The Sit-to-Stand Movement

recovery after stroke and
objective assessment

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The Sit-to-Stand Movement

recovery after stroke and objective assessment

Van zitten naar staan

het herstel na een beroerte en het objectief meten van de beweging

Proefschrift

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1

Introduction and Outline

The STS movement

The Sit-to-Stand (STS) movement can be described as the change in body posture from a sitting to standing position. In more biomechanical terms, it can be defined as a transitional movement to the upright posture requiring movement of the center of mass from a stable to a less stable position over extended lower extremities.[1] The STS movement is an important skill because it is related to functioning and mobility, and is a prerequisite for walking.[2-7] The execution of the STS movement varies within and between persons, because many factors influence the way how people perform an STS movement, e.g. seat height, arm rests, feet position, age, and lower extremity muscle strength.[8-14] Literature indicates that also health condition, e.g. neuromuscular disorders, joint disorders and stroke, can result in specific changes in the execution of the STS movement.[15-20]

Stroke and STS movement

A stroke can have a profound effect on the execution of the STS movement; in the acute phase but also in the chronic phase a stroke frequently results in a distorted STS movement. This has been described by several authors, who generally focused on the chronic stage.[21-25] This altered execution is characterized by, for example, slowing down, asymmetry of body weight support, asymmetry of joint moments produced, asymmetry of joint kinematics, and amount of support needed.[3,16,24-28]

However, studies on the recovery of the STS movement after an acute stroke are scarce, and therefore little is known about the patterns of recovery. In gait, for example, the pattern of recovery can be summarized by a strong initial recovery, followed by leveling off in the interval three to six months after start of recovery. It is not certain, however, whether this pattern is similar for the STS movement.[29,30]

From STS movement to STS-related functioning

A stroke does not only affect the execution of the STS movement, but also other aspects of functioning related to the STS movement. The International Classification of Functioning (ICF)[31] can be used to clarify these other aspects and their relation with the STS movement. In the perspective of the ICF the STS movement is primarily positioned in the domain 'Activities', associated with the part of maintaining and changing body posture as described in the chapter on Mobility. In this thesis, execution of the STS movement comprises data on being able to execute the STS movement, the rising speed, the balance control during the STS movement, and the number of STS movements during daily life. However, the STS movement is not only related to body functions (such as muscle strength and vestibular problems), but also to (sub) domains such as (experienced problems in) self-care and mobility, independent functioning, participation, and quality of life. To encompass all these constructs, we prefer to use the term 'STS-related functioning'. Thus, STS-related functioning comprises the execution of the STS movement, as well as all items related to the STS movement from the body function domain, the participation domain, and items (other than STS movement execution) from the Activity domain.

STS movement execution

Within the ICF domain of Activities a distinction is made between the qualifiers Capacity and Performance. The Capacity qualifier describes an individual's ability to execute a task or an action, whereas the Performance qualifier describes what an individual does in his or her current environment.[31] Both qualifiers can be applied to quantity aspects of activities (e.g. "how often can/does a patient perform an STS movement") and to quality aspects (e.g. "how can/does a patient perform an STS movement"). STS movement execution comprises the quality and quantity of the STS movement, in terms of both capacity and performance. In this thesis, it comprises data on being able to execute the STS movement, the rising speed, the balance control during the STS movement, and the number of STS movements during daily life.

An additional distinction that can be made is the difference between Actual and Self-reported information. This difference may concern the Capacity qualifier where actual STS capacity comprises both quality and/or quantity of STS movement execution as (objectively) measured in a 'standardized' environment (i.e. capability, rising speed). It is, however, especially relevant with respect to Performance: this latter item concerns the difference between what a patient actual does, objectively measured (actual performance), and what a patient thinks he/she does, and/or which problems he/she experiences in daily life, based on self-reports (self-reported performance). Figure 1 is a graphical representation of these various aspects.

A clinically relevant – perhaps even the most important – issue is whether or not a patient is actually able to perform an STS movement, which is related to the Capacity/Quantity/Actual[†] part of Figure 1 (not possible =0). Most studies on the STS movement in stroke patients focus on objectively measured STS movement execution in a movement laboratory, which can be related to the Capacity/Quality/Actual[*] part of Figure 1.

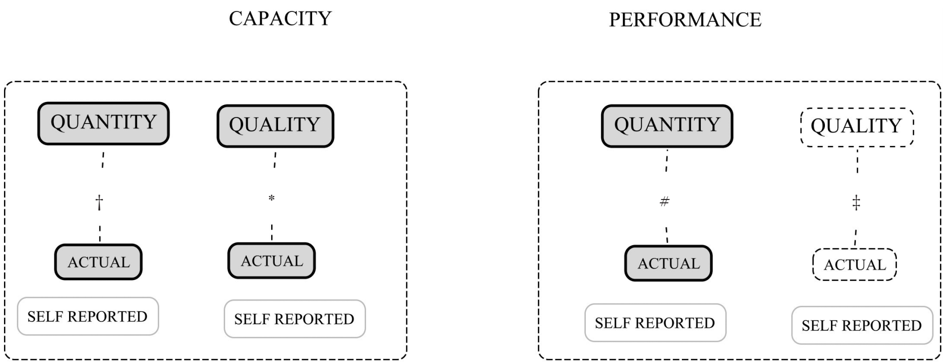

Figure 1 *Diagram showing execution of the STS movement.*
Areas that are covered by this study are shaded grey; for an explanation of the assessment of the actual quality of Performance see the text.

From Figure 1 and from the previous paragraph, it becomes clear that these capacity measures may provide potentially relevant information, but only on a specific part of STS-related functioning.

Because these measures do not directly provide information about Performance and/or other aspects of STS-related functioning, it is uncertain whether they are *an indicator* of other aspects of STS-related functioning.

Another issue is that assessments in the gait laboratory are based on the assumption that the execution within a movement laboratory is similar to that under daily life conditions, which is in fact questionable.[32,33] The discussion centers on the difference between execution under optimized and artificial conditions and execution under normal daily life conditions. From this perspective, it is questionable whether data obtained in a movement lab are sufficiently valid. Therefore, it seems worthwhile to strive for measurements in which the STS movement is performed in a way that is as natural and representative as possible. An example of this discrepancy between laboratory-based and community-based assessments is provided by Taylor and Dean;[34,35] they showed that clinic-measured gait speed (i.e. walking speed over 10 meters) was related to gait speed measured in the community in people with stroke, but for those who walked slowly gait velocity during daily life could be overestimated.

Additionally, even if ecological validity would not be an issue, its feasibility still might be questionable. For example, assessment of the STS movement in a movement lab requires complex and expensive instrumentation, such as force-plates, optoelectronic devices and video. Clinical bedside assessments, such as the timed Get-Up and Go Test[36] and the 5-repetition sit-to-stand test[9,14] are cheaper and more user friendly, but do not provide very specific information. Therefore, simple and cheap instruments that can be used to assess the quantity and quality of STS performance would be beneficial to STS research and clinical practice. Thanks to technological developments in the last decades, this had become within reach.

Actual STS Performance

In research performed so far in subjects with stroke, little attention has been paid to actual STS performance. Actual STS performance comprises the actual quality[‡] and/or quantity[≠] (Figure 1) of STS movement execution as (objectively) measured in the subject's own environment (in this thesis: number of STS movements). Objective assessment of quantity and quality of actual STS performance requires specific assessment techniques, which became available for daily life assessments with the introduction of body-fixed sensors such as accelerometers. This technique is increasingly used and potentially provides a basis for valid assessment of several aspects of actual performance of humans.

Quantitative data on STS movement during daily life are known from a study by de Bruin et al., based on long-term ambulatory monitoring in *healthy* elderly subjects in residential care.[37] Some research has focused on objectively assessing actual performance in *stroke* subjects during daily life; in these studies, primarily the number of steps taken during daily life was measured.[38-40] Data on the quantity and quality of STS movements in stroke subjects during daily life are scarce. Two interventional studies that aimed to influence STS movement capacity by extra (task-specific) training during inpatient rehabilitation reported preliminary data on the number of STS movements in daily life were presented.[41,42] These studies presented the number of STS movements (mean 10.6/24h[41] and 18.6/therapy day[42]) but did not assess qualitative data on STS movement, such as movement duration.

Accelerometry has been used for assessment of body posture and mobility in several groups of patients.[43-47] Its validity has been shown for the detection of body postures and motions, where it is valid to assess the duration of periods of body postures and motions, together with a classification of the type of activity.[43,47-49] More recently its possible value for detailed evaluation of gait, postural changes and balance control have received more attention.[50-58] However, the validity of accelerometry to assess spatio-temporal characteristics of the STS movement has yet to be determined.

Aim and outline of the current studies

The main objective of the current thesis was to describe the time course of the recovery of STS-related functioning in the first year after stroke in a cohort of subjects with a first-ever stroke. In these studies, actual performance and movement analysis outside the movement lab had special interest. Therefore, some initial investigations were needed. First of all, we needed to know which determinants potentially affect the execution of STS movement; therefore, we made an overview of the literature (**Chapter 2**). Based on the application of the accelerometry-based Activity Monitor, we aimed to assess not only the quantity aspects of STS performance, but also the quality aspects, such as duration of the STS movement and balance control during this movement. For that purpose, we needed to understand the content of the accelerometer signals during that movement (**Chapter 3**). Subsequently we explored the validity of accelerometry to assess STS duration (**Chapter 4**) and to assess balance control (**Chapter 5**). Recovery of the capability to rise, the STS movement duration, actual STS movement performance and other aspects of STS-related functioning during the first year after a stroke were explored in a prospective cohort study (**Chapter 6**). Finally, **Chapter 7** focuses on predictive variables for actual STS movement performance, and on the association between STS-related functioning and STS-movement recovery.

2

Determinants of the sit-to-stand movement: a review

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Abstract

Background and Purpose
The sit-to-stand (STS) movement is a skill that helps determine the functional level of a person. Assessment of the STS movement has been done using quantitative and semiquantitative techniques. The purposes of this study were to identify the determinants of the STS movement and to describe their influence on the performance of the STS movement.

Methods
A search was made using MEDLINE (1980-2001) and the *Science Citation Index Expanded* of the Institute for Scientific Information (1988-2001) using the key words "chair", "mobility", "rising", "sit-to-stand", and "standing". Relevant references such as textbooks, presentations, and reports also were included. Of the 160 identified studies, only those in which the determinants of STS movement performance were examined using an experimental setup (n=39) were included in this review.

Results
The literature indicates that chair seat height, use of armrests, and foot position have a major influence on the ability to do an STS movement. Using a higher chair seat resulted in lower moments at knee level (up to 60%) and hip level (up to 50%); lowering the chair seat increased the need for momentum generation or repositioning of the feet to lower the needed moments. Using the armrests lowered the moments needed at the hip by 50%, probably without influencing the range of motion of the joints. Repositioning of feet influenced the strategy of the STS movement, enabling lower maximum mean extension moments at the hip (148.8 N·m versus 32.7 N·m when the foot position changed from anterior to posterior).

Discussion and Conclusion
The ability to do an STS movement, according to the research reviewed, is strongly influenced by the height of the chair seat, use of armrests, and foot position. More study of the interaction among the different determinants is needed. Failing to account for these variables may lead to erroneous measurements of changes in STS performance.

Introduction

The sit-to-stand (STS) movement is one function people frequently use as they change from a sitting position to a standing position (and then often to walking). The ability to go from a sitting position to a standing position is an important skill; in elderly people, the inability to perform this basic skill can lead to institutionalization, impaired functioning and mobility in activities of daily living (ADL), and even death.[59-61] Changes in ability to perform the STS movement are found in elderly people and people with disabling diseases and are related to the determinants of the STS movement.[11,15-17,26,62-70] In a survey of independently living Dutch men and women aged 55 years and older, 25% of the men reported moderate disability and 5% of the men reported severe disability (as compared with 37.4% and 7.8% of the women, respectively) on the rising component of the Health Assessment Questionnaire.[59]

The manner in which the STS movement is defined depends to some extent on the aim of the study. Roebroeck et al.,[71] for example, defined the STS movement as moving the body's center of mass upward from a sitting position to a standing position without losing balance. Vander Linden et al.[1] defined the STS movement as a transitional movement to the upright posture requiring movement of the center of mass from a stable position to a less stable position over extended lower extremities. The STS movement also can be described using kinematic or kinetic variables, with definitions supplied for phases and events during this movement.[72-74] A definition of these phases that is used frequently is the one provided by Schenkman et al.[73] and is marked by 4 events. Phase I (flexion-momentum phase) starts with initiation of the movement and ends just before the buttocks are lifted from the seat of the chair. Phase II (momentum-transfer phase) begins as the buttocks are lifted and ends when maximal ankle dorsiflexion is achieved. Phase III (extension phase) is initiated just after maximum ankle dorsiflexion and ends when the hips first cease to extend; including leg and trunk extension. Phase IV (stabilization phase) begins after hip extension is reached and ends when all motion associated with stabilization is completed.[73]

Studying the STS movement, in our opinion, requires a basic knowledge of the factors influencing how the movement is performed. The determinants, we believe, should be independent from the techniques used to study movement. The extent of these determinants' influence can be small and detected only when using specific measurement or research techniques (eg, moments assessed by force plates). Knowledge of the determinants, we contend, is necessary in order to conduct research on the STS movement or to interpret results of reported studies, because the results can be, in part, a function of a determinant.

The STS movement has been studied using standardized clinical tests, which are used in epidemiological studies and clinical testing.[36,59-61,75-79] Measurements of aspects of the STS movement have been obtained using techniques such as use of force plates,[72] video analysis,[70,80-82] use of optoelectronic systems,[11,67,68,83-86] goniometry,[65,87] and accelerometry.[88]

Because the most recent review on the STS movement was published in 1991,[89] we believed an update was necessary to gain insight into studies on the effects of variables on the STS movement, especially in view of the new technology available to study the movement. The aims of our article are to review research on STS movement determinants and to describe the type and magnitude of their influence on the STS movement. In addition, we aimed to expose gaps in the literature and make recommendations for future research.

Methods

A search was made using Medline (1980–2001) and the Science Citation Index Expanded of the Institute for Scientific Information (1988–2001) using the key words “chair”, “mobility”, “rising”, “sit-to-stand”, and “standing”. References such as textbooks, presentations, and reports also were included. After reading the articles or abstracts, studies were included only when quantitative instrumental analyzing techniques were used to study STS movement performance in the subjects (patients and people without known impairments). The studies in this review were included on the basis of their design (ie, the design had to be experimental and aimed at elucidating the effect of determinants on the STS movement by manipulating the variables). Thus, descriptive and comparative studies were excluded, but because we included textbooks, presentations, and similar materials, there was not a requirement that articles be peer reviewed.

The STS movement determinants are factors that influence how the movement is performed. We categorized the studied determinants as chair related (eg, seat height), subject related (eg, age, muscle force), or strategy related (eg, speed or light conditions) (Table 1). Strategy-related determinants are those that are related to the execution of the STS movement. Although subject-related determinants can be investigated only by means of comparative studies, which was beyond the scope of our study, the types of patients investigated are indicated in Table 2. We judged studies according to the techniques used (eg, use of force plates, optoelectronic devices, or goniometers), number of movements analyzed, the determinants studied (ie, chair related, subject related, or strategy related), and the dependent variables (Table 2).

Table 1. *Number of Experiments Performed in the 39 Reviewed Studies Investigating Determinants of the Sit-to-Stand Movement*[a]

Chair-Related Determinants	n	Subject-Related Determinants	n	Strategy-Related Determinants	n
1. Height of chair seat	12	1. Age	0	1. Speed	11
2. With armrests	5	2. Disease[b]	0	2. Foot position	5
3. Chair special type	3	3. Muscle Force	0	3. Trunk position/movement	3
4. With backrest	0	4. No footwear	0	4. Arm use with armrest	5
				5. Terminal constraint	1
				6. Arm movement	1
				7. Dark versus light	2
				8. “Fixed” joints	1
				9. Knee position	1
				10. Attention	0
				11. Training	1

[a]In some studies, more than one determinant was investigated. The constrained determinants are indicated in Table 2 (numbers in columns under “Determinant Constrained” heading in Table 2 refer to the details of determinants listed in Tab. 1). [b]eg, stroke, arthritis, low back pain

STS Movement Determinants in the Reviewed Studies

Of the 160 studies identified, we found 39 studies that addressed the effects of determinants on the STS movement using an experimental design (Table 2). We did not examine whether the results could be obtained consistently by multiple researchers (ie, we did not examine reliability of these judgments).

Chair-Related Determinants

The literature indicates that the chair has an influence on the performance of the STS movement (eg, the height of the seat can make an STS movement impossible).[12] Most research has been focused on the height of the seat, and few studies tried to clarify the influence of the armrest position, use of armrests, or the type of chair on the STS movement.

Seat height.
Lowering the height of the seat makes the STS movement more demanding or even unsuccessful according to the literature we reviewed.[4,12,65,68,82,90-92] The minimum height for successful rising for elderly people (community-dwelling and nursing home residents 64–105 years of age) with chair rise difficulties appears to be 120% of lower leg length.[4] A lower seat apparently leads to increased angular velocity of the hip in order to stand[12,68,82,91] and to more repositioning of the feet (also called the “stabilization strategy”).[12,68] In young subjects (25–36 years of age) without impairments, lowering the seat of the chair from 115% to 65% of knee height results in

Table 2. *Details of the Experimental Studies That Addressed Determinants of the Sit-to-Stand (STS) Movement*[a]

Authors	Year Published	Study: Technique	Study: Rep	Subjects: N	Subjects: Type	Age (y): X	Age (y): Range	Determinant Constrained[b]: Chair Related	Determinant Constrained[b]: Subject Related	Determinant Constrained[b]: Strategy Related	Determinant	"Dependent" Variables
Alexander et al.[95]	1991	Video (3 samples in total) Handle dynamometry Chair instrumented	3(1)	51	17 y 23 o 11 o/d	23.2 72.4 84.4	19-31 63-86 75-92	1, 2, 4	4	2, 4	With/without arm use	Movement time, kinematic data, phase duration, hand forces
Arborelius et al.[93]	1992	Force plate (feet) Video (digitized) sEMG Borg Scale	2	9	9 m	26	23-34	2		4	3 chair heights (kh, kh +1/3 upper leg, kh +2/3 upper leg) Armrest	Difficulty Borg scale estimation of effort, load bearing seat, joint moments, sEMG
Burdett et al.[90]	1985	Force plate Cinecamera (digitized)	1	14	10 m 4 m/d	33.3 52	25-41 19-67	2, 3	4	2, 4	2 types chair (with/without arm use)	Kinematic data, joint moments
Carr[106]	1992	Force plate (feet) Video (digitized)	6	6	6 m		20-30	1		2, 4, 6	3 arm movement conditions	Movement time, COM position and horizontal and vertical linear momentum, support moment, angular displacement ankle joint
Doorenbosch et al.[98]	1994	Force plate (feet) Cinefilm, motion analyzer sEMG	5(2)	9	3 m, 6 f	27		1		1, 2, 4	Trunk movement (natural versus full flexion)	Movement time, kinematic data, joint moments, sEMG
Fleckenstein et al.[108]	1988	Cinecamera (digitized)	1	10	5 m, 5 f	25.4		1	4	4, 9	2 knee angles (75º, 105º)	Movement time, kinematic data, joint moments, phase duration
Goulart and Valls-Sole[88]	1999	Accelerometry sEMG Chair switch	5	20	12 m, 8 f	34.7	25-45	1		2, 3, 4	6 types STS (trunk straight, flexion of trunk, feet anterior, knees first, head supported, reference)	EMG phases

Authors	Year Published	Study: Technique	Study: Rep	Subjects: N	Subjects: Type	Age (y): X	Age (y): Range	Determinant Constrained[b]: Chair Related	Determinant Constrained[b]: Subject Related	Determinant Constrained[b]: Strategy Related	Determinant	"Dependent" Variables
Gross et al.[99]	1998	Force plates (2, feet/chair) Motion analysis system sEMG Dynamometry 1 repetition maximum	3 (1)	38	12 f/y 26 f/e	24.2 70.1	 64-84	1		1, 4	2 speeds (normal, as fast as possible)	Movement time, moments, torque hip/knee, ground reaction forces, kinematics
Hanke et al.[81]	1995	Force plates (2, feet/chair) 2 video	5	19	9 m 10 f	32.4 31.1	25-38 27-36	1	1, 2, 4		3 speeds (normal, as slow and as fast as possible)	Movement time, COM momentum, phase duration
Hesse et al.[102]	1996	Force plates (2, feet/chair)	5	20	9 m, 11 f	27.8	19-40	1	2, 4		3 speeds (natural, as slow and as fast as possible)	Movement time, COM displacement
Hesse et al.[109]	1998	Force plates (2, feet/chair)	15	35	17 m/h 18 f/h	64.8	59-79	1		2	Pretraining/post training	Movement time, time seat- off, body weight distribution, COM velocity/displacement
Hughes and Schenkman[11]	1996	Force plate (feet) Motion analysis system	1	18	18 o/d	74.8		1	4	4	2 chair heights (kh and lowest possible)	Movement time, hip flexion velocity, COM/ base of support separation at lift-off
Hughes et al.[12]	1994	1 video (digitized)	1	22	22 o/a/d	72.0	64-105	4	4	2, 4	6 chair heights (43.2–55.9 cm with an interval of 2.5 cm	Movement time, COM movement/velocity
Hughes et al.[85]	1996	Force plate (feet) Motion analysis system Dynamometer	1	21	5 m, 5f /y 5 m, 6f /od	25 78				2, 4	3 chair heights (58 cm, kh and lowest possible)	Joint moments, isometric quadriceps femoris muscle strength

Authors	Year Published	Study Technique	Rep	Subjects N	Type	Age (y) X	Range	Determinant Constrained[b] Chair Related	Subject Related	Strategy Related	Determinant	"Dependent" Variables
Itokazu et al.[87]	1998	Force plate Chair switch Goniometers (3)	3	46	tka 16 oa 30 art	ncs		1			Kneeflexion <100º and >100º 2 chair heights (100%, 120% of kh)	Movement time, hip/knee flexion angle and angular velocity
Kawagoe et al.[104]	2000	Force plates (2, feet) Motion analysis system sEMG	5	10	10 m	30.2		1		2	3 chair heights (30, 40, 50 cm) 3 foot positions	Movement time, temporal data, kinematics, center-of-gravity position, ground reaction force
Khemlani et al.[114]	1999	Force plate (foot right) sEMG Video Pressure-sensitive chair switch	6	9	9 m	29	20-35	1		2, 3, 4	2 foot positions (anterior/posterior)	Movement time, EMG phasing, extension moments, temporal data
Kotake et al.[74]	1993	Motion analysis system Isometric dynamometer	2	12	12 m	30.7	22-40	1	4	1, 2, 4	3 speeds (natural, fast, slow)	Movement time, kinematic data, phase duration, joint moments
Mourey et al.[97]	1998	Force plate (feet) Motion analysis system Cameras (3)	3	13	7 y 5 o	22.8 73.2	20-25 71-82	1, 4		4, 7	Young vs elderly, dark and light, 2 speeds (normal, fast)	Movement time, kinematic data, head stability
Mourey et al.[103]	2000	Force plate (feet) Motion analysis system Cameras (2)	3	7	4 m/y, 3 f/y 7 o	22.8 75.1	20-25 71-82	1, 4		3, 4, 7	Young vs elderly, dark and light, 2 speeds (normal, fast)	Movement time, kinematic data
Munro et al.[82]	1998	Force plate (feet) Video Arm rest load cell	3 (1)	12	12 o/f/ra.	65.5		2, 3, 4		3, 4	High (54 cm) vs low (45 cm) chair seat Eject vs noneject use	Movement time, kinematics, armrest force, exertion (Borg scale), pain (VAS)

Authors	Year Published	Study: Technique	Study: Rep	Subjects: N	Subjects: Type	Age (y): X	Age (y): Range	Determinant Constrained[b]: Chair Related	Determinant Constrained[b]: Subject Related	Determinant Constrained[b]: Strategy Related	Determinant	"Dependent" Variables
Munton et al.[65]	1984	Cinecamera Goniometer sEMG	1	9	5 4 art	ncs				2, 4	2 chair seat heights (42 and 59.5 cm) With/without arm use 2 foot positions (normal, posterior)	sEMG pattern
Pai and Lee[86]	1994	Force plate (stool) Motion analysis system	5	9	4 m, 5 f		27-39	1	4	5	3 speeds (natural, as fast as possible, fast+"fall")	Movement time, temporal parameters, COM displacement, moments
Pai and Rogers[67]	1991	Force plates (2, feet/chair) Motion analysis system	5	8	4 m, 4 f		26-38	1		1, 4	3 speeds (natural, slow, fast)	Movement time, kinematic data, joint moments
Pai and Rogers[83]	1990	Force plates (2, feet/chair) Motion analysis system	5	10	5 m, 5 f		26-38	1		1, 4	3 speeds (natural, slow, fast)	Movement time, COM movement, impulse momentum
Pai et al.[84]	1994	Force plates (2, feet/chair) WATSMART LED motion analysis system	5(3)	32	8 m/y, 8 f/y 8 m/o, 8 f/o	31.9 72.1	25-38 63-84	1		1, 4	3 speeds (natural, slow, fast)	Movement time, COM momentum, COM movement, phase duration
Papa and Cappozzo[100]	1999	Force plate (feet/chair) Load cell	5	12	6 m, 6 f		22-34	1		1, 2, 3, 4	2 speed (self-selected/maximal)	Movement time, kinematic data, rotational/linear actuator
Papa and Cappozzo[101]	2000	Force plate (feet/chair) Load cell	5	51	7 m/y, 9 f/y 12 m/o, 23 f/o		22-34 65-81	1		1, 2, 3, 4	2 speed (self-selected/maximal)	Movement time, kinematic data, rotational/linear actuator

Authors	Year Published	Study: Technique	Study: Rep	Subjects: N	Subjects: Type	Age (y): X	Age (y): Range	Determinant Constrained[b]: Chair Related	Determinant Constrained[b]: Subject Related	Determinant Constrained[b]: Strategy Related	Determinant	"Dependent" Variables
Rodosky et al.[91]	1989	Force plate Motion analysis system	2left, 2right	10	5 m, 5 f	25.5	20-35	4		2, 4	4 chair heights (65%, 80%, 100%, 115% of kh)	Kinematic data, joint moments
Schenkman et al.[68]	1996	Force plates (2, feet) Motion analysis system	2	21	11 y 10 o	28.9 67.3	25-36 61-79		4	1, 2, 4	4 chair heights (65%, 80%, 100%, and 115% of kh)	Kinematic data, phase duration
Scholz and Schoner[107]	1999	Motion analysis system Cameras (2)	10	9	5 m, 4 f		22-28	1		2, 3, 4, 8	3 conditions (normal/ rigid boots/ narrow base)	COM, head, hand trajectory
Seedhom and Terayama[94]	1976	Force plate (feet) Cinecamera UV oscillograph	2	2	ncs	ncs		1, 2, 4			With/without arm use	Moments, force (quadriceps femoris, hamstring, calf muscles)
Shepherd and Koh[70]	1996	Force plates (feet) Pressure switch Video	ncs	6	6 f	21.3	18-25	1		2, 4	3 foot positions	Movement time, kinematic data, joint moments, moment of support
Shepherd and Gentile[80]	1994	Force plate (stool) Video Chair-switch	6	6	6 m		20-30	1	4	2, 4	3 trunk positions	Kinematic data, phase duration, joint moments, moment of support
Stevens et al.[105]	1989	Force plate Photographic 1 sEMG	8	2	ncs	ncs		1		2, 4	Leg position guided/ unguided	Ground reaction forces, head movement, sEMG pattern
Su et al.[92]	1998	Force plates (2 feet) Motion analysis system	ncs	38	12 tka 4 m, 8 f 12 oh 5m, 6f 14 oa, 2 m, 12 f	64.8 61.7 ncs	57-75 54-75			4	4 chair heights (65%, 80%, 100%, 115% kh)	Movement time, kinematic data, COM displacement, joint flexion moments

Authors	Year Published	Study Technique	Study Rep	Subjects N	Subjects Type	Age (y) X	Age (y) Range	Determinant Constrained[b] Chair Related	Determinant Constrained[b] Subject Related	Determinant Constrained[b] Strategy Related	Determinant	"Dependent" Variables
Vander Linden et al.[1]	1994	Force plate Motion analysis system sEMG	5(4)	8	1m, 7f	68.8	61-77	1		1, 2, 4	2 speeds (self-selected, fast) 2 feet positions (5º or 18º dorsiflexion),	Movement time, kinematic data, phase duration, ground reaction forces, sEMG pattern
Weiner et al.[4]	1993	Motion analysis system VAS	1	22	o/a/d	ncs		4	4	2, 4	6 chair heights (17-22 in, interval 1 in)	Kinematic data, difficulty (VAS), STS success ratio
Wheeler et al.[96]	1985	Video 1 manual (1 sample) Electrogoniometer EMG	3	20	10 f/ y 10 f/o	24 75	22-28 67-81	1,4	4		2 types of chair	Movement time, kinematic data, sEMG pattern, hand/ foot placement

[a]Rep=repetitions (number analyzed in parentheses), m=male, f=female, y=young, o=old, a=able, d=disabled, h=hemiplegia, art=arthritis, ra=rheumatoid arthritis, oa=osteoarthritis, tka=total-knee arthroplasty, ncs=not clearly stated, sEMG=surface electromyography, kh=knee height, COM=center of mass, VAS=visual analog scale, WATSMART=Waterloo Spatial Motion Analysis Recording Technique, LED=light-emittingdiode; UV=ultraviolet.

[b]Numbers refer to determinants listed in Table 1.

an increase in trunk flexion angular velocity of almost 100% in order to stand.[68] A lower seat has been shown to increase trunk, knee, and ankle angular displacement.[82,91,92] Changing the seat height affects the maximum moment needed at the hip and knee.[11,91-93] Differences for hip and knee moments can be as large as 50% to 60%, with seat height having a greater influence on the moments needed at the knee than at the hip.[11,91-93] The changes in seat height can result in changing biomechanical demands (eg, the need to move the body's center of mass over a larger distance) or in an altered strategy (eg, "stabilization strategy", due to the imposed biomechanical demands by a different foot, trunk, or arm position).

Armrests.

Issues related to the armrest use include positioning of the hands on the armrests, height of the armrests, and the moments exerted. There is no research on the relationship among the height of the armrests, seat height, hand positioning, and their cumulative effect on performance of the STS movement.

Using armrests, according to the articles we reviewed, results in lower moments at knee and hip; at the hip, a reduction of about 50% of the extension moment needed to perform the STS movement has been calculated.[90,93,94] Burdett et al.[90] found no influence of the use of arms on joint angles in subjects without impairments (25–41 years of age). In a study by Alexander et al.,[95] young and old subjects without impairments used a hand bar positioned in front of them to perform the STS movement. They found no differences in body segment rotations in the young subjects (19–31 years of age). A difference in trunk rotation was observed in the old subjects (63–86 years of age), although this movement was analyzed only at the moment of maximum anterior head displacement.[95]

Chair type.

We found only 3 studies on the influence of specially designed chairs.[82,90,96] Different types of chairs designed to "ease" the STS movement were studied.[82,90,96] Wheeler et al.[96] suggested a negative influence of seat posterior slant because of tilting the body's center of mass farther backward. Use of an ejector mechanism lowered vertical impulses applied to the armrests by 47% in patients with arthritis, but no differences were found for knee and ankle moments.[82]

Backrests.

We found no experimental studies concerning the influence of backrests on STS movement. In only 8 studies[4,12,82,91,94-97] was a chair with a backrest used. When a backrest was used, it was to standardize the STS movement starting position. The influence of trunk position has been studied; however, this influence cannot necessarily be related to backrest use or backrest position, because the trunk position studied was not comparable to the trunk position using a backrest.[80]

Strategy-Related Determinants

Speed.

Increasing speed of the STS movement increases the hip flexion, knee extension, and ankle dorsiflexion joint moments.[67] To increase reproducibility and comparability of the results of their studies, some authors[68,71,98] did not allow subjects to rise at their self-selected speeds. Subjects had to rise at a preset speed indicated by, for example, a metronome.[71] Other researchers studied the influence of speed on strategy, peak joint moment, phase changes, and lateral displacement. Pai and colleagues[83,84] reported that a faster STS movement influences the peak vertical momentum of the center of mass while the peak horizontal momentum remains relatively unchanged (data were given in graphs). A faster STS movement gave a shorter flexion and momentum-transfer phase.[1,81] Vander Linden and colleagues[1] reported no influence of speed on joint excursions. Gross et al.[99] and Papa and Cappozzo,[100,101] however, described less hip flexion at the moment of seat-off in elderly subjects who stood rapidly. In several studies[97,99,102,103] elderly subjects (64 – 84 years of age) were less able to increase the speed of their STS movement.

Foot positioning.

Shepherd and colleagues[70] studied the effect of foot position (posterior, preferred, and anterior positions) prior to the start of the STS movement, and they showed a shorter movement time with feet placed posterior. With the posterior placement of the feet, hip flexion and hip flexion speed were lowered, whereas anterior placement of the feet increased the pre-extension phase.[70] Kawagoe et al.[104] also showed an influence of posterior foot placement. Positioning the feet more posteriorly enabled lower maximum mean extension moments at the hip (148.8 N·m versus 32.7 N·m) to be used for the STS movement.[104] Hughes et al.[12] described repositioning of the feet as a movement strategy to lower moments used for the STS movement, which they called "stabilization strategy". Munton et al.[65] found no difference in electromyographic (EMG) activity of 6 large lower-extremity muscle groups with feet placed normal or posterior. Stevens et al.[105] studied the effect of the initial lower-extremity posture, including foot posture, on the STS movement and reported that the preferred lower-extremity position gives less head movement and lower ground reaction forces.

Trunk positioning/movement.

According to Shepherd and Gentile,[80] changing the initial trunk position to have more flexion did not change the peak support moment, but the duration of maximum support moment did increase. The duration of the extension phase also became longer when the trunk initially was more flexed.[80] Starting from a trunk position different from erect alters the kinematics and kinetics of the STS movement. For the condition “flexion of the trunk” (first flex the trunk toward the knees, before rising from the chair), Goulart and Valls-Sole[88] described a longer movement time than for normal STS movement condition and delayed seat-off, without joint angular changes. This observation was supported by Schenkman et al.,[73] who described a momentum transfer strategy in which the momentum generated by the upper body is used during the extension phase.

Doorenbosch et al.[98] studied the effect of an STS strategy aimed at maximum flexion of the trunk during the STS movement. This strategy resulted in kinematic changes around the hip, but the range of motion of the knee and ankle did not change. Using the maximum flexion strategy, 27% lower (net) knee joint moments than in natural rising were found.[98]

Arm movement.

Study of the STS movement is often done with constraints on the use of the arms.[106] In most studies, use of the arms during the STS movement was not allowed. Subjects were often instructed to stand up with their hands in their lap, folded, sideways, placed on the knees, or fixating an object. Some authors[69,96] have reported that use of the arms during the STS movement is very common among elderly people and even among young people. Only Carr[106] studied the effect of arm movement strategy on the body’s center of mass. Arm position during the STS movement appears, based on the literature, to influence the position of the body’s center of mass.[106] The body’s center of mass moves forward at the end of the STS movement when subjects point with their arms.[106] Restricting the arms leads to a different pattern of ankle angular displacement, with a much higher mean standard deviation than occurs with the arms free. This finding suggests that more adjustment of the strategy of rising is needed, using ongoing adjustment at the ankle joint during restricted arm movement.[106]

Terminal constraint.

The terminal constraint is the required body position or activity at the end of the STS movement. The STS movement has been studied while the motion was aimed at standing quietly at the end of the movement. Pai and Lee[86] conducted a study with a constraint to fall after the movement instead of standing quietly at the end. No study has quantitatively explored the sit-to-walk movement.

Dark versus light.
Visual control was manipulated while subjects performed the STS movement in light and darkness at 2 speeds.[97,103] No effect on movement time was found in young (20–25 years of age) and elderly (71–82 years of age) people when visual control was varied.[97,103] The speed of the center of mass, however, was lower in the blindfolded condition for the elderly subjects.[103]

Fixed joints.
Only one study[107] concerned the influence of joint fixation on the level of control of STS movement performance using the so-called “uncontrolled manifold concept” (a cybernetic concept to describe results). This analysis showed that the position of the center of mass in the sagittal plane is controlled. No data on joint angle or angular velocity were given. Another study[87] analyzed the relationship between the active limitation in range of motion of the knee following total knee arthroplasty and the height of the seat when rising from a seated position. The subjects with larger limitations in active knee flexion (<100° of knee flexion) required a higher angular velocity of the hip to lift the trunk forward than did those with less limitation of knee flexion (>100° of knee flexion).[87]

Knee position.
Positioning the knee in more extension than preferred prior to the STS movement appeared to lead to an increase of the hip joint angular displacement, with an increase of hip extension moments of 77%.[108] This experimental setup is to some extent comparable to the foot -forward setup as used by Shepherd and Koh[70] because foot forward will result in more knee extension.

Attention.
No experimental study addressing the influence of attention on the performance of the STS movement in subjects without impairments could be found.

Training.
Training can be a determinant in an experimental study. Hesse et al.[109] studied the influence of 4 weeks training (4-week inpatient rehabilitation program; the physical therapists trained the patients to distribute equal weight on both legs and to avoid lateral compensatory tilt of the trunk) on the temporal and spatial variables of the STS movement. Only in a subgroup of people with left hemiparetic strokes was a difference noted.

Discussion and Conclusions.

Method

General.
In our review, we included only experimental studies. In an experimental study of the STS movement, the determinants are manipulated in order to explore their influence on performance. Not all of the studies reviewed, however, were completely experimental. Some articles included comparative or descriptive data. We believe that experimental studies are important because they provide the strongest evidence concerning the influence of the determinants. In these studies, only one determinant is usually manipulated while others are kept constant. In comparative studies, we believe conclusions are difficult to make because of the nonexperimental design. The relationship between subject-related determinants (eg, age, muscle force) and STS movement performance, in our view, is seldom unambiguous because subject-related determinants are generally examined in nonexperimental studies. For example, the influence of age on the ability to do an STS movement is often studied,[95,97,101,103,110-112] with age accounting for small differences in STS movement performance and a decreased ability to decrease movement time. Whether these differences in the STS movement are the result of increased age or of covariates such as muscle force, balance disturbances, neuromusculoskeletal changes, or changed motor control is not clear. Another example concerns muscular force as a determinant of STS movement performance. Less quadriceps femoris muscle force will affect the performance of the STS movement, and the time to do the STS movement will increase.[85,99,110,113] Related neuromusculoskeletal changes (eg, loss of trunk muscle force, loss of balance) may influence the performance of the STS movement to the same degree. When these related changes cannot be controlled for in a study, they can become confounding factors influencing the conclusions to be drawn from these studies.

Validity.
Our review of studies on the determinants of what makes the STS movement possible led us to believe that many studies have good internal validity, but we did not use evaluative criteria or examination by multiple authors. There is, in our view, also evidence for construct validity for the measures used, because clinical tests for STS movement performance appear to us to be highly correlated with physical functioning in elderly people.[60,61] We question, however, whether the reviewed studies are externally valid for predicting changes in standing up. Standing up from a chair is almost never aimed at standing alone but is part of a goal-oriented behavior, such as going for a walk or picking up an object. Nevertheless, there are examples in which standing up is aimed at simple standing (eg, in church, watching sports).

Variability.

There is intrasubject and intersubject variability in the performance of the STS movement. Variability can be the result of problems in defining the STS movement events, technical problems, or analysis of a low number of STS movements, or it can be considered as a sign of flexibility of performance during the STS movement. To lower variability and to ease analysis of the determinants, many constraints were used in the STS movement studies that we reviewed (Table 2). We contend that only in clinical physical performance is testing of the natural STS movement imitated (with self-selected speed and strategy).[15,36,75] Other explanations for variability may include a learning effect during performance of the STS movement, fatigue in repeating fast and frequent STS movements, and erroneous instructions leading to misinterpretation.

General conclusions

In our review, we found that in most studies (27 of the 39 studies), a combination of force plate(s) and a motion analysis system (varying from video to a type of optoelectronic system) was used. Surface EMG analysis was used in 10 of the 39 studies. The number of analyzed STS movements per subject ranged from 1 to 15. In 7 of the 39 studies, only one trial was used for statistical analysis. The number of subjects studied ranged from 2 to 51. We believe, however, that general conclusions can be drawn. The height of the chair seat, the use of armrests, and foot positioning has a major influence on STS movement performance. A higher chair seat results in lower moments at hip and knee level (up to 60% and 50%, respectively).[12,65,68,82,90-93] Lowering the chair seat will increase the need for generation of momentum or repositioning of the feet to lower the moments needed.[68] Comparison of the results of the studies is difficult because of differences in study design and the fact that chair seat height is not always based on lower-extremity length. Using armrests will lower the moments needed at the knee by 50%, probably without influencing the range of motion of the joints.[90,93,95] There were no reports on the interaction between the height of the armrests, chair seat height, or hand positioning and their cumulative effect on STS movement performance. Repositioning of feet appears to influence the STS movement strategy, enabling lower peak moments at the hip and knee.[1,12,70,104,114] No experimental study was found that addressed the influence of the use of a backrest. The influence of trunk position has been studied; however, trunk position cannot be related to backrest position, because the studied trunk position is not comparable to the trunk position using a backrest.[80]

Clinical Significance

The ability to perform an STS movement is an important skill. In elderly people, the inability to perform this basic skill can lead to institutionalization, impaired ADL functioning, and impaired mobility.[60,61] Consequently, this movement is frequently assessed in clinical practice. Knowledge of determinants of the STS movement, therefore, is important for clinicians interested in evaluating the ability to do an STS movement. For a proper evaluation of the STS movement in a clinical setting, we contend that standardization of the evaluation should be done in regard to type of chair, chair seat height, positioning of feet, and the use of armrests. Results of experimental studies show that these variables influence the performance of the STS movement. Neglecting these variables may result in an inability to measure actual changes in STS movement performance of a patient. Furthermore, problems in STS movement performance may be obscured without standardization. Another consequence may be that apparent changes or discrepancies may not actually be present. All of these factors can lead to suboptimal choices and decisions with respect to prognosis, planning, and therapy.

Recommendations

We believe that in both experimental and comparative STS movement studies, there needs to be control of variables that can influence STS movement performance. Some determinants (eg, chair seat height, speed, position of feet) have been studied extensively. Others (eg, the effects of footwear on STS movement performance) have not been well studied (although the footwear type does influence the performance of the Timed Up & Go Test[115]). The interaction among determinants has been studied to some extent.[1,82,97,103,104] More research is needed, however, on the interaction of variables such as use of armrests, chair seat height, and foot positioning.

All of the studies we examined were directed at the level of impairment. Studying functional performance, in our opinion, should also include testing at the level of skills.[32,116] To analyze the skill of a subject to perform the STS movement, it may be necessary to evaluate the abilities of that subject to cope with changing constraints (eg, STS movement at different speeds, at different chair seat heights, STS movement versus sit-to-walk movement, light versus darkness). To gain insight into the influence of the determinants on the STS movement may entail using other biomechanical models or paradigms.[32,117] New techniques (eg, ambulatory techniques that register body posture and movements in the real-life environment of the subject) raise new research questions. To enhance the validity of data obtained in future studies and the generalizability of the results, new methods of research (which can be used outside the gait laboratory[43]), we believe, should be evaluated.

3

Analysis and decomposition of accelerometric signals of trunk and thigh obtained during the sit-to-stand movement

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Abstract

Background and Purpose

Piezoresistive accelerometer signals are frequently used in movement analysis. However, their use and interpretation are complicated by the fact that the signal is composed of different acceleration components. The aim of the study was to obtain insight into the components of accelerometer signals from the trunk and thigh segments during four different sit-to-stand (STS) movements *(self-selected, slow, fast, and full-flexion).*

Methods

Nine subjects performed at least six trials of each type of STS movement. Accelerometer signals from the trunk and thigh in the sagittal direction were decomposed using kinematic data obtained from an opto-electronic device. Each acceleration signal was decomposed into gravitational and inertial components, and the inertial component of the trunk was subsequently decomposed into rotational and translational components.

Results

The accelerometer signals could be reliably reconstructed: mean normalised Root Mean Square (rms) trunk 6.5% (range 3-12 %), mean rms thigh 3% (range 2-5%). The accelerometric signals were highly characteristic and repeatable. The influence of the inertial component was significant, especially on the timing of the specific event of maximum trunk flexion in the accelerometer signal. The effect of inertia was larger in the trunk signal than in the thigh signal and increased with higher speeds.

Discussion and Conclusion

The study provides insight into the acceleration signal, its components and the influence of the type of STS movement, and supports its use in STS movement analysis.

Introduction

Knowledge of a patient's movement behaviour is relevant for rehabilitation to evaluate and guide rehabilitative interventions. The Sit-to-Stand (STS) movement is important because it is a prerequisite for the initiation of gait and postural changes. The STS movement is a change of posture normally consisting of three major and distinctive components: (1) flexion and (2) extension of the trunk, and (3) extension of the leg. Phase definitions of the STS movement vary owing to differences in the techniques used,[72,73,118] Schenkman et al. proposed three phases: flexion momentum, momentum transfer and extension momentum.[73] As a result of the extension of the legs the trunk is also moved forward and upwards. The STS movement at self-selected speed usually takes 2-3 seconds[1] and can be influenced by several factors, for example chair height and disorders.[11,93,103,119]

Quantitative research into the STS movement is performed using techniques such as force-plates, opto-electronic devices etc.[119] These techniques can only be used in a movement laboratory with its associated limitations, e.g. a limited generalisability to actual daily behaviour.[33] To overcome these problems an activity monitor (AM) has been developed and validated.[43,46-49,120] The configuration used consists of four piezoresistive accelerometers, attached to the trunk and thighs, and a portable data recorder. This instrument was initially aimed at long-term automatic detection of body postures and motions and proved to be valid with respect to this detection.[43,48,121] Accelerometry also has the potential to determine *how* an activity is performed, as is shown in studies on walking and balance control.[52,54,122-124]

Thus it can be postulated that ambulatory accelerometry is potentially feasible for quantification of the way the STS movement is performed, enabling assessment of the STS movement during daily life. An advantage of the AM would be the combination of prolonged detection of mobility-related activities, such as the number of STS movements, and the quantification of, for example, the duration of the STS performance. However, the interpretation of the accelerometer signals of the AM during movement is only partially obvious and is hampered by the intrinsic character of piezoresistive accelerometers. A piezoresistive accelerometer signal has a combination of components consisting of gravitational and inertial acceleration that cannot be distinguished in the signal itself.[125] This inertial acceleration is frequently named movement acceleration.[125] The description and decomposition of the signal have been provided for walking.[122,126] Before the AM and its obtained signals are used in the assessment of the STS movement, further knowledge about the accelerometer signals during the STS movement is mandatory: the characteristics of the signal, such as amplitude, timing and shape must be understood before they can be used to characterise the STS movement (i.e. temporal parameters). Defining temporal events and parameters will only be possible after the accelerometry signal has been studied in more detail and will be the second part of the validation of this technique.

The aim of this study was to acquire insight into the accelerometer signal and its components during different types of STS movement.

Methods

Subjects

Nine subjects participated in this study; all were healthy with no history of mobility limitations (mean age 29.9 years (range 21-43), four female and five male).

Protocol

All measurements were performed in a movement laboratory. The subject sat on a stool without a back rest, sitting upright, then rose with hands in lap, the foot position self-selected. The height of the chair was normalised to the knee height (measured from the lateral knee joint space to the ground, with the subject wearing shoes). The subjects performed STS movements using four conditions: self-selected speed, slow speed, fast speed and with full flexion of the trunk. Full flexion consists of moving the trunk in an exaggerated, flexed way before rising. This STS movement was added because this often forms part of STS movement training in patients with a stroke. Prior to each condition, subjects were allowed to practise the specific condition by performing at least three trials. Rising was performed without using the hands (hands in lap). The registration duration ranged from 60 to 80 s for registration of at least six STS movements per condition. Between each condition, a 5 min pause was allowed to prevent fatigue.

Instruments

Piezoresistive accelerometers

The Activity Monitor uses piezoresistive accelerometers attached to the thigh and trunk. Piezoresistive accelerometers consist of a mass connected to a frame by beams that can be represented by a damped spring. In the beams piezoresistors are mounted, forming a bridge circuit. The value of the resistors depends on the deformation of the beams, which depends on the magnitude of acceleration. As a consequence of this construction, these accelerometers are only sensitive to accelerations in line with their sensitive axis. If the sensor does not move, there is a signal that depends on

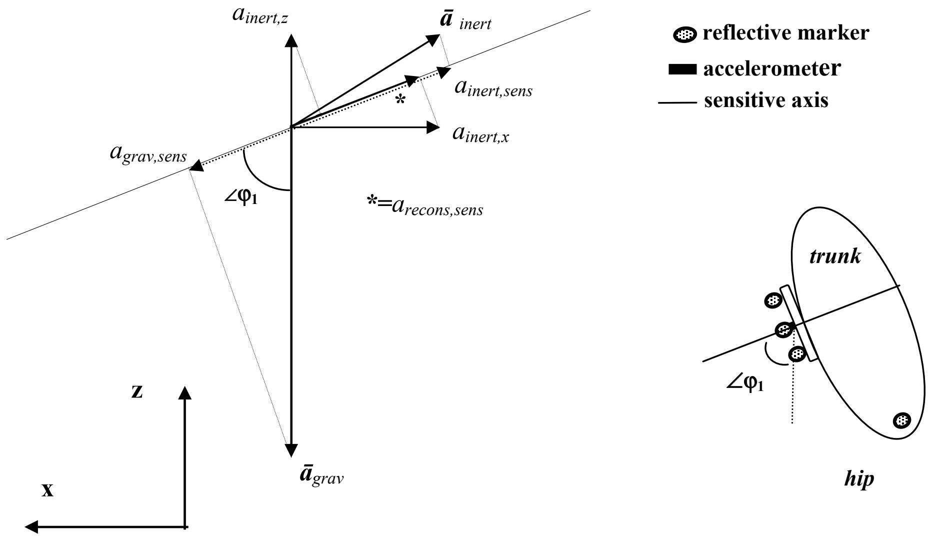

Figure 1 *Measurement set-up and graphical representation of components of sensed gravitational and inertial acceleration.*

the gravitational acceleration $\bar{a}_{grav}$ exerted on the mass. The part of the gravitational acceleration that is measured in the sensitive axis, $a_{grav,sens}$ depends on the angle ϕ_1 between the sensitive axis of the accelerometer and gravitational acceleration (see Fig. 1). If the accelerometer is moved also an inertial acceleration $\bar{a}_{inert}$ occurs. Like the gravitational acceleration, the part of the measured inertial acceleration that is measured, $a_{inert,sens}$ depends on the angle between the sensitive axis of the accelerometer and the inertial acceleration (see Fig. 1). The total measured accelerometer signal is the sum of $a_{grav,sens}$ and $a_{inert,sens}$. Assuming rigid segments, $\bar{a}_{inert}$ is the resultant of translational ($\bar{a}_{trans}$) and rotational ($\bar{a}_{rot}$) acceleration of the segment.

The subjects were instrumented with accelerometers and reflective markers for simultaneous accelerometer and kinematic recordings. Both recordings were synchronised using a photo-flash. The sensor and marker set-up was derived from the standard AM sensor configuration.[45] This AM configuration consists of a sensor on each thigh, halfway between the trochanter major and the lateral knee joint, which is sensitive in the sagittal direction during standing. The other two AM sensors are attached to the skin of the sternum, perpendicular to one another: during standing one sensor is sensitive in the sagittal direction and one is sensitive in the longitudinal direction. Because of the sagittally directed sensitive axis the sagittal trunk sensor is more sensitive to angular changes of the trunk during the STS movement than the longitudinal sensor. In the present study we only analysed the sagittal accelerometer signal.

To create two rigid segments we used two aluminium frames (trunk: 2.8 cm x 13 cm and thigh: 2.5 cm x 16 cm) on which the accelerometers and three reflective markers were fixed. We recorded the kinematic and accelerometric data of two segments, the trunk and thigh segments. The reflective markers were placed proximally and distally on the two frames and on top of the accelerometers which were attached centrally on the aluminium frame (see Fig. 1). The markers were placed proximally and distally at maximum distance to minimise possible errors in the calculation of the angular change of the frame. One reflective marker (marker$_{hip}$) was placed at the trochanter major. The frame was attached to the thigh and trunk with adhesive tape in such a way that the accelerometers were located as in the standard AM set-up as described above.[45] A Vitaport Recorder[a] placed in a waist belt recorded data with a sample frequency of 128 Hz. This recorder also recorded the flashes for synchronisation (using a photocell fixed on the left shoulder). The motion of the reflective markers was recorded using a Proreflex infrared 3D three-camera system[b] with sample frequency of 128 Hz. In the co-ordination frame used, the xz-plane was the sagittal plane in which the STS movement was analysed.

A calibration measurement was performed using one STS movement with prolonged standing for 10 s. This measurement was needed to correct for possible attachment error of the accelerometer on the frame.

The data files were stored on a personal computer. The Proreflex data were tracked and the markers were identified. The resulting files were exported in ASCII format for calculations in Matlab. To reveal the relevant parts of the Vitaport data these data were shortened using the flash signal for detection of the STS movement and exported in ASCII format. Positional data and accelerometric signals were filtered using a second order low-pass Butterworth filter with 6 Hz cutoff frequency. The Proreflex and Vitaport data were merged using the flash synchronisation. Kinematic data were calculated based on positional data of the Proreflex data. In the presented calculation model it is assumed that the sensitive axis of the accelerometer is perpendicular to the frame (see Fig. 1). To correct for a possible attachment error of the accelerometer on the frame, a correction angle (the angle between the line perpendicular to the accelerometer's sensitive axis and the line between the reflective markers on the frame) was calculated. This correction angle was used in the subsequent calculations of angles and accelerations.

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a. TEMEC

b. QUALYSIS

Normalisation of STS movement.

To enable analysis of six trials of STS movement per condition, we performed a normalisation procedure for the duration of the STS movement for the trunk and thigh segment separately based on the kinematic data of the frames. The start and end of the STS movement were detected using a threshold of an angular velocity of 5 degrees/s. for both the trunk and thigh segment.[99] To normalise the data, the durations of the trials were resampled towards the fastest trial. Mean data for angular change of frame and acceleration of the markers for six trials of STS movement were calculated.

Decomposition

As mentioned before, the measured acceleration signal consists of gravitational and inertial components. We decomposed the accelerations exerted on the accelerometers using the reflective markers fixed on the frames and the accelerometers. The proximal and distal markers were used to calculate angle ϕ_1 between the sensitive axis of the accelerometer and the direction of the gravitational acceleration, after correction for differences in angular attachment of the accelerometer to the frame (see Fig. 1).

The gravitational component in the sensitive axis $a_{grav,sens}$ was calculated according to

$$a_{grav,sens} = -9.81 \cdot \cos(\phi_1) \tag{1}$$

By double differentiation of the position data of the reflective marker placed on the accelerometer we calculated the acceleration in the horizontal ($a_{inert,x}$) and vertical ($a_{inert,z}$) directions. This allowed the calculation of the inertial component in the sensitive axis according to

$$a_{inert,sens} = a_{inert,x} \cdot \sin(\phi_1) - a_{inert,z} \cdot \cos(\phi_1) \tag{2}$$

By addition of the contributions of the gravitational and inertial accelerations in the sensitive axis, the reconstructed acceleration signal $a_{recons,sens}$ could be calculated, which should be equal to the measured acceleration signal $a_{meas,sens}$.

$$a_{recons,sens} = a_{grav,sens} + a_{inert,sens} \tag{3}$$

All calculations were the same for the trunk and the thigh segments.

Additionally, the inertial component of the trunk signal $a_{inert,sens}$ was further decomposed. The inertial acceleration of a body segment-fixed accelerometer can be regarded as a result of a translational acceleration by translational movement of the body segment $a_{transl,sens}$, and a rotational acceleration of that body segment by rotation around a chosen point of rotation, i.e. the hip ($a_{rot,sens}$).

The translational acceleration $a_{transl,sens}$ was calculated according to

$$a_{transl,sens} = a_{hip,x} \cdot sin\ (\phi_1) - a_{hip,z} \cdot cos\ (\phi_1) \tag{4}$$

The contribution of the rotational acceleration $a_{rot,sens}$ to $a_{inert,sens}$ was estimated by

$$a_{rot,sens} = a_{inert,sens} \cdot a_{transl,sens} \tag{5}$$

RMS

To study the similarity of a_{meas} and a_{recons}, these two signals were visually examined, but also the rms (= normalised RMS, square root of the mean squared difference) was calculated according to

$$rms = RMS/ \surd(\ \sum(\ a_{meas,sens}(i)^2)/N) \tag{6}$$

in which

$$RMS = \surd(\ \sum(\ a_{meas,sens}(i) - a_{recons,sens}(i))^2/N) \tag{7}$$

where $a_{meas,sens}(i)$ is the measured acceleration for the ith sample; $a_{recons.sens}(i)$ is the reconstructed acceleration for the ith sample; and N is the number of samples. Normalised RMS is dimensionless and expressed as a percentage.

The graphical representations of the decomposition were studied and, if indicated, additional quantitative analysis determining intervals between significant events in the different signals was performed.

Statistical analysis

Differences for the different conditions were tested for significance (significance level 0.05) using a non-parametric test (paired samples Wilcoxon test). We used a non-parametric test because of the small sample size and because we did not know whether the data had a normal distribution.

Results

General

Using the criterion of angular velocity of five deg s^{-1} for the start and end of the segment movements (which also was also used for normalisation of the STS movement), we measured distinctive trunk and thigh movement times for the different conditions. Mean trunk movement time ranged from 1.40 to 3.77 s, and mean thigh movement time ranged from 1.09 to 2.59 s (see Table 1). The trunk angle range differed significantly between the different conditions, except for the fast as against self-selected condition which just failed to reach significance (p = 0.051). Concerning the graphical representations all data are from one representative subject, with Fig. 2 giving separate STS movements, and Figs. 3-5 giving mean data of six trials.

Table 1 *Duration of trunk and thigh movements per condition (determined with opto-electronic device)*

Condition	trunk duration mean (sd) s	thigh duration mean (sd) s	trunk angle range mean (sd) °
self-selected	2.08 (0.47)	1.56 (0.32)	35 (7)
slow	3.77 (0.48)	2.59 (0.20)	45 (8)
fast	1.40 (0.18)	1.09 (0.13)	30 (8)
full-flexion	2.87 (0.57)	1.94 (0.35)	62 (12)

All differed significantly except trunk angle fast as against self-selected (p = 0.05 Wilcoxon test)

Measured accelerometer signal

Fig. 2 shows representative examples of the accelerometer signals of the trunk and thigh segments for one STS movement per condition for one of the subjects.

The accelerometer signal of the trunk always showed a distinct negative peak t_3 related to the flexion movement of the trunk. The presence of the positive peak of the signal at t_2 depended on the condition and subject, but was always present for the condition of self-selected and fast speeds. For the conditions of slow rising and fullflexion t_2 was not present or had low amplitude. In these cases t_1 could be defined as a visible change in signal amplitude. The trunk signal reached a stable state at t_4. Furthermore, the amplitudes of the trunk accelerometer signal differed for the distinct conditions.

The thigh signal changed at l_1, with a positive peak l_2 the presence of which depended on condition and subject, but was always present in the condition of self-selected and fast speeds. In the conditions of slow rising and fullflexion in general l_2 was present more than t_2. The signal showed a sharp decline, to stabilise at l_3. The moment of occurrence of l_3 and t_4 differed as a consequence of differences in the movement pattern of the trunk and thigh segments.

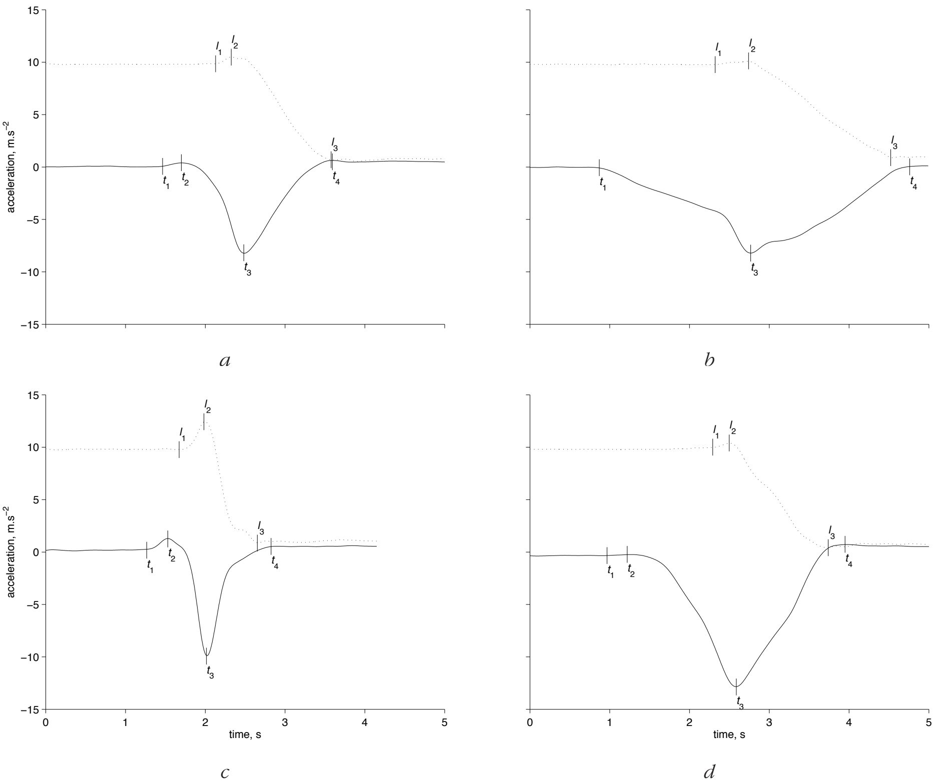

Figure 2 *Accelerometer signals during STS movement, (a) self-selected speed, (b) slow speed, (c) fast speed, (d) fullflexion condition. One trial, sample frequency 128 Hz, low-pass filter 6 Hz; (—) trunk; (...) thigh; l: thigh events; t: trunk event. Data for 1 representative subject*

The level of output from the thigh accelerometer was different during sitting and standing owing to the extension of the thigh segment. The trunk accelerometer signal showed only a small difference between sitting and standing owing to a small difference in the trunk position in the sitting and standing positions.

For the different conditions the same patterns were seen, although fast rising gave rise to a more pronounced positive peak in both signals. Clearly visible was the difference in time needed for the different conditions.

Accelerometer signal decomposition

Figs. 3 and 4 show representative examples of measured and reconstructed accelerometric signals, after the normalisation and averaging of the signals of one subject. A clear relationship between the reconstructed and the measured accelerometer signals was seen visually and quantitatively, although agreement was lower for the

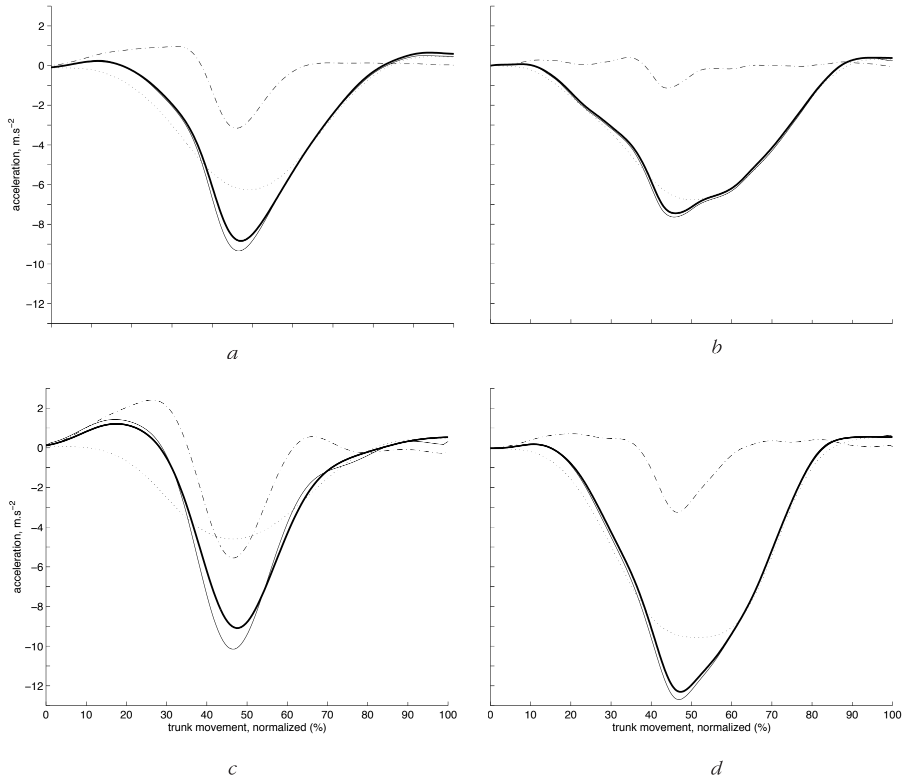

Figure 3 *Curves of measured and reconstructed accelerometric signals of the trunk; mean of 6 STS movements, (a) Self-selected speed; (b) slow speed; (c) fast speed; (d) fullflexion.*
(—) a_{meas}*;(—)* $_{recons}$*; (. . .)* $a_{grav,sens}$*; (- · - · -)* $a_{inert,sens}$ *Trunk movement normalized using angular velocity (see Section 2). Reconstructed signal is composed of gravitational and inertial components. Data from 1 representative subject.*

fast condition (see Figs. 3 and 4 and Table 2). Normalized root mean square for the trunk accelerometer signals significantly differed in the fast condition from the other three conditions; for the thigh accelerometer signal this difference was not present.

For the measured *trunk* accelerometer signal $a_{meas,sens}$ there was a clear, but not exact, relationship with the trunk flexion angle (Figs. 3 and 4). The gravitational acceleration component $a_{grav,sens}$ formed the main component of the measured trunk accelerometer signal. The inertial acceleration component was most pronounced in the fast STS movement condition, giving rise to the specified positive peak t_2 in the measured signal during the initial part of the STS movement. The mean interval between the occurrence of t_3 in the reconstructed signal and maximum trunk flexion ranged from 0.02 s. (fast) to 0.12 s. (slow); for the occurrence of t_3 in the reconstructed signal and the negative peak of inertial acceleration, it ranged from 0.003 s. (fast) to 0.04 s. (slow). Thus the moment of occurrence of t_3 in the

Table 2 *Normalised rms (and range) for trunk and thigh acceleration*

	trunk *rms*		thigh *rms*	
	mean, %	range, %	mean, %	range, %
Measured against calculated, self-selected	7	2 - 19	3	1 - 6
Measured against calculated, slow	3	2 - 6	2	1 - 5
Measured against calculated, fast	12	4 - 21	5	2 - 6
Measured against calculated, fullflexion	4	3 - 11	2	1 - 5

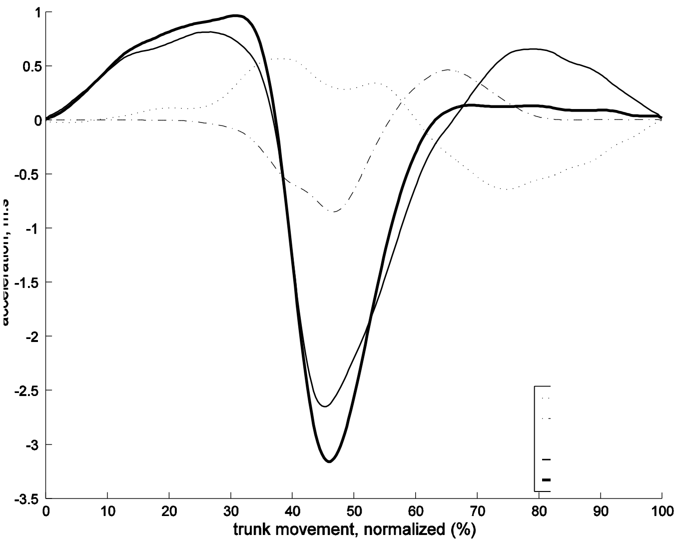

Figure 4 *Acceleration of trunk decomposed into different components during STS movement at normal speed (based on kinematic data). Trunk translational acceleration decomposed into two components x and z; all accelerations given for sensitive axis of accelerometer. Data are from same subject as Fig. 3.*
(· · ·) $a_{transl,sens,x}$*; (- · - · -)* $a_{transl,sens,z}$ *; (—)* $a_{rot,sens}$ *; (**—**)* $a_{inert,sens}$

reconstructed signal related somewhat better to the moment of maximum inertial acceleration than the moment of maximum trunk flexion. Further decomposition of the inertial component of the trunk signal (an example is shown in Fig. 4) showed that the amplitude of $a_{inert,sens}$ was considerably larger than that of $a_{transl,sens}$. This showed that $a_{inert,sens}$ is predominantly determined by $a_{rot,sens}$ and does not result from the translational movement of the trunk segment. This decomposition is comparable for the different conditions.

The measured *thigh* accelerometer signal was predominantly determined by the gravitational component. The inertial component became clearly visible in fast rising at t_2 (Fig. 2 and 5). This inertial component was also to a certain extent visible in the self-selected and full-flexion condition.

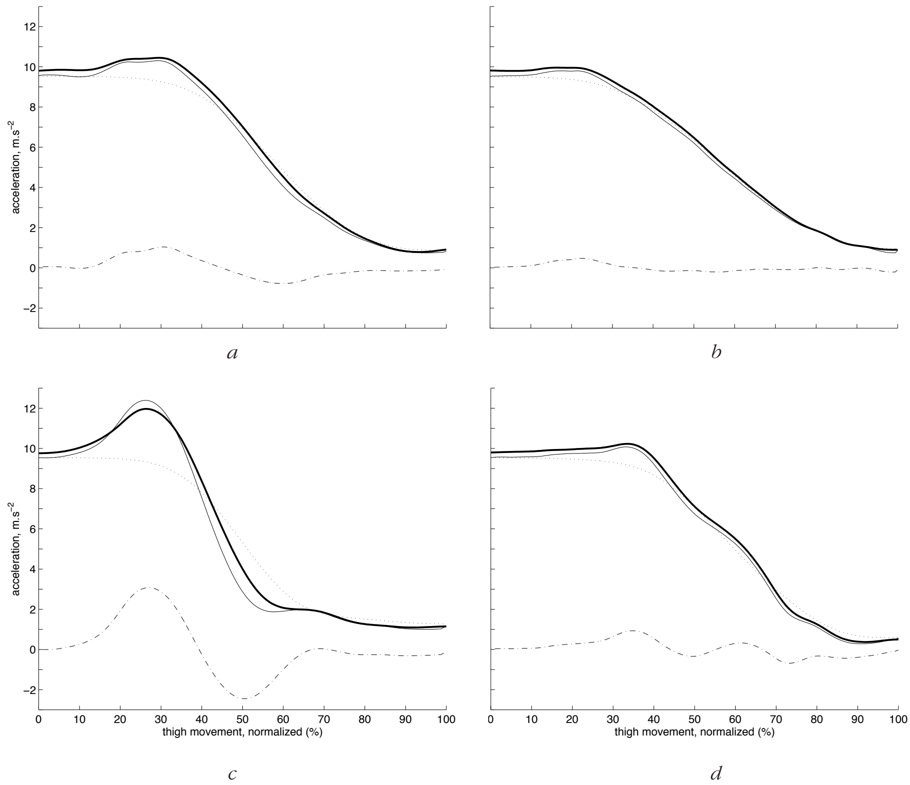

Figure 5 *Curves of measured and reconstructed accelerometric signals of thigh; mean of 6 STS movements, (a) Self-selected speed; (b) slow speed; (c) fast speed; (d) fullflexion.* **(—)** a_{meas}; (—) a_{recons}; (· · ·) $a_{grav,sens}$; (- · - · -) $a_{inert,sens}$ *Thigh movement normalised using angular velocity (see Section 2). Reconstructed signal is composed of gravitational and inertial components. Data from 1 representative subject.*

Discussion

The temporal parameters obtained for the duration of the STS movement are in agreement with the results of others.[1,127] The four conditions differed for temporal data and for trunk angular movement, with the lowest trunk angular excursion occurring for fast rising (see Table 1). Vander Linden et al. and Mourey et al. found no influence of speed on total angular excursion for the hip, knee and ankle.[1,97] Gross et al. however noted a more upright trunk at lift-off during fast speed trials.[99] We have to realise that we reported on trunk angular excursion, a composite angular excursion, whereas others reported on angular excursion of the separate joints. Because the STS movement is difficult to standardise we normalised the STS movement using the kinematic data for the trunk and thigh segment movements. This enabled us to calculate the mean of six STS movements and reduce the inter-trial variability of the STS movement.

In the present study we used a 6 Hz low pass-filtered accelerometric signal, because in future we want to determine specific events during the STS movement. Pilot measurements with the use of several cut-off frequencies showed this frequency to be adequate for that purpose. Although this cutoff frequency does change the measured accelerometric signal, the effect was shown to be small, owing to the relatively low movement frequency during the sit-to-stand movement. Therefore the results of the present study are not significantly affected by the chosen filtering procedures.

The accelerometric signals showed a similar pattern for the different conditions, although some changes are apparent. First of all, the temporal aspect of the STS movement changed, giving rise to another presentation of the signal, when not normalised, with steeper slopes (see Figs. 2a-d). Secondly, the presence of the initial peaks t_2 and l_2 is different for the distinct conditions, with a clear presence of these peaks in the fast and self-selected speed conditions. Thirdly, there is a difference of amplitudes with higher amplitudes in fast rising. Despite these changes the shape of the signals was comparable for the different conditions with the exception of the initial peaks t_2 and l_2. In all conditions we saw the leg signal starting to change (l_1) after t_2 and prior to t_3; this is in accordance with the momentum generation as reported by Kralj and colleagues, with the seat unloading beginning after sufficient momentum has been generated by trunk movement.[72,73]

Decomposition of the accelerometric data showed clearly that the measured and reconstructed accelerometer signals are closely related. The low values of normalised rms are an indication of the validity of the methods chosen in this study. Compared with similar studies on accelerometric measurements of gait our rms values are low.[122,126] An explanation for this is that in gait higher levels of inertial acceleration and more complex movement patterns regarding acceleration and deceleration are present. This explanation can also be used to explain the difference in rms between the different conditions and the trunk and the thigh segments. There can be several causes of the small discrepancy between the reconstructed and measured accelerometric signals. First of all, for the calculation of the reconstructed signal we made the assumption that all movement occurred in the sagittal plane. This assumption can be incorrect because of the changes in the total movement during the changing of speed and movement pattern, giving rise to movement in the other planes combined with rotational components. Secondly, we assumed the sensitive axis of the accelerometer to be stable during all experiments; however, we have to realise that angular changes (e.g. 1°) in the sensitive axis can give rise to significant changes in the amplitude of the accelerometer signal. The impact of the angular change in the sensitive axis of the accelerometer depends on the position of the accelerometer; the effect is most pronounced when the sensitive axis is parallel to the ground, so the trunk sensor is the most sensitive for this technical aspect. Thirdly, we performed one calibration per subject to correct for a possible discrepancy in the direction of the sensitive axis and the frame angle. Owing to movement of the frame or accelerometer this could have changed giving rise to a larger discrepancy. The low rms for the slow condition can be explained by the very low inertial component in this condition. The measured accelerometric signal is primarily determined by sensed gravitational acceleration; see Figs. 3b and 4b.

The signals of the accelerometers attached to the trunk and thigh segments are composed of inertial and gravitational components. Decomposition of the acceleration as performed clearly showed the relationship between the distinct components. For the trunk accelerometer the main contribution to the amplitude of the acceleration signal is from the gravitational component, except for the fast condition; in the fast condition, there is an increase in the inertial component and a decrease in the gravitational component (Fig. 3c). Therefore the acceleration signal of the trunk produces information on the moment of maximum inertial acceleration of the trunk combined with the amount of trunk segment flexion. The inertial component influenced the shape of the total reconstructed signal. First of all it contributed to the shape of t_2 and t_3 and, secondly, the moment of t_3 is determined by the moment of the negative peak of $a_{inert,sens}$. The moment of peak t_3 in the reconstructed signal is slightly better related to the moment of the maximum inertial component than to the moment of the maximum gravitational component. However, the difference in timing of these two moments is small. The inertial component is primarily determined by the rotational acceleration of the trunk segment. The moment and magnitude of the minimum of $a_{inert,sens}$ are determined by the rotational acceleration of the trunk segment (see Fig. 4). Peak t_2 is also determined by the rotational component of the trunk segment, although the amplitude and moment of occurrence of the peak are not as closely related as in peak t_3. In the thigh signal the inertial component was low compared with the gravitational component and contributed clearly in the fast condition (see Figs. 5a-d). As in the trunk, the inertial component also contributed to the shape of the signal giving rise to the peak l_2. The contribution of the gravitational component to the reconstructed leg segment signal was almost comparable for the four conditions. This is owing to the fact that the angular change of position of the thigh segment is comparable for the four conditions, contrary to the trunk segment in which the angular change is related to the condition. The inertial component can be masked by the gravitational component. In the fast-moving condition, peak l_2 was very prominent reflecting the magnitude of the inertial acceleration and because of the fact that the sensitive axis of the accelerometer is almost parallel to the direction of the vertical acceleration of the thigh segment.

This study showed which information is represented in the accelerometric signals for different types of STS movement. The timing and, to a lesser extent, the magnitude of the two components (gravitational and inertial), showed a fixed pattern that is present in the different conditions. Therefore accelerometer signals obtained from the trunk and the thigh segments during the STS movement give information concerning the temporal aspects of the performance of the Sit-to-Stand movement with some information on the magnitude of the inertial acceleration of the trunk and the thigh, together with information on the timing of maximum inertial acceleration of the trunk. Although we have to be aware that these signals contain two components, this technique will enable us to acquire information on the performance of the Sit-to-Stand performance that was unobtainable until now.

Conclusions

This study showed that the accelerometric signal contains information on kinematic events that will enable us to define time markers to describe the phasing and duration of the STS movement without the use of a gait laboratory. The low rms we found for slow conditions warrants application in clinical studies because of the slowing of the STS movements generally noticed in patients. There is now need for validation of time markers based on accelerometric signals and kinematic data. Together with the possibility of gathering data on the performance of the STS movement during unsupervised activities, this technique can provide us with relevant information on the motor behaviour of patients. This information can be combined with the information gathered by the Activity Monitor, such as the duration of mobility-related activities and the number of STS movements performed.[43,45] These data will enable us to evaluate changes, such as recovery and the effect of rehabilitative measures, in more detail.

4

Validity of accelerometry in assessing the duration of the sit-to-stand movement

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Abstract

Background and Purpose

Accelerometry is frequently used in movement analysis to assess body postures and motions. Here, we assessed the validity of ambulatory accelerometric measurement of the Sit-to-Stand (STS) movement duration.

Methods

We compared accelerometric and opto-electronic assessment of the STS movement duration under four conditions (comfortable, slow, fast movement and exaggerated trunk flexion) with six healthy subjects and six subjects with stroke who performed movements six times under each condition.

Results

Accelerometric and opto-electronic data of STS movement duration were strongly related (r = 0.98). Accelerometry showed a fixed bias of 0.07 s (95% CI 0.01, 0.14) in healthy subjects and 0.32 s (95% CI 0.22, 0.42) in stroke subjects. In healthy subjects, a significant negative proportional bias of 0.1 was detected (95% CI -0.16, -0.03). Accelerometry showed discriminative validity in comparing stroke subjects to healthy subjects, and in comparing speed conditions.

Discussion and Conclusion

Our results indicate that accelerometry can provide valid data on the STS movement duration; during its use additional information on the STS movement, such as balance control, can be recorded.

Introduction

The Sit-to-Stand (STS) movement, which consists of flexion and extension of the trunk combined with extension of the legs,[1] is a prerequisite for standing and walking, and critical for daily activities. Loss or decline of this function (e.g., in stroke patients) leads to functional limitations in self-care,[128] walking,[129] and independent functioning.[27,130,131] Many studies focus on the STS movement, its determinants,[132] and manipulation of the STS movement in healthy subjects and in subjects with disabling diseases, such as stroke,[22,24] cerebral palsy[133] or Alzheimer's.[134]

STS parameters are relevant in assessing functional recovery from stroke and the effects of treatment. The parameters reflect temporal aspects; the length can be determined by the start and stop of the movement. Additional events, e.g., maximum trunk flexion, may allow the assessment of phases within the STS movement.[1,72,135] The duration of the STS movement has shown to be related to chair-rise task demand,[95,97,132] the extent of paresis in stroke patients,[24] and strength of lower extremities.[136,137] Additionally, STS movement duration in subjects with a stroke was predictive for gait speed and symmetry.[22]

Several instruments are used to assess temporal STS characteristics, with a stopwatch assessing repeated STS movements or the Get up & Go time being the most simple and cheapest.[8,137,138] The instruments can have limitations in the subjectivity and accuracy of assessment (observation, clinical timed tests, combined actions) and range from relatively simple[36,136,139] to complex instrumentation and data analysis (force plates, opto-electronic systems).[32] Accelerometry is a method of interest and can potentially provide objective and accurate data on the STS movement in a inexpensive, simple and ecologically valid way.[140] All instruments are prone to the question of ecological validity; does the movement performed and assessed in the gait lab represent the normal movement of a subject during daily life.[32,48]

Accelerometry is frequently used to provide parameters on events such as gait, balance and falls, and is also applied in studies on aspects of the STS movement. Accelerometry can provide an accurate and simple method to describe kinematics during rising,[58] walking,[57] and shows a strong relationship with trunk flexion kinematics during rising.[141-143] The characteristics and content of the acceleration signals in relation to kinematics during the STS movement have been studied for agreement and error sources, both theoretically and clinically.[141-143] However, the validity of accelerometry to assess spatio-temporal characteristics of the STS movement is yet to be determined. Assessing the STS movement time by performing a fixed number of the STS movements, using a stopwatch, can provide low cost information, but this method also includes the time needed for sitting down and testing can be hampered in frail subjects. Furthermore, accelerometric assessment could provide data on the number of STS movements during normal daily life,[45,47,48] as well as on the speed of rising in daily life and other aspects of the STS movement, such as balance control.[134]

The aim of this study was to determine the validity of accelerometry to assess STS movement duration in healthy subjects and subjects who had suffered from a stroke. We studied the relationship between the STS movement duration obtained by accelerometry compared to data obtained by an opto-electronic device (reference method) in healthy subjects and subjects with stroke. Different conditions for the STS movements and groups with varying STS performance allowed us to determine the discriminative power of accelerometry compared to the reference method.

Methods

Subjects

A total of twelve subjects were included in this cross-sectional explorative study. Six were healthy subjects without any history of musculoskeletal diseases, and six subjects had previously suffered from a stroke. The mean age of the healthy subjects was 29.9 years (range 21-43 years), and two were female and four were male. The mean age of the subjects with a stroke was 64.6 years (range 45-76 years), and five were male and one was female. None of the subjects was obese (BMI >30.0). Functional Ambulation Category was 4 in five subjects and 5 in one subject. Mean time post stroke was 33 months ranging from 8 to 75 months. The study was approved by the local Medical Ethical committee and all subjects signed an informed consent.

Protocol

To assess the STS movement, subjects wore shoes and sports clothing, sat in a chair adjusted at knee height with the hands in lap, and feet in any position. Chair height ranged from 46.5 cm to 52.0 cm (mean 49.6 cm) Four types of STS movements were performed: at comfortable, low, and fast speed, and with exaggerated trunk flexion. The instruction on exaggerated trunk flexion was to rise with exaggerated flexion of the trunk, bringing the center of the body mass further forward, and an example STS movement was made to show the procedure. After each subject performed three tryout movements in each condition, six STS movements were registered for detailed analysis. Accelerometry and a synchronized opto-electronic device were used for measurement.

Instruments

Accelerometers

Accelerometers (ADXL202, Analog Devices, adapted by TEMEC Instruments, range ± 3 g) were attached onto aluminum strips (to create rigid segments) on the sternum (2.8 x 13 cm) and the left leg (or non-paretic leg in subjects with a stroke; 2.5 x 16 cm) using the regular setup of the Activity Monitor.[48] Briefly, one accelerometer was attached to the lower part of the sternum and one to the lateral side of each thigh halfway between the trochanter major and lateral knee joint. The accelerometers were placed in such a position and direction that their sensitive axis was in the sagittal (i.e., anterior-posterior) direction while standing. Data were stored with a sample frequency of 128 Hz using a Vitaport 2™ system (TEMEC Instruments). After the measurement, data were stored on the computer and converted to ASCII files for further analysis with Matlab.

Opto-electronic device (hereafter referred to as video)

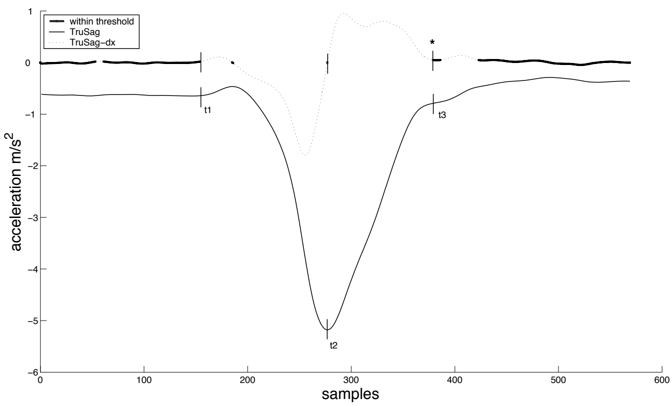
Opto-electronic measurements were performed with the ProReflex system (Qualisys AB, Gothenburg, Sweden; with a resolution of 1/30,000 of the view field used; Mac Reflex User Manual 3.2). Two reflective markers were applied to the sternum (proximal and distal on an aluminum strip) and two on the lateral leg (proximal and distal on the aluminum strip). Three cameras were used, and data were sampled at 128 Hz. In the healthy subjects, the left side of the body was recorded; in stroke subjects, the non-paretic side was recorded.

Data analysis and outcome measures

The analysis was based on the detection of the following events (Fig. 1): the start of trunk movement (t_1), the end of the trunk flexion phase (t_2), the end of trunk movement (t_3), the start of leg movement (l_1), and the end of the leg movement (l_2). The main parameter derived from these events was the *STS movement duration*: the length of time between the initiation of the STS movement (derived from t_1 or l_1) and the completion of the movement (derived from t_3 or l_2). The algorithm to determine the STS movement duration was optimized by several analyses on three healthy subjects and two subjects with a stroke. These subjects did not take part in this study.

Accelerometry

The accelerometer signals were processed by custom-made MatLab programs. Figure 1 shows an example of the acceleration signals of the trunk and leg during one STS movement of a subject with a stroke. After filtering (6 Hz second order low pass Butterworth filter), the derivative of the acceleration signals was calculated. First, t_2 was determined by the moment the derivative of the signal was 0. t_1 and t_3 were detected when the derivative passed the threshold of ± 0.05 m/s³ (see Fig. 1). The first instance prior to t_2 that the derivative passed the threshold was defined as t_1, and the first instance after t_2 when the derivative passed the threshold was defined as t_3. The automatic detection was visually checked (without knowledge of opto-electronic data). The MatLab algorithm allowed correction towards a prior or subsequent automatically determined point in time (see * in Fig. 1). For the leg, a comparable algorithm was defined to detect l_1 and l_2 (Fig. 1) using a comparable algorithm starting at the instance that the derivative was maximal.



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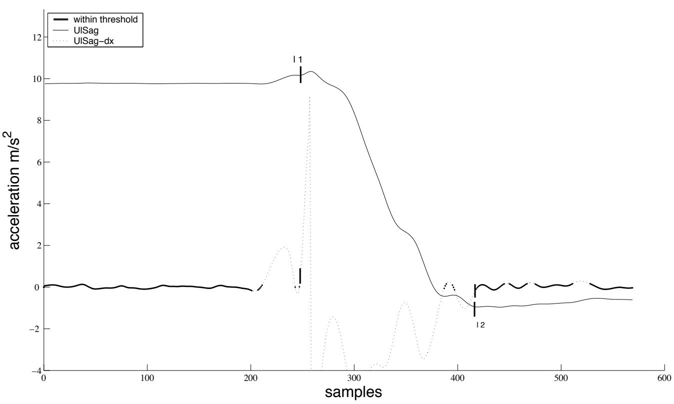


b

Figure 1 *a The trunk sagittal accelerometric signal with its derivative is shown. b The leg sagittal accelerometric signal with its derivative is shown. For reasons of representation the derivative signals are scaled to fit (factor 0.1). The detection of STS movement events are based on the defined algorithm in Matlab. Parts of the derivative signal that are within the threshold are marked bold. Automatic detection based on the algorithm is shown by an asterisk in the derivative signal with concomitant representation of that moment in the original signal for visual control. Sampling frequency is 128 Hz.*

Video

Qualysis Track Manager software (version 1.9.25 h; Qualysis AB) was used to determine trajectories of the reflective markers in the sagittal plane. After conversion to ASCII files, these data were analyzed by Matlab to calculate trunk and leg segment angular velocity relative to the horizontal plane. The initiation and completion of the movement of the trunk and leg segment were determined using the angular velocity of these segments with a threshold of 5° per second.

Statistical analysis

Statistical analysis was performed using the software package SPSS 12. The median of the temporal data of six STS movements was used to reduce the effect of outliers in the trials per subject. Data were analyzed for both subjects who suffered a stroke and healthy subjects. Data from accelerometry and video were compared by several approaches. First, for each condition, differences between accelerometry and video data were tested by a paired *t* test. Second, for the pooled data of each group (including the data of all conditions), the strength of the relationship between the two techniques was assessed by linear regression analysis, providing correlation coefficients and residual standard deviation (RSD). To test differences in intercepts and slopes between both groups, linear regression analysis was used. In this analysis, the patient data set was limited to data within the range of the healthy subjects; patient data outside the range were deleted before regression analysis. Third, Bland-Altman plots were created from the pooled data of each group. These data were tested for mean value of difference; if significantly different from 0, a fixed bias was present. The coefficient for the slope of the regression of differences on means of both techniques was tested; if significantly different from 0 a proportional bias for accelerometric assessment compared to the reference method was present.[144,145] To test the sensitivity to change we calculated the standardized response mean (SRM) for conditions of different rising speed for accelerometry and the reference method. For discriminative properties we used the *p* value from the *t* test for independent samples when comparing data of healthy and stroke subjects.

Results

We analyzed a total of 288 STS movements in twelve subjects. Data from six STS movements (by two subjects) were excluded because automatic detection failed for the ProReflex data. Using the reference method we could determine that in all STS movements the trunk moved earlier than the leg in all trials, and that the end of the movement was determined by the trunk in 48% and by the leg in 52% of the trials. Correction for assessment of STS movement initiation or completion, after visual check, was used in 5% of the STS movements for the start of the trunk and the end of leg movement, and in 15% for the end of the trunk movement.

The average duration of STS movements under all conditions was 2.50 (SD 0.95) s in healthy subjects and 3.36 (SD 1.58) s in stroke subjects as determined by video data. The standard deviation ranged from 0.11 to 0.64 s among conditions in healthy subjects, and 0.37 to 1.64 s in stroke subjects. Both groups could increase and decrease the speed of the STS movement.

Table 1 shows data on STS movement duration of both groups assessed for each condition separately and for the pooled data. Accelerometry overestimated STS movement duration in the pooled data of both groups, for the fast condition in healthy subjects, and the slow and exaggerated flexion condition in stroke subjects. The standard deviations of the two techniques were of the same magnitude (Table 1).

Table 1 *Data on STS movement duration using two assessment techniques in healthy subjects and subjects with stroke.*

		Healthy subjects			Stroke subjects		
	Condition	Video, s mean (SD)	Acc, s mean (SD)	*p*	Video, s mean (SD)	Acc, s mean (SD)	*p*
STS	Slow	3.60 (0.47)	3.66 (0.35)	0.386	5.01 (1.64)	5.37 (1.71)	0.003*
	Comfortable	2.15 (0.59)	2.25 (0.45)	0.25	2.72 (0.66)	3.05 (0.49)	0.07
	Fast	1.41 (0.11)	1.59 (0.10)	0.001*	1.92 (0.37)	2.16 (0.53)	0.065
	Exaggerated flexion	2.83 (0.64)	2.79 (0.57)	0.455	3.79 (1.31)	4.15 (1.46)	0.004*
	Pooled	2.50 (0.95)	2.57 (0.86)	0.030*	3.36 (1.58)	3.68 (1.65)	0.000*

*Video opto-electronic device, Acc accelerometry * Significant, $p < 0.05$, paired t test*

The relationships between the two assessment techniques for the pooled data for both groups are presented in Fig. 2. The coefficients of the slopes of the regression lines of healthy and stroke subjects did not differ significantly (1.09 and 1.05, respectively, $p = 0.64$), and neither did the intercept (-0.30 and -0.43, $p = 0.62$). After limiting the data range, the RSD changed from 0.23 to 0.20 (Table 2).

Bland-Altman plots (Fig. 3) show a fixed (0.07 s) and proportional (-0.10) bias in healthy subjects, and a fixed bias (0.32 s) in stroke subjects (Table 2).

The discriminative properties of both techniques for different speed conditions and type of subject group are indicated in Fig. 4. Under conditions of varying speeds, the SRM ranged from 1.59 to 4.79 for the accelerometric data, and from 1.36 to 2.70 for the video data (Table 3). Comparison of the groups revealed *p* values ranging from 0.008 to 0.029 for accelerometry, and 0.005 to 0.046 for video (Table 4). The video technique was non-discriminative for subject group in the comfortable condition, while accelerometry was discriminative in all conditions.

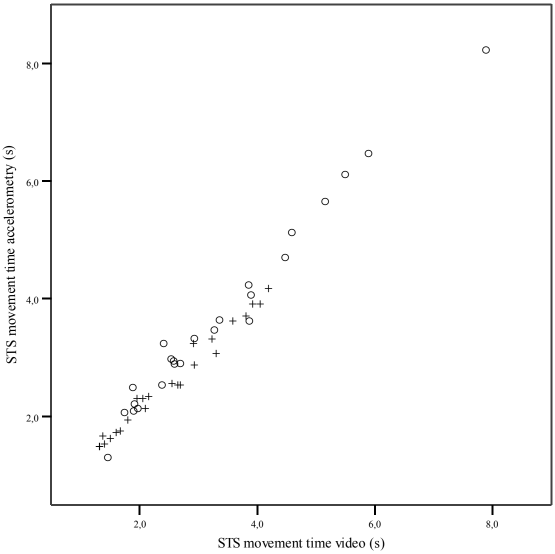


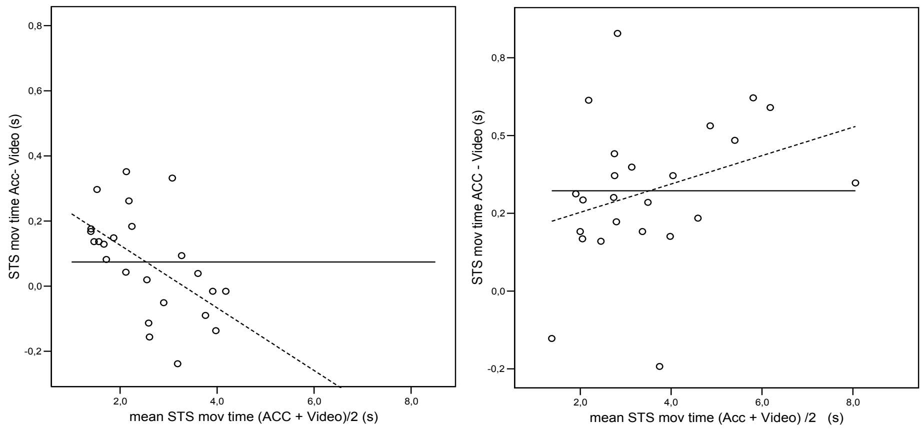
Figure 2 *Scatter plot of the STS movement duration assessed by accelerometric and video technique. Video opto-electronic device. Type subject: ○ stroke, + healthy*

Discussion

Overall, the results from our study support the validity of accelerometry to measure STS movement duration. The study was based in part on two assumptions: the STS movement duration differs depending on the speed of the movement, and differs between subjects who had suffered a stroke and control subjects. Both assumptions were shown to be valid. In this study, video was used as a reference method, as it is most frequently used and described in detail in STS movement studies. However, although we still feel that video can be a reference method, it must not be regarded as a gold standard due to present measuring error. Therefore, this must be taken into account while interpreting of the results of this study.

The STS movement durations for the healthy subjects in comfortable, fast and slow condition are in agreement with the results presented by Kotake et al.[74] and Hanke et al.,[81] but different from the data reported by Pai et al.[135] The assessed STS movement in comfortable conditions by subjects with stroke is shorter than that reported by Chou et al.[22] and longer than those reported by Hesse et al.,[109] which may be due to differences in study population and methods used. However, for the purposes of our study, durations of STS movement corresponding to previous studies are not essential.

The STS movement duration based on accelerometry was strongly related to the reference method ($r > 0.98$). However, in the pooled data we found a positive fixed



a *b*

Figure 3 *The Bland Altman plots for (a) the healthy subjects and (b) the subjects with stroke. Reference line for the mean of difference between the techniques is given. Video: opto-electronic device, Acc: accelerometry*

Table 2 *Data on relationship of the two assessment techniques for STS movement duration for pooled data.*

	r	*p*	RSD (s)	Mean difference (SEM)	95% CI	*p*	Coefficient BA plot	95% CI	*p*
Healthy	0.989	0.00	0.13	0.07 (.032)	0.008, 0.141	0.030	-0.096	-0.160, -0.032	0.005
Stroke	0.990	0.00	0.23	0.32 (.048)	0.223, 0.422	0.000	0.046	-0.016, 0.107	0.137

Mean value for the difference of techniques is given with 95% confidence interval (CI), and coefficient for slope of regression of differences on mean in BA plot is given with 95% CI; proportional bias is present if this coefficient differs significantly from 0

r multiple correlation coefficient, RSD residual standard deviation, SEM standard error of the mean

Table 3 *Data for discrimination of speed conditions with standardized response mean (SRM) in healthy and stroke subjects.*

	Slow	Fast
Accelerometry		
Healthy	3.71	1.59
Stroke	1.63	4.79
Video		
Healthy	2.70	1.36
Stroke	2.16	2.43

Standardized response means are given for conditions compared to the comfortable condition

Table 4 *Data for discrimination between types of subject.*

	Slow	Comfortable	Fast
Accelerometry			
Healthy versus stroke	0.029	0.008	0.014
Video			
Healthy versus stroke	0.046	0.070	0.005

p-value given using the t-test for independent samples

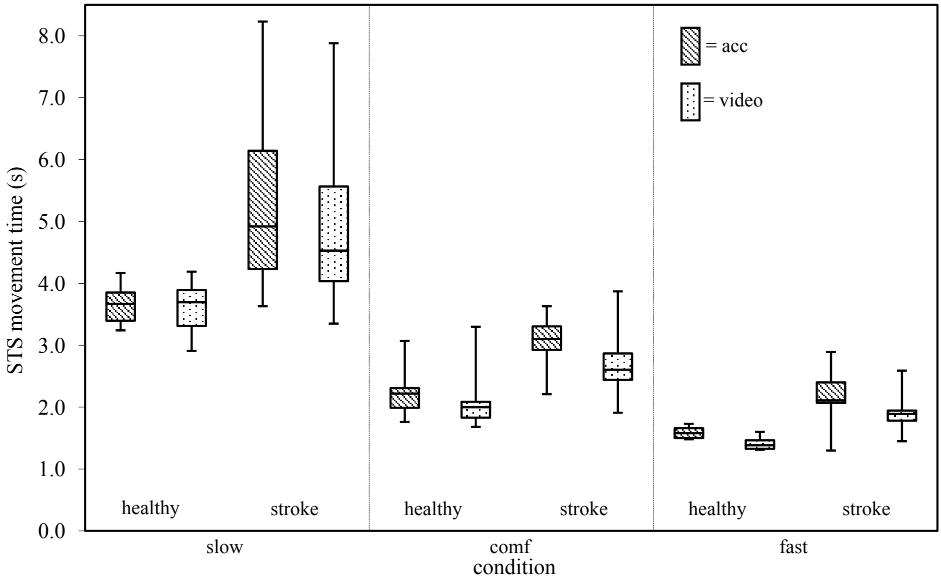


Figure 4 *Boxplot to show discriminatory properties of STS movement duration during various conditions using accelerometry and opto-electronic device in healthy subjects and subjects with a stroke. Video: opto-electronic device, acc: accelerometry*

bias of accelerometry in both groups (0.32 and 0.07 s for stroke and healthy subjects, respectively). Additionally, there was a negative proportional bias in the healthy group, resulting in an underestimation at lower rising speeds. No proportional bias was found in the subjects with stroke, although the difference between accelerometry and video tended to increase at lower rising speeds. Several explanations can be given for the fixed bias, which originated in the detection of both the initiation and completion of the STS movement. First, accelerometry and video fundamentally differ in the type of measurement technique, criteria of event detection, and method of data analysis. A main difference is that detection in accelerometry is based on the change in the acceleration signal, which contains components of both gravitational and inertial accelerations,[141] whereas the video system uses angular velocity as a criterion for detection. In a post hoc analysis we studied the background for the fixed bias in more detail, which showed that a complex of factors contributes to both fixed and proportional bias. For example, the mutual contribution of both acceleration components will depend on the rising speed; during fast rising, the inertial component will be more dominant than during slow rising, which could result in accelerometry detecting the start point earlier during fast rising.[141] However, this effect was only observed in healthy subjects. The fact that this effect was not found in stroke subjects could be due to a different range of STS rising speeds or to a different movement pattern. This may result in different patterns of accelerations and angular velocities at the start and the end of the STS movement.

Similarly, the presence of proportional bias could be due to the absence of consistent patterns, and the likely effects of different ranges of rising speeds and different methods of performance. Speed influences the magnitude of the components of the measured acceleration signals, with higher inertial acceleration at higher rising speed. As shown previously, healthy subjects move faster, so this phenomenon can be expected to be more prominent in this group; this is also suggested by the results presented in Fig. 3 that show an overestimation in fast rising of healthy subjects compared to slow rising. This effect was not as obvious in stroke subjects, and here this effect could have been hidden by the obvious fixed bias. Furthermore, the variance of performance of the STS movement in stroke subjects is higher (see RSD for both techniques in Table 2), which could have resulted in different patterns of inertial and gravitational accelerations per trial producing a different overall effect.

To estimate the presence of fixed and proportional bias we used the Bland-Altman technique[144] and not techniques such as Major Axis for line of best fit.[145,146] Every technique has advantages and disadvantages, and we are aware of the limitations of the Bland-Altman technique, such as not incorporating the measurement error in video data and our aim of describing the relationship between the two techniques without predicting STS movement durations by assessment of accelerometric data. However, we regarded the advantages of the Bland- Altman technique (e.g., clinicians are familiar with this technique) as more important. Furthermore, the presence and significance of fixed and proportional bias in this study requires further elaboration. In the present study, no complete agreement was found between video and accelerometry, as expressed by fixed and proportional bias, and we already discussed the background. The importance of these effects must also be discussed within the framework of clinical relevance. Information on the fixed and proportional bias is relevant when an exchange of assessment techniques is considered.[144,145] If only one technique is to be used in assessment, the information on bias will show aspects of the validity of a technique. If the discriminative properties increase due to a proportional bias, the arguments must be weighed in order to choose a measurement technique, which will depend on the aim of a particular study. The difference in fixed bias in the two subject groups will increase the possibility to discriminate between these groups. The proportional bias related to speed will increase the possibility to discriminate between different speeds. Despite the present fixed and proportional bias, we demonstrated a strong relationship between the results of accelerometry and an opto-electronic device for assessing the STS movement duration.

We studied the discriminative validity of accelerometry by two approaches: by the capability to discriminate between different speed conditions (“sensitivity for speed change”) using SRM, and by the capability to discriminate between healthy subjects and subjects with a stroke. With respect to the former approach, generally accelerometry showed good discriminative validity compared to video (see Table 3), although the results were not consistent over all conditions. This observation, however, could be due in part to the effect of the proportional bias, which in healthy subjects would increase the difference between different speed conditions. With respect to the discriminative power of accelerometry to distinguish between groups, *p* values based on a *t* test for independent samples were generally lower with accelerometry. This indicates that accelerometry is capable in discriminating groups with different STS performance. Again, however, here we have to realize that this could be due in part to the influence of fixed bias.

This study has some potential limitations. First, the number of subjects is small, which can result in a bias of the results. The limited number of subjects also limits the ability to generalize the results to other subgroups within the stroke population. However, the data on STS movement duration are consistent with results from other studies in healthy and stroke subjects despite differences in duration of the STS movement, and did not indicate that our group had an on average poor STS performance.[22,27,74,81,109,135] Secondly, the visual check component of the STS movement analysis may be considered; the initiation of the STS movement was corrected in 5% of all STS movements and the end of the STS movement in 10% of all STS movements, because the end of the STS movement was only determined by the trunk movement termination in 48% of the movements. This subjective component could potentially result in unreliability or inconsistency in the detection of the start and end of the STS movement. To diminish the influence of these corrections, the corrections were performed prior to the determination of the median of six STS movement trials. It is also important to note that these corrections implied a change of one automatically detected and selected moment to another automatically determined but not selected moment (see data analysis). Using the median of six movement trials additionally reduces the influence of the visual check. Therefore, we feel that the effect of the visual correction on the overall data of STS movement duration is small, although improvement of the automatic detection will also improve the clinical acceptance of the method.

The focus of the present study was the potential use of accelerometry in future ambulatory studies. Of course, accelerometry may also replace video systems in the movement lab to some extent, but accelerometry may be the basis of a simple, cheap, reliable, valid and precise instrument that allows prolonged quantitative measurements on STS movements outside a laboratory. Thus, not only could our system provide data on the number of STS movements during normal daily life,[45,47,48] but also on the speed of rising in daily life. Preliminary unpublished data show a considerable difference between movement behavior in a movement lab and movement behavior in daily life. Therefore, we are convinced that ambulatory STS research has the potential to increase the (ecological) validity of STS measurement.

Conclusions

STS movement duration assessment using accelerometry and an opto-electronic device give clearly related results in healthy subjects and subjects with a stroke, although the relationship is different for both groups concerning fixed and proportional bias. Accelerometry is at the least as equally discriminative as an opto-electronic device to distinguish different speeds in various groups.

5

Sensitivity of accelerometry to assess balance control during sit-to-stand movement

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In press

Abstract

Background and Purpose

Accelerometry has the potential to measure balance, defined as high frequency body sway, ambulatorily in a simple and inexpensive way. The aim of this study was to determine and compare the sensitivity of accelerometric balance parameters during the sit-to-stand (STS) movement.

Methods

Eleven healthy subjects (4 males, 28.2 ± 7.9 years) and 31 patients with stroke (21 males; 63.3 ± 12.8 years) were included. The healthy subjects performed STS movements in four conditions with different levels of difficulty. Data of the patients were compared a) with healthy subjects, b) between patient subgroups, c) between different phases of recovery to assess the sensitivity of accelerometry for differences in balance control. Accelerometers were attached to the trunk, and force plate measurements were simultaneously done in the healthy subjects. Main outcome measures were Root Mean Square (RMS) and Area Under the Curve (AUC) derived from the high frequency component of the transversal acceleration signal of the trunk.

Results

In all comparisons there was a significant difference in AUC data ($p < 0.05$), and AUC appeared to be more sensitive than RMS. Variability in AUC was not completely or mainly the result of changes and differences in the duration of the STS movement.

Conclusion

As a conclusion, accelerometry is a potentially valuable technique to measure balance during the STS movement.

Introduction

Rising from the seated to the standing position is a complex movement, that requires sufficient strength and control of especially the muscles of the lower limbs and trunk.[118,147] This sit-to-stand (STS) movement is regarded as a fundamental activity of daily living,[24,118,148] is a prerequisite for walking and standing,[22,27] and a stable performance is important from the view of safety. Therefore, improvement of the deteriorated STS performance is an important goal in rehabilitation medicine.

Stroke patients suffer balance disturbances during STS performance, as one of the results of the hemiparesis.[22,27] For this reason, many instruments used in clinical practice (generally based on observation or interview) focus on or include the assessment of the STS movement. More objective and quantitative methods using devices as force plates, opto-electronic cameras, and body-fixed sensors, are also applied.[149]

Balance control is an active process by the central nervous system to keep the body upright. Using an inverted pendulum model Maurer states that as a result of slight deviations from the upright position the resultant gravity-induced torque will cause accelerations of the trunk.[150] The resulting sway can be assessed in several ways, using several parameters.[22,27,54,151,152] Which body sway measures to use is still being discussed due to questions concerning validity and sensitivity of the sway parameters to describe balance control.[152] Interestingly, direct measurement of the acceleration at the trunk, as measure for balance control, is seldom performed.[54,57]

Although all these devices can provide data on or are related to movement pattern,[48] body-fixed sensors, such as accelerometers, have the advantage that they are relatively cheap and simple, and that they allow prolonged ambulatory measurements in a patient's own environment. It can be assumed that these prolonged measurements outside a gait lab provide ecologically valid data about a patient's performance.[48] Furthermore, accelerometry based measurements provide a base for combined detection of activities and assessment of postural control. The use of accelerometry is supported by both theoretical arguments and results of prior studies.[58,140-143] In addition it provides a portable system that allows standardized, short-term measurements outside a movement laboratory.

Several studies have used accelerometry to investigate different aspects of the STS movement, but none has addressed the issue of balance control during the STS movement.[58,140-143] In previous research we studied the ability of accelerometry to provide data on body postures and motions (including the number of STS movements) during prolonged measurements in daily life (Activity Monitor, AM), and to assess the STS movement in the sagittal plane.[45,48,141] Our data from measurements performed so far and findings reported in literature[51,54,57,141,153] indicate that trunk movements in the transversal plane during the STS movement may be a sensitive indicator of balance control which is related to motor recovery. Although lateral trunk movements may include low and medium frequency body sway (e.g. related to postural asymmetry), high frequency body sway is assumed to be the most directly related to balance control.[54,153] Balance control is mostly assessed during standing posture, although balance control is also of concern during the changing of posture. This aspect is also included in clinical assessment tools as PASS and Trunk Control Test (TCT).[154,155]

Accelerometry is potentially able to measure high frequency body sway, and offers a novel technique for assessing balance control during the STS movement, but its feasibility during rising from a chair is unknown. One of the aspects of feasibility is sensitivity, i.e. for example the ability to discriminate between groups, conditions and moments in time. The aim of the present study is therefore to determine and compare the sensitivity of accelerometric parameters in measuring balance control during the STS movement, using groups with different balance control.

Materials and Methods

To assess the *sensitivity* of accelerometric balance parameters, several high frequency lateral sway parameters were calculated and compared between 1) healthy subjects for whom balance was experimentally influenced in different conditions, 2) stroke patients and healthy subjects, 3) stroke patients with good and poor balance, and 4) different phases of motor recovery of stroke patients. To compare the sensitivity of accelerometry with a reference method, balance parameters were also determined with a force plate.

Subjects

The study included 11 healthy comparison subjects (4 males, 7 females; height 1.76±0.06 m; weight 70.7±7.4 kg; age 28.2±7.9 years) and 31 patients with stroke (21 males, 10 females; age 63.3±12.8 years). Healthy subjects were included if they had no history of neurological or musculoskeletal deficits or balance disorders. Patients participated in an ongoing prospective cohort study on the recovery of the STS transfer after stroke. They were recruited from the Stroke Unit of the Erasmus Medical Centre Rotterdam and were included if they met the following criteria: 1) a first episode of unilateral cerebrovascular accident with hemiparesis, 2) the ability to understand and follow simple verbal instructions, and 3) 20-80 years of age. They were excluded if they had a history of any other neurological pathology, previous stroke with persistent problems, and dizziness and/or balance problems that form a safety problem during assessment. The study was approved by the Medical Ethics Committee of the Erasmus Medical Centre. All participants signed an informed consent.

Instruments

Accelerometry:
A piezoresistive accelerometer (Analog Devices, ADXL202, Temec Instruments, Kerkrade, The Netherlands) was attached to the skin over the sternum, and was sensitive in transversal direction. According to the standard configuration of the Activity Monitor,[45] an additional accelerometer - sensitive in the sagittal direction - was attached at the same place. The accelerometers were attached with their sensitive axis as parallel as possible to the related anatomical axis. Each accelerometer was attached to a data recorder (Vitaport 2, Temec Instruments). Data were sampled with 128 Hz. Before each measurement, the accelerometers were calibrated (± 1 g; 9.8 m/s^2). After the measurements, data were downloaded onto a personal computer for analysis.

Force plate:
Data were collected from a force plate (AMTI OR6-7MA, Watertown, Massachusetts, USA). The parameter of interest was the medio-lateral component of the center of pressure (CoP). Data were AD converted with a 12-bit resolution DASH-16 PC board (640 Hz).

The Postural Assessment Scale for Stroke patients (PASS):
The PASS is developed and used for patients with stroke, even with poor postural performance,[154] and has shown good psychometric properties.[156] The PASS contains twelve 4-level items (0-3), which are used to grade the performance for situations of varying difficulty in maintaining (5 items) or changing (7 items) a given posture. The total score ranges from 0 to 36. The score on the PASS maintaining items was used to divide the patient group into two subgroups with a different functional level of postural control: good balance (score > 10), and poor balance (score ≤ 10).

Protocol

Healthy subjects:
Simultaneous measurements with the accelerometers and the force plate were performed in one session in a performance laboratory. The subjects sat upright on a chair without a backrest. The chair was placed just behind the force plate so that only the feet were on the force plate. The height of the seat was individually set at

knee height when standing with shoes. All subjects wore comfortable shoes. Each STS movement was made at self-selected rising speed, with the eyes focused on a target in front of the subject, and with the arms crossed in front of the trunk. At the end of each transfer, subjects stood still for at least 2 seconds. Four conditions were defined: 1) rising on 2 legs with the feet placed on a hard level surface (reference condition), 2) with the feet on 10-cm thick foam (40x40 cm), 3) with the feet on a Tumble Forms 2 vestibular board (48x45x15 cm), and 4) rising on 1 leg (dominant side) with the foot placed on a hard level surface. The foam and balance board were placed on the force plate. Six trials were performed for each condition. Sufficient rest was given in between trials and conditions to prevent fatigue.

Patients:

Patient measurements were performed either at the hospital or at home after discharge. Only accelerometric data were collected. Each patient was measured twice: the first measurement (T_0) took place between 4 days and 6 weeks after stroke, when a patient was first capable to independently perform a STS movement; the second measurement (T_1) was scheduled 12 weeks after stroke. Patients were asked to rise from a convenient chair (with armrests) that they were familiar with, at a self-selected rising speed, and with use of the arm rests. During rising, they were allowed to use other aids when this was necessary. The same aids were used at T_0 and T_1. Patients performed 6 STS movements during each measurement. The PASS was assessed before recording the STS movements using the protocol given for the PASS.[154]

Data Analysis

Accelerometry:

Accelerometric signals were converted into ASCII format and imported in custom-made Matlab programs (Matlab 6.5, The MathWorks Inc., Natick, Massachusetts, USA). The start and end – and thus also the duration – of the STS movement was derived from the sagittal trunk signal.[173] In short, after filtering (6 Hz second order low pass Butterworth filter) the derivative of the acceleration signals was calculated. By use of a matlab algorithm, the start of the STS movement was determined where the derivative signal first exceeded the 0.05 m/s^3 threshold (see Fig. 1). The one closest before minimum acceleration (t_2 for the trunk) was defined as t_1. The end of the STS movement was determined where the derivative signal entered the 0.05 m/s^3 threshold. The automatic detection was visually checked. The algorithm provided correction towards a prior or subsequent automatically determined point in time (see Fig. 1).

The transversal trunk signal was 2 Hz high-pass filtered (4th order recursive Butterworth filter) (Fig. 2). Two accelerometric outcome measures of lateral sway were calculated from the filtered transversal signal for each STS movement: the Root Mean Square (RMS_{acc}) and the area under the curve (AUC_{acc}). Both parameters were calculated

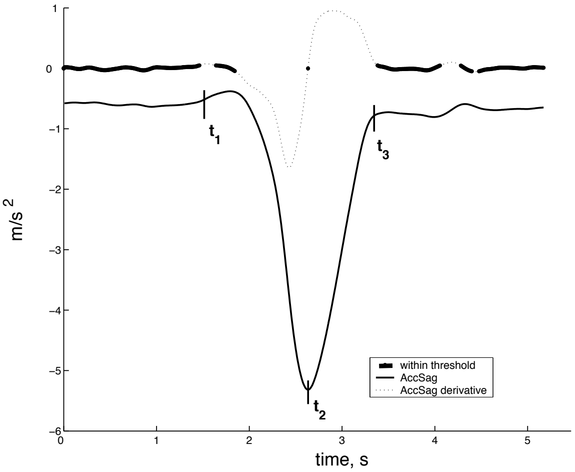

Figure 1. *Detection of trunk movement events using the defined algorithm in Matlab. In this figure the trunk sagittal signal with its derivative are shown. For reasons of representation the derivative signals are scaled to fit (factor 0.1). Parts of the derivative signal which are within the threshold are marked. Automatic detection based on the algorithm is shown by a | in the original signal.*

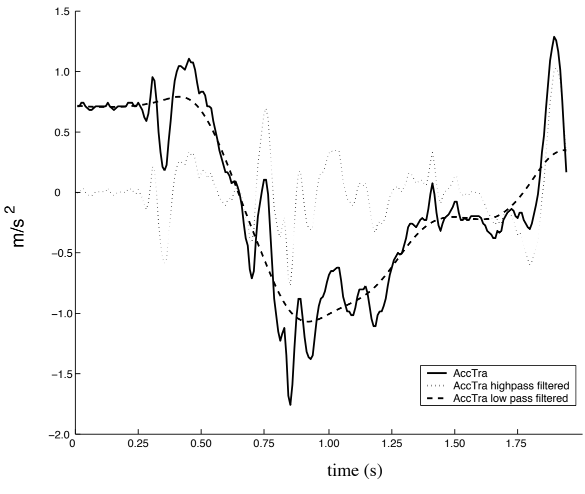

Figure 2. *The transversal trunk signal with 2 Hz high-pass filtering (4th order recursive Butterworth filter). This high pass filtered signal is used for further analysis of balance.*

for the part of the movement between the determined start and end. The RMS, being the square root of the mean of the squares of the amplitude, expresses the average magnitude of the amplitudes of higher frequency components, whereas the AUC expresses the summed amplitudes. As a result, the AUC is not only influenced by the amplitudes of higher frequency components, but also by the duration of the STS movement.

Force plate:
Data from the force plate were synchronised with the accelerometric data. Center of pressure was calculated according to the technique provided by the producer of the force plate (www.amtiweb.com/calculations.htm); to correct for the height of the balance board a correction was applied as reported by Latash (COPy = (-h·Fy+Mx)/Fz).[157] The transversal CoP signal from the force plate was also converted into ASCII format and imported in custom-made Matlab programs. This signal was low-pass filtered (30 Hz) and subsequently differentiated, resulting in a velocity profile.[152] This profile was high-pass filtered (2 Hz, 4th order recursive Butterworth filter). The start and end moment were derived from the accelerometer signal. Subsequently, the RMS (RMS_{fp}) and the AUC (AUC_{fp}) were also calculated for the force plate data.

Statistical Analysis

For both the accelerometric and force plate data the median of 6 STS movements was calculated. Paired samples Wilcoxon tests were used to compare the different conditions for healthy subjects, and to assess the sensitivity to change of accelerometric data in patients, comparing data of T_1 with T_0. Mann-Whitney U tests were used to compare patients with healthy subjects and to compare patients with good and poor balance. Linear regression analysis was used to examine the contribution of RMS_{acc} and duration to AUC_{acc}. All analyses were done with SPSS version 12.0.5 (SPSS Inc., Chicago, Illinois, USA), and an α level of 0.05 was chosen for significance testing.

Results

In the healthy subjects, in the three disturbed balance conditions AUC_{acc} differed significantly from the reference condition, whereas RMS_{acc} differed only in two conditions (Table 1). No significant differences were found for duration. The two force plate parameters were significantly different in all three disturbed balance conditions compared to the reference condition.

Duration, RMS_{acc} and AUC_{acc} all showed significant differences between patients and the healthy subjects (Table 2). When at T_0 patients were divided into two sub-

Table 1 *Median (range) and p-values of accelerometer and force platform parameters in healthy subjects (n=11)*

Instrument	Parameter	Reference (Ref) median (range)	Foam (F) median (range)	p (F vs.Ref)	Balance board (BB) median (range)	p (BB vs.Ref)	One leg (OL) median (range)	p (OL vs.Ref)
Accelerometry	Duration (s)	1.91 (1.54-2.25)	2.03 (1.64-2.27)	0.12	1.99 (1.83-2.68)	0.15	2.08 (1.77-2.53)	0.05
	RMS_{acc} (m/s^2)	0.15 (0.10-0.28)	0.19 (0.15-0.25)	0.21	0.23 (0.16-0.82)	0.01	0.41 (0.29-1.16)	<0.01
	AUC_{acc} (m/s)	0.22 (0.17-0.33)	0.26 (0.19-0.35)	0.02	0.29 (0.21-1.05)	0.02	0.62 (0.42-1.46)	<0.01
Force platform	RMS_{fp} (m/s^2)	0.08 (0.07-0.23)	0.14 (0.10-0.29)	< 0.01	0.18 (0.13-0.42)	<0.01	0.16 (0.11-0.25)	0.01
	AUC_{fp} (m/s)	72.8 (58.5-152.7)	122.8 (97.6-211.4)	<0.01	145.4 (95.68-310.06)	<0.01	141.5 (102.9-192.69)	<0.01

RMS: root mean square, AUC: area under the curve, significance tested for conditions versus reference condition

Table 2 *Comparison of the accelerometric parameters (median and range) of patients at T_0 (n=31) and the reference condition of healthy subjects (n=11)*

Accelerometric parameters	Patient group	Healthy group	p
Duration (sec)	2.87 (1.66-4.79)	1.91 (1.54-2.25)	<0.001
RMS_{acc} (m/s^2)	0.23 (0.13-0.30)	0.15 (0.11-0.28)	0.01
AUC_{acc} (m/s)	0.44 (0.23-0.06)	0.22 (0.17-0.33)	<0.001

RMS: root mean square, AUC: area under the curve

groups based on their PASS score, a significant difference was found for duration and AUC_{acc} (Table 3). A similar result was found when the data at different moments of recovery were compared (Table 4).

Linear regression analysis performed on the patient data revealed that the changes and differences found in AUC_{acc} were not completely or mainly the result of changes and differences in duration: standardized regression coefficients of both duration (range: 0.64 to 0.82) and RMS_{acc} (range: 0.59 to 0.99) were significant (all $p < 0.01$).

Discussion

Rising from a chair is one of the most commonly performed tasks of daily living and an important issue in the post-stroke treatment of many patients. A stroke frequently leads to disturbed movement control, in its turn resulting in problems related to the performance of the STS movement. The purpose of this study was to determine the

Table 3. *Comparison of accelerometric parameters, median (range), of two patient sub-groups (good balanced, n=13; poor-balanced, n=18)*

	Patient subgroup		
Accelerometric parameters	Good-balanced	Poor-balanced	p
Duration (sec)	2.58 (1.86-3.73)	3.04 (1.66-4.79)	0.04
RMS_{acc} (m/s^2)	0.21 (0.14-0.28)	0.24 (0.14-0.30)	0.49
AUC_{acc} (m/s)	0.39 (0.23-0.60)	0.48 (0.25-0.69)	0.05

RMS: root mean square, AUC: area under curve

Table 4. *Comparison of accelerometric parameters [median (range)] of patients in different phases of recovery (n=31)*

	Phase of recovery		
Accelerometric parameters	T_0	T_1	p
Duration (sec)	2.87 (1.66-4.79)	2.39 (1.32-3.51)	0.02
RMS_{acc} (m/s^2)	0.23 (0.13-0.30)	0.24 (0.12-0.31)	0.96
AUC_{acc} (m/s)	0.44 (0.23-0.68)	0.39 (0.20-0.59)	0.02

RMS: root mean square, AUC: area under curve

feasibility of accelerometry in measuring balance – defined as the high frequency lateral sway - during the STS movement of healthy subjects and patients with stroke. Accelerometry has been used previously to assess balance, but studies performed so far focused on postural control during quiet standing, and on differences due to disease and different standing surfaces.[22,50,51,54,57,158] Our study showed that accelerometry, and especially the AUC parameter, can be used to discriminate between different levels of postural control during the STS movement.

In the present study we focused on the ability of accelerometry to distinguish between conditions, groups, subgroups and moments in time, to illustrate the sensitivity of accelerometry in assessing balance control. The present study did not aim to describe specific groups of subjects or conditions. The characteristics of subjects and conditions were not exactly the same because we primarily wanted to describe the ability of accelerometry to distinguish between level of balance control compared to a reference method. The force plate was used as a reference, because it is a generally accepted technique for measuring balance,[22,27,57,97,151,159] and a relationship between force plate and accelerometer data can be expected. In our study, we found correlation coefficients of 0.77 (RMS, $p < 0.01$) and 0.58 (AUC, $p = 0.06$) between force plate and accelerometric data, respectively. Although force plate and accelerometric data were analysed in corresponding ways and analyses of both data focused on the same characteristics in the signal, it should be noted that the parameters obtained from both data are not expected to be exactly related, because the two methods measure constructs that are not completely the same.[57] The transversal trunk accelerometer measures directly the lateral accelerations of the trunk, whereas a force plate measures the position and movement of the CoP in the transversal plane, which is the result of the position and movement of different body parts. Therefore, the comparison of the sensitivity of force plate and accelerometric data does not only include different instruments, but also measured constructs that differ to some degree. We feel that in research focusing on lateral sway of the trunk, accelerometry is a more direct and probably a more valid method than a force plate. Furthermore there is still some discussion, using a force plate, as to which body sway measures to use due to questions concerning the validity and sensitivity of the sway parameters to describe balance control.[152]

Our measurement setup differed to some extent from the one used by Moe-Nilssen and Helbostad[54] and Mayagoitia et al.[57] because we attached the accelerometer to the sternum and not at the low back. Therefore, the accelerations we measured will not be as closely related to the acceleration of the CoM as in their studies. As the aim of our study was to determine and compare the sensitivity of accelerometric parameters in measuring balance control during the STS movement, we did not describe the relationship between acceleration of the trunk and CoM in our setup.

Several parameters can be defined and calculated from the accelerometric data to assess lateral sway. In the present study, we used the RMS and the AUC as the main parameters. By using these parameters, we actually focused on the high frequency movements of the trunk in the transversal plane, which is assumed to be related with disbalance.[54] This high frequency range with intermediate and higher frequency bands would include the closed-loop as well as the open-loop postural control as reported by Collins and DeLuca for quiet standing.[160] The RMS and AUC of the transversal accelerometer signal are only slightly influenced by slow (i.e. low and medium frequency) lateral movements of the trunk during the STS movement. These lower frequent movements – which can be assumed to express movement asymmetry (see Fig. 1) - can occur in stroke and may be relevant, but were not part of the analyses of the current study. Future study will also need to focus on this component.

The interpretation of the results is not straightforward because of the mutual dependencies between RMS, AUC and duration. Both RMS and the AUC focus on the high frequency components and were assumed to be related. This was shown by the correlation coefficients found: 0.97 ($p < 0.001$) and 0.93 ($p < 0.001$ for force plate and accelerometry, respectively. Despite these strong relationships, they partially provide different information. The RMS depends only on the magnitude of amplitudes, whereas the AUC depends on both amplitude and duration. The difference between

RMS and AUC was also clear in our study: the results showed that AUC_{acc} is clearly more sensitive than RMS_{acc}. This raises the question whether or not the duration of the STS movement is the only factor determining AUC_{acc}. However, linear regression analysis showed that AUC_{acc} is determined by both duration and amplitude (i.e. RMS), and the regression coefficients indicate that both aspects are almost equally important. Another argument that variability in AUC_{acc} can not be fully explained by duration is provided by the result that no significant differences in duration were found between the conditions in the healthy subjects, whereas AUC showed significant differences for all conditions. The clinical interpretation of these findings is that decreased balance can express itself in more high frequency lateral sway and/or a longer movement time. Further research is needed on the relevance and significance of the chosen strategy.

The results of the study indicate that accelerometry is able to provide a sensitive measure of balance during the STS movement. Accelerometry offers the benefit of low cost and portability.[51] An important advantage of accelerometry therefore is that it allows measurement outside a movement laboratory. These measurements may include short-term, standardized measurements – as in the present study - but also prolonged (e.g. 24-hour), unsupervised measurement in the patient's own environment or in a semi-natural setting.[45,52] Prolonged measurement in the patient's own environment can also provide data on the number of STS transitions during normal daily life. In this way assessment of the STS movement is possible in a more ecologically valid manner. Further studies in this area should include the added value of these types of measurements.

6

Recovery of the actual Sit-to-Stand performance after Stroke: a longitudinal cohort study

WGM Janssen, JBJ Bussmann, RW Selles, PJ Koudstaal, GM Ribbers, HJ Stam

Submitted to Stroke

Abstract

Objective

To explore the recovery of Sit-to-Stand (STS)-related functioning during the first year post-stroke. STS-related functioning was studied with a focus on the capability to perform an STS movement independently, the speed of rising, and actual STS performance during normal daily life.

Design

A prospective cohort study. Assessments were made at 0, 3, 6, 9, 12, 24 and 48 weeks post-stroke. Actual STS performance was assessed at 0, 12 and 48 weeks.

Participants

Fifty patients with a stroke were included.

Interventions

Not applicable.

Main Outcome Measures

Capability to rise independently; rising speed (Power chair stand up (Pcsu)); number of STS movements; and percentage walking and standing during daily life (using an Activity Monitor).

Results

During the first year post-stroke the percentage of patients able to rise increased from 54 to 83%. Most improvements occurred during weeks 0-12 (p = 0.00), whereas for weeks 12-48 no significant change was observed (p = 0.160). Rising speed showed a similar pattern: an increase from 0.15 to 0.27 s^{-1} during weeks 0-12 and up to 0.31 s^{-1} at week 48 (p = 0.00 and 0.01, respectively). The number of STS movements increased significantly during weeks 0-12 (from 10.6 to 17.7; p = 0.004), but not during weeks 12-48.

Conclusions

STS-related functioning improves significantly in the year after stroke, with the strongest improvements during the first 12 weeks. After 12 weeks, rising speed, gait speed and Barthel Index also show significant improvement.

Introduction

Stroke is a frequent cause of problems in body function and causes limitations in activity and participation. In the Netherlands, approximately 41,000 persons per year (28 per 100,000) suffer their first stroke.[161] Although neurological recovery can occur over time, many of these patients experience a persistent loss of function and related limitations of activities and participation.[30]

A stroke frequently results in an impaired Sit-To-Stand (STS) movement[3,22,24,128,162-164] ranging from inability to perform this movement independently to slow and/or asymmetric performance.[17,24,27,109,165] Stroke not only affects carrying out the STS movement, but also other constructs associated with it but positioned in other functional domains, e.g. muscle strength, balance, self-care and participation. In the present study, execution of the STS movement and its associated constructs are referred to as 'STS-related functioning'.

Studies on STS-related functioning after stroke have mainly focused on STS capacity and quality, i.e. the STS movement patterns under standardized circumstances with the duration of STS movement as a primary outcome measure. However, it is questionable whether this represents the STS movement patterns as performed in daily life, particularly since capacity is not necessarily related to performance aspects, such as the percentage of time walking and standing, the amount of help needed, and the number of steps taken.[3,30,38,40,42,166-170] Therefore, measurement of STS performance in daily life may be of added value in STS research on stroke. Other limitations of studies performed so far are that they are cross-sectional or experimental in design, or focus solely on the subacute or chronic stage of stroke.[21,24,26,109,162,171,172] In the absence of a prospective longitudinal study starting in the acute phase using a broad range of measuring instruments, knowledge on the recovery of STS-related functioning in acute stroke remains incomplete.

Therefore, the present study explored the recovery of STS-related functioning during the first year post-stroke. STS-related functioning was studied mainly from the perspective of the execution of the STS movement: i.e. the capability to perform an STS movement independently, the speed of rising, and the actual STS performance during normal daily life. Secondarily, recovery of STS-related functioning was evaluated using outcome measures from other domains of functioning.

Methods

Patients

Patients were recruited at the stroke unit of the Erasmus MC from January 2004 to December 2005. Patients were included when the following criteria were met: 1) recent stroke; 2) age 20-90 years; 3) inclusion within 4 days after stroke. Patients were excluded for the following reasons: 1) comatose on day 4; 2) previous stroke with persistent symptoms; 3) suffered a transient ischemic attack, 4) serious co-morbidity interfering with sit-to-stand transfer or walking; 5) severe deficit in communication and understanding.

The study was approved by the Ethics Committee of the Erasmus MC. All patients signed an informed consent.

Design and Procedure

The study had a prospective observational design. Each subject was scheduled for assessment at 0, 3, 6, 9, 12, 24 and 48 weeks. At each assessment the capability to perform an STS movement independently, as well as the secondary outcome measures, were evaluated. If the patient was able to perform the STS movement independently, the duration of the STS movement was measured using ambulatory accelerometry (at all assessments), and the actual STS performance during daily life was measured with an Activity Monitor (AM) at 0, 12 and 48 weeks.

Outcome measures

Primary outcome measures: STS movement execution

Capability to perform an STS movement independently
This was assessed using the Motor Assessment Scale item related to the STS movement. This item is scored from 0 (Unable to) to 6 (Sitting to standing to sitting with no stand-by help three times in 10 seconds). To provide a dichotomous outcome the scoring of this item was re-coded: scores 0 and 1 [Gets to standing with help from therapist (any method)] on this item were recoded as 0 (Not capable); scores 2 through 6 were recoded as 1 (Capable of independently performing the STS movement).

Duration of the STS movement

The duration of the STS movement was assessed using accelerometers attached to both upper legs and to the sternum.[173] This configuration is based on the sensors and sensor location of the AM.[48,141] Measurements were done in the most normal environment for the patient at that moment (e.g. their hospital room, or living room at home). The sensitive axis of the leg accelerometers was oriented into the sagittal direction while standing, whereas the accelerometers at the sternum were sensitive in three directions (sagittal, longitudinal, and transversal). Six STS movements were performed at comfortable speed, with and without use of the arms. Patients were allowed to rise with their feet in preferred position; the chair that was mostly used in daily life was used for the measurements. Data from accelerometry were stored on a Vitaport 2 recorder (sample frequency 128 Hz) and transferred to a PC for further analysis. The assessment of STS movement duration was based on the signals in sagittal direction of the non-paretic upper leg and sternum. A Matlab algorithm for detection of start and end of the STS movement was used.[173] For analysis, we used the inverse of the duration (1/duration), called Power chair stand up (*Pcsu*), which is equivalent to rising speed.[10] If a patient was not capable to perform the STS movement independently the rising speed was scored as 0.

Actual STS performance

The same AM sensor configuration (as described above) was used for the assessment of actual STS performance. If a subject was capable of independent rising, the AM was worn from 10 a.m. to 6 p.m. Data from daytime ambulatory accelerometry were transported to a PC and analyzed with the Kinematic Analysis Module of the Vitaport Analysis Package[a]. This program allows automatic detection of body postures, body motions, and transitions between body postures.[45] Outcome measures used in the present study were the number of STS movements during the measurement period (e.g. also including transitions from sitting to walking), and the percentage of time spent standing and spent walking during the measurement period. If a subject was not capable to perform these assessments these outcome measures were scored as 0.

Secondary outcome measures

In addition to the primary measures, STS-related functioning measures concerning function, activity, and participation were assessed. These outcome measures were the Motor Assessment Scale (MAS),[174,175] the Trunk Control Test (TCT),[148,176-178] and the Postural Assessment Scale for Stroke patients (PASS),[148,154,156,179,180] a test for postural

a. Temec Instruments, Kerkrade, The Netherlands control consisting of two components concerning maintaining and changing posture. In addition, the Barthel Index (BI), the Functional Independence Measure (FIM), and the five-meter walking test (5mWT) at comfortable speed were used.[181]

Statistical analyses

Most outcome measures were analysed in a similar way. First, data were examined by descriptive techniques. For all outcome measures (except for the capability to perform the STS movement), changes in outcome during the follow-up period were analyzed using random coefficient analysis[b]. This technique allows assessment of changes over time with incomplete data sets. A basic regression model was made with time as a set of dummy variables; the assessment at 12 weeks was chosen as the reference variable. Estimates of the variables at the other assessment moments were obtained using the regression coefficients of the dummy variables provided by the model. All seven assessment occasions were included in the basic model. The results of the basic models retrieved by random coefficient analysis are presented graphically. Patients were included if they were assessed on at least two occasions.

The data on actual STS performance were analyzed in a similar way, with the assessment at 12 weeks as reference, but with only two additional assessments. For this analysis we restructured the data by allocating the measurement at weeks 0, 3, 6 as first assessment, at week 12 as second assessment (with two assessments at 24 weeks also included), and week 48 as third assessment. Changes during these intervals were tested for significance using a p value < 0.05.

Results

In total 50 patients were included: 33 male, 17 female with a mean age of 62.2 (range 28-87) years (Table 1). All patients were included within 4 days after stroke during their stay on the Stroke Unit. Four patients were not included in the analysis of rising speed recovery because they left the study shortly after inclusion. Nine patients were not included in the daytime AM analysis: i.e. the 4 patients mentioned above, 3 that were not assessed according to the scheduled times, and 2 who refused the day time monitoring. These 9 patients did not differ significantly from those included in the analysis, except for gender.

b. MlwiN, version 1.1; Centre for Multilevel Modeling, Institute of Education, London, UK

***Table 1** Characteristics of the study population.*

	All patients mean (sd)	Activity Monitor daytime analysis mean (sd)
Age (years)	62.2 (13.5)	61.0 (13.0)
Gender	33 m/17 f	30 m/11 f *
MAS	22.2 (18.5)	20.0 (18.0)
BI	9.2 (7.6)	8.2 (7.5)
n	50	41

**Significantly different from patients included in the AM daytime analysis (using t-test for independent samples) (sd) standard deviation*

In the first year post-stroke, the percentage of patients capable of rising independently increased from 54 to 83% (Fig. 1). Of the 23 patients (65%) who were not capable of rising independently shortly after stroke, 14 regained independent rising in the subsequent year. The improvements were most prominent during weeks 0-12 (p < 0.01); during weeks 12-48 there was no significant improvement (p = 0.16; paired samples t-test). Table 2 presents data on recovery of the rising speed (Pcsu), with and without arm support. Figure 2 shows the recovery of rising speed estimated by random coefficient analysis.

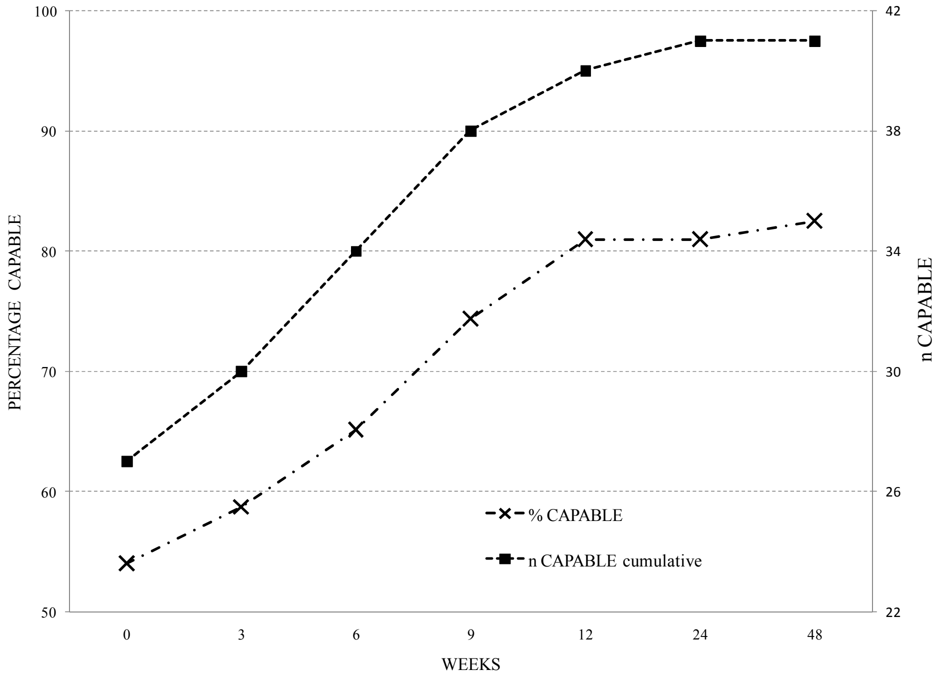

***Figure 1** Course of regaining independence for the STS movement, expressed as a percentage of the total number of patients evaluated at each assessment moment; the cumulative number of patients capable of independent rising is also shown.*

Table 2 *Data on capability of rising and rising speed (Pcsu), descriptive data.*

week	**0**	**3**	**6**	**9**	**12**	**24**	**48**
n[1]	50	46	43	39	42	42	40
Capable	27	27	28	29	34	34	33
Regained		3	4	4	2	1	
Pcsu using arms	0.15 (0.16)	0.21 (0.21)	0.20 (0.20)	0.23 (0.21)	0.26 (0.19)	0.29 (0.21)	0.30 (0.20)
	49	***44***	***39***	***35***	***41***	***39***	***40***
Pcsu without	0.15 (0.18)	0.23 (0.22)	0.23 (0.23)	0.25 (0.23)	0.25 (0.22)	0.29 (0.22)	0.31 (0.21)
using arms	***49***	***46***	***42***	***38***	***41***	***40***	***39***

[1]Numbers of patients at each assessment. Figures in italics are the numbers of the patients contributing to these data. All patients are included (also those with only one assessment); those not capable of rising are valued as rising speed (Pcsu) =0; ring speed s^{-1}

The estimated rising speed with arm support increased significantly from 0 to 12 to 48 weeks from 0.15 to 0.27 to 0.31 s^{-1} (p = 0.00 and 0.01, respectively), whereas the estimated rising speed without arm support increased significantly from 0.16 to 0.26 to 0.31 s^{-1} (p = 0.00 and 0.00, respectively). Table 3 presents data on actual STSperformance. The estimated number of STS movements and the percentage standing and walking during daily life changed significantly between 0-12 weeks, but remained stable thereafter (Fig. 2).

Table 4 presents data on the secondary outcome measures; Figures 3 and 4 show the results of the random coefficient analysis. For the purpose of presentation, all

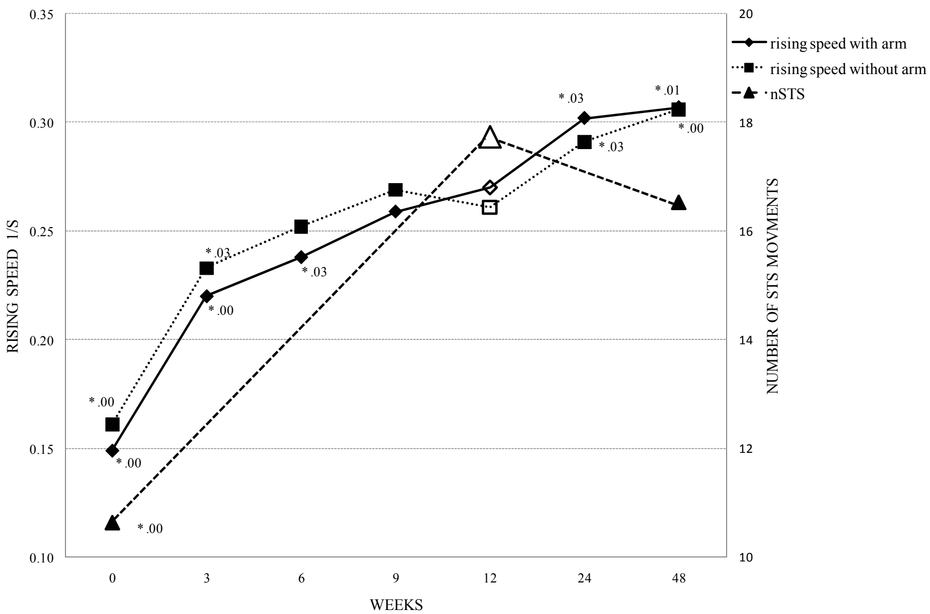

Figure 2 *Course of the estimated rising speed and number of STS movements during the first year post-stroke as calculated from basic RCA models. nSTS: number of STS movements during daytime assessment.*

Table 3 *Actual STS performance during daily life, descriptive data (mean and sd).*

Moment	t_1	t_2	t_3
n[1]	41	41	39
n STS	10.6 (11.6)*	17.7 (19.3)	16.6 (13.6)
%standing	5.2 (7.9)*	9.9 (10.9)	9.4 (8.8)
%walking	3.8 (6.9)*	7.6 (10.3)	8.3 (9.5)

*[1]Number of patients contributing to these data, *p < 0.05 compared with t_2*

Table 4 *Secondary outcome measures for the patients included in the RCA analysis for rising speed, descriptive data (mean and sd)*

week	0	3	6	9	12	24	48
n[1]	46	46	43	39	42	42	40
MAS	21.2 (18.1)	28.0 (19.2)	30.1 (18.6)	32.0 (17.6)	34.3 (17.4)	34.3 (17.2)	36.0 (17.2)
TCT	50.0 (38.8)	62.5 (35.1)	67.1 (33.2)	68.5 (30.1)	71.4 (31.4)	72.0 (31.5)	72.2 (31.1)
PASS	17.9 (13.7)	23.2 (12.2)	24.6 (10.9)	26.6 (9.6)	27.6 (9.1)	27.5 (10.3)	28.4 (9.4)
BI	8.7 (7.4)	13.1 (7.3)	14.4 (6.7)	15.2 (5.8)	15.8 (5.9)	16.3 (5.3)	16.8 (5.2)
FIM	64.9 (36.1)	89.1 (33.7)	93.5 (32.3)	99.1 (27.0)	103.7 (27.0)	103.1 (25.9)	106.9 (23.1)
Gait speed	0.31 (0.40)	0.49 (0.49)	0.51 (0.51)	0.54 (0.52)	0.57 (0.49)	0.60 (0.51)	0.66 (0.50)

[1]Numbers of patients for the MAS, TCT, PASS, BI, and FIM data

estimates for STS-related functioning clinical measures were normalized to a percentage of maximum range of recovery. For MAS, TCT and PASS there was a significant recovery during weeks 0-12 post-stroke, but not during weeks 9-12 weeks and

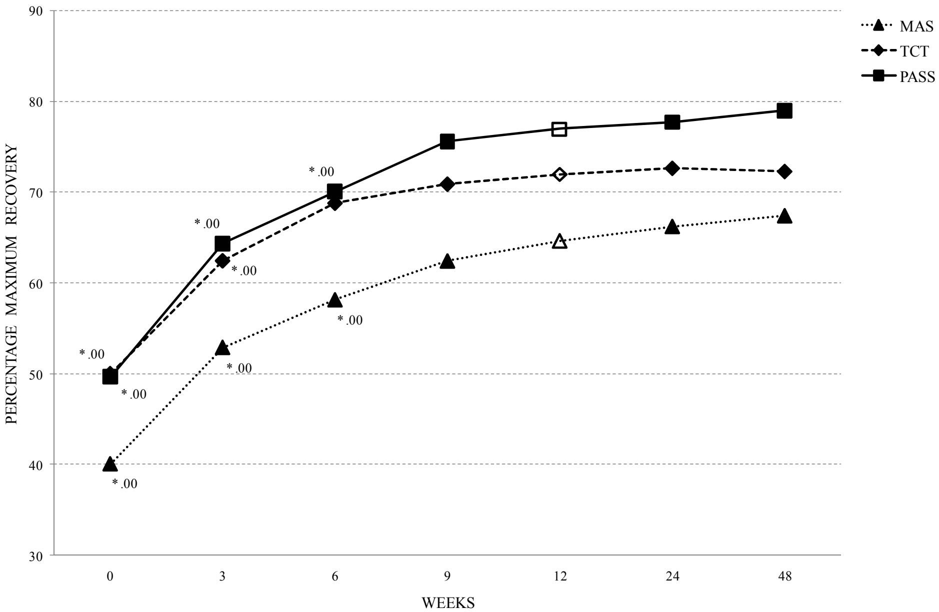

Figure 3 *Course of the recovery for the Motor Assessment Scale (MAS), the Trunk Control Test (TCT) and Postural Assessment Scale for Stroke (PASS) during the first year post-stroke as calculated from the basic model (for patients included in the RCA analysis).*

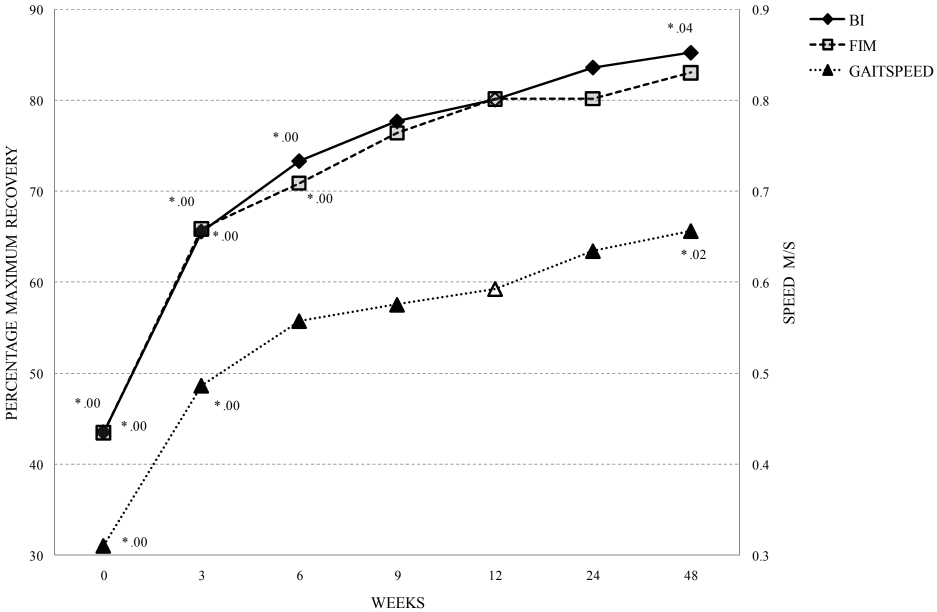

Figure 4 *Course of the recovery for the Barthel Index (BI), the Functional Independence Measure (FIM) and the gait speed during the 5-meter walking test as calculated from the basic model.*

thereafter (see Fig. 3). Estimates for BI, FIM and gait speed showed a similar course of recovery; however, for BI and gait speed a significant improvement was found during weeks 12-48 (p = 0.04 and 0.02, respectively) (Fig. 4).

Discussion

This prospective longitudinal cohort study examined the recovery of STS-related functioning during the first year post-stroke in 50 patients, focusing on STS movement execution. In general, all outcome measures showed strongest improvement during the first 12 weeks post-stroke, with a levelling off thereafter. However, besides this overall pattern, differences were also seen in the magnitude of change for outcome measures after 12 weeks. The general patterns found are similar to previous reports on other aspects of functioning after stroke. For example, Kollen et al. and Kwakkel et al.[30,182,183] presented data on recovery of balance, Fugl Meyer scores for leg, and gait after stroke, and found a similar course of changes during the first 3 months. Our patterns are also comparable to the time course for reaching stationary walking, best neurological outcome, and maximum ADL function, as found in a cohort study of Jorgensen et al.[184,185] These similar time courses suggest a common basic mechanism of recovery for STS movement and other aspects of functioning. One of the options might be the recovery of trunk and balance control.[30,148,186] For example, Kollen et al.[182] showed that primarily standing balance control and (to a lesser extent) leg strength are predictive for gait recovery. This might also be the case in recovery of the STS movement. The presence of an underlying mechanism for recovery of gait speed and rising speed is supported by an additional analysis of the present data. Mean gait speed and mean rising speed with arm support were clearly correlated (r^2=0.90) at different assessment moments using a linear regression technique: gait speed = 2.0 · Pcsu + 0.06. Furthermore, gait speed and rising speed showed the same course (a significant increase), during weeks 12-24 and weeks 12-48. This phenomena could be the result of functional recovery, without an underlying change of impairments, due to compensation strategies still acting in this period.[30] For gait speed this phenomenon has recently been addressed.[187] Nevertheless, in our study population, no increase in the number of STS movements or percentage walking occurred, showing that no direct relationship between rising speed, gait speed and number of STS movement may be expected.

The capability to rise independently is an important but rather rough indicator of STS functioning. Nevertheless, in the present study, the general pattern described above was also found for this independent rising. At the moment of inclusion 54% of our patients showed the capability to rise independently, and 18% did not regain this capability before the end of the first year post-stroke.

This means that there is a patient group with severe persistent function deficit which hampers an independent rising. To our knowledge, ours is the first study to examine the recovery of rising speed in the first year post-stroke. The rising speed changed significantly during the study period, both for rising with and without arm support. Mean rising speed values in our patients ranged from about 0.15 s^{-1} to 0.31 s^{-1}, depending on the assessment moment and whether rising with or without arm support. For those patients capable to rise independently, during the study period the rising speed increased from 0.31 to 0.40 s^{-1} with arm support, and from 0.35 to 0.41 s^{-1} without arm support. In the mean data, the effect of not being able to rise is included because not being capable is transformed to rising speed=0. This will result in lower mean values and an increase of the mean rising speed over time when more persons are able to perform the STS movement. However, when analysing the data for the patients capable to rise independently separately, a significant difference in rising speed was found between week 12 and weeks 0, 3 and 6; the rising speed with arm support being significantly lower than the rising speed without (p = 0.018, 0.034, 0.011, respectively; paired sample t-test). At later moments there were no significant differences in mean rising speed with or without arm support (see Fig. 2). The later disappearance of the differences present at weeks 0, 3 and 6 is probably due to recovery, or is a result of exercise with a related increase of rising speed.

Other studies on STS movement duration in the subacute and/or chronic phase reported durations of STS movement ranging from 1.45-4.8 s for comfortable rising without arm support, and rising speed values of 0.21-0.69 s^{-1}.[5,16,17,22,24,27,28,64,109,163,171,188] However, direct comparison between our study data and others is difficult because of differences in study definitions, techniques and time post-stroke, all of which hamper a valid comparison.

Actual STS performance data showed a significant increase in activity during the first interval (up to 12 weeks) but with no change during the second interval (after 12 weeks). The estimated mean number of STS movements, mean percentage standing, and mean percentage walking increased by 67, 88 and 98%, respectively, during the first interval. Assuming that an STS movement is a start of a walking period, the difference in the increase of the number of STS movements and the percentage walking indicate a lengthening of walking periods. Previous studies provide little information on actual STS performance during the first year post-stroke, except for studies investigating additional exercise of the STS movement in the subacute phase[41,42] and studies on the number of steps taken.[39,40] Britton et al., studying improvement of the STS movement, reported that the number of STS movements in the control group (receiving no extra training) was 18 for the time period 10 a.m. to 3 p.m.[42] Barecca et al. reported that the median number of STS movements was 10.6 per 24 hours.[41] Mean data from our study population are higher than that of Barrecca et al.[42] and similar to that of Britton et al.[41] These differences might be attributable to differences in inclusion criteria and assessment moments. The number of STS movements in our stroke patients was considerably lower compared with an earlier group of healthy subjects (nSTS 61.1/24 h ≈ 31.5/8 h[44,189]). Until now, for stroke patients, only the number of steps during daily life have been presented and no data on percentage walking. Shaughnessy et al.[40] and Michael et al.[39] assessed 2765 (±1677) and 2837 (± 1503) steps per day (range 66-5795 steps), respectively. Our increase in estimated percentage walking of 98% is similar to the 80% increase in steps reported by Shaugnessy et al. and underlines their statement concerning the magnitude of changes that can occur during measurement of behaviour in daily life.[40] In our study, the percentage of daytime walking at the end of the study period (8.3%) is lower than reported in healthy subjects (10.4-14.6%).[44,189] The mean percentage of walking for those who were capable was 10.7% (range 1-41.1%); the large variability indicates that some were either highly active or very inactive.

This study has some limitations which might influence the results and conclusions. First, the number of included patients was relatively small. Although this may have led to a lack of power and non-representativeness of the study sample, the data do not suggest a strong influence of either. Second, based on the selection and size of the cohort, full information on possible relevant subgroups was lacking; i.e. a large proportion of patients not capable to rise at the moment of study inclusion would have provided more insight into the course of recovery in these patients. Third, more frequent assessment of actual STS performance over a longer period could have provided more insight into the relationships between actual STS performance and rising speed and secondary outcome measures. Fourth, the STS movement and duration were only assessed under supervised and standardized conditions, although the measurements were made in the environment most familiar to the patient at that moment. Therefore, our approach is a first step towards assessment of the quality aspects of actual STS performance.

In conclusion, the present study emphasizes aspects of the recovery of STS-related functioning in the first year post-stroke which had not yet been explored, such as the rising speed and the number of STS movements during daily life. Mean rising speed increased significantly during the intervals 0-12 weeks and 12-48 weeks due to recovery of the capability to rise and to an increase of individual rising speed. Further studies on recovery of STS-related functioning are required to further elucidate the relationship between the currently used clinical measures and the recently developed daily life outcome measures.

7

Prognostic Determinants of the Sit-To-Stand movement execution after Stroke

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To be submitted

Abstract

Objective

To establish which baseline variables are prognostic for the (course of) capability to execute the Sit-to-Stand (STS) movement during 48 weeks post-stroke, and which baseline variables predict the rising speed and number of STS movements during daily life at 12 and 48 weeks.

Design

Prospective cohort study.

Patients

A total of 50 patients with a stroke were included within 4 days post-stroke at a stroke unit of a university hospital.

Methods

The capability to perform the STS movement was assessed at 0, 3, 6, 9, 12, 24, and 48 weeks post-stroke. At 12 and 48 weeks post-stroke, the rising speed and the number of STS movements during daily life were assessed. The independent variables included patient and stroke characteristics, function measures (Motor Assessment Scale, MAS; Postural Assessment Scale for Stroke patients, PASS; Trunk Control Test, TCT), and activity measures (Functional Independence Measure, FIM; the Barthel Index, BI).

Results

For the (course of) capability to execute the STS movement all function measures were prognostic. For rising speed and the number of STS movements both function and activity measures were predictive, with function measures showing higher standardized regression coefficients than activity measures.

Conclusions

In our study group, function measures were prognostic for the capability to execute the STS movement and the PASS score had the highest prognostic value. Both function and activity measures can be used for predicting the rising speed and number of STS movements.

Introduction

Stroke is a frequent cause of problems in bodily function and causes limitations in activity and participation. Although recovery can occur after a stroke, many patients with a stroke experience persistent loss of function and related activity limitations and participation restrictions, in part related to the Sit-to-Stand (STS) movement.[3,24,128,164] As a result a stroke may also affect STS movement execution, which we have previously defined as: *comprising the quantity and quality of the STS movement, in terms of both capacity and performance.*[190] In an earlier study we explored the following parts of the STS movement execution: the capability to perform the STS movement independently, the rising speed, and the number of STS movements during daily life.[190]

Shortly after stroke, patients, relatives and care providers need information about the recovery of functioning, including STS movement execution. Knowledge on the recovery of STS movement execution can assist in patient management, and can provide a basis for therapeutic interventions and evaluation.[3,41,42,128] Predicting recovery is an important theme in the acute phase after stroke.[3,128,191,192]

Until now, recovery of the STS-movement execution and its prognostic determinants have received little attention. Most earlier studies on the STS movement in stroke have a cross-sectional design, focus on the subacute or chronic phase, and on the quality of the STS movement in a movement laboratory.[21,24,26,109,162,171,172] Two recent studies focused on recovery of the STS movement in the subacute phase, but provided no information on its natural course or its prognostic determinants.[41,42] Perry et al. studied prognostic determinants for the change of required caregiver assistance for the Sit-to-Stand movement; however, that was a retrospective study and did not describe the STS movement.[3] Technological developments now allow us to make direct and objective measurement of the STS movement execution in a patient's own (current) environment.[173,190] Our earlier report on recovery of STS movement execution using objective measurements did not include an analysis of its prognostic determinants.[190] Thus, until now, no prognostic determinants for recovery of the STS movement execution have been established.

Therefore, the primary aim of the present study was to explore the prognostic determinants of the capability to perform an STS movement independently 48 weeks post-stroke in patients who had lost this capability, and to determine which determinants influence the course of recovery of this capability to perform an STS movement. In addition, we evaluated whether these determinants could function as predictors of rising speed and the number of STS movements during daily life, at 12 and 48 weeks.

Methods

Design/Procedure

This explorative study on the determinants of recovery of the execution of the STS movement used data from a previous study on the course of recovery of the STS movement execution in the first year post-stroke.[190] The latter study had a prospective longitudinal design and assessments were scheduled at 0, 3, 6, 9, 12, 24, 48 weeks. Assessments consisted of clinical tests with additionally ambulatory accelerometry to determine rising speed, and the number of STS movements during daily life (actual STS performance).

Subjects

Patients with an acute stroke were included at the stroke unit of the Erasmus MC from January 2004 to December 2005. For inclusion the following criteria had to be met: a) recent stroke, b) age 20-90 years, and c) within 4 days after stroke. Patients were excluded for the following reasons: a) comatose on day 4, b) previous stroke with persistent symptoms, c) transient ischemic attack, d) serious co-morbidity interfering with sit-to-stand transfer or walking, and e) severe deficit in communication and understanding.

The study was approved by the Ethics Committee of the Erasmus MC. All patients signed an informed consent.

Dependent variables

The following variables were designated as dependent: the capability to perform an STS movement independently, the rising speed, and the actual STS performance.

The capability to perform an STS movement independently was assessed using the item concerning the Sit-to-Stand movement in the Motor Assessment Scale (MAS).[174] To provide a dichotomous outcome the scoring of this item was recoded: scores of 0 and 1 [Gets to standing with help from therapist (Any method)] on this item were recoded as 0 (not capable), and scores 2 through 6 were recoded as 1 (Capable of independently performing the STS movement).

For rising speed the inverse of the STS movement duration was used (also called Power chair stand up; *Pcsu*).[10] If someone was not capable to perform the STS movement independently we scored the rising speed as 0.

The actual STS performance was assessed using the Activity Monitor (worn from 10 am to 6 pm) providing the number of STS movements during the measurement period (i.e., also including transitions from sitting to walking).[48] If a subject was not capable to perform these assessments, this outcome measure was scored as 0.

Independent variables

Based on clinical grounds and on previous studies on determinants of STS movement duration in chronic stroke[3,20,24,28] the following three prognostic variable categories were included, 1) subject and disease characteristics, 2) function measures, and 3) activity measures. These variables were assessed at the moment of inclusion.

Subject and disease characteristics included were: age, gender, side of stroke, and localisation of stroke (based on vascular territory: middle, anterior, posterior cerebral artery, and basal ganglia).

For function measures we used the MAS,[174,175] the Trunk Control Test (TCT)[148,176-178] and the Postural Assessment Scale for Stroke patients (PASS).[148,154,156,179,180]

At the activity level we used the Barthel Index (BI) and the Functional Independence Measure (FIM).[192-194]

Statistical analysis

For statistical analyses the statistical software package SAS was used.[a] For the patients not capable of rising at the moment of inclusion we explored the strength of the baseline variables to be prognostic for the capability to perform an STS movement independently after one year post-stroke, using logistic regression techniques.

The prognostic ability of the selected variables was evaluated using the area under the curve (AUC) of receiver operating characteristics (ROC) curves, odds ratios (ORs), and the *p*-value. ROC curves are the graphical representation of the sensitivity and specificity of a variable to discriminate, e.g., the capability; the AUC is related to the prognostic properties of this variable concerning the capability to rise. An AUC of 0.60-0.70 was considered to represent a poor variable, 0.70-0.80 fair, 0.80-0.90 good, and 0.90-1.00 to represent an excellent prognostic variable. Furthermore, we studied which variables influenced the course of recovery, using survival techniques, with resulting hazard ratios and *p*-values. The hazard ratio is the effect of the independent variable on the “risk” of recovery. For the determinant localisation of stroke, only the prognostic value for recovery of capability was calculated due to the high proportion in which location was unspecified.

To assess predictive baseline variables for rising speed and the number of STS movements during daily life at 12 and 48 weeks, regression techniques with stan-

a. SAS 9.1 SAS enterprise

dardized regression coefficients and *p*-values were used to select the variables with the best predictive ability. For this part of the study, data from patients capable to rise independently at study inclusion were also used. Statistical significance was assumed at a *p*-value ≤0.05.

Results

The study included 50 patients (33 male, 17 female) with a mean age of 62.2 (range 28-87) years. At the moment of inclusion, of the 50 patients 23 were not capable of rising independently; 14 of these recovered during the follow-up period.

Table 1 presents characteristics of all patients and of the two subgroups used for this analysis. The patients not capable to rise at inclusion showed lower function and activity measures.

The data on capability to rise at each assessment moment are shown in Table 2, together with data on rising speed and number of STS movements during daily life at 12 and 48 weeks post stroke.

Table 1 *Characteristics of the study population and baseline variables.*

	All patients mean (sd)	Capability analysis mean (sd)	Activity Monitor (AM) analysis mean (sd)
Patient characteristics			
Age (years)	62.2 (13.5)	62.1 (14.8)	61.0 (13.0)
Gender	33 m/17 f	14m/9 f	30 m/11 f *
Function measure			
MAS	22.2 (18.5)	4.4 (6.3)*	20.0 (18.0)
PASS	18.3 (13.7)	4.3 (5.4)*	16.9 (14.0)*
TCT	51,7 (39.5)	12.6 (16.1)*	46.7 (38.1)
Activity measure			
FIM	67.2 (37.6)	32.5 (13.7)*	62.1 (36.5)
BI	9.2 (7.6)	2.1 (2.1)*	8.2 (7.5)
n	50	23	41

**Significantly different from all patients (using t-test for independent samples, p value < 0.05)*
(sd) standard deviation, AM analysis: rising speed and number of STS movements during daily life

Table 2 *Data on capability of rising, rising speed, and number of STS movements, descriptive data.*

week	**0**	**3**	**6**	**9**	**12**	**24**	**48**
n[1]	50	46	43	39	42	42	40
Capable	27	27	28	29	34	34	33
Regained		3	4	4	2	1	
Rising speed using arms					0.26 (0.19)		0.30 (0.20)
n STS movements					17.7 (19.3)		16.6 (13.6)

[1]Numbers of patients at each assessment. All patients are included (also those with only one assessment); for those not capable of rising, rising speed was scored as 0, rising speed s^{-1}

Table 3 *Determinants for recovery of the capability to rise independently after one year.*

Variable	AUC ROC	Odds ratio*	p-value
Patient and stroke characteristics			
Age	0.73	0.43	0.09
Gender	0.61	0.64	0.36
Side paresis	0.62	0.61	0.33
Function measure			
MAS	0.82	> 999	< 0.01
PASS	0.89	> 999	< 0.01
TCT	0.78	67.61	0.03
Participation measure			
FIM	0.67	9.4	0.16
BI	0.70	29.76	0.10

*n=23 patients not capable of rising at inclusion in the study * odds ratio of standardized regression coefficient β, AUC Area under the Curve, a higher value represents a better prognostic property, ROC Receiver Operating Characteristic curve For localisation p = 0.28, see Statistical Analysis section for details*

Table 4 *Determinants influencing the course of recovery of the capability to rise independently in the patients not capable of rising at inclusion in the study.*

Variable	Hazard Ratio *	p-value
Patient and stroke characteristics		
Age	0.70	0.13
Gender	0.72	0.25
Side paresis	0.70	0.20
Function measure		
MAS	3.79	0.03
PASS	13.35	<0.01
TCT	9.21	<0.01
Participation measure		
FIM	3.29	0.12
BI	5.52	0.09

** Hazard ratio of standardized regression coefficient β*

Table 3 presents data on the AUC of the ROC of the independent variables, where the AUC for both MAS and PASS are ≥0.80. Also presented are the ORs of the standardized regression coefficient, with *p*-values. All function measures were prognostic, whereas no other variables showed a significant effect.

Table 4 shows the influence of the independent variables on the course of the recovery of the capability to rise, expressed by the hazard ratio. The results are similar to those shown in Table 3, i.e. there is no significant effect of the patient characteristics or the activity variables. From the function determinants, PASS had the highest hazard ratio. To illustrate the prognostic ability of the PASS on the course of recovery, we defined a cut-off point at the 33rd percentile (PASS ≤5) and calculated the *cumulative recovery distribution* function (Fig. 1). This graph clearly shows that

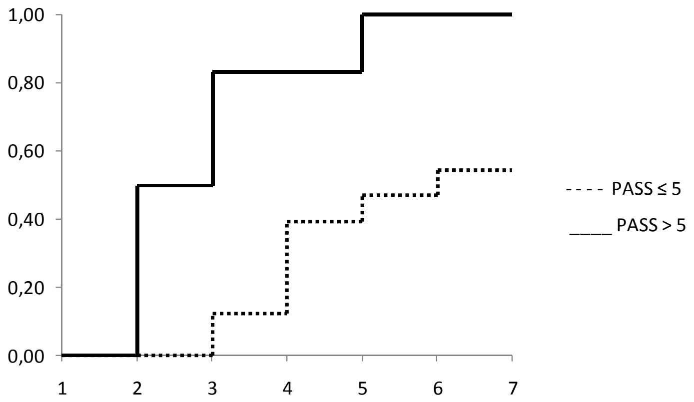

Figure 1 *Survival curve with two strata with cut-off PASS ≤5*
The Y-axis represents the cumulative recovery distribution function, the X-axis shows the seven assessment moments (see Methods section), p <0.001

Table 5 *Standardized regression coefficients (beta) for predicting rising speed and number of STS movements at 12 weeks and 48 weeks*

	Rising speed		n STS movements	
variable	12 weeks (n=31)	48 weeks (n=32)	12 weeks (n=41)	48 weeks (n=39)
Patient and disease characteristics				
Age	-0.01	0.01	-0.09	-0.12
Gender	-0.02	-0.01	0.66^{***}	0.42^{***}
Side paresis	-0.002	0.01	0.13	0.03
Function measure				
MAS	0.11^{***}	0.10^{***}	0.61^{***}	0.37^{***}
PASS	0.13^{***}	0.12^{***}	0.80^{***}	0.50^{***}
TCT	0.12^{***}	0.12^{***}	0.70^{***}	0.49^{***}
Activity measure				
FIM	0.09^{***}	0.10^{***}	0.57^{**}	0.37^{***}
BI	0.09^{***}	0.10^{***}	0.46^{***}	0.30^{***}

*For rising speed regression using truncated normal distribution, for n STS movement regression using Poison distribution. n=number of subjects, ***p < 0.0001, **p < 0.001*

with a PASS of ≤5 recovery occurs later and only a part of the patients in this group recovers, whereas all patients with a PASS >5 recover over time.

Table 5 presents prognostic data on the rising speed and the number of STS movements during daily life, as expressed by standardized regression coefficients. These analyses show that the patient and disease characteristics are not prognostic for rising speed; however, gender showed a significant effect for the number of STS movements. A significant effect was also found for the function and activity determinants. Again, the standardized regression coefficients for PASS were the highest.

Discussion

The present study explored the prognostic determinants of recovery of the STS movement execution in a cohort of 50 patients with an acute stroke. For analysis of the determinants of the (course of) the capability to rise we included 23 of the 50 patients (i.e. patients not capable to rise at study inclusion). For this subgroup of patients the function measures had a prognostic ability, with the PASS showing the highest prognostic values. For rising speed and the number of STS movements all function and activity measures had a prognostic value, with the highest standardized regression coefficients for the PASS. Gender also showed to have a prognostic value for the number of STS movements. Regarding generalizability, these results can be considered valid for patients in the acute phase after stroke (i.e. at a stroke unit).

Although other function variables also had prognostic value, the PASS was the strongest prognostic determinant for the capability to rise and its course of recovery. This might be explained by its characteristics; the PASS scale is composed of the maintaining posture (MP) part and the changing posture (CP) part, with the aim to assess both the ability to maintain a posture and to ensure equilibrium during the changing of posture.[154] Moreover, the test was developed to be used in patients with a large range of balance control, from poor to good. The PASS has shown good prognostic validity for total FIM and transfer FIM items,[154] and the PASS-CP showed to be strongly related to comprehensive ADL function.[148] Its prognostic value might be due to the fact that the PASS includes items on balance and on changing posture, and a relationship has been reported between muscle force, balance control and the STS movement.[24,28] Another possibility is that the change of posture part of the PASS will be related to muscle strength, because muscle strength is needed to execute these movements. This combination of items may have contributed to the prognostic ability of this variable. Although the MAS also includes items related to strength and balance, the number of balance items is smaller. The current measurements and analyses of data do not allow us to conclude which part of the tests used (i.e. strength or balance items) was the most relevant. Future studies could explore the items ‘maintaining posture’ and ‘changing posture’ separately. In the present study, the correlation coefficient between these two parts was 0.63, which indicates that both parts are not independent but also not totally comparable.

Besides exploring the prognostic ability of the determinants for STS capability, we also investigated other parts of the STS movement execution. Also for these latter variables, the PASS had the highest predictive value; however, the standardized regression coefficients for the MAS and TCT were very similar (especially for the rising speed). Some differences were found when comparing the data at 12 and 48 weeks; for rising speed the standardized regression coefficient was almost the same, whereas for the number of STS movements the coefficient differed. This difference is probably related to differences in the course of these two dependent variables.[190] Rising speed showed a significant increase between 12 and 48 weeks, whereas the number of STS movements did not change. Surprisingly, gender had a significant prognostic effect on the number of STS movements at 12 and 48 weeks, where gender had no effect on capability and rising. A possible explanation for this is the effect of social factors (e.g. more men having work) that might influence the amount of STS movements during daytime.

In the present study other constructs related to function (e.g. continence, or cognitive function) were not measured, although variables related to these constructs can have prognostic value. For example, Perry et al. showed that FIM cognitive scores are prognostic for caregiver assistance for the Sit-to-Stand movement; however, that study only included patients admitted for inpatient rehabilitation.[3] Although other constructs, such as strength and balance, were included in our study when using clinical measures related to function, these items were not studied in detail. The clinical measures used consist of a combination of items that cover both strength and balance.[154,174] Other clinical measures (e.g. the Berg Balance Scale) could have been selected, or we could have applied more quantitative techniques (e.g. for muscle strength or balance control assessment), using a dynamometer or force plates. Another possibility is to study isolated items (or subcategories of items) of the clinical measures we used. Proper selection of the items will enable to study in more detail the constructs of strength and balance; for example by studying the CP of the PASS, or the sitting balance item of the TCT to establish their prognostic value, as described earlier.[192]

Limitations

The present study has some potential shortcomings. First, the number of patients is relatively small because the study is primarily descriptive and concerns the course of recovery. A study including a larger number of patients would be more feasible if less outcome measures would be assessed. Because the small numbers have an impact on the statistical analysis, the results need to be interpreted with some caution.

Second, there was a close relationship between the clinical measures, almost all of which had significant correlations. For the subgroup not capable to rise at study entry, correlations for the three function measures ranged from 0.75 to 0.90, correlation for the two participation measures was 0.87, and correlations for the function and participation measures together ranged from 0.35 to 0.60. These high correlations emerge because these measures include items which are not purely related to one ICF domain[168,195-197] or to a construct such as strength and balance, thus resulting in overlap. These correlations combined with the small sample size hamper the use of multivariate analysis.

Future research

We explored the prognostic ability of determinants for STS movement execution, but could also have studied aspects of STS-related functioning with regard to activity (e.g. gait speed and FIM). Also, the relationship between STS movement execution and FIM, BI and gait speed could be evaluated to establish the relevance of STS movement execution in functioning. Furthermore, a study including a larger number of patients with more prognostic variables, but with less outcome measures and less frequent assessment moments, would allow to describe the prognostic variables in more detail.

Conclusions

The present study shows that function measures are prognostic for the capability to execute the STS movement, and that the PASS scale has the highest prognostic value. For rising speed and number of STS movements, both function and activity measures can be used. Further studies are needed to explore other determinants and to confirm these findings in larger cohorts.

8

General Discussion

The main objective of this thesis was to describe the time course of the recovery of STS-related functioning in the first year after stroke in a cohort of patients with a first-ever stroke. We included many aspects of STS-related functioning, but focused on objective assessment and on the STS movement outside the movement laboratory. More specifically, on STS movement execution, which is defined as: *the ability to perform the STS movement, and the quality and quantity of the STS movement, in terms of both capacity and performance.* Concerning quality we studied rising speed and balance control; concerning quantity we studied the number of STS movements during daily life. Prior to the cohort study we performed a literature review on STS movement determinants, and conducted several studies to examine the content of the accelerometric signals and its validity in assessing STS movement execution. In the chapters describing these studies, the results and their related topics have been discussed in detail. In this general discussion, some additional aspects will be addressed.

Importance of STS movement and its assessment

Rising from a seated position is a movement that is performed frequently during daily life. This movement, most often referred to as Sit-to-Stand (STS) movement, is a prerequisite for walking and is needed in self care. It is a complex movement from a stable (sitting) to an unstable (standing) position, requiring coordinated muscular contraction, strength, and balance control. The STS movement is frequently assessed in order to assess these constructs indirectly.[8,9,137,198] Assessment of the STS movement is also encouraged because its duration is associated with, for example, falls,[2,5] self-reported functioning,[9] and gait speed after stroke[22] Our longitudinal cohort study showed that 54% of the patients was not capable to rise shortly (<4 days) after stroke, and at one year post-stroke 17% was still not capable to rise independently.

The relevance of the STS movement is generally accepted in Physical and Rehabilitation Medicine (PRM), because it forms a more or less central part of several clinical assessment tools such as the Timed Get Up & Go Test, the 5 times STS test, the 30 s chair-stand test, the Motor Assessment Scale, the Postural Assessment Scale for Stroke patients, the Functional Independence Measure, etc. By assessment of the STS movement, physicians are able to identify functional limitations in patients, to describe specific STS movement changes in specific patient groups, and to plan and evaluate treatment in patients with STS movement limitations. Galli et al. even proposed to include assessment of the STS movement in every gait evaluation in a movement laboratory.[199]

STS movement assessment in stroke

However, it is remarkable that in outcome studies on stroke, objective assessment of the STS movement has received less attention than gait and mobility. The STS movement might be seen as being less important than gait or mobility in patients with stroke, because of the high incidence of gait disturbances in survivors of stroke. Also, the availability of several treatment strategies for gait disturbances that need evaluation could have contributed to this difference in attention. Furthermore, studying the STS movement in more detail may be hampered by the lack of available instruments that are easy to use, and that are also valid and sensitive to change.

From STS movement capacity to performance

In studies in which the STS movement is (part of) the focus of interest, STS movement *capacity* is usually the main issue. For example, it is a central part of frequently used clinical tests and questionnaires (i.e., the MAS, PASS and FIM), which mainly focus on the ability to (independently) perform the STS movement. These instruments are mostly ordinal and cannot quantify the STS movement capacity in detail or objectively. Objective, quantitative assessment of STS movement capacity is, in general, only applied in experimental studies. The STS movement is then assessed in a movement laboratory; however, this is a cumbersome method for general use because the techniques used are often complex and need technical support, and the subjects are required to come to a hospital for assessment.

Other characteristics of STS movement capacity studies in stroke patients performed so far are that they focus on the chronic phase, that they are mostly cross-sectional, and that they aim to assess the consequences of the stroke or the effects of specific training protocols on the STS movement.[26,27,42,109,131,171,200-202] Finally, studies on the STS movement in stroke patients and on functioning of PRM patients in general, fail to evaluate actual performance, i.e. what patients actually do in their own environment. One of the reasons for this is the limited number of instruments available for assessing performance. In our opinion, the lack of attention paid to actual performance is mainly due to the lack of instruments rather than to a lack of clinical relevance.

It might be assumed that actual performance can be predicted by assessment of capacity. However, there is no evidence for a strong relationship between movement laboratory capacity assessments and self-reported performance on the one hand, and actual performance on the other.[203] In a critical evaluation of clinical gait analysis[32,33] the validity of movement laboratory assessments was discussed from this perspective. Because capacity and performance are different constructs, it can not be assumed that assessment of capacity in the movement laboratory provides valid information on performance. This discussion results in part from the fact that movement behaviour in a movement lab might well differ from movement behaviour outside the lab. A good example of this is the study of Taylor and Dean[34,35] who showed that, although clinic-measured gait speed (i.e. walking speed over 10 meter) was related to gait speed measured in the community, gait velocity during daily life could be overestimated for those who walked slowly. However, assessments of performance are not more important than, nor can they replace, assessment of capacity in the movement laboratory.

Thus, there are several reasons to advocate assessment of the STS movement *outside* the movement laboratory, the most important being that it provides data on 'real' behaviour or actual performance. This information increases the knowledge about performance in relation to subject-related or disease-related variables. In turn, this knowledge can support physicians in their decision as to whether or not a patient performs as expected (based on the known relationships between performance and subject-related or disease-related variables), can help to provide more details to patients and their family, and to evaluate treatment programs. Additionally, these data can increase our understanding about the relationship between capacity assessed in the movement laboratory and performance during daily life.

STS movement assessment outside the movement laboratory

A less bothersome assessment technique, together with a shift of this assessment from inside to outside the laboratory, would be a great advantage in reducing the reported drawbacks of current STS movement capacity assessments. Ambulatory assessment systems using body-fixed sensors present new possibilities for evaluation of STS movement capacity outside the movement laboratory.[37,48,123,204] However, preparatory studies were needed before ambulatory accelerometry could be used for STS movement capacity assessment outside the movement laboratory. After studying the content of the acceleration signal in relation to kinematic data (Chapter 4), followed by a validation study using a reference technique (Chapter 5) and finally a study assessing balance control (Chapter 6), we concluded that ambulatory accelerometry was valid and sensitive for STS movement capacity assessment outside the movement laboratory.

In addition to the above-mentioned advantages, STS movement assessments can now be performed in different environments (e.g., the patient's home, or in a residential care setting); this means that the inclusion criteria for subjects participating in studies will also change. Furthermore, it will increase the willingness of subjects to participate in studies without being confronted with their mobility problems, or the requirement to visit a movement laboratory.

Capacity assessment using standardized conditions

Before using ambulatory accelerometry in our cohort study, criteria for the standardization of the STS movement had to be selected. Studies on STS movement capacity in the movement laboratory revealed several determinants that had a significant influence on the STS movement (Chapter 2). Of these, height of the chair seat, use of arms, and position of the feet were the most relevant. Therefore, if possible, for standardization purposes we used the same chair throughout the follow-up period (i.e. the chair the subject most often uses in daily life), two conditions related to arm use (i.e. with and without use of armrests), and we allowed patients to rise with feet in their preferred position. Patients were allowed to rise at their normal (usual) speed, and sitting upright before rising. In this way, we believe we were able to create a balance between standardization - necessary to assess changes over time - and self-preferred performance.

The conditions rising with and without use of the armrest were included in the protocol for several reasons. First, for assessment of the STS movement we felt it should reflect as closely as possible the daily life execution of this movement by our patients, who very often use their arms. In most experimental studies the use of arms was not allowed, which probably resulted in a change of movement behaviour. Besides that, assessing the movement without arms would have been too great a challenge for our patients, especially shortly after a stroke. In addition, using both these conditions provided information about the differences in STS movement execution. For example, we could show that the mean rising speed of both conditions changed over time (Chapter 6), with a decrease in the difference in rising speed between the two techniques.

Performance Assessment

In our studies, accelerometry was not only used to assess STS capacity outside the laboratory but also to assess actual STS performance, which was operationalised by the number of STS movements and the percentage of standing and walking. The ambulatory system used (the Activity Monitor) detects posture and motions using the signal of five accelerometers in a specific set-up, during prolonged periods of time. The validity of accelerometry (and specifically the system used in our studies) to assess these aspects of STS performance has been examined in several patient groups.[43,44,46,48,49,120] These latter studies showed that, despite deviations in walking patterns, the detection of activity using ambulatory monitoring was still valid. Therefore, we did not consider it necessary to systematically study the validity of the Activity Monitor in stroke patients separately.

As discussed earlier, the interpretation of performance assessments can be hampered by the fact that assessment might be performed without the control of, and out of sight of, the researcher. For example, compensatory movements or extreme slowing down of movements can hinder the interpretation.

Results of STS movement capacity and performance using an ambulatory system

To emphasise the type of data that can be provided by measurements outside the movement laboratory, some results of our applications in our cohort study will be discussed here in addition to the data presented and discussed in Chapter 6. It can be concluded that rising speed, gait speed, and other STS-related functioning measures show a similar course of recovery: i.e. a significant recovery in the first phase that levels off after three months (see Chapter 6). Rising speed still showed a significant recovery after three months, whereas measures on function showed a less pronounced and non-significant change. This could have resulted from compensatory strategies that were still active during this period, resulting in functional recovery but without an underlying change of impairments.[30]

As mean gait speed and mean rising speed showed the same pattern of recovery in our cohort study, we assumed that there is a basic recovery mechanism for both. This suggestion is supported by a supplementary analysis showing a strong relationship between the data for mean rising speed (Pcsu) and mean gait speed at the specific assessment moments, with r^2 0.90 with arm use (see Fig. 1).

Furthermore, an increasing rising speed over time was not related to an increase of the number of STS movements or percentage walking, which may lead to the conclusion that these constructs are independent from each other.

The number of STS movements during daily life showed a very wide range between patients, but the mean number was lower than in healthy subjects. A similar wide range was found for the percentage walking during daytime (Chapter 6). Other researchers also reported large ranges for the number of steps taken during daily life in patients with a stroke.[39,40]

These data could be presented because we were able to perform objective, quantitative assessments outside the movement laboratory. This enabled us to obtain follow-up data of the patients included in our cohort, despite their mobility problems.

Additional data from ambulatory assessment

In addition to the presented data, a number of relevant data on the STS movement execution can be explored, for example: phases of the STS movement, frequency and duration of walking periods, balance control, gait parameters, and asymmetry. As our ambulatory instrument captures acceleration signals in a specific but extensive setup, it can provide much valuable information on body postures and motions.[44,45,48]

Describing phases during rising will be possible, using the two segments (trunk and upper leg) as outlined in Chapter 4. Using the acceleration signals of these segments we are able to define phases such as flexion and extension of the trunk, and the interval between trunk and leg movement initiation, etc.

The frequency and duration of walking periods can also be determined, although we provided data only on the percentage of time walking during daily life. Also, the distribution of these periods can be of interest when related to subject or disease characteristics.

A study assessing balance control using accelerometry demonstrated its validity and sensitivity (Chapter 5). Using the described technique we can present data on balance control changes during the recovery after stroke, and study the relationship between balance control and other STS-related functioning measures.

Asymmetry of posture and movement can be described, owing to the extensive setup of the activity monitor. The magnitude of trunk asymmetry in standing can be analysed using the acceleration signal in the transversal plane; asymmetry will be represented by a change in offset of the signal.[54,57] Also, asymmetry of leg movement during gait can be assessed because the accelerometers are fixated to both legs.[205]

Apart from the additional data on the STS movement, we can also analyze other movements (such as the Stand-to-Sit movement).

These data will be of interest when addressing questions related to these outcome measures and to subject-related and/or disease-related characteristics. We plan to further analyze these topics in the future.

Other systems based on body-fixed accelerometer sensors

In our study we used the Activity Monitor to assess the STS movement execution, but other systems are also used in stroke research. For example, the Activpal[a] is a system based on accelerometry that classifies an individual's activities of daily living in periods of sitting, standing and walking; it also provides the number of steps taken. The typical setup is the attachment of the device to the ventral side of one

a. PAL Technologies Ltd, 141 St James Road, Glasgow G4 0LT, Scotland, UK

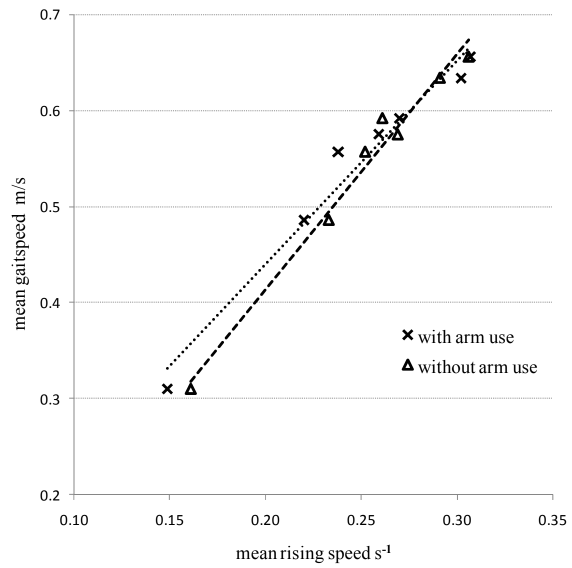

Figure 1 *Scatterplot showing the relationship between mean rising speed and mean gait speed, measured at the different assessment moments, with and without use of arms.*

thigh. It has been tested and used for STS movement detection in stroke patients.[42,206] However, the Activpal is only capable of assessing the number of STS movements during daily life; it does not assess the STS movement duration. Another example is the Stepwatch Activity Monitor (SAM)[b], which is an accelerometer-based activity monitor to assess the number of steps and stepping pattern during daily life, which has been validated for use in stroke patients.[40,207] The SAM is small and lightweight, and is worn at the ankle. Owing to its setup it is not capable of assessing a change of posture or an STS movement.

Finally, there are several other systems that potentially can be used in the assessment of STS movement execution; however, they have not yet been validated for STS movement duration assessment.

Other ambulatory techniques

The instrument we used in our studies and the instruments described in the previous paragraph were based solely on accelerometry. Other ambulatory techniques use accelerometers combined with other sensors to provide angular (gyroscopes) or positional (magnetometer) information to give detailed information on body posture and movement.

b. OrthoCare Innovations, LLC

For example Xsens Technologies b.v.[c] provides systems for ambulatory assessment of human motion (Xbus kit or Moven). Basically it consists of a sensor integrating data from an accelerometer, gyroscope and magnetometer. A real-time software algorithm provides the information on 3-D orientation. Its use has been validated in the movement laboratory, but no prolonged measurements have been presented.[208]

Another activity monitor system, which in fact is validated to assess the STS movement, is the Physilog.[37,121,123] The applied version consists of a combination of a biaxial accelerometer and a gyroscope, which are fixated to the sternum. Its clinical use, with the possibility to detect the STS movement duration during prolonged measurements, was recently described in a geriatric population.[37] It showed STS movement times ranging from 4.3 to 5.6 seconds. The described version can not, however, provide information on balance or asymmetry as no sensor was available for the transversal plane.[37]

Thus, although several ambulatory systems are available, only a few are validated for assessing STS movement capacity and performance during prolonged periods in patients with a stroke.

Why is this study different from previous ones?

The present study emphasizes aspects of the recovery of the STS movement execution in the first year post-stroke which have not yet been explored (such as the rising speed and the number of STS movements during daily life) by making assessments outside the movement laboratory.

By performing the current study we demonstrated that assessment of STS movement execution outside the movement laboratory is feasible. As discussed earlier, the assessment of STS movement execution in a movement laboratory can be complex and cumbersome, and the use of a system using body-fixed sensors provided an alternative. Ambulatory accelerometry has not only proven to be an alternative but also to be a valuable addition to the current assessment systems. As discussed in the section “Additional data from ambulatory assessment”, this system allowed us to collect data that were not attainable in the past. Thus, we presented the number of STS movements, the percentage of time walking and standing in patients with a stroke during daily life.

Our study design also differed from others in that we investigated patients that were included within four days after stroke at a stroke unit. Other studies on outcome following stroke often included patients in a later phase, or a specific selection

c. Xsens Technologies B.V., P.O. Box 559, 7500 AN Enschede, The Netherlands e.g., patients referred to a rehabilitation centre, or not being able to walk after one week. Our design will enable generalization of these results to patients seen at the stroke unit by physicians.

With the frequency of assessments selected in our study, we believe that we have measured the true recovery course because frequent assessments estimate true recovery to a higher degree.[209] This recovery course would have become less evident if assessments were performed, for example, only at 12 and 48 weeks post-stroke. Concerning actual STS performance assessments, we could have used a more frequent assessment which probably would have provided more insight into the relationship between actual STS performance and rising speed and secondary outcome measures.

Clinical implications for training of the STS movement in stroke

Several recommendations on training of the STS movement can be made, based on the results of our cohort study, on general concepts concerning training, and the results of other studies.

We recommend that, from early post-stroke, the patient should be stimulated to perform the STS movement as frequently as possible (if this can be done independently) in order to exercise balance, prevent loss of muscle strength and/or to improve strength. We also suggest to compensate for functional limitations in order to enable the STS movement, because of the task specificity of training.[210] Therefore, we suggest that arm support should be allowed, so that the patient can compensate for loss of balance and muscular strength. Using arm support with a hemiparesis will probably induce asymmetry, but this should be accepted when stimulating early exercise. Furthermore, earlier studies have reported the possibility to reduce acquired asymmetry at a later stage by the use of exercises. With feet repositioning a patient can compensate for loss of muscular strength and improve their balance; the patient should be advised on how to use compensatory techniques which reduce the necessary movement of the centre of body mass.[11,101] This is especially relevant when a wheelchair is used for mobility, because a wheelchair often inhibits feet repositioning due to the presence of calf support. Increasing the seat height will lower the required strength and speed, and reduce the amount of balance control needed at the end of rising. The seat height of a wheelchair is, however, often lowered to enable the use of the leg for wheelchair propulsion; this can hamper an appropriate starting position for STS movement execution and training. With the task specificity of training in mind[210] we also strongly recommend that all professionals involved use a standardised strategy or protocol. Moreover, the importance of the frequency of STS movement training has been emphasized in two studies.[41,42]

Therefore, the best training conditions for the STS movement requires a change from a wheelchair to a chair, frequent training, and a standardised strategy to be used by all professionals.

Clinical implications for STS movement assessment in general

For a systematic assessment of the STS movement and its determinants we suggest to thoroughly investigate the STS movement, together with muscular strength and balance, using standardized measures.[6,211-213] This includes performing the STS movement using a standardized seat height, with feet in the preferred position, and at a comfortable speed. In order to place more stress on the actual STS movement, the use of arms or the speed can be constrained, for example rising fast without the use of arms. Also, the height of the chair seat can be lowered. A pass or fail strategy could be applied in these circumstances.[13] Any modifications to the task should be assessed and recorded.[11,13,14] Furthermore, it is advised to assess the 5-STS time as a measure for STS movement execution,[8,9] together with a Timed Up & Go test.[36] Because STS movement execution is obviously related to the level of functioning of a subject, its quantitative assessment is justified by the ease of application and relevance. Implementation of these measures will increase awareness of this important movement and its relevance for functioning.

Systematic assessment of the STS movement should, preferably, form a part of the diagnostic and evaluative tools of physicians in Physical and Rehabilitation Medicine.

Future research

The present study has focused on several aspects of the recovery of the STS movement execution, but many aspects still remain to be explored or presented. From a clinical point of view additional questions need to be addressed about recovery, as well as the relationship between STS movement execution and STS-related functioning in the first year after a stroke. Also, the relationship between limitations of strength and balance on the one hand, and STS movement capacity and performance on the other needs further research in larger study samples. In addition, specific STS movement training programs for patients with a stroke need to be developed and evaluated.

Summary

A stroke results in a decrease of the capability to perform a Sit-to-Stand (STS) movement and, in the acute phase, often in complete impossibility to do so. The STS movement is regarded as a fundamental activity of daily living, a prerequisite for walking and standing; moreover, a secure execution of the STS movement is an important component in relation to physical safety. In the acute period after the stroke, recovery can occur in several bodily functions resulting in a change of the capability to execute an STS movement. Most people regain the ability to execute this movement, but it frequently remains distorted in the sense of a slowing down or being asymmetric with regard to weight bearing. As a result of the changed bodily functions, the distorted STS movement and together with other consequences of the stroke, patients are frequently limited in their activities and social participation. Several studies have focused on the STS movement, but almost all were cross-sectional in design, and measurements were done in the sub-acute or chronic phase.

To study the course of the recovery of the STS movement in the first year post stroke and its determinants, we designed and conducted a longitudinal prospective cohort study. To assess the STS movement execution we used ambulatory accelerometry, which enables objective assessment of body postures and motions during daily life (to assess the number of STS movements during daily life) as well as assessment of STS movement outside a movement laboratory.

The work for this thesis was made possible by a personal grant from the Netherlands Organization for Health Research and Development (ZonMw) which was awarded for a clinical research position within the program Rehabilitation Research.

Chapter 1 provides an introduction of the STS movement and its position in the International Classification of Functioning (ICF). In this model the movement is positioned in the Activity domain with two qualifiers: capacity and performance. Assessments can focus on objective measurements of what a person actually is able to do or does, or on subjective self-reports about one's capabilities and behaviour. In the ICF model actual STS *capacity* comprises both quality and/or quantity of STS movement execution as (objectively) measured in a 'standardized' environment (i.e. rising speed and capability). The actual STS *performance* also comprises the actual quality and/or quantity of STS movement execution as (objectively) measured, but now in the subject's own environment during normal daily life (in this thesis: the number of STS movements). We planned to study the capability to execute the STS movement, the rising speed, and the number of STS movements during daily life, which together we called '*STS movement execution*'. Until now, most studies on STS movement after stroke focused on the aspect 'quality' of the actual STS capacity in movement laboratories. Secondly, it was studied to a certain extent as part of clinical outcome measures which have items on the STS movement, for example in the Motor Assessment Scale (MAS), the Postural Assessment Scale for Stroke Subjects (PASS), and the Functional Independence Measure (FIM). Therefore, this current study on recovery of the STS movement will serve to counteract the current deficit in information on STS movement execution.

To gain insight into the factors that might influence the STS movement in ambulatory conditions, we reviewed the literature on the constraints used in STS movement research. A total of 39 experimental studies were studied in detail and we were able to define chair-related, subject-related and strategy-related determinants of the STS movement. The results of this review are presented in ***Chapter 2***; also provided are data on the subjects studied, the assessment techniques used and the parameters studied in the movement laboratory. Most studies were performed in healthy subjects, investigating small groups of persons using several constraints. Almost all studies were performed without the use of arm support. These studies indicated that chair seat height, use of armrests, and foot position have a major influence on the way the STS movement is executed. Using a higher chair seat resulted in lower moments at knee level (up to 60%) and hip level (up to 50%); lowering the chair seat increased the need for momentum generation or repositioning of the feet to lower the needed moments. Using the armrests lowered the moments needed at the hip by 50%, probably without influencing the range of motion of the joints. Repositioning of feet influenced the strategy of the STS movement, enabling lower maximum mean extension moments at the hip.

As we planned to use ambulatory accelerometry in our longitudinal study, in ***Chapter 3*** we addressed a well-known problem that arises when using piezo-resistive acceleration signals for movement analysis. The acceleration signal is composed of different acceleration components, i.e. a gravitational and inertial component. The gravitational component depends on the position of the sensor in the gravitational field, therefore it can provide angular information of the body part on which it is fixated. However, the inertial component may complicate the interpretation of the acceleration signals if one is primarily interested in the angular information. For our study we were interested in both angular information related to the STS movement and in the shape of the acceleration signal related to event detection. To understand the relationship between acceleration signal and angular positions, we needed to gain insight into the components of accelerometer signals from the trunk and thigh segments during STS movements under four conditions (self-selected speed, slow speed, fast speed, and full-flexion strategy). We decomposed accelerometer signals from the trunk and thigh (in the sagittal direction) using kinematic data obtained from an opto-electronic device. This was studied in nine healthy subjects, performing each condition six times. The acceleration signals were decomposed into the gravitational and inertial components. Subsequently, the inertial component of the trunk was decomposed into rotational and translational components. The accelerometer signals could be reliably reconstructed, and were highly characteristic and repeatable for the STS movement. The contribution of the inertial component was larger in the trunk signal than in the thigh signal and increased with higher speeds. This study was performed using the trunk and thigh segments, without relating the trunk and thigh segment in, for example, trunk-thigh angle. This was primarily done because the aim of the study was to gain information about the contributing acceleration components per segment to evaluate the possibility to use the acceleration signal for angular information, and to explore the start and the end of the movement of the individual segments. The results supported the use of acceleration signals in STS movement analysis.

Since these acceleration signals have not yet been used for determination of the STS movement duration we needed to perform a validation study, which is presented in ***Chapter 4***. First we had to develop an algorithm to detect the start and end of the movement of the trunk and thigh segment using the acceleration signals in the sagittal direction. We compared accelerometric and opto-electronic assessment of the STS movement duration under four conditions (comfortable speed, slow speed, fast speed, and exaggerated trunk flexion) with six healthy subjects and six subjects with stroke who performed the movements six times under each condition. The data of the accelerometric and the reference method of STS movement duration were strongly related (r=0.98). Accelerometry showed a fixed bias of 0.07 s in healthy subjects and 0.32 s in stroke subjects. In healthy subjects, a significant negative proportional bias of 0.1 was detected. Thus in stroke subjects accelerometry overestimated the STS movement duration but without significant proportional bias. Accelerometry showed the ability to discriminate between stroke and healthy subjects, and between speed conditions. In ***Chapter 4*** we concluded that, although there is no total agreement, accelerometry can provide valid data on the STS movement duration in a longitudinal study in stroke subjects, where fixed bias would be present for all.

Accelerometry is a method of interest because potentially it can provide information on more objective STS parameters than the STS movement duration alone. ***Chapter 5*** presents a study on the potential of accelerometry to ambulatorily measure balance control in a simple and inexpensive way. Balance control was defined as high frequency body sway. From the transversal acceleration signal of the trunk the high frequency component was used to calculate balance control parameters: the Root Mean Square (RMS) and Area under the Curve (AUC). We wanted to determine and compare the sensitivity of these accelerometric balance parameters during the STS movement. Eleven healthy subjects (4 males, 28.2 ± 7.9 years) were included. The healthy subjects performed STS movements in four conditions with different levels of difficulty. Accelerometers were attached to the trunk, and simultaneously force plate measurements were made. An additional 31 patients (21 males, 10 females; 63.3 ± 12.8 years) participating in the cohort study on acute stroke were included in this study. Data of the patients were compared: a) with healthy subjects, b) between patient subgroups, and c) between different phases of recovery, to assess the sensitivity of accelerometry to measure differences in balance control. In all comparisons there was a significant difference in AUC data ($p < 0.05$), and AUC appeared to be more sensitive than RMS. Variability in AUC was not completely or mainly the result of changes and differences in the duration of the STS movement. We concluded that ambulatory accelerometry might be a valuable extension of the set of instruments currently used in balance control studies. It has the advantage that it can be used in the assessment of static and dynamic balance, outside the laboratory, and during a prolonged time period.

Chapter 6 presents the results of the longitudinal prospective cohort study on the course of recovery of the STS movement execution and the STS-related functioning after acute stroke. The course of the recovery was primarily studied from the perspective of the capability to carry out an STS movement independently, the speed of rising (Pcsu), and the actual STS performance during normal daily life. A total of 50 subjects (mean age 62.2 years, range 28-87) were included within 4 days after stroke during their stay on the Stroke Unit of the Erasmus MC. Assessments were performed at week 0, 3, 6, 9, 12, 24, and 48 weeks post-stroke. We measured the capability to rise independently, Power chair stand up (Pcsu=1/STS movement duration), and the number of STS movements during daily life. The secondary outcome measures were percentage walking and standing during daily life, and several STS-related functioning measures. These included the Motor Assessment Scale (MAS), the Trunk Control Test (TCT), and the Postural Assessment Scale for Stroke patients (PASS), a test for postural control consisting of two components related to maintaining and changing posture. Furthermore the Barthel Index (BI), the Functional Independence Measure (FIM), and the five-meter walking test (5mWT) at comfortable speed were used.

During the first year post stroke the number of patients capable to rise independently increased significantly during the first three months (from 54 to 81%, $p = 0.00$); in the same time interval the mean Pcsu changed from 0.15 to 0.27 s^{-1} ($p = 0.00$) while the number of STS movements increased from 10.6 to 17.7 ($p = 0.004$) during daytime. Also, the percentage of walking and standing during daily life changed significantly in that period. STS-related functioning changed for the larger part during the first three months after stroke. During the interval 12-24 and 12-48 weeks a significant change of mean Pcsu, gait speed and BI was noted. This could have resulted from compensation strategies, because the MAS, PASS and TCT scores did not show changes of the same magnitude.

Regarding the course of recovery, we could conclude that the main change occurred during the first three months with a levelling off at three months; however, outcome measures such as the Pcsu and gait speed still showed significant changes after three months but without simultaneous changes in the MAS or PASS.

In ***Chapter 7*** we studied the variables related to the recovery of the STS movement in more detail. In those subjects not capable to rise at inclusion in the study, we analysed which variable at the moment of inclusion was the best predictor for the capability to perform the STS movement one year post stroke. The Postural Assessment Scale for Stroke patients (PASS), a test for postural control consisting of two components related to maintaining and changing posture, had the highest prognostic value. The influence of the variables on the course of recovery of the capability to rise was assessed using survival techniques, and assessing hazard ratios. The variable PASS showed the most significant effect, expressed as hazard ratio, on the course of recovery. Furthermore PASS showed to be predictive for rising speed and number of STS movement in daily life at 12 and 48 weeks.

Finally, we should mention that this analysis on the determinants of the recovery of the STS movement was of an explorative character due to the limited number of patients included in this study, and the number of subjects recovering. Therefore, further research on these topics is still required.

Samenvatting

Een beroerte verstoort de wijze waarop een patiënt vanuit zit opstaat, en in de acute fase van een beroerte is er vaak een volledig onvermogen om op te staan. Het vanuit zit opstaan is een essentiële activiteit van het dagelijks leven en is een voorwaarde voor het gaan staan en lopen. Veilig opstaan vanuit zit is een belangrijk onderdeel van het zelfstandig functioneren. In de periode na een beroerte treedt herstel op van een aantal lichamelijke functies, zoals kracht en balans, samen met een verandering van de activiteiten, waaronder het vanuit zit opstaan. De meeste patiënten kunnen na verloop van tijd vanuit zit opstaan, maar vaak is er een blijvende verandering: de snelheid neemt af of er is sprake van een asymmetrische belasting van de benen tijdens deze beweging. Patiënten worden door de combinatie van veranderd lichamelijk functioneren, een verstoord opstaan vanuit zit en andere gevolgen van de beroerte vaak beperkt in hun participatie. Het vanuit zit opstaan na een beroerte is diverse malen onderwerp van onderzoek geweest, maar voornamelijk in cross-sectionele studies. Deze studies vonden plaats in de subacute of chronische fase van de beroerte.

In een longitudinale prospectieve cohort studie hebben we het beloop en de determinanten van het herstel van het opstaan vanuit zit in het eerste jaar na een beroerte bestudeerd. Om het opstaan vanuit zit te beoordelen hebben we gebruik gemaakt van draagbare versnellingssensoren, waarmee we de houdingen en de bewegingen van de patiënten objectief vast konden leggen. Hierbij kon zowel de duur van het opstaan vanuit zit als het aantal malen opstaan vanuit zit worden vastgelegd, buiten een bewegingslaboratorium.

Hoofdstuk 1 beschrijft het opstaan vanuit zit en haar plaats binnen de 'International Classification of Functioning, Disability and Health' (ICF) van de Wereld Gezondheids Organisatie. In dit model is het opstaan vanuit zit gepositioneerd in het Activiteiten domein met twee typeringen: '*capacity*' (vermogen) en '*performance*' (uitvoering), ofwel wat iemand kan en doet. Een activiteit kan worden beoordeeld door a) objectief bepalen wat iemand kan of doet of b) subjectief bepalen van vermogen en uitvoering met behulp van zelfrapportage. Het begrip '*capacity'* (vermogen) omvat zowel de kwaliteit als de kwantiteit van het opstaan uit zit, zoals deze objectief is te meten onder "standaard" omstandigheden (bijvoorbeeld de opsta snelheid en het vermogen om op te staan). De '*performance'* (uitvoering) omvat ook de kwaliteit en kwantiteit, maar dan van de werkelijk uitgevoerde opstabeweging in de omgeving van de patiënt gedurende het dagelijkse leven. In dit proefschrift is dit geoperationaliseerd als het aantal malen opstaan per dag. We bestudeerden het vermogen om vanuit zit op te staan, de snelheid van opstaan en het aantal malen opstaan vanuit zit gedurende het dagelijkse leven. Als overkoepelende term voor '*capacity*' en '*performance*' hebben we de term '*STS movement execution*' gebruikt. De meeste onderzoeken van het vanuit zit opstaan hebben zich tot op heden gericht op de 'kwaliteit' van de opstabeweging in bewegingslaboratoria. Daarnaast wordt deze beweging ook gebruikt als onderdeel van klinische uitkomstmaten, die vaak items met opstaan vanuit zit bevatten, zoals bijvoorbeeld in de Motor Assessment Scale (MAS), de Postural Assessment Scale voor Stroke Subjects (PASS), en de Functional Independence Measure (FIM). Onze studie, gericht op het herstel van het opstaan vanuit zit na een beroerte, heeft een aanvulling gegeven op de huidige kennis van de 'STS movement execution'.

Om goed geïnformeerd te zijn over de factoren die het opstaan vanuit zit in het dagelijks leven kunnen beïnvloeden hebben we een literatuurstudie verricht naar de meetomstandigheden bij het onderzoek van deze beweging in bewegingslaboratoria. We bestudeerden in totaal 39 studies die gebruik maakten van manipulaties bij het opstaan (bijvoorbeeld de hoogte van de stoel). We deelden de relevante factoren in drie groepen in: factoren toe te schrijven aan de stoel, de proefpersoon en de strategie. De resultaten van deze studie worden gepresenteerd in ***Hoofdstuk 2***, waarin ook het type proefpersoon, de meettechniek en de bestudeerde aspecten van het opstaan vanuit zit worden beschreven. De meeste studies werden verricht bij kleine groepen gezonde proefpersonen met diverse manipulaties van het opstaan zonder het gebruik van de armen bij het opstaan. De studies toonden aan dat de hoogte van de zitting van de stoel, het gebruik van de armleuningen en de positie van de voeten een belangrijke invloed hebben op de wijze waarop deze beweging wordt uitgevoerd. Het gebruik van een hogere stoelzitting resulteerde in lagere momenten op knie niveau (tot 60 % reductie) en heup niveau (tot 50 % reductie). Het verlagen van de zitting van de stoel leidde tot een toename van 'momentum generation' strategie en een 'herpositionering' van de voeten om de benodigde momenten te kunnen leveren. Het gebruik van de armleuningen verlaagde de benodigde momenten op het niveau van de heup met 50 %, zonder een duidelijke beïnvloeding van de bewegingsuitslag van de gewrichten. Herpositioneren van de voeten beïnvloedde de strategie van het opstaan waardoor lagere extensiemomenten bij de heup noodzakelijk waren voor deze beweging.

Aangezien we in onze longitudinale studie ambulante accelerometrie wilden gebruiken hebben we in ***Hoofdstuk 3*** een bekend probleem van het gebruik van versnellingssignalen bij bewegingsanalyse onderzocht en beschreven. Het versnellingssignaal is samengesteld uit verschillende versnellingscomponenten, te weten een zwaartekracht- en inertiecomponent. De zwaartekrachtcomponent hangt samen met de positie van de versnellingsopnemer in het zwaartekrachtveld, waardoor het informatie geeft over de hoek van het lichaamssegment waarop deze opnemer is bevestigd. De inertiecomponent kan het interpreteren van het versnellingssignaal bemoeilijken, zeker wanneer men is geïnteresseerd in de informatie over de positie (hoek) van lichaamssegmenten. In verband met het vaststellen van bepaalde ‘sleutel’ momenten (tijdstippen) voor het romp- en beensegment tijdens het opstaan waren wij zowel geïnteresseerd in informatie over de hoek, gerelateerd aan het opstaan vanuit zit, als in de vorm van het versnellingssignaal. Voor het begrijpen van de relatie tussen het versnellingssignaal en de hoekinformatie was het noodzakelijk inzicht te krijgen in de componenten van de versnellingssignalen van het romp- en beensegment bij het opstaan onder vier condities (zelfgekozen snelheid, langzaam, snel, sterke romp flexie conditie). Met aanvullende kinematische informatie verkregen door middel van opto-electronische apparatuur waren wij in staat de versnellingssignalen (in het sagittale vlak) te analyseren. Deze studie werd verricht bij negen gezonde proefpersonen die per conditie zes maal opstonden. De versnellingssignalen werden ‘ontbonden’ in de zwaartekracht- en inertiecomponent. De versnellingssignalen konden goed worden gereconstrueerd en waren karakteristiek en reproduceerbaar voor het opstaan vanuit zit. In het versnellingssignaal van het rompsegment was de bijdrage van de inertiecomponent groter dan in het beensegment, en deze bijdrage nam toe bij hogere snelheden. De resultaten ondersteunen het gebruik van accelerometrie bij de analyse van het opstaan vanuit zit.

Aangezien versnellingssignalen tot op heden niet zijn gebruikt bij het meten van de duur van het opstaan vanuit zit was het noodzakelijk de mogelijkheid hiervan te bestuderen. Dit wordt in ***Hoofdstuk 4*** beschreven. Allereerst hebben wij een algoritme ontwikkeld om met behulp van versnellingssignalen in het sagittale vlak het begin en het einde van de beweging van het romp- en beensegment te bepalen. We vergeleken de duur van het opstaan bepaald met ambulante accelerometrie en een opto-electronisch apparaat onder vier condities (zelfgekozen snelheid, langzaam, snel en met sterke romp flexie). Deze studie vond plaats bij zes gezonde personen en bij zes personen met een beroerte, waarbij per conditie zesmaal werd opgestaan. De meetgegevens van de accelerometrie en de referentiemethode waren sterk gecorreleerd (r=0.98). Accelerometrie vertoonde een ‘fixed bias’ van 0.07 s bij gezonde personen en van 0.32 s bij de personen met een beroerte. Bij de gezonde personen was er een significante ‘proportional bias’ van 0.10. Bij personen met een beroerte gaf accelerometrie een overschatting van de duur van het opstaan, zonder significante ‘proportional bias’. Accelerometrie was in staat onderscheid te maken tussen gezonde personen en personen met een beroerte, evenals tussen condities met verschillende snelheden. We concludeerden dat ondanks de afwezigheid van volledige overeenstemming, accelerometrie gebruikt kan worden in een longitudinale studie bij personen met een beroerte, waar de ‘fixed bias’ voor allen geldt.

Accelerometrie is een techniek die in potentie meer objectieve parameters over het opstaan vanuit zit kan geven dan alleen de duur van deze beweging. In ***Hoofdstuk 5*** wordt een studie gepresenteerd naar een 'eenvoudige' en goedkope mogelijkheid van het ambulant meten van de balans controle met behulp van accelerometrie. Balans controle werd gedefinieerd als 'body sway' met een hoge frequentie. De hoogfrequente component van het versnellingssignaal in het transversale vlak van de romp werd gebruikt om de parameters van de balans controle te berekenen: de 'Root Mean Square' (RMS) en de 'Area under the Curve' (AUC). De gevoeligheid van deze parameters gedurende het opstaan vanuit zit werd vastgesteld en vergeleken. Gezonde personen (n=11, leeftijd 28.2 ± 7.9 jaar) stonden vanuit zit op onder vier omstandigheden, met een toenemende moeilijkheidsgraad. Er werden gelijktijdig metingen verricht met accelerometrie en een krachtenplatform. Verder werden 31 patiënten met een beroerte, deelnemend aan een cohort studie, geïncludeerd (21 mannen, 10 vrouwen, leeftijd: 63.3 ± 12.8 jaar). Om het onderscheidend vermogen van accelerometrie te bepalen, werden de resultaten van de patiënten vergeleken: a) met gezonde proefpersonen, b) tussen subgroepen van patiënten en c) tussen verschillende fasen van herstel. Bij alle vergelijkingen toonde de AUC een significant verschil en bleek de AUC meer onderscheidend te zijn dan de RMS. De verschillen in de AUC waren niet alleen het gevolg van verschillen in de duur van het opstaan. We concludeerden dat ambulante accelerometrie een waardevolle aanvulling is op de instrumenten die gebruikt kunnen worden bij studies naar balans controle. Accelerometrie heeft het voordeel dat het gebruikt kan worden voor het meten van zowel statische als dynamische balans, buiten een bewegingslaboratorium en gedurende langdurige metingen.

De resultaten van de longitudinale prospectieve cohort studie naar het beloop van het herstel van het opstaan vanuit zit na een beroerte en het aan het opstaan gerelateerde functioneren worden gegeven in ***Hoofdstuk 6***. Het beloop van het herstel werd primair bestudeerd met aandacht voor het vermogen onafhankelijk vanuit zit op te staan, de snelheid van het opstaan en het aantal malen opstaan gedurende het dagelijks leven. In totaal werden 50 patiënten (gemiddelde leeftijd 62.2 jaar, range 28-87 jaar) geïncludeerd binnen vier dagen na een beroerte, gedurende hun verblijf op de Stroke Unit van het Erasmus MC. Metingen werden verricht op het tijdstip van de inclusie en op 3, 6, 9, 12, 24 en 48 weken na de beroerte. Tevens werd het aan het opstaan gerelateerde functioneren beoordeeld met de Motor Assessment Scale (MAS), de Trunk Control Test (TCT) en de Postural Assessment Scale for Stroke patients (PASS). Verder werden de Barthel Index (BI), de Functional Independence Measure (FIM), en de loopsnelheid (comfortabele snelheid, 5 meter looptest) bepaald. In de eerste 3 maanden van onze follow-up nam het aantal patiënten dat onafhankelijk kon opstaan significant toe (van 54 tot 81%, p = 0.00). In dezelfde periode nam de snelheid van opstaan toe van 0.15 naar 0.27 s^{-1} (p = 0.00), terwijl het aantal malen opstaan gedurende de dag toenam van 10.6 tot 17.7 (p = 0.004). Ook het percentage van de tijd dat patiënten lopen en staan gedurende het dagelijks leven nam significant toe in deze periode. De grootste verandering van het aan het opstaan gerelateerde functioneren vond plaats in de eerste drie maanden na de beroerte. Gedurende het interval 12-24 en 12-48 weken vonden wij een significante verandering van de snelheid van opstaan, de loopsnelheid en BI. We vonden geen veranderingen in dezelfde orde van grootte in de MAS, PASS en TCT, wat erop zou kunnen wijzen dat er een effect is van compensatiestrategieën. Ten aanzien van het beloop van het herstel konden wij concluderen dat de belangrijkste verandering optrad gedurende de eerste drie maanden na de beroerte met een afvlakking van herstel bij drie maanden, waarbij echter na drie maanden nog significante verandering van de snelheid van opstaan werd vastgesteld

In ***Hoofdstuk* 7** worden de variabelen die zijn gerelateerd aan het herstel van het opstaan vanuit zit verder bestudeerd. Wij bestudeerden welke variabele op het moment van inclusie de beste voorspeller was voor het onafhankelijk kunnen opstaan een jaar na de beroerte; bij die patiënten die bij inclusie niet in staat waren op te staan. De PASS (een test voor balans, bestaande uit twee componenten gerelateerd aan houdingscontrole en houdingsverandering) had de hoogste voorspellende waarde. Door middel van survival technieken, waarbij de hazard ratio's werden bepaald, werd de invloed van de variabelen op het beloop van het herstel van het vermogen op te staan onderzocht. De variabele PASS vertoonde het meeste effect op het beloop van het herstel. Verder was de PASS voorspellend voor de snelheid van opstaan en het aantal malen opstaan gedurende de dag bij 12 en 48 weken. Deze analyse van de voorspellende waarde van de verschillende variabelen had een exploratief karakter tengevolge van het aantal geïncludeerde patiënten en het aantal patiënten dat herstel vertoonde. Verder onderzoek naar voorspellende variabelen en de relaties tussen de diverse variabelen op de verschillende momenten is gewenst.

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Curriculum vitae

The author was born in Son en Breugel on 1st September 1959. From 1971 to 1977 he attended *Het Mgr. Zwijsen College* (VWO) at Veghel. In 1977 he started his medical education at the *Katholieke Universiteit* Nijmegen, where he obtained his medical degree in 1985.

The author worked in the rehabilitation centre "*de Kastanjehof*" in Apeldoorn for one year. He started his residency for Physical Rehabilitation & Medicine in 1986 in Arnhem, followed by Zwolle and Enschede (with drs. HWCM Vos, drs. CGM Warmerdam and prof.dr. G Zilvold, respectively, as trainer), and completed his training in 1990. From 1990 to 1994 he worked as staff member of the Rehabilitation Center "*Het Roessingh*" at Enschede, allocated to the *Medisch Spectrum Twente* hospital in Enschede. In October 1993 he became qualified to practice the specialty of Physical Medicine and Rehabilitation by meeting the requirements of the European Board of Physical Medicine and Rehabilitation. Since 1994 the author has been a staff member of the department of Rehabilitation Medicine of the Erasmus MC in Rotterdam, with fields of interest focusing on neuromuscular disorders, treatment of spasticity, and congenital malformations of the upper extremity. He became deputy trainer (*waarnemend opleider*) in Rehabilitation Medicine for the Erasmus MC in 2005.

In 1996 and 1997 he participated in the national Scientific Course "VRA SGO". From 2001 to 2005 he was a board member of the VRA (*Vereniging voor revalidatieartsen*) and on behalf of this board a member of the *Landelijke Commissie Revalidatieonderzoek*.

From 1994 onwards he has been a member of the advisory board of the Dutch Neuromuscular Society, and after being chair of the board for several years is now vice-chair. Since 2004 he has been medical advisor of the working group 'Charcot-Marie-Tooth' of the Dutch Neuromuscular Society, and is also a member of the ISNO Dutch Neuromuscular Research Support Centre.

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The author is married and has two daughters.

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