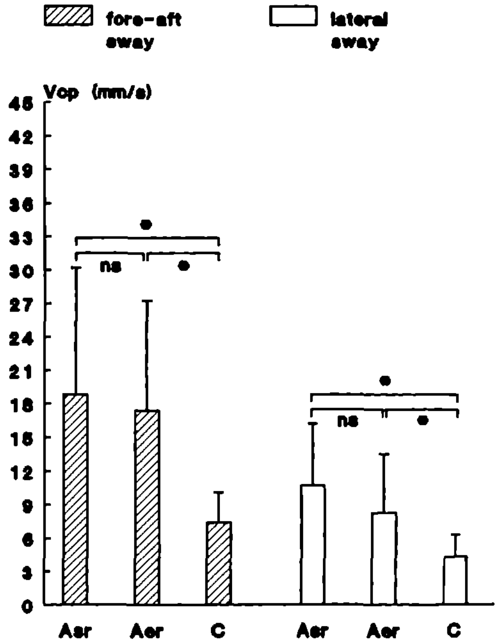


Figure 1. The group mean and SD of the RMS CP velocity (Vcp) in the single-task condition are shown for the amputation group at the start (Asr) and at the end (Aer) of rehabilitation and for the eight control (C) subjects; * = $p<.05$, ns = not significant (Wilcoxon test).

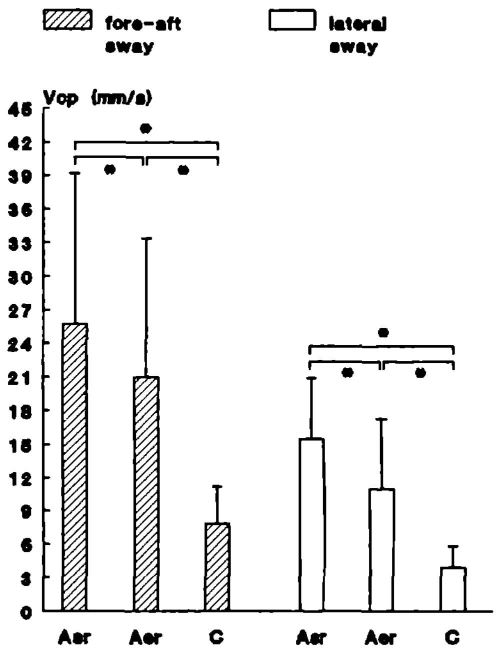


Figure 2. The group mean and SD of the RMS CP velocity (Vcp) in the dual-task condition are shown for the amputation group at the start (Asr) and at the end (Aer) of rehabilitation and for the eight control (C) subjects; * = $p<.05$, ns = not significant (Wilcoxon test).

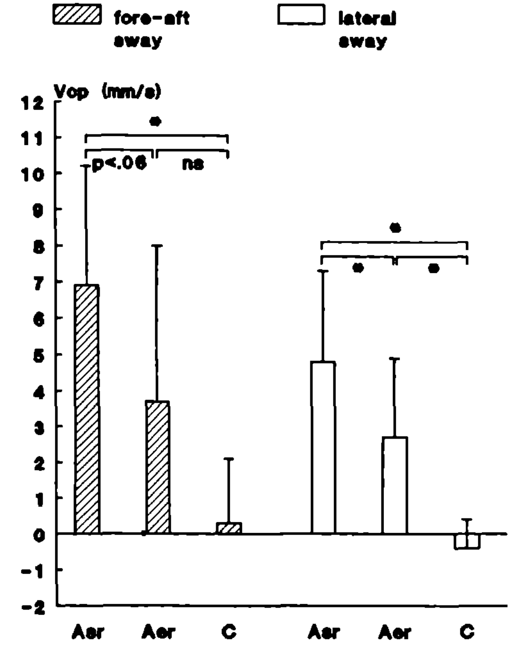


Figure 3. The group mean and SD of the RMS CP velocity (Vcp) of the dual- minus single-task condition (differential score) are shown for the amputation group at the start (Asr) and at the end (Aer) of rehabilitation and for the eight control (C) subjects; * = $p<.05$, ns = not significant (Wilcoxon test).

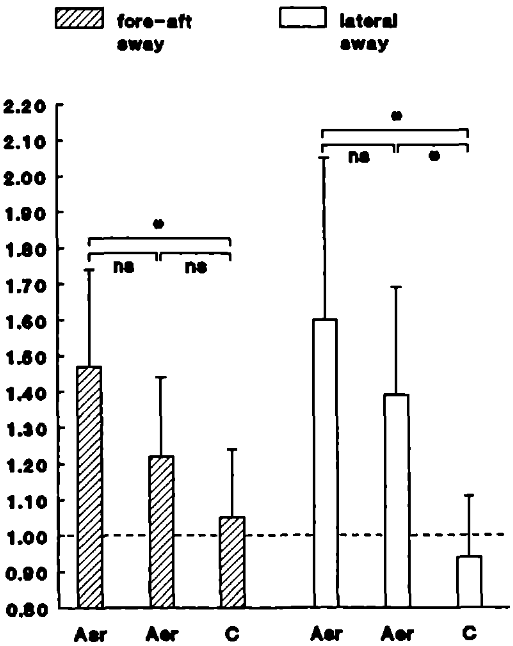


Figure 4. The group mean and SD of the RMS CP velocity of the dual- divided by single-task condition (quotient) are shown for the amputation group at the start (Asr) and at the end (Aer) of rehabilitation and for the eight control (C) subjects; * = $p<.05$, ns = not significant (Wilcoxon test).

## DISCUSSION

The aim of this study was to demonstrate that an important characteristic of the central reorganization process after a lower limb amputation is the gradually decreasing need of attentional resources to perform the same motor task. A dual-task procedure was developed to estimate the level of automaticity of a quiet, upright standing task. It was shown that a complex dual-task condition provides different information compared to a simple single-task condition. In comparison with control subjects, persons with lower limb amputation showed a lower level of postural efficiency and interference effects caused by a concurrent attention-demanding task This interference was clear at the start of rehabilitation, although less severe, it could still be measured at the end of rehabilitation. The interference effects were present in both directions of sway, which is in agreement with the fact that a lower limb amputation affects postural control in the sagittal plane as well as in the frontal plane

The degree of interference, expressed as the absolute difference between the balance performance in the dual- and single-task condition, decreased during the rehabilitation process In this experiment, only the dual-task performance clearly improved in time These findings suggest that it is not the motor output as a separate entity, but the motor behavior as the integrative outcome of perceptual, cognitive, and motor processes that gradually became more efficient

However, some precautionary remarks must be made The lack of improvement on the single-task performance may be the result of the relatively small number of patients Therefore, no generalizations can be made Furthermore, estimating interference effects using the absolute differences between conditions is somewhat hazardous It is questionable whether the same differential scores must be valued equally in someone with low initial velocities of sway as in someone with high velocities in the single-task condition Therefore, the balance performance in the dual-task condition was divided by the single-task performance to obtain quotients that might allow a more reliable comparison of the interference effects among subjects and among different moments for the same subject Although the quotients provided the same pattern as the differential scores, they did not show a significant change between the start and the end of rehabilitation It is plausible, however, that this lack of significance was also due to the limited number of patients Nevertheless, the relative differences on the FA sway are at least indicative of a meaningful improvement in time.

It is important to note that fatigue or other time-related factors can be excluded as being responsible for the observed differences of balance performance between the single- and dual-task conditions. The same amputation group did not show any tendency toward heightening velocities of sway in a control measurement without a concurrent attention-demanding task. Hence, the general conclusion is drawn that the diminution of interference in the dual-task condition is related to the central sensorimotor reorganization. Unfortunately, a precise explanation of the neural mechanisms underlying the observed task interference is more difficult to give. For example, it is not unequivocally clear what stages of information processing are responsible for the interference effects. Although this problem is not of primary concern to the clinical theme of this paper, some remarks should be made.

Both the Stroop test and the arithmetic task require totally different motor responses from those needed for standing balance. It is, therefore, tentative to think that the reported task interference is mainly caused by resource competition due to a limited undifferentiated processing capacity. However, performing the Stroop test also requires the processing of visual information. Because vision is involved in postural control,[15] and because persons with lower limb amputation suffer from an increased dependency on visual information,[16] it cannot be denied that interference of a more specific nature may have played an additional role. Although some authors[17,18] have argued that there are two visual subsystems that operate in truly independent, parallel fashion (central vision designed for object recognition, active in the Stroop test, versus peripheral vision designed for spatial orientation, active in postural control), there is evidence that the role of vision in balance control is not exclusively dominated by the peripheral part of the visual field.[19]

Therefore, part of the interference effects may be due to competitive factors within the uptake or processing of visual information needed to perform the Stroop test and the balance task simultaneously. Consequently, the diminution of interference may partly reflect a decrease of visual dependency during the rehabilitation process in the amputation group.[20] Indeed, a lessening dependency on vision for postural control can be seen as an integral part of the central reorganization process after a lower limb amputation.

The question about the exact neural basis is, however, of secondary importance in comparison with the actual occurrence of disproportional interference effects in the amputation group. The notion that the restoration of postural control in persons with lower limb amputation is partly based on cognitive processes that are not directly accessible for motor assessment procedures using simple-task conditions has clear clinical implications. It can be seen as an argument to implement a dual-task procedure into the functional assessment of persons with lower limb amputation and to specifically train these patients in more complex, attention-demanding environments.

## CONCLUSIONS

Clinical rehabilitation is aimed at the patient's participation in ADL. To perform safely under many daily circumstances, one needs at least some automaticity of standing and walking so that simultaneous demands from the environment can be met. A lack of automaticity may lead to a locomotor disability, as defined by the World Health Organization.[21] Thus, the estimation of automaticity of gross motor behavior by means of dual-task procedures can be useful to operationalize an essential characteristic of well-developed locomotor skills.[22] Insight to a motor skill will result from research in which movements are studied together with other variables related to perception and information processing. It is argued that the information obtained in this way is necessary to fully understand all aspects of central reorganization processes after peripheral sensorimotor damage, such as a lower limb amputation. This consideration has, of course, general implications for monitoring rehabilitation processes and for the assessment of therapy outcome, not only with regard to standing, but also concerning other forms of gross motor behavior.

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CHAPTER 5

# POSTURAL REORGANIZATION FOLLOWING LOWER LIMB AMPUTATION:

## Possible Motor and Sensory Determinants of Recovery

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**ABSTRACT**

Postural control was assessed in persons with a unilateral lower limb amputation before and after their rehabilitation. The centre-of-pressure fluctuations during quiet upright standing on a dual-plate force platform were registered with and without visual information in order to indentify relevant determinants of balance restoration. In addition, static (weight distribution) as well as dynamic (control activity) asymmetry characteristics were examined. Besides a small improvement in balance control with full visual information (fore-aft sway, $p<0.06$; lateral sway $p<0.05$), there was a major decrease in visual dependency (fore-aft and lateral sway, $p<0.05$) indicating a somatosensory re-integration process. Postural asymmetry in comparison with matched control subjects was most apparent and only significant in dynamic terms and remained constant across rehabilitation. It is concluded that after a lower limb amputation a central reorganization of postural control takes place, in which sensory determinants of motor recovery may play a critical role.

## INTRODUCTION

Of all the research which has been focused on stance in subjects with lower limb amputation, relatively few studies have dealt with the control characteristics of maintaining posture with a lower limb prosthesis (1,4,6,7,10,11,32). Even fewer studies have focused on the restoration of balance control after a lower limb amputation (7,10). Balance in persons with lower limb amputation has, however, been recognized as a relevant clinical problem, particularly during the early phases of rehabilitation (18,19).

Nevertheless, reduced balance control in subjects with lower limb amputation has not been unequivocally demonstrated in conditions with full perceptual information (4,6,7,11). On the other hand, increased visual control of posture seems to be a more characteristic consequence of both below- and above-knee amputation, caused by the reduced availability of somatosensory information (4,6,11).

Little is known about postural control during the early phases following lower limb amputation. Gauthier-Gagnon et al. (7) reported a diminution of visual dependency in five subjects with below-knee amputation who had been trained with auditory feedback from a limb load monitor. However, this was not found in a control group of six subjects who had received traditional training. There was no decrease in sway during quiet stance with the eyes open in either group.

Although a lower limb amputation is primarily a peripheral disorder, a central reorganization must take place to adapt to the peripheral sensory and motor impairments (10). By studying balance recovery, information can be obtained about the most critical determinants of sensorimotor reorganization after lower limb amputation to improve assessment procedures and rehabilitation programmes.

Therefore, quiet standing was examined in a heterogeneous group of subjects with unilateral lower limb amputation before and after a traditional training programme. The basic level of balance control efficiency was assessed, as well as the degree of visual dependency to examine the contribution of motor and sensory processes to the central reorganization following lower limb amputation. Some preliminary data on this issue were reported earlier (8).

In addition, it was evaluated whether balance restoration would coincide with a reduction in postural asymmetry. Such asymmetry is generally expressed in terms of weight bearing (static asymmetry) (16,29), although the relevance of weight distribution between the feet for monitoring balance skills in persons with lower limb amputation is still unclear (7,28,30). In this study, postural asymmetry was also assessed in dynamic terms (dynamic asymmetry) by recording the centre-of-pressure fluctuations under each foot. In this way, the compensatory control activity of the non-amputated limb was evaluated before and after the rehabilitation process.

## SUBJECTS AND METHODS

### Subjects

The amputation group ($n = 10$) was aged 25 to 84 years (mean age 67.7 ± 18.1) and consisted of 7 males and 3 females with either a below- ($n = 4$), a through- ($n = 3$) or an above-knee ($n = 3$) amputation. The four patients with a below-knee amputation had a vascular (diabetic) cause of amputation. Of the six patients with a through- or above-knee amputation, three subjects had a non-vascular and three subjects had a vascular (non-diabetic) cause of amputation.

During a period of eight months, persons with a recent unilateral limb amputation above the ankle and below the hip joint participated in the study. Subjects suffering from serious cognitive (e.g. disorders of memory or attention) or perceptual (e.g. cataract or visual field loss) deficits were excluded, as well as those suffering from pain problems. In addition, a group of ten healthy subjects were tested, matched for age and gender (mean age 65.6 ± 16.5).

### Equipment

Balance measurements were made with a firmly secured force platform which consisted of two separate aluminium plates, each placed on three force transducers (hysteresis and non-linearity < 1%) recording the vertical ground reaction forces. The force platform was connected to a microprocessor, which determined the virtual centre of the ground reaction forces at a sampling rate of 20 Hz and with a maximum error of ± 1 mm in both directions of sway. The coordinates of this centre-of-pressure (CP) were led through a digital low-pass 5-Hz filter to eliminate erroneous readings due to noise.

### Procedure

During the balance recordings, the subjects stood erect on the force platform with their feet against a foot frame (medial sides of the heels 8.4 cm apart; each foot toeing-out at a 9° angle from the sagittal midline). The subjects were repeatedly instructed to stand as still and symmetrically as possible with their hands folded behind their back.

Each test procedure consisted of three conditions-- standing with the eyes open, with blurred vision (wearing milky-white spectacles preventing visual anchoring) and with the eyes closed (reinforced by closed dark spectacles). With their eyes open, the subjects faced a white wall at a distance of 1.5 meters. The blurred-vision condition was employed, in addition to the eyes-closed condition, because there is evidence that the effect of discongruent visual input is different from the effect of visual suppression (17,21,25).

In every condition, balance was recorded for 20 sec with at least one minute's rest between conditions. A fixed sequence of tests--eyes open, blurred vision, eyes closed--was employed, so that order effects were kept constant. A complete sample of 20-sec registration represented postural control in each condition.

Each balance assessment consisted of two consecutive test procedures. In all the patients, balance was assessed one or two days after the first training with the definitive type of prosthesis (start of rehabilitation) and repeated just before the completion of the training programme (end of rehabilitation). The matched control subjects were tested once, with bare feet, to obtain reference values.

Before the start of rehabilitation, every patient was given exercises with an airboot for a variable period from three to six weeks. The period between the start and the end of rehabilitation varied considerably between subjects from three to fourteen weeks (mean 10.6 weeks). During this period the amputation group was trained for two hours daily.

The training programme followed a gradual transition from erect standing, weight-shifting and stepping between parallel bars, followed by walking on regular surfaces with two crutches or sticks, to walking on irregular surfaces without aids. In the latter stages, balance and ambulation were also trained while the patient was performing a secondary visuo-motor task such as throwing and catching a ball. Therapists provided verbal instructions and manual corrections, but no artificial sensory feedback was employed.

All the decisions on the rehabilitation of individual subjects were made independently of the research team. In order to determine whether substantial functional progress had been made, the activity level of every patient was assessed at the end of rehabilitation by means of an amputee activity list (2). The items concerned wearing and handling a prosthesis, indoor and outdoor ambulation and wheelchair use, walking aids, stair climbing, employment and household activities.

**Data Analysis**

Three types of parameters were derived from the CP fluctuations under both feet together (overall CP) in the fore-aft (FA) and lateral (LAT) directions: the mean CP position (Pcp), the root mean square (RMS) amplitude of the CP displacements (Acp) and, after a first-order differentiation, the RMS of the CP velocities (Vcp).

Because the Vcp integrates amplitude and frequency into a single measure of regulatory activity, this measure was primarily used to detect differences in control efficiency between the conditions. It has been demonstrated that parameters related to the average CP velocity show acceptable reliability as well as discriminating power between different physiological and pathological conditions (3,5,13,15,23,31). The

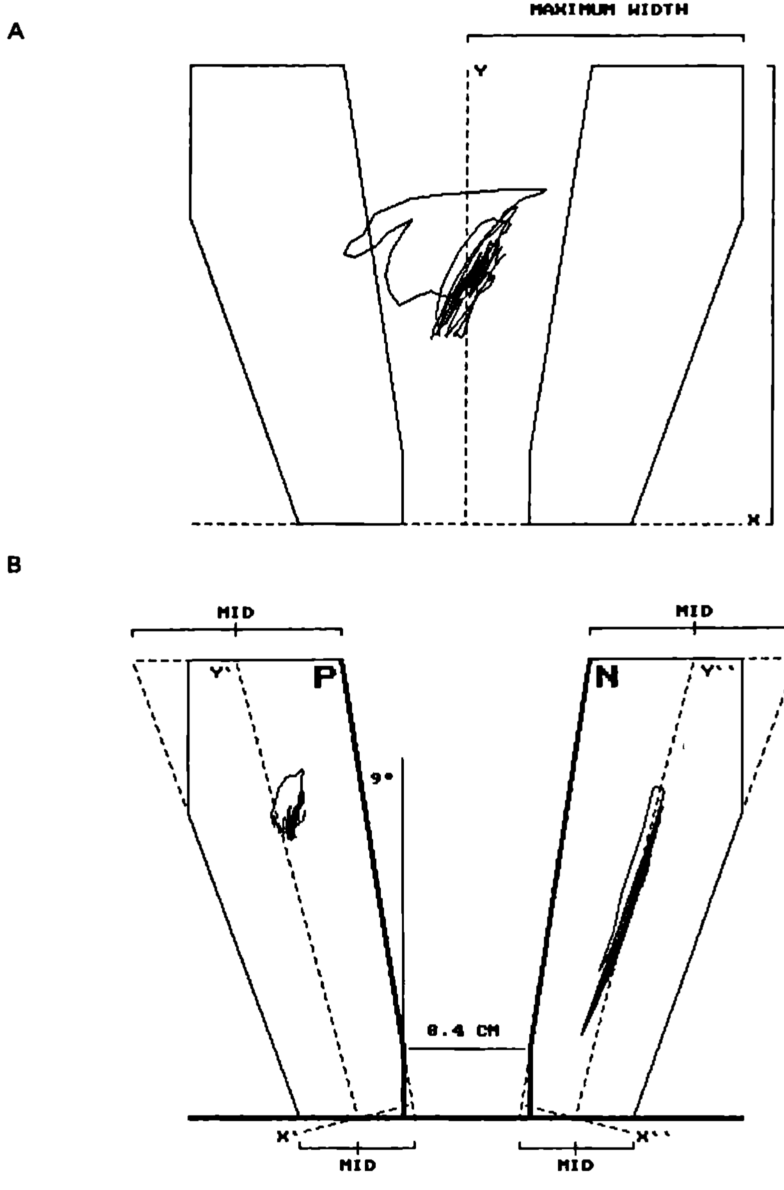


**Figure 1.** (A) The trajectory of the overall CP is shown for one of the amputees standing with the eyes closed at the start of rehabilitation; the uninterrupted lines indicate the position of the feet; the dashed lines X and Y refer to the LAT and FA directions of the body frame of reference, respectively. (B) The trajectory of the CP under each individual foot is shown for the same registration; the dashed lines X' and Y' / X'' and Y'' refer to the LAT and FA directions of the frames of reference for the prosthetic (P) and normal (N) foot, respectively; the bold lines show the position of the foot frame.

Acp was regarded as an additional measure of postural sway. Furthermore, the ratio between the Vcp and Acp served to estimate the mean frequency (Fcp) using the following approximation: $Fcp = Vcp / (Acp \times 4 \times \sqrt{2})$ (13).

The Pcp was expressed as a percentage of the maximum length (FA direction) and the maximum width (LAT direction) of support with the zero point at the rear in the sagittal midline (see Figure 1A). The base-of-support measures were determined by drawing the circumference of the feet on a piece of paper while the subjects stood on the force platform.

The Vcp was also calculated in two directions under each individual foot. First, the frame of reference was digitally translated and rotated towards the anatomical position of each foot to be able to reliably compare the FA and LAT CP fluctuations under the prosthesis to those under the non-amputated limb. The method which was applied to estimate the longitudinal axis through each foot (FA) from the base of support, and the line perpendicular to this axis (LAT), is visualized in Figure 1B.

For each assessment, the comparable results on the two test procedures were averaged into a single score. Thus, three groups of data were obtained--the results of the amputation group at the start of rehabilitation, the results of the amputation group at the end of rehabilitation and the results of the control group.

**Statistical Analysis**

Differences among groups or between conditions within each group were analysed by means of a distribution-free Wilcoxon matched-pairs signed-ranks test.

## RESULTS

All the subjects with an amputation had achieved an acceptable level of independent ambulation at the end of rehabilitation. Nonetheless, two patients were excluded from further analysis because of confounding factors. One patient with a (vascular) through-knee amputation developed progressive vascular deficiency of the non-amputated foot. A second patient with a (vascular) above-knee amputation kept falling at unexpected moments during his rehabilitation, which was probably related to mild hemiparesis of the amputated side.

Consequently, the data from eight patients were analysed statistically and compared to the matched control data. The activity assessment showed that four patients had reached "average," two patients "high" and two patients "restricted" activity levels at the end of rehabilitation. As a consequence of the criteria applied in the activity assessment, all the patients were "inactive" at the start of rehabilitation.

Occasionally, a balance test in the absence of visual information had to be interrupted, because the patient came close to falling. In such cases, a second trial was performed. Nevertheless, one patient was unable to stand with blurred vision. Because the blurred-vision condition did not show significant group differences compared to the eyes-closed condition, only the latter condition will be further discussed.

Table 1 presents the Vcp values of the overall CP for the three groups of data. In comparison with the controls, the persons with an amputation showed less postural control efficiency at all moments in both conditions (FA and LAT sway, $p<0.05$). There was a marked improvement in balance control within the amputation group between the start and the end of rehabilitation assessed by the eyes-closed condition (FA and LAT sway, $p<0.05$). In contrast, the eyes-open condition revealed only a minor improvement in time, which was merely marginally significant on the FA sway (FA sway, $p<0.06$; LAT sway, $p<0.05$). There was a moderate (non-significant) negative correlation between the activity level and the CP velocity at the end of rehabilitation.

**Table 1: Means and Standard Deviations of the RMS Velocity and Amplitude and the Mean Position of the Centre of Pressure under Both Feet (overall CP) in the Amputation Group and Control Group ($n=8$).**

| | Amputation group | | | | Control group | |
|---|---|---|---|---|---|---|
| | Start rehabilitation | | End rehabilitation | | | |
| | Eyes open | Eyes closed | Eyes open | Eyes closed | Eyes open | Eyes closed |
| *Fore-aft sway* | | | | | | |
| Vcp[a] | 21.4±11.4 | 42.5±26.8 | 18.9±11.3 | 26.7±17.9 | 7.8±3.0 | 9.3±3.3 |
| Acp[b] | 5.1±1.7 | 9.8±4.2 | 4.6±1.6 | 5.8±2.7 | 3.8±1.5 | 4.7±2.1 |
| Pcp[c] | 41.9±6.4 | 42.3±5.9 | 41.1±5.6 | 42.5±5.2 | 41.8±6.3 | 45.2±6.6 |
| *Lateral sway* | | | | | | |
| Vcp[a] | 12.8±5.6 | 25.9±18.7 | 9.8±5.1 | 12.5±9.9 | 4.2±1.3 | 5.2±2.5 |
| Acp[b] | 4.8±2.6 | 6.5±3.4 | 2.9±0.8 | 3.8±1.5 | 2.1±1.0 | 2.8±1.5 |
| Pcp[c] | 9.5±10.4 | 10.1±8.8 | 5.4±11.3 | 7.1±11.1 | -6.0±5.7 | -6.7±6.3 |

[a]RMS (amplitude) of the CP displacements (mm). [b]RMS of the CP velocities (mm/sec). [c]Mean CP position relative to the base of support (%); in the amputation group positive values in the lateral direction correspond with a deviation towards the normal foot, in the control group towards the right foot.

The CP velocity indicated that in every group the FA sway control was less efficient in the eyes-closed than in the eyes-open condition ($p<0.05$). With respect to the LAT sway control, such reduced efficiency in the absence of vision was only significant in the amputation group at the start of rehabilitation ($p<0.05$).

To determine the degree of visual dependency in both planes of balance control, the absolute difference in Vcp between the eyes-closed and eyes-open performance (differential score) was calculated for each test procedure, as well as the relative difference by dividing the eyes-closed by the eyes-open performance (quotient) (23).

The group means and standard deviations of the differential scores are shown in Figure 2. During the course of rehabilitation, the mean visual dependency for eight persons with an amputation decreased from 21.1 to 7.8 mm/sec on the FA sway ($p<0.05$) and from 13.0 to 2.7 mm/sec on the LAT sway ($p<0.05$). Figure 3 demonstrates that the same significance was derived from the visual dependency quotients, which decreased from 2.01 to 1.45 on the FA sway ($p<0.05$) and from 2.00 to 1.22 on the LAT sway ($p<0.05$).

In the amputation group at the start of rehabilitation, all the visual dependency scores were greater than in the control group ($p<0.05$). At the end of rehabilitation, only the differential visual dependency scores for FA sway differed significantly from the control values ($p<0.05$).

Relating the Acp to the Vcp values for the three groups of data (Table 1) revealed that the reduction in postural control efficiency in the absence of vision was largely caused by an increase in Acp in all the groups. Indeed, the mean Fcp in the amputation group remained fairly constant between conditions, varying from 0.6 to 0.7 Hz on the FA sway and from 0.5 to 0.6 Hz on the LAT sway. The control group showed mean Fcp values of between 0.3 and 0.4 Hz in both directions of sway.

Also, the improvement of postural control within the amputation group across rehabilitation was mainly based on a decrease in Acp. Both the FA sway in the eyes-closed condition and the LAT sway in the eyes-open and eyes-closed conditions showed a significant decrease in Acp ($p<0.05$). The decrease in the CP amplitude on the FA sway in the eyes-open condition was not significant.

The Pcp values of the overall CP are also listed in Table 1. At the start of rehabilitation, the mean Pcp in the amputation group deviated approximately 10% of the support width towards the normal foot. This deviation corresponded with a mean value of 43% weight bearing onto the prosthesis. At the end of rehabilitation, the mean Pcp was located approximately 6% of the support width towards the normal foot (mean value 45% weight bearing onto the prosthesis). No significant difference in the LAT CP position was found between the start and the end of rehabilitation.

The control group also demonstrated static asymmetry by bearing less weight on the right limb (mean 45%). Static asymmetry in the amputation group was never significantly greater than in the control group. In all groups, the Pcp was within 41%

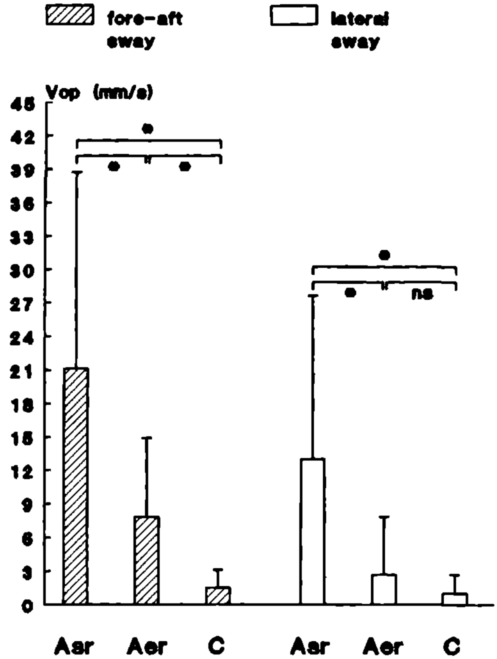


Figure 2. The group means and SDs of the RMS CP velocity (Vcp) in the eyes-closed minus the eyes open performance (differential score) are shown for the amputation group at the start (Asr) and at the end (Aer) of rehabilitation and for the control group (C) ($n=8$); $^*p<0.05$, ns = not significant (Wilcoxon test).

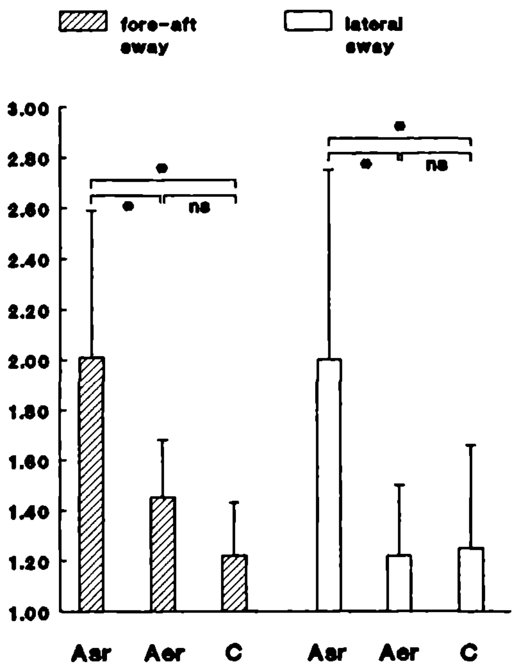


Figure 3. The group means and SDs of the RMS CP velocity in the eyes-closed divided by the eyes-open performance (quotient) are shown for the amputation group at the start (Asr) and at the end (Aer) of rehabilitation and for the control group (C) ($n=8$); $^*p<0.05$, ns = not significant (Wilcoxon test).

and 46% of the foot length. No significant differences in the FA CP position were found between groups or conditions.

Table 2 presents the Vcp values under each individual foot for the amputation group before and after the rehabilitation. The control activity in the FA direction was always significantly greater under the normal than under the prosthetic foot ($p<0.05$). Similar dynamic asymmetry was found with respect to the LAT CP fluctuations under each foot, but was only significant in the eyes-closed condition ($p<0.05$). In contrast, dynamic asymmetry was absent in the control group.

**Table 2: Means and Standard Deviations of the RMS Velocity of the Centre of Pressure under Each Foot in the Amputation Group ($n=8$).**

| | Start rehabilitation | | End rehabilitation | |
|---|---|---|---|---|
| | Eyes open | Eyes closed | Eyes open | Eyes closed |
| *Fore-aft sway* | | | | |
| Vcp-N[a] | 33.6±22.4 | 68.1±53.7 | 30.0±21.7 | 43.4±34.4 |
| Vcp-P[b] | 8.7±2.5 | 14.2±3.9 | 8.6±2.7 | 10.5±3.5 |
| *Lateral sway* | | | | |
| Vcp-N[a] | 5.6±2.7 | 11.7±4.5 | 3.8±1.7 | 6.7±4.5 |
| Vcp-P[b] | 3.2±1.2 | 5.6±2.5 | 2.8±0.8 | 3.3±1.3 |

[a]RMS of the CP velocities under the normal foot (mm/sec).
[b]RMS of the CP velocities under the prosthesis (mm/sec).

To examine whether the degree of dynamic asymmetry showed any change across rehabilitation, symmetry quotients were calculated. For each test procedure, the Vcp under the normal foot was divided by the Vcp under the prosthetic foot (Vcp-N / Vcp-P) in both the FA and LAT directions. However, this analysis did not reveal any significant decrease in the dynamic asymmetry in either direction.

## DISCUSSION

The purpose of this study was to identify relevant motor and sensory determinants of balance restoration following lower limb amputation. Due to the limited number of subjects, the results may not be generalized to all persons with a lower limb amputation. It is, on the other hand, legitimate to lay emphasis on the mutual relations between various characteristics within the same amputation group.

In contrast with some of the earlier studies (4,6,7,11,32), the amputation group showed less postural control in both directions of sway at all moments. Apart from inter-subject differences, this result can be explained by different instrumentation and parameter choice. Tables 1 and 2 indicated that the higher overall-CP velocities in the amputation group were largely caused by higher CP velocities under the normal foot, which means that generally, not the centre of gravity of the body, but the centre of the ground reaction forces as a physical control variable (20,24) was moving at a relatively high speed.

Furthermore, the high CP velocities in the amputation group were substantially determined by relatively high frequencies. This finding also supported the notion that during stance with the eyes open, there was not so much destabilization as loss of control efficiency in the amputation group. Hence, studies in which the velocity of the centre of gravity was registered (4,6,11) or studies which used CP parameters which are less sensitive to frequency (7,32) may have led to different results.

The activity assessment was primarily meant to determine whether considerable functional progress had been made to presume a general improvement in balance performance. As the items incorporated a broad area of daily activities, some of them completely unrelated to balance (such as putting on and taking off the prosthesis), the lack of a significant correlation between the activity score and body sway was not surprising in such a limited number of subjects.

Visual control of posture was found not only in the amputation group, but also in the control group, at least on the FA sway. Indeed, the influential role of vision in balance control is well-known (14,26,27). In accordance with earlier reports (4,6,11), an increased dependency on visual information was found to be a marked feature in persons with a lower limb amputation in both directions of sway. This phenomenon can be attributed to a unilateral loss of somatesthesis, in particular from the ankle joint and foot sole. In all the groups, the reduction in postural control efficiency in the absence of visual information coincided with a considerable increase in the CP amplitude, which may reflect a compensatory mechanism to increase the excitation of other input sources (e.g. the vestibulum).

Between the start and the end of rehabilitation, there was much more improvement in balance control assessed with the eyes closed than with the eyes open. The degree of visual dependency in the amputation group, expressed either as an absolute or as a relative measure, showed a major decrease in time in both directions of sway, approaching normal values at the end of rehabilitation. This decrease in visual dependency clearly indicated a central integration of sensory input from the amputated limb into the multi-sensory control of posture.

This study provided little indication of a symmetrization process following lower limb amputation, either in static or in dynamic terms. Apparently, the amputation group had learned to bear substantial weight on the airboot during the previous phase of rehabilitation. Weight-bearing asymmetry in the amputation group was not significantly greater than in the control group, which was partly due to relatively large inter-subject variability.

In contrast, the amputation group showed significant dynamic asymmetry both at the start and at the end of rehabilitation. It has been shown that a physiological ankle joint and intact lower leg muscles are essential output structures for the utilization of ground reaction forces to maintain equilibrium (12,22). This study indicated that unilateral damage to these structures causes permanent compensatory control activity of the contralateral limb.

## CONCLUSION

Processes which enhance the availability of sensory information from the amputated limb may substantially contribute to balance recovery in persons with a lower limb amputation. This conclusion provides an argument to implement various sensory conditions in the physical training of subjects with lower limb amputation. The ability to process somatosensory information seems to be particularly important in situations of sensory conflict and reduced visual information (21,23,25) and may, therefore, be related to the safety of balance performance. This conclusion also has implications for a valid assessment of motor recovery. Unless sensory manipulations are used, afferent aspects of central reorganization processes remain unrecognized in clinical motor assessment. For future research, it seems important to also appreciate such "hidden" determinants of motor recovery (9).

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# CHAPTER 6

## POSTURAL ORGANIZATION IN PATIENTS WITH

## HEREDITARY MOTOR AND SENSORY NEUROPATHY

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**ABSTRACT**

The postural organization in patients with hereditary motor and sensory neuropathy (HMSN) type I or II was studied clinically by means of a force platform. Balance was registered in 14 barefooted HMSN patients and healthy matched control subjects during quiet stance. The effect of visual deprivation was tested to determine the degree of visual dependency. The effect of the simultaneous performance of a concurrent attention-demanding task (Stroop task) was tested to estimate the level of balance automaticity. In comparison with control subjects, the HMSN patients showed a basically decreased efficiency of postural control ($p<.01$) as well as an increased visual control of posture ($p<.05$) in both directions of sway. No loss of balance automaticity was found, which suggested an on-line central adaptation to the slowly developing peripheral impairments. The results provide a starting point for understanding the balance problems and gross motor disabilities in HMSN patients.

## INTRODUCTION

Patients with hereditary motor and sensory neuropathy (HMSN) type I or II generally have distal muscle weakness and atrophy and loss of somatesthesis, especially of deep sensibility.[1,2] Due to a basically specific pattern of lower limb muscle atrophy, foot-ankle deformities develop such as pes cavus, transversoplanus, equinovarus, and hammer toes.[3-5] Progressive degenerative conditions are currently a major cause of acquired foot-ankle deformities.[6]

Lack of postural control is one of the most important functional consequences of the above-mentioned impairments caused by HMSN, which may lead to serious gross motor disability. Nonetheless, little work has been reported on this subject. This study focuses on the organization of postural control in type I and II HMSN patients to gain insight into some consequences of the underlying sensorimotor dysfunctions.

Balance was measured by means of a force platform permitting unrestrained upright standing. To obtain a system-oriented assessment of postural control, equilibrium reactions were studied in interaction with sensory input and central information processes using two separate task sets. In task set one, the basic level of control efficiency and the degree of visual dependency were determined.[7] Task set two used a dual-task procedure to evaluate the automaticity of balance control. Because a limited processing capacity was assumed,[8,9] the interference of a concurrent attention-demanding task (Stroop task[10,11]) served to estimate the amount of attentional resources needed to maintain balance.

## METHODS

### Subjects

Fourteen HMSN patients, eight with type I HMSN and six with type II, participated in the study. The nine females and five males had a mean age of 31.8 ± 11.1 years. All subjects were self-reliant and mobile without aids. No one had any other neurologic pathology. Every subject attended the outpatient department of the rehabilitation clinic at least once during a one-and-a-half-year period. An equal number of healthy individuals, matched for age (mean age = 31.2 ± 9.5 years) and gender, served as a control group. Informed consent was obtained from all subjects.

### Equipment

Balance measurements were made with a firmly secured force platform consisting of two separate aluminum plates, each placed on three force transducers (hysteresis and nonlinearity < 1%) recording the vertical ground reaction forces. Signals were processed by six DC-amplifiers (nonlinearity < 0.1%) and first-order low-pass filters(cut-off frequency 100Hz); then they were stored in a 8086 microprocessor after a 14-bit AD-conversion at a sampling rate of 20Hz.

By means of digital moment-of-force calculations, the virtual center of the ground reaction forces was determined for each sample, with a maximum error of ±1mm in both directions of sway. The coordinates of this center of pressure (CP) were passed through a digital low-pass 5-Hz filter.

The force platform was placed 1.5m in front of a white projection screen (120 x 110cm). In task set two, a remote-controlled slide projector was used to project samples of the Stroop test onto the screen.

**Procedure**

During all balance recordings, subjects stood barefoot on the force platform with their feet positioned against a foot frame (the medial sides of the heels 8.4cm apart; each foot toeing-out at a 9° angle from the sagittal midline). The subjects were repeatedly instructed to stand as still and symmetrically as possible with their hands folded in back of them. In all conditions with normal vision, no specific instructions were given concerning visual attention.

Task set one consisted of three conditions--standing with eyes open, with blurred vision (wearing milky white spectacles to prevent visual anchoring) and with eyes closed (wearing dark spectacles). Each of the three conditions was recorded for 20 seconds with at least one minute rest between conditions. A fixed sequence of conditions--eyes open, blurred vision, eyes closed--was used. In this way, order effects were kept constant, which ensured reliable comparisons among subjects. The complete sample of the 20-second registration represented postural control in each condition.

In task set two, standing still was assessed with and without the simultaneous performance of a secondary attention-demanding task (modified Stroop test[10,11]). Balance was registered for 30 seconds. During the first 15 seconds, balance was tested in a single-task condition. After 15 seconds, a sample of the Stroop test was added to the primary task. Subjects were instructed to start this concurrent task as soon as possible, with emphasis on maintaining the same balance strategy.

The modified Stroop test was projected onto the screen at eye level at a size of approximately 70cm (width) x 35cm (height). It consisted of a presentation of 25 colored words (five lines of five words), representing color names that were always incongruent with the printed ink colors. For instance, the word "blue" was printed in the color red. In every sample, the same four colors--yellow, red, blue and green--for words and inks were used in random order. The subjects were instructed to name the colors of the inks as quickly as possible until the end of the test procedure (30th second). At the same time, they had to suppress the strong tendency to read the words. The first five seconds of the second half of the registration were discarded from the analysis because undesired visual or transitional effects might have been an influence. Consequently, the last ten seconds represented balance in a dual-task condition and were compared to the ten seconds before the introduction of the secondary task, which represented the single-task condition.

In each assessment, task set one was completed first. Then the modified Stroop test was practiced three times in a sitting position. After practicing, it was recorded in this position to obtain a reference measure of the single-task performance. After ten minutes rest, task set two was performed. The Stroop test performance was assessed by the number of colors named in 15 seconds. The number of mistakes was noted also.

**Data Analysis**

The following parameters were derived from the CP fluctuations in the fore-aft (FA) and lateral (LAT) directions: the mean CP position (Pcp), the root mean square (RMS) amplitude of the CP displacements (Acp), and, after a first-order differentiation, the RMS of the CP velocities (Vcp). In this experiment, these parameters were derived either from a 20-second (task set one) or 10-second (task set two) registration period.

Because the Vcp integrates amplitude and frequency into a single measure of regulatory activity, this measure was primarily used to detect differences in control efficiency between conditions. The Acp was regarded as a measure of postural stability. In addition, the ratio between the Vcp and Acp served to estimate the mean frequency (Fcp) using the following approximation: $Fcp = Vcp/(Acp \times 4 \times \sqrt{2})$.[12] If not specified, in this article, the data on balance control refer to the Vcp values.

The Pcp was expressed as a percentage of the maximum length (FA direction) and the maximum width (LAT direction) of support with the zero point at the rear in the sagittal midline. Thus, the width of support was the distance between the sagittal midline and the lateral edge of the fifth metatarsophalangeal joint, whereas the length of support was the distance between the heel and the tip of the first toe. The base-of-support measures were determined by drawing the circumference of the feet on a piece of paper while the subjects stood on the force platform.

**Statistical Analysis**

Differences among groups or between conditions within the same group were analyzed by means of a distribution-free Wilcoxon matched-pairs signed-ranks test.

## RESULTS

The results of task set one and two are shown, respectively, in Tables 1 and 2. All Vcp and Acp data were significantly greater in the HMSN group than in the control group, with the only exception of the FA amplitude in the single-task condition (Table 2).

In both the HMSN and the control groups, no significant difference of sway was found between the eyes-open and single-task condition. With respect to the FA sway under these basic circumstances, the mean Fcp was significantly higher in the patient group (0.6Hz) than in the control group (0.4Hz) ($p<.05$). Conversely, the mean Fcp on the LAT sway in these conditions revealed no group differences (0.4Hz).

**Table 1: Results of Task Set One ($n$ = 14).**

| | HMSN (mean ± SD) | Controls (mean ± SD) | Two-tailed Probability Wilcoxon Test |
|---|---|---|---|
| *Eyes open, fore-aft direction* | | | |
| Vcp[a] | 13.2±6.4 | 6.4±1.6 | $p<.005$ |
| Acp[b] | 4.1±1.1 | 3.3±1.3 | $p<.05$ |
| Pcp[c] | 41.9±8.1 | 39.0±6.6 | NS |
| *Eyes open, lateral direction* | | | |
| Vcp[a] | 7.4±3.9 | 3.8±1.1 | $p<.005$ |
| Acp[b] | 3.9±1.8 | 1.7±0.6 | $p<.005$ |
| Pcp[c] | -4.3±7.7 | -1.1±3.9 | NS |
| *Blurred vision, fore-aft direction* | | | |
| Vcp[a] | 26.9±16.1 | 7.9±3.0 | $p<.005$ |
| Acp[b] | 6.8±2.9 | 3.9±1.5 | $p<.05$ |
| Pcp[c] | 42.2±7.3 | 42.5±5.1 | NS |
| *Blurred vision, lateral direction* | | | |
| Vcp[a] | 12.3±8.1 | 3.9±2.0 | $p<.005$ |
| Acp[b] | 5.4±3.2 | 1.7±0.9 | $p<.01$ |
| Pcp[c] | -3.1±8.4 | 0.1±3.3 | NS |
| *Eyes closed, fore-aft direction* | | | |
| Vcp[a] | 25.3±15.0 | 8.6±3.2 | $p<.005$ |
| Acp[b] | 6.9±2.8 | 4.2±1.9 | $p<.01$ |
| Pcp[c] | 43.1±8.3 | 42.3±4.2 | NS |
| *Eyes closed, lateral direction* | | | |
| Vcp[a] | 10.9±6.6 | 4.2±1.8 | $p<.005$ |
| Acp[b] | 4.9±3.8 | 1.9±0.9 | $p<.005$ |
| Pcp[c] | -2.9±9.1 | 0.5±2.9 | NS |

[a]RMS of the CP velocities (mm/sec). [b]RMS amplitude of the CP displacements (mm). [c]Mean CP position relative to the base of support (%); in the lateral direction, negative values correspond with a deviation toward the left foot, positive values toward the right foot.

**Table 2: Results of Task Set Two ($n$ = 14).**

| | HMSN (mean ± SD) | Controls (mean ± SD) | Two-tailed Probability Wilcoxon Test |
|---|---|---|---|
| *Single task, fore-aft direction* | | | |
| Vcp[a] | 13.8±6.6 | 5.6±2.1 | $p$<.005 |
| Acp[b] | 3.4±1.2 | 2.6±1.5 | NS |
| Pcp[c] | 43.2±6.8 | 38.9±6.0 | $p$<.05 |
| *Single task, lateral direction* | | | |
| Vcp[a] | 7.7±6.6 | 3.9±1.8 | $p$<.05 |
| Acp[b] | 2.9±1.4 | 1.5±0.7 | $p$<.005 |
| Pcp[c] | −3.6±6.7 | 0.2±3.6 | NS |
| *Dual task, fore-aft direction* | | | |
| Vcp[a] | 14.0±7.1 | 6.2±2.0 | $p$<.005 |
| Acp[b] | 3.2±1.1 | 2.4±0.5 | $p$<.05 |
| Pcp[c] | 42.4±7.3 | 38.7±5.9 | NS |
| *Dual task, lateral direction* | | | |
| Vcp[a] | 7.5±3.3 | 4.5±1.1 | $p$<.01 |
| Acp[b] | 2.9±1.2 | 1.6±0.4 | $p$<.005 |
| Pcp[c] | −4.1±7.0 | 0.0±3.6 | NS |

[a]RMS of the CP velocities (mm/sec). [b]RMS amplitude of the CP displacements (mm). [c]Mean CP position relative to the base of support (%); in the lateral direction, negative values correspond with a deviation toward the left foot, positive values toward the right foot.

A comparison between conditions (task set one) within groups showed significantly less efficiency of postural control in HMSN patients with eyes closed than with eyes open (FA sway: $p<.01$; LAT-sway: $p<.05$). A similar effect of eye closure was found in the control group; however, it was less pronounced and only significant with respect to the FA sway ($p<.05$). Because the blurred-vision condition provided identical results as the eyes-closed condition, these data were not further elaborated.

To compare the degree of visual dependency in both planes of balance control between the HMSN and control group, for every subject the absolute difference of Vcp was calculated between the eyes-closed and eyes-open performance (differential score) as well as the relative difference by dividing the eyes-closed by the eyes-open performance (quotient).[13] The mean quotient differed significantly between groups on the FA sway (HMSN: 2.03; controls: 1.43, $p<.05$) as well as on the LAT sway (HMSN: 1.52; controls: 1.09, $p<.05$). The mean differential scores showed the same pattern. There was a clear difference between groups on the FA sway (HMSN: 12.1; controls: 2.2, $p<.05$), which was a little less significant with respect to the LAT sway (HMSN: 3.5; controls: 0.4, $p<.08$). No relation was found between the degree of visual dependency and type of HMSN.

Relating the Vcp to the Acp values revealed that both in the HMSN and in the control group the decreased postural efficiency in the absence of visual information was mainly caused by an increase of sway amplitude, particularly in the FA direction ($p < .01$). In the absence of vision, the mean Fcp remained fairly constant.

Significant differences between balancing with and without the simultaneous performance of the Stroop test (task set two) were found neither in the HMSN group nor in the control group. In both groups also the Stroop test was performed equally in the single-task (sitting) as in the dual-task (balancing) condition. Consequently, no dual-task interference was present in any group.

The Pcp data derived from task set one and two are also presented, respectively, in Tables 1 and 2. No significant differences among groups were found, except in the single-task condition in the FA direction (Table 2). This group difference, although less clear in the eyes-open condition (Table 1), was related to a forward-shifting phenomenon, which occurred only in the control group. In conditions without visual information, healthy subjects showed a mean Pcp that was a little anterior with respect to the eyes-open condition (Table 1). This forward shift was small but significant (blurred vision, $p < .02$; eyes closed, $p < .01$). Conversely, HMSN patients showed a mean Pcp which was already relatively anterior with eyes open. There was no further shift with visual deprivation.

## DISCUSSION

The purpose of this study was to provide insight into some motor, sensory, and cognitive aspects of postural organization in HMSN patients. Because the study focused on a selected group of type I and II HMSN patients who had been referred to a rehabilitation clinic to prevent further loss of gross motor ability, some precautions must be made in generalizing the results to other HMSN patients. In addition, only standing still on a firm base of support was examined. It is conceivable that more dynamic gross motor tasks might have provided different results. Nevertheless, a few general considerations will be mentioned.

A decreased postural control efficiency in HMSN patients, even with full perceptual information, has been demonstrated. This basic lack of balance control can be attributed to lower limb muscle pareses and foot-ankle deformities, through which the efficacy of the ankle strategy, an essential mechanism for controlling the FA body sway,[14,15] is considerably reduced. A relatively high mean frequency in the FA direction may also result from distal muscle weakness. To prevent large torques around the ankle joints, a higher regulatory activity was adopted. Despite this strategy, the FA amplitude, which was greater in the HMSN group than in the control group, also reflected a loss of postural stability.

Besides a basic loss of postural control in HMSN patients, the data demonstrated a disproportional decrease of regulatory efficiency in conditions without visual information. This increased visual dependency indicated a substantial loss of somatesthesis both in type I and type II patients. The finding that the decrease of postural efficiency coincided with an increase of amplitude suggested a further postural destabilization in the absence of vision.

Despite the clear input and output deficiencies in most HMSN patients, the lack of dual-task interference indicated a substantial level of balance automaticity. This result can be explained by assuming a continuous central reorganization process, which ensures an optimum adaptation to the slowly developing peripheral impairments. It forms an interesting contrast with earlier findings in persons with lower limb amputation, who demonstrated a marked dual-task interference on the same balance procedure, particularly during the early phases of rehabilitation.[16]

The relatively anterior Pcp in HMSN patients in the presence of normal vision may be interpreted as a safety strategy. Indeed, the anterior tibial muscles, which prevent backward falling, are among the first muscles affected by HMSN.[1,3] Because the triceps surae muscles remain relatively strong in relation to their antagonists, a few degrees forward body inclination, although energetically unfavorable, appeared to be an effective balance control strategy.

## CONCLUSION

A serious lack of postural control results from HMSN, which may account for a substantial part of the gross motor disabilities observed in these patients. Although at the impairment level the sensory deficits are commonly considered to be subsidiary to the motor consequences of HMSN,[1,2,17] this study indicates that balance problems in HMSN patients may have an afferent as much as an efferent origin. Because the visual system in particular increases its contribution to postural control in cases of sensory conflict or underspecification,[18,19] an abnormally high visual control of posture is an important functional consequence of HMSN.

The considerable level of balance automaticity found in the HMSN group illustrates the adaptability of the central nervous system to peripheral pathology. This adaptability probably would prevent a much more serious breakdown of skills if the same impairments were abruptly imposed on the sensorimotor system. It may also account for the fact that HMSN patients may become so well accommodated to their disease, that its insidious onset is often missed both by the patient and the clinician.[1]

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# CHAPTER 7

# INFLUENCE OF ORTHOPEDIC FOOTWEAR ON POSTURAL CONTROL IN PATIENTS WITH HEREDITARY MOTOR AND SENSORY NEUROPATHY

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## ABSTRACT

The influence of orthopedic footwear on postural control was examined in ten patients with hereditary motor and sensory neuropathy (HMSN), type I or II, at the start and at the end of an individual training programme. Balance was registered during quiet stance on a force platform. In addition to the basic level of postural control, the degree of visual dependency was tested (Romberg test), as well as the level of balance automaticity (dual-task procedure). Besides a tendency towards stabilization of lateral body sway, no long-term influence of orthopedic footwear on postural control was found. However, before the start of the training programme, new footwear caused a marked loss of fore-aft sway control, which became overt during the simultaneous performance of an attention-demanding task. This loss of balance automaticity was explained on the basis of a central adaptation of the postural organization to the sudden alteration of peripheral constraints. Some recommendations are made to facilitate the central adaptation process to new orthopedic footwear in neurologically disabled patients.

## INTRODUCTION

Patients with hereditary motor and sensory neuropathy (HMSN) suffer from a slowly progressive neuronal degeneration which affects the peripheral motor, sensory and autonomic neurons. Following the classification by Dyck (1984), types I and II (respectively the hypertrophic and neuronal type of Charcot-Marie-Tooth) are the more frequent expressions of HMSN, commonly with an onset early in life. Apart from pathophysiological differences, the clinical signs of both these types are quite similar and relatively benign (Harding and Thomas, 1980; Joosten, 1988).

As for the lower extremities, the most prominent efferent impairments in types I and II HMSN are a predominantly distal muscle atrophy, starting in the intrinsic foot muscles and peroneus brevis, followed by the anterior tibial muscle, peroneus longus and the long toe extensors (Dyck, 1984; Mann and Missirian, 1988). With regard to afferent impairments, a loss of somatesthesis can be demonstrated in many patients (Harding and Thomas, 1980). The typical pattern of lower limb muscle atrophy gradually causes muscular imbalance inside the foot and across the ankle joint. As a result, foot deformities develop such as hammer toes and equinus of the first metatarsal leading to pes cavus and pes transversoplanus with a marked adduction of the forefoot. Eventually, subtalar inversion causes pes varus (Jahss and Lusskin, 1982; Sabir and Lyttle, 1983; Mann and Missirian, 1988).

Serious gross-motor disabilities may result from the above-mentioned impairments in HMSN, in particular a lessened control of stance and gait (clumsiness, footdrop, steppage gait, toe-walking, medial-lateral ankle instability). To prevent further deformity and pressure sores and in order to preserve functional abilities, orthopedic footwear is recommended (Dyck, 1984; Joosten, 1988). This made-to-measure footwear basically consists of a corrective inlay to counterbalance and compensate for foot deformities (low footwear). The inlay can also be incorporated in an ankle-foot socket which is built up inside the shoe to redress a varus deviation and to keep the talocrural joint in a 0° dorsiflexion position (high footwear). It has been shown that an ankle-foot orthosis can reduce ankle instability and prevent footdrop or initial forefoot contact in gait due to paresis of the foot dorsiflexors (Lehmann et al., 1986). Usually, a roll-off correction (early, biphasic) is made in the shoe to ensure smooth centre-of-pressure advancement during ambulation (see for a detailed prescription Rijken, 1991).

Although orthopedic footwear is frequently prescribed in patients with HMSN (as well as in other neurological pathologies), the influence of this footwear on the control of stance and gait is not well understood. For instance, whereas some patients profit from their footwear almost immediately, many others seem to experience marked adaptation problems particularly with regard to maintaining balance. Because balance has seldom been studied in relation to orthotic appliances, this study is focused on the short- and long-term influence of orthopedic footwear on postural control in type I and II HMSN patients. A major goal is to provide a better insight into the mechanisms that underlie the adaptation process (or problem) of the patient to new orthopedic footwear. Such knowledge would help to understand the interaction between this footwear and the sensorimotor system and may, indirectly, improve the footwear prescription and rehabilitation in HMSN. Before we specify in more detail the variables under investigation as well as the assumptions made in this study, we will briefly discuss some theoretical notions that have led to the employed postural tasks.

Orthopedic footwear will almost inevitably influence the size of the support base as well as the sensory feedback from the footsole, both of which are relevant factors in postural control (Horak and Nashner, 1986; Magnusson et al., 1990). High footwear will also reduce ankle mobility and proprioceptive feedback from the ankle joint and lower leg muscles; mechanisms that play an essential role in the control of fore-aft body sway ("ankle strategy") (Nashner, 1976; Nashner and McCollum, 1985; Horak and Nashner, 1986). In other words, orthopedic footwear creates a substantial change of mechanical conditions and sensory feedback, which could influence both the efferent and afferent control of posture.

The sudden change of peripheral constraints will force the central nervous system (CNS) to rapidly adjust to the novel situation (see for similar argumentation Frank and Earl, 1990). It is predicted here that such a central reorganization may (temporarily) increase the central-processing demands in postural control and, thus, reduce the degree of balance automaticity. Because the central-processing (or attentional) capacity is limited (Schmidt, 1988), interference may occur with other attention-demanding tasks. Such reduction in balance automaticity may be particularly apparent in HMSN patients, because a subtle equilibrium exists between slowly developing peripheral impairments and long-term central adaptations which probably is rather vulnerable (Geurts et al., 1992).

It is important to note that this study searches for general principles in the influence of new orthopedic footwear on postural control, which implies that no attempts are made to evaluate the effects of specific footwear components in individual subjects. Instead, we have assumed that all patients were provided with appropriate footwear which could differ between subjects in many details. Emphasis is laid on the *change* of peripheral constraints induced by the orthopedic footwear, rather than on the type of footwear. We will only make a (descriptive) distinction between high and low footwear because of the importance of ankle mechanisms in normal standing. We examined the adaptation to new footwear by making balance assessments at the start and at the end of an individual training programme. Again, we have no intention to evaluate the training programme in itself, but to assess the effects of motor learning, whether the result of "spontaneous" adaptation or of rehabilitation treatment.

Quiet standing was registered while subjects stood upright on a force platform wearing orthopedic footwear. In addition, barefoot standing was recorded as a control condition employing a within-subjects design. We examined only the dynamic aspects of the centre-of-pressure fluctuations under the feet reflecting the *regulation* of posture. Besides the basic level of postural control, the degree of visual dependency was determined (Romberg test) as an indirect measure of the somatosensory contribution to the afferent control of posture (task set one). To estimate the level of balance automaticity, the interference of a concurrent attention-demanding task (modified Stroop test) with balance was assessed (task set two). The theoretical background of this dual-task procedure has been discussed earlier (Geurts et al., 1991; Mulder and Geurts, 1991).

**Table 1: Characteristics of HMSN Patients ($n = 10$) and Type of Orthopedic Footwear.**

| Subject | Gender | Age | Type HMSN | Type footwear |
|---|---|---|---|---|
| 1 | M | 20 | I | low |
| 2 | M | 39 | I | high |
| 3 | F | 21 | I | high |
| 4 | F | 25 | I | high |
| 5 | F | 12 | I | low |
| 6 | F | 44 | I | low |
| 7 | F | 32 | I | high |
| 8 | M | 18 | II | low |
| 9 | M | 34 | II | high |
| 10 | M | 32 | II | high |

## METHODS

### Subjects

Ten HMSN patients participated in the study. All subjects were self-reliant and mobile without aids. No one suffered from any concomitant neurological pathology. Every subject attended the out-patient department of the rehabilitation clinic at several moments; all within a period of one-and-a-half year. During this period, six patients (mean age 30.5 years) received high orthopedic shoes or boots with a rigid ankle-foot socket. This socket had either a circular fit or a stiffened dorsal tongue and was built up inside the footwear to well above the ankle joints (18 to 20 cm from the base of the calcaneus). Four patients (mean age 23.5 years) received low orthopedic shoes. All subjects experienced the use of orthopedic footwear for the first time Table 1 summarizes some relevant patient characteristics, together with the type of footwear.

**Equipment**

Balance was measured with a dual-plate force platform recording the vertical ground reaction forces. Force signals were stored into a microprocessor at a sampling rate of 20 Hz. By means of digital moment-of-force calculations, the point of application of the resultant of the ground reaction forces in a transverse plane was determined for each sample, with a maximum error of ± 1 mm in both directions of sway. The coordinates of this "centre of pressure" (CP) were passed through a digital low-pass 5-Hz filter. The force platform was placed 1.5 m in front of a white projection screen (120 x 110 cm) A remote-controlled slide projector was used to project samples of the Stroop test onto the screen (task set two).

**Procedure**

During all balance recordings, subjects stood on the force platform with their feet against a foot frame (the medial sides of the heels 8.4 cm apart; each foot toeing-out at a 9° angle from the sagittal midline). The subjects were repeatedly instructed to stand as still and symmetrically as possible with their hands folded on the back. In all conditions with normal vision, no specific instructions were given concerning visual attention.

Task set one consisted of three conditions in the following order--standing with eyes open, with blurred vision (wearing milky white spectacles preventing visual anchoring) and with eyes closed (wearing dark spectacles). Each of the three conditions was recorded for 20 seconds with at least a one minute's rest between conditions. The tests were performed in a fixed sequence to keep order effects constant, which ensured reliable comparisons between different situations.

In task set two, standing still was assessed with and without the simultaneous performance of an attention-demanding task (modified Stroop test (Stroop, 1935)). Balance was registered for 30 seconds. During the first 15 seconds, balance was tested as a single task. After 15 seconds, a sample of the Stroop test was added to the primary task. Subjects were instructed to start this concurrent task as soon as possible, with emphasis on maintaining the same balance strategy.

The modified Stroop test was projected onto the screen at eye level (70 x 35 cm). It consisted of a presentation of 25 coloured words (5 x 5), representing colour names that were always incongruent with the printed ink colours. For instance, the word "green" was printed in yellow. In every sample, the same four colours (yellow, red, blue, green) for words and inks were used in random order. The subjects were instructed to name the colours of the inks as quickly as possible until the end of the test procedure, while suppressing a strong tendency to read the words. The first five seconds of the second half of the registration were discarded from the analysis because undesired visual or transitional effects might have been an influence. Consequently, the last ten seconds represented balance in a dual-task condition, whereas the ten seconds before the introduction of the secondary task represented the single-task condition (see Figure 1).

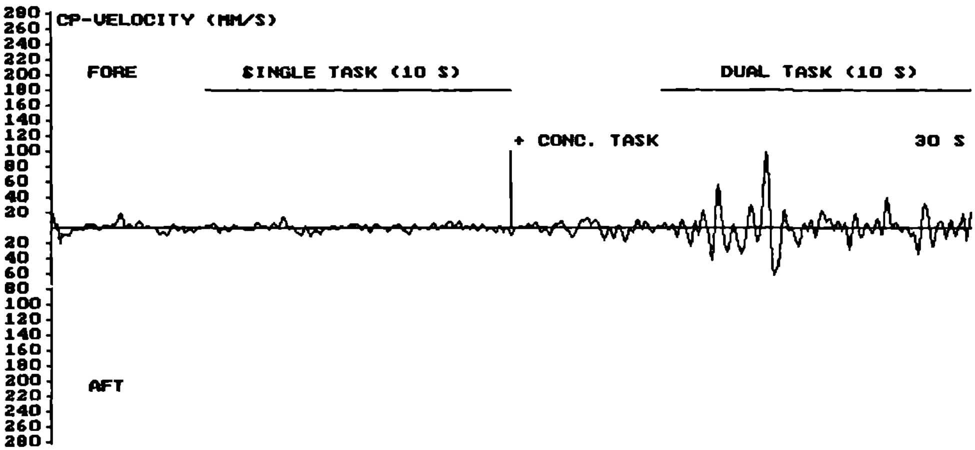


**Figure 1. A balance registration (fore-aft sway) of one of the HMSN patients wearing new rehabilitation footwear before the start of a training programme.** At the 15th second, the concurrent Stroop task is introduced. Note the difference between the single- and dual-task performance (task set two).

In each assessment, task set one was completed first. Then, the modified Stroop test was practised three times in a sitting position. After practising, it was recorded in this position to obtain a reference measure of its single-task performance. After ten minutes physical rest, task set two was performed. The Stroop test performance was assessed by the number of colours named in 15 seconds. Also, the number of mistakes was noted.

With each subject, the above-described assessment was completed twice: with orthopedic footwear and with bare feet. When necessary, subjects were allowed to adjust to each situation for several minutes. Because most patients arrived at the experimental room wearing their orthopedic footwear, balancing with footwear was selected as the first condition to ensure optimal patient compliance.

Each patient was tested one or two hours after the application of new orthopedic footwear. In the first instance, try-out "rehabilitation footwear" was provided, which is a modifiable prototype of the definitive orthopedic footwear with most of its mechanical properties. After an individual programme of stance and gait training (varying from two to four months), all patients were retested following the same procedure. At this time, they had already been provided for one or two weeks with definitive orthopedic footwear. The definitive footwear was similar to the rehabilitation footwear with regard to the presence of an ankle-foot socket. However, in most cases, modifications to the inlay, socket or roll-off correction were made during the training period. All decisions on the footwear prescription and training were made independently of the research team.

**Data Analysis**

The following parameters were derived from the CP fluctuations in the fore-aft (FA) and lateral (LAT) directions: the root mean square (RMS) amplitude of the CP displacements (Acp) and, after a first-order differentiation, the RMS of the CP velocities (Vcp). These parameters were derived from either a 20-second (task set one) or a 10-second (task set two) registration period.

Because of its frequency-dependency, the Vcp was primarily used to detect differences in postural control. Indeed, specifically the higher frequency components of the CP fluctuations reflect the compensatory torques which are applied at the support surface by muscular activity to counterbalance the accelerations of body mass (Murray et al., 1967; Hufschmidt et al., 1980). In contrast, the Acp is often determined by the lower frequency components of the CP fluctuations (well below 1 Hz) and is, therefore, believed to be more related to body sway or "instability". In addition, the ratio between the Vcp and Acp served to estimate the mean frequency (Fcp) using the following approximation: $Fcp = Vcp / (Acp \times \sqrt{2} \times 4)$ (Hufschmidt et al., 1980).

## RESULTS

All data were tested in a within-subjects design using a three-way multivariate analysis of variance (MANOVA) of Footwear (barefoot vs. orthopedic footwear) x Training (before vs. after) x Condition with repeated measures for each factor.

Separate analyses were done for different dependent variables. Because there were never significant differences between the blurred-vision and eyes-closed performances (paired $t$ tests), the data of task set one were tested for a Footwear x Training x Vision (eyes open vs. closed) interaction. The balance data of task set two were tested in a Footwear x Training x Attention (single vs. dual task) analysis, whereas the Stroop data were tested in a Footwear x Training x Balance (sitting vs. standing) analysis of variance (see e.g. Maxwell and Delaney, 1990). Only a descriptive distinction was made between low and high footwear due to the small numbers in each subgroup. The Vcp and Acp values derived from task sets one and two are presented in Tables 2 and 3, respectively.

**Table 2: Balance Results of Task Set One (Romberg Test) ($n = 10$).**

| | Before training | | After training | |
|---|---|---|---|---|
| | Barefoot | Footwear | Barefoot | Footwear |
| *Eyes open, fore-aft direction* | | | | |
| Vcp[a] | 11.1±5.7 | 10.8±5.2 | 11.2±4.0 | 12.0±5.8 |
| Acp[b] | 3.9±1.1 | 3.9±1.3 | 4.2±1.6 | 4.1±1.3 |
| *Eyes open, lateral direction* | | | | |
| Vcp[a] | 6.1±2.7 | 5.0±2.4 | 6.6±2.5 | 5.8±2.6 |
| Acp[b] | 4.0±2.1 | 2.2±0.9 | 3.0±1.2 | 2.6±1.6 |
| *Blurred vision, fore-aft direction* | | | | |
| Vcp[a] | 21.6±14.9 | 21.6±11.8 | 20.9±13.0 | 25.5±23.6 |
| Acp[b] | 6.0±2.6 | 6.0±2.5 | 6.7±3.4 | 6.9±4.3 |
| *Blurred vision, lateral direction* | | | | |
| Vcp[a] | 9.3±6.0 | 7.9±3.5 | 9.5±3.9 | 9.9±10.7 |
| Acp[b] | 5.0±3.4 | 2.8±1.2 | 4.2±1.7 | 3.7±3.4 |
| *Eyes closed, fore-aft direction* | | | | |
| Vcp[a] | 22.5±14.6 | 24.9±18.9 | 23.9±17.7 | 22.0±12.4 |
| Acp[b] | 6.4±2.6 | 5.9±3.2 | 7.6±4.2 | 6.6±2.6 |
| *Eyes closed, lateral direction* | | | | |
| Vcp[a] | 9.3±6.7 | 8.4±5.7 | 10.8±6.6 | 8.6±4.9 |
| Acp[b] | 5.3±4.4 | 2.9±2.4 | 5.6±4.1 | 3.0±2.1 |

[a]RMS of the CP velocities (mm/s). [b]RMS amplitude of the CP displacements (mm).

**Task Set One**

With respect to the Vcp data, only vision demonstrated a main effect both on the FA, $F(1,9) = 8.63$, $p < .05$, and on the LAT, $F(1,9) = 6.66$, $p < .05$, sway, indicating less postural control with eyes closed. No significant two- or three-way interaction in either direction was found. Hence, there was no influence of footwear or training on the degree of visual dependency. The Acp data were also greater in the absence of vision, however, the main effect of vision was only significant for the FA sway, $F(1,9) = 10.89$, $p < .01$. In addition, a significant increment in Fcp contributed to the greater CP velocities without vision in both the FA, $F(1,9) = 5.91$, $p < .05$, and LAT, $F(1,9) = 8.10$, $p < .05$, directions.

**Table 3: Balance Results of Task Set Two (Dual-Task Procedure) ($n = 10$).**

| | Before training | | After training | |
|---|---|---|---|---|
| | **Barefoot** | **Footwear** | **Barefoot** | **Footwear** |
| *Single task, fore-aft direction* | | | | |
| Vcp[a] | 12.1±6.7 | 9.6±5.2 | 14.0±6.9 | 11.5±5.7 |
| Acp[b] | 3.2±1.3 | 3.5±1.8 | 4.1±1.7 | 3.3±1.6 |
| *Single task, lateral direction* | | | | |
| Vcp[a] | 5.7±3.0 | 4.6±2.2 | 7.4±3.6 | 5.9±4.2 |
| Acp[b] | 2.5±1.2 | 1.7±1.0 | 2.8±1.5 | 2.1±1.9 |
| *Dual task, fore-aft direction* | | | | |
| Vcp[a] | 13.8±8.3 | 19.1±12.8 | 15.4±11.0 | 13.7±8.3 |
| Acp[b] | 3.2±1.3 | 4.0±1.7 | 3.3±1.3 | 3.1±2.0 |
| *Dual task, lateral direction* | | | | |
| Vcp[a] | 7.2±3.2 | 6.9±3.8 | 7.4±4.3 | 6.2±3.6 |
| Acp[b] | 3.0±1.3 | 2.2±1.1 | 2.6±1.5 | 2.2±1.4 |

[a]RMS of the CP velocities (mm/s). [b]RMS amplitude of the CP displacements (mm).

The increase in Fcp in the LAT direction with eyes closed was strongly related to the influence of orthopedic footwear. Indeed, there was a main effect of footwear on the Fcp in this direction, $F(1,9) = 41.16$, $p < .001$, as well as a Footwear x Vision interaction, $F(1,9) = 9.14$, $p < .05$. Hence, the footwear caused a shift towards higher Fcp values in the LAT direction, which was most apparent when no visual information was available. This shift coincided with a stabilizing main effect of footwear on the LAT CP amplitude, $F(1,9) = 13.76$, $p < .01$. Thus, orthopedic footwear induced a marked trade-off between CP amplitude and frequency with respect to the LAT sway. The net result was a (non-significant) tendency towards lower CP velocities with orthopedic footwear in this direction.

**Task Set Two**

With regard to the Vcp data in the FA direction, only attention showed a main effect, $F(1,9) = 7.42$, $p < .05$. However, there was a significant Footwear x Training x Attention, $F(1,9) = 10.32$, $p < .05$, interaction, as well as a Training x Attention, $F(1,9) = 10.86$, $p < .01$, a Footwear x Attention, $F(1,9) = 5.79$, $p < .05$, and a

Footwear x Training, $F(1,9)=4.89$, $p=.054$, interaction. This complex interactive pattern is illustrated in Figure 2. It shows that the poorest control of posture occurred before the training period when the subjects stood with new rehabilitation footwear while simultaneously performing the Stroop task.

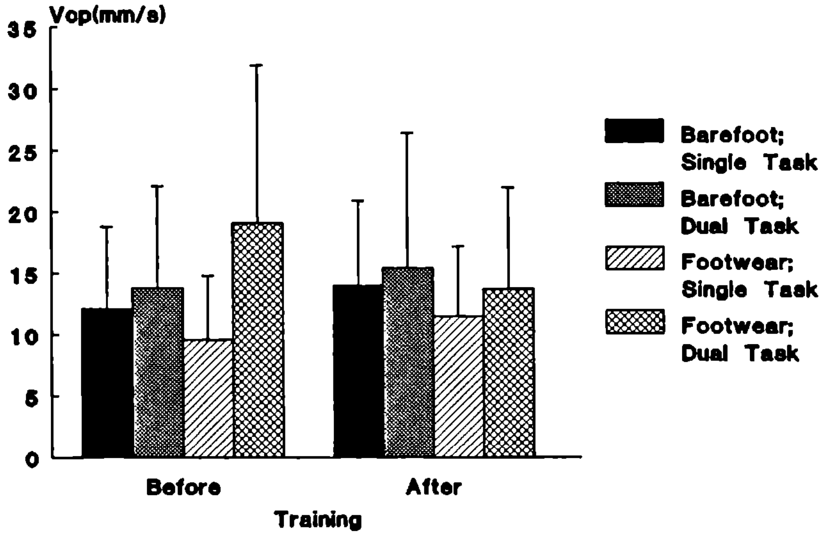


Figure 2. The group mean and SD of the RMS CP velocity (Vcp) in the fore-aft direction of body sway, derived from task set two, are shown for the HMSN patients (n = 10) standing with and without orthopedic footwear before and after an individual training programme. Note the dual-task interference with new footwear before the training period.

Indeed, before the start of the training, there was a main effect of attention, $F(1,9)=10.65, p<.05$, as well as a Footwear x Attention interaction, $F(1,9)=11.32$, $p<.01$. Further analysis revealed that the influence of the concurrent attention-demanding task on postural control was only significant for the footwear condition, $F(1,9)=12.37$, $p<.01$. Figure 1 gives an individual example of such dual-task interference. A main effect of footwear or attention could no longer be demonstrated at the end of the training period, nor a Footwear x Attention interaction.

Based on the Acp data in the FA direction, no significant main or interaction effects were found. In contrast, the Fcp data in this direction yielded a main effect of attention before the training period only with new rehabilitation footwear, $F(1,9)=7.62, p<.05$. These results indicate that the above-mentioned increase in FA

CP velocity caused by the concurrent task performance was mainly caused by a shift of the mean frequency towards higher values. The average degree of dual-task interference (expressed as the difference of FA CP velocity between the dual- and single- task condition) was larger in subjects with high (11.4 mm/s) than in subjects with low (6.6 mm/s) rehabilitation footwear, suggesting a marked influence of ankle stabilization.

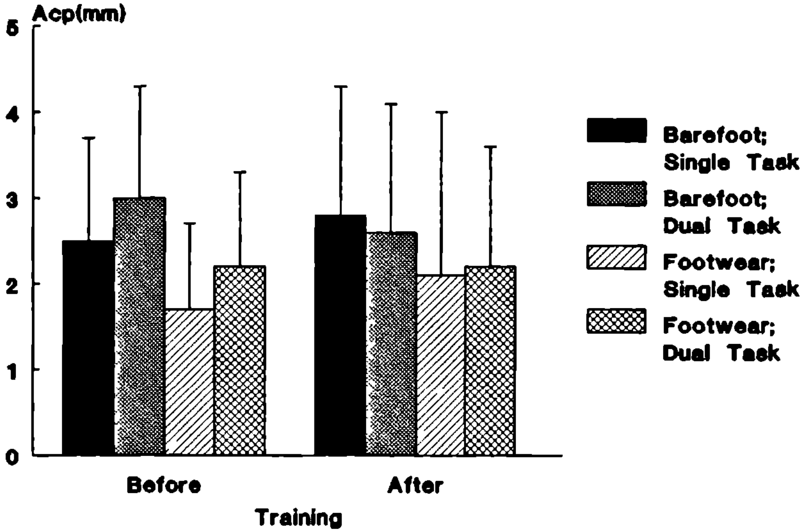


Figure 3. The group mean and SD of the RMS CP amplitude (Acp) in the lateral direction of body sway, derived from task set two, are shown for the HMSN patients ($n = 10$) standing with and without orthopedic footwear before and after an individual training programme Note the stabilizing effect of orthopedic footwear

The Vcp data in the LAT direction did not reveal any clear main or interaction effect. However, the Acp and Fcp data in this direction showed a pattern similar to the results of task set one. There was a stabilizing main effect of footwear on the Acp, $F(1,9) = 8.46$, $p < .05$, which is illustrated in Figure 3. Again, footwear enhanced the LAT Fcp, $F(1,9) = 10.25$, $p < .05$. The net result of this trade-off was a tendency towards lower LAT CP velocities, which was now marginally significant, $F(1,9) = 3.79$, $p =$ 084.

As for the Stroop task performances, no significant main or interaction effects were found that might neutralize the above-mentioned dual-task interference.

## DISCUSSION

In a related study in which type I and II HMSN patients were compared to healthy individuals (Geurts et al., 1992), HMSN patients showed a basically reduced control of posture, as well as an increased visual dependency in both directions of sway. There was no effect of concurrent task performance in either group indicating a substantial degree of balance automaticity also in the HMSN group in spite of their relative imbalance. The present study again revealed a marked effect of visual deprivation in both directions of sway, whereas dual-task interference was absent during barefoot standing. The latter result suggests an optimal adaptation of the CNS to the slowly developing peripheral impairments.

The emphasis in the present study was, however, on the short- and long-term influence of orthopedic footwear on postural control in type I and II HMSN to obtain a better insight into the interaction between this footwear and the sensorimotor system. We were particularly interested in whether a *postural* reorganization process might (in part) determine the adaptation of the patient to new footwear or, in different terms, to a sudden alteration of peripheral constraints. Although the experimental group was small and heterogeneous with respect to the severity of the balance problems and the individual footwear prescriptions, two significant phenomena deserve further discussion.

The first phenomenon concerned the control of FA body sway. The concurrent Stroop task performance caused a considerable loss of postural control when the subjects stood with new footwear just before the start of a training programme. Apparently, the footwear induced a marked decrease in balance automaticity which led to a disproportional degree of dual-task interference. It is improbable that (aspecific) order effects are responsible for such a significant influence of the footwear, because the single-task performance with footwear was always slightly better than the similar performance with bare feet (see Figure 2). Instead, the finding that dual-task interference was larger in subjects with high footwear and the fact that it was only found for the FA sway much more suggest an adverse influence of the footwear on ankle-strategy control (Nashner and McCollum, 1985; Horak and Nashner, 1986).

Indeed, both a roll-off correction and a rigid ankle-foot socket prevent the normal utilization of vertical ground reaction forces to maintain anteroposterior equilibrium, which is essential to the ankle strategy. These footwear components exert such an influence by shortening the length of the support base and by reducing talocrural mobility, respectively. They may also reduce proprioceptive feedback from the ankle joint and lower leg muscles due to a decrease in joint motion and muscle stretch. Because these ankle mechanisms are already impaired in HMSN by distal muscle atrophy, sensory impairments and foot deformities, a central reorganization of postural control, e.g. a shift towards "hip-strategy" control (Horak and Nashner, 1986), seems inevitable.

Remarkably, the data indicate that the adaptation process was not reflected in a (temporary) deterioration of the basic level of postural control or in an increased visual dependency. The adverse influence of orthopedic footwear on the use of ankle mechanisms was completely compensated for by the CNS, however, at the cost of considerable attention demands. Hence, this influence became overt only during the simultaneous performance of an attention-demanding task by a significant increase in the FA CP velocity. A relatively small effect on the CP amplitude in most of the patients indicates that maximum effort was made to prevent large displacements of the centre of body mass, which would enhance the need for effective ankle torque application at the support surface.

After a period of stance and gait training, the observed dual-task interference had completely disappeared. The fact that, at this moment, the subjects wore definitive orthopedic footwear seems to be of little importance in comparison with the fact that they had been given the opportunity to learn. Indeed, the definitive footwear diminished the use of ankle mechanisms in the sagittal plane in equal proportion. The reduction in dual-task interference, therefore, reflects the adaptability of the sensorimotor system to alterations of peripheral constraints. It is probable that neurologically disabled patients, suffering from combined motor and sensory impairments, need ample time to adapt to such a novel situation compared to subjects with an intact nervous system.

The second phenomenon of interest concerned the control of LAT body sway and had a more permanent character. The orthopedic footwear induced an increase in stability assessed by the average amplitude of the CP displacements. This influence was probably caused by a decrease in medial-lateral ankle instability. Based on the average CP velocity, however, the control of LAT body sway appeared to improve considerably less due to a concurrent increase in the mean frequency of the CP displacements. This amplitude-frequency trade-off indicates that stabilization at one level does not necessarily lead to a significant improvement of postural control of a multi-segment body (Keshner, 1990). Nevertheless, the results suggest a small beneficial effect of orthopedic footwear on LAT sway control during quiet standing. Naturally, the data of this study do not preclude relevant effects of this footwear on more dynamic postural skills.

## CONCLUSION

It appears that a central adaptation of the postural organization takes place after the application of new orthopedic footwear in HMSN patients. This adaptation process coincides with a temporary increase in the attention demands in postural control. It is argued that a reduced capacity to use ankle mechanisms in the control of fore-aft body sway may be an important factor determining the need for such a central adaptation. Based on these conclusions, the following tentative recommendations are made with regard to the footwear prescription and rehabilitation in HMSN, which might also be valid for other types of neurologically disabled patients.

To facilitate the adaptation of the patient to new ankle-stabilizing footwear, it is recommended to use a socket that permits sufficient talocrural mobility, especially dorsiflexion, thus redressing a varus deviation more specifically at the subtalar level. Hereby, also the control of gait would be improved, because anteroposterior postural strategies are very similar during in-place standing and during the support phase of locomotion (Nashner and Forssberg, 1986). If nonetheless, for the sake of ambulation, a pronounced early roll-off correction is needed, it seems appropriate to first provide a longer (stabilizing) support base to ensure sufficient postural stability during the initial phases of motor learning. Naturally, such a policy requires the use of modifiable (rehabilitation) footwear. Although this study does not give information about how fast central adaptation processes take place, we would recommend to combine the application of new orthopedic footwear with an individually tailored programme of stance and gait training to facilitate the learning process. In this way, also a premature evaluation of the functional outcome is prevented.


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# CHAPTER 8

## VULNERABILITY OF POSTURAL CONTROL TO

## SECONDARY TASK PERFORMANCE IN HEALTHY SUBJECTS

## BALANCING ON DIFFERENT SUPPORT-SURFACE CONFIGURATIONS

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**submitted for publication.**

## ABSTRACT

This study investigates whether the human posture-control system is vulnerable to secondary task performance dependent on the complexity of the primary balance task, i.e. the extent to which the task is not subserved by (a combination of) well-developed synergies. The fluctuations of the centre-of-pressure (CP) under the feet were registered by means of a force platform in 24 healthy adults while they maintained three different postures both in a single- and dual-task condition: besides quiet standing with the feet apart (*comfortable* stance), balancing on a pair of seesaws was examined (*seesaw* stance), as well as balancing on seesaws with the feet placed in a tandem position (*tandem* stance). As the concurrent activity, three different Stroop tasks were used of increasing complexity: the word card, the colour card, and the colour-word card. It was hypothesized that seesaw stance would only require a change in parametrization within well-developed ankle synergies working in the sagittal plane, whereas tandem stance would induce a radical shift towards poorly developed ankle mechanisms working in the frontal plane. Hence, little or no dual-task interference was predicted with regard to *fore-aft* sway control during seesaw stance. In contrast, with increasing complexity of the Stroop task, an increasing amount of dual-task interference was predicted for *lateral* sway control during tandem stance. The results of this study globally confirm these predictions. There was, however, no effect of the type of Stroop test on the amount of dual-task interference during tandem stance. This finding is attributed to the fact that the number of completed Stroop items decreased with increasing complexity of the Stroop task, thus leaving the central-processing demands unaltered. It is argued that the observed interference of Stroop task performance with lateral sway control during tandem stance is probably best accounted for by simultaneous attention demands on a general-purpose central processor due to a lack of automatic postural activity in the frontal plane. It is concluded that the automaticity of the posture-control system in man may decrease as a function of the novelty of the balance task.

## INTRODUCTION

It is generally accepted that the control of normal upright posture in man is a highly automated skill requiring little if any involvement of attention-demanding central processes (Nashner, 1977; Nashner & Cordo, 1981). Postural control is subserved by numerous neural pathways at spinal and supraspinal levels that constitute elementary reflexes and initially learned synergies which form the basis for fast, automatic responses to body perturbations (Brooks, 1986). These reflexes and synergies provide a continuous parametric control of gain and phase of feedback sensorimotor loops directed at maintaining a certain state of equilibrium (Droulez & Berthoz, 1986). This "lower" level mode of control is usually regarded as independent of attention-demanding cognitive processes and, thus, resistant to simultaneous central-processing demands because it requires only a minimum of computational activity (Nashner & McCollum, 1985).

The involvement of cognitive processes in the control of posture becomes more apparent if one considers the role of feedforward control in the adaptation to varying motor goals. Cognitive influence on postural control is reflected in the effect of prior knowledge on the timing of anticipatory postural adjustments to self-initiated arm movements (Horak, Esselman, Anderson, & Lynch, 1984; Lee, Buchanan, & Rogers, 1987) as well as in the modulation of the magnitudes of automatic postural responses to externally induced body perturbations by "central set" based on prior experience (Horak, Diener, & Nashner, 1989). Rather than feedback control, feedforward control is dependent on centrally stored information (internal representation) about the task and the body in terms of its geometry and dynamics (Frank & Earl, 1990; Gurfinkel & Levick, 1991). Still, cognitive influence on postural control is usually discontinuous, that is, during short periods of adaptation to new equilibrium states, e.g. during an alteration of support-surface configuration (Droulez, Berthoz, & Vidal 1985; Massion, 1992). Even in such instances, marked vulnerability of postural activity to secondary task performance on the basis of central capacity interference is not likely to occur, because probably few if any noticeable demands on attentional mechanisms are made. In most situations, healthy subjects automatically adopt an appropriate postural strategy by selecting (a combination of) prestructured synergies (Brown & Frank, 1987; Horak & Nashner, 1986).

Nevertheless, it may be that the human posture-control system is only *relatively* autonomous, since task concurrence costs have been reported during the execution of a difficult, poorly trained balance task (Kerr, Condon, & McDonald, 1985). Dependency on attention-demanding central processes seems even more apparent when the nervous system is impaired as in elderly subjects (Stelmach, Meeuwsen, & Zelaznik, 1990). Accordingly, in previous work (Geurts, Mulder, Nienhuis, & Rijken, 1991), we have found a detrimental influence of secondary Stroop task performance (Stroop, 1935) on standing balance in persons with acquired peripheral sensorimotor deficits due to lower limb amputation compared to matched controls. In contrast, similar dual-task interference was not observed in persons suffering from a slowly developing inherited peripheral neuropathy (HMSN type I or II) (Dyck, 1984), despite the fact that this degenerative disease also leads to a combination of sensory and motor impairments (Geurts, Mulder, Nienhuis, Mars, & Rijken, 1992). These results are interesting because they indicate the adaptability of the central nervous system to alterations of peripheral task constraints.

Hence, the available data suggest that the novelty of the postural task determines the processing demands imposed upon multi-purpose capacity-limited central operators, in interaction with the integrity of the nervous system. It is hypothesized here that, whenever existing strategies fail to cope with the task in an adequate way, there is a need to rely on novel or adapted motor patterns which requires a more continuous cognitive involvement in the control of posture. In these instances, there is a (temporary) shift towards attention-invested error detection and correction of inappropriate postural responses. During the early phases of learning, such cognitive, attention-invested control is characterized by slowness, errors, as well as by an increased vulnerability to concurrent task performance. If these theoretical considerations are valid, it should be possible to evoke a detrimental influence of concurrent (Stroop) task performance on balance control also in healthy subjects by reducing the efficacy of existing strategies, whereas such dual-task interference should be absent during the performance of balance tasks that are still adequately subserved by a combination of prestructured synergies. To test the validity of the above-mentioned hypothesis, this study addresses the question if and to what extent postural control can be influenced by secondary task performance in healthy adults balancing on different support-surface configurations. Until now, very few experimental data are available on this subject.

Besides quiet upright standing on a firm and flat support surface (*comfortable* stance), we examined standing on two other support-surface configurations in order to interfere with the efficacy of commonly employed postural strategies. Firstly, subjects were requested to balance on a pair of *seesaws*, thus complicating the utilization of vertical ground reaction forces through ankle torque generation to control fore-aft body sway (Horak & Nashner, 1986; Nashner & McCollum, 1985). Because it was believed that this manipulation would merely require a change in parametrization (timing and gain) within well-developed synergies, little or no interference of Stroop task performance with the control of *fore-aft* sway was predicted for this balance task. Secondly, subjects were asked to stand in a *tandem* position on the same seesaws, thus completely eliminating the intrinsic mechanical stability of lateral balance which is normally provided by double-limb support in the frontal plane. The seesaws again complicate the control of fore-aft sway by reducing the efficacy of ankle mechanisms in the sagittal plane, however biped stability is now available in this plane. Hence, tandem stance was primarily expected to induce a radical shift of lateral sway control towards the generation of high-frequency ankle torques working in the frontal plane. Because this control mechanism is only poorly developed in non-athletic subjects, a clear dual-task effect on *lateral* sway control was predicted for this task.

With every subject, we combined each balance task with each of the three cards of the Stroop test: the word card, the colour card and the colour-word card (within-subjects design). It was assumed that, in this sequence, the three Stroop tasks would demand an increasing amount of information processing capacity and, therefore, would increasingly interfere with lateral sway control during the tandem stance task. The Stroop task was selected because its performance requires a considerable amount of attention even after many repetitions and because it comprises three discrete levels of complexity. Using the Stroop task as the concurrent activity also facilitates comparisons of the results with previously collected data in persons with lower limb amputation or HMSN.

## METHODS

### Subjects

Twenty-four healthy individuals, 12 females and 12 males, aged 20 to 40 years, participated on the basis of voluntariness. The experimental group consisted of students, therapists, as well as technical and civil personnel of a rehabilitation clinic. Subjects with more than average balance skills acquired by special activities such as dancing or gymnastics were not included. Every subject had normal or corrected-to-normal visual acuity, as well as unimpaired colour perception.

### Equipment

Balance measurements were made with a dual-plate force platform recording the vertical ground reaction forces. Force signals were amplified and led through first order low-pass filters with a cut-off frequency of 30 Hz. After a 12-bit AD-conversion, they were stored into a personal computer (COMPAQ 386SX, 16 MHz) at a sampling rate of 60 Hz. By means of moment-of-force calculations, the point of application of the resultant of the ground reaction forces in a two-dimensional transverse plane was determined for each sample with a maximum error of ± 1 mm in both directions. The coordinates of this centre of pressure (CP) were passed through a digital low-pass Fourier filter with a cut-off frequency of 6 Hz to eliminate high-frequency components due to noise or tremor. A remote-controlled slide projector was used to project samples of the Stroop test onto a white projection screen, which was placed 1.5 m in front of the platform.

**Balance Tasks**

A quiet upright posture had to be maintained on three different support-surface configurations: (a) with each foot positioned against a foot frame and placed on one of the two force plates (distance between the medial sides of the heels 8.4 cm, toeing-out angle 0°) (*comfortable* stance), (b) standing in the same position but now with each foot supported by a seesaw placed on one of the two force plates (*seesaw* stance), and (c) standing on the same seesaws but now in a tandem position with the heel of the anterior foot directly ahead of the toes of the posterior foot (*tandem* stance). The selected balance tasks are schematically illustrated in Figure 1.

The seesaws used in the latter two balance tasks consisted of wooden platforms (30 x 10 cm) with a curved base (radius 40 cm). With the platform in a horizontal position, the line of contact with the ground was at a distance of 11 cm from the top and was located asymmetrically at 40% length (12 cm) from the rear.

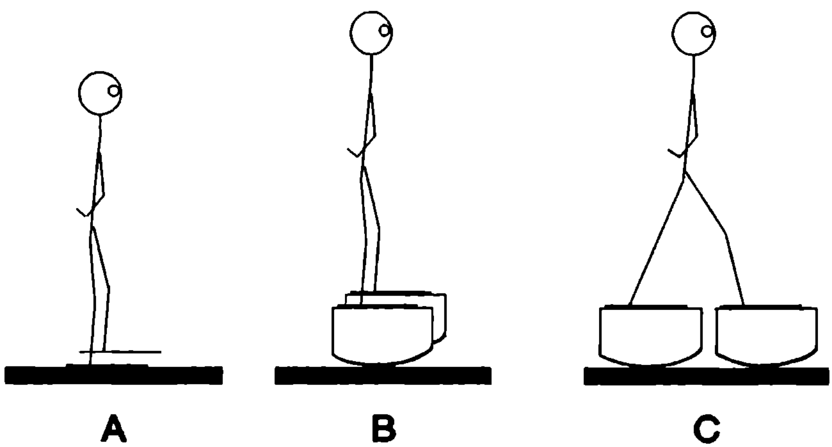



**Figure 1. A schematic illustration of the three balance tasks with the subject standing on the force platform: (A) comfortable stance, (B) seesaw stance, and (C) tandem stance; during tandem stance the subject is rotated 90° to the right with respect to the position during comfortable and seesaw stance.**

At the bottom of each seesaw, a longitudinal groove was made to guide its rolling motions along a rail fixed to the platform. In this way, rotations along the vertical axis were prevented. Foot placement on the seesaws was individually adjusted in such a way that the ankle joint was slightly behind the rotation point of the seesaw in a horizontal position (see Figure 1). As for the tandem stance task, the direction of orientation of the subject was rotated 90° to the right so that, again, each seesaw could be placed on separate force plates. Sufficient space (3 cm) was left between the seesaws to permit rolling movements. Every subject was requested to select a preferred anterior foot. Once selected, the anterior foot was kept constant during all tandem stance registrations.

With regard to each balance task, subjects were instructed to stand as still as possible with their hands clasped behind their back for a period of 22 seconds. During the tandem stance task, they were also requested to bear at least 35% of their body weight on the anterior foot, which was exemplified during the practice trials by means of auditory feedback. If a balance task was performed without a concurrent Stroop task (single-task condition), a visual reference was projected onto the screen in the form of a white cross on a dark background, yet no specific instructions were given with respect to visual attention.

**Stroop Tasks**

Three modified (shortened) versions of the Stroop test were used--the *word* card, the *colour* card, and the incongruous *colour-word* card--each consisting of 25 items randomly arranged in a 5 x 5 matrix of evenly spaced rows and columns. The word card was made of colour names printed in black, whereas the colour card was made of rectangular blocks printed in different inks. The colour-word card consisted of colour names that were incongruent with the printed ink colours, for instance the word "GREEN" printed in yellow. Except for the word card, the same four colours for words and inks were used: yellow, green, red, and blue. Every card was projected onto a white projection screen at eye level with a size of approximately 70 cm (width) x 35 cm (height). In this way, the projection cone formed a visual angle with the subject's head of about 26° in the horizontal plane and 14° in the vertical plane.

Subjects were instructed to read the colour names on the word card and to name the colours on the colour card, from the left to the right and from the top downwards. As for the colour-word card, the colours of the inks must be named while suppressing a strong tendency to read. With each task, subjects were requested to complete as many items as possible in a period of 22 seconds. After completion of the last (25th) item before the end of the registration, they had to start again with the first. If noticed by the subject, errors had to be immediately corrected.

**Procedure**

Each balance test was recorded both as a single task and in a dual-task condition with each Stroop test. In addition, each Stroop test was performed as a single task in a sitting position. In this way, four balance levels were obtained (sitting [ST], comfortable stance [CS], seesaw stance [SS], and tandem stance [TS]) as well as four Stroop levels (no card [NC], word card [WC], colour card [CC], and colour-word card [CW]). The 24 subjects were randomly assigned to the 24 possible sequences of conditions for either factor.

Before starting the measurements, subjects received verbal instructions about the different Stroop and balance tasks. Each of the three Stroop tests was then practised three times. Thereafter, the TS task was practised for a short period to select the anterior foot and in order to learn to bear sufficient body weight on the anterior foot (> 35%) by means of auditory feedback.

In addition, each single- and dual-task condition was practised once, directly before the performance of three consecutive trials that were used for definitive analysis. After each trial, a one-minute rest was permitted. In all (nine) dual-task conditions, the Stroop card was projected immediately at the start of the balance registration. In the (three) single-balance-task conditions, the visual reference was also available from the start of the registration. The total procedure lasted, including instructions and practise trials, approximately one-and-a-half hour for every subject.

**Data Analysis**

For every 22-second balance registration, also the first time derivative was calculated from the CP displacements in the fore-aft (FA) and lateral (LAT) directions separately. Then, the first 176 and last 120 of the 1320 samples were discarded from every data array leaving records of 1024 samples (≈ 17 s) for further analysis. In this way, undesired effects on balance at the start and at the end of each registration were excluded. All balance parameters were calculated for the FA and LAT sway components separately. With regard to the TS task, the FA and LAT axes were rotated 90° to the right according to the direction of orientation of the subject.

The root mean square was derived from both the CP displacements (RMS amplitude or Acp) and the CP velocities (RMS velocity or Vcp). Because the Vcp is strongly influenced by the higher frequency components of the CP fluctuations which are related to the accelerations of body mass, this parameter was primarily used to detect differences in postural control between conditions. Because the Acp is mainly influenced by the lower frequency large-amplitude components reflecting the

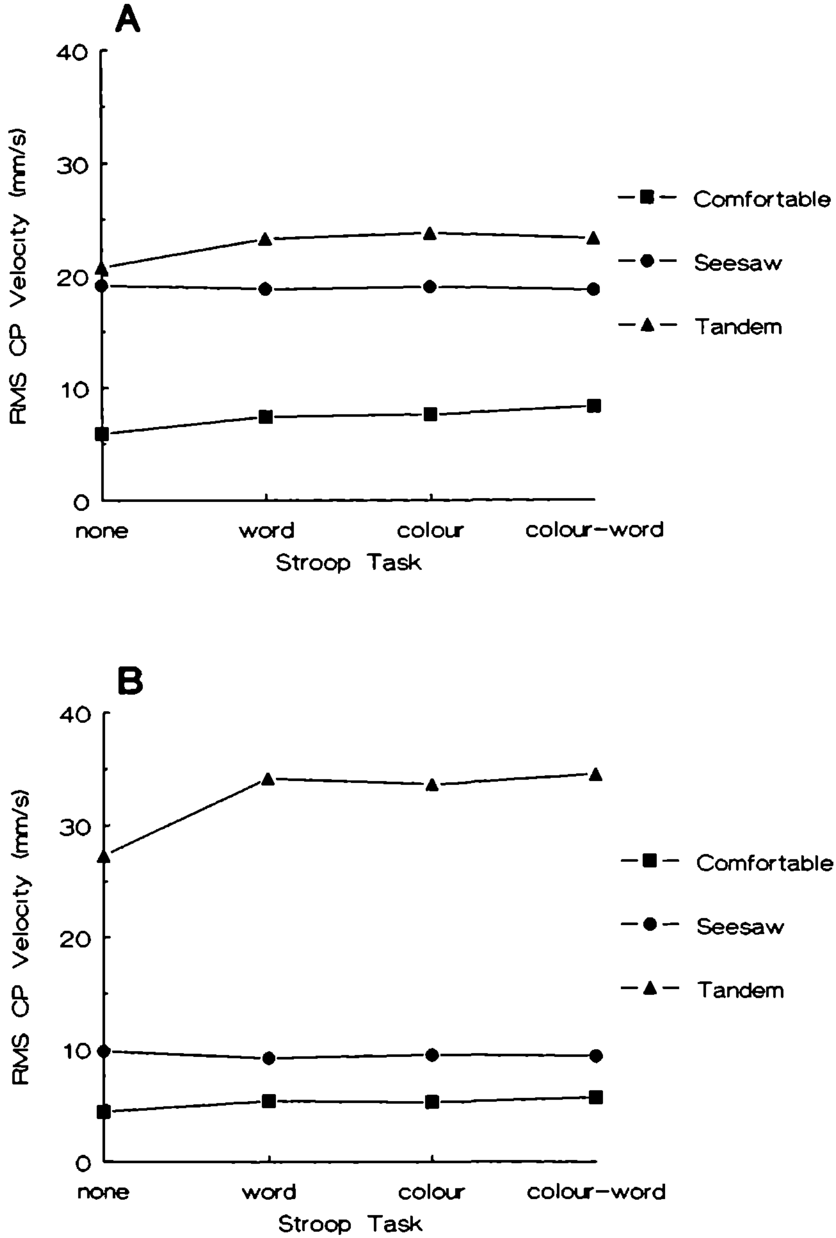



**Figure 2. The RMS CP velocity is presented for the three single- ("none") and nine dual-task conditions in the fore-aft (A) and lateral (B) directions separately.**

displacements of body mass, it was regarded as a measure of body sway. The number of items completed in 22 seconds on each of the Stroop tests served as the main measure of performance. Also, the number of uncorrected errors was noted.

## RESULTS

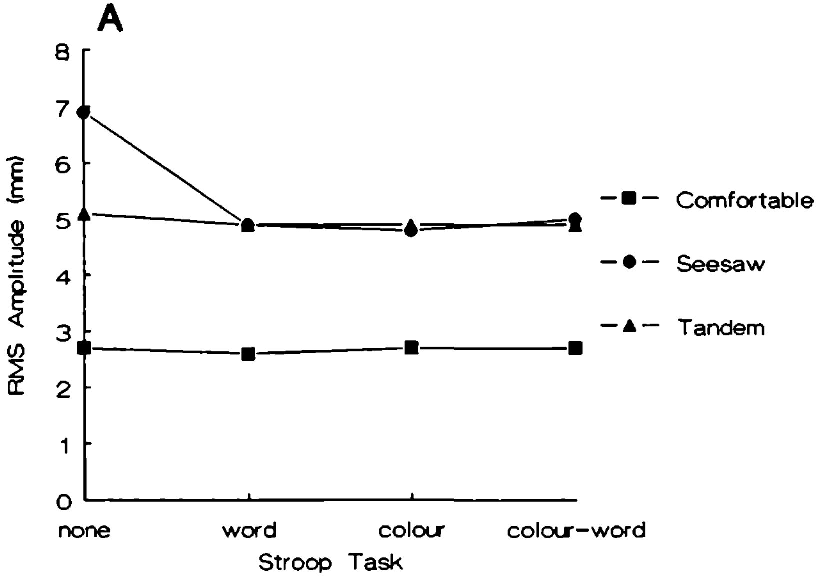
All individual balance and Stroop data from the three consecutive trials within each condition were averaged into single scores to reduce intrasubject variability. The balance parameters were tested in a Balance (CS, SS, TS) x Stroop (NC, WC, CC, CW) analysis of variance (ANOVA), whereas the Stroop data were tested in a Stroop (WC, CC, CW) x Balance (ST, CS, SS, TS) ANOVA with repeated measures on both factors ($\alpha < .05$).

### Balance Data

The group means of the Vcp values are presented in Figure 2 for both directions of sway. With respect to the FA sway, there was a large main effect of balance, $F(2,22) = 59.70$, $p < .001$. In particular, the Vcp values were much higher for the SS and TS tasks than for the CS task. There was no main effect of Stroop, nor a Balance x Stroop interaction (Figure 2A).

With regard to the LAT sway, there was again a main effect of balance on the Vcp, $F(2,22) = 123.35$, $p < .001$. According to expectation, the most important difference was now found between the TS task on the one hand and the CS and SS tasks on the other hand. Furthermore, there was a main effect of Stroop, $F(3,21) = 13.19$, $p < .001$, as well as a Balance x Stroop interaction, $F(6,18) = 9.15$, $p < .001$, indicating that the Stroop task effect differed for the various balance tasks. Figure 2B clearly demonstrates that LAT equilibrium control during *tandem* stance deteriorated as a result of simultaneous Stroop task performance. The data suggest, however, that the same amount of interference occurred irrespective of the type of Stroop task. This observation is statistically underscored by the finding that, after exclusion of the NC level from the analysis, the main effect of Stroop as well as the Balance x Stroop interaction were no longer significant. Separate paired *t* tests on the Vcp data reflecting LAT sway control during the TS task showed substantial differences between the NC task on the one hand and the WC, CC, and CW tasks on the other hand, $t(23) = 6.00, 3.43, 5.29$, $p < .005$, respectively. Similar comparisons among the WC, CC, and CW levels yielded no significant results.

The group means of the Acp values, which are presented in Figure 3, revealed a quite different perspective. With regard to LAT sway control (Figure 3B), the stable or perhaps slightly decreasing trends in the Acp for every balance task indicate that the increase in the LAT Vcp as a result of simultaneous Stroop task performance



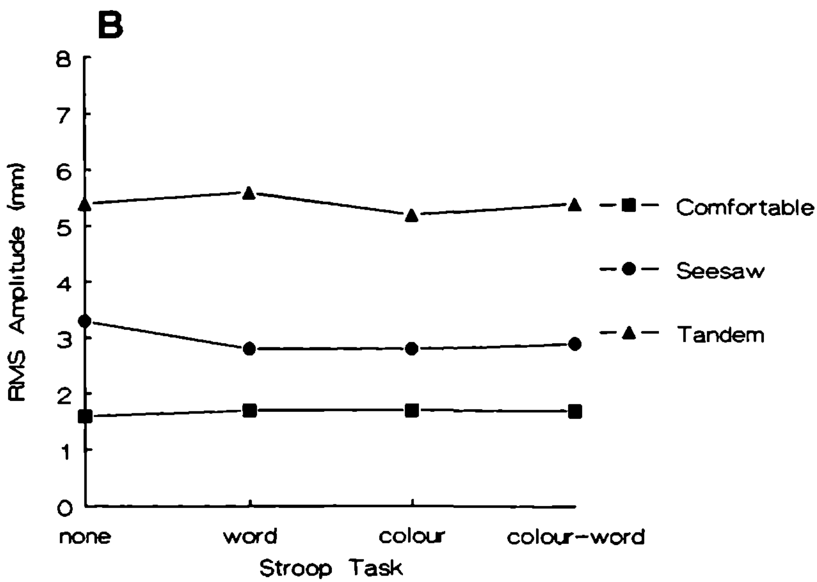




**Figure 3. The RMS CP amplitude is presented for the three single- ("none") and nine dual-task conditions in the fore-aft (A) and lateral (B) directions separately.**

during the TS task was almost exclusively determined by an increase in the mean frequency. Indeed, merely a main effect of balance was found, $F(2,22) = 181.26$, $p<.01$.

As for FA sway control, there was a main effect of balance on the Acp, $F(2,22) = 162.32$, $p<.01$. Besides, we found a main effect of Stroop, $F(3,21) = 5.58$, $p<.01$, as well as Balance x Stroop interaction, $F(6,18) = 3.96$, $p<.05$. Figure 3A clearly demonstrates that the latter two results were related to a stabilizing effect of simultaneous Stroop task performance on the FA sway during *seesaw* stance. Again, this effect occurred irrespective of the type of Stroop task. Discarding the NC level from the analysis resolved the main effect of Stroop as well as the Balance x Stroop interaction. Individual paired *t* tests on the Acp data reflecting FA sway control during the SS task showed substantial differences between the NC task on the one hand and the WC, CC, and CW tasks on the other hand, $t(23) = 5.31, 5.01, 4.69$, $p<.001$, respectively, whereas comparisons among the WC, CC, and CW levels were not significant.

**Stroop Data**

Because the mean number of uncorrected Stroop errors in all conditions never exceeded one, we only analysed the number of completed Stroop items. Figure 4 presents the group means of the number of completed items for the various Stroop tasks. There was a large main effect of Stroop, $F(2,22) = 355.56$, $p<.001$, with the slowest performance on the CW task and the fastest performance on the WC task. There was no main effect of balance. The statistics also produced a significant Stroop x Balance interaction, $F(6,18) = 4.13$, $p<.01$. Further analysis revealed that this interaction was related to a general improvement on the CC task during balancing compared to sitting. However, Figure 4 shows that the magnitude of this effect was very small.

## DISCUSSION

This study was conducted to investigate if and to what extent postural control in healthy adults is vulnerable to secondary (Stroop) task performance when subjects are confronted with different support-surface configurations. It was predicted that dual-task interference would occur when there would be a need to rely on poorly developed control strategies, but not when the execution of a postural task would still be adequately subserved by (a combination of) well-developed, prestructured synergies. The results of this study corroborate these global predictions based on the novelty of the postural task.

When subjects balanced on the *seesaws*, there was no detrimental influence

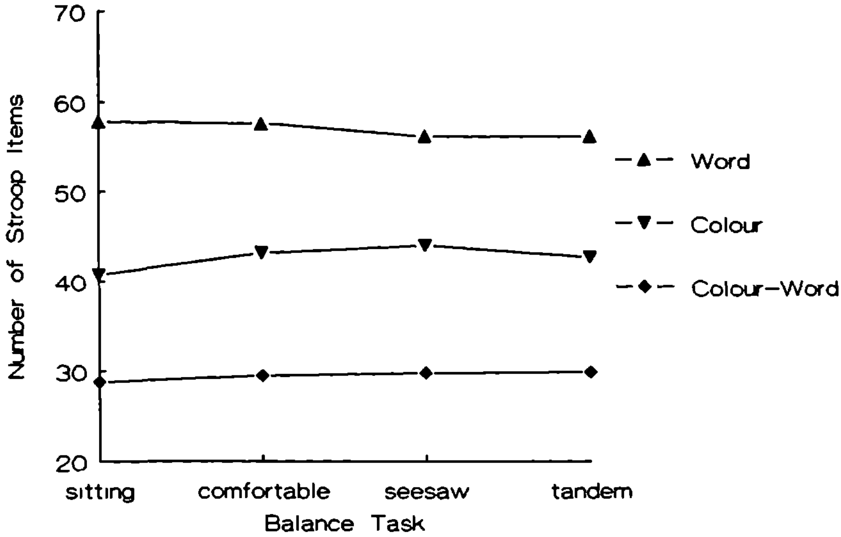



**Figure 4. The mean number of completed Stroop items is presented for the three single- ("sitting") and nine dual-task conditions.**

of simultaneous Stroop task performance on the control of *fore-aft* body sway as assessed by changes in the Vcp. There was, however, a marked amplitude-frequency trade-off as a result of the Stroop task performance leaving the Vcp unaltered. This phenomenon is accounted for in terms of an adaptation of timing and gain of normally employed postural synergies to achieve a more critical stabilization of posture in order to facilitate the uptake of visual information from the Stroop cards.

In contrast, when subjects balanced on the seesaws placed in a *tandem* position, there was an adverse influence of simultaneous Stroop task performance on *lateral* equilibrium control reflected by an increase in the Vcp (see Figure 2B). This finding corroborates our prediction, but contrasts with the results reported by Kerr et al. (1985), who found interference effects of maintaining balance in a tandem Romberg position (blindfolded) solely with the performance of a spatial memory task. Moreover, task concurrence costs were found only on the basis of a deteriorated performance of the memory task, while the balance performance remained constant. However, Kerr et al. (1985) merely used CP measures that are insensitive to changes of frequency (mean absolute distance and standard deviation from the mean position, in addition to the absolute total maximum deviation) whereas this study shows that the influence of secondary task performance on postural control during tandem stance is almost a pure *frequency* effect. This consideration also attenuates Kerr's conclusion that the observed dual-task interference was specifically based on competition within spatial cognitive processes. Our finding that the Acp remained unaffected can be regarded as a correct adaptation to the small medial-lateral size of the support base when standing with the feet in a tandem position. Indeed, the possibility to utilize vertical ground reaction forces to control LAT equilibrium is limited by the fact that ankle torques must be made effective through the short lever constituted by the foot in the medial-lateral direction. Hence, increasing the frequency of the CP fluctuations is the only effective mechanisms whenever LAT sway control becomes precarious.

Remarkably, we did not find any influence of the *type* of Stroop test on the degree of dual-task interference during the TS task, although we had predicted that the WC, CC and CW tasks would increasingly interfere with LAT sway control. This lack of discrimination between the different dual-task conditions related to tandem stance is attributed to the fact that the speed of the Stroop task performance (i.e. the number of completed items) was consistently lowest for the CW and highest for the WC task (see Figure 4). Differences in task complexity were neutralized by differences in the speed of performance, thus leaving the central-processing demands unaltered. As a consequence, the results give no unambiguous clue as to what neural mechanism are responsible for the task concurrence costs. Indeed, only a gradual increase in dual-task interference over the various (WC < CC < CW) Stroop tasks would have provided a strong indication of competition for use of a general-purpose central processor or, in other words, for use of nonspecific attentional resources.

The finding that there was never any influence of the type of Stroop test is suggestive of a possible role of *structural* interference mechanisms, i.e. interference due to competition either for use of the same input or output systems, or for use of common modality-specific information processing or storage systems (Schmidt, 1988). As for the combination of tasks used in this study, structural interference can only apply to the *visual* system which is involved in the control of posture on the one hand and in reading and colour perception on the other hand. The possibility of visual interference mechanisms as a (partial) explanation for the observed dual-task interference should be considered because, in comparison with normal upright standing, the importance of visual information in the control of posture is increased during tandem stance, especially with regard to LAT sway control (Amblard, Crémieux, Marchand, & Carblanc, 1985). In particular, the influence of eye movements on the visual stabilization of posture should be discussed, as well as the influence of concurrent visual information processing.

Firstly, it is a relevant question whether there is evidence for an effect of horizontal eye movements on (LAT) equilibrium control. Although some authors have reported a *stabilizing* effect of voluntary eye saccades on postural control (Kikukawa & Taguchi, 1985; Oblak, Gregoric, & Gyergyek, 1985), others merely emphasized the absence of a *destabilizing* effect as long as the frequency of the horizontal saccades is lower than 0.5 Hz and the amplitude smaller than 20° to 30° (Brandt, Paulus, & Straube, 1986). White, Post, and Leibowitz (1980) reported that, in contrast with externally induced retinal image motion, similar image motion due to voluntary saccadic eye movements does not easily affect postural control even while standing on one foot. They inferred from their results that efferent commands to move the eyes may play a critical role in modifying and suppressing sensory activity evoked by retinal image motion during voluntary eye movements. The data of the present study are in accordance with this conclusion because the WC task did not lead to more interference with tandem stance control than the CW task, although it generally coincided with higher velocities of horizontal eye movements.

A second important question is whether part of the observed dual-task interference in this study may have been caused by competition for use of the same channels for visual information processing. Such a type of interference is coherent with a multiple-resources model of attention assuming many different (limited-capacity) central operators, each related to a specific sensory or motor modality and processing stage (Wickens, 1989). As for the visual system, it is widely accepted that there are two subsystems, focal or object vision for processing of visual *form* vs. ambient or spatial vision for processing of visual *motion*, which are subserved by distinct cortical pathways (Leibowitz & Post, 1982; Mishkin, Ungerleider, & Macko, 1983). Even within the system for motion perception, there are specific visuo-cortical areas that respond to externally induced retinal image motion but not to self-induced visual motion during ocular tracking, thus providing visuo-spatial stability (Erickson & Thier, 1991). Whereas focal vision primarily involves the central parts of the retina, the ambient system is mediated by the entire visual field (Leibowitz & Post, 1982). Apparently, a high degree of modularity is characteristic for the various parts of the visual system. In view of this modularity, it is apparent that visual control of posture concerns ambient vision, whereas Stroop task performance depends on focal vision. As a consequence, structural (visual) interference is not likely to occur, even though postural control is not exclusively dominated by information provided by the peripheral parts of the visual field (Paulus, Straube, & Brandt, 1984; Stoffregen, 1985; Van Asten, Gielen, & Denier van der Gon, 1988). This conclusion is supported by the observation that ambient functions basically operate in the absence of awareness without any attentional effort (Gielen & Van Asten, 1990; Stoffregen, 1985), whereas focal functions are typically attention-driven. The fact that our data concerning the postural organization in HMSN patients always showed a dissociation between the effect of visual deprivation and the effect of simultaneous Stroop task performance (Geurts, Mulder, Nienhuis, & Rijken, 1992; Geurts, Mulder, Nienhuis, Mars, & Rijken, 1992) further underlines the notion that visual factors play a minor role in the observed dual-task interference.

To summarize, it is inferred from the present results on the one hand and from the literature on the other hand that the observed dual-task interference in this study is probably best accounted for by competition for use of a general-purpose central processor, which is somehow limited in capacity (Schmidt, 1988) This competition is attributed to a lack of automaticity of LAT sway control when the feet are placed in a tandem position due to the relative impotence of centrally stored postural strategies (see for a review of "automaticity" as a measure of motor control Neumann, 1984). It is improbable that other, nonspecific effects such as an enhanced arousal or an increased respiration rate during the Stroop task performance contributed to the dual-task interference because this would not readily explain why the dual-task interference specifically concerned lateral sway control during tandem stance.

This study indicates that merely reducing the efficacy of ankle mechanisms working in the sagittal plane does not lead to a loss of balance automaticity in healthy adults, probably because postural control can be dealt with by a change in parametrization within well-developed ankle synergies. In contrast, previous observations in HMSN patients indicate a temporary loss of balance automaticity when they are provided with new orthopedic footwear which also leads to a reduced efficacy of ankle mechanisms in the sagittal plane (Geurts, Mulder, Nienhuis, & Rijken, 1992). This discrepancy could imply that, besides the *novelty* of the task, the *integrity* of the nervous system further determines the attention demands in postural control. Indeed, Stelmach et al (1990) have demonstrated an increased vulnerability of postural stabilization to concurrent arithmetical task performance in elderly subjects after self-generated arm swings. They speculated that, due to a decrease in the availability of proprioceptive information, elderly subjects must allocate more attention to the control of posture to derive the necessary sensory information. Hence, we not only conclude that the automaticity of the human posture-control system may decrease as a function of the novelty of the balance task, but also that its vulnerability to secondary task performance during poorly trained balance tasks may increase as the result of an impaired nervous system

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# CHAPTER 9

# ATTENTION DEMANDS IN BALANCE RECOVERY FOLLOWING LOWER LIMB AMPUTATION

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**ABSTRACT**

The attention demands in the control of posture after structural damage to the peripheral sensorimotor system were studied in 12 persons with a recent lower limb amputation. The interference of an arithmetic task with two postural tasks of different complexity (quiet standing and weight shifting) was examined while the subjects stood on a force platform at several times during their rehabilitation. Control data were obtained from healthy subjects. For both postural tasks, persons with amputation performed worse than controls. Quiet standing as a relatively simple balance task revealed clear dual-task interference only in the amputation group. Evidence was found for a reduction in dual-task interference across rehabilitation. In contrast, voluntary (feedback-controlled) weight shifting as a complex balance task revealed an equal amount of dual-task interference in persons with amputation and in control subjects without clear changes in interference across time. The results indicate that attention-demanding processes may be involved in postural control depending on the novelty and complexity of the task. Moreover, a reduction in central-processing demands for quiet standing may reflect a central adaptation of the postural organization to the peripheral sensorimotor impairments due to lower limb amputation.

## INTRODUCTION

Vulnerability to secondary task performance is a striking feature of any poorly developed motor act, whereas the increase in the ability to perform concurrent tasks is one of the essential aspects of motor learning. The theoretical explanations for dual-task interference, however, show great diversity with respect to the mechanisms of task concurrence costs (Wickens, 1989). This study is focused on dual-task interference as a result of competition for a "general-purpose limited-capacity central processor" being the most frequently used operational definition of attention (Brown & Carr, 1989; Schmidt, 1988). Many motor control theories assume a transition from attention-invested to attention-free performance during the acquisition of a novel skill (Adams, 1971, 1981; Fitts & Posner, 1967; Neumann, 1984; Norman & Shallice, 1986; Schmidt, 1975). From this perspective, the reduction in interference of a simultaneous attention-demanding (cognitive) task with a primary motor task over practice may reflect the level of motor skill acquisition, in particular the degree of automaticity (Neumann, 1984; Brown & Carr, 1989).

The role of attention in the control of action has not been extensively investigated (Norman & Shallice, 1986). Attention demands have been studied even less in motor *recovery*, e.g. after structural damage to the peripheral sensorimotor system. Instead, in many clinical experiments basic motor skills such as standing and walking are assessed under single-task and optimal sensory conditions, thus, discarding the role of higher order central processes from the scope of investigation (Mulder & Geurts, 1991; Patla, 1991). However, major structural damage to the neuromuscular and skeletal systems causes a disruption of the premorbidly employed muscular synergies and sensorimotor strategies and confronts the central nervous system (CNS) with altered mechanical constraints and distorted sensory feedback. This loss of coordinative patterns inevitably requires a central adaptation of the sensorimotor organization in order to restore functional abilities. Consequently, an important aspect of motor recovery after peripheral lesions may be reflected in a changing interaction between lower level (automatic) control mechanisms and higher order (attention-demanding) central processes.

From a cognitive or computational approach to skill acquisition (see for a review Masson, 1990), it is often presumed that novel sensorimotor strategies develop on the basis of an adapted central representation of knowledge that captures invariant components between input and output patterns. That is, new linkages between sensory input and motor output are formed by continuous repetition of an act. As a result, the need for attention-demanding processes concerned with error detection and correction during the motor-task performance becomes less, leading to a reduction in task concurrent costs.

The primary aim of this study is to gain more insight into a theoretically important aspect of *balance* recovery after peripheral sensorimotor damage, viz. the changing involvement of attention-demanding central processes in balance performance across time. We have focused on postural control, because the ability to maintain an erect posture in a fast and automatic fashion is a prerequisite for the unrestrained performance of many daily activities. Normally, body perturbations during upright standing are counterbalanced by fixed patterns of rapid postural responses requiring little if any attentional involvement (Nashner, 1977; Nashner, Woollacott, & Tuma, 1979). Self-initiated rapid arm movements are automatically preceded by anticipatory postural adjustments (Belen'kii, Gurfinkel, & Pal'tsev, 1967; Lee, 1980). Even under changing support-surface conditions postural actions remain fast and largely automatic because they appear to be selected from a limited repertoire of preprogrammed strategies (Horak & Nashner, 1986; Massion, 1992).

Nonetheless, interaction of the regulation of posture with cognitive processing has been indicated by the detrimental influence of a difficult standing task (tandem Romberg position) on a secondary spatial memory task in young adult subjects (Kerr, Condon, & McDonald, 1985). Stelmach, Meeuwsen, and Zelaznik (1990) reported an adverse effect of a secondary arithmetic task on the postural restabilization time after destabilizing activity in elderly persons. These results suggest that cognitive involvement in the control of posture is strongly dependent on the novelty and complexity of the balance task. Indeed, Nashner and Cordo (1981) have argued that (automatic) postural stabilization in healthy subjects is only relatively independent of voluntary (attention-driven) control of movement, each function representing a different organizational level in a hierarchical structure.

We started from the assumption that the relative autonomy of the postural control system is considerably reduced in peripheral disorders through which essential neuromuscular and skeletal structures have been damaged. From the many possible pathological conditions, we have studied balance recovery after lower limb amputation because of the irreversibility and the combined efferent-afferent nature of the disorder, whereas the CNS remains unaffected.

Persons with a unilateral lower limb amputation lack a physiological ankle joint and its adjacent muscular structures to actively apply ankle torque at one side of the body (Murdoch, 1969). If the amputation is through or above knee level, a prosthetic knee joint will not allow active control of knee flexion such as during vertical movements of the body center of gravity (Nashner & McCollum, 1985). In addition, the somatosensory input from the amputated side is always severely distorted (Fernie & Holliday, 1978). Consequently, postural strategies which are mediated by somatosensory input or which are executed by lower limb muscles are considerably impaired. There is also a severely reduced ability to transfer body weight onto the prosthetic limb which is at least partly due to inadequate sensory information from the stump tissues (Gauthier-Gagnon, St-Pierre, Drouin, & Riley, 1986).

Based on the abovementioned considerations, a radical sensorimotor reorganization must take place in subjects who recently underwent a lower limb amputation. In this perspective, even the learning of a relatively simple act such as standing with a prosthesis should be regarded as the acquisition of a skill. Hence, a clear lack of balance automaticity was predicted during the early phases of skill acquisition. To study the attention demands in the relearning of balance control after a lower limb amputation, we examined the interference of an arithmetic task with two postural tasks of different complexity at several times during the rehabilitation process.

Quiet upright standing was considered to be a relatively simple task of maintaining a vertical posture, for which an almost complete recovery to a normal level of automaticity was expected. Conversely, weight shifting on the basis of artificial visual feedback was tested as a complex task of maintaining postural stability while making voluntary (destabilizing) body-mass displacements. It was expected that automatized performance of this task would not occur. By minimizing the similarity between the cognitive task on the one hand and the postural tasks on the other hand, structural interference due to common receptor or effector systems or to common (modality-related) processing channels was prevented. As a result, possible dual-task interference became attributable to simultaneous capacity demands on a higher level general-purpose central processor.

## METHODS

### Subjects

Twelve persons who had recently underwent a unilateral limb amputation above the ankle and below the hip joint and who were fit for the first time with a lower limb prosthesis participated in the study. Subjects suffering from serious cognitive (e.g. disorders of memory or attention) or sensory (e.g. cataract, visual field loss, or deafness) dysfunctions were excluded, as well as those suffering from persistent pain problems or dystrophy. The amputation group was aged 23 to 78 years (mean age 59.4 ± 18.3) and consisted of nine males and three females. They had either a below- ($n$ = 4), a through- ($n$ = 5) or an above-knee ($n$ = 3) amputation. Nine patients had a vascular cause of amputation, two of them suffered from diabetes mellitus. The other three patients had a nonvascular cause of amputation. In addition, a control group of 12 healthy subjects who had been matched for age (mean age 58.9 ± 18.3) and gender participated in the study, as well as a nonmatched control group of four females and four males, aged 22 to 30 years (mean age 24.9 ± 2.4).

**Apparatus**

Balance measurements were made with a force platform consisting of two separate aluminum plates, each placed on three force transducers (hysteresis and nonlinearity < 1%) recording the vertical ground reaction forces. Signals were processed by six DC-amplifiers (nonlinearity < 0.1%) and first-order low-pass filters (cut-off frequency 30 Hz); then they were stored into a microprocessor (COMPAQ 386SX, 16 MHz) after a 12-bit AD-conversion at a sampling rate of 60 Hz. By means of digital moment-of-force calculations, the point of application of the resultant of the ground reaction forces in a two-dimensional transverse plane was determined for each sample, with a maximum error of ± 1 mm in both directions of sway. The coordinates of this "center-of-pressure" (CP) were passed through a digital low-pass 6-Hz (Fourier) filter to eliminate high-frequency components due to noise or tremor.

An additional color monitor (diameter 34 cm) was connected to the computer and was placed a little below eye level on a height-adjustable table one meter in front of the standing subject. A white screen (300 cm wide x 260 cm high) that was positioned at a distance of two meters in front of the platform covered the greater part of the visual field. A portable tape recorder with a headphone was used to present subjects with different sets of arithmetic problems.

**Balance Tasks**

During all balance tests, subjects stood on the force platform with the medial sides of their heels 8.4 cm apart and with each foot toeing-out at a 9° angle from the sagittal midline. Before the first assessment, the *length* of the support surface in the sagittal plane was determined with a measuring tape, as well as the distance between the anterior borders of the right and left tibia (*width* of support). These measures (in centimeters) were then fed into the computer.

Every test series incorporated four balance tests in a fixed sequence (a) quiet standing as a single task, (b) quiet standing while performing an arithmetic task, (c) weight shifting as a single task, and (d) weight shifting while performing an arithmetic task. In both the static and dynamic dual-task conditions, the concurrent arithmetic task was started simultaneously with the balance task. For reasons of reliability, the easier condition always preceded the more difficult condition to minimize insecurity, thus, ensuring optimal patient compliance. In every condition, the CP fluctuations were recorded for 30 seconds. Each 30-second registration was preceded by an anticipatory period of five seconds, which were visualized on the computer monitor by presenting the figures "1" to "5" on a horizontal line.

During quiet standing ("static" task), subjects were instructed to stand as still and symmetrically as possible with their hands folded behind their back. No specific instructions were given concerning visual attention. After the anticipatory period, the color monitor turned into uniform gray until the end of the registration.

During weight shifting ("dynamic" task), the CP position was displayed on the computer monitor as a small white cursor moving on a gray background. In addition, two stationary goals were presented at either side of the virtual vertical through the middle of the screen which corresponded to the sagittal midline of the body. Each goal was a square of four blue lines. The length of these lines as well as the lateral deviation of the centers of the goals from the sagittal midline was individually determined at 15% of the *width* of support. In this way, weight bearing of approximately 65% on each limb was required to bring the cursor in the middle of the corresponding goal. In the fore-aft direction, the centers of the goals were given fixed positions at a distance of 40% of the *length* of support from the rear.

With all monitor displays, 1 mm on the screen was adjusted to 1 mm in the real world in both directions. Thus, a real-time, real-size visual feedback was provided in which up and down movements of the cursor corresponded to forward and backward CP displacements, respectively. Left and right movements corresponded to the actual CP displacements in the lateral direction. The target in which direction body weight had to be transferred was indicated by filling up one of the goals with the color yellow. In order to make a correct weight shift, subjects had to maintain their CP during at least one second within the target. As soon as an effective weight shift was made toward one side, the contralateral goal became the target. The first target was randomly assigned by the computer. Subjects were instructed to start from a comfortable position and to make as many alternating weight shifts as possible in 30 seconds, as fluently as possible. Because no fixed rhythm was imposed, each person determined his own speed and accuracy, which ensured a low threshold of performance.

**Arithmetic Task**

The concurrent activity was a simple arithmetic task in which subjects were verbally presented with a set of eight single-digit additions equally timed over a 30-second period. The additions were selected at random and followed by a sum that could be either correct or incorrect, e.g. "2 + 3 = 5," "7 + 4 = 13." The subjects were instructed to verbally indicate the correctness of each addition problem (response time ≈ 2 s). The number of incorrect decisions was noted. A similar task has been employed by Stelmach et al. (1990) requiring subjects to mentally count the number of correct additions. We have required immediate "good" or "fault" decisions to eliminate the possibility to guess the number of correct additions at the end of the task, as well as an influence of individual differences in short-term memory.

**Procedure**

First, the arithmetic task was practiced three times in a sitting position to familiarize the subjects with the task. Then, the dynamic balance task (without concurrent activity) was practiced for several minutes until the subject showed an optimal understanding and performance of the task. Every assessment consisted of two consecutive (identical) test series, starting with the performance of the arithmetic task in a sitting position (single task performance to be used for comparison with the performance during the balance tasks), and followed by the four balance tests. A one minute's rest was given after each test, whereas a longer pause was allowed between the two test series.

The amputation group was assessed five times at two-weekly intervals, starting one or two days after the first training with their prosthesis (week 0) and ending after a rehabilitation period of eight weeks (week 8). During this period, the amputation group received a regular training program for two hours daily following a gradual transition from erect standing, weight shifting and stepping between parallel bars, to walking with and without aids. The patients were always tested wearing the type of footwear to which the prosthetic components and alignment had been adjusted.

Similar to the amputation group, the nonmatched control (NMC) group was assessed five times at two-weekly intervals to detect possible learning or time effects unrelated to rehabilitation. For practical reasons, the (elderly) matched control (MC) group was tested just once to obtain age-corrected reference values for balance automaticity. All control subjects were tested with bare feet.

**Data Analysis**

From a 30-second static balance registration, the root mean square (RMS) amplitude of the CP displacements (Acp) and (after a first-order differentiation) the RMS of the CP velocities (Vcp) were derived in the fore-aft (FA) and lateral (LAT) directions separately. Because of its frequency-dependency, the Vcp was primarily used to detect differences in balance control. Indeed, specifically the higher frequency components in the CP fluctuations reflect the compensatory torques which are applied at the support surface by muscular activity to counterbalance the body accelerations caused by gravity (Hufschmidt, Dichgans, Mauritz, & Hufschmidt, 1980; Murray, Seireg, & Scholz, 1967; Shimba, 1984). In contrast, the Acp is often determined by the lower frequency components in the CP fluctuations (well below 1.0 Hz) with a relatively large amplitude and is, therefore, believed to be related to the actual body sway (Hufschmidt et al., 1980).

As for a dynamic balance registration, the number of correct weight shifts represented the speed of performance. As a measure of the fluency of weight shifting, the length of the optimal CP pathway in the LAT direction (the distance between the centers of the two goals) was subtracted from the length of the actual LAT CP pathway per weight shift to determine the average length of the *surplus* CP pathway in this direction (SPcp) according to the following equation:

$$SPcp = \frac{\sum_{i=a}^{b-1} |x_{i+1} - x_i|}{n-1} - d \qquad (1)$$

a = sample of first correct weight shift
b = sample of last correct weight shift
x = LAT CP coordinate
d = distance between the centers of the goals
n = number of correct weight shifts

The CP trajectory in the FA direction was not incorporated in this analysis, because the exact relationship between FA and LAT sway control is unknown, in particular with respect to weight shifting.

## RESULTS

For each balance assessment, the comparable parameters derived from the two test series were averaged into a single score. Because balance performance in the (elderly) MC group was generally worse than in the (younger) NMC group, we did two separate statistical analyses with respect to either control group. To compare the MC data with the first assessment of the amputation group, each dependent variable was entered into a Group (amputation vs. MC) x Condition (single vs. dual) repeated measures analysis of variance. To compare the NMC data with the amputation data, a three-way analysis of variance of Group (amputation vs. NMC) x Condition (single vs. dual) x Time (week 0, 2, 4, 6, 8) with repeated measures on the last two factors was used for each of the dependent variables. It must be mentioned beforehand that all persons with an amputation reached an acceptable level of standing and walking ability at week 8 during their rehabilitation, which indicated a substantial improvement of overall balance skills for all subjects.

### Static Balance Tests

At week 0, the amputation group generally showed greater Vcp values than the MC group for both the FA, $F(1,11) = 19.27$, $p<.005$, and LAT sway, $F(1,11) = 16.16$, $p<.005$. Moreover, a Group x Condition interaction in the FA, $F(1,11) = 9.61$, $p<.05$, and LAT directions, $F(1,11) = 5.25$, $p<.05$, indicated a significant amount of dual-task interference in the amputation group. As for the Acp values, there was merely a main effect of group in both directions (FA: $F(1,11) = 7.40$, $p<.05$; LAT: $F(1,11) = 21.54$, $p<.005$), reflecting more sway in the amputation group. The number of arithmetic errors did not produce significant results. Table 1 summarizes the data of the amputation group at week 0 and those of the MC group.

**Table 1: Balance Parameters and Arithmetic Errors in Different Groups and Conditions: Static Balance Tests.**

| Condition | | Group | |
|---|---|---|---|
| | | Amputation[a] | Control[b] |
| | | *Fore-aft sway* | |
| Single | | | |
| | Vcp[c] | 18.0 ± 9.4 | 7.5 ± 2.6 |
| | Acp[d] | 4.2 ± 1.7 | 3.2 ± 0.7 |
| Dual | | | |
| | Vcp[c] | 22.9 ± 11.2 | 8.1 ± 2.9 |
| | Acp[d] | 5.3 ± 2.7 | 3.6 ± 1.3 |
| | | *Lateral sway* | |
| Single | | | |
| | Vcp[c] | 11.9 ± 7.2 | 5.0 ± 2.2 |
| | Acp[d] | 4.0 ± 1.8 | 2.3 ± 0.7 |
| Dual | | | |
| | Vcp[c] | 14.4 ± 7.9 | 5.0 ± 1.5 |
| | Acp[d] | 4.2 ± 2.0 | 1.8 ± 0.5 |
| | | *Number of arithmetic errors* | |
| Sitting | | 0.5 ± 0.9 | 0.2 ± 0.4 |
| Balancing | | 0.9 ± 1.4 | 0.3 ± 0.4 |

[a]Data of the amputation group at week 0 ($n = 12$). [b]Data of the matched control group ($n = 12$). [c]RMS CP velocity (mm/s). [d]RMS CP amplitude (mm).

In a three-way analysis of the Vcp data, the amputation group also showed poorer balance control than the NMC group for both the FA, $F(1,18) = 12.06$, $p<.005$, and LAT sway, $F(1,18) = 9.15$, $p<.01$. Again, a Group x Condition interaction in the FA, $F(1,18) = 4.71$, $p<.05$, and LAT directions, $F(1,18) = 4.46$, $p<.05$, revealed clear dual-task interference in the amputation group. There was, however, no main or interaction effect involving time. For both groups, the CP velocities in the FA and LAT directions are illustrated in Figure 1 and 2, respectively.

Because the NMC group was substantially smaller than the amputation group, a subsequent two-way (Condition x Time) analysis was done for each of the groups separately. Whereas the NMC data did not produce any main or interaction effect, the amputation group showed a Condition x Time interaction for the FA sway, $F(4,15) = 3.74$, $p<.05$, as well as main effects of time (FA: $F(4,15) = 3.60$, $p<.05$; LAT: $F(4,15) = 3.95$, $p<.05$), and condition (FA: $F(1,18) = 10.48$, $p<.01$; LAT: $F(1,18) = 7.99$, $p<.05$). Postural control in the amputation group deteriorated in the dual-task condition, whereas dual-task interference became less over practice, at least for the FA sway (see Figure 1). Paired $t$ tests of the single- versus dual-task performance at each of the rehabilitation times revealed a significant amount of dual-task interference for the FA sway only at week 0, $t(11) = -3.26$, $p<.01$.

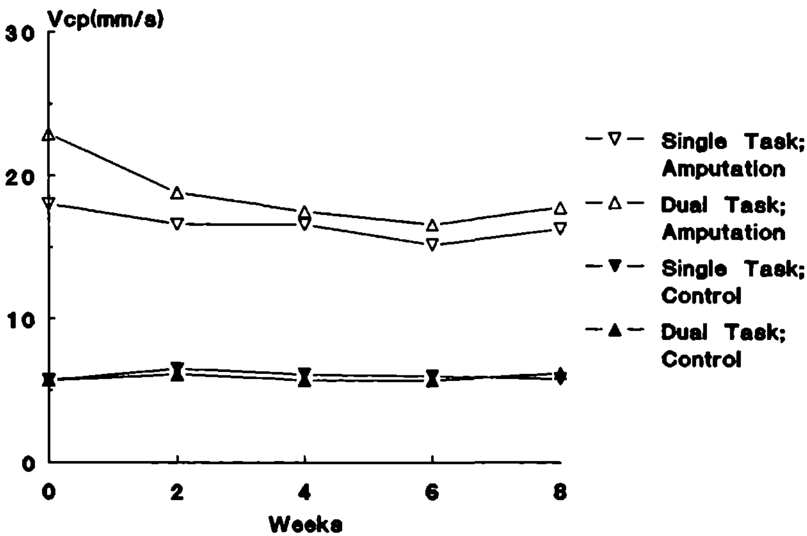


**Figure 1. RMS CP velocity (Vcp) (mm/s) in the FA direction as a function of time (weeks) and condition (single and dual task) for the amputation group ($n$ = 12) and nonmatched control group ($n$ = 8).**

A three-way analysis of the Acp data merely yielded a main effect of group with greater values in the amputation group than in the NMC group for both the FA, $F(1,18) = 4.33$, $p<.06$, and LAT sway, $F(1,18) = 13.68$, $p<.005$. As for the number of arithmetic errors, there was a Group x Condition interaction, $F(1,18) = 5.02$, $p<.05$, as well as a main effect of group, $F(1,18) = 5.49$, $p<.05$, indicating more errors during stance in the amputation group than in the NMC group.

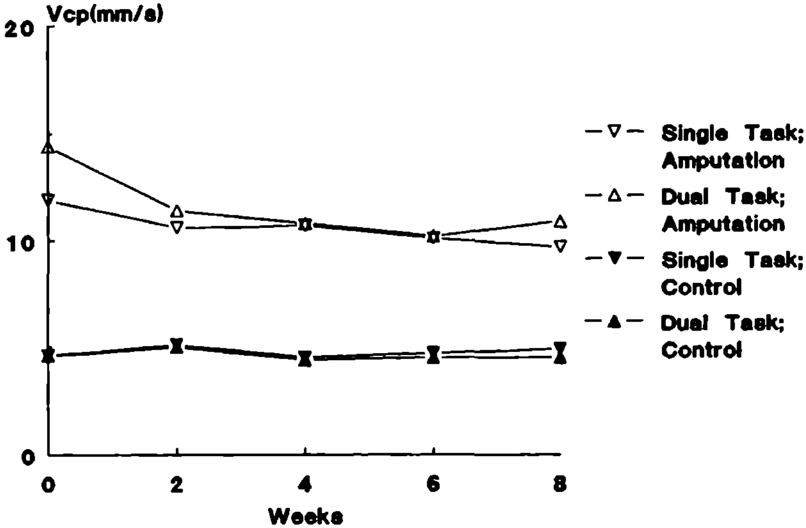


**Figure 2. RMS CP velocity (Vcp) in the LAT direction as a function of time (weeks) and condition (single and dual task) for the amputation group ($n$ = 12) and nonmatched control group ($n$ = 8).**

**Dynamic Balance Tests**

At week 0, eight patients failed to make minimally three weight shifts in at least one single-task registration, whereas at week 2 only two patients failed to meet this criterion. These two patients were excluded from further analysis, as well as all assessments made at week 0, to prevent that floor effects would obscure dual-task interference. Thus, a reduced data structure of ten patients and four assessments was formed to analyze the speed of weight shifting. The SPcp values were calculated only for those assessments within the abovementioned reduced data structure ($n$ =

10, weeks 2 to 8) with three or more weight shifts in each single- and dual-task registration. This additional reduction was necessary to calculate the SPcp for each registration with a denominator greater than one (Equation 1). Therefore, only seven patients were included in the analysis of the fluency of weight shifting. With the exclusion of amputation data, the corresponding MC data were also excluded from the statistical tests.

At week 2, the amputation group generally made less weight shifts than the MC group, $F(1,9) = 22.02, p < .005$, whereas the number of weight shifts was lower in the dual-task condition, $F(1,9) = 5.54, p < .05$. A nonsignificant Group x Condition interaction indicated that the amount of dual-task interference was equivalent for both groups. As for the fluency of weight shifting, the amputation group merely showed a tendency toward higher SPcp values compared to the MC group, $F(1,6) = 5.24, p < .07$. The number of arithmetic errors did not produce significant results. Table 2 presents the data of the amputation group at week 2 as well as the corresponding data of the MC group.

**Table 2: Balance Parameters and Arithmetic Errors in Different Groups and Conditions: Dynamic Balance Tests.**

| | | Group | |
|---|---|---|---|
| Condition | *n* | Amputation[a] | Control[b] |
| | | *Number of weight shifts*[c] | |
| Single | 10 | 7.6 ± 4.2 | 12.2 ± 2.7 |
| Dual | 10 | 6.8 ± 5.2 | 11.2 ± 2.9 |
| | | *Surplus CP pathway per weight shift (mm)*[d] | |
| Single | 7 | 77.8 ± 26.6 | 43.8 ± 24.1 |
| Dual | 7 | 90.9 ± 68.6 | 34.7 ± 12.9 |
| | | *Number of arithmetic errors*[c] | |
| Sitting | 10 | 0.4 ± 0.5 | 0.2 ± 0.3 |
| Balancing | 10 | 0.8 ± 1.0 | 0.3 ± 0.5 |

[a]Data of the amputation group at week 2. [b]Data of the matched control group. [c]The numbers of weight shifts and arithmetic errors are given for those patients (and corresponding controls) with minimally three weight shifts in one of the single-task registrations at weeks 2-8. [d]The average length of the *surplus* CP pathway per weight shift in the LAT direction is given for those patients (and corresponding controls) with minimally three weight shifts in each single- and dual-task registration at weeks 2-8.

A three-way analysis of the speed of weight shifting again revealed less weight shifts in the amputation group than in the NMC group, $F(1,16) = 21.22$, $p<.001$. Main effects of condition, $F(1,16) = 27.42$, $p<.001$, and time, $F(3,14) = 15.30$, $p<.001$, indicated that the number of weight shifts was lower in the dual-task condition and increased across time. A marginally significant Group x Time interaction, $F(3,14) = 2.89$, $p<.08$, suggested a more substantial learning effect in the amputation group compared to the NMC group (see Figure 3). All interaction effects involving condition were not significant. Apparently, dual-task interference remained constant over time irrespective of the group.

A three-way analysis of the SPcp data revealed less fluency of weight shifting in the amputation group than in the NMC group, $F(1,13) = 7.80$, $p<.05$. Although the mean SPcp values of the amputation group showed a clear tendency toward more fluency with practice (see Figure 4), the main effect of time was not significant. The main effect of condition was also nonsignificant, as were all two- and three-way interactions. The number of arithmetic errors merely showed a main effect of group, $F(1,16) = 5.99$, $p<.05$, with generally more errors in the amputation group than in the NMC group.

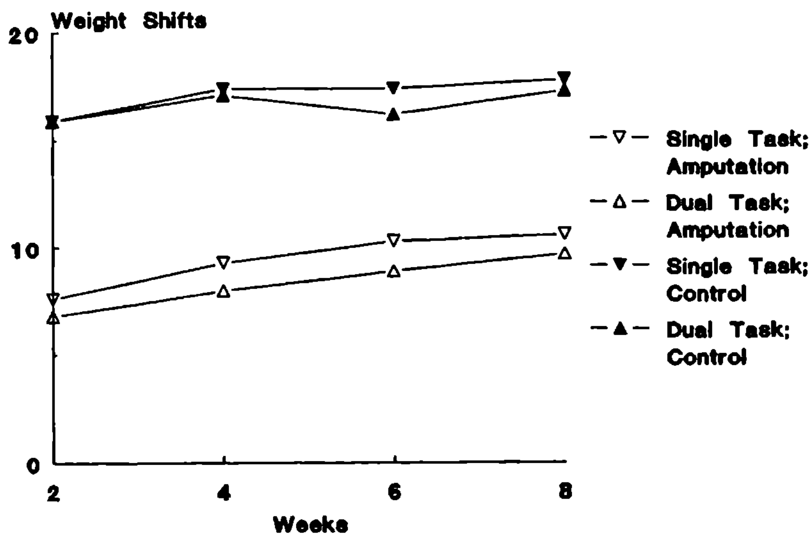


Figure 3. Number of weight shifts as a function of time (weeks) and condition (single and dual task) for the amputation group ($n$ = 10) and nonmatched control group ($n$ = 8).

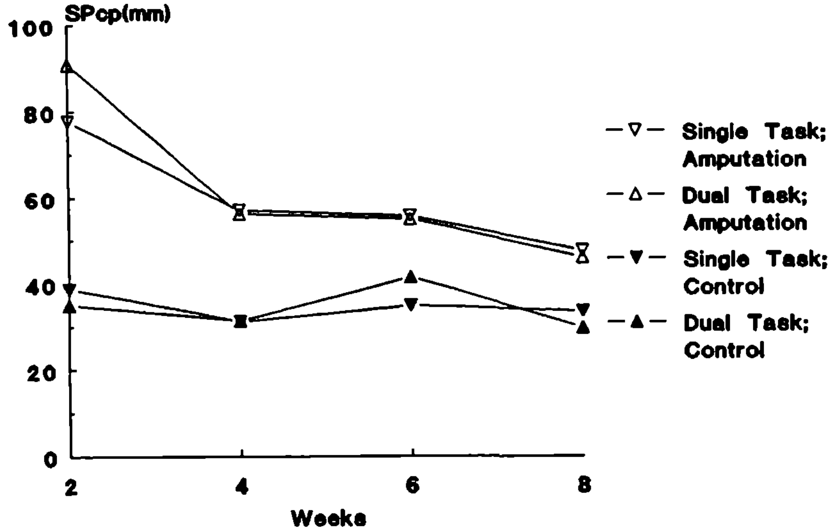


**Figure 4.** Average length of the *surplus* CP pathway per weight shift (SPcp) in the LAT direction as a function of time (weeks) and condition (single and dual task) for the amputation group ($n$ = 7) and nonmatched control group ($n$ = 8).

## DISCUSSION

The primary aim of this study was to gain insight into the role of attention-demanding processes in the restoration of postural control after structural damage to the peripheral sensorimotor system, in particular following a recent lower limb amputation.

Both the Vcp and Acp data during the static tests indicate that, compared to healthy controls, persons with an amputation suffer from a basically reduced control of posture in both directions of sway. This lack of balance control can be attributed to the irreversible loss of input and output structures at the side of amputation. The Vcp data further demonstrate that the vulnerability of balance control to the simultaneous performance of a simple arithmetic task is clearly enhanced, particularly early in rehabilitation. The nonsignificant increase in Acp as a result of concurrent task performance indicates that the dual-task effect is at least partly due to a shift of the mean CP frequency toward higher values. This result suggests reduced balance control rather than increased body sway in the dual-task condition.

The results of the dynamic balance tests show that persons with an amputation have a lower weight-shifting speed and tend toward greater disfluency than healthy control subjects indicating a grossly diminished ability to voluntarily displace body mass. The performance of the amputation group, however, may have been disproportionally affected by a possible reduction in the visual stabilization of posture as a result of the visual monitoring of the feedback signal. Indeed, such structural interference within the dynamic task may have been relatively important for the persons with an amputation, because they strongly depend on vision due to a loss of kinaesthetic information. In all groups, the number of weight shifts was significantly reduced during the simultaneous performance of an arithmetic task, whereas the SPcp as a fluency measure was not affected. Apparently, most subjects adopted a slower weight-shifting strategy during simultaneous attention demands while they maintained their accuracy of performance.

We believe that the observed dual-task interference during both postural tasks was free from competition with the concurrent arithmetic task either for use of the same input or output systems, or for use of common modality-specific information processing or storage systems. The balance tasks, on the one hand, require visual, vestibular and somatosensory inputs, spatial processing, and motor activity in postural muscles. The arithmetic task, on the other hand, demands auditory input, nonspatial processing, and oral motor output. Dual-task interference due to outcome conflict or confusion (Navon & Miller, 1987) can also safely be excluded, whereas a possible influence of the arithmetic task output (eight verbal responses in 30 s) on the rate of respiration (and indirectly on postural control) remains within a normal range (12-16 respirations per minute). Lastly, short-term learning effects due to the fixed-sequential performance of the balance tests may have decreased rather than increased the degree of dual-task interference.

Hence, dual-task interference in this study is attributed to competition for a general-purpose central processor which is somehow limited in capacity. Because an adequate performance of the arithmetic task requires a more or less constant amount of attention, the observed dual-task interference is interpreted as a lack of balance automaticity. With the exception of quiet standing in healthy control subjects, which revealed no task concurrence costs, the results provide an argument for an influential role of attention-demanding processes in postural control depending on the novelty (standing with a prosthesis) and complexity (feedback-controlled weight shifting) of the task.

As for quiet standing with a lower limb prosthesis, this interpretation implies that attention-invested control mechanisms are involved in safely maintaining a vertical posture during the early phases of skill acquisition. Such a "cognitive bypass" in balance control may be needed to detect body disturbances or executional errors that are insufficiently compensated by automatic prestructured responses due to the lack of somatosensory cues and the loss of output structures at the side of amputation. Not until an adapted repertoire of sensorimotor strategies has been formed, balance behaviour is slow, jerky and probably strongly dependent on immediate (e.g. visual) feedback. Consequently, basic equilibrium reactions remain vulnerable to all sorts of changes in the environmental context. Clinically, this idea implies that in the case of sudden damage to the peripheral sensorimotor system balance behaviour should be tested under various environmental conditions to reliably assess the level of skill acquisition.

During the rehabilitation process, the amputation group showed a tendency toward a reduction in dual-task interference during quiet standing, which was significant for the FA sway. A similar reduction in interference of a secondary attention-demanding task with quiet standing was found in an earlier study on balance recovery after lower limb amputation (Geurts, Mulder, Nienhuis, & Rijken, 1991). Because learning to stand safely with a lower limb prosthesis can be considered as the acquisition of a *novel* skill, it is probably better accounted for by intratask automaticity (Brown & Carr, 1989) than by task combination strategies such as attention switching (Broadbent, 1982) or task integration (Hirst, Spelke, Reaves, Caharack, & Neisser, 1980). In particular, automaticity of balance control may be based on a process of information encapsulation (McLeod, McLaughlin, & Nimmo-Smith, 1985) through which invariant relationships between input and output patterns are captured into a computational structure or knowledge representation. That is, new efferent patterns develop that are gradually associated with the appropriate sensory cues through learning. Whereas at the start of rehabilitation visual and vestibular cues predominate balance control, a gradual integration may occur of proprioceptive and exteroceptive inputs, in particular from the amputated leg, into the multi-sensory control of posture.

A related conception of (distributed) knowledge representation in skill acquisition is used in neural network models with a strong emphasis on error detection and correction (see e.g. Rumelhart & McClelland, 1986). In most cognitive theories of skill acquisition, some form of knowledge representation plays an essential role, such as in the formation of production systems (Anderson, 1982), schemata (Schmidt, 1975), memory instances (Logan, 1988), as well as in the transition of closed-loop to open-loop control in classical motor-program theories (Adams, 1971; Keele, 1968). The basic idea is that, once a representation of the task has been formed, motor output can be reliably specified without input from capacity-limited mechanisms such as a general-purpose central processor or working memory (Brown & Carr, 1989). McLeod et al. (1985) argued that practice reduces the need for interaction between general high-level resources and low-level resources as the lower level gradually acquires the correct parameters for coping with the task.

Remarkably, the results of the dynamic tests contrast in essential aspects with those of the static tests Whereas dual-task interference during quiet standing was significant only for the amputation group, the weight-shifting task revealed an equal and relatively small amount of interference in each of the groups. This discrepancy can be explained by the complexity of the dynamic balance task in relation to the easiness of the arithmetic task

During the applied dynamic task, subjects had to maintain a vertical posture while, at the same time, they were requested to actively perturb equilibrium by making voluntary body-mass displacements on the basis of visual feedback Because the transformation of this artificial feedback into well-adjusted weight shifts requires a considerable amount of closed-loop attention-invested visuomotor processing and because this study shows that simply maintaining a vertical posture with a lower limb prosthesis is not an attention-free activity, it must be concluded that for recently amputated subjects the dynamic balance task has dual-task characteristics already in the single-task situation In view of this consideration, the rather easy arithmetic task probably imposes few additional attention demands Hence, lack of automaticity of the basic equilibrium reactions in the amputation group may have caused a deterioration of the single-task weight-shifting performance, rather than a large amount of interference with the arithmetic task

Similarly, reacquisition of automaticity of the basic equilibrium reactions in the amputation group may have led to an improved weight-shifting performance, rather than to a reduction in interference with the arithmetic task. Because the visuomotor processing during the dynamic balance task will never be fully automatic, it must be expected that some degree of interference with a concurrent attention-demanding task will always occur This notion corroborates our prediction and is underscored by the control data It is also in conformity with Adams' (1981) conclusion that cognitive involvement in a complex motor task may disappear only when there is no external indication of error and when the subject perceives his performance as optimal and error-free.

Only in the amputation group, a substantial learning effect on the speed of weight shifting was found besides a (nonsignificant) tendency toward an improvement of weight-shifting fluency. It is plausible, however, that the latter lack of significance has been influenced by the rather small number of (seven) subjects We, therefore, conclude that the amputation group improved their control of voluntary body-mass displacements, which is perhaps (partly) based on a re-automatization of the basic equilibrium reactions

Nonetheless, in general, the data provide few indications of improved balance performance in the amputation group with practice. Besides the possibility that the assessment period has been too short, this result could be due to the fact that some balance training took place already before week 0 when subjects stood with a provisional prosthesis (airboot). Hence, the assessment period of 8 weeks that was used in this study does not represent the complete rehabilitation process. A second reason could be that the selected tasks or variables lack validity, although this idea is contradicted by the fact that clear differences were found between the amputation group and the control subjects for every variable in each of the balance tasks. It is possible, however, that a more difficult arithmetic task would have provided more pronounced changes in dual-task interference across time.

In conclusion, this study provides an indication of balance automatization following lower limb amputation which results from a central adaptation to the structural lesions of the peripheral sensorimotor system. It is assumed that the need for attention-demanding processing subsides as an adjusted repertoire of postural strategies develops on the basis of adaptable sensorimotor representations (Droulez, Berthoz, & Vidal, 1985; Massion, 1992). However, such a reduction in central-processing demands does not yet give a clue as to what neural mechanisms might underlie the central adaptation after lower limb amputation. In this perspective, it is interesting to note that the somatosensory cortex maintains a "lifelong" plasticity as it shows a rapid functional reorganization following sectioning of peripheral nerves or amputation of a body part (Kaas, Merzenich, & Killackey, 1983). Such a reorganizational capacity has recently been reported also for the motor cortex after upper limb amputation (Cohen, Bandinelli, Findley, & Hallett, 1991). As for the somatosensory cortex, it has been argued that the adaptation to changes of peripheral input is based on an input-driven self-organizing capacity within neural networks (Ritter & Schulten, 1986).

Hence, the decrease in attention demands in postural control after lower limb amputation does not necessarily imply that the central reorganization following peripheral sensorimotor damage is under direct attentional control. Instead, it may be that a self-organizational capacity within central representations *indirectly* reduces the need for attention-controlled error detection and correction. It is doubtful, however, whether self-organization within primary somatosensory (or motor) cortex can fully account for the formation of new linkages between sensory input and muscular coordination which is essential to motor learning. The formation of such linkages, which takes place through active experience with a task, requires changes within spatial (kinematic) representations that form a critical link between the sensory and motor systems (Kalaska, 1991). Whether such changes are *directly* mediated by attentional processes is still subject to discussion (Masson, 1990).

Yet, the results of this study corroborate the conclusion by Nashner and Cordo (1981) that the postural control system is only relatively autonomous. During the early phases of (re)acquiring balance skills, capacity-limited central processes appear to be involved in balance control. Such attention-invested control (temporarily) causes an increased vulnerability of the basic equilibrium reactions to secondary task performance. This basic lack of balance automaticity may also deteriorate voluntary control of body-mass displacements during more dynamic postural tasks.

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# CHAPTER 10

# IMPLICATIONS FOR REHABILITATION

**ABSTRACT**

This chapter considers the implications of the work reported in this thesis for rehabilitation medicine. First, the implications for clinical practice are discussed in terms of balance training and assessment in persons with an acquired lower limb amputation. It is inferred from the results of previous chapters that the implementation of sensory, cognitive and motor manipulations in the evaluation of basic sensorimotor skills can provide additional and sometimes indispensable information about functional recovery. Through a *process*-oriented approach it is possible to detect changes in the role of different output-organizing processes in the control of action. Then, the procedure for the assessment of balance performance and recovery as described in this thesis is discussed in view of the International Classification of Impairments, Disabilities, and Handicaps (WHO, 1980). Current proposals for revision of this ICIDH, in which the disability classification is subdivided into functional limitations (i.e. reductions in basic sensorimotor *skills*) and activity restrictions (i.e. reductions in daily *activities*) are endorsed. Whereas outcome-oriented assessment procedures are probably well adapted to describe human activities, process-oriented procedures are recommended to assess context-dependent functional limitations at the level of skills. Finally, some possible directions for future research are presented with regard to the development of novel assessment procedures for monitoring balance recovery following structural damage to the peripheral sensorimotor system.

## INTRODUCTION

The first part of this chapter considers the practical implications of the theory and the empirical data reported in the previous chapters with regard to the rehabilitation of persons with an acquired lower limb amputation. After the implications for training, emphasis is laid on the implications for the clinical assessment of standing balance from the perspective of "routine" evaluation of individual performance. A brief overview of these considerations has earlier been given in a theoretical paper on postural reorganization after lower limb amputation (Geurts & Mulder, in press). As for the practical implications of this study for the rehabilitation of patients with hereditary motor and sensory neuropathy, the reader is referred to the discussion and the conclusion of chapter 7. In the latter two sections, the most important recommendations with regard to orthopedic footwear prescription and accompanying stance and gait training have been described. For the sake of brevity, they will not be repeated here.

The concepts and recommendations concerning the assessment of balance recovery following lower limb amputation may just as well apply to other peripheral pathologies that are characterized by both an efferent and afferent ("bottom-up") disruption of the postural organization, such as after serious peripheral nerve injury of a lower extremity or after (partial) damage to the lumbosacral plexus or spinal nerves. In all these situations, there is a suddenly impaired efficacy of the prestructured muscular synergies, complicated by a reduced or distorted somatosensory input from (parts of) the legs. As a consequence, postural strategies have to be adapted through a central reorganization process, although with a different prognosis for different types of peripheral pathology. Some of the concepts regarding balance assessment may even be useful in cases of disorders of the central nervous system (CNS), such as after (incomplete) spinal cord dissection or after (mild) brain injury. Yet, the implications for assessment will mainly be aimed at gross-motor recovery following peripheral lesions. Similarly, even though some notions could be valid for the assessment of other gross-motor abilities like sitting and walking, the primary focus will be on (the recovery of) standing balance.

This thesis has been directed at the postural adaptation to irreversible alterations of peripheral constraints in patients with sensorimotor impairments of the lower limbs. This theme has been elaborated against a clinical background of balance assessment in rehabilitation. Because the International Classification of Impairments, Disabilities, and Handicaps (ICIDH) (WHO, 1980) is now a widely accepted method for the registration of the consequences of disease, the most essential elements of the assessment of postural reorganization as described in this thesis will be discussed in view of the ICIDH in the second part of the present chapter. The final section discusses some of the interesting possibilities for future research. Here too, emphasis is laid on the development of valid assessment procedures for monitoring balance recovery following structural damage to the peripheral sensorimotor system.

## CLINICAL PRACTICE

### Training

As for the re-education of postural skills after a lower limb amputation, it is possible to draw a few indirect conclusions from the results of this thesis First of all, balance should be trained under increasingly attention-demanding environmental conditions in order to stimulate the development of *automaticity* of postural reactions and adjustments to movement It seems most appropriate to gradually increase the complexity of the task and its context by creating "natural" training situations such as through games or sports Although automaticity of postural control may also develop in less variable training conditions, variability of practice is necessary to improve the transfer of balance skills from the therapy situation to daily life (Mulder, 1991b) Similarly, the re-integration of *somatesthesis* from the stump tissues could be facilitated by training balance in various sensory conditions, especially in those with reduced visual information. In this way, the necessity to rely on alternative sources of input is provoked with the advantage of being in a well-controlled therapy environment By creating complex task conditions, a persistent inability to perform flexibly under various circumstances can already be recognized during the rehabilitation process, so that preventive measures can be taken in terms of additional diagnosis and treatment or, more frequently, in terms of behavioural advice, aids and adaptations in and about the home.

In the early phases of rehabilitation, visual or auditory *feedback* of global output parameters (e g. weight distribution or centre of ground reaction forces) may help to learn how to bear various amounts of weight on the prosthetic limb. Such *extrinsic* feedback may be particularly useful when the availability of intrinsic feedback is grossly insufficient, for instance in patients with diabetes mellitus or with different forms of polyneuropathy In this way, the CNS is supported in matching the (reduced and distorted) intrinsic feedback with the more easily available extrinsic feedback in the expectation that the need for artificial feedback will gradually decrease. It is beyond the scope of this thesis to discuss in detail the role of feedback in motor learning, or the proper design of the learning situation Instead, the reader is referred to the many recent publications on this subject by other authors (see e.g. Holding, 1989, Salmoni, Schmidt, & Walter, 1984, Schmidt, 1988), some of who specifically address the field of rehabilitation (Mulder 1991a, 1991b; Winstein, 1991).

**Assessment**

In chapter 1 the assessment of gross-motor skills, like standing and walking, has been discussed from the perspective of clinical rehabilitation. It was argued that also these forms of motor behaviour, despite their relative autonomous control in healthy individuals, can be regarded as the end-result of a fine-tuned interaction between sensory, cognitive and motor processes; the *ways* of interaction being strongly dependent on the level of skill acquisition (see also Holding, 1989). Furthermore, it was argued that manipulation of the complexity of the task context is a necessity to reliably evaluate the changing contribution of different output-organizing processes to the gross-motor performance. Against this background, four complementary types of task manipulation (sensory, cognitive, motor and mechanical) have been proposed to obtain a fairly complete picture of the skill under evaluation. As for the rehabilitation after a lower limb amputation, the results of chapters 4 and 5 provide convincing evidence for the added value of balance assessment by means of a dual-task and a visual-deprivation condition respectively. As a matter of fact, a significant improvement of balance performance was only found for the latter complex task conditions, whereas standing with eyes open and without additional central-processing demands merely showed a small, non-significant improvement across rehabilitation. The data of chapter 9 corroborate these earlier findings because, also in this study, improvement was only found for the more complex conditions, i.e. balancing in a dual-task condition and feedback-controlled weight shifting. In contrast, the data derived from balancing as a single task are suggestive of a ceiling effect. It may be that, already early in the process of functional recovery, improvement of the more basic balance reactions is constrained by irreversible effects of the amputation. Therefore, it appears that, after reacquisition of the basic motor abilities, task manipulation can be an absolute necessity to detect further functional progress.

Hence, the results of this thesis indicate that the implementation of sensory (visual deprivation), cognitive (dual task) as well as motor (controlled weight shifting) manipulations in the evaluation of a basic task (balancing) can provide additional or even indispensable information about motor recovery. Perhaps the most striking result is that, otherwise unnoticeable, sensory and cognitive processes play a substantial role in the restoration of motor skills and, thus, must be considered as relevant determinants of functional recovery. The theoretical relevance of this inference is apparent from the perspective of the investigator who is interested in the functional characteristics of reorganization processes (this thesis) or in the differential effects of various treatment strategies on the speed and extent of these processes. However, this conclusion also has a *practical* value for the "routine" assessment of individual balance performance and recovery in the clinic. On the basis of similar considerations, the need for a skills laboratory in rehabilitation has already been addressed in the final part of chapter 1. It was hypothesized that repeated assessments of (the automaticity of) postural control might be of value for providing insight into the course of an individual learning process and, thus, about the individual effects of physical interventions, training and medication It was also suggested that a (single) process-oriented analysis of balance performance could provide useful diagnostic information about the functional consequences of disease; a functional diagnosis aimed at understanding the nature and severity of the observed motor dysfunctions at the level of skills (Mulder & Geurts, 1991) These practical aspects of balance assessment relate to an essential, but still poorly developed element of rehabilitation medicine, i.e. the standardized evaluation of individual performance and functional progress.

## BALANCE ASSESSMENT AND ICIDH

The above-mentioned considerations on balance performance are typical of a rehabilitation approach to motor dysfunctions Instead of focusing on the underlying organic disorder, the consequences of the disorder are put forward as the core elements for making a functional diagnostic profile. Generally, the so-obtained rehabilitation diagnosis forms the starting point for making a functional prognosis as well as for selecting the most useful interventions to achieve an optimal participation of the patient in daily life A proper rehabilitation diagnosis should include all relevant sensory, cognitive, motor, and behavioural aspects of human activity assessed at the three distinct levels which have been defined by the World Health Organization in the International Classification of Impairments, Disabilities, and Handicaps (WHO, 1980). In this ICIDH, *impairment* refers to the loss or abnormality of function at the level of the organ or anatomical structure, *disability* refers to the reduction in human activity at the level of the person, whereas *handicap* refers to the disadvantage at the level of social and economic roles As for the relationship between the ideas on balance assessment presented in this thesis on the one hand and the ICIDH on the other hand, two considerations deserve further discussion, both concerning the classification of human disability.

The first consideration concerns the *level* of disability classification. Lack of standing balance certainly does not refer to a dysfunction of one specific anatomical structure, nor does it directly refer to a social or economic disadvantage Hence, the conclusion that a patient suffers from a lack of postural control relates to the disability level. In the present ICIDH classification, lack of standing balance can best be placed in the category of body disposition disabilities (D-5). It can, however, hardly be argued that standing upright is a purposeful activity in itself. Instead, it represents a basic sensorimotor ability or skill on top of which goal-directed movements (e.g.

grasping) and compound activities (e.g. dressing) can be executed. In view of this differentiation, many others have felt the need for distinguishing, besides several categories of human disability, more than one level of disability (see Voorwerk, 1991). For instance in the Netherlands, the Foundation for Science and Education in Physiotherapy has recently published a thoroughly elaborated proposal for a revised ICIDH, in which the first category of the disability classification deals with basic functional abilities, i.e. changing and maintaining body posture and making elementary movements with the trunk and the extremities (SWSF, 1991). Together with the second category on basic psychological and social abilities, it constitutes the "lower" level of the disability coding comprising the elementary *skills* being the keystones for the development of the more compound daily *activities*. In this way, functional limitations are distinguished from activity restrictions.

As a WHO representative, Chamie (1990) also adopted the differentiation between functional limitations and activity restrictions, emphasizing the distinction between specific reductions in bodily functions on the one hand (e.g. hearing, walking and grasping) and in daily activities on the other hand (e.g. dressing, bathing and housework). The level of functional limitations in the latter classification is probably somewhat broader defined than in the SWSF classification. It is apparent that the criteria for coding disabilities through each of the revised classifications need to be further developed and integrated. So far, the aforementioned proposals for revision of the ICIDH agree in their attempt to distinguish a reduced *ability* to perform basic skills from a reduced *performance* of daily activities. The ideas and data reported in this thesis strongly support the need for such a subdivision. Indeed, reduced balance performance as assessed in this thesis would fit well into a category of functional limitations, whereas it would only be indirectly related to certain restrictions in daily activity; a conclusion which is supported by the data on the correlation between balance and ADL described in chapter 5.

The second consideration concerns the theoretical basis of disability classification. In principle, the existing ICIDH is descriptive in nature or, in different terms, outcome-oriented. That is, the assignment of (dis)ability to individuals is based on whether a particular activity is performed, irrespective of how the task is completed. The underlying neural and psychological processes that are involved in the task performance are not taken into consideration, because the end-result of the individual's attempts forms the sole criterion for assessment. The reader will remember that in the previous part of this chapter a way of balance assessment was proposed in which specific manipulations of the task context serve to estimate the (changing) role of various output-organizing processes in the restoration of postural control. Such an analytical approach to balance assessment can be classified as process- or skill-oriented, because it emphasizes the role of sensory, cognitive as well as motor processes in the control of movement. It is argued that a process-oriented approach incorporates physiological and psychological aspects of human motor control and, therefore, provides insight into the context dependency of a skill. For instance, an increased dependency on visual information in standing and walking may constitute a functional limitation in situations in which the visual surround is moving, remote, or scarcely illuminated. Similarly, a substantial involvement of attention-demanding processes in basic sensorimotor abilities will limit the smooth performance of the numerous dual tasks encountered in daily life.

A *process*-oriented approach to disability assessment, aimed at detecting context-dependent functional limitations, seems most appropriate with respect to sensorimotor abilities such as sitting, standing and straight walking. Indeed, particularly these abilities constitute the basis for the development of more complex, goal-directed activities and should therefore be well automated and relatively unaffected by the many, diverse types of change in the task context. Conversely, an *outcome*-oriented approach is probably quite adequate for assessing the compound activities of daily life (ADL) such as dressing and household activities, because a strong influence of the task context is inherent in ADL performance. It can be argued that a process-oriented approach to skill assessment provides insight into common sensorimotor processes underlying different skills. For example, balancing the trunk is a prerequisite for maintaining an erect posture; hence, lack of sitting balance can predict lack of standing and walking balance. It is speculated here that such an analytical approach might reduce the need for a detailed classification of all sorts of "sub-skills" and, thus, increase the clinical applicability of the classification system.

## FUTURE RESEARCH

Adaptation of the muscular coordination in postural control must take place after a lower limb amputation in order to perform safely and flexibly in various situations. This adaptation concerns both the spatio-temporal organization (sequence and timing) and the parametrization (force) of the postural reactions and anticipatory adjustments to movement. As was explained in chapter 2, leg- and trunk-muscle activity must be rearranged into novel synergies, inappropriate muscle activation must be suppressed or counteracted, and contralateral muscles must compensate for the loss of output structures at the side of amputation. Data on these *efferent* aspects of postural reorganization have not been presented in this thesis, because the force-platform registrations merely allowed an interpretation of the motor output in terms of overall efficacy and left-right asymmetry of muscular activity. In order to find changes in the spatio-temporal organization of posture, one would need a different experimental set-up including the application of surface electromyography and the use of balance tasks through which the body is sufficiently perturbed to elicit measurable postural responses. Such perturbations could be induced either actively (e.g. by rapid arm movements) or passively (e.g. by a tilting platform).

Perhaps the most illustrative example of adaptation of muscular coordination following lower limb amputation concerns the active stabilization of the prosthetic knee joint by muscles that act across the hip in persons with a through- or above-knee amputation. Especially the acquisition of control over the hip flexors seems to be essential because activation of these muscles facilitates destabilization of the prosthetic knee. Yet, such activation is part of a highly automated response sequence that is meant to stabilize the body in reaction to or in anticipation of a backward perturbation (Nashner, 1977). In contrast, activation of the hip extensors, which normally takes place in the case of forward body perturbation, increases the stability of the prosthetic knee. In general, it would be of interest to examine whether, and to what extent, inadequate components within prestructured muscular synergies can be selectively suppressed by the CNS. Indeed, an entirely different strategy could be that inadequate muscle activity is not so much suppressed as counteracted by co-activation of antagonistic muscles. It can, for instance, be speculated that younger subjects are able to selectively suppress inappropriate muscle activations at the affected body side, whereas elderly subjects tend towards massive preparatory co-contractions to secure stability of the prosthesis. The type of efferent reorganization could provide relevant information about the adaptability of the CNS in individual subjects. By examining the postural *reactions* at the side of amputation to externally induced fore-aft body perturbations, it should be possible to obtain such information. Implementation of a dual-task procedure in this research design may yield additional information about the development of automaticity (or safety) of the adapted coordination patterns.

In anticipation of a self-initiated disturbance of body equilibrium, leg and trunk muscles are activated to minimize the effect of mass displacement and acceleration on balance. For instance, when making a rapid arm flexion movement, the ipsilateral biceps femoris is one of the first muscles to be activated prior to the initiation of arm movement by the anterior deltoid muscle (Horak, Esselman, Anderson, & Lynch, 1984). In the case of an amputation above the ankle, the efficacy of such *anticipatory* postural activity of the thigh muscles is probably impaired at the side of amputation, because the passive stability provided by the prosthesis is less than the active stability provided by the ankle joint muscles. It is unknown under what circumstances an alternative strategy is adopted by changing the sequence, relative timing, and strength of the postural adjustments in favour of the muscles of the intact limb. Moreover, it would be of particular interest to establish the (changing) role of cognitive processes (e.g. prior knowledge of the task) on such efferent reorganization. Again, this information could provide insight into the context-dependent flexibility of the CNS in individual subjects.

Besides these various efferent aspects of balance reorganization, there is much more to discover about the *afferent* reorganization following peripheral sensorimotor lesions. For instance, one could study in more detail the (changing) role of different visual cues in the control of balance and gait, or one could investigate the influence of various degrees of somatosensory de-afferentiation on the reacquisition and automation of basic sensorimotor skills. All such process-oriented studies will provide fundamental knowledge about central adaptation processes following structural damage to the peripheral sensorimotor system. It seems almost trivial to infer that the so-obtained information can be used to develop valid procedures for the assessment of gross-motor recovery, or that these procedures can be implemented into clinical research programmes aimed at elucidating the effects of different treatment strategies. Still, it cannot be argued too often that, also for rehabilitation medicine, fundamental knowledge about restoration processes constitutes the necessary basis for the evaluation of individual performance as well as for the planning of individually tailored intervention programmes aimed at functional recovery.

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# SUMMARY

# SAMENVATTING

**SUMMARY**

The ability to maintain an upright posture under various circumstances is essential to the unrestrained performance of many daily activities such as walking, stair climbing, bathing and dressing. Although the control of vertical posture is usually not a purposeful activity in itself, it is a basic sensorimotor skill that subserves the execution of other, goal-directed tasks. It is, therefore, hardly surprising that the reacquisition of well-automated postural reactions and adjustments to movement is a keystone in the rehabilitation of many patients suffering from the consequences of lesions of the sensorimotor system. Lesions involving the *peripheral* neuromuscular system often give rise to a combination of motor and sensory impairments of the lower limbs, leading to an efferent-afferent disruption of the postural organization. It is argued that such a "bottom-up" disorganization, which may be caused by many different underlying pathologies, requires a *central* adaptation of postural strategies by linking altered input patterns to novel patterns of muscular activity. Adaptation of the muscular coordination concerns both the spatio-temporal organization (sequence and timing) and the parametrization (force) of the postural reactions and anticipatory adjustments to body perturbations. The main aim of this thesis is to obtain insight into the central processes that determine the adaptation of postural organization to structural alterations of peripheral motor and sensory conditions. Secondly, an attempt is made to translate the theoretical considerations into practical implications for the assessment of balance performance and recovery in persons with irreversible sensorimotor deficits of the lower limbs.

As a consequence of a sudden breakdown of well-automated input-output patterns, there is a (temporary) need to rely on "higher level" attention-invested control strategies as well as on unaffected sensory systems in order to maintain safe balance. Because only part of the central reorganization following peripheral lesions is concerned with the reprogramming of the (observable) motor output per se, sensory and cognitive processes are additionally examined as equally important (but less visible) factors in balance restoration; these factors include the (changing) role of attention-demanding processes as well as of visual information in gross-motor control. Chapter 1 is a general introduction to the importance of task manipulation in the clinical assessment of gross-motor skills in order to detect the changing contribution of output-organizing processes to the observed motor behaviour. Based on such a *process*-oriented approach, four types of task manipulation (sensory, cognitive, motor and mechanical) are proposed to obtain a fairly complete picture of the skill under evaluation.

The focus is then directed at balance performance and recovery after unilateral lower limb amputation. Chapter 2 starts with a discussion of the most important

peripheral impairments as a consequence of lower limb amputation above the ankle and below the hip joint, i.e. a lack of ankle torque generation to restore equilibrium in the sagittal plane, a lack of weight-shifting capacity to control posture in the frontal plane, and a distorted somatosensory input from the side of amputation. In the second part of this chapter, a reduction in the vulnerability to secondary (non-motor) task performance as well as a decrease in visual dependency are put forward as two of the most critical determinants of the postural reorganization following amputation and as relevant predictors of the safety of gross-motor skills in general.

Because in all the reported studies quiet two-legged standing is repeatedly assessed by means of a force platform recording the point of application of the resultant of the ground reaction forces or centre-of-pressure (CP), the intrasubject variability of several selected force-platform parameters in healthy adult subjects is examined in chapter 3. It is argued that the root mean square (RMS) velocity of the CP fluctuations is mainly influenced by the power of the higher frequency components (0.5 - 5 Hz) reflecting the accelerations of the centre of gravity of the body (CG), whereas the RMS CP amplitude is much more related to the actual amount of (low-frequency) body sway (< 0.5 Hz). It was found that, in the fore-aft direction of sway, the RMS CP velocity was a more consistent parameter over ten repeated tests than the RMS CP amplitude. This result suggests that during normal upright standing it is not so much the absolute amplitude of the CG displacements that is minimized, but rather the relative CG displacement in time.

Based on the above-mentioned theoretical and methodological notions, the first empirical evidence for a significant reduction in dual-task interference during the rehabilitation of persons with a recent lower limb amputation is given in chapter 4. Compared with a matched control group, the amputation group showed increased interference of concurrent Stroop task performance with postural control as assessed by the RMS CP velocity in either direction of sway, however most clearly at the start of rehabilitation. Improvement of postural control was significant only for the dual-task condition, indicating a restoration of balance *automaticity* rather than a return of the basic level of balance control to normal values. Chapter 5 presents data from the same amputation group dealing with the changing effect of visual deprivation on postural control. A marked reduction in the dependency on visual information across rehabilitation was identified as a critical *sensory* determinant of balance recovery. This result is accounted for by a central integration of somatosensory input from the stump tissues into the multi-sensory control of posture. Postural asymmetry during quiet standing was, compared to matched controls, more apparent in dynamic (control activity) than in static (weight distribution) terms and remained unaltered during the rehabilitation period.

In contrast to the recovery from an *acute* disruption of the efferent and afferent organization of posture (amputation), the adaptation of the postural organization to *slowly* developing peripheral sensorimotor impairments is also investigated, viz. in patients with hereditary motor and sensory neuropathy (HMSN) type I or type II (chapter 6). Comparable to persons with a lower limb amputation, HMSN patients showed a basically impaired postural regulation as well as an increased visual control of posture. However, unlike the amputation group, their standing balance was not disproportionately vulnerable to a secondary Stroop task indicating a substantial degree of balance automaticity. Apparently, a continuous central reorganization process ensures an optimal adaptation to the slowly developing impairments. In chapter 7, it is shown that the equilibrium between central adaptations and peripheral impairments in HMSN patients is rather precarious, because it is easily disrupted by imposing novel peripheral constraints. For instance, there was a (temporary) loss of balance automaticity immediately after the application of new orthopedic footwear, probably because of a reduction in the efficacy of ankle mechanisms working in the sagittal plane.

Because the above-mentioned results suggest that the human posture-control system is only *relatively* autonomous, chapter 8 investigates whether a loss of balance automaticity can be provoked in healthy adult subjects by changing the configuration of the support surface. In contrast to the HMSN group, a mere reduction in the efficacy of fore-aft ankle mechanisms by balancing on a pair of seesaws was not sufficient to elicit interference with a concurrent Stroop task. Such interference was observed only when there was a need to rely on poorly developed postural strategies such as in lateral sway control while maintaining a tandem stance position. This combination of results implies that, besides the novelty of the task, the integrity of the nervous system further determines the attention demands in postural control.

Chapter 9 returns to the issue of balance recovery following lower limb amputation from a theoretical perspective. The changing influence of an arithmetical task with two postural tasks of different complexity is examined to study the role of attention-demanding processes in the control of posture after structural damage to the peripheral sensorimotor system. Whereas quiet standing as a relatively simple balance task revealed a reduction in dual-task interference over practice, voluntary (feedback-controlled) weight shifting as a complex balance task revealed a constant amount of dual-task interference across time. The results corroborate the hypothesis that attention-demanding processes may be involved in postural control depending on both the novelty and the complexity of the task. The reduction in central-processing demands for quiet standing underscores previous data reported in chapter 4 indicating a return of balance automaticity following lower limb amputation.

Chapter 10 presents some preliminary notions on the implications for clinical rehabilitation. In conformity with the ideas mentioned in chapter 1, it is inferred from the results presented in this thesis that the implementation of sensory, cognitive and motor manipulations in the evaluation of basic sensorimotor skills can provide additional and sometimes indispensable information about functional recovery. Through a process-oriented approach it is possible to detect changes in the role of different output-organizing processes in the control of action. In the latter half of this chapter, the described procedure for the assessment of balance performance and recovery is discussed in view of the International Classification of Impairments, Disabilities, and Handicaps and of some current proposals for revision of this ICIDH. The work reported in this thesis seems to support the need for a subdivision of the Disability classification into functional limitations (for coding reductions in basic sensorimotor *skills*) and activity restrictions (for coding reductions in daily *activities*); reduced balance performance being a functional limitation which may indirectly lead to certain restrictions in personal activity. Whereas outcome-oriented procedures are probably well adapted to describe human activities, process-oriented procedures are recommended to assess context-dependent limitations at the level of skills. Such an analytical approach to skill assessment adds a new dimension to making an individual functional diagnosis and prognosis, as well as to evaluating functional progress in rehabilitation medicine.

## SAMENVATTING

Het vermogen om onder wisselende omstandigheden rechtop te blijven staan is essentieel voor het ongehinderd uitvoeren van vele dagelijkse activiteiten zoals lopen, traplopen, wassen en kleden. Ofschoon het handhaven van een verticale lichaamshouding doorgaans geen doelgerichte activiteit op zich is, is het een basale sensomotorische vaardigheid welke in dienst staat van andere doelgerichte activiteiten. Het is daarom nauwelijks verbazingwekkend dat het herleren van goed geautomatiseerde houdingscorrecties en -aanpassingen aan beweging centraal staat in de revalidatie van vele patiënten die lijden aan de gevolgen van een beschadigd sensomotorisch apparaat. Beschadigingen die het *perifere* neuromusculaire apparaat betreffen leiden dikwijls tot een combinatie van motorische en sensorische stoornissen van de onderste extremiteiten met als gevolg een efferente-afferente ontregeling van de houdingsregulatie. Een dergelijke "bottom-up" ontregeling, welke veroorzaakt kan worden door diverse onderliggende aandoeningen, vereist een *centrale* adaptatie van balansstrategieën waarbij veranderde sensorische (input) patronen worden gekoppeld aan nieuwe patronen van spieractivering (output). Adaptatie van de musculaire coördinatie betreft zowel de spatio-temporele organisatie (volgorde en timing) als de parametrisatie (kracht) van de posturale reacties en anticipaties op lichaamsverstoringen. Het belangrijkste doel van deze dissertatie is het verwerven van inzicht in de centrale processen die de adaptatie van de houdingsregulatie aan structurele veranderingen van perifere motorische en sensorische condities bepalen. Daarnaast wordt getracht de theoretische overwegingen te vertalen in praktische implicaties voor de beoordeling van balanshandhaving en -herstel bij mensen met irreversibele sensomotorische stoornissen van de onderste extremiteiten.

Door een plotselinge ineenstorting van goed geautomatiseerde input-output patronen bestaat er een (tijdelijke) noodzaak om over te schakelen op hogere orde aandachtvragende motorische strategieën evenals op onbeschadigde sensorische systemen voor een veilige balanshandhaving. Omdat slechts een deel van de centrale reorganisatie na perifeer letsel het herprogrammeren van de (zichtbare) motorische output betreft, worden tevens sensorische en cognitieve processen bestudeerd als even belangrijke (evenwel minder zichtbare) factoren die het herstel van de houdingsregulatie bepalen; deze factoren omvatten de (veranderende) rol van aandachtvragende processen evenals van visuele informatie bij de sturing van de grove motoriek. Hoofdstuk 1 vormt een algemene inleiding op het belang van taakmanipulatie bij de klinische beoordeling van grof-motorische vaardigheden om de veranderende bijdrage van voorbereidende processen aan het geobserveerde motorische gedrag te kunnen vaststellen. Op basis van een dergelijke *proces*-georiënteerde benadering worden vier typen van taakmanipulatie (sensorisch, cognitief, motorisch en mechanisch) voorgesteld om een goed beeld te verkrijgen van de te onderzoeken vaardigheid.

Het accent wordt vervolgens verlegd in de richting van balanshandhaving en -herstel na enkelzijdige beenamputatie. Hoofdstuk 2 begint met een verhandeling over de belangrijkste perifere stoornissen als gevolg van een beenamputatie boven het niveau van de enkel en onder het niveau van de heup, te weten een beperkt vermogen om enkelmoment te genereren ten behoeve van houdingscorrecties in het sagittale vlak, een beperkt vermogen om gewicht te verplaatsen in het frontale vlak, en een verstoorde somatosensorische input vanuit de geamputeerde lichaamshelft. In de tweede helft van dit hoofdstuk wordt betoogd dat een verminderde kwetsbaarheid voor de gelijktijdige uitvoering van een (niet-motorische) taak evenals een verminderde visuele afhankelijkheid twee van de meest kritische determinanten zijn van het balansherstel na beenamputatie en belangrijke predictoren voor de veiligheid van grof-motorische vaardigheden in het algemeen.

Omdat in alle gerapporteerde studies tweebenig stilstaan herhaaldelijk wordt gemeten met behulp van een krachtenplatform dat het aangrijpingspunt van de resultante van de grondreactiekrachten (CP) registreert, wordt in hoofdstuk 3 de intrasubject-variabiliteit van verscheidene geselecteerde platform-parameters bij gezonde volwassen mensen onderzocht. Er wordt gesteld dat de "root mean square" (RMS) van de snelheid van de CP fluctuaties vooral wordt beïnvloed door het vermogen van de hoog-frequente componenten (0.5 - 5 Hz), welke de versnellingen van het lichaamszwaartepunt (CG) weerspiegelen, terwijl de RMS van de CP amplitudo meer gerelateerd is aan de werkelijke hoeveelheid (laag-frequente) lichaamsverplaatsing. Over tien herhaalde metingen bleek in de voor-achterwaartse richting de RMS van de CP snelheid minder variabel dan de RMS van de CP amplitudo. Deze bevinding bevestigt de hypothese dat tijdens rustig rechtop staan niet zozeer de absolute amplitudo van de CG verplaatsingen wordt geminimaliseerd, als wel de relatieve CG verplaatsing in de tijd.

Op basis van de bovenstaande theoretische en methodologische overwegingen worden in hoofdstuk 4 de eerste empirische gegevens aangeleverd voor een significante vermindering van dubbeltaak-interferentie gedurende de revalidatie van mensen met een recente beenamputatie. In vergelijking met een controlegroep toonde de amputatiegroep meer interferentie van een gelijktijdig uitgevoerde Stroop taak met de houdingsregulatie beoordeeld op basis van de RMS van de CP snelheid in beide richtingen; dit was echter het meest uitgesproken aan het begin van de revalidatie. Verbetering van de houdingsregulatie was alleen significant voor de dubbeltaak-conditie, hetgeen eerder duidt op een herstel van balans-*automaticiteit* dan op een normalisatie van het basisniveau van balanshandhaving. In hoofdstuk 5 worden gegevens gepresenteerd van dezelfde amputatiegroep die de veranderende invloed van visuele deprivatie op de balanshandhaving betreffen. Een opvallende vermindering van de visuele afhankelijkheid gedurende de revalidatie wordt gevonden als een kritische *sensorische* determinant van het balansherstel. Deze bevinding wordt verklaard op basis van een centrale integratie van somatosensorische input vanuit de stompweefsels in de multi-sensorische regulatie van de lichaamshouding. Houdingsasymmetrie tijdens stilstaan bleek in vergelijking met controlepersonen duidelijker in dynamische (regelactiviteit) dan in statische (gewichtverdeling) zin en bleef onveranderd gedurende de revalidatie.

Naast het herstel van een *acute* ontregeling van de efferente en afferente balansorganisatie (amputatie) wordt ook de adaptatie van de houdingsregulatie aan *langzaam* ontwikkelende perifere sensomotorische stoornissen onderzocht, namelijk bij patiënten met erfelijke motorische en sensorische neuropathie (HMSN) type I of type II (hoofdstuk 6). Vergelijkbaar met mensen met een beenamputatie toonden de HMSN patiënten een verminderd basisniveau van de houdingsregulatie evenals een toegenomen visuele afhankelijkheid. Echter anders dan bij de amputatiegroep bleek hun balanshandhaving niet onevenredig kwetsbaar voor een gelijktijdig uitgevoerde Stroop taak, hetgeen duidt op een aanzienlijke mate van balansautomaticiteit. Klaarblijkelijk garandeert een voortdurende centrale reorganisatie een optimale adaptatie aan de zich langzaam ontwikkelende stoornissen. In hoofdstuk 7 wordt aangetoond dat het evenwicht tussen centrale adaptaties en perifere stoornissen bij HMSN patiënten tamelijk precair is, aangezien het gemakkelijk wordt verstoord door het opleggen van nieuwe perifere condities. Er ontstond bijvoorbeeld een (tijdelijk) verlies van balansautomaticiteit onmiddellijk na de verstrekking van nieuw orthopedisch schoeisel, waarschijnlijk ten gevolge van een verminderde effectiviteit van enkelmechanismen werkzaam in het sagittale vlak.

Aangezien de bovenstaande resultaten voeding geven aan de idee dat de houdingsregulatie bij de mens slechts een *relatief* autonoom regelsysteem betreft, wordt in hoofdstuk 8 onderzocht of bij gezonde volwassen proefpersonen een verlies van balansautomaticiteit kan worden geprovoceerd door de configuratie van het dragend oppervlak te wijzigen. In tegenstelling tot de HMSN groep, bleek het verminderen van de effectiviteit van voor-achterwaartse enkelmechanismen door het balanceren op een tweetal rollende blokken onvoldoende om interferentie met een gelijktijdig uitgevoerde Stroop taak te ontlokken. Een dergelijke interferentie werd pas gevonden wanneer men was aangewezen op weinig ontwikkelde balansstrategieën zoals bij de houdingsregulatie in het frontale vlak tijdens het staan met de ene voet voor de andere (tandem positie). Dit geheel aan bevindingen impliceert dat, naast de onbekendheid van de taak, ook de integriteit van het zenuwstelsel bepalend is voor de aandachtsbelasting van een balanstaak.

Hoofdstuk 9 grijpt terug op het balansherstel na een beenamputatie vanuit een theoretisch perspectief. Om de betekenis van aandachtvragende processen voor de houdingsregulatie na structureel letsel van het sensomotorische apparaat nader te bepalen, werd de veranderende invloed van een rekentaak op twee balanstaken van verschillende complexiteit onderzocht. Terwijl het stilstaan als relatief eenvoudige balanstaak een vermindering van dubbeltaak-interferentie over de trainingsperiode liet zien, toonde het willekeurig (feedback-gestuurd) gewicht-verplaatsen als complexe balanstaak een constante hoeveelheid dubbeltaak-interferentie in de tijd. Deze resultaten zijn in overeenstemming met de hypothese dat aandachtvragende processen betrokken kunnen zijn bij de houdingsregulatie afhankelijk van de onbekendheid en de complexiteit van de taak. De vermindering van aandachtsbelasting welke werd gevonden voor het stilstaan bevestigt eerdere bevindingen beschreven in hoofdstuk 4 die wijzen op een terugkeer van balansautomaticiteit na een beenamputatie.

Hoofdstuk 10 behandelt enkele voorlopige ideeën met betrekking tot de implicaties voor klinische revalidatie. In overeenstemming met de inhoud van hoofdstuk 1 wordt op basis van de gepresenteerde resultaten geconcludeerd dat de implementatie van sensorische, cognitieve en motorische manipulaties bij de beoordeling van sensomotorische basisvaardigheden aanvullende en soms onmisbare informatie kan verschaffen over functioneel herstel. Middels een proces-georiënteerde benadering is het mogelijk om veranderingen in de rol van verschillende voorbereidende processen bij de bewegingssturing vast te stellen. In de tweede helft van dit hoofdstuk wordt de beschreven procedure voor de beoordeling van balansvaardigheid en -herstel besproken in het licht van de Internationale Classificatie van Stoornissen, Beperkingen, and Handicaps (ICIDH) en van enkele recente voorstellen tot wijziging van deze ICIDH. Het hier beschreven onderzoek lijkt de noodzaak van een onderverdeling van de Beperkingen classificatie in functionele beperkingen (voor het vastleggen van tekorten in sensomotorische *basisvaardigheden*) en restricties in het activiteitenniveau (voor het vastleggen van vermindering van dagelijkse *activiteiten*) te ondersteunen; hierbij moet een verminderde balansvaardigheid worden gezien als een functionele beperking welke indirect zou kunnen leiden tot bepaalde restricties in het persoonlijke activiteitenniveau. Ofschoon resultaat-georiënteerde procedures waarschijnlijk geschikt zijn om menselijke activiteiten te beschrijven, worden proces-georiënteerde procedures aanbevolen om context-afhankelijke beperkingen in basisvaardigheid te beoordelen. Een dergelijke analytische benadering van vaardigheidsbeoordeling voegt een nieuwe dimensie toe aan het stellen van een individuele functionele diagnose en prognose, als ook aan het evalueren van functioneel herstel in de revalidatiegeneeskunde.

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## CURRICULUM VITAE

De schrijver van dit proefschrift werd op 17 augustus 1961 geboren te Nijmegen. In 1979 behaalde hij het ongedeeld VWO examen aan het Canisius College in zijn geboortestad. Van 1979 tot 1987 studeerde hij geneeskunde aan de Katholieke Universiteit Nijmegen, in welke periode hij tevens de diploma's tennisleraar A en B van de KNLTB verwierf. In de periode van april tot augustus 1986 werkte hij als co-assistent in Sumve Hospital, Sumve, Mwanza region, Tanzania. Na het behalen van het artsexamen op 20 maart 1987, was hij van 23 maart 1987 tot 1 maart 1988 werkzaam als wisselassistent in Ziekenhuis Overvecht te Utrecht. Tevens nam hij waar als docent chirurgie aan de opleiding tot A-verpleegkundige van het Academisch Ziekenhuis Utrecht.

Van 1 maart 1988 tot 1 maart 1992 is hij als AIO verbonden geweest aan het Nijmeegs Instituuut voor Cognitie en Informatie (NICI), sectie motoriek en revalidatie (Prof. dr. A.J.W.M. Thomassen). In deze hoedanigheid was hij gedetacheerd bij de Afdeling Research en Ontwikkeling van de St. Maartenskliniek te Nijmegen (dr. T.W. Mulder), waar het in dit proefschrift beschreven onderzoek werd uitgevoerd. In dezelfde periode nam hij verscheidene maanden waar als zaalarts op de revalidatie afdelingen van de St. Maartenskliniek, begeleidde hij een tweetal doctoraalstudenten Bewegingswetenschappen alsmede een keuzevakstudent Geneeskunde, en werkte hij mee als gastdocent motoriek en revalidatie aan diverse NDT-cursussen voor paramedici en verpleegkundigen in de regio's Eindhoven en Nijmegen. Sinds 1 april 1992 is hij in opleiding tot revalidatiearts in het opleidingscluster Nijmegen gevormd door de St. Maartenskliniek en het Canisius Wilhelmina Ziekenhuis (R.A.J. Rijken).

**STELLINGEN**

**behorende bij het proefschrift**

**CENTRAL ADAPTATION OF POSTURAL ORGANIZATION TO PERIPHERAL SENSORIMOTOR IMPAIRMENTS**

**door**

**Alexander C. H. Geurts**

**Nijmegen, 18 november 1992**

I. Bij de bestudering van functioneel herstel na perifeer sensomotorisch letsel is taakmanipulatie van belang om veranderingen in de sensorische en cognitieve regulatie van de motoriek te kunnen vaststellen. (dit proefschrift)

II. Na een beenamputatie dient er een automatisatie van de houdingsregulatie plaats te vinden als basis voor het herleren van meer complexe vormen van grove motoriek. Het uitblijven van een dergelijke automatisatie moet worden gezien als een complicatie in het revalidatieproces (dit proefschrift)

III. Na een beenamputatie kan er slechts in beperkte zin sprake zijn van re-symmetrisatie. Symmetrie is derhalve niet het meest geschikte criterium voor het herstel van de houdingsregulatie. (dit proefschrift)

IV. Bij het voorschrijven van een orthese verdient het aanbeveling om, naarmate de integriteit van het zenuwstelsel sterker is aangedaan, meer rekening te houden met bijkomende effecten van centrale adaptatie. (dit proefschrift)

V. Naarmate een motorische taak een groter beroep doet op weinig ontwikkelde balansstrategieën, wordt de evenwichtshandhaving bij de mens sterker beïnvloed door centrale processen die aandachtscapaciteit vergen. (dit proefschrift)

VI. De neuropsychologische en neuropsychiatrische aspecten van mensen die lijden aan de gevolgen van cerebrale beschadiging zijn tot op heden in de revalidatiegeneeskunde onderbelicht gebleven.

VII. Omdat elke vorm van wetenschap *inperking* van haar aandachtsgebied vereist, bestaat er een principieel spanningsveld tussen revalidatie-geneeskundig onderzoek en revalidatiegeneeskunde als klinische discipline welke per definitie een *breed* aandachtsgebied wil bestrijken.

VIII. Eenvoudige observatie van de peutervoet tijdens het staan en lopen toont de cruciale rol van de teenflexoren bij de ontwikkeling van de houdingsregulatie.

IX. Wanneer blijkt dat personeel uit alle geledingen van een kliniek voor orthopedie, reumatologie en revalidatie zonder schroom parkeert op essentiële doorgangs- en standplaatsen voor rolstoelgebruikers kan men daaruit afleiden dat professionele omgang met personen met functionele beperkingen geenszins een garantie vormt voor een daarbij passende sociale attitude.

X. Er is in de revalidatie meer behoefte aan criteria voor het *staken* van functionele training dan voor het *starten* van dergelijke therapie.

XI. De grote betekenis die, getuige het veelvuldige gebruik van statistische toetsen, in de wetenschap aan het toeval wordt toegekend staat in schril contrast met de werkelijkheid van alledag waarin voor elk maatschappelijk verschijnsel een oorzaak wordt gezocht.

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