Ambulatory monitoring of mobility-related activities in rehabilitation medicine

Het onderzoek dat in dit proefschrift is beschreven en de publicatie van dit proefschrift is financieel ondersteund door:

Algesiologisch Instituut
Jaeger Toennies Benelux
Ministerie van Economische Zaken
Pijn Kennis Centrum Rotterdam
Rotterdamse Stichting voor Cardiologische Revalidatie
TEMEC Instruments

ISBN 90 5166 664 0

Uitgeverij Eburon
Postbus 2867
2601 CW Delft
info@eburon .nl

Illustratie omslag: Coureur muni de chaussures exploratrices et portant l' appareil incripteur du rhytme de son allure. Uit: E.J. Marey, La méthode graphique dans les sciences expérimentales et principalement en physiologie et en médecine. Paris: G. Masson, 1885

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Ambulante registratie van mobiliteit-gerelateerde activiteiten binnen de revalidatiegeneeskunde

Proefschrift

ter verkrijging van de graad van doctor
aan de Erasmus Universiteit Rotterdam
op gezag van de Rector Magnificus
Prof. dr P.W.C. Akkermans M.A.
en volgens besluit van het College voor Promoties.

De openbare verdediging zal plaatsvinden op
donderdag 26 november 1998 om 13.30 uur

door

Johannes Bernardus Josephus Bussmann

Geboren te Etten-Leur

Promotiecommissie

Promotor:	Prof. dr H.J. Stam
Overige leden:	Prof. dr A.P. Hollander
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	Prof. dr ir C.J. Snijders

Het duurt altijd langer dan je denkt,
ook als je denkt
het zal wel langer duren dan ik denk
dan duurt het toch nog langer
dan je denkt.

Judith Herzberg, *Liedje*

Contents

Manuscripts

The following manuscripts are part of this thesis:

1. Bussmann JBJ, Stam HJ. Techniques for measurement and assessment of mobility in rehabilitation medicine: a theoretical approach. Clin Rehabil 1998; 12: 513-522 *(chapter 2)*

2. Bussmann JBJ, Veltink PH, Koelma F, Lummel van RC, Stam HJ. Ambulatory monitoring of mobility-related activities; the initial phase of the development of an Activity Monitor. Eur J Phys Med Rehabil 1995; 5: 2-7 *(chapter 3)*

3. Veltink PH, Bussmann HBJ, Vries de W, Martens WLJ, Lummel van RC. Detection of static and dynamic activities using uniaxial accelerometers. IEEE Rehab Eng 1996; 4: 375-385 *(chapter 3)*

4. Bussmann HBJ, Reuvekamp PJ, Veltink PH, Martens WLJ. Validity and reliability of measurements obtained with an 'Activity Monitor' in people with and without a transtibial amputation. Phys Ther 1998; 78: 989-998 *(chapter 4)*

5. Bussmann JBJ, Tulen JHM, Herel van ECG, Stam HJ. Quantification of physical activities by means of ambulatory accelerometry: a validation study. Psychophysiology 1998; 35: 488-496 *(chapter 5)*

6. Bussmann JBJ, Laar van de YM, Neeleman MP, Stam HJ. Ambulatory accelerometry to quantify motor behaviour in patients after failed back surgery: a validation study. Pain 1998; 74: 153-161 *(chapter 6)*

7. Berg van den-Emons RJG, Bussmann JBJ, Keijzer-Oster DJ, Balk AH, Stam HJ. Everyday physical activity in chronic congestive heart failure.as measured with a novel Activity Monitor. Am J Cardiol (submitted) *(chapter 8)*

8. Bussmann JBJ, Damen L, Stam HJ. Analysis and decomposition of signals obtained by thigh-fixed accelerometry during walking. Med Biol Eng Comp (to be submitted) *(chapter 9)*

9. Bussmann JBJ, Hartgerink I, Woude van der LHV, Stam HJ. Feasibility of accelerometer signals for measurement of physical strain in ambulation. Med Sci Sports Exerc (to be submitted) *(chapter 10)*

10. Bussmann JBJ, Berg van den RJG, Angulo S, Moerland C, Stam HJ. Feasibility of ambulatory measurement of prosthetic gait in the early phase of rehabilitation. Am J Phys Med Rehab (to be submitted) *(chapter 11)*

General introduction and outline

Introduction

Rehabilitation medicine is fundamentally aimed at the consequences of diseases, trauma, congenital anomalies, and impairments.[e.g. 8,36,40,48,55,80,92,93] It is obvious, therefore, that many measurement instruments within medicine are focussed on these consequences; questionnaires and scales are the techniques most used. Data provided by these instruments are, for example, used to evaluate interventions and natural recovery of patients, and to aid in deciding on treatment strategy and in prognosis.[e.g. 8,34,36,49,58,93]

Other medical disciplines traditionally also use discipline-specific measures: e.g. the cardiologist measures heart function by means of Doppler techniques, the orthopaedic surgeon measures joint mobility and muscle force, and the anaesthesiologist measures pain and analgesia. Also, within many medical disciplines interest is shifting towards the focus of rehabilitation: namely, disability and handicap.[74] However, this tendency can not be dealt with using the traditional measures, which do not have a direct relation with the measures on more functional levels.[e.g. 12,56,80] Therefore, development of reliable, sensitive, and valid instruments at the level of disability and handicap is of utmost importance, especially as there is a lack of these instruments, even within rehabilitation medicine.[e.g. 21,30,34,37,40,49,81]

Disability is closely related to daily functioning or daily activities, which can be operationalised in several ways. The perspective used in this thesis is to regard daily functioning as a complex whole of postures, transitions between postures, and movements. In line with the terminology used in rehabilitation medicine, we have chosen for the concept *mobility-related activities*, where mobility is defined as 'the process of moving oneself, and of maintaining and changing postures'.

Ambulatory monitoring means continuous observation of free-moving subjects in everyday life,[27] and enables measurements to be performed on persons without being space-bound by instruments, cables etc. This technique is, therefore, potentially suitable for objective measurement of mobility-related activities during daily life; quantitative data are stored directly in a memory unit, without the intervention of a patient, researcher or observer. Due to technological developments (more and more possibilities of smaller and smaller instruments and sensors) an instrument to measure mobility-related activities could be developed. This instrument, the Activity Monitor (AM), will be the thread which runs throughout this thesis. The AM is an instrument based on long-term ambulatory accelerometry, and aimed at the measurement of quantity, quality, and physical strain of mobility-related activities. *Quantity* refers to which activity is performed, when, for how long, and how often. *Quality* is defined by 'the way activities are performed'; examples of quality are speed, symmetry, and phasing of movements and transitions. *Physical*

strain is regarded as the reaction of the body due to the performance of an activity. The relationships between these constructs can be clarified using a stress-strain-capacity model:[20,61,99] the quantity and quality of activities can be regarded as stressors which, dependent on factors such as the physical work capacity of the person involved, cause a certain level of physical strain. These three concepts (quantity, quality, physical strain) can be considered important in rehabilitation: e.g. treatment and training is often aimed at improvement of activity level (quantity) or movement pattern and co-ordination (quality), or at reduction of physical strain by increasing physical work capacity.

In the selection of the set of activities that need to be studied, a number of criteria have been used: the activities have to be related to mobility; the set of activities has, as far as possible, to cover all commonly occurring daily activities; and the set has to be manageable. Rehabilitation handbooks, papers, instrument descriptions, and the opinion of rehabilitation specialists were used to select the mobility-related activities that best satisfied the above criteria. The set finally defined consisted of the following activities: lying, sitting, standing, walking, climbing stairs, cycling, driving a wheelchair, and the transitions between different postures. Lying, sitting, and standing were combined into the category *static activities*, the others into the category *dynamic activities*. The transitions from one posture to another were regarded as a separate relevant category.

Static activities, dynamic activities, and transitions are aimed to be measured with the AM, which consists of accelerometers, a portable data recorder, and a computer for measurement control and data analysis. The major purpose of this thesis is the development and experimental evaluation of a method for the assessment of mobility-related activities during normal daily life. This purpose will mainly be related to patients with a leg amputation, as well as to rehabilitation medicine, although the AM has also been used in studies with other populations and within other fields. Some of these studies are also discussed in this thesis.

Ambulatory monitoring in literature

Ambulatory monitoring has a rich tradition in cardiovascular research. The larger part of the papers in the Journal of Ambulatory Monitoring (recently integrated into the Journal of Medical Engineering & Technology) concerned the ambulatory measurement of ECG and blood pressure.

The concept of ambulatory measurement of mobility, gait, and physical activity is not new. The figure at the cover of this thesis is found in a book published in 1885 and shows us a person carrying a recorder that monitors locomotion measured by means of air-filled units under the feet.[60] Although that instrument may not have

been developed for long-term purposes, it fundamentally reflects an ambulatory application. However, from the 1970s, real ambulatory systems were more frequently described. For example, Halstead[36] described the monitoring of functional variables, such as wheelchair mobility and time-out-of-bed. Snijders et al.[86] reported ambulatory measurement of the curvature of the spine in an ergonomic study. Mechanical movement counters were designed to measure physical activity and movement,[17,83] followed by the use of instruments based on accelerometers (see 'Ambulatory measurement of physical strain'). These instruments generally were attached at the wrist, ankle, or waist, and their output was usually related to activity level or energy expenditure, but not to the type of activity performed.

Therefore, instruments (activity monitors) have been developed which provide additional information on the activities performed. Stock and colleagues[88,89] used a 'microcomputer-based system for the assessment of post-operative fatigue', consisting of a posture timing module, an activity module, and a heart rate module. Anastasiades and Johnston[1] used EMG to discriminate between static and dynamic activities. Tuomisto et al.[91] applied accelerometers, a hydrostatic tube and EMG to distinguish activities. Instruments for measuring walking periods by footswitches, accelerometers, and mechanical sensors, are also developed.[7,41,42,76,83,87] These instruments only measure walking and (probably) climbing stairs, but not cycling and different body positions. Miyazaki[65] described an ambulatory instrument using gyroscopes for the measurement of stride length and walking velocity. The instrument used by Diggory and colleagues[19] and Follick and colleagues[32] was based on a tilt switch to detect the upright position.

Some other systems described are focussed on a more extended set of activities. Walker et al.[95] described an activity monitor based on mercury switches and accelerometers, for the measurement of posture and number of steps. Kiani et al.[50,51] and Groeneveld et al.[35] have reported an AMMA system (Ambulatory Monitoring of Motor Activities) using accelerometers, an artificial neural network and fuzzy logic. An accelerometer-based activity monitor to measure postures and movement is also reported by Busser et al.[13] Fahrenberg et al.[28,29] studied the possibilities of accelerometry to detect postures and movement from a psychophysiological viewpoint.

It can be concluded that many ambulatory systems have been designed and used. Generally, however, most of the mentioned instruments distinguish a relatively small set of activities, and validation studies regularly have serious limitations. Due to developments in data recorder and sensor technology, advanced ambulatory systems that measure (during) daily activities have become within reach.

Accelerometry in movement analysis

Accelerometer signals are frequently used for the analysis of human movement, and most frequently for the analysis of gait.[e.g. 15,18,26,38,67,71,75,84,85] In most of these studies the accelerometers are attached to the lower back. Accelerometers are also applied in research on shock absorption,[57] in research related to functional electrostimulation,[98,99] and in research about quantification of bradykinesia.[22] Another application is the use of accelerometers in studies aimed at movement co-ordination and phase relations between segments.[23,94] It can, therefore, be concluded that accelerometer signals have a potential to provide data on quality of activities.

Ambulatory measurement of physical strain

Several techniques exist to measure physical strain or energy cost during daily life activities, each with their pros and cons. Measurement of oxygen uptake is a reliable, valid and frequently used method, but has practical disadvantages and can not or not easily be performed ambulatory.[14,33,68,73,77] The doubly labelled water method is also a valid technique to determine energy cost, but is very costly, can only be used long-term (mostly 2 weeks), and can not be directly related to activities performed at certain moments in time.[5,9] Subjective data from diaries and questionnaires can be associated with data from more objective instruments,[17,45] but serious limitations in reliability and validity are reported.[2,23,31,62,73]
Heart rate or ECG – which can also be measured with the AM – has been used to provide data on cardiac and physical strain, overall workload, or energy expenditure.[1,3,43,59,69,78,79] However, some problems exist concerning reliability and validity:[3,59,73,77] heart rate is, for example, sensitive to mental processes, stress, fear, illness, medication, temperature, body position, and type of movement. Nevertheless, heart rate has shown to be of value in the ambulatory measurement of physical strain during activities of daily living.[16,46,82]
Accelerometry-based movement sensors have been used to measure the amount and intensity of body segment movements, which is called *motility*. Some of the instruments described in literature are the Caltrac,[2,6,53,64,72,97] Tracmor,[10,11] LSI,[44,66,96] Tritrac,[23,25,54,70,97] AMS,[52] Actigraph,[90] CSA,[47,64] and motion sensitive instruments developed by Meijer et al.,[62,63] and van Hilten and colleagues.[39] Although motility can not directly be regarded as a measure of physical strain, and discrepancies with physical strain measures may therefore exist, the relationship between motility-related measures on the one hand, and heart rate, oxygen uptake, or energy expenditure on the other, has frequently been studied and found, although the relationship is usually not unambiguous.[11,25,47,52,63,64,66,73,91,97] The sensors in these

studies are generally attached to the wrist, waist, or ankle. In the analysis program of the AM, motility signals are routinely derived from the accelerometer signals for the detection of activities. In view of the literature, these signals may also be of value in the assessment of physical strain during normal daily life, as may the simultaneous measurement of ECG or heart rate.

Outline of the thesis

This thesis is structured to correspond with the three main aspects of mobility-related activities: quantity (chapters 3-8), quality (chapter 9), and physical strain (chapter 10); in chapter 11 all three aspects are studied.

Chapter 2 gives an overview of current techniques used in rehabilitation to measure mobility-related activities. Instruments are classified and assessed according to relevance, aspect of mobility they measure, methodological criteria, and practical criteria. In *chapter 3* the AM is described in more detail, mainly addressing quantity. The focus is on the requirements for and description of sensors, a performed feasibility and master study, and a technical description of the AM.

Chapters 4, 5, and 6 present three validation studies. These validity studies are similar in design, but differ in setting and population: healthy subjects and subjects with a trans-tibial amputation (*chapter 4*), healthy subjects within a psychopharmacological study (*chapter 5*), and failed back surgery patients (*chapter 6*). The data of these studies were initially processed using a first version of the AM analysis program, which was restricted to several static activities (several types of lying, sitting, and standing), all transitions, and dynamic activities as one group. Recently, algorithms to distinguish dynamic activities have been implemented, and the structure of the analysis program has been changed. This extended AM version – which is described in chapter 3 – is validated with the existing signals of the three validity studies. The results are presented in *chapter 7*.

In the validity studies the measurements were relatively short-term (0.5 - 4 hours), whereas the AM is developed for long-term (from one to several days) measurements. *Chapter 8* provides an example of such measurements in patients suffering from congestive cardiac failure and in healthy subjects, to obtain insight in the activities performed by both groups, and the between-day variance.

The accelerometer signal is rather complex. The signal is constructed of the gravitational acceleration, as well as of inertial accelerations. To use the signal as a source of quality variables, knowledge of and insight in the signal is necessary. In *chapter 9* an experimental study is described, which was aimed at the decomposition of the signal from the tangential accelerometer attached at the thigh during walking,

at its relation with temporal events, and at the influence of subject variability, walking speed, walking surface, and sensor attachment on the signals.
The relation between motility-related measures and physical strain measures is frequently investigated. The accelerosignals measured with the AM may, therefore, have the potential to measure physical strain. The study described in *chapter 10* was aimed at the feasibility of AM motility signals in the evaluation of physical strain in walking at different walking speeds and in walking with a brace. Motility data are compared to heart rate and oxygen uptake data.
In *chapter 11* evaluation of quantity, quality, and physical strain are combined in one study. Due to previous and foreseen studies with an 'Activity Monitor', questions arose about the feasibility of measurements with the AM in the early phase of rehabilitation of persons with an amputation. This study was therefore aimed at the detection of walking and climbing stairs, the reliability of gait quality and physical strain variables, the sensitivity of these variables to differences and changes, and the potential of motility to predict physical strain.
Chapter 12 presents a general discussion on the content of this thesis.

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Techniques for measurement and assessment of mobility in rehabilitation: a theoretical approach

Summary

Mobility is an important construct in rehabilitation; many instruments have emerged which measure or assess (aspects of) mobility. In the selection or development of an appropriate technique, knowledge about the fundamentals of rehabilitation medicine is needed, as well as about essential characteristics of techniques and fundamental differences between them. The aim of this paper is to classify, assess, and discuss current techniques that are or can be used to measure aspects of mobility.

Eight techniques (physical science techniques, clinimetry, observation, diaries, questionnaires, actigraphy, physiological techniques, and activity monitors) are classified, assessed and discussed, based on the level of outcome measures, the aspect of mobility they measure, and methodological and practical criteria.

It is stated that rehabilitation medicine has a particular need for instruments that enable measurement of outcome measures on the level of activity and role fulfilment. Techniques differ in type and number of mobility aspects they measure. Furthermore, important differences exist based on methodological and practical criteria, although one optimal technique does not exist. The choice of a technique always has to depend on a complexity of factors, such as clinical problem, research question, mobility aspect of interest, required methodological strength, costs, and availability.

(8) *Ambulatory activity monitors*: portable systems to measure specific activities related to mobility, such as postures and/or movements (e.g. walking). This technique is, like actigraphy, characterised by an unrestrained range of motion of the measured subject, long-term and (semi-)continuous measurement, and the possibility of measurement in a person's own environment.[1,22,27] The instruments range from simple and/or providing one or few measures[1,19,27,58,59] to complex and/or providing several measures.[14,23,36]

In deciding on this classification one criterion has been most important: there should not be a complete or almost complete overlap in studied characteristics between two or more categories. Each category had to be significantly different from the others. Therefore, spoken (interviews) and written questionnaires were not distinguished, and some subcategories of physiological techniques were not discussed separately. The used classification is most appropriate with regard to the purpose of this paper.

Level of outcome measures

In rehabilitation medicine the hierarchical levels of impairment, disability and handicap – according to the ICIDH – although often discussed, are widely used.[3,10,20,28,33,51,66,67] In this paper the positively defined[57] counterparts will be used: function, activity and role fulfilment, respectively. One focus of discussion concerns the definition of the levels used in the ICIDH model. The distinction between the levels is not always clear[3,10,67] and additional (sub)levels are sometimes introduced.[3,10,64] In this paper the division of activity in 'simple' and 'complex' activity will be used; although using different terminology, this is also done or propagated by others.[3,10,50,64]
Techniques and outcome measures can – despite the semantic problems – to a certain extent be classified according to the (adapted) ICIDH model. Outcome measures on *function* level concern (local) structures or functions related to the locomotor system, e.g. angular movement of the knee, contraction of the quadriceps, curvature of the spine. Outcome measures on the level of *simple activity* concern the performance of a co-ordinated set of functions, which forms a component (and prerequisite) of functional, purposeful activities: examples include walking, climbing stairs, reaching and standing. Outcome measures on the level of *complex activity* concern the performance of functional, purposeful activities of daily life, such as dressing, feeding, housekeeping and working. *Role fulfilment* outcome measures are related to the disadvantage(s) resulting from an impairment or disability.

Table 2.1 shows the relationships between technique categories and level of the outcome measure. *Physical science techniques* usually have outcome measures on the function level (e.g. torque during isokinetic muscle testing) or on the simple activity level (e.g. ground reaction force while standing, joint moments while walking). *Clinimetric techniques* also measure on these levels; e.g. muscle testing according to the Medical Research Council, and manual measurement of joint mobility (both function level), the up&go test and clinical observation of gait (both simple activity level). *Observation*, *diaries* and *questionnaires* usually concern the performance of complex activities (e.g. questions about mobility activities a person can or does perform; observation of actual behaviour); observation and questionnaires can also be used to obtain information on role fulfilment (e.g. questions about problems related with limited mobility). The outcome measures from *actigraphy*, *physiological techniques*, and *activity monitor systems* are usually related with and/or validated during simple activities.

Table 2.1. Relationship between measurement technique and level of the outcome measure (+ can measure on this level; – can not measure on this level).

Technique	Outcome measure level			
	Function	Simple activity	Complex activity	Role fulfilment
Physical science	+	+	–	–
Clinimetry	+	+	–	–
Observation	–	–	+	+
Diary	–	–	+	–
Questionnaire	–	–	+	+
Actigraphy	–	+	–	–
Physiological markers	–	+	–	–
Activity monitor	–	+	–	–

On the function level there is a relatively large number of reliable and valid instruments; instruments testing functions and simple activities predominate.[34] Outcome measures on the level of activity and role fulfilment, however, are considered more relevant in rehabilitation, because rehabilitation goals on these levels are frequently reported as being the most important.[4,25,27,29,31,33,40,55,67] Adequate instruments on these levels, however, are relatively scarce and seldom

used.[18,24,25,28,29,34,56] Within the field of activity, complex activity outcome measures are considered to be more relevant than the simple activity outcome measures, due to their more functional character. Outcome measures on the function level may be (more) relevant if there is an unambiguous relationship between these and more functional outcome measures.[18,28,42,56] Although some association was found,[29,42,63] other times this relationship appeared to be or is assumed to be absent or complex[12,18,41,55]

Relevance determines, in part, the validity of an outcome measure, and therefore is in fact also a methodological criterion. The relevance (and with it the validity) of outcome measures is not only determined by their level, but also by the functional degree of the act, during which, or on the basis of which (i.e. retrospectively, e.g. by means of questionnaire), measurements are conducted.[29,35] For example, symmetry of walking (simple activity level) can be measured during walking on a treadmill and during shopping; the latter is assumed to be more valid. The functional degree of the act, in its turn, is related to the setting of the act; e.g. shopping can not take place in a lab. It can generally be stated that the more functional the act is, the more natural the setting has to be, and the greater will be the ecological validity[22] and thus the relevance.
The characteristics of observational techniques, questionnaires, diaries, actigraphy, physiological techniques, and activity monitor systems are in line with this reasoning. Physical science techniques and clinimetry are inadequate with regard to this; if one is interested in functionality, then these techniques are not useful.

Aspects of mobility

The concept of mobility is rather general: instruments that measure 'mobility' measure a specific aspect of mobility; techniques differ in the aspect they measure.[45] Two types of classification will be discussed: (1) the distinction in quantity, quality, and strain; (2) the distinction in performed, possible and preferred mobility from the viewpoint of a professional or a patient.

Quantity, quality and strain
Quantity of mobility concerns items such as: when, how often, and how long. The main item of *quality* is: how, i.e. the way of performance. *Strain* concerns the physical and psychological reaction of the body due to activity. The classification in quantity, quality or strain is especially useful in outcome measures on the activity level. Table 2.2 shows the eight techniques and the mobility aspect that each technique generally measures.[47]

Performed, possible and preferred

Another difference between techniques concerns the following three aspects of measured mobility:

(1) mobility a person actually performs or has performed, the 'do do' part of mobility (*Performed mobility*); e.g., a patient walks 750 meter without stops for shopping.

(2) mobility a person is actually able to do at a certain moment, the 'can do' part (*Possible mobility*). The construct capacity is almost equivalent to this; e.g., a patient can walk 1000 meter without stops at a certain moment.

(3) mobility which a person wants to perform or is supposed to perform compared to others in the same situation, the 'will do' part (*Preferred mobility*).

All aspects of measured mobility have two perspectives which are called *Professional* and *Patient*.[12,30,54] *Professional* entails that the mobility of a subject is assessed by a (more or less) objective expert or objective instrument, (more or less) independent of personal feelings or prejudices.[13] *Patient* entails that a subject's mobility is assessed by the subject himself.

Table 2.2. Relationship between measurement technique and aspect of mobility (quantity, quality, strain) that can be measured (+ possible; ± questionable or indirectly; – not possible).

Technique	Mobility aspect		
	Quantity	Quality	Strain
Physical science	–	+	+
Clinimetry	–	+	–
Observation	+	+	–
Diary	+	–	±
Questionnaire	+	+	+
Actigraphy	+	–	±
Physiological markers	–	–	+
Activity monitor	+	±	±

Table 2.3 lists the 6 fields of mobility together with their related techniques. When performed mobility is measured, indirectly insight is obtained in possible mobility: what a person does, he is able to. These cases are marked with '±'. *Questionnaires* have the advantage that a variety of aspects can be assessed with a single instrument;[47] performed, possible and preferred mobility can be measured,

depending on the type of questions formulated. A few examples: 'I do not walk up or down hills' (performed mobility according to the patient;[11] 'Patient is able to go up and down a flight of stairs safely without help or supervision' (possible mobility according to the professional);[44] 'I can't walk at all' (possible mobility according to the patient).[21] Note that Professional and Patient perspectives are not always neatly separated: the Professional perspective is sometimes strongly based on patient information (or on 'say do').[12,54]

Table 2.3. Relationship between measurement technique and aspect of mobility (performed, possible, preferred, according to the professional and the patient) that can be measured (+ possible; ± questionable or indirectly; – not possible).

Technique	Mobility aspect					
	Performed Profess.	Performed Patient	Possible Profess.	Possible Patient	Preferred Profess.	Preferred Patient
Physical science	–	–	+	–	–	–
Clinimetry	–	–	+	–	–	–
Observation	+	–	±	–	–	–
Diary	–	+	–	±	–	–
Questionnaire	±	+	±	+	–	+
Actigraphy	+	–	±	–	–	–
Physiological marker	+	–	±	–	–	–
Activity monitor	+	–	±	–	–	–

The classifications presented in Tables 2.2 and 2.3 are not a question of 'good' or 'bad': the aspect of mobility an instrument measures is mainly a matter of classification and not a matter of assessment. If the choice for a specific instrument has to be assessed, then the relation between the chosen instrument on the one hand, and the aspect one is interested in and the research question on the other, should be of major importance. If one has formulated a clinical or research question, it is important to clarify the aspect of mobility one is interested in, i.e., the aspect which corresponds with the posed questions.

Methodological quality and practical feasibility

Important methodological properties of and requirement for an instrument are responsiveness (or sensitivity), reliability and validity.[6,13,17,43,51,60,67] Here, we use the

following more concrete methodological criteria to assess the techniques: possible influence of subjective factors on the part of the subject and researcher, possible influence of measurement on acts performed ('reactivity'[22,47] or 'perturbation effect'[13]), retrospectivity and required motivation. The results of the classification of techniques based on these criteria are given in Table 2.4.

Table 2.4. Assessment of measurement techniques according to methodological criteria (+ good; ± questionable; – bad).

Technique	Methodological criteria				
	Influence Subject	Influence researcher	Influence measurement	Retro-spectivity	Required motivation
Physical science	+	+	–	+	+
Clinimetry	+	–	–	+	+
Observation	+	±	–	+	+
Diary	–	+	–	–	–
Questionnaire	–	±	+	–	+
Actigraphy	+	+	±	+	+
Physiological marker	+	+	+	+	+
Activity monitor	+	+	±	+	+

The major drawback of *physical science techniques* is that they may cause a reactivity effect: the activities performed by a patient are influenced by the measurement instruments, setting, etc. The same holds for *clinimetry*, in which the presence of a clinician or researcher may play an additional role. Although *observational techniques* are considered reliable and a good reference method,[1] an observation may influence a person's performance,[7,15,45,47,49,53] and subjective influences of the observer (e.g. observer fatigue[45]) cannot be ruled out.[7] Moreover, intra-observer and inter-observer differences may exist.[7,15] A person may be asked to keep a *diary*; however, a diary interrupts and influences their activities,[45,47,65,69] demands a high level of compliance,[16,45,47,48,49,69] and doubts remain about its reliability due to the subjective[53,65] and generally retrospective[7,48] nature of a diary. The reliability and validity of *questionnaires* is threatened by their retrospective and subjective character.[7,39,45,47,52,60,69] In verbal questionnaires or interviews the subjective character is even stronger due to the possible influence of the interviewer.

On the other hand, questionnaires have the characteristic of non-reactiveness.[39] One of the main characteristics of *ambulatory systems* is objectivity,[47,53] but some effect of the instrument on the activities a patient performs may exist.[22] *Physiological techniques* are methodologically strong, offering accuracy and objectivity.[47]

The practical feasibility has been operationalised as: convenient for researcher or clinician (e.g. simplicity of use, possibility to determine measurement characteristics), convenient for the patient (e.g. comfortable, painless), costs of instrument and costs of measurement.[32] These and other factors will play an important role before an instrument may be introduced in practice, i.e. in a clinical setting by physiatrists, physiotherapist, etc.[9,18,25,28,37,55]

Table 2.5. Assessment of measurement techniques according to practical criteria (+ good; ± questionable; – bad).

Technique	Practical criteria			
	Convenience Researcher	Convenience patient	Costs instrument	Costs measurement
Physical science	–	±	–	±
Clinimetry	+	+	+	+
Observation	–	±	+	–
Diary	+	–	+	+
Questionnaire	+	+	+	+
Actigraphy	+	+	±	+
Physiological markers	+	+	–	–
Activity monitor	+	±	–	+

Table 2.5 shows the results of the assessment based on practical criteria. From a practical feasibility viewpoint *clinimetric* tests and instruments have major advantages. *Physical science techniques* are generally difficult to use and expensive. For assessment of *observational techniques* practical shortcomings are important; they are time-consuming for the researcher, therefore costly and being used for a limited duration only.[45,49,53] A *diary* interrupts and influences the activities of a person,[65,69] and demands a high level of compliance.[47,48,49,69] *Questionnaires* are unobtrusive and generally easy to use and inexpensive.[39,47] The costs of *actigraphy*

are moderate;[45] more advanced *activity monitors* may be more costly.[47] The high cost of *physiological techniques* is an important disadvantage.[47]

When the techniques are assessed using methodological and practical criteria, the picture is rather complex. Especially here, there is great variability between instruments, and any generalisation should be made with caution.
The concepts of activity and, especially, social roles partly concern subjective matters of experience;[67] but even the more objective aspects of activity and role fulfilment are often assessed with techniques prone to undesired subjective influences. However, it will be important to use – if possible – instruments which measure functional performance more objectively.
In general, the more functional the outcome measure of interest is (activity/role fulfilment), the lower the level of objectivity and reliability.[62,67] Instruments measuring on the level of activity and role fulfilment often have an undetermined reliability and validity; if validity is investigated, the results are often difficult to generalise.[18,24,25,28,56] Moreover, these instruments often lack responsiveness and are thus inadequate to measure small but essential effects. It will be clear that performed reliability and validity studies are important arguments in the selection of an instrument.
After having decided which element of mobility should be measured, methodological and practical requirements should determine the choice for one technique. The choice of a specific technique or instrument is generally a matter of weighing up both qualities: the choice of a technique depends on the relative importance of the various criteria. Finally, for an instrument to be used in clinical practice, the data it provides must be understandable;[33] data which are not clear to the clinician will not be accepted.

Discussion and conclusions

The aim of this paper was to present a theoretical framework in which current techniques used in rehabilitation to measure mobility could be placed. Classifying and assessing techniques is useful, though we are aware of three facts: (1) one *perfect* technique does not exist: the flaws of one technique are often the strengths of another;[15] (2) within one type of technique there will be a great variability in quality and characteristics of specific instruments; and (3) in the classification and assessment of techniques some grey areas exist.
An 'ideal' instrument should: measure on the level of activity or role fulfilment, measure one or, if possible, more well-selected aspects of mobility, be methodologically strong, and practically feasible. For each technique the most

important disadvantages can be given. *Physical science and clinimetric techniques* have the disadvantage of measuring functions and simple activities, during lower functional acts in an artificial setting; their relevance with regard to the central issues in rehabilitation medicine is limited. *Observation* is time-consuming and therefore costly and, besides, may influence a patient's behaviour. *Diaries* require a high level of motivation and are subjective; both of which threaten reliability and validity. *Questionnaires* will remain an important technique in obtaining information on activity and role fulfilment, though ongoing research on reliability, validity and responsiveness will be necessary. Due to the characteristics of *ambulatory techniques*, these techniques may offer new possibilities for research in rehabilitation, possibly in combination with other techniques. Especially their costs and the type of outcome measures they can provide will determine their additional value. *Physiological techniques* are methodologically strong, but the costs per measurement, and their limitation to strain are serious disadvantages.

It can further be concluded that the choice of a technique will inevitably depend on a complexity of factors, such as clinical problem, research question(s), mobility aspect of interest, required methodological strength, costs and availability. We are convinced that the criteria presented in this paper can support the process of selecting techniques and instruments. It is important to realise that relevant measures that are not reliable and valid are useless, as are reliable and valid measures that are not relevant.

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Detection of mobility-related activities using accelerometry: development and characteristics

Summary

Within rehabilitation medicine and other related areas, objective measurement of activities of daily living may provide relevant information. Due to developments in data recorder and sensor technology, advanced ambulatory systems that measure (during) daily activities have become within reach. One such instrument is the Activity Monitor (AM): an instrument based upon long-term ambulatory accelerometry, and aiming at the assessment of the activities performed by a person during the measurement period. This chapter will focus on the requirements for sensors, the characteristics of piezo-resistive accelerometers, a completed feasibility study, a study to define the various settings of the instrument ('master study'), and a technical description of the AM.

Piezo-resistive accelerometers are assumed to be adequate for ambulatory activity detection. A feasibility study (n=10) showed the applicability of these sensors and a theoretical detector scheme. Following this feasibility study, a first version of the AM analysis software was developed, followed by an extended version. The procedures and settings within the analysis software of both versions were guided by a 'master study' in which healthy subjects (n=19) performed an extensive set of activities according to a strict protocol. The selected procedures and settings were tested on the data of this study, although optimisation of the so-called 'Activity Detection Knowledge Base', which contains the settings of the instrument, will remain an ongoing process. A more detailed technical description of the extended version of the AM is provided in the last section of this chapter.

Introduction

Rehabilitation is primarily directed at the functional status of patients. Although not the only important outcomes of a rehabilitation program, locomotion, ambulation or mobility are considered important.[1,2,4,8,13,16,17] Many techniques are used for acquisition of data on (aspects of) mobility – including questionnaires, observation, diaries, kinetic and kinematic systems, actometers, and types of activity monitors (see chapter 2).[3] If unobtrusive, objective, and valid measurements are required to capture a large and specific set of mobility-related activities during normal daily life in a person's personal environment, currently available techniques are inadequate.
Especially within the area of ambulatory systems, recent technological developments have led to advanced measurement systems. Small, portable, digital data logger systems have become available in the recent years, with increased data processing and data storage capacities. Due to simultaneous developments in sensor technology, measurement during normal daily life has become within reach.
Technology may provide means, but provides no answer to what should be measured and how it should be measured. The focus of rehabilitation medicine, the relevance of mobility-related activities, and the technological developments have been the motive for the development of the Activity Monitor (AM): an ambulatory instrument to measure mobility-related activities during normal daily life. The following activities were regarded to be of major interest: lying, sitting, and standing ('static activities'), walking, climbing stairs, cycling, and driving a wheelchair ('dynamic activities'), and the transitions between different postures.
This chapter will focus on the requirements for sensors, the characteristics of piezo-resistive accelerometers, a completed feasibility study aimed at the applicability of these sensors and a theoretical detector scheme, a study to define the various settings of the instrument ('master study'), and a technical description of the AM.

Biomechanical characteristics of activities

Biomechanically, the human body can be considered to consist of a number of rigid bodies or body segments, linked together by joints.[6,15,19] In the case of a static activity, the positions and orientations of the segments do not vary significantly with time. The static activities can therefore be identified by the orientations of the segments with respect to the gravitational field. In contrast, in the case of a dynamic activity, the positions and orientations of the segments do vary with time. Body movements occurring over distances that are large in comparison with the length of the body segments are most naturally achieved by moving the segments in a cyclical

fashion.[9] Non-cyclical movements are present during e.g. transitions between postures.

Requirements for sensors

Ambulatory measurements place great demands on the sensors used. They have to be sufficiently small (a few square centimetres) and light (a few grams) such that they can be taped on the skin without introducing relative resonance movements. Furthermore, they should be robust, have a low energy demand, and they should be easily mountable on the body, and stay in place for the duration of the measurement. Their alignment should not be critical, and they should be comfortable for the subject and not impede the activities of daily living. Therefore, they should not cross joints and only require short cables for connection to the measurement unit. Their signals should contain maximal information about relevant kinematic quantities, and the number of sensors required for sufficient evaluation of daily life activities should be small. Among the alternatives, piezo-resistive accelerometers satisfy most of these conditions.

Piezo-resistive accelerometers: general principles

Uni-axial piezo-resistive accelerometers consist of a mass, connected to a frame by beams, which can be represented by a damped spring (Figure 3.1).[19] In the beams piezo-resistors are mounted; they form a bridge circuit, and the value of the resistors depends on the magnitude of acceleration. Due to their structure, the sensors are sensitive for accelerations in only one direction. Uniaxial piezo-resistive accelerometers measure combined a component of the gravitational acceleration ($\bar{a}_{grav}$, 9.81 m.s^{-2}) as well as a component of the inertial acceleration ($\bar{a}_{inert}$). In static situations, the accelerometer signal yields only gravitational information, while in dynamic situations this information is combined with inertial information.[7,11,12,18] The part of $\bar{a}_{grav}$ that is measured ($\bar{a}_{grav,sens}$) depends on the angle φ_2 between the sensitive axis of the sensor and $\bar{a}_{grav}$:

$$a_{grav,sens}(t) = a_{grav}(t) \cdot \cos(\varphi_2(t))$$

The part of the $\bar{a}_{inert}$ that is measured ($\bar{a}_{inert,sens}$) depends on the angle φ_1 between the sensitive axis of the sensor and $\bar{a}_{inert}$:

$$a_{inert,sens}(t) = a_{inert}(t) \cdot \cos(\varphi_1(t))$$

The gravitational force acts on the mass, while the inertial forces act on the frame of the sensor. Therefore, $\bar{a}_{grav}$ and $\bar{a}_{inert}$ have opposite effects on the spring and therefore on the signal provided by the sensor (a_{sens}):

$$a_{sens}(t) = a_{inert,sens}(t) - a_{grav,sens}(t)$$

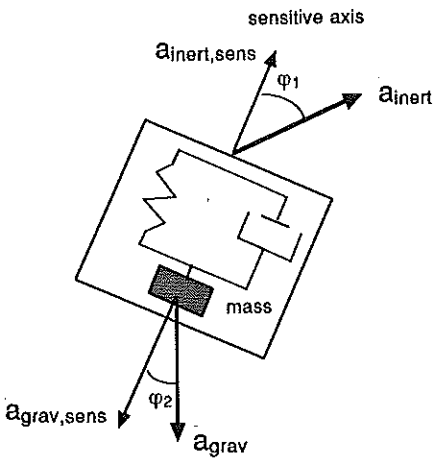

Figure 3.1. Uniaxial piezo-resistive accelerometers measure the component $\bar{a}_{inert,sens}$ *and* $\bar{a}_{grav,sens}$ *of the inertial (*$\bar{a}_{inert}$*) and gravitational (*$\bar{a}_{grav}$*) acceleration, respectively.* φ_1 *is the angle between the sensitive axis of the accelerometer and* $\bar{a}_{inert}$*;* φ_2 *is the angle between the sensitive axis and* $\bar{a}_{grav}$*.*

A constant a_{sens} which is smaller than 9.81 $m.s^{-2}$ can, theoretically, be the result of a constant $\bar{a}_{inert}$, or the result of $\bar{a}_{grav}$; with a single accelerometer this can not be determined. However, during the performance of normal dynamic activities the occurrence of a constant $\bar{a}_{inert}$ having a duration in the order of a few seconds is unlikely. Therefore, activities will be assumed to be dynamic if the uniaxial accelerometers mounted on the body yield a time-varying signal and static if the signal is constant. For static activities the constant a_{sens} can be written as:

$$a_{sens} = -a_{grav} \cdot \cos(\varphi_2)$$

The angle between the accelerometer axis and the gravity vector can be determined from the constant accelerometer signal, which gives information about the orientation of the accelerometer. Note that detection of static activities on the basis of this assumption will only give reliable results if the environment does not accelerate.

Distinguishing activities by accelerometry: initial phase

Detection of the static or dynamic nature of activities

Assuming that dynamic activities yield time-varying accelerometer signals, while static activities yield constant accelerometer signals, the static or dynamic nature of

activities can be detected by determining whether the signal varies with time. For this purpose, the static-dynamic detector scheme depicted schematically in Figure 3.2 was proposed.

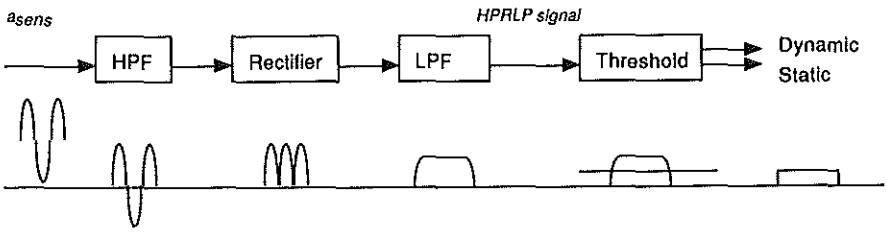

Figure 3.2. Detector of the static or dynamic nature of activities. The signal a_{sens} of an uni-axial accelerometer mounted on the body is high-pass filtered (HPF), rectified and low-pass filtered (LPF) to obtain an HPRLP signal that is a measure of the time variation in the signal a_{sens}. A static activity is detected if the HPRLP signal is lower than a set threshold, otherwise an activity is detected as dynamic.

The acceleration signal is high-pass filtered (HP, 0.5 Hz), rectified (R), and subsequently low-pass filtered (LP, 0.1 Hz), resulting in an HPRLP signal. The high-pass filter will eliminate the current offset of the signal. Rectification and low-pass filtering at a cut-off frequency of approximately 0.1 Hz will yield a measure for the averaged signal deviation from the mean, and is therefore a measure of time variation or 'acceleration energy'. The static or dynamic nature of activities can be discriminated by applying a threshold to the resulting signal. If the signal is above the threshold the activity is detected as being dynamic, while those below are deemed static. The threshold should be set such that small movements within static activities are not detected as dynamic, while still detecting all dynamic activities.

Distinguishing static activities

If an activity has been detected as static, the type of static activity can be identified from the orientation of the body segments. These orientations can be obtained from the accelerometer readings, which give only inclination information if the position and orientation of the body segments do not vary with time.[5] Even in static activities, some small and irrelevant higher frequency movements may take place. To make the static activity identifier less vulnerable for this, the signal can be low-pass filtered (e.g. 0.1 Hz; LP signal). Most static activities can be discriminated by observing the orientations of the trunk and the thigh only.

Distinguishing dynamic activities

Dynamic activities are normally achieved by cyclical movements,[9] such as walking, climbing stairs, and cycling. The accelerometer signals per cycle may differ in several aspects: morphology, mean, standard deviation and cycle time. Therefore, these properties may be used to distinguish between dynamic activities.

Feasibility study

To investigate the feasibility of a small set of uniaxial accelerometers placed on several body segments to distinguish several static and dynamic activities, a feasibility study was performed. The study was focussed on (1) the number and location of sensors to be used, (2) the potential of the HPRLP signal to distinguish between static and dynamic activities, (3) the potential of the LP signal to distinguish between all activities, and (4) the potential of features related with morphology of the measured signal to distinguish between dynamic activities. This study is described in more detail in Veltink et al.[14]

Methods

In the experiments, ten male able-bodied subjects participated (age range 23-42 years; height range 1.72-1.87 m; mass range 60-88 kg). The experiments comprised two protocols: the first protocol consisted of a number of standardised static activities (several types of standing, sitting, lying prone, lying on the side right/left, lying supine), while the second protocol consisted of several dynamic activities (several types of walking, ascending stairs, descending stairs, and cycling; driving a wheelchair was not included). During the experiments, a maximum of 6 uniaxial piezo-resistive accelerometers were mounted on different locations (shank, thigh, sternum, shoulder) and in different directions (perpendicular to the long axis of the segment: tangential; or along this axis: radial).

After initial evaluation, only the accelerometers mounted tangentially and radially in the sagittal plane on the sternum and tangentially in the sagittal plane on one thigh were further evaluated. The tangential thigh accelerometer signal was chosen for the detection of the static or dynamic nature of activities (HPRLP signal, Figure 3.2), because leg movements are pronounced in all investigated dynamic activities. To enable a secure detection of cycles in the analysis of dynamic activities, in addition membrane switches were mounted under the heels and the balls of the feet to detect foot contact and clearance. A videotape recording was made simultaneously.

Results and discussion

The results indicated that it was possible to distinguish static from dynamic activities using the static-dynamic detector of Figure 3.2. The optimal setting of the detector parameters and subsequent performance evaluation requires explicit performance criteria. Development of such criteria, which may depend on the area of application, was not a topic in this study.

All selected static activities were well classified by combining the radial sternum and tangential thigh LP signals. Only lying on the left and right sides could not be distinguished with these two accelerometers. Although all static activities could be classified, it must be noted that only a few activities were examined in this standardised study. Further study was recommended to examine the validity of accelerometers in discriminating other static activities under more natural circumstances.

LP signals of the thigh and tangential sternum signal during cyclic movements appeared to be significantly different in several dynamic activities. The tangential trunk accelerometer is more sensitive for variation in average trunk orientation when the trunk is nearly vertical than the radial accelerometer,[5] due to the sinusoidal character of the signal when the accelerometer is rotated with a constant angular speed. This should be taken into account if static or dynamic activities are to be distinguished on the basis of relatively small angular differences of body segments. It seemed that the inter-subject differences in LP signal values during dynamic activities were such that a single detection threshold value for all subjects could not be used. Results on cycle duration and standard deviation of the signal of the tangential thigh accelerometer indicated that these measures contribute to the distinction of the speeds of walking and cycling. Distinction of dynamic activities by morphological comparison of signals was investigated for the tangential thigh accelerometer. The signals of ascending and descending stairs showed a higher variability than the signals of walking. The results further indicated a proper distinction between walking at any speed and ascending or descending stairs on the basis of accelerometer signal morphology.

Implementation: the Activity Monitor

The results of the feasibility study showed the potential feasibility of the accelerometers to discriminate between several activities, and yielded the starting point of the activity detection by means of the Activity Monitor. In the phase between the first feasibility study and further implementation, some choices were made concerning number of sensors and activity detection scheme.

Number of sensors

From the feasibility study (see also Veltink et al.)[14] it was concluded that two accelerometers (one tangential on the thigh, and one radial accelerometer on the sternum) were sufficient to discriminate between the static activities performed. Although such a configuration is able to distinguish a lot of static activities, some others will not be distinguished. For example, standing with the trunk flexed and sitting with the trunk flexed will be classified as lying on the side and lying supine, respectively. Furthermore, tangential accelerometers are more sensitive to changes in inclination in 'vertical' body positions than radial accelerometers, which may be especially important in discriminating several dynamic activities. Therefore, it was decided to use at least 3 accelerometers in the standard configuration: one tangential thigh accelerometer, one radial and one tangential trunk accelerometer. Another advantage is that, due to the fact that both trunk sensors can be fixed perpendicularly to each other, the validity of activity detection will increase. In later phases, a fourth sensor was used which was attached at the other thigh; in this way information could be obtained on walking pattern, and validity be increased as well.

First version Activity Monitor

The algorithms to calculate the HPRLP and LP signal could relatively easily be implemented in the time-series oriented language that was used: Signal Processing and Inferencing Language (S.P.I.L.).[10] In the AM analysis, the LP signal was converted to angles by means of an arcsine transformation (LP/angular signal). Furthermore, the HPRLP signal was not low-pass filtered in the final stage, but smoothed (HPRS signal). The procedures proposed to distinguish dynamic activities were more difficult to implement, and needed further study.

Therefore, a first version of the Activity Monitor analysis program was developed, based on the activity discrimination of dynamic versus static activities as general categories, and on discrimination of several static activities (Figure 3.3). In the analysis program the static-dynamic detection was based on a Boolean procedure applied on the HPRS signal of one or both legs. Each static activity was divided into two or more subcategories in the analysis procedure; in total 14 static subcategories were distinguished. For each subcategory, a minimum and maximum value was predetermined for each (three or four, depending on the number of sensors used) LP/angular signal. These values were pre-set in a so-called 'Static Activity Detection Knowledge Base' (SADKB). So each subcategory had a unique set of three or four ranges. Every second, the distance from each LP/angular sample to the corresponding minimum or maximum value was calculated and, finally, added for all three or four sensors, giving rise to a 'total distance'. The shorter the total distance, the higher the possibility that an activity was estimated correctly; if the

value of the sample is between the maximum and minimum value, the distance is zero. This version of the AM was used in the validity studies described in chapters 4, 5, and 6.

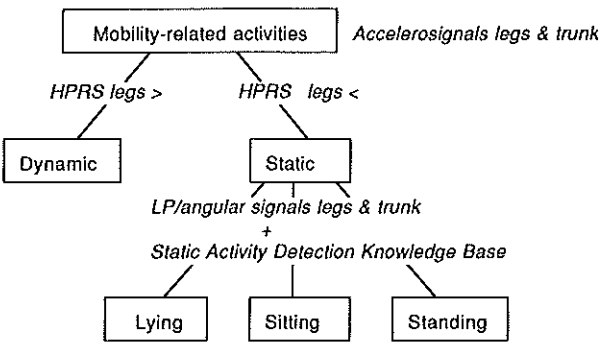

Figure 3.3. Activity detection scheme used in the first version of the Activity Monitor. Different static activities and all dynamic activities together could be distinguished. Detection of the groups 'static' and 'dynamic' occurred by means of a pre-set threshold on the HPRS signal of one or both legs. The detection of several static activities occurred by means of the LP/angular signals from all sensors and pre-set ranges in the Static Activity Detection Knowledge Base.

Ongoing developments

Besides the limitation of detecting all dynamic activities together, some other limitations were present in the first version of the AM. First, due to the hierarchical character of the detection scheme, errors in the first Boolean step ('static' versus 'dynamic') were transported to the lower levels; the quality of this first step strongly determined the validity of the instrument. Second, distinguishing static from dynamic activities by only the variability of the accelerometer signal from the thigh(s) is a rather imperfect method; it should be possible to use more subtle methods for this. Recently, the extended version of the AM analysis program has been implemented and tested. In this version the hierarchical structure is no longer present. All activities are characterised by (1) a LP/angular feature (corresponding to the LP/angular signals in the first version); (2) a motility feature (corresponding to the HPRS signal in the first version); and (3) a frequency feature, which was not used in the first version and which is based upon a type of instantaneous frequency analysis: Fast Time Frequency Transform (FTFT).[10] Each activity subcategory in the analysis program is characterised by specific ranges of each feature; these values are set in the so-called 'Activity Detection Knowledge Base' (ADKB). As in the

detection of static activities in the first version of the AM, the principle of the smallest total distance is used to detect an activity. Figure 3.4 shows the schematic representation of this extended version, which will be further described in the Technical description section of this chapter. In chapter 7 the validity results of this extended version will be presented.

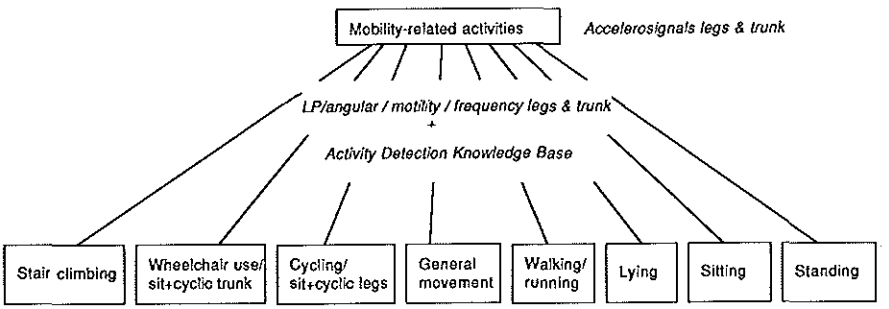

Figure 3.4. Activity detection scheme used in the extended version of the Activity Monitor. From all sensors three features are derived. Based on these three features and the corresponding activity-specific ranges for each feature in the Activity Detection Knowledge Base, an activity subcategory is detected.

Master study

Background

From the feasibility study it was concluded that the accelerosignals could be used to distinguish static activities, and that they had the potential to distinguish static from dynamic activities. In a later phase, the frequency feature appeared to have possibilities in detecting cyclic activities. However, a lot of 'fine-tuning' needed to be done. Therefore, a master study was designed, aimed at the optimisation of the feature signals and the settings in the (static) activity detection knowledge base.

Methods

Nineteen healthy subjects (8 female, 11 male; mean age 23 yrs, range 20-30; mean height 1.74 m, range 1.58-1.88; mean mass 65 kg, range 48-88) volunteered, and performed a standardised activity protocol. The protocol consisted of about 40 types of static and dynamic activities. Each activity was performed during about 15 seconds. Because cycling was not included, in another experiments four healthy subjects (2 female, 2 male; mean age: 26 years, range 25-27) performed a protocol

consisting of different types of cycling (varying cycling speed, underground, bicycle, saddle height, inclination).

During the measurement four accelerometers were used: one at each thigh (tangential), two at the trunk (radial and tangential). The signals were AD-converted and stored without pre-processing on a memory card with a frequency of 32 Hz (Vitaport1™). Besides the accelerometers, a special type of keyboard (Vitaport emopad) was connected to the recorder. Each activity had its own code, and before an activity was performed a key was pressed; the activity-specific code was assigned to a separate event channel on the same recorder. Signals were analysed on a Macintosh computer with Vitagraph™ and Signal Processing and Inferencing Language (S.P.I.L.)[10] routines and programs.

Results and discussion

There was a large range of movement frequencies within the several dynamic activities. This large range yielded problems in the band-pass filtering: especially during lower frequency movements, higher order/harmonic frequencies appeared in the band-pass signal, which disturbed its sinusoid shape (with the same frequency as the movement frequency), and therefore the FTFT procedure. This problem was (partly) solved by lowering the upper cut-off frequency and, at the same time, making the filter less steep at the upper part of the filter.

The signals of the dynamic activities performed (e.g. walking) appeared to differ significantly in shape when measured during different types of performance (e.g. walking fast, walking slow). Furthermore, a considerable inter-subject variability was found, and the signals during some activities (e.g. climbing stairs, walking fast, and running) contained a lot of higher frequency components. Therefore, it was concluded not to use morphology characteristics in detecting activities, but rather just the fundamental frequency.

The scaled settings within the ADKB (of the extended version of the AM), which were the result of examining the data of the master study, are shown in Table 3.1. The static activity subcategories are characterised by unique combinations of LP/angular data from the legs and trunk, with no or small motility, and without frequency. The LP/angular settings of the static activity subcategories are almost equal to the settings used in the SADKB of the first version of the AM. General movement is characterised by undefined LP/angular data, no frequency data (except lower frequencies of the legs), and motility data within a certain range. In walking, all features have a certain range; walking with increasing speed is characterised by increasing motility and frequency.

Table 3.1 Settings within the Activity Detection Knowledge Base, with the low-pass/angular, motility, and frequency ranges per activity subcategory and per sensor. The motility and frequency values are scaled; tan = sensor sensitive in tangential direction, rad = sensor sensitive in radial direction.

Activity subcategory	Feature settings								
	Low-pass/angular (degrees)			Motility (g, scaled)			Frequency (Hz, scaled)		
	thigh (tan)	trunk (tan)	trunk (rad)	thigh (tan)	trunk (tan)	trunk (rad)	thigh (tan)	trunk (tan)	trunk (rad)
Lying supine									
Standard	30 / 90	60 / 90	-30 / 30	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Lying on the side									
Strongly backwards	30 / 45	30 / 45	-15 / 15	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Backwards	0 / 30	0 / 30	-15 / 45	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Forwards	-30 / 0	-30 / 0	-15 / 15	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Strongly forwards	-45 / -30	-45 / -30	-15 / 15	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Lying prone									
Standard	-90 / -30	-90 / -60	-30 / 30	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Trunk slightly raised	-90 / -30	-60 / -45	30 / 45	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Standing									
Standard	-15 / 15	-30 / 30	60 / 90	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Trunk flexed	-15 / 15	-60 / 30	30 / 60	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Trunk strongly flexed	-5 / 20	-90 / -60	-30 / 30	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0

Sitting									
Backwards	45 / 90	30 / 45	45 / 60	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Standard	45 / 90	-30 / 30	60 / 90	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Trunk flexed	45 / 90	-60 / -30	30 / 60	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Trunk strongly flexed	45 / 90	-90 / -60	-30 / 30	0 / 25	0 / 25	0 / 25	0 / 0	0 / 0	0 / 0
Dynamic									
General movement	-90 / 90	-90 / 90	-90 / 90	50 / 500	50 / 250	50 / 250	0 / 30	0 / 0	0 / 0
Walking slow	5 / 9	-7 / 0	66 / 86	70 / 130	20 / 60	30 / 110	30 / 90	20 / 100	30 / 70
Walking	5 / 10	-10 / 2	61 / 90	120 / 250	30 / 80	40 / 140	70 / 100	55 / 100	70 / 100
Walking fast	5 / 10	-5 / 0	71 / 90	230 / 350	60 / 100	140 / 300	100 / 120	70 / 250	80 / 120
Climbing upstairs	15 / 30	-22 / -12	60 / 81	100 / 220	30 / 70	65 / 155	50 / 100	0 / 100	50 / 100
Climbing downstairs	15 / 20	-7 / 0	76 / 87	85 / 350	20 / 50	100 / 175	50 / 90	0 / 100	50 / 110
Cycling / sit+cyclic legs	40 / 70	-50 / -5	50 / 90	40 / 600	20 / 150	20 / 225	30 / 140	0 / 250	0 / 150
Driving wheelchair / sit+cyclic trunk	60 / 90	-30 / 30	60 / 90	30 / 125	25 / 50	30 / 70	0 / 0	30 / 110	0 / 150
Running	0 / 10	-25 / -5	66 / 86	240 / 700	60 / 200	300 / 500	100 / 120	0 / 250	130 / 160

Walking stairs is different from walking with respect to the mean value of the signals of the trunk and legs. Although in all subjects the LP/angular feature during walking stairs differed from the LP/angular feature during walking, and this difference was rather stable within and between subjects, there was some inter-subject variability in the LP/angular signal during these activities. This is most probably due to initial attachment differences in angular position of the sensors. Because the LP/angular differences between walking, walking upstairs, and walking downstairs are rather small, the inter-subject differences in attachment disturb the discrimination between these activities. The solution used for this problem is to correct the LP/angular features for initial angular deviations due to attachment.
Driving a wheelchair is characterised by the LP/angular data of all sensors, small motility values, and a detected frequency from the tangential trunk signal. During cycling a relatively large range of motility data is possible; motility therefore is not very specific for cycling. A frequency is generally detected from the leg signals; regularly a frequency was also detected from especially the tangential trunk sensor, but this detection was not always present. Running is characterised by a highly variable signal; therefore the motility settings are high; the frequency in the tangential trunk signal is not always detected very well due to signal distortion from high accelerations; therefore the minimum setting is 0.

After optimising the feature signals and settings, the analysis software was applied on the master data. Although testing the AM in this way can not be regarded as validating, the results will provide an indication of the functioning of the AM.
Static activities were detected well: of the 348 performed static activities, 344 were correctly detected (99%). Errors were the detection once of lying on the side while reading as standing, and the threefold detection of lying prone with leg(s) flexed as standing. Sitting and picking something up from the ground three times, was detected as general movement. The squat position was detected as sitting; this is not remarkable, because the position of trunk and legs in the squat position resembles the position in sitting. Walking is correctly detected in 185 of the 190 walking periods (97%). Only walking with crutches (with minor loading of one leg) was 5 times detected as climbing stairs, due to the more flexed position of trunk and/or legs. Climbing upstairs and downstairs only differ in trunk position. This difference appeared to be too small to obtain reliable data. Therefore, these two subcategories were further combined in the analysis (activity category climbing stairs). Normal stair climbing (foot for foot) was generally well detected: in 73 of the 76 cases correct (96%). Other types of stair climbing (foot besides foot; climbing downstairs with face to the stairs; with crutches) were less well determined: correct in 80 of the 114 cases (70%). Error detections were most frequently walking. Driving a

wheelchair (19 periods) was in 7 periods detected as sitting, although in all cases alternating with the detection of driving a wheelchair. Cycling was correctly determined in 64 of the 80 cases (80%). In the other cases, the detection as cycling alternated with error detections as sitting, general movement, and driving a wheelchair. These error detections were generally due to the fact that the cyclic nature was not very well presented in the signal, which caused imperfect band pass signals, and thus no detection of frequencies in the thigh signals. This phenomenon must receive further attention.
The limits of driving a wheelchair and cycling in the ADKB could be adjusted so that they should be detected more often correctly. However, especially driving a wheelchair already has a relatively low 'threshold', and small cyclic movements of the trunk in a sitting position will probably lead to the selection of driving a wheelchair. Therefore, we propose to rename the category 'driving a wheelchair' to 'sitting with cyclic movements of the trunk' (with low to moderate motility), and 'cycling' to 'sitting with cyclic movements of the legs' (with moderate to high motility).

Extended version Activity Monitor: technical report

Measurement set-up
The standard configuration of the Activity Monitor consists of four IC-3031 uniaxial piezo-resistive accelerometers (about 1.5×1.5×1 cm). The sensors are fixed on Rolian Kushionflex™ by double-sided tape; Rolian Kushionflex can be fixed directly on the skin. Two sensors are attached midfront the thighs, halfway the spina iliaca anterior superior and the upper side of the patella and two on the lower part of the sternum, perpendicular to each other (Figure 3.5). All accelerometers have to be attached as parallel as possible to the vertical or horizontal plane; a deviation of 15 degrees is allowed. This requirement is usually no problem at the legs; at the trunk a kind of wedge sometimes has to be used. Each accelerometer is attached to a data recorder, by means of separate Lemo-jackets or with one connector (Vitaport2™ or RAM, respectively; see next paragraph). Before measurements are started, the accelerometers are calibrated (+1g, −1g).

Recorder
The type of data recorder is in fact not crucial, although some requirements have to be met. The data logger should allow measurements for at least one day (data storage, energy supply), be able to measure (at least) three accelerosignals and ECG or heart rate, have low dimensions and weight, and be easy to handle by researchers and clinicians.

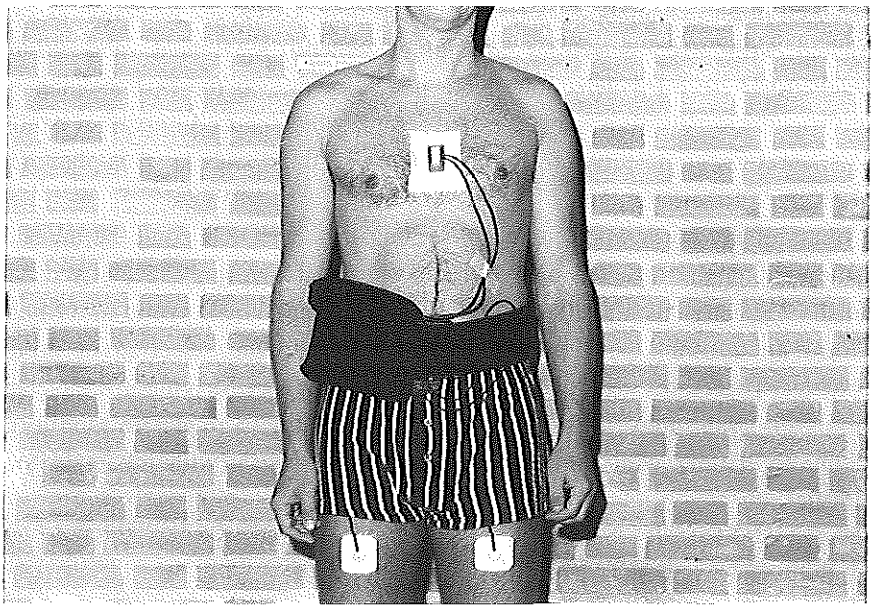

Figure 3.5 Person with the sensors attached according to the standard configuration of the AM, and wearing the recorder.

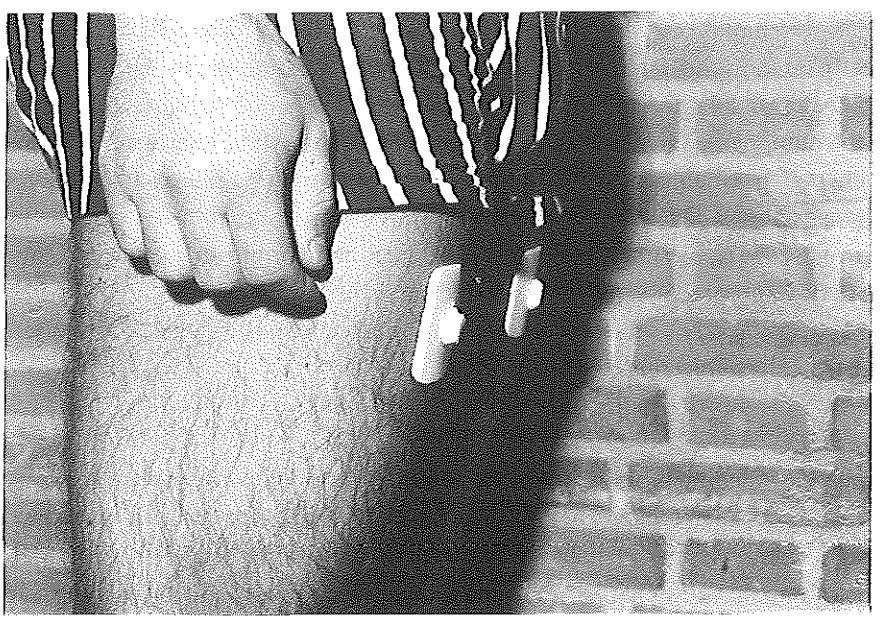

Figure 3.6. The thigh sensors fixed on Rolian Kushionflex™, which is fixed on the skin.

The Vitaport1™ data recorder was the starting point of the AM. This digital recorder (6×11×3 cm, 500 gr) was energy supplied by a dedicated battery, and allowed the simultaneous measurement of up to 8 channels; data were stored on a flash card of 1 MB. Therefore, long-term measurements were not possible with this recorder.

The Vitaport2™ followed the Vitaport1™. From the activity monitoring point of view, the most important differences were the use of 4 penlites batteries, the use of a PCMCIA hard disk of flash card (with a memory capacity of up to 360 MB), and the larger size and the greater weight (9×15×4.5 cm, 700 gr.). Continuous measurement (without changing batteries or disks) up to 2 days was now possible.

However, many of the features of this recorder were redundant for activity monitoring, and the size, weight, and costs were disadvantages. Therefore, recently 5 prototypes of the so-called 'Rotterdam Activity Monitor' (or 'RAM') are developed: a recorder based on Vitaport2™ technology, but more dedicated to activity monitoring, allows up to 5 accelerometers, ECG, a marker signal, and is smaller and lighter (9×15×3.5 cm, 500 gr).

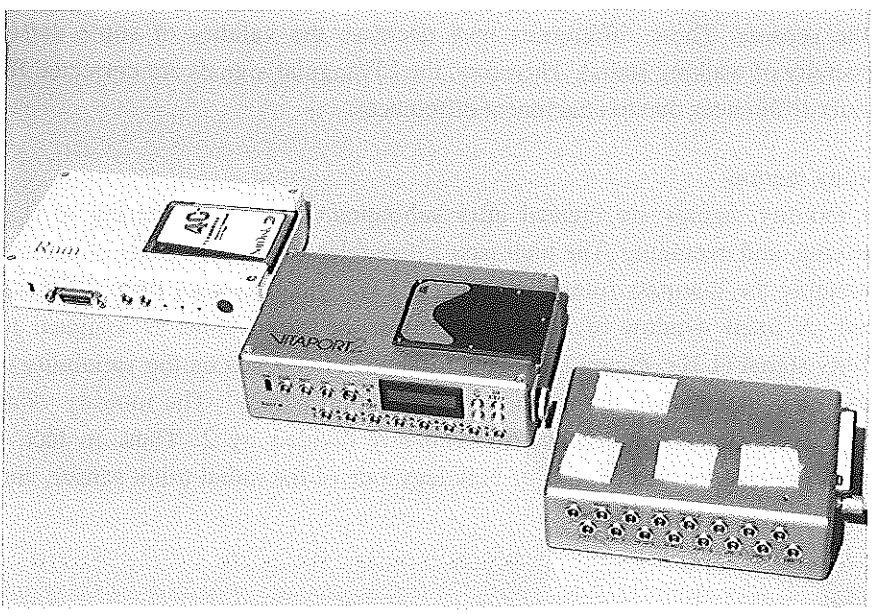

Figure 3.7. Three data recorders: the Vitaport1, the Vitaport2, and the RAM.

The data recorder must contain a so-called 'definition file', which contains the measurement set-up. The measurement set-up consists, among other things, of calibration and offset factors, sample frequency (32 Hz), resolution (12 bits), and filters (30 Hz low pass).
After the measurements, the data are downloaded onto a Macintosh computer or PC for analysis. In the analysis, three parts can be distinguished: (1) feature processing, i.e. new signals with specific characteristics are derived from the measured signal; (2) activity detection, i.e. based on the feature channels, activities are determined; and (3) post-processing, i.e. output signals of the activity detection are processed in such a way that readable and relevant information is provided. These three parts will be discussed in the following sections.

Feature processing
For activity detection, three feature signals are derived from each measured accelerometer signal.

LP/angular feature
The LP/angular signals are created after low-pass filtering (Finite Impulse Response filter, cut-off frequency 0.3 Hz). The signal is subsequently converted to 1 Hz and to angles via an arcsine transformation (range: –90 to +90 degrees; Figure 3.8). Some variability between measurements may exist due to different (angular) attachments of the sensors. These differences may decrease the validity of some detections (especially of walking stairs). Therefore, the LP/angular signals are corrected for the (initial) differences due to attachment differences.

Motility feature
The motility signals are created after subsequent high-pass filtering at 0.3 Hz, rectifying, and averaging. The high-pass filtered derivative is actually effectuated by subtracting the low-pass filtered signal (see LP/angular feature) from the measured signal. A fixed window of data is averaged; the mean value is assigned to the motility signal (1 Hz; Figure 3.8). This mean value depends on the variability of the measured signal around the mean, or 'acceleration energy'. The motility values of both leg sensors are added and subsequently divided by 2. Thus, in fact, three motility signals are created: one of both legs, one of the radial trunk sensor, and one of the tangential trunk sensor.

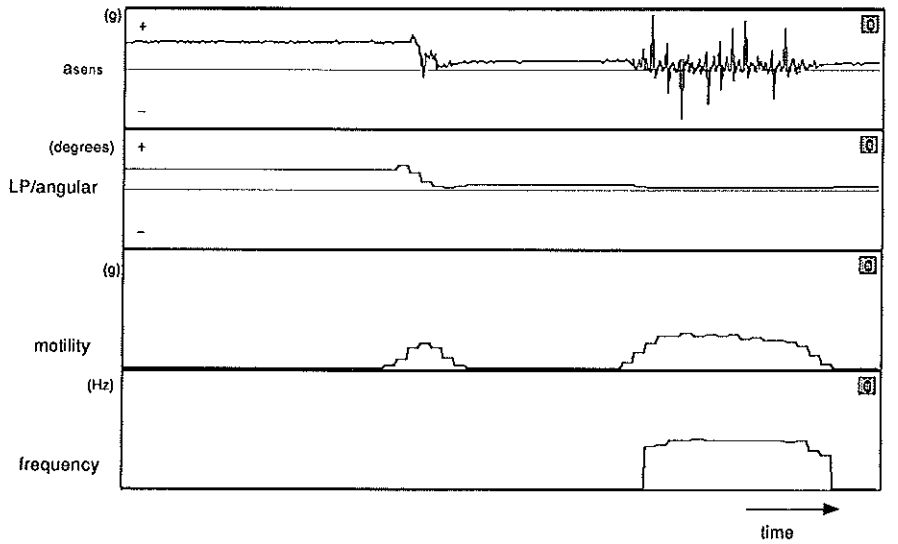

Figure 3.8 . The measured signal of one thigh during subsequent sitting (squat position), standing, walking and standing, and the calculated LP/angular, motility, and frequency signals.

Frequency feature

The frequency feature signal is based upon a band-pass filtered derivative (0.3–2 Hz for the legs; 0.6-4 Hz for the trunk), also with the use of Finite Impulse Response filters. This band-passed signal, which ideally has a sinusoid shape with the movement frequency of the segment the sensor is attached to, is the input of the FTFT procedure. This procedure consists of an instantaneous frequency analysis, and it determines instantaneous the frequency and amplitude/envelope of the band-passed signal. Whether the calculated frequency will be regarded adequate or not, depends on three pre-set criteria in the so-called FTFT Knowledge Base: the frequency range, the amplitude ('power') range of the band-passed, and the variability of the detected frequency. If the current signal does not meet all pre-set criteria, no valid frequency is assigned; otherwise the frequency is assigned to a so-called 'frequency signal' and compressed to 1 Hz. Like the motility signals, three frequency signals were constructed: one of both legs (the mean of both leg frequencies), one of the radial trunk sensor, and one of the tangential trunk sensor.
So overall, ten signals were calculated: 4 angular signals, 3 motility signals, and 3 frequency signals, all with a time resolution of 1 second.

Activity detection

In the analysis program, 23 activity subcategories were distinguished. So, ten input features are used to distinguish 23 activity subcategories. For each subcategory, a minimum and maximum value is pre-set in the ADKB (see Table 3.1). For consecutive moments in time (1 second), for each subcategory and for each feature, the 'distance' is calculated from the actual feature value to the pre-set range. The three features have different units: degrees, g, and Hz. To allow a proportional influence of all features, some features are blown-up, or the calculated distance is multiplied. If an actual feature value is within the pre-set range of a specific activity subcategory, it does not add for the distance for that activity. The calculated distances of the 10 features are added for each subcategory; the activity with the shortest distance in the end will be selected. If an activity is detected, but the distance is above a pre-set general threshold, indicating a relatively high degree of unreliability, the category 'unknown' is selected.

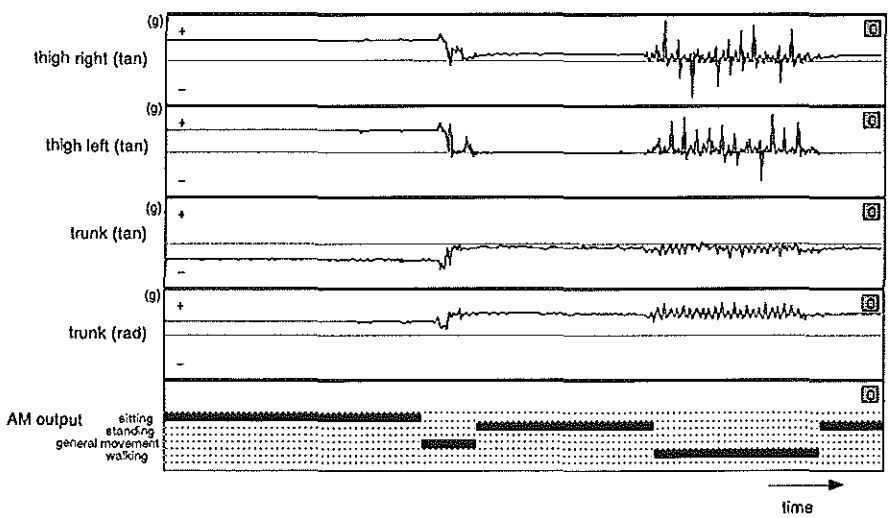

Figure 3.9. Four measured signals and the Activity Monitor output signal of the same measurement period as in Figure 3.8.

Post-processing

After the activity detection, some (optional) post-processing procedures take place.

From subcategories to main categories

Although most of the 23 activity subcategories are required initially to avoid mis-detection, not all 23 subcategories have to be of interest in a later phase. Reducing

the number of activities by taking some activities together may be desirable. In the studies performed, the subcategories are reduced to the main categories described, although other choices can be made.

Duration threshold

Each second an activity is selected. The advantage is the possibility to trace short-lasting activities as walking only a few steps, and consecutive transitions in a short time. This high resolution also has two disadvantages. First, error detections mostly are of short duration, and thus will be present if a one-second resolution is used. Second, the number of activities will increase considerably, while (very) short activities may not be of interest. Therefore, a post-processing procedure is included by which activities below a certain duration are 'deleted'. In fact, each sample a frame of a number samples is examined; the activity that is most frequently detected in that frame is assigned to that sample. The size of the frame determines the duration threshold. In our studies, a threshold of 5 seconds was applied.

Detection of transfers

All activity categories, with the exception of general movement, can be associated with body postures, e.g. walking with standing, cycling with sitting, and so on. In this way the type of transfer, e.g. sitting to standing, lying to siting, can be determined. Some activities performed may be in 'grey areas' between two postures. For example, sitting slopped in an easy chair may be detected as sitting, but also as lying on the back. It may also happen, that another posture is selected with only very small changes in angular position of thigh and trunk. Therefore, if a change in posture is noticed, the change in angular position of the thighs and trunk is calculated also, and added to an overall measure. Only if this measure exceeds a pre-set threshold, a change in posture is determined as a transition.

Statistics

Several measures can be derived from the activity detection; the measures discussed in this paragraph are only examples of several options. For each activity the total duration can be calculated. Furthermore, a frequency histogram per activity can be made, with duration categories on the X-axis (e.g. 0-10 seconds, 10-30 seconds, etc). The number of each transition category can be calculated. Motility data can be used as outcome measures. Furthermore, heart rate data can be combined with activity categories; e.g. mean heart rate during each activity category. The measures may comprise the whole measurement period, or one or more parts of it.

Discussion

In this chapter the development and characteristics of the AM are described, from the first version of the AM to the recently available extended version of the AM. Development of such an instrument is an ongoing process, and will continue over a long time. Especially the knowledge bases may change; this is no problem, because the structure of the analysis program allows user-specific or measurement specific settings. The settings within the knowledge bases are usually a matter of balance: changing the settings may improve the detection of one or more activities, but mostly worsens the detection of other activities at the same time. It depends on the research question, which settings should be used.
In the initial phase of development, standardised and controlled measurements are necessary for studying the feasibility of the instrument. The data from the master study can not be used as validity data. To show the validity of the AM, separate validity studies need to be performed. These studies should also include more natural activities in a more natural environment. Three such validity studies are described in chapters 4, 5, and 6, based on the first version of the AM analysis program. In chapter 7 the extended version of the AM is applied on the data of the studies described in chapters 4, 5, and 6.

Acknowledgements

I would like to thank Wim Martens for his contribution to this chapter. The work described in this chapter was financially supported, in part, by the *Algesiologisch Instituut*, Rotterdam, the Netherlands.

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Validity and reliability of measurements obtained with an 'Activity Monitor' in people with and without a trans-tibial amputation

Summary

In this study the validity and reliability of measurements obtained with an 'Activity Monitor' (AM) were examined. The instrument is designed to monitor ambulatory activity by use of accelerometer signals, and it detects several activities associated with mobility (standing, sitting, lying, transitions, and dynamic activities).

Four men with a trans-tibial amputation and 4 men without a trans-tibial amputation participated. The subjects performed normal daily activities, during which accelerations were measured and videotape tape recordings were made (reference method). Validity was assessed by calculating agreement scores between AM and videotape output, and by comparing the number of transitions and the duration of activities determined by both methods.

The overall agreement between the AM and the videotape was 90%. Other agreement scores and the determination of the number of transitions and the duration of activities were generally within a range of error of 0% to10%.

The reliability and validity of the AM measurements appeared to be good, which supports its potential use in rehabilitation and physical therapy.

Introduction

Locomotion, ambulation or mobility are important aspects of rehabilitation.[4,7,13,20,33,38,41,42] Many techniques are used for acquisition of mobility data including the use of questionnaires,[2,5,8,31,34,45] observation,[5,12,18,30] diaries,[28,35] kinetic and kinematic systems,[6,16,18] mechanical and electronic motion sensors,[21,24,25,30] and types of activity monitors.[1,3,15,17,36,37,39] The selection of a technique depends, among others, on the kind of information required. If unobtrusive, reliable, and valid measurements are required of a large and specific set of mobility activities during normal daily life in a person's own environment, current techniques fail to some extent. We therefore developed an 'Activity Monitor' (AM), an instrument that can be used for long-term monitoring of ambulatory activity by use of accelerometer signals and for assessment of the quantity (when, how long, how often) and quality (how performed) of several mobility activities. The activities include static activities (i.e. standing, sitting and three different modes of lying), dynamic activities (i.e. walking, climbing stairs, cycling, using a wheelchair), and the transitions between the static activities. Among other types of subjects, our studies will include people with amputation of the leg, because restricted mobility is a major problem for these persons. Until now, the development of the instrument comprised the selection of the type, number and location of sensors, and optimisation of analysis algorithms (see chapter 3),[11,40] based on data for subjects without disease or impairment. The validity of measurements obtained with the instrument, however, needs much attention (i.e. whether AM-derived measurements actually reflect the subject's activities). Validity may be influenced by factors such as age, gender, height, weight, disease or impairment, phase of rehabilitation, amputation level, and setting. The aim of this study, therefore, was to investigate the reliability and validity of AM-derived measurements, obtained for persons with and without an amputation. The main research question was: Can the type and duration of activities, and the number of transitions be validly measured by the AM? A secondary research question was: Does the AM function at the same level of accuracy (1) when measurements are repeated, (2) when the instrument is used with different subjects, and (3) when the instrument is used with persons with and without a trans-tibial amputation?

Methods

Activity Monitor

The AM consists of accelerometers, a portable data recorder and a computer with analysis programs. In this study IC–3031 uni–axial piezo–resistive accelerometers (1.5×2×0.5 cm) were used. The signals of these sensors consist of both a component of

the gravitational acceleration and a component of other accelerations (see chapter 3).[11,40] The magnitude of these components depends on the direction of these accelerations with regard to the sensitive axis of the sensor, and their magnitude.
Four sensors were fixed on the skin by means of double-sided tape. Two were attached on the thighs, halfway the spina iliaca anterior superior and the upper side of the patella, and 2 sensors were attached on the lower part of the sternum, perpendicular to each other. The trunk sensors were also held in place by means of a rubber belt. All accelerometers were attached as parallel as possible to the vertical or horizontal plane; a maximal deviation of 15 degrees was allowed. Each accelerometer was connected to a portable Vitaport1™ data recorder (13x9x4 cm; 480 gr, battery included) by a cable (under the clothes) and a Lemo-jacket. The recorder was worn on a belt around the subject's waist. Power was delivered by a rechargeable battery (270 mAh, 4.8 V). Raw signals were digitally stored on a removable memory card, with a sample frequency of 25 Hz.
After the measurements the data was downloaded onto a Macintosh computer for analysis. Although the signals of 4 sensors were measured, the signal of the left or amputated leg was not used in the analysis. The signal of this leg was measured to study the quality of walking. The data were analysed by means of Vitagraph™ and Signal Processing and Inferencing Language (S.P.I.L.™).[22]
The output of the AM is the automatic 1-second selection of one type of activity (AM output, Figure 4.1). To achieve this output, 2 types of signals were derived from each sensor signal: (1) a low-pass filtered (0.5 Hz) signal, converted to angles (3 LP/angular signals), and (2) a successively high-pass filtered (0.5 Hz), rectified, and smoothed (3 HPRS signals). The LP/angular signals are used to distinguish 5 static activities, because these activities have a unique combination of three LP/angular signals. The HPRS signal of the thigh was used to distinguish between dynamic and static activities: dynamic activities are characterised by variability of the accelerometer signal; the more 'energetic' an activity, the more variable the accelerometer signal, and the higher the value of the HPRS signal. The way in which the dynamic activities can be distinguished from each other is still under investigation. This study, therefore, was restricted to the global categories 'static' and 'dynamic', the 5 static activities, and the transitions.

Reference method

Videotape recordings were chosen as the reference method, or standard. During all measurements videotape recordings (with video clock) were made, together with the monitoring of the acceleration signals. To allow a correct comparison of the videotape and AM data, the timing of both instruments was synchronised. The videotape recordings were made and analysed by the same person, a medical student during her

research traineeship, independent from the AM analysis. In a later study (see chapter 5),[9] we investigated the inter-rater reliability of data from the videotape analysis. An overall agreement of 99.7% was found between 2 raters, indicating the reliability of the data from the videotape analysis.

The classification categories of the videotape analysis were the same as the classification categories of the AM, and the output signals of both instruments had the same 1-second time resolution. The guidelines for videotape analysis, however, were different from the guidelines for the AM analysis. The videotape analysis of lying, sitting, and standing was based on the presence and position of supporting surfaces, whereas the detection of posture with the AM was based on the angular position of the thighs and the trunk. Furthermore, in the videotape analysis, only cyclic activities (walking, climbing stairs, cycling) were determined as dynamic, whereas the AM may also determine non-cyclic activities as dynamic. After synchronisation to the signals in the AM file, the videotape recording time was converted to sample numbers. These sample numbers, and their corresponding category codes were edited so that they could be transferred to a signal in the AM file (Figure 4.1).

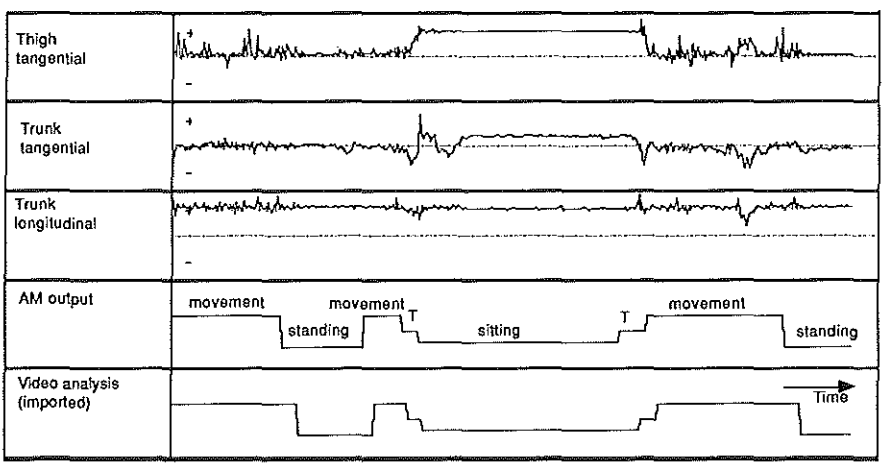

Figure 4. 1. Example of the measured accelerosignals during a 2-minute measurement period with some activities. The fourth curve is the output of the Activity Monitor (AM), each activity category represented by its own level (the words are added by way of illustration; T = transition). The bottom curve shows the imported videotape analysis, used as reference in the calculation of the agreement scores.

Protocol

The measurements were taken in a semi-natural setting in the occupational therapy department, in which a complete (representative) apartment had been installed. During

the measurements, the subjects performed several functional activities, including dressing, going to bed, preparing breakfast, peeling and cooking potatoes, watching television, reading a newspaper, shopping on another floor after taking a stairway, and riding a bicycle (using a wheelchair was not included). These activities were selected by an occupational therapist. Before the measurements, the protocol was explained to the subject. When the measurements were taken, subjects were allowed to do the activities in their own way and at their own pace. The measurements were planned to last approximately 45 minutes.

Subjects

The following inclusion criteria were used for the subjects with an amputation: one-sided trans-tibial amputation, recent (<6 months) discharge from the outpatient rehabilitation clinic, age greater than 18 years, no use of assistive devices, be able to complete the protocol, and no diseases or impairments disturbing locomotion. A rehabilitation specialist selected four male persons from a file of discharged patients (mean age 32 years, range 19-57; mean height 1.82 m, range 1.75-1-85; mean weight 74 kg, range 63-83). For each patient a person without an amputation and of the same gender, age (± 10 years), weight (± 10 kg), and height (± 0.10 m): mean age 32 years, range 23-53; mean height 1.81 m, range 1.72-1.88; mean weight 75 kg, range 65-85. The patients performed the protocol once, the comparison subjects performed the protocol twice on different days to determine test-retest reliability. A total of 12 measurements were taken.

Data analysis

Both the AM output signal (with AM activity category codes) and the videotape analysis signal (with the videotape activity category codes) had a time resolution of 1 second. Every second, the codes of both signals could be compared. In this way the number of corresponding and non-corresponding counts (1 count=1 second), and agreement scores could be calculated. Because the videotape analysis can be regarded as a standard, the following agreement scores – as validity measures of the AM – were used (research question 1):

(1) *Agreement:* the percentage agreement between all samples of videotape and AM data. Agreement was calculated according to the equation: agreement = (number of identical samples of videotape recording and AM data / total number of samples)×100%.

(2) *Sensitivity:* the degree to which each videotape activity category (representing the activities actually performed) was detected correctly by the AM. Sensitivity was calculated according to the equation: sensitivity for videotape activity category A=(number of identical samples of videotape recording and AM data when

videotape activity category is A/total number of samples for videotape activity category A)×100%.

(3) *Predictive value:* the degree to which each AM activity category agreed with the videotape activity category (representing the activities actually performed). Predictive value was calculated according to the equation: predictive value of AM activity category A=(number of identical samples of videotape recording and AM data when AM activity category is A/total number of samples for AM activity category A)×100%.

The 1-second output of the AM and videotape recording analysis allowed calculation of duration (in seconds) per activity. The number of transitions within each transition category was calculated from identified changes in posture. All calculations and comparisons were done automatically by means of SPIL software.

Simple descriptive statistical measures, such as weighted (corrected for duration of activities) mean and standard deviation were used to describe group results. The Wilcoxon matched-pairs signed-ranks test was used to show systematic differences in results between the videotape recording analysis and the AM analysis. The Mann-Whitney *U* test was used to show systematic differences in results between the patient group and the first measurement of the comparison group. All statistical analyses were done with SPSS 5.0 for MS Windows. A probability value of $P< 0.05$ was considered to indicate a significant effect.

Results

The overall agreement between videotape data and the AM output was 90%. The overall (weighted) mean of most sensitivities and predictive values equalled or exceeded 90% (Tables 4.1 and 4.2). The overall sensitivity for lying on the side and dynamic activities was somewhat lower (88% and 85%, respectively), as was the overall predictive value of the AM activity categories of sitting and dynamic activities (88% and 89%, respectively). During 890 seconds, standing (determined by videotape analysis) was detected by the AM as a dynamic activity; during 916 seconds, on the contrary, dynamic activities (determined by videotape analysis) were detected by the AM as standing. Table 4.1 also provides insight on the distribution of activities during the measurements. No significant differences in distribution existed between the groups, although the measurements in the patient group lasted, on average, longer than those in the comparison group (41 min and 34 min, respectively).

The AM slightly overestimated the total number of transitions compared with the videotape recordings (overall difference: +16 (+7%), $P<0.05$, Table 4.3). The overall duration of activities determined by the AM did not deviate from the duration of

activities determined by videotape analysis for standing (Table 4.4). The duration of sitting was overestimated by the AM (mean difference=2.2%, P<0.01).

Table 4.1. Number of corresponding (bold/italic) and non-corresponding (plain) counts (1 count=1 second) of videotape activity and Activity Monitor output, added for all measurements. The last column shows the overall sensitivity of the AM for each videotape activity category; the bottom row shows the overall predictive value of each AM activity category. The percentage bottom-right represents the overall agreement.

Video	Activity Monitor (counts)							Sensitivity (%)
	Lying back	Lying side	Lying prone	Standing	Sitting	Dynamic activity	Total	
Lying back	***853***	0	0	0	53	15	921	93
Lying side	0	***1057***	0	5	118	20	1200	88
Lying prone	0	0	***67***	0	0	1	68	99
Standing	0	0	0	***8755***	83	890	9728	90
Sitting	0	0	0	0	***4809***	108	4917	98
Dynamic activity	27	20	5	916	421	***8068***	9457	85
Total	880	1077	72	9676	5484	9102	26291	
Predictive value (%)	97	98	93	90	88	89		90

The standard deviation of the difference in agreement between the first and second measurements for the comparison subjects was 3.9% (range –5 to 4%). The standard deviations for the percentage of agreement ranged from 2.3% (range 88%-92%) in the patient group to 3.6% (range 85%-93%) for the second measurement in the comparison group. When all measurements were included, the standard deviation was 2.6% (range 85%-93%).

To examine differences in functioning of the AM in persons with and without trans-tibial amputation, variables were compared between the patient group and the comparison group (first measurement); the agreement did not differ between both groups. Some percentages for sensitivity and predictive value did, however, differ (Table 4.2). The sensitivity for standing and the predictive value for dynamic activities were higher in the patient group, and the sensitivity for dynamic activities was higher in the comparison group than in the patient group (all P<0.03).

The functioning of the AM, expressed as the correct determination of the number of transitions, did not differ between both groups (Table 4.3). The duration of dynamic activities was overestimated by the AM in the comparison group (+2.1%) and underestimated in patients (–5.9%). The duration of standing was underestimated in the comparison group (–3.1%) and overestimated in patients (+3.5%).

Table 4.2. Percentages per measurement, representing the sensitivity (S) and predictive value (PV). '-' means that the activity was not performed or detected. The weighted means for each subgroup are calculated. In the last column the agreement per measurement is shown (Agr), in the last row the weighted overall means are shown. Measurement codes: H1-H4=subjects without amputation; A1-A4=subjects with trans-tibial amputation; M1 and M2=measurement 1 and 2.

Measurement	Agreement scores (%)												
	Lying on back		Lying on Side		Lying prone		Standing		Sitting		Dynamic activity		
	S	PV	S	PV	S	PV	S	PV	S	PV	S	PV	Agr
H1M1	-	-	99	99	-	-	90	90	97	95	86	87	91
H2M1	100	97	100	98	-	-	77	91	98	97	93	85	90
H3M1	94	91	100	91	-	-	91	97	95	82	91	90	92
H4M1	-	-	93	99	-	-	77	90	97	87	88	80	86
Mean	97	93	97	98	-	-	85	92	97	92	90	85	90
H1M2	100	99	95	100	-	-	92	91	99	94	88	92	93
H2M2	100	95	16	88	-	-	77	96	98	74	94	86	85
H3M2	97	98	-	-	-	-	90	92	98	92	89	90	92
H4M2	-	-	99	100	-	-	85	88	94	91	90	88	90
Mean	98	98	70	99	-	-	86	92	97	87	91	89	90
A1M1	-	-	98	98	99	93	97	89	98	97	80	93	92
A2M1	61	100	-	-	-	-	97	92	97	69	73	92	88
A3M1	100	97	95	95	-	-	95	94	100	87	80	92	92
A4M1	99	96	100	98	-	-	96	80	100	95	71	94	88
Mean	86	97	99	98	99	93	96	89	99	86	76	93	90
Mean (overall)	93	97	88	98	99	93	90	90	98	88	85	89	90

Table 4.3. The number of 6 transition categories, determined by videotape analysis (V) and Activity monitor (AM). The data are shown per measurement, for each subgroup, and for all measurements together. Measurement codes: H1-H4=subjects without amputation; A1-A4=subjects with trans-tibial amputation; M1 and M2=measurement 1 and 2.

Measurement	Transitions (number)													
	Lying-sitting		Lying-standing		Sitting-lying		Sitting-standing		Standing-lying		Standing-sitting		Total	
	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM
H1M1	0	0	1	1	0	0	8	9	1	2	9	9	19	21
H2M1	1	2	0	0	0	1	9	9	1	1	8	8	19	21
H3M1	1	1	0	0	0	1	9	9	1	0	8	9	19	20
H4M1	0	0	1	2	1	1	7	8	0	1	8	9	17	21
Total	2	3	2	3	1	3	33	35	3	4	33	35	74	83
H1M2	0	0	1	1	0	0	7	7	1	1	7	7	16	16
H2M2	1	1	0	0	0	0	12	12	1	1	11	11	25	25
H3M2	1	1	1	1	1	1	8	8	1	1	8	8	20	20
H4M2	0	0	1	1	1	1	5	6	0	0	6	7	13	15
Total	2	2	3	3	2	2	32	33	3	3	32	33	74	76
A1M1	1	1	0	0	1	1	8	7	0	0	8	7	18	16
A2M1	2	2	0	0	2	2	8	10	0	0	8	10	20	24
A3M1	1	1	0	0	1	1	5	5	0	0	5	5	12	12
A4M1	0	0	1	1	0	1	7	8	1	0	7	9	16	19
Total	4	4	1	1	4	5	28	30	1	0	28	31	66	71
Total (overall)	8	9	6	7	7	10	93	98	7	7	93	99	214	230

Table 4.4. Duration (as percentage of the measurement time) of each activity category, determined by videotape analysis (V) and Activity Monitor (AM). The data are shown per measurement, and weighted means are calculated for the subgroups and for all measurements together. Measurement codes: H1-H4=subjects without amputation; A1-A4=subjects with trans-tibial amputation; M1 and M2=measurement 1 and 2.

Measurement	Duration (% measurement time)													
	Lying back		Lying side		Lying prone		Standing		Sitting		Dynamic activity		Total	
	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM
H1M1	0.0	0.0	9.0	8.9	0.0	0.0	36.9	36.8	23.0	23.4	31.1	30.9	100	100
H2M1	1.7	1.8	5.7	5.9	0.0	0.0	29.9	25.2	19.0	19.1	43.7	48.1	100	100
H3M1	2.6	2.7	2.5	2.8	0.0	0.0	44.5	41.9	10.9	12.6	39.6	40.1	100	100
H4M1	0.0	0.0	9.3	8.8	0.0	0.0	35.3	29.9	17.5	19.6	37.8	41.7	100	100
Mean	1.0	1.1	6.7	6.6	0.0	0.0	36.6	33.5	17.8	18.8	37.9	40.0	100	100
H1M2	5.3	5.4	4.3	4.1	0.0	0.0	30.2	30.7	26.4	27.6	33.8	32.3	100	100
H2M2	3.9	4.1	6.7	1.2	0.0	0.0	27.4	22.2	21.1	27.6	40.9	44.9	100	100
H3M2	10.9	10.8	0.0	0.0	0.0	0.0	33.5	33.0	15.5	16.6	40.1	39.6	100	100
H4M2	0.0	0.0	9.3	9.2	0.0	0.0	34.3	32.9	15.1	15.7	41.3	42.2	100	100
Mean	5.1	5.1	5.0	3.5	0.0	0.0	31.3	29.6	19.6	22.1	39.0	39.8	100	100
A1M1	0.0	0.0	2.7	2.7	3.0	3.1	39.4	43.0	25.5	26.0	29.4	25.1	100	100
A2M1	4.9	3.0	0.0	0.0	0.0	0.0	49.2	51.6	14.4	20.3	31.6	25.2	100	100
A3M1	7.3	7.6	1.9	1.9	0.0	0.0	36.9	37.2	23.5	26.9	30.4	26.4	100	100
A4M1	4.6	4.8	5.8	5.9	0.0	0.0	40.3	48.1	13.0	13.7	36.3	27.5	100	100
Mean	4.2	3.8	2.4	2.5	0.7	0.7	42.0	45.5	18.7	21.6	31.9	26.0	100	100
Mean (overall)	3.5	3.3	4.6	4.1	0.3	0.3	37.0	36.8	18.7	20.9	36.0	34.6	100	100

Discussion

The agreement scores in this study were generally within a range of error of 0% to 10%. A comparison of the results of our study with those of other studies and instruments is not possible because instruments discriminate different activity categories,[1,3,15,43] protocols consisted of different activities,[17,39] or validity was calculated following another or unknown method.[23,36,37] Kiani et al.[23] have developed the 'Ambulatory Monitoring of Motor Activities' (AMMA) system using accelerometers and an artificial neural network. The set of activities that can be detected are similar to the activity categories of the AM. Although the technique seems promising and a 95% reliability is reported, the used validation technique is questionable. The measured accelerosignals were the input of both the AMMA system and the reference method (visual interpretation of the signals). Walker et al.[43] reported a validation study of an activity monitor based on mercury switches and accelerometers. Validation was studied in terms of steps counted; validation of body positions was not reported. Stock and colleagues[36,37] used a 'microcomputer-based system for the assessment of postoperative fatigue' consisting of a posture timing module, an activity module, and a heart rate module. Although they reported maximum error percentages of about 5%, they did not clearly describe how these percentages were obtained, and the results, therefore, are difficult to interpret. Anastasiades and Johnston[1] used electromyography to discriminate between static and dynamic activities. Static activities could not be distinguished from each other and inter-individual comparison appeared to be difficult. Fahrenberg et al.[17] applied acccelerometers and a hydrostatic tube to monitor their subjects' ambulatory activity. The same sensors were used by Tuomisto et al.,[39] who used also electromyography. In both studies, only a small number of standardised activities were performed to determine validity. Instruments for measuring walking periods were validated.[3] Although the measurements they obtained showed reasonable validity, the instruments they used measured only walking and stair-climbing performance; these instruments were not designed to measure cycling and different body positions. Several workers[19,26,30,32] used accelerometry to distinguish between static and dynamic activities and to determine the level of activity. Due to the limited scope of these instruments, comparison with the AM is not useful. The instrument used by Diggory et al.[15] is designed with a tilt switch to measure time in the upright position. Although their validity study gave good results, possible errors were reported when lying prone, cycling, and climbing stairs. Furthermore, dynamic activities formed no part of the output of the instrument.

Some limitations of the present study can also be posed. The activities were prescribed and limited in number, and they were performed in the same environment. The

subjects selected for the patient group had trans-tibial amputations and had finished rehabilitation. Furthermore, they were relatively good walkers, and 3 of the 4 subjects were fairly young. In our view, the generalisability of the results of our study to unsupervised measurements is enhanced by some other characteristics of the study. The measurements were done in a semi-natural setting, and the activities were functional and selected by an occupational therapist who did not participate in the study. The subjects could perform the activities in their own manner, and the method for assessing the validity of the measurements was critical. If the AM is to be used in the real-world environment of patients (e.g. with external vibrations due to car or train), extension of the kinds of movements and postures,[27,30] and with patients with other movement patterns[29] (e.g. other impairments, other amputation levels, other phases of rehabilitation, different ages), however, further study of the reliability and validity of the measurements obtained with the device is needed. In later validation studies of the AM, used with patients with failed back surgery in their own environment (see chapter 6)[10] and in subjects without known disease or impairment in the setting of a psychophysiological study (see chapter 5),[9] results similar to those in our study were obtained. Overall percentages of agreement of 87% and 88% were found, respectively, supporting the validity of the measurements and the robustness of the AM.

While we were taking the measurements in our study, there were no problems with either the sensors or the recorder system, and neither the cables nor the recorder interfered with any of the subjects' activities. We observed, however, a tendency of the sensors (especially the trunk sensors) to come somewhat loose from the skin, probably due to chest hair, perspiration, or the rubber belt. In later studies, therefore, we taped the sensors onto the skin with other material (e.g. Kushionflex). The method of fixation of the sensors is still a focus of our research interest. Recently, we obtained measurements (1-2 days) with a recorder of a slightly larger size and weight than used in the present study and found that the measurement system may cause some discomfort. The system cannot be used in a wet environment (e.g. while bathing or taking a shower). Some people find the recorder or the cables disturbing while sleeping or while dressing or undressing, and some people dislike being seen wearing the instrument. Therefore, we are investigating methods that will allow patients to easily attach and remove the recorder and sensors themselves, without compromising measurements. Reducing the weight and size of the recorder may also increase comfort and applicability. Five prototypes of a recorder (RAM), of approximately the same size and weight as the recorder used in our study, has been developed that enables continuous measurement of accelerosignals for more than 48 hours (compared with approximately 2 hours for the recorder used in our study), without replacement of flash cards or batteries. The recorder receives power from 4 penlight batteries and allows

measurement of up to 8 signals. Data are stored on a flash card of 40 MB (or more). Reading the data from the flash card takes about 1 minute. The analysis of the data (for a 2-day measurement) takes less than one hour. Generally, we start the measurements in person's home, and the data are downloaded in our laboratory; downloading the data by telephone is now an option.

The AM was not free of errors. The investigation of even small errors can be used to increase knowledge about the functioning of the AM.
Relatively often, standing according to videotape analysis is detected as a dynamic activity by the AM, and, on the contrary, some dynamic activities according to videotape analysis are detected as sitting or standing by the AM (see Table 4.1). The distinction between the two global categories static and dynamic occurs early in the analysis program. Errors at this point have irreversible consequences for subsequent phases of the activity detection. Furthermore, small time shifts between videotape and AM signals, infrequent activities that were to difficult to analyse using videotape recordings, and timing inaccuracies in videotape analysis (as shown in Figure 4.1) may have some effect on these errors. The inter-rater reliability of data from the videotape analysis in a similar study performed later, showed a percentage of agreement of 99.7%. The effect of videotape analysis errors on the data will be small, and that result showed that the videotape recording can be used as a standard. Other causes were shown to be more important.
The detection of an activity as static or dynamic strongly depends on an adjustable threshold applied on the HPRS signal of the leg, and the setting of this threshold is a matter of optimisation. The detection of standing as a dynamic activity (e..g. due to leg movement during standing, but not walking) and the detection of dynamic activities as standing (e.g. due to shuffling) are quantitatively important misinterpretations; they lead to a decrease in overall agreement of about 6%. Generally, however, the threshold seemed well-chosen, as can be concluded from the small difference (–1.4%) between the duration of dynamic activities determined by AM and videotape recordings, and the almost equal number of counts of the detection of dynamic activities as standing and of the detection of standing as a dynamic activity.
A dynamic activity is rather frequently (421 seconds) determined as sitting (Table 4.1). Cycling appeared to be the main activity that caused this discrepancy. During cycling, there were periods when the subject does not move his legs. These periods were periods of dynamic activity according to videotape analysis, while the AM detected sitting. Furthermore, if the legs were moving during cycling, the acceleration energy was sometime too low to cause the detection of dynamic activities. To quantify the effect of these errors, we studied the occurrence of these errors. The determination of a dynamic activity as sitting was, for about 300 seconds of output, explained by these

errors. If these errors had not occurred, the sensitivity of dynamic activities would have increased from 85% to 88%, and the predictive value for sitting would have increased from 88% to 93%.

If decision errors for static and dynamic activities are excluded (i.e. the 'dynamic activity' row and column in Table 4.1 are not included in the analysis), the quality of the static activity detection can be assessed. The overall agreement then increases to 98%, and most sensitivities and predictive values equal, or come close to, 100%. Only the overall sensitivity for lying on the back (94%) and lying on the side (90%), and the overall predictive value of sitting (95%) are then clearly lower than 100%. This finding is due to two misinterpretations. In one comparison subject (H2M2, 118 seconds) lying on the side with the upper side of the trunk elevated and legs rotated (while reading a book) was detected as sitting. In one patient (A2M1, 53 seconds) lying on the back (according to the criteria of the videotape analysis) with the trunk supported by a few pillows was also detected as sitting.

The total number of transitions is slightly, but systematically, overestimated by the AM. This finding was mainly due to overestimation of the sitting-to-standing and standing-to-sitting transitions (e.g., foot on a chair while donning and doffing shoes and socks [videotape: standing; AM: standing-sitting-standing]), and to overestimation of more complex transitions (e.g. from standing to lying on the back via sitting and lying on the side).

The duration of sitting also was overestimated. When data are corrected for the 2-fold misinterpretation of static activities and the determination of cycling as sitting, a considerable improvement is reached. Existing differences in duration (last row, Table 4.4) will almost disappear.

The differences in the agreement scores in the first and second measurement of the 4 comparison subjects were small, as were the differences in agreement scores among subjects within the same group, even though considerable individual differences in the way of performance were observed.

The mean agreement was the same for the patient group and the comparison group (90%). Comparison of the 2 groups, however, revealed some differences in results. In general (also after correction for the determination of cycling as sitting), dynamic activities were overestimated in the comparison group (+2.1%), but underestimated in the patient group (–5.9%). Standing, however, was overestimated in the patient group (+3.5%) but underestimated in the comparison group (–3.1%). These differences between groups were statistically significant. The data suggest that the persons with trans-tibial amputations in this study walked, cycled and climbed stairs less energetically (i.e., with less acceleration variation) than did the comparison subjects.

The potential of the AM can be increased by simultaneous measurement of, for example, electrocardiographic activity in quantifying strain, or markers. Furthermore,

we assume that the accelerometer signals contain much information on the quality (how performed) of the activities. Our research focuses especially on the quality of walking (spatio-temporal variables, stability).
There may be a considerable inter-day and inter-subject variability,[3,14,45] which may differ by patient group. This variability will determine the number of days over which measurements should be taken and the number of subjects included in group studies. Future studies should investigate these 2 types of variability.
The AM is an instrument that provides data on the activities a patient actually performs during daily life. Generally, in many medical disciplines and also in physical therapy, the decisions about treatment and the evaluation of that treatment are increasingly tuned to and related to functional performance. For example, the effectiveness of physical therapy is not only determined by the change in joint mobility, but also by changes in the performance of activities of daily living. The usefulness of the AM in physical therapy intervention, therefore, will depend on the treatment goals. The AM can be a powerful instrument in evaluative studies, especially when the formulated goals are related to the quantity of movement and postures. We believe that the AM will first be used in research. After making the instrument more dedicated, improving its usability, and reducing its costs, however, we expect that the AM will also be used in the practice of physical therapists.

Conclusion

Activity monitoring by means of accelerometry proves to be a promising method to obtain reliable and valid measurements of the activities a patient actually performs during daily life, which is essential in rehabilitation and physical therapy. Research with less obtrusive devices in real-world settings is now needed.

Acknowledgements

The research described in this paper was part of the EUREKA project DYNAPORT, which was financed, in part, by the Dutch Ministry of Economic Affairs.

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Quantification of physical activities by means of ambulatory accelerometry: a validation study

Summary

The objective of the study was to assess the validity of an 'Activity Monitor' (AM) within a psychophysiological study. The AM was based on four body-fixed accelerometers and discriminated postures, transitions, and dynamic activities.
Three subjects participated in each of two 4-hour sessions. During each session, consisting of two protocols, ambulatory accelerometer and heart rate measurements were made. The output of the AM was compared with simultaneously recorded videotapes.
An overall agreement between AM and video of 88% (spontaneous part) and 96% (standardised part) was found. The number of transitions and dynamic periods, and the duration of activities were well determined. Posture-related heart rate changes were demonstrated. A three-sensor configuration had minimal influence on the validity scores.
The AM appeared to be a valid instrument to quantify aspects of physical activity, and offers new possibilities for ambulatory psychophysiological research.

Introduction

Reliable and accurate quantification of the number, duration, and intensity of physical activities during normal daily life is becoming increasingly important from two perspectives.

First, there is a growing interest in ambulatory studies in psychophysiological research due to recently formulated hypotheses that quantification of cardiovascular responses to real-life stressors may be more salient than laboratory reactivity assessments in determining individual risk factors for cardiovascular disease.[12,17] Because ambulatory cardiovascular measurements are greatly influenced by changes in body posture and locomotor activity,[e.g. 1] and cardiovascular stress responses have been shown to be posture-dependent,[e.g. 11] quantification of postural changes and physical activity is essential for the interpretation of cardiovascular data.

Second, apart from functioning as a control for the interpretation of ambulatory recorded physiological signals, measures on the quantity of body postures and locomotor activity can also be important effect variables. A detailed assessment of the physical activities performed during daytime ambulation offers an interesting approach to studying behavioural processes and to evaluating the effects of clinical or experimental manipulation on everyday life.[10] Additionally, relationships between activity variables and psychophysiological variables can then be studied.

Current methods of measuring the actual activities that a person performs during daily life – such as questionnaires, observation, diaries, and kinetic and kinematic systems – have practical or methodological shortcomings.[9,10,19] By and large, mechanical and accelerometer-based actometers are the solution to these shortcomings. Their output, however, provides general information on activity and energetic turnover, and no detailed information on the quantity (when, how often, how long) of specific activities.[e.g. 8,9,10,14] Some activity monitors have already been described, which are able to provide more detailed information on postures and activities.[1,5,6,13,16] In general, however, these activity monitors are not thoroughly validated, concern rather complex configurations or different types of sensors, or provide data with other purposes (e.g., to predict heart rate). Due to progression in technology, the development of small and more advanced portable AM systems has now become possible.

The above arguments have led to the development of an AM: an instrument based on ambulatory monitoring of accelerometer signals, and aimed at the long-term (>12 hr) assessment of the quantity (when, how often, how long) and quality (how performed) of several mobility-related activities during normal daily life (see chapter 3).[4] These activities include the static activities (or postures) of standing,

sitting, and different modes of lying (prone, on the back, on the side), the transitions between these postures, and the dynamic activities of walking, climbing stairs, cycling, and driving a wheelchair. The AM consists of four accelerometers, a portable data recorder, and a computer with programs for analysis. By processing and combining the accelerometer signals, it is possible to distinguish between the different postures and activities. The theoretical background, design, and the developmental phase of this instrument has been described elsewhere (see chapter 3).[4,18]

The AM has been developed from a physical medicine and rehabilitation point of view, to be used in descriptive and evaluative studies. However, the application of the AM in other fields has been considered from the beginning. Before the instrument can be applied in various clinical or research settings, however, its validity needs to be studied thoroughly. The selection of the setting and subjects of the validity studies will depend on the future use; a good criterion validity of the AM in one kind of study does not automatically result in a good validity in other fields of research.

So far, the validity of the AM has been investigated with healthy subjects and amputees;[3] the agreement scores between AM output and video analysis ranged from 85 to 93%. The present study was performed within the context of a psychopharmacological experiment, and differed from the previous validation study in environment, activities performed, type of subjects, and use of medication; a separate validation study was therefore justified. Furthermore, in a phase of development and validation, knowledge of the error sources is important. This knowledge can be used to make adaptations to the analysis software to increase the validity of the AM, or to find and understand the limitations of the instrument. Detailed validation studies are essential to study error sources, their effect on validity, and the adaptations that may be required.

Although the development of the AM started with a 3-sensor configuration, nowadays a 4-sensor configuration is our standard, due to our interest in the quality of walking. However, when interest is only on the quantity of activities, a 3-sensor version may be appropriate from validity arguments, and preferable from the standpoint of power supply, data storage and usability. The validity of the AM when using a simpler configuration, therefore, needs to be studied.

The aim of the present validation study was two-fold:

(1) To assess the validity and feasibility of the AM within the setting of a psychophysiological study, and to detect possible sources of error; and
(2) To assess the validity of the AM within this study if a simpler (i.e. using fewer sensors) configuration is used.

Our measurements were performed within the context of a study that aimed to assess the effects of benzodiazepines on subjective mood and cardiovascular functioning, in relation to normal daily activities. The circumstances under which the ambulatory measurements for the validation study were made equalled the ambulatory measurement situation of the psychopharmacological study as much as possible.

Methods

Context of the validation study: psychophysiological effects of benzodiazepines
The effects of two oral dose levels of alprazolam (0.5 and 1 mg) and one dose level of lorazepam (2 mg) were compared in a double-blind randomised placebo-controlled, cross-over study. During the morning part (4 hr), cardiovascular and catecholaminergic responses were studied in a standardised laboratory schedule of several conditions. During the afternoon (for 4 hr), spontaneous body movements and postural changes were evaluated by means of ambulatory accelerometry, in addition to the ambulatory measurement of heart rate. The subject stayed in a living room in the hospital, where he could move around freely, study, relax, or sleep. Details of the study have been published elsewhere.[2,15]

Subjects
Three young, healthy male volunteers (age range 19-24 years; height range 1.85-1.93 m; mass range 72-81 kg) entered the validation study. The inclusion criteria used were the same as used in the psychopharmacological study: the subjects were male, aged between 18 and 40, not allergic to benzodiazepines, no drug users, non-smokers, and did not suffer from diseases of the locomotor system. For 3 days before each measurement the subject was not allowed to perform excessive physical or mental tasks, or to drink more than one unit of alcohol a day. The subjects underwent a medical examination by an independent clinician, and were asked to sign an informed consent form.

Design
The design of the validity study resembled the design of the psychopharmacological study as much as was possible. The three subjects were each measured twice, with an interval of at least 1 week between measurements. At 8:00 a.m. the subject took a 2-mg oral dose of lorazepam or the placebo, administered double-blind. In the morning the subject could move around freely, within the hospital. The measurements took place in the afternoon. The measurements consisted of two protocols. In the spontaneous protocol the subject had to stay in a living room (4×6 m), which contained a writing desk and a chair, a bed, and an easy chair with a

coffee table. The subject was free to choose his own activities. These measurements lasted about 4 hr. Because the subject would probably not perform an extended set of activities, he had to perform 40 different forms of standing, sitting, lying, and walking (standardised protocol) in about 15 min at the end of each spontaneous protocol.

Instruments

Activity Monitor

In this study, four IC-3031 uniaxial 3g-piezo-resistive accelerometers (1.5×2×0.5 cm) were used. The signals were composed of a vector of the gravitational acceleration (giving absolute angle information), and a vector of the actual acceleration of the sensor. In static activities, when no actual accelerations occur, the value of the signal ranges from −1 g to +1 g (1 g=9.81 $m.s^{-2}$), which depends on the position of the sensor as compared with the vertical gravitational force.

Two sensors, their sensitive axis almost parallel to a sagittal axis (or sensitive in X-direction) while standing, were each attached to the skin of the front of upper legs. The other two sensors were attached to the skin of the sternum, perpendicular to one another. While standing, the sensitive axis of one trunk sensor was almost sagittal (or in X-direction), the other almost longitudinal (or in Y-direction). The sensors were fixed by double-sided tape, and the placement of the sensors was standardised. The accelerometer on the upper leg was attached as vertically as possible during normal standing of the subject, approximately halfway between the spina iliaca anterior superior and the upper side of the patella. A maximal deviation of 15 degrees was allowed. The two sensors on the trunk were attached as vertically/horizontally as possible while standing, with the same deviation allowed.

The accelerometers were connected to a portable Vitaport1™ data recorder; the signals were digitally stored on a memory card, with a sampling and storage frequency of 16 Hz and a 12-bits resolution. After the measurement the data were downloaded onto a Macintosh IIci computer. Analysis took place by means of the signal processing and inferencing language (S.P.I.L.™).[7] For the analysis described in this paper, all the signals were (a) low-pass filtered (Finite Impulse Response, 0.5 Hz) and converted to angles (4 LP/angular signals), and (b) successively high-pass filtered (the 0.5 Hz low-pass filtered signal subtracted from the original signal), rectified and smoothed (4 HPRS signals). These derived time series had a frequency of 1 Hz. Figure 5.1 shows an example of an upper leg signal and its two derivatives for subsequent sitting, standing, walking, standing, walking, standing and sitting.

Theoretically, four LP/angular signals are sufficient to distinguish different postures from one another (see chapter 3).[4,18] Each posture was divided into two or more sub-postures in the analysis procedure. For each sub-posture, a range with a minimum

and maximum value was predetermined for each LP/angular signal; each sub-posture had a unique set of four ranges. Every second, the distance from each LP/angular sample to the corresponding minimum or maximum value was calculated, and added for all four sensors to a 'total distance'. The shorter the total distance, the higher the possibility that a (sub)posture was estimated correctly; if the value of the sample was between the maximum and minimum value, the distance was zero. By decreasing the number of input signals and running the analysis program again, the effect of the use of fewer sensors on the results could be investigated. The detection of transitions was derived from changes in posture; however, a transition was detected only if a significant change in posture occurred.

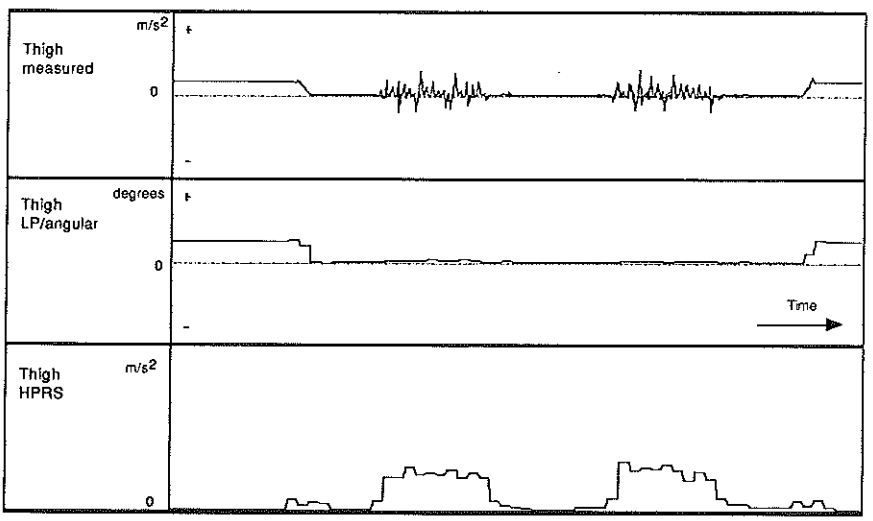

Figure 5.1. Example of the measured signal of the thigh sensor during subsequent sitting, standing, walking, standing, walking, standing, and sitting; the 0.5 Hz low-passed (LP)/angular derivative of the measured signal (middle graph); and the high-passed, rectified, and smoothed (HPRS) derivative of the measured signal (lower graph).

Discrimination between static and dynamic activities was achieved by applying a threshold to the HPRS signals of the upper leg accelerometers (Figure 5.1). The more 'dynamic' an activity was, the more variable the accelerometer signals, and the higher the value of the HPRS signal. In static activities the HPRS signal equalled or came close to zero. If the HPRS signals of both leg sensors were above the applied threshold of 0.05 g, a dynamic activity was detected. If an activity was detected as dynamic, it overruled the posture detection. The analysis program contains a

procedure in which all activities – except transitions – lasting less than 5 s were deleted. The way in which the dynamic activities can best be distinguished from each other is still under investigation. This study, therefore, restricts itself to the global categories 'static' and 'dynamic', the transitions, and the five postures. Figure 5.2 shows the (X) signals of the leg and trunk, and the automatic AM output for the same example as Figure 5.1.

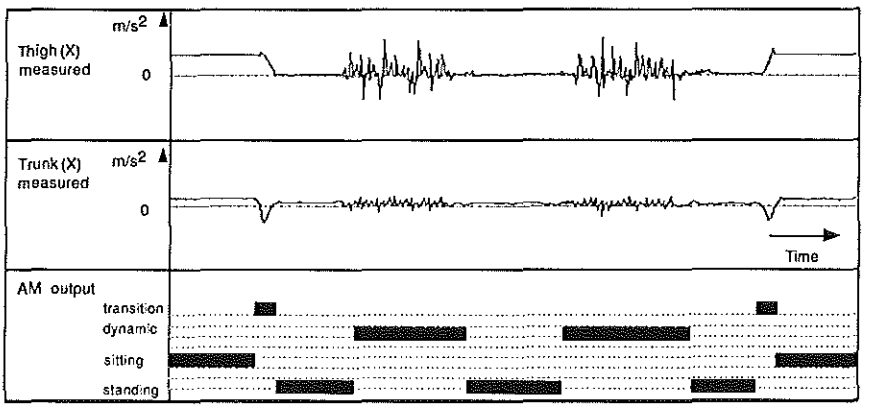

Figure 5.2. The measured X signals of the upper leg (upper graph) and trunk (middle graph) during subsequent sitting, standing, walking, standing, walking, standing, and sitting. The bottom graph represents the output of the activity monitor.

Heart rate

Heart rate (beats per minute) was derived from a pre-cordial electrocardiogram (ECG) lead. The ECG was processed in the Vitaport recorder (R-wave triggering, 2-ms accuracy), transposed to heart rate time series, and stored at a sample frequency of 4 Hz.

Reference method

With a camera placed in the living room, video recordings were made as the reference method, or gold standard. All video recordings were analysed independently from the AM output by the same person, with a time resolution of 1 s. To determine the inter-rater reliability, a second rater analysed the first 2 hr of each placebo measurement. An overall agreement of 99.7% was found between the two raters.

Though the output categories of the video analysis were the same as the output categories of the AM, the general rule of the formulated guidelines for video

analysis was different from the one of the AM. For example, the video analysis of lying, sitting, and standing was based on the presence and position of supporting surfaces, whereas the posture detection of the AM was based on the angular position of legs and trunk. The video recordings were synchronised with the accelerometer recordings. Again, activities – except transitions – lasting less than 5 s were disregarded. The analysed video recordings were transferred to a signal in the AM-file: all calculations and comparisons were done automatically by means of S.P.I.L.™ software.

Data analysis

Spontaneous protocol

The output of the AM was compared with the synchronised output of the video analysis, with a time resolution of 1 s. Because the video analysis was a gold standard, the following agreement scores – as validity measures of the AM – were calculated:

(1) *Agreement*: the percentage of agreement between all samples of video and AM. Agreement was calculated according to: agreement=(number of identical samples of video and AM/total number of samples)×100%.

(2) *Sensitivity*: the degree in which each video activity category (representing the activities actually performed) was detected correctly by the AM; the sensitivity percentage was calculated according to: the sensitivity for video activity category A=(number of identical samples of video and AM for video activity category A/total number of samples for video activity category A)×100%.

(3) *Predictive value*: the degree in which each AM activity category agreed with the video activity category (representing the activities actually performed); this value was calculated according to: the predictive value of AM activity category A=(number of identical samples of video and AM for AM activity category A/total number of samples for AM activity category A)×100%.

Sensitivity and predictive value were not calculated if the number of AM or video samples of a specific activity category was less than 60 (60 samples equals 1 min). The weighted mean (i.e., corrected for duration of activities) was used to describe group results. The number of dynamic periods (lasting 5 s or longer) and the number of transitions per transition category were calculated. The duration was calculated per activity category. The (relative) difference between video and AM in number and duration was calculated as follows: the difference AM–video=[(number AM–number video)/number video]×100%. For each activity category (all types of lying combined) the mean heart rate was calculated. Due to technical problems, data of one subject had to be discarded from analysis.

Standardised protocol

The standardised protocol consisted of 40 different activities; these activities were performed at the end of each recording session. This protocol included several forms of standing, lying, sitting, standing, walking, and climbing stairs. A total of 237 activities were analysed. Again, predictive value and sensitivity percentages were calculated, but these percentages were not based on 1-second comparisons. Due to the standardised character of this protocol, it was possible to categorise an AM detection as correct or false; if a detection was alternated (partly correct, partly false), it was classified as false.

Results

Spontaneous protocol

Agreement measures

The overall agreement between video analysis and AM output was 88%. Table 5.1 shows the one-second comparisons between video and AM output of all measurements together, as well as sensitivity and predictive value scores. The overall sensitivity and predictive value scores ranged from 58 to 100%. The overall predictive value of lying on the back and of dynamic activities (69 and 64%, respectively), and the sensitivity for dynamic activities (58%) were the lowest.

The agreement per measurement ranged from 59 to 100% (Table 5.2). One subject sat flopped in an easy chair with his legs on a coffee table for an essential part of the measurement time. This posture, occurring in both conditions (1-plac, 1-lor), was detected as lying on the back for most of the time, whereas it was recorded as sitting in the video analysis. The effect of this discrepancy was quantified by recalculation of the results after redefinition of this posture as lying on the back in the video analysis. This redefinition resulted in an increase of the overall agreement from 88 to 99%, and an increase of the predictive value of lying on the back (from 69 to 100%) and the sensitivity for sitting (from 85 to 99%) (Table 5.2).

Number of dynamic periods and transitions

The overall number (the sum of all measurements) of dynamic activities (≥ 5 s) according to video analysis and AM output was 73 and 70, respectively (Table 5.3). The differences within measurements ranged from +2 to −3 (AM output versus video analysis).

Table 5.1. Overall number of corresponding (bold/italic) and non-corresponding (plain) counts (1 count = 1 second) of video and Activity Monitor during both conditions (spontaneous protocol). The last column shows the overall sensitivity of the AM for each video activity category, the bottom row the overall predictive value of each AM activity category. The percentage bottom-right represents the overall agreement.

Video	Activity Monitor (counts)							Sensitivity (%)
	Lying back	Lying side	Lying prone	Standing	Sitting	Dynamic activity	Total	
Lying back	***18155***	1	0	0	0	38	18194	100
Lying side	0	***2574***	0	24	212	15	2825	91
Lying prone	0	0	***0***	0	0	0	0	-
Standing	0	0	0	***1851***	21	41	1913	97
Sitting	8238	48	0	4	***47551***	213	56054	85
Dynamic activity	35	7	0	281	66	***546***	935	58
Total	26428	2630	0	2160	47850	853	79921	
Predictive value (%)	69	98	-	86	99	64		88

Table 5.2. Summary sensitivity (S), predictive value (PV) and agreement (Agr) scores (spontaneous protocol) per measurement, and calculated over all measurements. The last row shows the overall data after video redefinition. Measurement code: e.g., 1-lor = subject 1, lorazepam condition; plac = placebo condition.

Measurement	Agreement scores (%)										
	Lying back		Lying side		Standing		Sitting		Dynamic activity		
	S	PV	S	PV	S	PV	S	PV	S	PV	Agr
1-lor	-	0	-	-	94	94	59	100	78	62	59
2-lor	100	100	88	100	90	66	99	87	61	62	97
3-lor	100	100	-	-	98	85	100	100	65	81	99
1-plac	-	0	-	-	99	89	77	100	46	61	78
2-plac	100	100	97	99	93	73	99	100	49	51	99
3-plac	-	-	-	-	90	98	100	100	77	71	100
Mean (overall)	100	69	91	100	97	86	85	99	58	64	88
Mean (video redefinition)	100	100	91	100	97	86	99	99	58	64	99

The overall number of transitions was 85 according to video analysis and 86 according to AM. There were some differences in number per transition category, but these differences were for the greater part caused by the 'sitting flopped posture'. After redefinition of the video, the differences per transition category became smaller, although the difference in overall number of transition increased.

Table 5.3. The number of dynamic periods (≥ 5 s) and six transition types during the spontaneous protocols, determined by video analysis (V) and Activity Monitor (AM). The data are shown per measurement, and for all measurements together. Measurement codes: e.g. lor-1=subject 1, lorazepam condition; plac=placebo condition. The last row shows the overall data after video redefinition.

Measure-ment	Dynamic periods (number)	Transitions (number)															
		Lying-sitting		Lying-standing		Sitting-lying		Sitting-standing		Standing-lying		Standing-sitting		Total	
	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM
1-lor	5	5	0	2	0	1	0	2	3	2	0	0	3	3	6	10
2-lor	11	13	4	4	1	1	3	2	4	4	2	2	3	3	17	16
3-lor	18	16	0	0	1	1	1	1	5	5	0	0	7	7	14	14
1-plac	16	13	0	3	0	2	0	1	7	5	0	0	7	7	14	18
2-plac	15	13	1	1	2	2	1	2	9	6	2	1	9	8	24	20
3-plac	8	10	0	0	0	0	0	0	5	4	0	0	5	4	10	8
Total	73	70	5	10	4	7	5	8	33	26	4	3	34	32	85	86
Total (video re-definition)	73	70	10	10	7	7	9	8	30	26	4	3	34	32	94	86

Duration of activities

The overall duration of activities (as a percentage of the measurement time) determined by the AM and video analysis is shown in Table 5.4. Lying on the back was overestimated (+10.3%) and sitting was underestimated (–10.2%). Again, redefining 'sitting flopped with the legs on a coffee table' as lying improved the results substantially.

Table 5.4. Duration data (as percentage of the measurement time) of each activity category for the spontaneous protocol, determined by video analysis (V) and Activity Monitor (AM). The data are shown per measurement, and for all measurements together (weighted means). The last row shows the overall results of the analysis after video redefinition. Measurement codes: e.g. lor-1=subject 1, lorazepam condition; plac=placebo condition.

Measurement	Duration (% measurement time)													
	Lying back		Lying side		Lying prone		Standing		Sitting		Dynamic activity		Total	
	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM
1-lor	0.0	39.9	0.0	0.4	0.0	0.0	0.7	0.7	98.8	58.4	0.5	0.6	100	100
2-lor	72.3	72.2	14.1	12.4	0.0	0.0	0.7	0.9	11.7	13.2	1.2	1.2	100	100
3-lor	35.8	35.8	0.0	0.0	0.0	0.0	3.3	3.8	59.2	59.0	1.7	1.4	100	100
1-plac	0.0	20.9	0.0	0.0	0.0	0.0	7.1	7.9	91.3	70.0	1.7	1.2	100	100
2-plac	29.1	29.1	6.8	6.6	0.0	0.0	1.6	2.0	61.2	61.0	1.3	1.3	100	100
3-plac	0.0	0.0	0.0	0.0	0.0	0.0	1.0	1.0	98.4	98.4	0.6	0.6	100	100
Mean (overall)	22.8	33.1	3.5	3.3	0.0	0.0	2.4	2.7	70.1	59.9	1.2	1.1	100	100
Mean (video redefinition)	33.1		3.5		0.0		2.4		59.8		1.2		100	100

Heart rate

The heart rate increased from lying (overall mean 62.8 beats.min^{-1}, range between measurements 59.2-71.8 beats.min^{-1}) to sitting (mean 68.3 beats.min^{-1}, range 65.7-74.6 beats.min^{-1}), and from sitting to standing (mean 78.3 beats.min^{-1}, range: 74.0-80.7 beats.min^{-1}). The mean heart rate during dynamic activities equalled the heart rate during standing (mean 78.3 beats.min^{-1}, range: 73.0-80.6 beats.min^{-1}).

Standardised protocol

Of the standardised protocol, 228 of the 237 activities performed were correctly detected. Table 5.5 shows the data per measurement and the activities that were falsely detected. The overall agreement of the standardised protocol was 96%. Only the predictive value of standing (79%) was somewhat lower.

Table 5.5. Measurement data for the standardised protocol. Values shown are number of activities correctly detected (/) against the number of activities performed (/*). The last column shows the agreement per measurement, the last two rows the overall sensitivity (Sens.) and predictive value (PV) scores. Measurement codes: e.g. lor-1=subject 1, lorazepam condition; plac=placebo condition.*

Measurement	Activities well detected/activities performed						Agreement
	Lying back	Lying side	Lying prone	Standing	Sitting	Dynamic activity	
1-lor	3/3	4/4	2/2	5/5	16/17[c]	7/7	97
2-lor	3/3	3/4[a]	2/2	3/3	19/20[d]	7/7	95
3-lor	3/3	4/4	2/2	6/6	17/18[e]	6/7[f]	95
1-plac	3/3	4/4	2/2	4/4	20/20	7/7	100
2-plac	3/3	4/4	2/2	4/4	19/20[d]	7/7	98
3-plac	3/3	4/4	2/2	5/6[b]	17/18[d]	6/7[f]	93
Sensitivity (%)	100	96	100	95	95	95	96
Predictive value (%)	95	96	100	79	99	100	

False detections: [a] lying on side while reading book - partially as standing; [b] kneeling one leg with support seat - as standing; [c] sitting flopped in easy chair - as lying back; [d] sitting on bed, back against wall, knees pulled up - as standing; [e] sitting on bed, back against wall, knees pulled up - as lying side; [f] strolling in small area - partially as standing

Effect of sensor configuration

The results of the 3-sensor configuration differed only slightly from the 4-sensor results. The overall agreement was equal, in both the spontaneous and standardised protocol. The overall sensitivity and predictive value scores of the 3-sensor configuration did not deviate more than 4% from the 4-sensor scores, except for dynamic activities: the predictive value of dynamic activities decreased from 64 to 50%. This decrease was due to the fact that in the case of a 3- or 2-sensor configuration, the detection of dynamic activities was based on the HPRS signal of one leg sensor. Using a 2-sensor configuration considerably affected the agreement results, especially the sensitivity and predictive value scores of lying and standing: overall percentages of 29, 47, 38, 45, and 27% were found. In the spontaneous protocol, the overall agreement was 88% (Xleg-Ytrunk configuration) and 84%

(Xleg-Xtrunk configuration); in the standardised part the overall agreement was 92% (X-Y) and 83% (X-X). Analysis of the effect of sensor configuration on the number of transitions and the duration of activities showed the same pattern.

Discussion

In the spontaneous and the standardised protocols we found an overall agreement of 88% and 96%, respectively. These results are satisfying and support the previously found validity results of the AM (see chapter 4).[3]
However, some results of our study need further attention. An important source of discrepancy between video and AM results appeared to be the effect of the detection of sitting flopped in an easy chair with legs on a coffee table as lying, as occurred in one subject. The analysis criteria of the video recording are based on the presence and position of supporting surfaces, whereas the AM output is – especially in postures – based on the position of upper leg and trunk. If a sitting person leans far backwards, the trunk signals approach the predetermined range of LP/angular values for lying on the back. The question, then, is: Is the detection of the described posture as lying on the back really an error? The answer to this question depends on the formulated research questions and is a matter of definition. Therefore, we are of the opinion that use of the agreement results after redefinition of the video analysis (data presented in the results section and Tables 5.2-5.4) are valuable. We can change the posture-bound ranges of sitting backwards; this alteration, however, would possibly result in a lower sensitivity for lying. We will study this point in the future. One of the further options is the implementation of a category 'uncertain': if a person maintains a posture in the transitional area between the two postures, then no posture will be detected. Furthermore, optimisation of the sensor fixation – in almost all subjects the trunk sensor on the sternum was slanted backwards slightly – may improve the functioning of the AM.

After redefinition of the video analysis, almost all results were satisfying. The sensitivity and predictive value of dynamic activities, however, stayed relatively poor. The detection of an activity as static or dynamic strongly depends on an adjustable threshold. A well-set threshold is characterised by a minimum number of standing-dynamic activity misdetections, but also by an equal number of misdetections to both sides. Standing was detected as a dynamic activity in 41 s, and a dynamic activity as standing in 281 s, which suggests that the set-point of the threshold was too high. Nevertheless, the main results – the number and duration of dynamic periods (see Tables 5.3 and 5.4) – were determined well. This finding was because of a relatively small, but in absolute numbers considerable, number of

samples in which sitting was detected as dynamic (e.g., when legs were moved corresponding to the rhythm of music). As has been concluded previously (see chapter 4),[3] the setting of the threshold will need further attention. On the other hand, when – in the future – several types of dynamic activities can be distinguished, more criteria will be involved and a non-hierarchical detection will be used in the selection of an activity as dynamic and deciding which type of dynamic activity it is. The problem will then be less important.

In the standardised protocol, we found an overall agreement of 96%. Most of the (few) errors in the standardised protocol were logical and explicable. With only a relatively simple set of four sensors, misdetection of some activities cannot be avoided. The results of this study do not, however, necessitate an immediate adaptation of the configuration or predetermined ranges. The analysis of the different configurations shows that the results hardly change when only one leg sensor is used (3-sensor configuration). This finding suggests that discarding one sensor is possible; however, information on the quality of walking (e.g., symmetry, co-ordination) cannot then be obtained. Discarding another sensor (2-sensor configurations) decreases some of the agreement scores considerably. It is predominantly the percentages for lying on the side and standing that are influenced. Here the choice also depends on the formulated research questions and the validity required.

The HPRS signal changes with the variability of the measured signal; the level of the HPRS signal is related to the intensity of motion (see Figure 5.1), which we call motility. Most of the currently available actometers are based on the principle of variability of the accelerometer signal. These actometers are generally attached to the human body at the wrist, the anklé or the waist.[8,9,10,14] The simultaneous use at different locations of three or four accelerometers in our instrument may give more precise and reliable information regarding motility, as was also suggested by Patterson et al.[10] Motility variables can be calculated for all sensors separately, or can be combined to indicate total bodily motility. The mean motility can be calculated over all activities, per activity category, per time period, and per posture. In the per posture case, no dynamic activity category exists; for example, walking then is standing with much motility. In our study, we calculated the motility variables, but we did not validate them. Validation of these variables depends on the characteristics in which one is interested. We believe, however, that motility variables are relevant extensions of the output of the AM. They are also interesting to study in relation to ambulatory recorded physiological processes. The portable recorder allows simultaneous recording of accelerometer signals and

psychophysiological signals (e.g. ECG, heart rate, blood pressure), which enlarges its applicability and surplus value. Reducing the size and weight of the recorder, which is planned for the near future, will also contribute to the usability of the AM.

In this validation study, we observed clear posture-related heart rate changes, on the basis of the classification of the AM. The heart rate changes were similar to those obtained in standardised situations.[2,6,14,16] The heart rate during dynamic activities was not different from the heart rate during standing. This finding could be explained by the short duration of almost all dynamic activities. The results underline the validity of our approach to assess the activity-related variability aspects of heart rate by means of accelerometry. Therefore, the possibilities and potential advantages of ambulatory activity monitoring justify further exploration and application of the AM in psychophysiological research.

Conclusion

In this validation study, the 4-sensor AM appeared to be a valid instrument for quantifying aspects of normal daily activities. The agreement scores between video and AM analysis were high. Duration of activities and number of transitions and dynamic periods were determined well. Leaving out one leg sensor had hardly any influence on the results, indicating that it is still possible to obtain reliable activity indices with a 3-sensor system. Ambulatory psychophysiological and activity monitoring during normal daily life in a person's own environment may stimulate further theoretical and clinical developments in biomedical and behavioural research.

Acknowledgements

We thank Frans van den Berg and Hugo G. van Steenis for their contribution to this study.

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Ambulatory accelerometry to quantify motor behaviour in patients after failed back surgery: a validation study

Summary

In the treatment of patients with pain, measures related to (pain) behaviour are of major importance. Ambulatory activity monitoring can be used to obtain insight into actual behaviour. This study was designed to validate the Activity Monitor (AM), an instrument based on long-term ambulatory monitoring of accelerometer signals, to assess several physical activities during normal daily life.

Ten failed back surgery (FBS) patients performed a number of functional activities in and around their own houses. During the measurements, continuous ambulatory registrations of accelerometer signals were made, based on four body-mounted accelerometers (one on each upper leg, two on the trunk). Video recordings made simultaneously with the measurements were used as a reference. The continuous output of the AM (postures, transitions, dynamic activities) was compared with visual analysis of the videotapes.

The overall results showed an agreement between AM output and video analysis of 87% (inter-subject range 83-88%). The maximal error in the determination of the mean duration of activities was 0.3%. The overall number of dynamic periods was determined well (AM 359, video 368), while the number of transitions was slightly overestimated (AM 228, video 205). The results when using the 3-sensor version of the AM were somewhat less accurate (overall agreement decreases from 87 to 82%).

The AM appeared to be a valid instrument to quantify aspects of behaviour of FBS patients, such as duration of activities and number of transitions. This new technique of ambulatory measurement of mobility activities seems to be a relevant and promising extension of the techniques currently used in the evaluation of pain treatment.

Introduction

Pain behaviour has been defined as 'anything a person does or does not do that you and I would interpret as likely to be due to tissue damage'.[e.g. 18] In the treatment of pain, assessment of actual behaviour of patients is considered important;[e.g. 9,10] pain behaviour is one of the conceptual levels of pain, together with nociception, pain and suffering.[18,19,25] Although some relationship may exist between these levels, direct connections will certainly not exist. For example, Fordyce et al.[11] found no correlation between severity of pain and pain behaviour.

In the measurement of pain behaviour several components can be distinguished. These components are often related to the quantity (which activity, when, how often and how long performed) and quality (how performed) of movements and postures, e.g. 'functional limitation or restricted movement because of pain';[10] 'distorted ambulation or posture' and 'avoidance of activity';[29] 'daily mobility avoidance', 'activities avoidance' and 'daily exercise avoidance';[22] and 'distorted mobility and posture' and 'fatigue'.[31] It can be concluded that the amount ('quantity') of postures and movements can be regarded as a valid operationalisation of these constructs.

In the past, instruments were developed and used to obtain insight in the behaviour-related effects of treatment. In most cases these instruments were based on questionnaires, self-reports or diaries.[e.g. 11,12,17] The most important drawback of these techniques is that they do not measure actual behaviour, but measure retrospectively the patient's behaviour according to the patient. This may lead to discrepancies between actual behaviour and reported behaviour.[e.g. 11,24]

Some instruments are able to measure actual behaviour more objectively; e.g. mechanical and accelerometer based actometers are used to measure physical activity. Their output provides general information on the level of activity and energy turnover, but no detailed information on the quantity (when performed, how often, how long) of specific activities.[16,20,21,27] Ambulatory activity monitor systems provide more specific data on postures and activities,[1,7,8,24,26,28] but generally lack information on validity, do not measure a complete and varied set of postures and movements, or are rather complex systems with several types of sensors.

Due to recent advances in technology, the possibilities for ambulatory (activity) monitoring have increased enormously. For example, the Activity Monitor (AM) is a newly developed instrument based on ambulatory monitoring of four accelerometer signals, and aimed at the assessment of the quantity and quality of several mobility-related activities during normal daily life.[6] These activities include the static activities (or postures) of standing, sitting and different modes of lying (prone, on the back, on the side), the transitions between these postures, and the dynamic activities of walking, climbing stairs, cycling and driving a wheelchair.

Long-term (up to several days) and continuous registrations are possible with this system. The validity of the AM has been investigated in both healthy subjects and amputees (see chapter 4),[5] and in healthy subjects in a pharmacological study (see chapter 5);[4] in these studies the setting of the measurements was semi-artificial and non-pain patients were involved.
In the present study, the validity of the AM was investigated when applied in failed back surgery (FBS) patients, performing functional activities in their own environment. This setting was chosen, because the AM will be used in a prospective evaluation study on the effects of some treatment methods of FBS patients. In addition, the source of any errors was investigated, as well as their effect on validity and any adaptations that may be required in the software. Due to power supply, data storage, and usability arguments, a decrease in the number of sensors may be advisable or necessary in future research. Therefore, we also examined the effect on validity of omitting one of the four sensors.

Methods

Subjects
Ten FBS patients participated in this study. All were selected from a file of FBS patients at the Pain Expertise Centre (Rotterdam, the Netherlands). The inclusion criteria were: having undergone back surgery one or more times without success; almost continuous pain (local and/or radiating); eligible for symptomatic treatment only; and assumed to be able to perform the greater part of the protocol.

Table 6.1. Subject characteristics and their scores on the Tampa Scale for Kinesiophobia (TSK) and the Roland Disability Questionnaire (RDQ).

Subject	Gender	Age (yrs)	Height (m)	Mass (kg)	TSK	RDQ
1	Male	58	1.8	78	46	19
2	Female	60	1.65	67	41	9
3	Female	42	1.72	76	35	13
4	Female	56	1.68	84	31	10
5	Female	45	1.6	60	37	13
6	Female	55	1.65	65	39	15
7	Female	47	1.64	55	37	10
8	Male	33	1.81	91	40	13

Measurements from eight of the ten subjects could be used for analysis; the characteristics of these 8 subjects are given in Table 6.1. For descriptive purposes, the (Dutch version) of the Tampa Scale for Kinesiophobia (TSK)[31] was used to assess fear of movement (scale range 17-68), and the (Dutch version) of the Roland Disability Questionnaire (RDQ)[3,13,23] to assess disability (scale range 0-24). In general, the TSK and RDQ scores suggest that the subjects in this study were considerably afraid to move and disabled. The subjects were invited to participate in the study by their specialist; before the measurements they signed an informed consent form.

Protocol

Based upon existing questionnaires about activities of daily life, an extensive list of functional activities was compiled (Table 6.2). Before the measurements took place, the patients were asked which activities in the list they normally performed; only these activities were included and performed during the measurement (mean number 32, range 29-34). The patients were asked to perform the activities in their own way and at their own pace. All measurements were performed in the patient's own environment in and around his or her house. The measurements were done and analysed continuously; the mean duration of the analysed measurement time was 47 minutes (range 28-82 minutes).

Instruments

Activity Monitor

Four IC-3031 uni-axial piezo-resistive accelerometers (1.5×2×0.5 cm) were used. These sensors measure accelerations related to changes in velocity, as well as the gravitational acceleration. The acceleration actually measured depends on (a) the direction of both types of accelerations with regard to the sensitive axis of the accelerometer, and (b) their magnitude. On each leg, one sensor was attached to the skin at the front of the thigh. The other two sensors were attached to the skin of the sternum, perpendicular to one another. The sensors were attached so that, with the subject standing, their axes were as close as possible to the vertical or horizontal plane; a maximal deviation of 15 degrees was allowed.

The accelerometers were connected to a portable Vitaport1™ data recorder; the signals were digitally stored on a memory card, each with a sampling frequency of 16 Hz. Analysis took place after the measurements by means of the Signal Processing and Inferencing Language (S.P.I.L.™).[14]

Postures are distinguished from each other by using the low-pass filtered (0.5 Hz) and to angles converted derivatives of each signal (LP/angular signal). Theoretically,

four LP/angular signals are sufficient to distinguish different postures from each other. The detection of transitions is derived from changes in posture.
A second type of signal that is derived from each measured signal is the result of a high-pass filtering (0.5 Hz), rectifying and smoothing procedure (HPRS signal). Dynamic activities are discriminated from static activities by applying a threshold to the HPRS signal of the upper leg accelerometers. The more 'energetic' an activity is, the more variable the accelerometer signals, and the higher the value of the HPRS signal (see chapter 3).[6,30] If an activity is detected as dynamic, it overrules the posture detection. The output of the AM – the continuous selection of an activity – has a time resolution of 1 second. Figure 6.1 shows an example of the four accelerosignals during subsequent activities, and the output of the AM.

Table 6.2. List of functional activities included in the protocol.

List of activities	
Mobility indoor	*Sleeping*
Walk to letterbox	Put on nightclothes
Take newspaper from ground	Sit on edge of bed
Walk to living room	Walk to bathroom
Read newspaper	Brush teeth
Visit toilet	Lie comfortably
Housekeeping activities	Lie in different postures
Use dustpan and brush	Get dressed
Take objects from the ground	Make up the bed
Vacuum the living room	*Mobility outdoors*
Move plant/table	Put household refuse outside
Put laundry in washing machine	Walk outdoors
Hang out the wash	Cycle outdoors
Use kitchen steps to place book in cupboard	Drive a car
Clean windows	*Leisure time*
Walk to kitchen	Watch television
Peel and cook potatoes	Perform own hobby (max. 3)
Take herbs from kitchen cabinet	Write a letter
Set the table	
Wash dishes, pan, mugs	

The way in which the dynamic activities can best be distinguished from each other is still under investigation. This study is, therefore, restricted to the categories 'static' and 'dynamic', the transitions and (within the static category) the five postures (lying prone, on the back, on the side, sitting and standing). By leaving the left leg sensor out of the analysis, the effect of a more limited configuration can be investigated.

Reference method

During the performance of the protocol video recordings were made. After the measurements the video recordings were analysed with a time resolution of one second, all by the same person. Though the output categories of the video analysis were the same as the output categories of the AM, the general rule of the formulated guidelines for video analysis was different from the one of the AM. For example, the video analysis of lying, sitting and standing is based on the presence and position of supporting surfaces, while the posture detection of the AM is based on the angular position of legs and trunk.

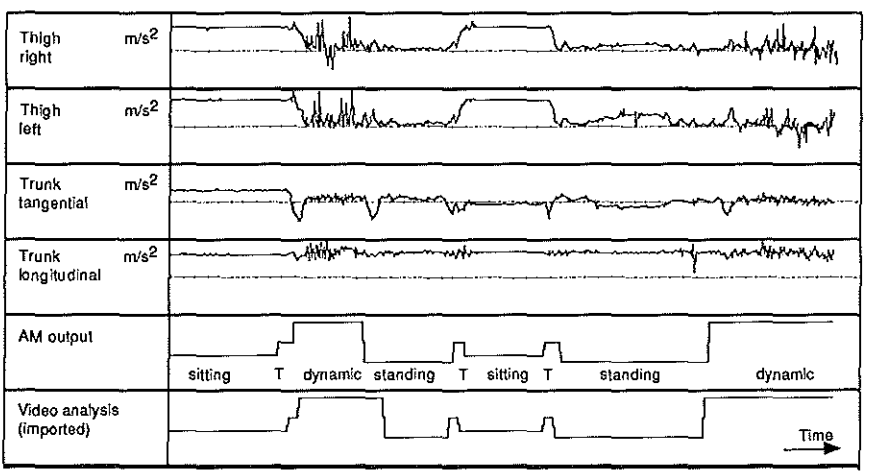

Figure 6.1. Example of the four raw accelerometer signals (two leg sensors, two on the trunk) and the output of the Activity Monitor (AM). This 2-minute part shows a sequence of activities as indicated by the level of the AM output (T = transition); each activity category has its unique level (the words are added by way of illustration). The bottom signal is the imported video analysis; in this way the output of the AM could be compared automatically with the video analysis.

The inter-rater agreement of the video analysis was investigated in a previous study, and was 99.7%. The video analysis served as reference method or gold standard. The video recordings were synchronised with the accelerometer recordings by means of a photo flash. If video recordings were inadequate, these parts were not used in the analysis. The analysed video recordings were transferred to a signal in the AM file (Figure 6.1): all calculations and comparisons were done automatically by means of S.P.I.L.™ software.

Data analysis

The continuous output of the AM was compared with the synchronised, continuous output of the video analysis, with a time resolution of one second. Because the video analysis is a gold standard, the following agreement scores -- as validity measures of the AM – were calculated:

(1) *Agreement*: the percentage of agreement between all samples of video and AM. Agreement is calculated according to: agreement=(number of identical samples of video and AM/total number of samples)×100%.

(2) *Sensitivity*: the degree to which each video activity category (representing the activities actually performed) is detected correctly by the AM; the sensitivity percentage is calculated according to: the sensitivity for video activity category A=(number of identical samples of video and AM for video activity category A/total number of samples for video activity category A)×100%.

(3) *Predictive value*: the degree to which each AM activity category agrees with the video activity category (representing the activities actually performed); this is calculated according to: the predictive value of AM activity category A=(number of identical samples of video and AM for AM activity category A/total number of samples for AM activity category A)×100%.

Sensitivity and predictive percentages are not calculated if the number of AM or video samples of a specific activity category is less than 20 (1 sample equals 1 second). The number of dynamic periods (lasting 5 seconds or longer) and the number of transitions per transition category were calculated. The duration was calculated per activity category. The Wilcoxon matched-pairs signed-ranks test was used to show systematic differences in number and duration results between video and AM. This test was done with SPSS 5.0 for MS Windows; a P-value of < 0.05 was considered to indicate a significant effect.

Results

Eight of the 10 measurements could be used for analysis. In one subject the combined trunk sensors came loose from the skin. After measurement in another subject, one trunk sensor appeared to have been defect during the measurement.

Agreement measures

The overall agreement between video analysis and AM output was 87%. The overall sensitivity and predictive value ranged from 84 to 96%, with the lowest percentages for standing and dynamic activities (Table 6.3).

Table 6.3. Number of corresponding (bold/italic) and non-corresponding (plain) counts (1 count = 1 s) of video (rows) and AM (columns) added for all measurements. The last column shows the overall sensitivity of the AM for each video activity category, the bottom row the overall predictive value of each AM activity category

Video	Activity Monitor (counts)							Sensitivity (%)
	Lying back	Lying side	Lying prone	Standing	Sitting	Dynamic activity	Total	
Lying back	***216***	2	0	0	0	33	251	86
Lying side	0	***359***	0	14	0	24	397	90
Lying prone	0	0	***0***	0	0	0	0	-
Standing	0	7	7	***6626***	186	1044	7870	84
Sitting	0	0	0	20	***4101***	254	4375	94
Dynamic activity	10	16	0	1271	120	***8290***	9707	85
Total	226	384	7	7931	4407	9645	22600	
Predictive value (%)	96	93	-	84	93	86		87

Only in a few cases – and if so for a very short time – static activities were mutually misinterpreted (Table 6.3). Lying on the side was detected as standing (14 seconds) while lying on the side with the trunk strongly erected; standing was detected as lying on the side (7 seconds) due to standing with the trunk bent strongly sidewards. Standing was detected as lying prone (7 seconds) whilst a subject was making the bed: his trunk was bent forward and one leg was extended backwards. Sitting was

sometimes detected as standing (20 seconds); sitting on the saddle of a bike with one foot on the ground was sitting according to the video analysis, and standing according to AM.
More often (186 seconds), standing was detected as sitting. In all cases this was due to squat positions, during which the seat was not supported by the feet or lower legs: standing according to the video analysis, sitting according to AM.
The variability between subjects of the agreement was small: the range was 83 to 88% (Table 6.4). The other percentages in Table 6.4 show no extremes, except the sensitivity for lying on the side in subject 5. In this measurement a short and dynamic period of lying on the side (26 seconds) was detected by the AM as dynamic (12 seconds) and standing (14 seconds), resulting in a sensitivity percentage of 0.

Table 6.4. Percentages per measurement, representing the sensitivity (S) and predictive value (PV). '–': the activity is not performed or detected, or is less than 20 seconds. In the last column the agreement per measurement is shown, in the last row but one the weighted overall means; the last row represents the overall data of the three-sensor configuration.

Measurement	Agreement scores (%)										
	Lying back		Lying side		Standing		Sitting		Dynamic activity		
	S	PV	S	PV	S	PV	S	PV	S	PV	Agr
1	-	-	95	81	82	85	96	96	86	83	88
2	89	100	97	96	86	82	96	97	83	86	87
3	-	-	97	96	93	73	95	94	77	93	86
4	-	95	-	-	89	83	98	99	75	82	87
5	71	100	0	-	87	83	79	88	83	80	83
6	96	89	86	96	76	89	95	96	92	83	87
7	100	96	95	95	74	83	96	84	92	90	87
8	85	94	100	94	85	87	96	89	87	88	88
Mean	86	96	90	93	84	84	94	93	85	86	87
Mean (3 sensors)	84	96	93	94	80	76	92	94	78	81	82

Number of dynamic periods and transitions

The overall number (total of all measurements) of dynamic activities (≥ 5 seconds) according to video and AM was 368 and 359, respectively. The overall number of transitions (Table 6.5) was 205 according to video and 228 according to AM (P=0.063). The differences within measurements were generally small (Table 6.5). The only statistically significant difference found was the overestimation by AM of the sit-to-stand (105 versus 92, P=0.043) and stand-to-sit transition (105 versus 94, P=0.046). The differences found were almost entirely related to the squat movement and position.

Table 6.5. The number of dynamic periods (≥ 5 seconds) and six transition types, determined by Activity Monitor (AM) and video analysis (V). The data are shown per measurement (Meas.), and added for all measurements. In the last row the overall data of the three-sensor configuration are shown.

Meas.	Dynamic periods (number)		Transitions (number)													
			Lying-sitting		Lying-standing		Sitting-lying		Sitting-standing		Standing-lying		Standing-sitting		Total	
	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM
1	46	46	1	1	0	0	0	0	9	11	1	1	10	12	21	25
2	31	29	0	0	0	0	0	0	10	11	1	1	9	11	20	23
3	36	44	1	0	1	1	1	0	11	11	1	1	12	12	27	25
4	34	32	1	1	0	0	1	1	9	9	0	0	9	9	20	20
5	54	54	0	0	2	1	1	1	11	16	1	1	13	17	28	36
6	48	40	1	1	0	0	0	0	11	11	1	1	9	8	22	21
7	40	41	0	0	1	1	0	0	10	13	1	0	10	12	22	26
8	79	73	1	1	0	2	0	0	21	23	1	2	22	24	45	52
Mean	368	359	5	4	4	5	3	2	92	105	7	7	94	105	205	228
Mean 3 sensors		382		5		4		3		100		8		101		221

Duration of activities

The overall duration of activities (as a percentage of the measurement time) determined by the AM differed from video analysis by –0.3% for dynamic activities to +0.3% for standing (Table 6.6). None of the differences were statistically significant. In general, the differences within measurements were small.

Effect of sensor configuration

The effect of sensor configuration is shown in the last row of Tables 6.4 to 6.6. When using the 3-sensor version the overall agreement decreased from 87 to 82%, mainly due to the standing and dynamic activities. Analysis of the effect of the 3-sensor configuration on the number of transitions and dynamic activities and the duration of activities showed, in general, slightly less accurate results.

Table 6.6. Duration (as percentage of the measurement time) of each activity category, determined by video analysis (V) and Activity Monitor (AM). The data are shown per measurement, and for all measurement together (weighted means). In the last row the overall results of the three-sensor configuration are shown.

Measurement	Duration (% measurement time)												
	Lying back		Lying side		Lying prone		Standing		Sitting		Dynamic activity		Total
	V	AM	V	AM	V	AM	V	AM	V	AM	V	AM	V/AM
1	0.3	0.3	1.2	1.4	0.0	0.0	29.6	28.5	34.9	34.7	34.0	35.0	100
2	3.4	3.0	4.3	4.3	0.0	0.0	34.2	35.9	18.2	18.0	40.0	38.8	100
3	0.5	0.2	4.5	4.6	0.0	0.0	29.1	37.3	16.2	16.5	49.6	41.4	100
4	1.0	1.1	0.0	0.0	0.0	0.0	42.8	46.0	24.6	24.1	31.6	28.8	100
5	1.5	1.1	0.8	0.0	0.0	0.0	37.0	38.7	19.2	17.2	41.5	43.1	100
6	1.0	1.1	1.2	1.1	0.0	0.0	34.6	29.5	12.9	12.8	50.2	55.6	100
7	1.7	1.7	0.8	0.8	0.0	0.0	28.3	25.2	10.9	12.6	58.4	59.7	100
8	0.7	0.6	1.9	2.0	0.0	0.1	40.9	39.8	16.7	18.0	40.0	39.5	100
Mean	1.1	1.0	1.8	1.7	0.0	0.0	34.8	35.1	19.3	19.5	43.0	42.7	100
Mean 3 sensors		0.9		1.7		0.0		36.7		19.0		41.6	100

Discussion

In this study, an overall agreement of 87% was found between AM output and video analysis. This result is almost identical to the results of previous validation studies, which reported overall agreement of 90% (3-sensor configuration) and 88% (4-sensor configuration) (see chapters 4 and 5).[4,5] The results of the present study support the assumption that the validity of the AM is general; the results so far do not indicate a strong effect of the patient group on the data.

Eight out of ten (80%) measurements could be used for analysis. Because this percentage seems too low, some comments should be made. In one subject, one trunk sensor appeared to have been defect during the measurement. In four years of measuring with the AM, this is only the second time that a defect has occurred; one of the advantages of the accelerometers is that they are robust. In another subject, the trunk sensors came partially loose from the skin, mainly due to perspiration and/or chest hair. Fixation of the sensors on the skin still needs to be improved; especially during long-term measurements greater demands are placed on it.

It is not easy to compare the results of this study with the validity data of other activity monitors. Activity monitors described in the literature have other, or less, possibilities.[1,2,7] Furthermore, validation techniques were not applied, were different, or were not clearly described.[8,26,28] Thus, the assessment of the usability of the AM can not be achieved by comparison with other instruments.

In every validation study new aspects of mobility become apparent, which can be used for the optimisation of the technique. In the present study particularly the squat position and squatting movement gave rise to discussion. Squatting without support of the seat is sitting according to the AM, and is standing according to the video analysis. Due to the short periods of this squat position, the effect on the sensitivity and predictive value is relatively small. If, in the video analysis, squatting is redefined as sitting, then the overall agreement increases 1% (from 87 to 88%), while the sensitivity for standing increases from 84 to 86%, and the predictive value of sitting increases from 93 to 97%. The effect of this redefinition on the number of transitions is more dramatic. The existing differences almost disappear: total number of transitions AM versus video: 228 versus 229 (was 228 versus 205); total number of sit-to-stand transitions AM versus video: 105 versus 104 (was 105 versus 92); total number of stand-to-sit: transitions AM versus video: 105 versus 105 (was 105 versus 94).

From the first validation study onwards, we have used the same video analysis criteria. These criteria were not based on the expected output of the AM, but on a different frame of reference: the presence and position of supporting surfaces, while the posture part of the AM is based on the position of the sensors with regard to the

gravitational acceleration. This position of the sensors is largely determined by the position of the body segments to which they are attached. Therefore, it can be expected that the squat position would be misinterpreted. However, the squat position has many similarities with sitting, and the AM analysis may often be regarded adequate; this will depend on the objectives of the study. We plan to study the possibilities to distinguish squatting from the standing-to-sitting and sitting-to-standing transition. The angles of upper legs and trunk in the squatting position will not be unique, but the movement pattern during the transition may be.

If the dynamic activities are excluded from the analysis – to obtain insight in the quality of the static activity detection – the overall agreement increases to 98%, with the overall sensitivity and predictive value ranging from 96 to 100%. It is important that the misdetections which occur are explicable. One type of misdetection warrants further explanation: lying on the side is sometimes detected as standing and vice versa. Although these static activities appear to be different, the signals of the accelerometers are not dissimilar: only the radial trunk sensor determines the difference between these postures. Therefore, mostly in short-lasting and 'dynamic' postures, a misdetection may occur.

The detection of an activity as static or dynamic strongly depends on an adjustable threshold. A well-set threshold is characterised by a minimum number of standing-dynamic activity misdetections, and by an equal number of misdetections on both sides. Standing is detected as a dynamic activity in 1044 seconds (13.3%), and a dynamic activity as standing in 1271 seconds (16.0%). The threshold seems to be set well. However, we are implementing algorithms in the software to distinguish dynamic activities from each other. In that program, more techniques and criteria will be involved (e.g. Fast Time Frequency Transform)[14] and a non-hierarchical detection will be used in the selection of an activity as dynamic and in deciding which type of dynamic activity it is.

The inter-subject variability of the agreement measures is small, as is the variability in sensitivity and predictive value percentages between activities. Again, this supports the general validity of the instrument. If in some subjects the validity showed to be low, or if some activities had clearly lower agreement percentages, the statements on the usability of the AM would be much more complex.

The number of dynamic activities, the number of transitions and the duration of activities is determined with a high level of accuracy. Only the described squat position leads to overestimation of the sitting-to-standing and standing-to-sitting transitions.

Omitting one leg sensor from the analysis results in a decrease in the overall agreement from 87 to 82%. Most results are slightly lower compared with the 4-

sensor version. It depends on the required validity whether such a configuration is a feasible option or not.

Apart from monitoring of activities, other signals or events can be simultaneously measured, e.g. ECG or heart rate, EMG, and other accelerometer signals. It is also possible to combine activity monitoring with the registration of perceived pain, as suggested previously.[11] Furthermore, other measures can be derived from the accelerosignals. First, until now the results are summarised to global categories. In the analysis program, however, many subcategories can be distinguished. For example, standing can be divided in standing upright, standing with the trunk bent slightly forward, and standing with the trunk bent strongly forward. In some studies, there may be special interest in specific subcategories, e.g. standing forward with the trunk bent strongly forwards. It is possible to obtain this kind of specific information. Second, accelerosignals contain much information on the way activities are performed (e.g. velocity, co-ordination, symmetry), which we call the quality of activities. Pain behaviour may be characterised by changes in quantity of activities (activity level or activity pattern), but also by changes in quality (movement pattern). For example, Keefe and Hill[15] found significant differences between walking patterns of patients and controls. We can already measure some relevant quality variables with the AM (step duration, symmetry, stability, phases in transitions), but we will continue efforts to quantify more quality aspects. Third, a signal can be derived which changes with the variability of the measured signal; the level of this signal is related to the intensity of motion, which we call motility. Traditional actometers are based on the principle of variability of the accelerometer signal. They are generally attached to the human body at the wrist, the ankle or the waist.[16,20,21] The simultaneous use at different locations of three or four accelerometers in our instrument may give more precise and reliable information regarding motility, as is also suggested by others.[21] Reducing both the size and weight of the recorder, which is planned for the near future, will also contribute to the usability of the AM.

The AM will be used in a prospective, randomised clinical trial in which different treatments of patients after FBS will be evaluated. As stated in the introduction, the output of the AM will present an objective picture of (changes in) the activities a patient actually performs, and will therefore be a source of behavioural measures. From the present study it can be concluded that the AM can validly measure the activities of FBS patients, performed in their own environment.

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Validity of measurements obtained with the extended version of the Activity Monitor; a new analysis program applied on existing data

Summary

After the development and validation of the first version of the Activity Monitor analysis program, an extended version was developed as a sequel to that first program. This extended version is based upon a non-hierarchical decision scheme and three input features, and allows also the detection of several dynamic activities. In this chapter, the validity of measurements with this extended version is described. The signals from three previously performed and reported validity studies were used (see chapter 4, 5, and 6). In these studies accelerations were measured and videotape recordings were made (reference method). Validity was assessed by calculating agreement scores between AM output and video output, and by comparing the number of walking periods, and the duration of activities determined by both methods.

The overall agreement between AM output and videotape analysis for the three studies was 89%, 93%, and 81%, respectively. In the studies with considerable dynamic periods (study 1 and 3) walking had agreement scores per measurement ranging from 67 to 95%. In climbing stairs, especially the sensitivity scores were lower (mean 24% and 76%, respectively; range 0-87%), generally due to the misdetection as walking. Generally, the duration of walking was slightly underestimated (–0.8% in both studies). The number of walking periods was well determined (total number 169 versus 170, and 255 versus 240, respectively). The agreement score per measurement for cycling ranged from 51-100%.

It is concluded that the extended version of the AM is a valuable extension of the first AM version. The detection of static activities remains stable, and walking and cycling were well determined. The ability to distinguish climbing stairs was less powerful, although some gain may be reached by a more valid determination of angular attachment deviations. The predictive value of driving a wheelchair will be low, because other sitting activities may also lead to cyclic movements of the trunk. Further study has to examine the sensitivity of the AM in detecting driving a wheelchair.

Introduction

To measure the activities a person performs during normal daily life an Activity Monitor (AM) has been developed. The first version of this instrument was able to distinguish several static activities (standing, sitting, and different forms of lying) and dynamic activities (e.g. walking, climbing stairs, cycling) as one group. The validity of this version was studied thoroughly (see chapters 4, 5, and 6).[2,3,4] However, the way dynamic activities were distinguished from static activities was based on one Boolean algorithm, the activity detection procedure was hierarchical, and several dynamic activities could not be distinguished from each other (see chapter 3).

As a sequel to the first version of the AM, an extended version of the AM has been developed (see chapter 3). This extended version does not differ from the first version in number of sensors and their location; only the analysis program has been changed. The analysis of the videotape recording was – at least partly – tuned to a future validation of an extended version. Therefore, the data obtained during the three validity studies described in this thesis could also be used to examine the validity of measurements with the extended version of the AM, which was the aim of the data analysis described in this chapter. Only small differences exist between the first version and the extended version of the AM with respect to the detection of static activities. Because the extended version differs from the first version mainly in the detection of dynamic activities, this chapter will focus on that part of the AM output.

Methods

The design and methods of the validity studies are described in detail in chapters 4, 5, and 6, while chapter 3 contains a technical description of the extended version of the AM. Therefore, this chapter will only present a summary of these items.

Protocols

Validity study 1

Four persons with a trans-tibial amputation and four persons without an amputation were involved. They performed a number of normal daily activities in and around an occupational therapy department, in which a complete (representative) apartment had been installed. The non-amputation subjects performed the protocol twice, with a one-week interval.

Validity study 2

The second validity study was performed within the context of a psychophysiological study. Three subjects participated in each of two 4-hour sessions. Each session consisted of two protocols. In the spontaneous protocol the subject had to stay in a living room. The subject was free to choose his own activities. At the end of the spontaneous protocol, he/she performed 40 forms of standing, sitting, lying, walking and climbing stairs (standardised protocol). Walking included walking slow, normal, and fast; climbing stairs included normal climbing upstairs and downstairs.

Validity study 3

In this study, 10 failed back surgery patients (the data of 8 of them were analysed) performed a number of functional activities in and around their own house.

Instruments

Extended version Activity Monitor

Four IC-3031 uni-axial piezo-resistive accelerometers were used. The sensors were attached to the skin at the front of each thigh, and the other sensors were attached to the skin of the sternum, perpendicular to one another. The accelerometers were connected to a portable Vitaport 1™ data recorder. Analysis took place after the measurements by means of the Signal Processing and Inferencing Language (S.P.I.L.).[5]

From each measured signal, a low-pass (LP)/angular, motility, and frequency feature signal was derived. Because the motility and frequency signals of both legs were combined, 10 feature signals were derived. For each of the 23 activity subcategories in the analysis program and for each feature signal, a minimum and maximum value is pre-set in an Activity Detection Knowledge Base (ADKB). So for each subcategory the distance between the current feature signal value and the pre-set range is calculated. The distance of each feature signal value is added for each activity subcategory. The subcategory with the lowest added distance will be selected. For the analysis described in this paper, 23 activity subcategories were reduced to 9 AM output categories. All lying subcategories were reduced to the AM output category lying; all other subcategories related to static activities were reduced to the output categories sitting or standing. All walking-related subcategories were summarised in the output category walking; climbing upstairs and downstairs in the AM output category climbing stairs. In addition, the activity categories sitting with cyclic movements of the leg (representative for cycling), sitting with cyclic movements of the trunk (representative for driving a wheelchair), running, and

general movement (representative for non-cyclic movements) were used in the analysis. A duration threshold of 5 seconds was applied.

Reference method
During the performance of the protocol video recordings were made. After the measurements the video recordings were analysed with a time resolution of one second. As in the AM analysis, there were more subcategories distinguished than output categories used. For example, standing was divided into standing quiet (which was regarded as standard), standing with leg movements, and standing with trunk movements; walking was divided into normal walking (standard), shuffling, walking with a bike, etc. The standard form as well as the non-standard forms were all joined into one output category. Thus, in the video analysis, 9 output categories were distinguished: lying, sitting, standing, walking, climbing stairs, cycling, driving a wheelchair, running, and transitions.

Data analysis
The output of the AM was compared with the synchronised output of the video analysis, with a time resolution of one second. The following agreement scores – as validity measures of the AM – were calculated: agreement, sensitivity, and predictive value (for calculations see chapters 4-6). The video output categories 'cycling', 'driving a wheelchair', and 'transitions' were compared to the AM categories 'sitting with cyclic movements of the legs', 'sitting with cyclic movements of the trunk', and 'general movement', respectively. Furthermore, the number of walking periods determined by videotape analysis and AM were compared.

Results

The overall agreement in the three studies was 89%, 93%, and 81%, respectively. Table 7.1a-c shows the overall data distribution for each study; Table 7.2 provides data on overall sensitivity and predictive value percentages, and their range within each study. The instances that a static activity was falsely detected as another static activity, generally equalled the instances that the first AM version falsely detected a static activity. The largest difference concerned the detection of sitting as lying on the back in the second validity study: the number of samples (seconds) that sitting was detected as lying on the back decreased from 8238 to 4425. The sensitivity percentages for walking were the lowest in the second validity study (mean sensitivity 39%). In this study, however, walking was hardly performed (0.8% of the measurement time). In the other studies the sensitivity score per measurement for

walking ranged from 69 to 95%, and the predictive value scores from 67 to 92%. In the standardised protocol of study 2, all walking periods were correctly detected.

Table 7.1a-c. The number of corresponding (bold/italic) and non-corresponding (plain) counts (1 count = 1 second) of video (rows) and AM (columns), added for all measurements. (a) validity study 1; (b) validity study 2; (c): validity study 3.

(7.1.a)

Video	Activity Monitor (counts)								
	Ly-ing	Sit-ting	Stand-ing	Walk-ing	Climb-ing stairs	Sit + cyclic trunk	Sit + cyclic legs	General move-ment	Total
Lying	***2016***	121	14	0	0	1	17	29	2198
Sitting	1	***4770***	26	27	0	151	27	96	5098
Standing	0	192	***8712***	615	0	1	61	298	9879
Walking	0	63	750	***4861***	14	2	8	116	5814
Climbing stairs	0	0	0	76	***244***	0	0	0	320
Driving wheelchair	0	0	0	0	0	***0***	0	0	0
Cycling	0	32	1	6	0	6	***2238***	14	2297
Transition	56	5	9	23	0	41	103	***163***	400
Total	2073	5183	9512	5608	258	202	2454	716	26006

(7.1.b)

Video	Activity Monitor (counts)								
	Ly-ing	Sit-ting	Stand-ing	Walk-ing	Climb-ing stairs	Sit + cyclic trunk	Sit + cyclic legs	General move-ment	Total
Lying	***20742***	225	23	0	0	0	0	50	21040
Sitting	4425	***51007***	43	13	0	71	205	263	56027
Standing	0	24	***1889***	14	0	0	0	13	1940
Walking	1	35	280	***236***	0	1	0	57	610
Climbing stairs	0	0	0	0	***0***	0	0	0	0
Driving wheelchair	0	0	0	0	0	***0***	0	0	0
Cycling	0	0	0	0	0	0	***0***	0	0
Transition	40	32	26	36	0	0	0	***36***	170
Total	25208	51323	2261	299	0	72	205	419	79787

(7.1.c)

Video	Activity Monitor (counts)								
	Ly-ing	Sit-ting	Stand-ing	Walk-ing	Climb-ing stairs	Sit + cyclic trunk	Sit + cyclic legs	General move-ment	Total
Lying	***596***	14	16	0	0	0	0	27	653
Sitting	0	***3894***	81	57	0	399	5	84	4520
Standing	0	254	***6713***	785	0	1	35	222	8010
Walking	0	20	1474	***6301***	9	7	5	121	7937
Climbing stairs	0	0	22	531	***175***	0	0	11	739
Driving wheelchair	0	0	0	0	0	***0***	0	0	0
Cycling	0	6	4	10	0	0	***335***	9	364
Transition	24	27	30	51	7	10	26	***107***	282
Total	620	4215	8340	7735	191	417	406	581	22505

In the first validity study the sensitivity scores for climbing stairs were considerably better than in the third validity study: mean score 76% versus 24%. If climbing stairs was misdetected, then this activity was mostly detected as walking. The predictive value scores for climbing stairs were higher and closer to each other: 95 and 91%, respectively. In the second validity study, climbing stairs was only performed in the standardised protocol. Climbing upstairs and downstairs were each correctly detected in three of the six subjects. In the other subjects climbing stairs was detected as walking or general movement.

Cycling was part of the protocol in study 1 and 3. The sensitivity for cycling (mean score 97 and 92%, respectively) was higher than the predictive value (91 and 83%, respectively). The video activity categories transitions and standing contributed most frequently – after cycling itself – to the AM category sitting with cyclic movement of the legs. The detection of standing as sitting with cyclic movement of the legs was in most cases due to a repetitive (cyclic) squat movement. In the second validity study, in one subject sitting was detected as sitting with cyclic leg movements for about 205 seconds; this was due to the subject moving his legs to the rhythm of the music.

Driving a wheelchair was not performed in the studies. Nevertheless, the video output category sitting was, respectively, in 3.0%, 0.1%, and 8.8% of its total duration detected as sitting with cyclic movements of the trunk.

The video category transitions, and the AM category general movement were categories with low sensitivity and predictive value percentages, respectively. The false detections were divided over a large range of activities.

Table 7.2. The sensitivity (Sens) and predictive value (P.V.) percentages for each activity category, and the agreement percentage. The overall percentages as well as the range between measurements within each study are provided. Cycling (video) is compared with sitting with cyclic movements of the legs (AM), and transitions (video) with general movement (AM).

Activity category	Agreement scores: mean (range) (%)					
	Study 1		Study 2		Study 3	
	Sens	P.V.	Sens	P.V.	Sens	P.V.
Lying	92 (47-100)	97 (87-100)	99 (98-100)	82 (0-100)	91 (50-100)	96 (83-100)
Sitting	94 (89-100)	92 (78-99)	91 (79-100)	99 (87-100)	86 (73-95)	92 (81-100)
Standing	88 (76-97)	92 (85-97)	97 (89-99)	84 (64-91)	84 (72-92)	80 (76-87)
Walking	84 (69-95)	87 (79-92)	39 (28-75)	79 (62-92)	79 (68-89)	81 (67-92)
Climbing stairs	76 (56-87)	95 (61-100)	-	-	24 (0-44)	91 (71-100)
Cycling/sit + cyclic legs	97 (91-100)	91 (84-99)	-	0 (0)	92 (84-100)	83 (51-100)
Transitions/ movement	41 (15-76)	23 (9-38)	21 (0-61)	9 (0-100)	38 (0-64)	18 (0-28)
Agreement		89 (84-93)		93 (79-99)		81 (78-84)

Table 7.3 presents an overall view on the duration of activities, as a percentage of the measurement time; Table 7.1 provides additionally insight into sources of differences in duration between videotape analysis and AM output. In all validity studies the duration of walking was slightly underestimated. An important reason

was that walking was more frequently detected as standing than standing was detected as walking. In validity study 2 the relative difference between both pairs was most obvious; as mentioned previously, however, in that study walking was hardly performed. Climbing stairs was underestimated, due to the rather frequent detection of climbing stairs as walking. Sitting with cyclic movements of the trunk was detected in all studies, although driving a wheelchair was not performed. General movement was overestimated in the three studies.

Table 7.3. Duration of activities as percentage of the measurement time, according to videotape analysis and Activity Monitor (AM). Cycling (video) is compared with sitting with cyclic movements of the legs (AM), driving a wheelchair (video) with sitting with cyclic movements of the trunk (AM), and transitions (video) with general movement (AM).

Activity category	Duration (% measurement time)					
	Study 1		Study 2		Study 3	
	Video	AM	Video	AM	Video	AM
Lying	8.5	8.0	26.4	31.6	3.0	2.7
Sitting	19.6	19.9	70.4	64.3	20.1	18.7
Standing	38.0	36.6	2.4	2.8	35.6	37.0
Walking	22.4	21.6	0.8	0.4	35.2	34.4
Climbing stairs	1.2	1.0	0.0	0.0	3.3	0.9
Cycling/sit + cyclic legs	8.8	9.4	0.0	0.3	1.6	1.8
Wheelchair/sit + cyclic trunk	0.0	0.8	0.0	0.1	0.0	1.9
Transitions/ movement	1.5	2.8	0.2	0.5	1.3	2.6

In the first study the total number of walking periods with a duration longer than 10 seconds was overestimated by 1: i.e. 170 versus 169 walking periods. The deviation from the number determined by video analysis ranged from –3 to +3 (mean number of walking periods per measurement was 14). In validity study 3 the AM underestimated the number of walking periods: 240 versus 255. The difference per measurement ranged from 0 to –4, with a mean number of walking periods of 32.

Discussion

In this chapter, the analysis software of the extended version of the AM was applied to already available data of three validity studies. The results have provided insight in the current characteristics and possibilities of the AM, but also in some problems and limitations. However, before the results are interpreted, something has to be said about the analysis method used.
First, the categories of the video analysis were not exactly the same as the AM output categories. For example, the AM output categories sitting with cyclic movements of the legs, sitting with cyclic movements of the trunk, and general movement, which will be discussed later in this section, do not entirely correspond with the video categories they were compared with: cycling, driving a wheelchair, and transitions. Second, it has to be noted that some grey areas exist: some performed activities can be classified into two activity categories. For example, shuffling is walking according to video analysis, standing with movement is standing. Both activities are close to each other, and the difference is not always entirely clear. The detection by the AM of such an activity is, in fact, always disputable. The assignment of an activity to one category and not to another, is not a matter of good or false. However, the quantitative part of the data analysis did not take this into account. Third, the increase in the number of activity categories leads to a smaller number of samples per category, which makes the results more sensitive for e.g. analysis errors (e.g. synchronisation error). Fourth, small differences in video analysis data existed between the first AM version (see Tables 4.1, 5.1, and 6.3) and the extended version (see Table 7.1a-c), whereas the same videotape data could be expected. These discrepancies are due to differences in the way transitions are handled, post-processing a larger number of categories, and two files that were partly damaged and could not be used in their entirety for re-analysis.

Although further optimisation of the analysis software can and will take place, the overall results show that the extended version of the AM is able to detect a large set of mobility-related activities. The overall agreement in the three studies is 89%, 93%, and 80%, compared with 90%, 88%, and 87%, respectively, in the first AM version (see chapters 4, 5, and 6).[2,3,4] Taking into account the four items discussed in the above paragraph, these percentages are satisfactory.
Static activities are rarely mutually misdetected. The occasions on which this happens and may happen are extensively discussed in chapters 4, 5, and 6. One significant difference from the first AM version was the decrease of time that sitting flopped in an easy chair was detected as lying on the back. This difference can be attributed to the correction for angular attachment deviations, which did not take

place in the first AM version, and to a small adaptation of the settings in the Activity Detection Knowledge Base (ADKB) with respect to the angular position of the trunk during lying on the back and sitting with the trunk backwards.

Walking generally has been detected well, with the exception of the second validity study. In that study, however, (a) the total walking duration was relatively short (0.8% of the measurement time, in contrast to 22.4% and 35.3% in validity study 1 and 3, respectively); (b) only short walking periods in a living room took place; and (c) these walking activities were usually in the grey area between walking, shuffling, and standing with movement. The lower agreement scores for walking must, therefore, not receive too much attention.

The duration of walking was underestimated in the three studies: –0.8%, (–0.4%, study 2), and –0.8%, respectively, whereas these percentages include the detection of climbing stairs as walking. This underestimation can be attributed to the fact that the walking category of the video analysis also contained less clear types of walking, such as shuffling and strolling. The AM possibly less easily detects these types of locomotion as walking. Future research will show whether the settings of walking within the ADKB need adaptation or not. An immediate adaptation does not seem to be justified at this moment, because the number of walking periods is detected quite well. In the first validity study the total number of walking periods determined by video and AM is almost equal. In the third study the number of walking periods is slightly (–6%) underestimated. This underestimation can be completely explained by the misdetection of climbing stairs as walking in the AM analysis. Regularly, a subject started to walk, climbed upstairs or downstairs, and then continued walking: two times a walking period according to video, one walking period according to AM (if misdetected). If the number of walking periods was calculated not considering the walking periods starting from climbing stairs, the original difference of –15 changed to a difference of +2 (+1%).

The detection of climbing stairs is clearly less well attained than the detection of walking. Climbing stairs had relatively high thresholds, i.e. that a high predictive value percentage is considered more important than a high sensitivity percentage. Background of this reasoning is, that climbing stairs is close to walking, that walking is usually more often performed than climbing stairs, and that the detection of walking as climbing stairs has to be avoided. The results of the validity studies indeed showed these effects in study 1 and 3: high (95 and 91%, respectively) mean predictive value percentages were found, while the mean sensitivity percentages were lower (76 and 24%, respectively). From the studies described in chapter 3 it appeared that, in most cases, normal types of climbing stairs could be distinguished from walking. That result was supported by the first validity study: 76% of climbing stairs was correctly determined; even climbing upstairs and downstairs usually were

correctly distinguished. In the second (standardised part) and third validity study, however, an important part of climbing stairs was determined as walking. A number of explanations can be given for this. First, due to the fact that the signals of climbing stairs and walking resemble each other, and due to the settings in the ADKB as mentioned above, this error can be expected. Second, the LP/angular signals are corrected for angular attachment discrepancies from an optimal 'in-plane' attachment while standing. This procedure may not have been optimal. At this moment we are still examining the best way to determine the correction angles. However, using correction factors appeared to have a beneficial effect on the detection of climbing stairs, without a negative effect on the detection of other activities. Third, the stairs at home and the way they are climbed may be different from the (climbing) stairs in the hospital, where validity study 1 and 2 as well as the master study described in chapter 3 were performed. We noticed that the periods of stair climbing in validity study 3 were shorter than in the other studies. Together with the finding that the AM generally starts the detection of climbing stairs one or two seconds later than video, and stops the climbing stairs detection one or two seconds earlier, this may also explain a part of the detection of climbing stairs as walking.

Cycling and, even more, driving a wheelchair, are activities with relatively non-specific characteristics. This means that the settings in the ADKB are crucial: low thresholds will mean high sensitivity and low predictive value percentages, high thresholds will mean low sensitivity and high predictive value percentages. The settings for these activities in the current ADKB – chosen deliberately – have a tendency to low thresholds: if cycling or driving a wheelchair is performed, the chance of a correct detection must be high, but other activities may also be detected as sitting with cyclic movements of the legs (cycling) or sitting with cyclic movements of the trunk. (driving a wheelchair). However, the category 'sitting with cyclic leg movements' appeared to be rather specific for cycling. Two types of performed activities were especially due to lower predictive value percentages of cycling: cyclic squat movement, i.e. to squat several times, and a subject moving his legs to the rhythm of the music. In all studies, sitting with cyclic trunk movements was for the major part detected during the video category sitting. This category therefore can be regarded as a refinement of the sitting category. Whether this category will be used or not in studies with non-wheelchair users, can be determined by the user by changing the settings in the ADKB or in the phase of post-processing (see chapter 3). Further study is needed to investigate the feasibility of this output category in studies where wheelchair use is focus of interest. An extra sensor attached on the arm may be an option in such studies.

'Transitions' in the video analysis is compared with the 'general movement' output category of the AM. The AM category contains all non-cyclic movements with a considerable degree of motility in legs and trunk, and will include more activities than just transitions. The low predictive value and sensitivity percentages of this category must therefore not receive too much attention.

Recently a new validation study has been performed in subjects with chronic congestive heart failure.[1] The results of that study are in agreement with the results presented in this chapter. It can be concluded that the extended version of the AM is a valuable extension of the first AM version. The detection of static activities remains stable, and walking and cycling were well determined. The ability to distinguish climbing stairs was less powerful, although some gain may be reached by a more valid determination of angular attachment deviations. The predictive value of driving a wheelchair will be low, because other sitting activities may also lead to cyclic movements of the trunk. Further study has to examine the sensitivity of the AM in detecting driving a wheelchair.

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Everyday physical activity in chronic congestive heart failure as measured with a novel Activity Monitor

Summary

Information on everyday physical activity is important in chronic congestive heart failure (CHF). To measure everyday physical activity objectively, an Activity Monitor (AM) has been developed. The aim was to obtain detailed information on everyday physical activity as measured with the AM, and on between-day variance in physical activity in patients with CHF class II or III (n=7). In addition, results found in the CHF group were compared with measurements in healthy matched comparison subjects (n=5).

Physical activity was measured with the AM, by which moment and rate of occurrence, and duration of several mobility-related activities can be determined. Motility (~ intensity) of physical activity can also be calculated. In the CHF group, measurements were performed during 2 consecutive weekdays and during one of these days of the subsequent week; in the healthy group, measurements were performed during 2 consecutive weekdays.

Mean duration of dynamic activities (as a percentage of the duration of the measurement day) was 3.9% (SD 1.5%) in the CHF group and 11.3% (SD 3.0%) in the comparison group (P=0.02). Mean motility of everyday physical activity and the number of walking periods (> 10 sec) were lower in the patients (0.06 versus 0.15 g, P=0.02; and 63 versus 181, P=0.01, respectively). Between-day variance in the duration of dynamic activities in the CHF group was significantly smaller (P<0.05) for different weekdays (0.80%) than for similar weekdays (5.49%).

The results indicate that the level of everyday physical activity as measured with the AM is considerably lower in patients with CHF than in healthy comparison subjects. The assumption that monitoring physical activity in CHF during similar weekdays would reduce the between-day variance was not supported.

Introduction

The most important symptoms in chronic congestive heart failure (CHF) are dyspnea and fatigue.[11,14,16] Because of these symptoms, patients are restricted in the performance of normal everyday activities such as walking, housekeeping, and gardening. Gradually then, a negative spiral may develop: hypoactivity → reduced fitness → early fatigue → further hypoactivity.

Measurement of everyday physical activity is important in CHF because it provides valuable information on disability in daily functioning and on the prognosis of the patient.[21] Futhermore, it can be assumed that everyday physical activity is related to quality of life. Until now, only a few studies are available on everyday physical activity in CHF. Methods that have been used include an actometer,[8,14] pedometer,[9,21,22] calorimeter,[17] and the doubly labelled water technique.[18,19] However, these methods only provide information on the level (or intensity) of everyday physical activity and no specification of the performed activities can be given.

In our department, an Activity Monitor (AM) has been developed,[3,4,5,6] by which detailed information on several aspects of everyday physical activity can be obtained. Briefly, the AM is based on long-term (more than 24 hours) ambulatory monitoring of signals from body-fixed accelerometers by which duration, rate and moment of occurrence of mobility-related activities [lying, sitting, standing, walking, walking stairs, running, cycling (sitting with cyclic movements of the legs), wheelchair driving (sitting with cyclic movements of the trunk)], and transitions between body postures can be detected.[1] Information on the variability of the acceleration signal (motility), which is assumed to be related with the intensity of activities, can also be obtained from the device. Apart from monitoring accelerations, other signals such as heart rate or ECG can be recorded simultaneously. The AM has recently been validated in patients with chronic CHF.[1]

The aim of the present study was to obtain insight in the everyday physical activity of patients with mild to moderate chronic CHF, as measured with the AM. Furthermore, the between-day variance in physical activity was assessed because this information is important in intervention studies for the determination of the optimal number of activity monitoring days and the required sample size. We also studied whether the between-day variance in physical activity can be reduced by monitoring on similar weekdays. The research questions were:

(1a) What is the everyday physical activity of patients with mild to moderate CHF as measured with the AM?

(1b) Is there a difference in everyday physical activity between patients with mild to moderate CHF and healthy matched comparison subjects?

(2a) What is the between-day variance in physical activity in patients with mild to moderate CHF, and does the between-day variance of similar weekdays differ from the between-day variance of different weekdays?

(2b) Is there a difference in the between-day variance between patients with mild to moderate CHF and healthy matched comparison subjects?

Methods

The study was part of a large screening project in patients with chronic CHF, performed at the Heartcenter Rotterdam (Cardiology Departments of University Hospital Rotterdam and of the Zuiderziekenhuis). Approval was given by the Medical Ethics Committee of the University Hospital Rotterdam and informed consent was obtained from all participants.

Subjects

Seven male patients with stable CHF were recruited from patient records of the Cardiology Department of the University Hospital Rotterdam. Four patients were in New York Heart Association (NYHA) class II, 3 patients were in class III (Criteria Committee of the New York Heart Association 1973.[7] All patients had symptoms of CHF for at least 1 year. Age ranged from 60 to 72 years (median 63 years). The most frequent cause of CHF was ischemia (n=5); in 1 patient the cause was cardiomyopathy, in 1 patient hypertension. Most subjects were taking diuretics (n=6) and ace inhibitors (n=7). Digoxin, cordarone, and nitrates were taken by respectively 3, 3, and 2 patients. Left ventricular ejection fraction ranged from 22.7% to 36.0% (median 28.4%). All patients were retired from work; 5 patients were living with a partner, and 2 patients lived alone.

In addition, for 5 of the patients, a healthy male comparison subject was selected (these 5 patients were the ones who lived with a partner; within the measurement period for practical reasons it was not possible to include comparison subjects for the 2 patients who lived alone). The comparison subjects had no diseases or impairments which disturbed everyday physical activity and were of the same age (±5 years) as the patients (age range 61 to 71 years, median 65 years). Furthermore, they were in a comparable living situation as the patients (lived with a partner and were retired).

Instrument

Four IC-3031 uniaxial piezo-resistive accelerometers were used (size 1×1×1 cm or 2×2×0.5 cm). Two sensors were attached to each thigh, and two were attached to the skin over the sternum. The sensors on the leg were fixed on the skin by Rolian

Kushionflex, adhesive medical tape was used to consolidate the attachment. The sensors on the trunk were fixed on the skin by silicone-based stickers. For a more detailed description of the sensors and the attachment see Bussmann et al.[3,4] and Veltink et al.[20]

The accelerometers were connected to a Vitaport2™ data recorder (Temec Instruments, Kerkrade, The Netherlands; size 15x9x4.5 cm; weight 700 g) or a Rotterdam Activity Monitor (RAM, Temec Instruments, Kerkrade, The Netherlands; size 15x9x3.5 cm; weight 500 g), which were worn in a padded bag round the waist; for practical reasons different devices were used. Accelerometer signals were stored digitally on a PCMCIA hard disk or flash card with a sampling frequency of 32 Hz. After the measurement, data were downloaded onto a Macintosh computer for analysis. Corrections were made for angular deviations of the sensors from the vertical plane (deviation was assessed during standing periods). In the analysis (Signal Processing and Inferencing Language (S.P.I.L.),[10] three parts can be distinguished:

(1) *Feature processing,* i.e. new signals with specific characteristics are derived from measured signals. From each measured signal, 3 feature signals were derived: (a) Low-pass (LP)/angular feature: the LP/angular signals are created after low-pass filtering (0.3 Hz) of the measured signals. This signal is subsequently converted to angles (range –90 to +90 degrees); (b) Motility feature: the motility signals are created after subsequent high-pass filtering (0.3 Hz), rectifying, and smoothing. This signal depends on the variability of the measured signal around the mean, or 'acceleration energy'; (c) Frequency feature: the frequency signal is based upon a band-pass filtered derivative (0.3-2 Hz for the legs, 0.6-4 Hz for the trunk) of the measured signal; this band-passed signal is the input of the Fast Time Frequency Transform (FTFT)[10] procedure (a type of instantaneous frequency analysis which determines the frequency of the band-passed signal). Pre-set criteria determine whether the frequency is valid. All features had a time resolution of 1 second.

(2) *Activity detection*, i.e. based on the feature channels, 23 activity subcategories are determined (e.g. walking fast, walking slow, standing upright, standing with trunk flexed). For each subcategory, a minimum and maximum value is pre-set for each feature signal in an activity detection knowledge base. Each measurement sample (1 second), for each subcategory and each feature, the distance is calculated from the actual feature value to the pre-set range. If the actual value is within the range, the distance is 0. The calculated distances of the features are added for each activity and the activity with the shortest distance is selected. If an activity is detected, but the distance is above a pre-set general threshold, indicating a relatively high degree of uncertainty, the category 'unknown' is selected.

(3) *Post-processing*, i.e. output signals of the activity detection are processed in such a way that readable and relevant information is provided. For example, the 23

subcategories are reduced to a smaller number of AM output categories. In this study, the static activities that were distinguished were lying, sitting (including sitting with cyclic movements of the trunk), and standing; the dynamic activities that were distinguished were walking (including climbing stairs and running), cycling, and general movement (all non-cyclic movements with a considerable degree of motility in the legs and trunk, e.g. moving around in the kitchen between table and dresser while cooking). Sitting with cyclic movements of the trunk is representative for wheelchair driving: a sitting position with cyclical movements of the trunk. However, because in our previous validation study of the AM in CHF[1] it was found that in 37% of the time during car driving, sitting was interpreted as sitting with cyclic movements of the trunk by the AM, and because none of the subjects of the present study had been driving a wheelchair during the measurements, the activities sitting and sitting with cyclic movements of the trunk were grouped together as sitting. Because the detection of climbing stairs was found to be poor in the validation study (climbing stairs was often detected as walking), the activities walking and climbing stairs were grouped together as walking. Also running was added to this activity category. Short-lasting activities (< 5 seconds) were disregarded. Values of the four motility signals were added and divided by 4 to obtain the body motility (~intensity or level of everyday physical activity). The automatic analysis of a 24-hour measurement lasted about 30 minutes. The output of the AM – the continuous selection of an activity – in this study had a time resolution of 1 sec.

Protocol

In order to obtain information on the between-day variance in physical activity and on possible differences in between-day variance of similar weekdays and of different weekdays, the patients were measured with the AM during 2 consecutive weekdays (and nights, 48-hour measurement) and during one of these days of the subsequent week (24-hour measurement). In the comparison subjects, for practical reasons, we performed measurements only during 2 consecutive weekdays (and nights). Measurements of both groups were performed in the same period.

To interfere as little as possible with the normal daily activity patterns, subjects were fitted at home with the AM (mostly between 10:00 a.m. and 11:30 a.m.). If necessary, the subjects were visited during the measurement period to replace the AM batteries and memory card, or to consolidate the attachment of the sensors. During the activity monitoring, subjects were not allowed to swim or take a shower. After the measurements, we visited the subjects again to remove the instrumentation and to ask them questions about the kind of activities they had performed, about the convenience of the AM, and about possible abnormalities (such as illness) during the measurements.

In order to avoid bias, the real aim of the study was initially not explained to the subjects; they were told that the measurements would be used to obtain insight in the practical problems subjects may experience during long-term measurements with the AM. Furthermore, subjects were instructed to continue their ordinary daily life. After the measurements, complete information on the aim of the study was given and also the reason for not having given that information before the measurements. All subjects agreed with this procedure, so all measurements could be included in the analysis.

Data analysis

In the analysis, only corresponding measurement periods were used between days or between patients and comparison subjects; e.g. in case patient data were missing say between 12:00 and 13:00 on day 1, data of this period on the other measurement days were excluded from analysis and the same was done in the comparison subject. To obtain information on everyday physical activity, the following variables were assessed: duration of static activities (as a percentage of the duration of the measurement day); duration of dynamic activities (as a percentage of the duration of the measurement day); mutual distribution of activities within the static activity category and within the dynamic activity category, number of transitions; number of sit-to-stand transitions; mean motility; mean motility during walking; number of walking periods, and distribution of the duration of walking periods. For the total CHF group (n=7), data are presented as mean (SD) over 3 measurement days; for 5 of the patients and their comparison subjects, data are presented as mean (SD) over two consecutive weekdays. Differences in the mutual distribution within the static activity category and within the dynamic activity category, or in the distribution of the duration of walking periods between patients and their comparison subjects were tested with MANOVA; other differences between both groups were tested with the Mann-Whitney *U* test. Comparisons within the study groups were made using the Wilcoxon test or the Friedman test.

The variable that was used for the assessment of the between-day variance in physical activity was the duration that dynamic activities were performed, as a percentage of the duration of the measurement day. In addition, also the between-subject variance in the duration of dynamic activities was assessed. Information on the between-day and between-subject variance in duration of dynamic activities was obtained from one way analysis of variance. Depending on which weekday was measured in the second week, in some of the patients the first 24-hour part of the 48-hour measurement (is measurement day 1) was assigned as weekday 1, whereas in others the second 24-hour part (is measurement day 2) was assigned as weekday 1. Measurement days in the comparison group were similar to weekdays. Between-day variance of similar weekdays and of different weekdays (CHF group) was based on measurements with

one week in between. Differences in variance between the CHF group and the healthy group or within the CHF group (similar weekdays versus different weekdays) were tested with the F-test.
All statistics were done with SPSS/PC$^+$; statistical significance was assumed when the P-value was less than 0.05.

Results

The mean (SD) time of a measurement day that was used for analysis was 20.9 (2.6) hours (CHF, n=7) and 19.6 (2.0) hours (CHF, n=5 and comparison group).

Everyday physical activity
Tables 8.1 and 8.2 present variables that are related to everyday physical activity. The percentage of the day that subjects performed dynamic activities, the mean motility, and the number of walking periods (>10 s) were significantly smaller in the patients than in their healthy comparison subjects (Table 8.1). The average duration patients spent with dynamic activities was 0.8 hour per 19.6 hours of measurement, whereas in the comparison subjects the average duration was 2.2 hours per 19.6 hours. The total number of transitions and the total number of sit-to-stand transitions tended to be lower (P=0.05) in the patient group than in the healthy group (Table 8.1). The mean motility during walking did not differ between both groups.
There were no significant differences between both groups in the mutual distribution of the durations of lying, sitting, and standing (as percentages of the duration of the static activity category), and in the mutual distribution of the durations of walking, cycling, and general movement (as percentages of the duration of the dynamic activity category) (Table 8.2). The percentage of the time during a day that subjects spent walking was 3.7 (SD 1.6) in the CHF group (n=5) and 9.1 (SD 3.3) in the healthy group (P=0.03). In Table 8.3, seven walking categories (from 0 - 10 seconds up to 10 - 30 minutes) and the time (as a percentage of the total walking time) spent in these categories are shown. There were no significant differences in the time spent in the different walking categories between both groups.

Between-day variance
There were no significant differences in duration of dynamic activities between the weekdays in both groups (P=0.87 and 0.50, respectively) (Table 8.4). There was also no significant difference in duration of dynamic activities between the first and second measurement day of the consecutive measurement in the CHF group: 4.2% (SD 3.2%) and 4.7% (SD 3.0%), respectively (P=0.99). The between-day variance in the duration of dynamic activities in the CHF group was significantly (0.01<P<0.05) smaller for

different weekdays (0.80%) than for similar weekdays (5.49%). The between-day variance tended to be smaller in the patients than in their comparison subjects (3.94% versus 16.18%, 0.05<P<0.10). The between-subject variance in the duration of dynamic activities in the CHF group was 6.37% (n=7) and 1.09% (n=5); the between-subject variance in the comparison group was 3.06%, which was not significantly diferent from CHF (P>0.10).

Table 8.1. Mean (SD) duration of the static activities (lying, sitting, standing) and the dynamic activities (walking, cycling, general movement), number of transitions between postures, motility, and number of walking periods per measurement day in patients with congestive heart failure (CHF) and healthy comparison subjects (H) [ranges within brackets].

Variable	mean (SD) [range]			P-value*
	CHF (n=7)	CHF (n=5)	H (n=5)	
Duration static activity cat. (% of measurement time)	95.6 (2.7) [89.6-98.3]	95.9 (1.6) [93.4-97.9]	88.8 (3.0) [85.9-93.4]	0.01
Duration dynamic activities cat. (% of measurement time)	4.3 (2.5) [1.7-9.9]	3.9 (1.5) [2.2-6.7]	11.3 (3.0) [6.6-14.1]	0.02
Number of transitions	105 (30) [49-159]	90 (27) [40-120]	133 (36) [99-198]	0.05
Number of sit-to-stand Transitions	38 (12) [12-57]	33 (12) [10-45]	54 (19) [41-89]	0.05
Mean motility (g)	0.06 (0.03) [0.03-0.12]	0.06 (0.02) [0.03-0.08]	0.15 (0.06) [0.08-0.21]	0.02
Mean motility during walking (g)	0.91 (0.07) [0.80-1.00]	0.96 (0.14) [0.78-1.11]	1.02 (0.19) [0.89-1.35]	0.92
Number of walking periods (> 10 s)	78 (50) [27-189]	63 (17) [35-82]	181 (62) [91-234]	0.01

* CHF (n=5) versus H (n=5)

Table 8.2. Mutual distribution of the duration of the static activities lying, sitting and standing within the static activity category, and of the dynamic activities walking, cycling, and general movement, within the dynamic activity category. The data are calculated as a percentage of the duration of the static and dynamic activity category, respectively, in patients with chronic congestive heart failure (CHF) and their healthy control subjects (H).

Activity category	Duration (% of static or dynamic category) mean (SD)			P-value*
	CHF (n=7)	CHF (n=5)	H (n=5)	
Static activities				
Lying	52.5 (9.3)	54.4 (10.5)	52.0 (11.1)	
Sitting	34.1 (8.0)	31.5 (8.6)	26.3 (6.6)	0.29
Standing	13.3 (3.3)	14.2 (3.4)	21.7 (6.7)	
Dynamic activities				
Walking	88.7 (6.4)	87.7 (8.0)	79.4 (9.9)	
Cycling	2.5 (3.2)	3.2 (4.9)	12.0 (9.2)	0.10
General movement	8.7 (5.1)	9.2 (6.2)	8.7 (6.1)	

* mutual distribution activities CHF (n=5) versus H (n=5)

Table 8.3. Seven walking categories and the time (as a percentage of the walking time) spent on these categories by patients with chronic congestive heart failure (CHF) and by healthy control subjects (H).

Walking category	Time spent (% walking time) mean (SD)		
	CHF (n=7)	CHF (n=5)	H (n=5)*
0 – 10 sec	22.9 (11.2)	19.8 (10.8)	16.0 (4.9)
10 – 30 sec	40.7 (17.0)	44.1 (19.3)	37.8 (7.9)
30 – 60 sec	15.1 (8.7)	15.1 (8.0)	23.9 (3.5)
1 – 2 min	5.7 (6.6)	4.2 (6.3)	10.9 (5.0)
2 – 5 min	10.3 (12.7)	9.9 (14.6)	6.2 (5.5)
5 – 10 min	2.5 (5.9)	1.7 (3.3)	2.2 (5.5)
10 – 30 min	2.8 (10.9)	5.1 (15.3)	3.0 (7.3)

* differences between CHF and H were not statistically significant (P=0.16)

Table 8.4. Mean duration of dynamic activities (walking, cycling, general movement; as a percentage of the measurement time during a day), and between-day variance in duration of dynamic activities (as a percentage of the measurement time during a day) in patients with chronic congestive heart failure (CHF) and healthy matched control subjects (H).

Measurement day(s)	Duration dynamic activities (% measurement day)			P-value**
	CHF (n=7)	CHF (n=5)	H (n=5)	
Mean duration				
Week 1, day 1*	4.8	3.6	12.5	0.05
Week 1, day 2*	4.5	4.1	10.0	0.05
Week 2 day 1*	3.7	-	-	-
Between-day variance				
Weekday 1,1 - 1,2 - 2,1	3.14	-	-	-
Weekday 1,1 – 1,2	2.42	3.94	16.18	<0.10
Weekday 1,1 – 2,1	5.49	-	-	-
Weekday 1,2 – 2,1	0.80***	-	-	-

* Weekdays 1,1 and 1,2 are consecutive days, weekday 2,1 is similar to weekday 1,1, but measured one week later; ** CHF(n=5) versus H (n=5); *** significantly different (0.01<P<0.05) from between-day variance of weekdays 1,1 – 2,1

Discussion

In order to obtain information on the everyday physical activity and the between-day variance in physical activity in CHF, physical activity was measured during 3 weekdays in 7 patients with CHF (class II or III) with the AM. In addition, measurements in this group were compared with measurements in healthy matched comparison subjects (n=5).

The AM has been found to be valid in healthy subjects, in patients after failed back surgery, and patients with an amputation of the leg (see chapters 4, 5, and 6).[4,5,6] However, in these validation studies, a first version of the AM analysis software was used (e.g. it was not possible to distinguish between dynamic activities in the former

validation studies). Recently, the AM (the extended version) has been validated in CHF.[1] The overall agreement between the AM and the reference method (video) was found to be 90%, and percentages of sensitivity and predictive value were for the detection of most activities over 80%. Furthermore, misdetections that did occur were all explicable and mainly due to methodological problems.

For practical reasons, in the present study we used the Vitaport2™ and the RAM data recorder. The only difference between the devices is that the RAM is smaller and lighter than the Vitaport2™ (sensors, signal processing, and analysis of signals are identical). Therefore, the RAM is more convenient for subjects, particularly during long-term measurements. Subjects experienced the AM (both Vitaport2™ and RAM) generally as rather convenient, also during the night. Some subjects disliked being seen with the instrumentation, so they wore the monitor under their shirt or jacket. However, particularly during gardening, the devices (both RAM and Vitaport2™) were in some subjects of (minor) hindrance. Because this hindrance was experienced with both devices and in both the patient and control group, we do not expect the use of different devices to have interfered with the results.

In spite of the fact that we used special silicone-based stickers on the trunk, the trunk sensors came loose from the skin in one comparison subject during running (extensive perspiration). In some other subjects (patients and comparison subjects), the trunk sensors came slightly loose from the skin at the end of the (48-hour) measurement (mainly in subjects with extensive chest hair, or during measurements on warmer days). No problems occurred with the attachment of the sensors on the leg. Future research at our department will focus on the (further) optimisation of the sensor attachment (particularly of the trunk sensors) and the convenience of the RAM (e.g. (further) reducing the size and weight of the monitor, adapting the monitor and sensors in a way that subjects can take a shower during the measurement period).

It was our aim to obtain 3 and 2 24-hour measurements in the patients and the comparison subjects, respectively. However, in most of the subjects, the period that could be used for analysis was less than 24 hours (~20-21 hours). This is because, in the comparison subjects, it was impossible (for practical reasons) to change the memory cards during the measurement period, which resulted in a measurement period in these subjects of 44 hours (on 2 consecutive days). Currently, flash cards are available with more memory capacity. Because we used only the corresponding periods in the patients for analysis, the average duration of measurement in the CHF group was also smaller than 24 hours. Furthermore, in one CHF patient a battery problem occurred during the measurement and, as mentioned above, in one healthy subject the trunk sensor came loose from the skin during running.

Everyday physical activity

Based on the percentage of the time that dynamic activities were performed, it can be concluded that the patients with CHF were considerably less active than healthy subjects of the same age and living situation (Table 8.1). In both the patient and comparison group, walking was the main dynamic activity (Table 8.2). The patients spent significantly (P=0.03) less time during a day with walking than their healthy comparison subjects, but there were no differences in the distribution of the duration of walking periods between both groups (Table 8.3). Also the motility during walking (Table 8.1), which is assumed to be related with walking speed,[2,12,13,15] did not differ between groups. In the comparison group, cycling was also a relatively important dynamic activity, but the difference in the mutual distribution of dynamic activities between patients and comparison subjects was not statistically significant (P=0.10). Because of the small sample size, from the results of this study no conclusions can be drawn on possible differences in everyday physical activity between NYHA class II and III patients.

The number of transitions and sit-to-stand transitions tended to be lower in the CHF group than in the comparison group (see Table 8.1). However, the number of transitions and sit-to-stand transitions may not be representative for the level of everyday physical activity. For example, one patient had on average only 10 sit-to-stand transitions per day, but based on the percentage of the time spent on dynamic activities, he was the most active person in the patient group. This indicates that the duration of dynamic activities and the number of transitions reflect different aspects of everyday physical activity.

The low level of everyday physical activity in CHF, as found in the present study, has also been reported by others: recently, Toth et al.[19] measured free-living energy expenditure in 25 patients with CHF [cachectic and noncachectic, aged 73 (SD 6) years and 67 (SD 5) years, respectively] and in 50 healthy comparison subjects [age 69 (SD 6) years] with the doubly labelled water technique. It was found that the energy expenditure for physical activity was significantly lower in the CHF group [269 (SD 307) kcal/day in chachectic patients and 416 (SD 361) kcal/day in noncachectic patients] than in the healthy group [728 (SD 374) kcal/day]. Walsh et al.[21] reported significantly lower pedometer scores in patients with CHF than in healthy comparison subjects (258 (45) $\times 10^2$ versus 619 (67) $\times 10^2$ steps/week). Also Hoodless et al.[9] found a reduction of about 50% in pedometer scores in CHF in comparison with healthy controls.

Between-day variance

Because there was no difference in the mean duration of dynamic activities between the first and second measurement day of the consecutive measurement in both the

CHF and comparison group, it can be concluded that the effect of habituation to the AM on everyday physical activity can be ignored. In the CHF group, the between-day variance in the duration of dynamic activities between different weekdays (measured with one week in between) was significantly ($P<0.05$) smaller than the between-day variance between similar weekdays measured with one week in between (0.80% versus 5.49%, respectively). This finding was in contrast with our expectation that on similar weekdays, similar activities would be performed (e.g. shopping on Mondays (with a relatively long duration of walking), housekeeping activities on Tuesdays, etc.). Apparently, a weekly pattern of physical activity did not exist in the patients. Based on this result it can be concluded that monitoring physical activity in CHF during similar weekdays (e.g. in intervention studies) does not reduce the between-day variance. We have no explanation for the finding that the between-day variance of similar weekdays was significantly larger than the between-day variance of different weekdays.

The between-day variance in duration of dynamic activities in the CHF group was relatively large (3.14%, Table 8.4), but tended to be smaller in the patients than in their healthy comparison subjects (3.94% versus 16.18%, $P<0.10$, Table 8.4). The tendency for a smaller between-day variance in the CHF group than in the healthy group is probably due to low levels of everyday physical activity and to low functional capacity in CHF.

There was a relatively large difference in between-subject variance in duration of dynamic activities between the total (n=7) patient group (6.37%) and the n=5 patient group (1.09%). This can be explained by the fact that one of the patients who was excluded from comparison with healthy subjects was the most active patient and the other one was the least active patient.

Information on the variance in everyday physical activity in CHF is essential in order to assess the number of activity monitoring days that is required to get insight in the customary daily physical activity in this group. In intervention studies with paired comparisons, particularly the between-day variance is important. Based on this variance, the magnitude of the effect one wants to detect, and the available number of subjects, the required number of sampling days can be assessed. In the near future, an intervention study on the effects of aerobic training on everyday physical activity in patients with CHF (class II and III) will be performed at our department (in co-operation with the Heartcenter Rotterdam). Based on (a) the between-day variance in the CHF group of 3.14% (when assessing differences between one day before and one day after an intervention, two times the between-day variance should be taken into account), and (b) a (relative) increase of 33% (~16 minutes) in duration of dynamic activities, which we consider clinically relevant, the required number of patients in our intervention group (with monitoring physical activity

during one day before and one day after the intervention) is 70 (power is 90%). When increasing the number of sampling days up to 2 days before and 2 days after the intervention, the between-day variance is halved, and an intervention group of 35 patients should be sufficient. However, one should be aware that in the above described calculations the long-term between-day variance in everyday physical activity is not considered.

Conclusion

In contrast with available studies, the present study provides detailed information on different aspects of everyday physical activity in CHF as measured with the AM. The results indicated that levels of everyday physical activity are considerably lower in patients with mild to moderate CHF than in healthy matched comparison subjects. In both the patients and the comparison subjects, walking was the main dynamic activity, and the motility during walking did not differ between the groups. In the comparison group, cycling was also a relatively important dynamic activity, but the mutual distribution of the duration of the dynamic activities was not statistically significant between the patient group and comparison group. The between-day variance in the CHF group was relatively large, but tended to be smaller (but not statistically significant) than the between-day variance in the comparison group. Activity monitoring during similar weekdays did not reduce the between-day variance.

Acknowledgements

We wish to express our thanks to Fokke Jonkman and Anke Wiebenga (Heartcenter Rotterdam) for their helpful comments and valuable assistance in subject recruitment. This study was supported, in part, by the Rotterdam Foundation for Cardiac Rehabilitation.

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Analysis and decomposition of signals obtained by thigh-fixed accelerometry during walking

Summary

The use of piezo-resistive uni-axial accelerometer signals in the assessment of the quantity and quality of walking is complicated by the fact that the measured signal is composed of different types of acceleration, and that the signal may vary between or within measurements. The aim of the study was to obtain insight in the signal from a tangential accelerometer attached at the thigh during walking, and in its components; to investigate its relation with temporal events; and to investigate the influence of the factors subject variability, walking speed, walking surface, and sensor attachment on the signal.

Six subjects walked with three different speeds (comfortable, slow, and fast) and under several conditions (different types of walking surface and sensor attachment). Simultaneous measurements were performed with accelerometers, footswitches, and an optoelectronic system. From the optoelectronic system the components of the acceleration signal were calculated. The effect of factors was determined by means of the root mean square (rms), compared to the rms values on intra-subject variability.

The acceleration signal was generally characterised by relatively high frequency components with regard to the movement frequency, and low amplitudes. Most pronounced was a high positive acceleration peak just before heel strike. The inclination component was characterised by a sinusoidal shape. The inertial component and inclination component had opposing contributions to the signal. The effect of deformation of the thigh on the signal was small and not systematic. If the hip joint marker was taken as point of reference, both the rotational and the translational acceleration components contributed to the inertial acceleration. A significant increase of the rms values occurred due to inter-subject variability, different walking speeds, and displacing the sensors 2 cm medially.

Insight in the acceleration, its components, and its sensitivity to subject variability, walking speed, walking surface, and sensor attachment was obtained. This knowledge will contribute to the further use of these signals in the assessment of gait quality, and in the detection of walking.

Introduction

Objective measurement of the activities actually performed by subjects during their normal daily life can be regarded as a new and relevant technique. One such instrument described in literature is the Activity Monitor (AM):[3,4,5] an instrument based on long-term ambulatory accelerometry, and aimed at the measurement of the quantity and quality of several mobility-related activities. The configuration normally used consists of four uni-axial piezo-resistive accelerometers attached to the trunk and thighs, a portable data recorder, and analysis software.
The accelerometer signals can be used to detect and distinguish static activities, such as sitting, standing and lying, and dynamic activities, such as walking, cycling and climbing stairs. Due to the importance of walking in daily functioning, this activity receives special interest. Specific features (low-pass/angular, motility, and frequency) of the accelerometer signals have been used with respect to the detection of walking (see chapter 3). However, the detection of walking may be optimised using other or additional features.
The accelerometer signals can also be used to measure the way in which activities are performed ('quality'). Walking is a cyclic movement, and the accelerometer signal measured will show a cyclic pattern. This acceleration signal contains a lot of information on the quality of walking, e.g. the walking pattern and movement co-ordination. For example, a study performed by our research group in elderly subjects showed that shape-related measures derived from accelerometer signals correlated with age.[19] Other researchers also used accelerometry to obtain data on movement quality or as a type of gait analysis.[2,6,8,16,20,21] In research with the AM, the signals are currently used to measure stride frequency and shape-related stability parameters. The potential to provide other quality variables, such as stride length, walking speed, and duration and timing of swing and stand phase, is still under study.

To use the accelerometer signals as a well-founded provider of gait variables, and of additional features in the detection of walking, further knowledge on the accelerometer signal during gait cycles is required. To obtain this insight, the characteristics (such as amplitude, timing and shape) of the accelerometer signal during walking should be understood and related to specific events in the gait cycle. These requirements were not fulfilled so far: interpretation of the measured accelerometer signals is not simple and is not immediately obvious. Piezo-resistive accelerometers provide a combined measurement of components of the gravitational acceleration and inertial accelerations (see chapter 3),[24] which can not be distinguished by means of the accelerometer signals themselves. To obtain insight in

the signal, these components need to be quantified. Furthermore, the signal has to be related to temporal events, such as heel strike and toe off: in literature, these events are widely used to describe and assess the walking pattern.[23,27] The determination of these events from the accelerosignal should increase its value in the assessment of gait quality. Due to the known influence of walking speed on the gait pattern, the issues discussed above need to be studied at different walking speeds.

The accelerometer signal may be sensitive to changing factors and conditions. From the ambulatory measurement viewpoint, some of them are considered most relevant. (a) Intra-subject variability should be small to allow reliable statements on (changes in) walking quality. To allow the use of stable, walking-specific characteristics the intra-subject variability as well as the inter-subject variability should be small. (b) Walking speed may vary within measurements, and walking speed is known to influence the movement co-ordination,[25] and thus also the accelerometer signals. (c) During ambulatory measurements persons may walk over different surfaces, which may have an effect on the accelerometer signals measured. (d) Sensor location may differ within or between measurements. Furthermore, different methods of attachment may be used to secure the sensor to the body. Measuring the influence of these factors on the accelerometer signals will provide insight in the practical feasibility of accelerometer signals to measure actual changes in gait quality, and in the possibility to use walking-specific characteristics in the detection of walking.

The aim of the study was (1) to obtain insight in the signal from a tangential accelerometer attached at the thigh during walking, and in its components; (2) to investigate its relation with temporal events; and (3) to investigate the influence of subject variability, walking speed, walking surface, and sensor attachment on the signal.

Theory

Uni-axial piezo-resistive accelerometers

A uni-axial piezo-resistive accelerometer consists of a mass, connected to a frame by beams which can be represented by a damped spring (Figure 9.1).[1,24] In the beams piezoresistors are mounted, forming a bridge circuit.[27] The value of the resistors depends on the deformation of the beams which, in turn, depends on the magnitude of acceleration. Due to their structure, these uni-axial sensors are only sensitive for accelerations perpendicular to their surface. Contrary to piezoelectric accelerometers, piezo-resistive accelerometers measure DC as well as AC components. Therefore, even if the sensor does not move, part of the gravitational

acceleration ($\bar{a}_{grav}$) is measured. The part of vector $\bar{a}_{grav}$ (9.81 m.s^{-2}) that is measured ($\bar{a}_{grav,sens}$) depends on the angle φ_2 between the sensitive axis of the sensor and the vertical $\bar{a}_{grav}$. If the accelerometer is accelerated, an inertial acceleration ($\bar{a}_{inert}$) will occur. The part of $\bar{a}_{inert}$ that is measured ($\bar{a}_{inert,sens}$) depends on the angle φ_1 between the sensitive axis of the sensor and $\bar{a}_{inert}$ (see chapter 3). The gravitational acceleration acts on the mass, while $\bar{a}_{inert}$ acts on the frame of the sensor; $\bar{a}_{grav}$ and $\bar{a}_{inert}$ therefore have opposite effects. If they act in opposite directions (as in Figure 9.1), they both make contributions to the measured signal in the same direction. If $\bar{a}_{grav}$ and $\bar{a}_{inert}$ act in the same direction, they make opposing contributions.

The standard configuration of the AM consists of four accelerometers, attached at the thighs and trunk. The thigh sensors are sensitive in anterior-posterior direction while standing. Because the thigh sensors are assumed to be more informative about quality of walking than the trunk sensors, in this study only the signals from the thigh sensors were studied. These sensors are attached mid-front the thigh, halfway spina iliaca anterior superior and upper side of the patella. The sensor is positioned in such a way, that angular positions with the hip flexed will cause positive signals (due to $\bar{a}_{grav}$), as will inertial accelerations in forward movements and decelerations in backward movements.

The accelerometer signals measured on the thigh are thus a combination of $a_{grav,sens}$ and $a_{inert,sens}$. Although the distinction between these two components is most important, the inertial acceleration in its turn can be additionally decomposed. The inertial component can be the result of three types of accelerations: (a) translational accelerations of the segment ($a_{transl,sens}$); (b) rotational accelerations of the segment ($a_{rot,sens}$); and (c) accelerations due to non-rigidity (deformation) of the segment itself, like vibrations and muscular contractions ($a_{deform,sens}$). The mutual distribution between the translational and rotational components depends on the chosen point of reference.

Accelerometer signals from the thigh sensor

Accelerometer signals are often difficult to interpret or predict – especially during activities such as walking – in contrast to signals on position and speed. The signals from piezo-resistive accelerometers are especially complex, because they contain different acceleration components. To obtain some theoretical insight in the different components of the accelerometer signals, the movement of the thigh during walking, and the accelerations measured by the accelerometer attached at the thigh, can be modelled (Figure 9.2). In the used model the amplitudes of the components are normalised and, in fact, not taken into account.

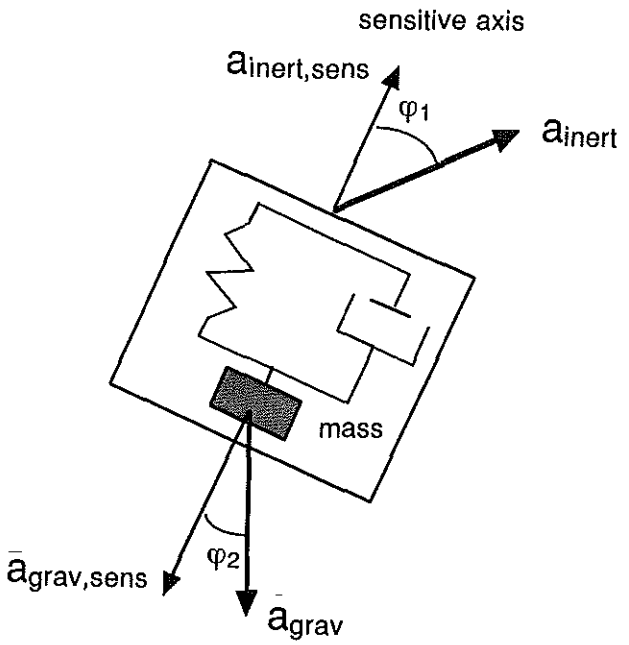

Figure 9.1. Uni-axial piezo-resistive accelerometers measure a combination of the components $\bar{a}_{inert,sens}$ *and* $\bar{a}_{grav,sens}$ *of the inertial and gravitational acceleration, respectively.* φ_1 *is the angle between the sensitive axis of the accelerometer and* $\bar{a}_{inert}$; φ_2 *is the angle between the sensitive axis and* $\bar{a}_{grav}$.

During the stance phase (phase ph_0), the leg is regarded as an inverse pendulum, rotating around the ankle. When the hip joint is taken as reference, as in the present study, the movement of the thigh can be expressed as a combination of a backward rotation and a forward translation of the thigh. During the swing phase (phase ph_1), the thigh can be regarded as a pendulum, which rotates in forward direction and also translates in forward direction, due to the inverse pendulum movement of the opposite leg. The gravitational component measured by the accelerometer according to the model ($a_{grav,model}$) will have the highest value during the cycle at moment t_0, to decrease during phase ph_0, and to be most negative at moment t_1. During ph_1 $a_{grav,model}$ will increase and become positive, to be maximal positive again at t_2; moment t_2 equals t_0, just as phase ph_2 equals ph_0. A sinusoidal curve will be the result (Figure 9.3).

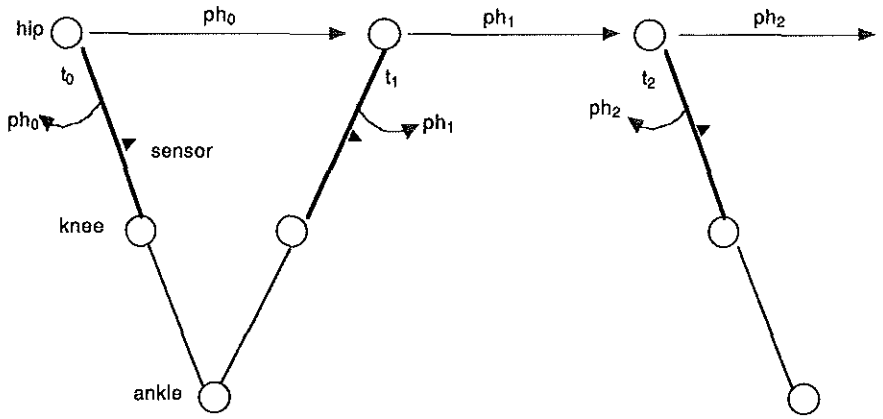

Figure 9.2. Model of the thigh movement during walking (Saunders et al.[18]) at time moments t_0, t_1, and t_2 (t_2 equals t_0), and phases ph_0, ph_1 and ph_2 (ph_2 equals ph_0). The hip joint, represented by the upper circle, is taken as point of reference. The other circles represent the knee and ankle. The accelerometer is indicated by a triangle.

Due to the perpendicular orientation of the accelerometer with respect to the radius, the accelerations due to rotations around the hip ($a_{rot,model}$) can be found by double differentiation of $a_{grav,model}$; this curve is also sinusoidal, but opposite to the curve of $a_{grav,model}$ (Figure 9.3). If both curves are equal in amplitude, they will eliminate each other. During ph_0, ph_1, and ph_2 not only does a rotation occur, but also a translation; this is due to the inverse pendulum movement of both legs and to the fact that the thigh is studied and the rotation axis is round the ankle. Theoretically, this translation could be uniform. In the literature, however, it is reported that the fastest translational speeds take place in or around double support phases, while the translational speed is slower in the phases in between.[10,12,13] Based on an assumed sinusoidal curve of the translational speed, with the highest translation speeds at t_0, t_1, and t_2, and ignoring the effect of thigh angle, the curve of the translational acceleration can be modelled ($a_{transl,model}$).

It is clear that the leg is not a perfect (inverse) pendulum, and that Figure 9.2 is only a rough representation of the actual walking pattern. For example, the leg is not a rigid link due to angular movements in the knee and ankle joint. Furthermore, the amplitude and timing of accelerations is indicated roughly. It may be expected that especially these facts will lead to discrepancies between the measured signal and signals presented in Figure 9.3.

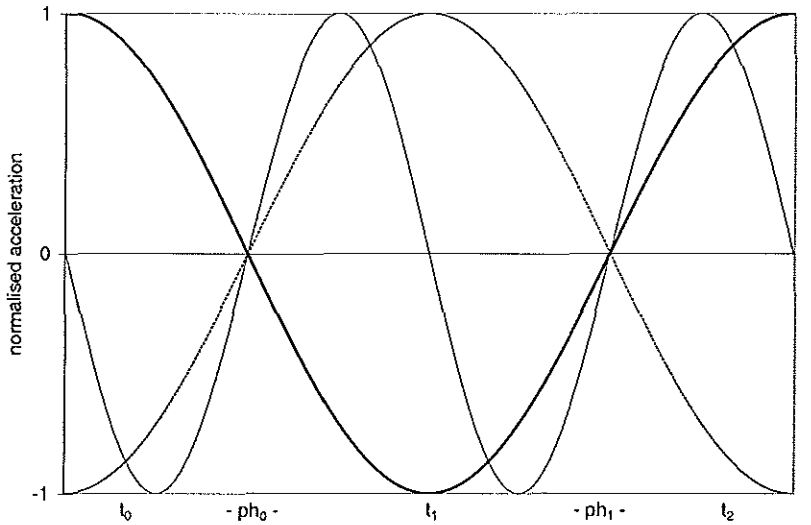

Figure 9.3. Graphical representation of the modelled, normalised accelerations due to inclination ($a_{grav,model}$ —), rotation ($a_{rot,model}$ ---), and translation ($a_{transl,model}$ —).

Methods

Subjects

Six subjects were included. The inclusion criteria were: male, and aged between 18 and 50 years. Subjects with diseases or impairments of the locomotor system or other diseases or impairments with possible consequences for the movement pattern, were excluded. The mean age of the subjects was 27.3 (range 24-42) years, the mean height 1.83 (range 1.76-1.89) meter, and the mean mass was 76.3 (range 73-93) kg.

Protocol

The measurements of this part took place in a gym. An oval trajectory of ± 30 meters was set out. The measurements were done at one straight part (± 12 meters). The subject first passed a light gate, then entered the measurement field of the optoelectronic system, left that field, and subsequently passed a second light gate (distance between gates: 8 meter). Accelerometric, optoelectronic, and footswitch data were obtained. To allow correction for angular attachment differences between optoelectronic markers and accelerometer, the subject sustained three thigh positions without movement for 15 seconds. Then the subject walked the trajectory several times to get used to the instrumentation and setting of the study. The comfortable

walking speed and comfortable stride frequency were calculated over three trajectories by means of the light gates and a stopwatch. A metronome was set at the calculated comfortable stride frequency and, after getting used to the stride-pacing by means of the metronome, the subject walked the trajectory 6 times at comfortable speed. The measurements were then repeated with slow (–20% of the comfortable stride frequency) and fast (+15% of the comfortable stride frequency) speed. The protocol (with the same stride frequencies) was repeated with accelerometers and optical markers on an aluminium stick (25-35 cm, depending on leg length), fixed with Velcro at the front of the thigh. The position of the accelerometer was the same as during the measurements with the accelerometer skin-fixed. The measurements with the stick were aimed at insight in the influence of deformation of the thigh, and in analytical errors. It was assumed that sensors and markers attached on an aluminium stick are not, or to only a small extent, vulnerable to acceleration due to deformation.

In the second part of the study measurements were performed without optoelectronic instruments. The measurements of the first part of the study were repeated under the same condition (intra-subject variability), while walking on a carpet or a treadmill (factor: walking surface), and with the accelerometers displaced 2 cm distally and 2 cm medially (factor: sensor location). The speed of the treadmill (Biodex Rehabilitation TreadMill) was the same as the comfortable walking speed in the gym. In all measurements a metronome imposed the stride frequency.

Instruments

Two piezo-resistive IC-3031 accelerometers were attached to each thigh (see Theory). The sensors were sensitive in anterior-posterior or X-direction while standing (for reference system see Winter[27]). The accelerometers were placed with their sensitive axis as close as possible to the transverse and sagittal plane while standing; a deviation of ± 15 degrees was allowed. The sensors were fixed with double-sided tape on Rolian Kushionflex, which in its turn was attached on the skin. Membrane footswitches were placed in the shoe, under the heel, ball of the big toe, and big toe itself. A light-sensitive sensor was placed on the clothes of the subjects. All these sensors were connected to a Vitaport2™ recorder, which was worn in a belt around the waist, and all signals were AD-converted with 100 Hz.

Optical markers were placed on the trochanter major (m_1), on the lateral epicondyle of the femur (m_3), and just beside the accelerometer (m_2) (Figure 9.4). In the measurements in which a rigid aluminium stick was used, the markers were placed at the top (m_1), middle (m_2 and accelerometer), and end (m_3) of that stick. To allow calculation of absolute angles, two markers were placed horizontally on the ground.

The co-ordinates of the optical markers were measured with a MacReflex™ system, with a sample frequency of 50 Hz. Both systems were synchronised using an infrared light in the measurement field of the optoelectronic system, and a photoflash that was registered by the light sensor on the clothes of the subject. The data from both the Vitaport and MacReflex were downloaded on a Macintosh PowerMac 7600.

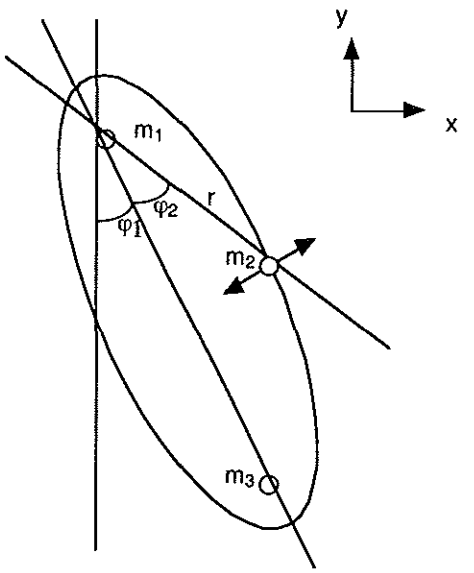

Figure 9.4. Schematic lateral representation of the thigh and the attached optoelectronic markers and sensors (m_1=proximal marker; m_2=sensor marker; m_3=distal marker; φ_1=angle between line m_1-m_3 and the vertical; φ_2= angle between line m_1-m_2 and m_1-m_3; r=distance m_1-m_2. The arrows represent the sensitive axis of the accelerometer.

Data analysis

The light gates - their flash registered on the Vitaport recorder - were used to calculate speed. The synchronisation procedure of the Vitaport and MacReflex was used in the selection of corresponding cycles. Each time the subject entered the measurement field of the MacReflex system, one cycle was selected and analysed. The start and end of each cycle was determined by the heel strike of the right foot (HSR), determined by means of the footswitches. The footswitches were also used for the determination of heel strike left, toe off right, and toe off left (HSL, TOR, and TOL, respectively).

In general, 6 cycles were analysed per condition. Mean cycles were calculated after duration normalisation of the 6 cycles. In the same way mean cycles were calculated

over all subjects. For normalisation and calculation of mean curves MatLab procedures were used.

From the optoelectronic markers, the different types of acceleration were calculated. Based on marker m_1 and m_3, angle φ_1 of the thigh with the vertical could be calculated with WingZ for MacReflex. After correcting φ_1 for the differences in angular attachment with the accelerometer, $a_{grav,sens}$ could be calculated.

$$a_{grav,sens} = 9.81 \cdot sin(\varphi_1)$$

By double differentiation of the position of optical marker m_2 (the 'sensor marker', with WingZ for MacReflex), the acceleration in (anterior-posterior) X-direction ($a_{m2,x}$) and (vertical) Y-direction ($a_{m2,y}$) was calculated, followed by the calculation of $a_{inert,sens}$.

$$a_{inert,sens} = a_{m2,x} \cdot cos(\varphi_1) + a_{m2,y} \cdot sin(\varphi_1)$$

By adding $a_{grav,sens}$ and $a_{inert,sens}$, the reconstructed accelerometer signal ($a_{recon,sens}$) could be calculated.

$$a_{recon,sens} = a_{grav,sens} + a_{inert,sens}$$

This reconstructed signal should be equal to $a_{meas,sens}$.

To obtain additional insight in $a_{inert,sens}$, this component was further decomposed, the hip marker taken as reference point. The translational acceleration component of the accelerometer could be calculated by double differentiation of the position of optical marker m_1 in X- and Y-direction (with WingZ for MacReflex).

$$a_{transl,sens} = a_{m1,x} \cdot cos(\varphi_1) + a_{m1,y} \cdot sin(\varphi_1)$$

The rotational acceleration component could be calculated according to:

$$a_{rot,sens} = (\ddot{\varphi}_1 \cdot r) \cdot cos(\varphi_2) + (\dot{\varphi}_1^2 \cdot r) \cdot sin(\varphi_2)$$

The translational and rotational acceleration components together should be equal to $a_{inert,sens}$. Discrepancies between the translational and rotational acceleration on the one hand, and $a_{inert,sens}$ on the other, may be the result of acceleration due to deformation (a_{deform}). The influence of $a_{deform,sens}$ was also assessed by comparing the measured signal from the stick-fixed condition with the one from the skin-fixed condition.

φ

Agreement between signals was visually examined, and operationalised by means of the root mean square (RMS).

$$RMS = \sqrt{\Sigma(a_{meas,sens}(i) - a_{recon,sens}(i))^2 / N}$$

with $a_{meas,sens}(i)$ = measured acceleration, ith sample, and $a_{recon,sens}(i)$ = reconstructed acceleration, ith sample, and N = number of samples. The RMS value was also used to assess the effect of intra-subject and inter-subject variability, walking speed, walking surface, and sensor attachment. In that case a measured signal was used in the equation, instead of $a_{recon,sens}$.

With the exception of the MatLab and statistics procedures, all calculations took place on the Macintosh. Differences in RMS values between conditions were tested with the Wilcoxon test (SPSS 7.5 for Windows).

Results

General

The mean comfortable speed was 1.3 (range 1.1-1.4) $m.s^{-1}$, the mean stride frequency 51.5 (range 48.5-55) per minute, and the mean stride duration 1.15 (range 1.07-1.21) s. When walking slow, the mean walking speed was 1.0 (range 0.9-1.2) $m.s^{-1}$, and the mean stride frequency was 41.5 (range 39-44) per minute. When walking fast, the mean walking speed was 1.5 (range 1.3-1.7) $m.s^{-1}$, and the mean stride frequency 59.5 (range 56-63.5) per minute.

Description of the accelerometer signal

Figure 9.5 shows from one subject a typical example of a measured accelerometer signal of one gait cycle during comfortable speed. The acceleration curve is generally characterised by relatively high frequency components with regard to the movement frequency, and low amplitudes. Despite the considerable variability between subjects (see Table 9.1), twelve peaks were found in almost all individual curves, and also in the overall mean curve (Figure 9.6). Most striking is the positive peak (P12) at the end of the cycle (mean: 97% of the cycle; range between subjects: 96-98%). After heel strike (0%), three negative peaks (P1: 4 (2-6)%, P3: 10 (9-11)%; and P5: 16 (14-18)%), and two positive peaks (P2: 7 (5-8)%; P4: 14 (12-15)%) occurred. Generally, P3 is most clear and most negative, while especially P5 is sometimes less pronounced. From ±25 to ±40% of the cycle the curve is rather flat, with small negative accelerations. From ±40 to ±60% two smoothed positive peaks can be seen [P6: 45 (43-47)%; P8: 56 (54-58)%]; with the negative peak P7 in between: 50 (48-53)%. Then the curve crosses the 0 line (±60%), followed by a negative peak [P9: 63 (62-65)%]. The curve becomes less negative [P10: 69 (66-72)%], after which the signal gradually becomes more negative with a, sometimes unclear, negative peak [P11: 89 (84-93)%] just before the positive peak (P12) before heel strike.

The same pattern can roughly be seen at slow and fast speed, although the amplitudes are generally smaller (and sometimes absent) in slow gait and generally higher in fast walking speed (Figure 9.6a-c).

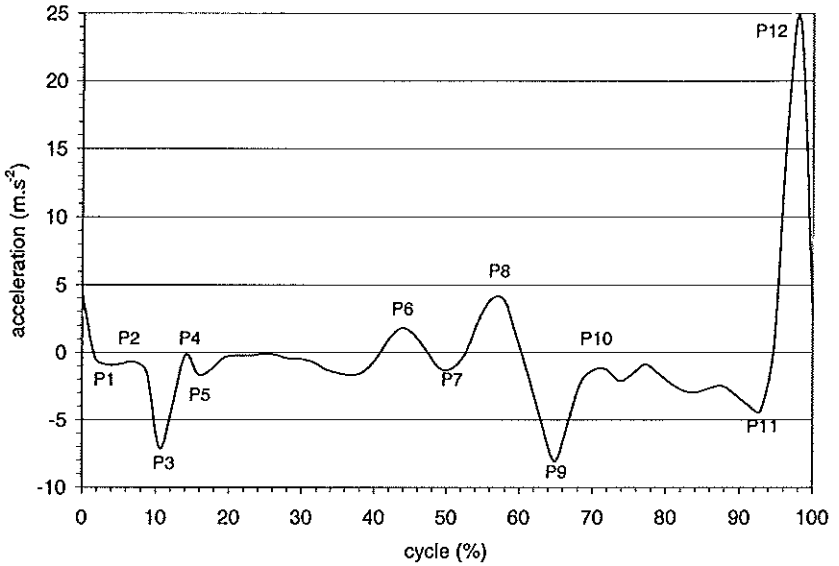

Figure 9.5. Typical example of the measured acceleration signal of one gait cycle at comfortable speed. Peaks generally found in the signals are indicated and consecutively numbered (P1-P12).

Measured and reconstructed accelerometer signal

Visual analysis showed high agreement between the measured and reconstructed accelerometer signals, although agreement decreased with increasing walking speed. This was reflected in the RMS values: at comfortable speed the mean RMS value was 2.11 (range 1.24-4.13) $m.s^{-2}$, at slow speed the mean RMS value was 1.27 (range 0.76-2.28) $m.s^{-2}$, while at high walking speed the mean RMS value was 2.41 (range 1.27-3.69) $m.s^{-2}$. Compared to the mean RMS values at several conditions (Table 9.1), these RMS values indicate again a relatively high agreement.

The RMS values of the measured and reconstructed accelerometer signal when using the aluminium stick were significantly lower (P=0.03) compared to the skin-fixed condition. At comfortable speed the mean RMS value was 1.10 (range 0.82-1.46) $m.s^{-2}$, at slow speed the mean RMS value was 0.70 (range 0.45-0.95) $m.s^{-2}$, while at high walking speed the mean RMS value was 1.58 (range 0.99-2.55) $m.s^{-2}$.

Figure 9.6a-c shows the overall mean curves of the reconstructed and measured data in normal, slow and fast speed, and demonstrates the high agreement between the measured and reconstructed curves. Small differences can be seen, e.g. peak P1 is less pronounced in the reconstructed signal (all speeds); peak P5 is not present in the

reconstructed signal (all speeds); and peak P12 is less pronounced in the reconstructed signal (normal speed).

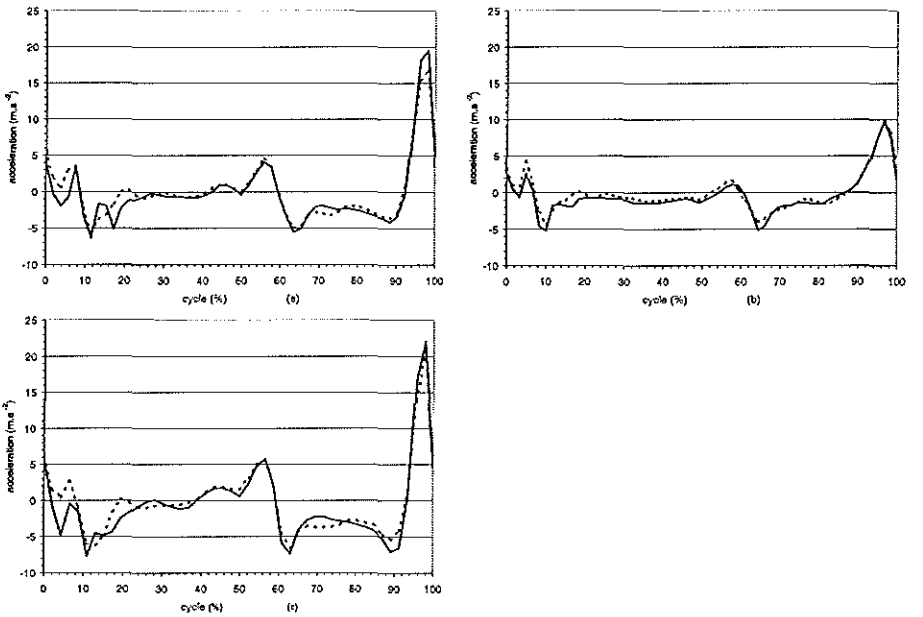

Figure 9.6a-c. The overall mean curve of the measured (—) and reconstructed signal (----) at (a) normal, (b) slow, and (c) fast speed, with markers and accelerometer attached to the skin. Root mean square values are 1.23, 0.66, and 1.46 $m.s^{-2}$, respectively.

Components of the accelerometer signals

The gravitational component ($a_{grav,sens}$) is directly associated with the angular position of the thigh, and is roughly characterised by a sinusoidal curve (Figure 9.7a). However: (1) the positive part of the curve is extended, and contains a dip; and (2) the negative part of the curve is asymmetrical and skewed. The gravitational acceleration component ($a_{grav,sens}$) will contribute positively to the measured acceleration signal from about 0 to 30%, and from about 70 to 100% of the gait cycle. Although $a_{grav,sens}$ thus will influence the amplitudes of $a_{meas,sens}$, the shape and peaks of the curve are mainly determined by $a_{inert,sens}$.

Both $a_{transl,sens}$ and $a_{rot,sens}$ appeared to contribute to $a_{inert,sens}$, and these components were not independent from each other (Figure 9.7b). The curves are roughly

opposite in direction to the curve of a_{grav}, with the exception of the moment of peak P12.

The translational and rotational components together generally strongly resembled $a_{inert,sens}$. Differences found mainly concerned the amplitude of the signal. In the same way the accelerometer signal measured in the stick-fixed condition strongly resembled the signal measured in the skin-fixed condition. The RMS values of the measured signal with the sensor mounted on an aluminium stick compared with the measured signal with the sensor on the skin, were not significantly different from the RMS values expressing intra-subject variability (Table 9.1). Additional visual analysis showed that the differences that were present were not systematic, i.e. they were only present in some subjects.

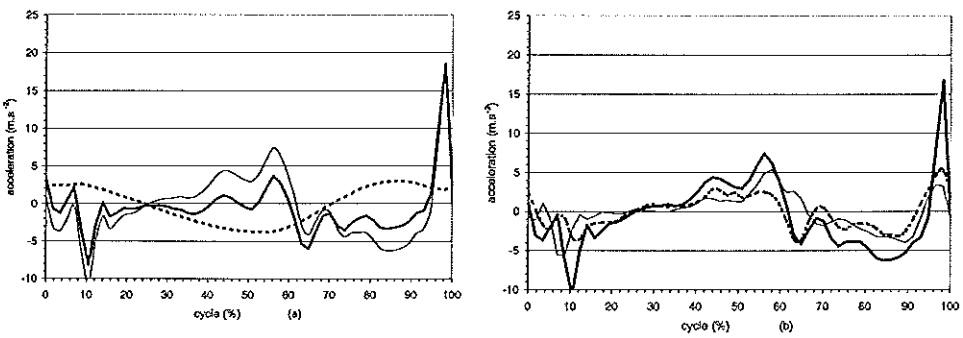

Figure 9.7a,b (a) Typical example of the reconstructed curve (—) at normal speed, decomposed into the gravitational (---) and inertial component (—); (b) the inertial acceleration component (—) decomposed into the translational (---) and rotational (—) component. Markers and accelerometer were attached to the skin.

Temporal events

Overall, during comfortable speed, the HSR occurred at 0% and 100% of the gait cycle, TOL at 11.7 (range 9-13) %, HSL at 50.1 (range 49-51) %, and TOR at 61.3 (range 60-62) % of the gait cycle. During slow and fast walking speed, these percentages are almost the same: TOL occurred at 12% and 11.1%, HSL at 50.2% and 50.0%, and TOR at 61.6% and 60.9%, respectively. These temporal events were associated with events in the measured acceleration curve: HRS took place just after the most pronounced acceleration peak in the curve (P12), while TOL occurs at P2. HSL is associated with a negative peak (P7); and TOR occurs simultaneously with a clear transition from acceleration to deceleration (between P8 and P9).

Influence of factors

Table 9.1 shows the mean RMS values and their range between measurements. When the test-retest RMS values were taken as reference, then the RMS values of the inter-subject variability, of walking slow versus walking comfortable, of walking fast versus walking comfortable, and of displacement of the sensor 2 cm medially, were significantly higher. The RMS values for walking on the carpet were significantly lower (all P = 0.03).

Table 9.1. The mean Root Mean Square (RMS) values with their ranges between measurements for the several conditions.

Factor	Mean RMS value (range)
Subject variability	
Intra-subject	1.88 (1.10–2.66)
Inter-subject	3.98 (2.45–5.65)
Walking speed	
Comfortable versus slow	3.52 (2.50–4.39)
Comfortable versus fast	2.92 (2.57–3.57)
Walking surface	
Carpet versus normal	1.31 (0.86–2.08)
Treadmill versus normal	2.03 (1.85–2.33)
Sensor attachment	
2 cm medial versus standard	3.09 (1.49–5.58)
2 cm distal versus standard	1.75 (1.58–2.00)
On stick versus on skin	2.38 (1.71–3.85)

Discussion

General

The mean comfortable, self-selected walking speed was 1.3 $m.s^{-1}$, which is in agreement with the comfortable walking speed of male subjects in this age group found in literature for male subjects of the same age group.[16] We wanted to standardise the slow and fast speed, to facilitate inter-subject comparison. Because walking speed is difficult to standardise in a free trajectory, we chose a metronome

to standardise stride frequency. Although most subjects had to get used to it, this method generally was appropriate.

Characteristics of the signal

The accelerometer signal measured during a walking cycle can be described as a signal with relatively high frequency components with regard to the movement frequency, and with relatively small amplitudes during the major part of the cycle. These low amplitudes are, in part, caused by the fact that the gravitational and inertial components make opposing contributions to the measured signal.

Despite the considerable variability between persons, 12 peaks could be distinguished in almost all subjects. Most clear, in all subjects, is the positive peak (P12) just before heel strike; at this moment both the gravitational and the inertial component have the same direction. The amplitude of this peak increases with increasing walking speed. In the slow walking speed (mean 1.0 $m.s^{-1}$), this peak is still present, although less clear. The presence and characteristics of peak P12, such as amplitude and slope of the flank, may be used in the automatic detection of walking, in selecting separate gait cycles, and in determining stride frequency. However, the reliability of this criterion in very slow speeds or in distorted gait patterns, needs further study.

Components of the accelerometer signal

For the interpretation of the components of the accelerometer signal, data or figures from Lohman,[12] Murray[14] and Murray et al.,[13] Spivack,[22] Sutherland et al.,[23] Whittle,[26] Winter,[27] and Wu[28] were used.

In the pendulum model, the angular position of the thigh was modelled as a sinusoid. The present study shows that the gravitational acceleration $a_{grav,sens}$ is indeed roughly a sinusoid (see Figure 9.7a). However, the peak of the curve is extended and contains a dip (at 95% of the stride cycle) with two small positive peaks on each side (at 85% and 10%), and the curve is asymmetrical and skewed. This curve is generally in correspondence with the curves of hip angle reported in literature. Immediately after heel strike the angle of the thigh, which is already in a flexed position, increases (until about 10% of the cycle). Although this increase is also reported by Murray et al.[13] and Wu,[28] others reported a decrease in hip angle,[12,22,23] or a plateau[26] at this phase. The increase of thigh angle at this phase corresponds with the loading response of the leg, with fore foot strike, flexion of the knee, and toe off of the opposite leg; the increase of thigh angle can be explained by this. Hereafter, the thigh angle gradually decreases in the course of the stance phase, and passes the frontal plane at about 25-30% of the cycle. At 50-55% of the cycle – about or shortly after heel strike of the opposite leg – the thigh angle reverses; most

authors report this point at 50% of the cycle. The start of the flexion movement of the hip (at 50-55% of the cycle) follows the start of the flexion movement of the knee and the plantar flexion movement of the ankle (at 40%). The thigh angle rapidly increases during the swing phase, and crosses the frontal plane at about 70% of the cycle. At about 85% of the cycle the thigh inclination is maximal, and decreases somewhat until shortly (at 95% of the cycle) before the moment of heel strike.
Remodelling the $a_{rot,sens}$ curve after inclusion of these differences, results in a curve with the following subsequent characteristics: (1) a negative peak (about the positive peak following the dip), contributing to peak P3; (2) a relative flat part (by the skewness of the curve); (3) a large and extended positive peak (about the turning point of $a_{grav,sens}$), contributing to peak P8; (4) a negative peak (about the positive peak preceding the dip), contributing to P11; and (5) a positive peak (about the dip), contributing to P12. All these characteristics can be found in the curve of $a_{rot,sens}$ in Figure 9.7b.

The translational and rotational components together strongly resembled $a_{inert,sens}$. Differences found concerned especially the amplitude of the some peaks. This finding indicates that the influence of acceleration due to deformation of the thigh is absent or small. This conclusion is supported by the absence of significant differences in the signal when the accelerometer is attached on a rigid stick. When the present study was started, we assumed an effect of accelerations due to deformation. It was hypothesised that contraction of the quadriceps muscle and/or vibration of the quadriceps mass could be present in the accelerometer signal. Therefore, measurements were also performed with the optoelectronic markers and the accelerometers on a rigid aluminium stick. Visual analysis of the signals of both conditions, and the RMS values showed, however, that the stick-fixed sensors did not provide significant different signals. Although some differences were found in the measured signals from the skin and the aluminium stick, they were not systematic, did not notably affect the peaks as shown in Figure 9.5, and no significant effect was found on the RMS values of walking with the stick compared to the RMS values expressing the intra-subject variability.
The RMS values of the measured and reconstructed signals when the sensor and markers were attached on the aluminium stick were significantly lower than the RMS values from the sensor and marker attached on the skin. This is probably due to the fact that some measurement errors, e.g. shifting of markers, were not present when a rigid stick was used.

Another component of the pendulum model was the translational acceleration around the double support phases. We assumed that at the moment of heel strike the translational speed was maximal, and that this speed curve was sinusoidal. Inman et al.[10] reported that shortly after heel strike the forward speed of the body is greatest, which means an acceleration of the body before, and a deceleration after that moment. The translational movements of the body will be related to the translational movements of the hip, although especially rotations of the pelvis may contribute to discrepancies between them. The acceleration effect before the first double support phase (this phase occurs about 0-10% of the gait cycle), might be seen in Figure 9.6b: $a_{transl,sens}$ contributes considerably to peak P12. The deceleration is less clear, but may contribute to peak P1 and peak P3. The acceleration before the second double support phase (this phase starts at 50% of the gait cycle), may contribute to P6, but continues during the second double support phase. The deceleration in that phase is not visible; the deceleration may be 'overruled' by the acceleration during the second double support phase due to plantar flexion of the ankle and the flexion of the knee, leading to a rotational acceleration (contributing to P8), but also to a forward translation of the thigh. Only just after the end of this phase, a translational deceleration occurs (contributing to P9).
Another event that may play a role in $a_{transl,sens}$ (and therefore in $a_{meas,sens}$) occurs in the first double support phase. In this phase the subject sets down his forefoot and flexes his knee, which leads to a rotation of the thigh, but especially to a translation of the thigh. This may result in a positive acceleration peak. If so, this event will coincide with peak P2, which may possibly be an interruption in an extended deceleration phase. Peak P4 and P5 may be the result of the knee that starts to extend. After this, the translational acceleration will be small, until the moment the push off of the right leg will start.

Accelerometer signals related to temporal events
The timing of temporal events (heel strike, toe off) within a gait cycle in the present study, was in agreement with the timing data found in literature.[11,22,23] Due to the complexity of the accelerometer signals, we have to be careful about drawing conclusions; at this moment we have to speak of associations within the present data.
The heel strike of the right leg is not clearly represented in the signal. But the timing of HSR is clear: HSR is in the lower part of the right flank of P12. The top of peak P12 might be assumed to be the moment of heel strike, but this is certainly not the case. The relative duration between toe off and heel strike reported in literature ranges from 8 to12.5%. In the present study the mean values were 11.7% and 11.3%. The best estimate for the toe off of the leg is provided by the moment that the

measured curve becomes negative after P8. The heel strike of the opposite leg is best estimated by P7; the toe off of the opposite leg by P2. However, further study of the relationship between temporal events and accelerometer signal will be necessary before the accelerometer signals can be reliably used to determine temporal events.

Influence of factors

Inter-subject differences, walking speed, and displacement of the sensor medially appeared to provide significantly higher RMS values than the RMS values based on intra-subject variability. Whether these findings will have serious consequences or not is difficult to assess. Nevertheless, some implications can be discussed. First, the differences between subjects are advantageous when different types of walking are measured, but may limit the use of accelerometer signals in the detection of walking. However, the peaks of Figure 9.5 were present in all subjects, which demonstrates that the accelerometer signals of the persons in this study also have common characteristics. The differences between subjects may also have been the result of differences in walking speed: each subject walked at his self-chosen speed, and walking speed appeared to considerably influence the shape and amplitude of the signal. Walking surface appeared to be a factor of minor influence, although walking on a carpet even decreased the RMS values. The results suggest that the signal obtained whilst walking on a carpet resembles the signal from the first part of the study more than the signals obtained at another moment but under the same conditions. This may be due to an order effect within the protocol. Although differences are reported between walking or running overground and on a treadmill,[15] in the present study no significant and systematic differences were found. On mechanical grounds no differences can be expected, as long as the belt speed is constant.[9] Displacement of the sensor 2 cm distally, and attachment of the sensor on a stick did not significantly affect the signal. Displacement of the sensor 2 cm medially, on the contrary, appeared to be a significant factor. Due to the curvature of the leg, this displacement had a relatively strong impact on the angular position of the sensor with respect to the sagittal plane. Therefore, there is a significant effect on especially the amplitude of the acceleration signal.

Acknowledgements

I would like to thank Peter H. Veltink for his contribution to this article.

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Feasibility of accelerosignals for measurement of physical strain in ambulation

Summary

Due to the development of an Activity Monitor, with its use of accelerosensors and its possibility to measure heart rate, questions arose about the feasibility of accelerosignals to measure physical strain, compared to heart rate. The aim of the study was to study the feasibility of accelerometry in the evaluation of physical strain in walking at different walking speeds and different levels of economy.

Twelve able-bodied subjects performed a walking test on a treadmill with increasing walking speed. After a 6-week period these measurements were repeated, whereas additional measurements were performed during perturbed walking with a brace. Motility (a measure of the variability of the acceleration signal) and Percentage Heart Rate Reserve (%HRR) were calculated. For motility, its relation with the reference measure oxygen uptake, the inter-subject variability of that relation, its sensitivity to changes, and its test-retest reliability were studied. Feasibility was determined by comparing the motility data with the oxygen uptake and heart rate data.

On average, motility, oxygen uptake, and %HRR all increased with increasing speed, although no complete linearity between the measures existed. The pooled r^2 value for the motility-VO_2 and %HRR-VO_2 relation was 0.91 and 0.85, respectively. The motility-VO_2 relation showed lower inter-individual differences, while both relations showed strong individual correlation. The sensitivity to changes in physical strain due to an increase in walking speed, and the test-retest reliability was the highest for motility. The pooled changes in all three variables due to walking with a brace, significantly correlated with each other; now the relation between changes in %HRR and VO_2 showed the highest explained variance (0.66 versus 0.31). The sensitivity to changes in physical strain due to perturbed walking was generally the highest for VO_2 and the lowest for motility. The relation between motility and VO_2 found during increasing speed could not be applied in the perturbed gait condition.

Motility appeared to be a feasible alternative to measure physical strain during walking, with advantages as well as limitations with respect to %HRR. Motility appeared to correlate well with oxygen uptake, and showed low intra-subject and inter-subject variability. On the other hand, especially its response to changes and differences in economy requires further study.

Introduction

Rehabilitation medicine is especially aimed at the functional consequences of disease, trauma, and congenital anomalies. An important area of interest is daily functioning of patients.[8,70] In the performance of daily activities, such as walking, three aspects can be distinguished: quantity (when are activities performed, for how long, how frequent), quality (how are activities performed, e.g. how fast, how stable, how economical), and physical strain (the measurable reaction of the body due to the performance of the activity) (see chapters 1 and 2).[14] The relationships between these constructs can be clarified using a stress-strain-capacity model:[20,44,72] the quantity and quality of activities can be regarded as stressors which, dependent on factors such as the physical work capacity of the person involved, cause a certain level of strain. Although strain can be studied from different viewpoints, the cardiovascular and respiratory systems and the corresponding physical strain have been the focus of several studies of human ambulation.[17,34] Rehabilitation treatment and training of patients such as amputees is often aimed at reduction of physical strain, e.g. by improving walking pattern, movement co-ordination, or physical work capacity.[1,3,18,28,36,39,57,60,67] Therefore, accurate measurement of physical strain is an important tool in evaluating the responses to, and the efficacy of rehabilitation treatment or training.

An Activity Monitor (AM) has been developed to objectively measure quantity, quality and physical strain under natural and daily circumstances.[11,12,13] The AM is an ambulatory instrument based on accelerometry and aimed at the assessment of mobility-related activities. Activities such as standing, sitting, lying, walking, climbing stairs, driving a wheelchair and cycling can be distinguished and measured. Simultaneous to signals of accelerometry, heart rate can be monitored as a measure of physical strain. Unsupervised, long-term measurement of the activities spontaneously performed during normal daily life, as well as measurement of activities during short-term and more standardised protocols outside a laboratory have been made possible. In research on the AM till now, especially the quantitative and qualitative aspects have been studied;[11,12,13,62] in the present study the ambulatory measurement of physical strain in ambulation was the central issue.

Several techniques exist to measure physical strain or energy cost during daily life activities, each with their limitations. Measurement of oxygen uptake can not or not easily be performed ambulatory.[15,27,52,55,58,59] The doubly labelled water method is very costly, can only be used long-term (mostly 2 weeks), and can not be directly related to activities performed at certain moments in time.[6,9] Subjective data from

diaries and questionnaires have serious limitations in reliability and validity.[4,21,24,45,55] Heart rate or ECG – which can also be measured with the AM – have been used to provide data on cardiac and physical strain, overall workload, or energy expenditure.[3,5,30,43,51,59,61] However, some problems exist in reliability and validity.[5,43,55,58] Nevertheless, heart rate has shown to be of value in the ambulatory measurement of physical strain. Accelerometry-based movement sensors have been used to measure the amount and intensity of body segment movements (motility). The relationship between motility-related measures on the one hand, and heart rate, oxygen uptake, or energy expenditure on the other, has frequently been studied and found, although the relationship is usually not unambiguous.[9,22,37,46,47,50,55,65,69] The sensors in these studies are generally attached to the wrist, waist, or ankle. In our studies, motility signals are routinely derived from the accelerometer signals of the AM – with sensors on thighs and trunk – and motility is one of the features used in the automatic detection of activities. In the current study these motility signals are used as an indicator of physical strain. If motility shows to be a feasible measure in the assessment of physical strain, in future research the monitoring of ECG/heart rate may be redundant, or be combined with the motility data.

The assessment of a measure can include the relation with a reference measure, the sensitivity to changes, and the test-retest reliability. Furthermore, if the relations between variables are equal between subjects, this is advantageous from a practical point of view: individual calibration is then not necessary.
To study the first two issues, physical strain has to be manipulated while simultaneously measuring the different variables for physical strain. In the current study two strategies are used: changing walking speed[e.g. 10] and changing economy of walking by perturbation of gait. The economy of walking is suggested to be influenced by walking with and without a brace.[25,42,66] Although both strategies have an effect on physical strain, they may have different effects on the measures involved. The feasibility can be assessed by examining the motility data with respect to the data of the reference measure oxygen uptake, and with respect to the heart rate data. Furthermore, the strength of the motility-oxygen uptake relationship can be compared to the more common relationship between heart rate and oxygen uptake. Because heart rate will be used as reference measure in future studies (see chapter 11), the relation between motility and heart rate will also be studied. Although these measures are aimed to be used in research with patients, in this initial phase examining their feasibility in non-patient subjects is necessary.
The aim of the study therefore was to examine the feasibility of accelerometry to measure physical strain in walking, at different speeds and different levels of economy.

The research questions were:

(1) What is the strength and inter-subject variability of the relation between motility, oxygen uptake, and heart rate, with increasing walking speed?
(2) What is the sensitivity of motility to speed changes, and its intra-subject variability, compared to oxygen uptake and heart rate?
(3) What is the strength and inter-subject variability of the relation between motility, oxygen uptake, and heart rate when gait is perturbed?
(4) What is the sensitivity of motility to decreased economy due to perturbed walking with a brace, compared to oxygen uptake and heart rate?

Methods

Design

The study consisted of two sessions. In session 1 the subjects performed a submaximal walking test on a treadmill; walking speed was increased to induce changes in physical strain. In this way the relations could be studied, their inter-subject variability, as well as the sensitivity of variables. After a 6-week period (session 2), the subjects performed a similar treadmill walking test to determine test-retest reliability. In addition, in this session a second walking test was conducted wearing a knee brace. With the latter test it was suggested to perturb gait and thus influence economy of walking. Test-retest reliability, the sensitivity of the physical strain variables, and the relations between the changes of the three measures, were examined. During all tests, accelerations, heart rate/ECG, and oxygen uptake were simultaneously measured. All measurements were done at the same time in the morning. A few days before the first session, the subjects exercised on the treadmill to get used to walking on the treadmill.

Subjects

Twelve male untrained subjects participated in the study. During one year before the measurements they had participated less than once a week in activities that require a great deal of physical strain,[38] and their job had to have a mean MET value of 3.0 or less.[2] Subjects with disorders or complaints of the locomotor system, hypertension, liver or kidney disorders, influenza, fever or infectious diseases, morbid obesitas (Quetelet index > 40.0 $kg.m^{-2}$), cardiovascular or pulmonary diseases, electrolytes abnormalities, or with anaemia, were excluded from the experiments. Before participation in the study the subjects underwent medical examination and they all signed informed consent. The characteristics of the subjects were: mean age 54.8 (SD 3.4) years; mean height 1.82 (SD 0.08) m; mean mass 82 (SD 13.6) kg. One subject did not finish the study, due to an illness. Furthermore, the motility or heart

rate or data could not always be used for analysis due to technical problems. The subjects were asked to avoid physical strain and to use no more than 2 glasses of alcoholic drinks the day before each session. Twelve hours before the measurements the subjects were not allowed to smoke or to drink coffee.

Tests

Walking on the treadmill

After getting used to the treadmill (Biodex Rehabilitation TreadMill), the subjects were asked to choose their comfortable walking speed. The speed of the belt was adapted according to the subject's preference, with intervals of 0.2 $km.hr^{-1}$. If the comfortable treadmill speed was selected, it was held for 2.5 minute, followed by 2.5 minutes walking with the comfortable walking speed previously selected on the ground. This period was followed by a standardised part: each 2.5 minute the walking speed was increased by 0.8 $km.hr^{-1}$, starting with 0.8 $km.hr^{-1}$, up to 7.2 $km.hr^{-1}$, without running. To get used to the transition of normal speed to slow speed, the first period of the standardised part (0.8 $km.hr^{-1}$) lasted 5 minutes. In the second session, the two highest speeds (6.4 and 7.2 $km.hr^{-1}$) were not performed to avoid transfer effects between walking and walking with brace. Furthermore, between walking and walking with brace a rest interval of minimally 30 minutes was applied.

Walking on the treadmill with brace

Immobilisation of the knee and use of a hinged cast brace is reported to increase physical strain.[25,42,66] To decrease economy, and thus increase physical strain, in the present study the knee was stabilised in extension with a brace consisting of rigid strips, closed with Velcro. Then a standardised test began: each 2.5 minute the walking speed was increased with 0.8 $km.hr^{-1}$, starting with 0.8 $km.hr^{-1}$, up to 5.6 $km.hr^{-1}$ (corresponding to the walking test on the treadmill in the second session). The first period (0.8 $km.hr^{-1}$) lasted 5 minutes.

Instruments and variables

Accelerometry and motility

In this study, accelerometry was conducted with four IC-3031 uni-axial piezo-resistive accelerometers (1.5×1.5×1 cm). The accelerometers were attached as in the standard configuration of the Activity Monitor.[11,12,13] A sensor, sensitive in anterior-posterior direction while standing, was attached to the skin of the ventral side of each thigh, halfway spina iliaca anterior superior and upper side of the patella. The

other two sensors were attached to the skin of the sternum, perpendicular to one another. While standing, one trunk sensor is sensitive in anterior-posterior direction, and the other in longitudinal direction. The sensors were fixed by double-sided tape. The accelerometers were connected to a portable Vitaport1™ data recorder; the signals were AD converted and stored with a frequency of 32 Hz. After the measurements analysis took place by means of the Signal Processing and Inferencing Language (SPIL™).[31] For the analysis described in this paper, all four signals were successively high-pass filtered, rectified and smoothed. The high-pass filtered signal was calculated by subtracting the low-pass filtered derivative (Finite Impulse Response, 0.3 Hz) from the measured signal; smoothing occurred by moving average and downscaling the sample frequency to 1 Hz.
The more 'dynamic' an activity is, the more variable the accelerometer signals, and the higher the acceleration energy of these signals. Therefore, these signals are assumed to have a relation with the intensity of an activity (or 'motility').[10,37,46,55,65] In this study the motility signals of the legs and trunk (expressed in 'g'; 1 g = 9.81 $m.s^{-2}$) were studied separately. 'Motility legs' is motility right leg plus motility left leg, divided by 2; 'motility trunk' is motility of both trunk sensors, divided by two; 'motility body' is motility legs plus motility trunk, divided by two.

Heart rate

The Vitaport recorder was used for the simultaneous measurement of ECG (V5 bipolar lead, according Mason-Likar). Heart rate was calculated from the R-R intervals. The heart rate during a specific speed interval (HRspeed) was derived from the mean heart rate during the last 30 seconds of a speed interval. The resting heart rate (HRrest) was derived from the mean heart rate in minutes 13-14 of the 15-minute resting period at the beginning of each measurement. The maximum heart rate (HRmax) was determined from the maximum heart rate during a maximum bicycle ergometer test. Subsequently, the percentage heart rate reserve (%HRR) was calculated. The %HRR is the task-related heart rate minus the resting heart rate, divided by the maximal heart rate minus the resting heart rate (×100%).[e.g. 16,35,72]

Oxygen uptake

An Oxycon Champion was used to determine oxygen uptake (VO_2). Because the energy cost of walking increases directly with body mass,[33,43,58] and VO_2max depends on body weight,[23] oxygen uptake was converted to $ml.kg^{-1}.min^{-1}$.

Data analysis and statistics

The relationships between changes in motility, oxygen uptake, and heart rate, motility due to increasing walking speed and walking with a brace were studied by

means of regression analysis. Individual optimal linear regression equations were calculated, as well as the linear regression equations over the pooled data with the corresponding explained variance (r^2), standard error of the estimate (SEE), and P-value. By linear regression with random coefficients also a SEE was calculated. The smaller the SEE calculated in this way, in relation to the SEE of the pooled data, the greater the inter-subject variability component within the pooled data. The sensitivity of variables to changes was examined by calculating the standardised difference: the mean difference divided by the standard deviation (SD) of that difference. To allow comparison between variables, only measurements with a complete data set were used for this part of the analysis. The test-retest reliability was assessed by means of a Reliability Coefficient (RC), the standard error of measurement (SEM), and Pearson's correlation coefficient. The RC was calculated according to: between-subject variance, divided by the between-subject variance plus the within-subject variance.
The statistics were done with SPSS 7.5.2 for Windows; only the linear regression analysis with random coefficients was done with SAS 6.12 for Windows. An alpha-level of 0.05 was used to indicate a significant effect.

Results

There were small differences in the motility results when calculated from the legs, trunk, or body: the median individual correlation coefficient with oxygen uptake was 0.93, 0.94, and 0.94, respectively; the pooled correlation coefficient was 0.83, 0.88, and 0.89, respectively. In this results section, only the data based on the body will further be presented.

Relationships – increasing walking speed
On average, motility, VO_2, and %HRR all increased with increasing speed (Figure 10.1). VO_2 increased curvilinear with increasing speed; the %HRR curve was stronger curvilinear, while motility was more linearly related with speed. These results also become apparent in the pooled scatter plots of Figure 10.2a-c.
The explained variance (r^2) of the pooled relation between motility and VO_2 was higher (linear 0.91, quadratic 0.94) than the explained variance of the relation between %HRR and VO_2 (linear 0.85, quadratic 0.86). The inter-subject differences in regression curves were relatively large in the %HRR-VO_2 relationship (SEE decreased from 1.83 to 1.01 $ml.kg^{-1}.min^{-1}$ when linear regression with random coefficients was applied); the inter-subject differences in regression curves were relatively small in the motility-VO_2 relationship (SEE decreased from 1.44 to 1.22 $ml.kg^{-1}.min^{-1}$). The SEE decreased from 9.64 to 5.35 $beats.min^{-1}$ in the motility-

%HRR relationship. The strength, expressed in r^2, of the individual relations between motility-VO_2 (mean 0.94, range 0.90-0.97) and %HRR-VO_2 (mean 0.94, range 0.86-0.99) did not significantly differ. The r^2 values of the individual relations were significantly lower for the motility-%HRR relationship (mean 0.89, range 0.78-0.96).

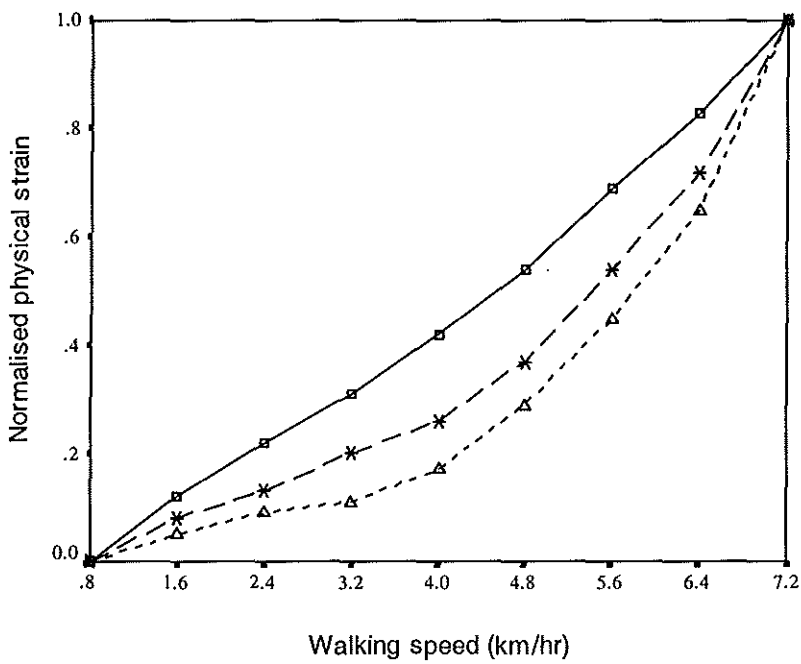

Figure 10.1. Normalised mean curves of motility (—□—), oxygen uptake (VO_2) (-----), and percentage heart rate reserve (%HRR) (····Δ····) plotted against walking speed.*

Sensitivity – increasing walking speed

The sensitivity to changes in physical strain due to an increase in walking speed was generally the best for motility (Table 10.1): the mean standardised difference was 5.11 for motility, versus 2.43 for VO_2, and 2.05 for %HRR.

The test-retest reliability was the highest for motility (RC=0.948; SEM=14.0 g; r^2=0.97), and the lowest for %HRR (RC=0.727; SEM=4.5%; r^2=0.52); for VO_2 these values were 0.869, 0.91 $ml.kg^{-1}.min^{-1}$, and 0.75, respectively.

Parameter – mean (SD)	Walking speed ($km.hr^{-1}$)								
	0.8	1.6	2.4	3.2	4	4.8	5.6	6.4	7.2
Motility (g)									
Session 1-normal (N=9)	47 (7.0)	78 (6.7)	107 (11.6)	130 (12.3)	160 (10.9)	193 (15.5)	231 (16.0)	270 (19.5)	315 (22.8)
Standard. difference-speed (N=9)		5.77	2.91	6.05	3.6	3.83	7.64	7.24	3.87
Session 2-normal (N=6)	48 (7.1)	81 (7.9)	108 (6.7)	131 (9.0)	158 (12.0)	197 (8.9)	241 (12.6)		
Session 2-brace (N=6)	50 (8.3)	82 (6.3)	110 (8.8)	135 (8.5)	167 (11.2)	213 (12.0)	267 (19.2)		
Standard. difference-brace	0.28	0.3	0.33	0.98	0.83	1.49	1.35		
VO_2 ($ml.kg^{-1}.min^{-1}$)									
Session 1-normal (N=12)	5.9 (0.4)	7.2 (0.8)	7.9 (0.9)	9.0 (0.9)	9.8 (0.8)	11.5 (0.6)	14.0 (1.1)	16.7 (1.6)	20.9 (2.2)
Standard. difference-speed (N=9)		1.84	2.68	1.63	1.88	1.78	2.81	3.07	3.71
Session 2-normal (N=6)	6.4 (1.0)	7.2 (1.0)	8.2 (1.0)	9.0 (1.0)	9.9 (1.2)	11.3 (1.0)	13.7 (1.5)		
Session 2-brace (N=6)	6.3 (0.6)	7.6 (0.9)	8.7 (1.1)	10.3 (0.9)	11.7 (0.9)	13.2 (1.0)	16.4 (1.7)		
Standard. difference-brace	-0.36	1.10	0.90	2.46	1.95	2.02	2.26		
HRR (%)									
Session 1-normal (N=12)	14.7 (6.5)	17.3 (7.0)	19.0 (6.6)	20.4 (7.3)	23.2 (6.5)	28.9 (8.9)	36.7 (8.2)	46.6 (10.5)	63.9 (12.6)
Standard. difference-speed (N=9)		0.59	0.89	0.47	2.00	1.29	4.63	2.56	3.98
Session 2-normal (N=6)	15.0 (5.6)	19.4 (6.2)	19.4 (3.5)	21.6 (6.2)	26.8 (5.9)	28.9 (6.4)	35.3 (8.1)		
Session 2-brace (N=6)	13.6 (4.5)	16.8 (5.6)	20.3 (6.6)	24.6 (6.6)	29.3 (7.3)	35.1 (5.8)	44.2 (9.4)		
Standard. difference-brace	-0.44	-1.75	0.27	1.54	0.61	2.29	1.77		

Table 10.1. Data on motility, oxygen uptake (VO_2), and heart rate reserve (HRR) for each walking speed, for normal walking (without brace) and for walking with brace, at the first and second session. The standardised difference (Standard. difference) is provided for changes due to an increase in walking speed, and for changes due to walking with a brace.

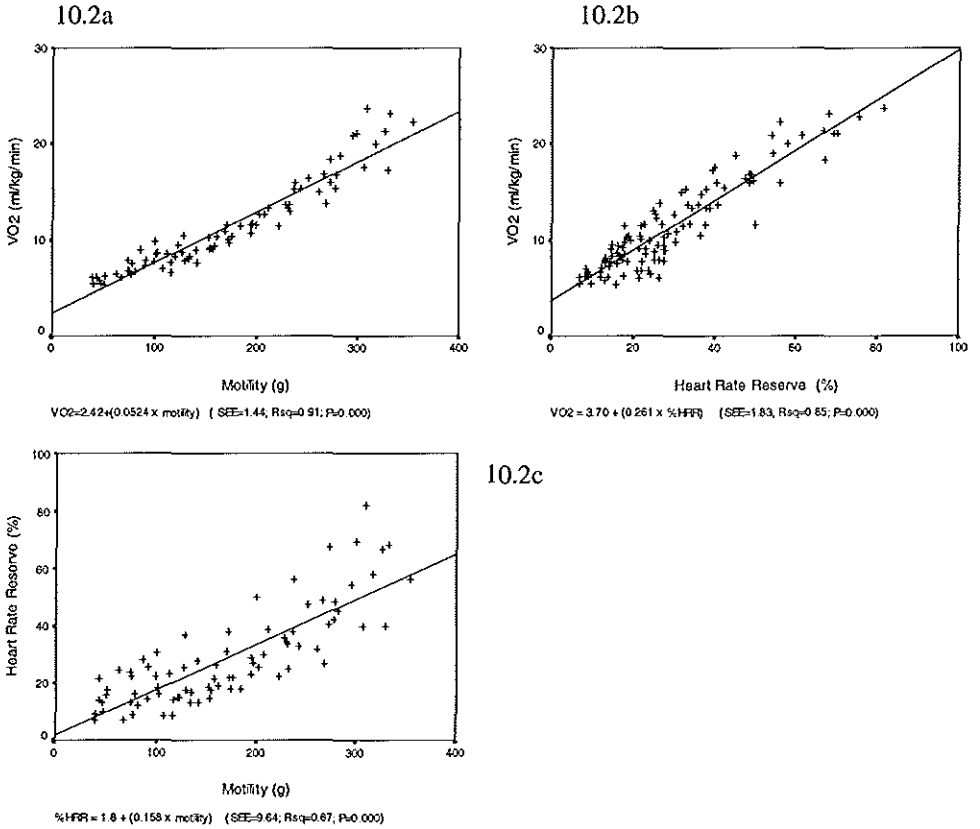

Figure 10.2a-c. (a) Motility plotted against oxygen uptake (VO2); (b) Percentage heart rate reserve (%HRR) plotted against VO2; and (c) motility plotted against %HRR (c). The data points are from all subjects at the first session, and from all walking speeds. The pooled linear curve estimation functions with their standard error of estimate (SEE), explained variance (Rsq), and P-value are provided.

Relationships – perturbed gait

Figure 10.3a-c shows pooled scatterplots of the effect of walking with a brace on motility, VO_2, and %HRR. The changes in all three variables significantly correlated with each other. The pooled explained variance (r^2) between the changes in %HRR and VO_2 was higher (0.66) than the explained variance of the delta motility-delta VO_2 relation (0.31). The SEE of all relations decreased relatively little after linear regression with random coefficients, indicating a small inter-subject component (motility-VO_2: SEE from 1.12 to 0.96 $ml.kg^{-1}.min^{-1}$; %HRR-VO2: from 0.84 to 0.83 $ml.kg^{-1}.min^{-1}$; motility-%HRR: from 4.1 to 3.5%). The individual r^2 values of the HRR-VO_2 relation (mean 0.70, range 0.28-0.93) showed a tendency (P=0.08) to be

higher than the r^2 values of the motility-VO_2 relation (mean 0.50, range 0.08-0.91). For the relation motility-%HRR the mean r^2 value was 0.60 (range 0.15-0.86).

Sensitivity – perturbed gait

At low walking speeds, walking with a brace did not, or had hardly any, influence physical strain. From 3.2 $km.hr^{-1}$ and upwards, however, a clear effect on the physical strain variables could be observed (Table 10.1). The relative change in these speeds was the lowest in motility (+6.9%), compared to the change in VO_2 (+17.3%) and %HRR (+17.5%). The sensitivity to changes due to walking with a brace was the highest for VO_2; the mean standardised difference from 3.2 $km.hr^{-1}$ and upwards was 2.17 for VO_2, 1.55 for %HRR, and 1.16 for motility (Table 10.1).

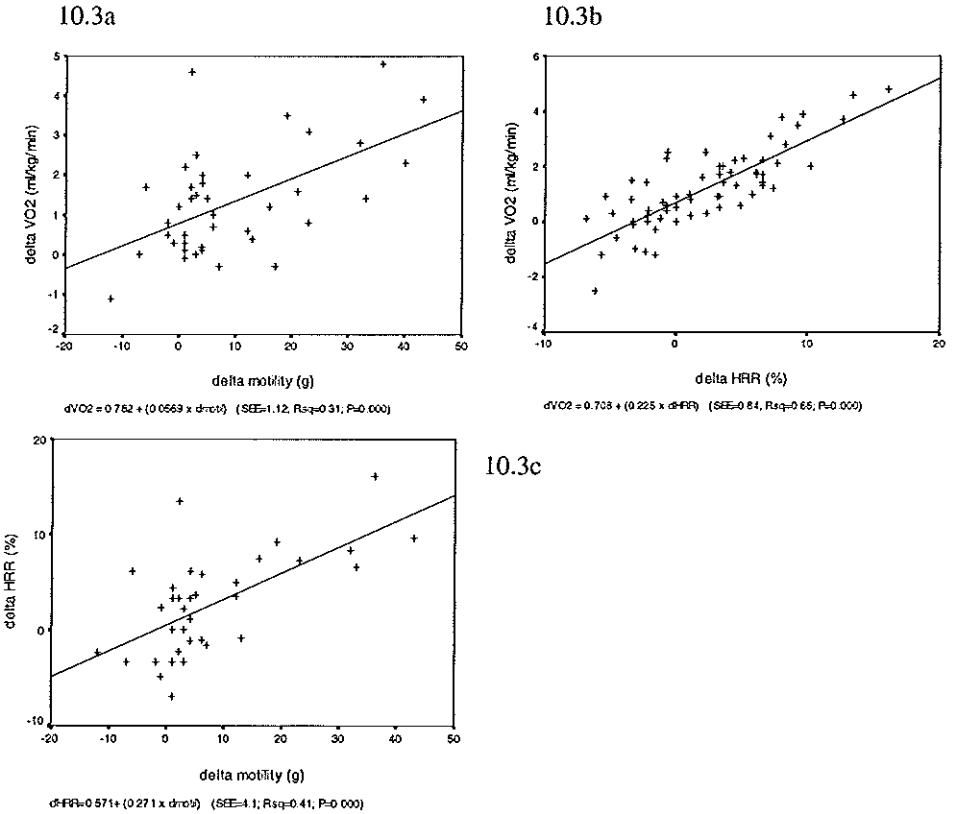

Figure 10.3a-c. Pooled scatter plot of the change in strain variables due to walking with a brace (a) motility (d(elta) motility) plotted against oxygen uptake (d(elta) VO2); (b) percentage heart rate reserve (d(elta) %HRR) plotted against oxygen uptake (d(elta) VO2); (c) motility plotted against percentage heart rate reserve. The linear curve estimation equation of the pooled data, with its standard error of estimate (SEE), explained variance (Rsq), and P-value, are provided.

Discussion

The aim of this study was to examine the feasibility of heart rate measurement and accelerometry to measure physical strain in walking of able-bodied subjects, where strain was manipulated by means of changing walking speed and perturbed gait. Measures were assessed by means of their relation with the reference measure oxygen uptake, the inter-subject variability of this relation, their sensitivity to changes caused by increasing walking speeds and perturbed walking with a brace, and the test-retest reliability of the variables. Physical strain measures are known to suffer from some unreliability, even a reference measure such as oxygen uptake. Therefore, in assessing and discussing the feasibility of the measures examined in this study, the data of the measures were compared with each other.

Relationships – increasing walking speed

As expected from literature, the variables measured in the present study increased with increasing walking speed. Generally, a curvi-linear response of physical strain measures – especially oxygen uptake – on increasing walking speed is reported.[10,29,33,53,56,58,64,71] Although the motility and %HRR are strongly and significantly related to VO_2 in the present study, their response to speed changes was not exactly the same (see Figure 10.1). The response of VO_2 was clearly curvilinear, whereas motility increased only slightly curvilinear with speed, and increased relatively more at lower loads and relatively less at higher speeds (see Figures 10.1 and 10.2a). This non-linear relationship between motility and VO_2 is in agreement with the findings of others,[22,46,47,53] although linear relationships are also reported.[10,50] However, comparison with data from literature is not easy, due to differences in sensor type, number of sensors, location of sensors, walking speeds, activities included in the protocol, and in data processing. In the present study, non-linear curve estimation models (e.g. quadratic) increased the pooled r^2 considerably (from 0.91 to 0.94).

A linear relationship between heart rate and oxygen uptake is described in literature.[5,23,26,32,43,46,49,54,58] However, as can be seen in Figure 10.1 and Figure 10.2, in the present study both parameters were not completely linearly related: at the lower and higher physical strain levels some discrepancy from linearity exists. At the lower physical strain levels VO_2 increased relatively more than %HRR, while at the highest physical strain levels the %HRR increased more. At lower and higher physical strain levels the relation between heart rate and oxygen uptake is reported to be less clear,[5,46,58,69] therefore non-linear relationships may exist at these loads. Despite these deviatons from linearity, curve estimation models other than the linear

one that was used to describe the relation %HRR-VO_2, do not provide higher r^2 values.
The above results have to lead to an even less linear relation between motility and %HRR (see Figure 10.2c); the pooled r^2 value (0.67) is clearly lower than the other two pooled values (0.85 and 0.91). This lower r^2 value is due to the discrepancy from linearity, but even more to the greater variability around the regression line: even after calculating a more optimal (quadratic) regression equation, the r^2 value increased relatively slightly to 0.72.

The relation between motility and VO_2 is characterised by strong individual relationships and low between-subject variability. The variability of the %HRR-VO_2 data around the pooled regression line is for a considerable part due to between-subject differences in regression curves: the SEE of the %HRR-VO_2 relationship based on individual regression lines (1.01 $ml.kg^{-1}.min^{-1}$) is even lower than the SEE of the motility-VO_2 relationship (1.22 $ml.kg^{-1}.min^{-1}$). The variability of the motility-%HRR data around the pooled regression line is due to weaker individual relations, as well as to between-subject variability of the regression equations. The SEE of the motility-%HRR relationship decreased from 9.64 to 5.35 $beats.min^{-1}$ when it was based on individual regression lines. The data on test-retest reliability fit into the picture described in this paragraph: the highest test-retest reliability for motility, and the lowest reliability for %HRR, with the data of VO_2 in between.
The SEE of motility in the prediction of oxygen uptake in the present study can be roughly compared with the standard errors found in other studies,[10,22,46,47,50] although some recalculation to uniform units (e.g. to $J.kg^{-1}.min^{-1}$) has to be done. The SEE found in the present study appears to be the lowest (about 30 $J.kg^{-1}.min^{-1}$), which is close to the SEE found by Bouten et al.[10] (about 30 and 40 $J.kg^{-1}.min^{-1}$), but clearly lower than the standard errors found by Montoye et al.,[50] Meijer et al.,[46] Melanson and Freedson,[47] and Eston et al.,[22] with standard errors up to 130 $J.kg^{-1}.min^{-1}$. However, interpretation of this comparison has to be done with care, because the SEE depends strongly on the activities included in the protocol.

Generally, the data indicate that motility has the advantage of low within-subject and low between-subject variability; the gain of individual calibration – i.e. calculating individual regression equations – is relatively small. The %HRR suffers greater variability, but individual calibration here considerably enhances its reliability. The %HRR is based upon the resting heart rate (HRrest), a speed-related heart rate (HRspeed), and the maximum heart rate (HRmax). HRmax was calculated from a maximal test on a bicycle ergometer, and was not varied in the calculation of the %HRR of the two sessions. HRrest was determined at each session, and appeared to

vary considerably between sessions in some subjects. However, except the data on the test-retest reliability, all data were derived from measurements within one session. The relatively large variability therefore can not be the result of unreliability of HRrest and HRmax, but has to be attributed to HRspeed.

Sensitivity – increasing walking speed

The sensitivity to changes in physical strain due to increasing walking speeds (see Table 10.1) was the highest for motility, with the highest standardised differences. To facilitate interpretation of the data: a standardised difference of about 0.95 or higher can be related to a significant effect when a non-parametric test is used. The %HRR showed the lowest standardised differences, especially at the lower walking speeds. This is due to the relatively smaller increase of heart rate at the speed changes in the lower velocities, and to the larger within-subject and between-subject variability of the %HRR, when compared to motility and VO_2.

Relationships - perturbed gait

As assumed in the Introduction section of this study, perturbed walking due to a brace was another way of increasing physical strain, other than increasing walking speed, considering the variables of interest. Motility depends on the movements of the segments involved, whereas VO_2 and %HRR depends on physiological responses of the body. On theoretical grounds it could be expected that increasing walking speed would affect both the movements of the segments and the physiological response. The effect of walking with a brace, however, was less easy to predict: an effect on physical strain (and thus on heart rate and oxygen uptake) could be expected, but the effect on motility was less certain.

The changes of the three variables due to perturbed walking were significantly correlated with each other (see Figure 10.3). However, in contrast to the situation when walking speed increased, the pooled relation between motility and oxygen uptake was the weakest. Although no significant differences between the three relations were found on the individual level, the r^2 values of the motility-VO_2 relationship tended to be lower than the r^2 values of the %HRR- VO_2 relationship. In the interpretation of the data it has to be noted that the lower correlation coefficients obtained from perturbed walking do not mean that the relation between measures is weaker: the range of data when walking speed increased was considerably larger.

Sensitivity – perturbed gait

Walking with a brace had the largest effect on VO_2 and %HRR, and the smallest effect on motility. Despite the same relative increase of %HRR and VO_2, the increase in oxygen uptake (20.8%) is underestimated when the regression equations

are used which are calculated from the walking test without brace (see Figure 10.2). For the two highest speeds, the mean predicted increase of VO_2 was 14.1% (based on %HRR), and 9.0% (based on motility). The increase in %HRR (20.9%) in its turn was underestimated on the basis of the motility data (10.2%). Apparently, perturbed walking has other effects on the variables involved than changes in walking speed. This corresponds with the finding that oxygen uptake increases with increasing grade of the treadmill, while motility is not or less sensitive for changes in grade.[47,50]
The increase of oxygen uptake corresponds with changes reported in literature: using a hinged cast brace, the submaximal oxygen uptake increased 9%, while heart rate increased 7%.[66] Mattson and Brostrom[42] reported an increase in oxygen uptake of 23% after immobilising the knee, while Fisher and Gullickson[25] reported an increase in energy expenditure of 13%.
Oxygen uptake appeared to be most sensitive to changes in physical strain because of perturbed walking, due to the largest effect of the brace relative to the relatively low variability. Motility, on the contrary, showed the smallest effect relative to the between-subject variability of that effect. The effect of perturbed walking on the physical strain data was most clear at the higher speeds: these data suggest that measurement of physical strain due to changes in economy can best be done at the higher speeds. At the lower speeds the data even tended to be negative, indicating a decreased physical strain during walking with a brace. Apparently, when walking at slow speeds a brace may even increase economy.

General

For a considerable part the present study is related to an ambulatory instrument that has been developed: the Activity Monitor.[11,12,13] Measurement of VO_2 was chosen as reference, and appeared indeed to be a reliable, valid, and sensitive measure, but from the ambulatory background, measurement of VO_2 is not an alternative, although portable respiratory gas exchangers exist.[63]
In the Results section, only the data on body motility were presented. An initial analysis revealed that if motility data of only the legs or trunk were used, only a slightly less strong relation was found with other measures. Apparently, the motility measures are closely related to each other; the added value of the combination of motility signals may be limited in some situations, which was also found or suggested by others.[9,22,69]
In the literature, the relationship between the output of motion-sensitive sensors – mostly accelerometers – and energetic turnover is described. Therefore it could be expected that some relationship between motility on the one hand, and velocity, %HRR and VO_2 on the other, would exist. However, during the walking test we found relatively low variability between subjects, relatively high test-retest

reliability, and a high correlation with VO_2. In the case of perturbed walking, there was an effect on motility at most speeds. Furthermore, motility data from the overground part of the study – not described in the results section – appeared to correspond with the motility data of the treadmill part.

Despite the satisfying results concerning motility, the use of motility also has its limitations. First, the relation between motility and oxygen uptake found during the walking protocol appeared not be valid during perturbed walking. Motility is less sensitive to perturbed gait than oxygen uptake and heart rate. Heart rate and oxygen uptake are physiological variables – indicating, more or less directly, physical strain on some systems of the human body – while with accelerometry movements are registered. This has to mean that different concepts are measured, which may explain some of the differences found between measures. For example, motility may be more associated with mechanical output (e.g. walking speed), while heart rate and oxygen uptake are really physical strain measures. However, different concepts may be associated in some or many circumstances. Therefore, an actually non-physical strain measure may still be a valid measure for certain types of physical strain. Following this line of reasoning it is worth to further explore the potential of motility to measure or predict physical strain. One of the issues of interest will be the combined measurement of heart rate and motility; the combination of these measures is also done or suggested by others,[40,47,55] although also a limited added value of this combination is reported.[22] However, the combined measurement of these variables can be used to enhance interpretation of the mechanical output and physical strain due to walking. Another point of interest is the use of motility and heart rate during activities other than walking, such as activities of daily living. In activities other than walking, some reliability problems may exist with respect to motility. Motility may be prone to external vibrations,[46,55] e.g. while sitting in a car, and physical strain during some activities may be underestimated (e.g. cycling, static work, weight lifting). Generally, the relation between motility and oxygen uptake may depend on the type of activity performed.[e.g. 7,37,46] At the present time it is not clear if the motility measure as used in the present study – different from measures in literature with regard to filter techniques and sensor location – will suffer these problems.

One of the patient groups involved in the research with the AM are persons with an amputation of the lower extremity. In this group the physical strain due to walking is increased due to the amputation and the prosthesis. Furthermore, decreased physical work capacity due to vascular problems and poor physical condition will increase physical strain during ambulation. The relationship between oxygen uptake and heart rate on the one hand, and motility on the other, also need further study in this particular group. Besides experimental research, more theoretical work has to be

done to study the relationship between these parameters, in healthy subjects as well as in patients.

Conclusion

Motility appeared to be a feasible alternative to measure physical strain during walking, with advantages as well as limitations with respect to %HRR. Motility appeared to correlate well with oxygen uptake. Further advantages of motility are the relatively high test-retest reliability and the low variability between subjects. Motility is relative low sensitive to changes in physical strain due to perturbed walking, and different relations between oxygen uptake and motility exists during normal walking and perturbed walking. These findings, and the use of motility in activities other than walking, will need further study.

Acknowledgements

We would like to thank R.J.G. van den Berg-Emons and Th. Stijnen for their contribution to this study and paper.

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Feasibility of ambulatory measurement of prosthetic gait in the early phase of rehabilitation

Summary

Due to previous and foreseen studies with an 'Activity Monitor', some questions arose about the feasibility of measurements with the AM in the early phase of rehabilitation of persons with an amputation. The present study was aimed at the detection of walking and climbing stairs, the reliability of measurements, the sensitivity of measurements to differences and changes, and the potential of motility to predict physical strain.

Ten patients with an amputation of the leg (having their prosthesis less than 2 months) and 10 matched comparison subjects performed, with an interval of one month, the same protocol three times: including comfortable walking speed, climbing stairs, and a test with fixed walking speeds. Signals from accelerometers and electrocardiographic (ECG) signals were monitored. Walking speed, stride frequency, stride length, movement variability, percentage heart rate reserve (%HRR), physiological cost index (PCI), and motility were variables of interest. The test with fixed walking speeds of the second session was used as calibration for the motility-%HRR relationship.

In the comparison group 98% (patient group 95%) of the walking periods was completely correctly determined; this percentage was 48% (patient group 0%) for the climbing stairs periods. At comfortable walking speed, all variables differed between groups. At fixed walking speeds, only %HRR and PCI differed. No improvement over time could be measured in the patient group. Reliability coefficients were generally the highest in walking speed, stride frequency, and stride length, and the lowest in PCI and movement variability. Strong relations were found between motility and %HRR; this relation in the patient group was different from that in the comparison group. The accuracy to predict %HRR from motility was highest in the session in which the calibration equation was determined.

The present study has provided further insight in the potential of ambulatory measurement in the rehabilitation of persons with an amputation of the leg. The detection of walking can be regarded as valid. The ambulatory measurement of gait quality and physical strain variables must be able to cope with a considerable within-subject and between-subject variance. If motility is used as a predictor of physical strain in these patients, individual calibration has to take place.

Introduction

In rehabilitation medicine interest focuses on the activities of patients related to normal daily life. An 'Activity Monitor' (AM) has, therefore, been developed.[6,7,8] The AM is based on long-term ambulatory accelerometry, and aimed at the assessment of mobility-related activities, such as walking, performed during normal daily life or under semi-natural conditions. The aspects *quantity* (which activity is performed, when, for how long, and how frequent), *quality* (how the activity is performed), and *physical strain* (the physical reaction of the body due to the performance of the activity) are being distinguished.
The AM is foreseen to be used in evaluative studies on the treatment of persons with an amputation. Persons with a leg amputation are an important patient group within rehabilitation medicine. In their daily functioning, walking is one of the most important activities, and impairments and disabilities related to walking often exist.[10] These disabilities are usually related to *quality* (e.g. distorted gait),[3,20,21,23,38,39] to *physical strain* (e.g. energetically less economical),[3,15,16,18,24,29,32,38,39] and to *quantity* (e.g. a restricted mobility range). These three aspects, therefore, are important in their treatment and in the evaluation of that treatment. Following previous studies (see chapters 7 and 10), and preceding future studies, three questions had to be answered with respect to the three distinguished aspects of walking (see I, II, III).

I. Detection of walking/climbing stairs
The validity of AM measurements to distinguish activities, was examined in healthy subjects (see chapter 4),[8] in patients with failed back surgery (see chapter 6),[7] in patients with heart failure,[2] in students participating in a psychophysiological study (see chapter 5),[6] and in persons with a trans-tibial amputation, after finishing their rehabilitation treatment (see chapter 4).[8] When using the AM for detecting activities in patients with an amputation of the leg, there might be some factors influencing the validity, such as distorted movement pattern, use of assistive devices, low ambulation speed, and higher levels of amputation. In addition, it can be hypothesised that the influence of these factors will depend on the type of activity and the rehabilitation phase the patient is in. Walking and climbing stairs are determined by pre-set values concerning low-pass (LP)/angular, motility, and frequency features (see chapter 3). In the early phase of rehabilitation of persons with an amputation, the motility might be lower than the pre-set values for walking, and the stride frequency might not be detected due to a too distorted signal, a too low frequency, or too low power of that frequency. For future use of the AM in

patients with an amputation, it is important to investigate the validity of the detection of walking and climbing stairs in the early phase of rehabilitation.

II. Sensitivity to measure changes and differences

Gait quality variables that are aimed to be measured with the AM, are walking speed, stride length, stride frequency, and variability of movement co-ordination. At this moment, the latter two can be derived from the accelerometer signals. Physical strain and economy, in walking usually defined as physical strain per distance, can be derived from ECG or heart rate, which can be recorded simultaneously. The feasibility of measurements of the above variables, based on a specific measurement protocol and using the AM in persons with an amputation, is unknown yet. Feasibility can be determined by test-retest reliability, and the sensitivity to changes (e.g. over time) or to differences (e.g. between groups). Generally, literature lacks data on these issues concerning patients with an amputation: in literature transversal studies about quality and physical strain of prosthetic gait are frequently described, but relevant longitudinal studies have not been found. Furthermore, the AM, accelerometry, and the protocol used in our study, are not used in comparable studies, and therefore no or little reference exists. Finally, especially variability of movement co-ordination, expressed as the variability of phase relations between limbs, has not been studied previously in persons with an amputation. In patients with Parkinson's disease and in stroke patients, for example, this variable has shown to provide relevant information.[12,36]

III. Prediction of physical strain by motility

From each accelerometer signal measured with the AM, a motility signal is derived, which depends on the variability of the measured signal (see chapter 3). Although this signal is initially used for the activity detection, it can also be related to physical strain and walking speed, although the theoretical background of these relations is not yet fully clarified. In a previous study (see chapter 10), the feasibility of the motility variable to predict physical strain was studied in healthy subjects. It was found that motility correlated strongly with oxygen uptake and heart rate when physical strain was manipulated. At the same time, the relation between motility on the one hand, and oxygen uptake and heart rate on the other, appeared to depend on the type of manipulation. The results of that study could not be extrapolated to persons with an amputation, because of possible differences in movement pattern, strain responses, and economy. Therefore, the relation between motility and physical strain has to be studied in this patient group, to be compared with a comparison group, and examined for within-subject and between-subject variance. Because

motility depends on movement and therefore on walking speed, additional study of the relation between motility and walking speed is demanded.
A strong individual relationship between variables indicates that, for the subject involved, a variable can be well predicted from the other. The less between-day variance in the relation, the less the individual calibration has to be repeated. Finally, the less between-subject variance in the relation, the less individual calibration is needed.

The three research questions of the study were:
(1) Can walking and climbing stairs be validly detected in early rehabilitation phases of an heterogeneous group of persons with an amputation?
(2) What is the reliability of measurements of gait quality and physical strain variables, and what is their sensitivity to changes and differences, when using the AM, and in persons with and without an amputation?
(3) What is the relation between motility, physical strain, and walking speed in persons with and without an amputation, and can motility be used to predict physical strain?

Methods

Subjects
Ten persons with amputation and 10 persons without an amputation participated in the study. Patients were recruited from the departments of Rehabilitation of the University Hospital Rotterdam and Zuiderziekenhuis, and the rehabilitation centre 'De Hoogstraat'. All patients had received a prosthesis and had started gait training. The inclusion criteria were: having undergone a uni-lateral trans-tibial (TTA), through-knee (TKA) or trans-femoral (TFA) amputation; being able to perform at least two of the speed levels used in the protocol; having their prosthesis less than 2 months. Exclusion criterion was: diseases and impairments influencing the walking pattern. Table 11.1 shows some characteristics of the persons with an amputation. For each patient a comparison subject without amputation was selected: mass ± 10 kg; height ± 0.10 meter; age ± 10 years. Before the measurements the subjects signed an informed consent form.

Protocol
The subjects were asked to avoid excessive activities and not to drink more than 2 units of alcohol the day preceding the measurements. On the day of the measurement they were not allowed to drink coffee, to smoke, or to perform heavy tasks. All measurements took place in the morning. In a corridor a trajectory of

about 40 meters was created, bordered by light gates. After passing these light gates the subjects turned smoothly, to pass the light gate again to continue for another 40 meter walk.

Table 11.1. Characteristics of the patients (P), and summary data of the comparison persons (C). TTA: trans-tibial amputation; TKA: through knee amputation; TFA: trans-femoral amputation.

Subjects	Gender (m/f)	Age (years)	Height (m)	Mass (kg)	Level
P1	m	66	1.96	81	TTA
P2	m	62	1.78	80	TTA
P3	m	65	1.75	70	TFA
P4	f	83	1.70	69	TTA
P5	m	48	1.68	80	TKA
P6	m	60	1.78	77	TFA
P7	m	76	1.75	75	TTA
P8	m	62	1.82	70	TTA
P9	m	58	1.75	71	TTA
P10	m	66	1.84	96	TKA
P1-P10 mean (SD)		64.6 (9.6)	1.78 (0.08)	76.9 (8.1)	
C1-C10 mean (SD)		61.3 (11.4)	1.77 (0.08)	77.0 (10.2)	

After a 15-minute resting period – to obtain a steady resting state – the subjects walked 2 minutes at their comfortable speed. After this, the subjects climbed upstairs and downstairs on a flight of stairs with 25 steps, again for 2 minutes. Then a resting period of 10 minutes was prescribed. After this, a fixed-speed test was performed, in which walking speed was increased every minute with 0.278 $m.s^{-1}$ (1 $km.hr^{-1}$), starting with 0.278 $m.s^{-1}$. A researcher, using a one-wheeled distance-meter to which a cycle speed-computer was attached, prescribed the walking speed. The walking speed was increased to the subject's maximum walking speed, or until 1.93 $m.s^{-1}$ was reached. The measurements of each session had a mean duration of about 45 minutes. After the initial session, the measurements were repeated two times (session 1, 2, and 3), with an interval of 1 month.

Instruments

Activity Monitor

Four IC-3031 uni-axial piezo-resistive accelerometers (1.5×1.5×1 cm) were used. On each leg, one sensor was attached to the skin at the front of the thigh, halfway spina iliaca anterior superior and upper side of the patella; while standing these sensors were sensitive in anterior-posterior direction. The other two sensors were attached to the skin of the sternum, perpendicular to one another: while standing one sensor was sensitive in anterior-posterior direction and one in longitudinal direction. The sensors were attached such that, with the subject standing, their axes were as close as possible to the vertical or horizontal plane; a maximal deviation of 15 degrees was allowed.

The accelerometers were connected to a portable Vitaport1™ data recorder; the signals were digitally stored on a memory card, each with a sampling and storage frequency of 32 Hz. Analysis took place after the measurements by means of the Vitagraph™ software and by routines written in Signal Processing and Inferencing Language (S.P.I.L.™).[22]

ECG

ECG (V5 bipolar lead, according Mason-Likar) was simultaneously recorded on the same recorder, with a sample frequency of 128 Hz.

Other registrations and data

During the protocol, synchronised videotape recordings were made. Flashes of the light gates were recorded on videotape as well as on the Vitaport recorder, using a photocell attached to the clothes of the subject. Furthermore, manually created light flashes were used to distinguish several parts of the protocol. The study was partly based on the assumption that the subjects improved their quality of walking during the measurement period.

Data analysis

Detection of walking/climbing stairs

The validity of measurements for which the AM is used to classify periods of walking and climbing stairs is studied by running the extended activity detection software as described in chapter 3, and as validated in chapter 7. By comparing the AM classification with the analysis of the videotape recording, the AM output was classified by a researcher as correct, partly false/correct, or false.

Gait quality and physical strain

The accelerometer signals of the legs during walking are characterised by a large, positive amplitude (see chapter 9). This peak was used to detect the beginning and end of subsequent gait cycles. Stride frequency could thus be calculated. For calculating the speed and (mean) stride length during the comfortable part of the protocol, the light flashes were used.

Movement variability was operationalised as the variability of the phase relations between the legs during 10 subsequent cycles. The analysis of the accelerometer signals for movement variability is almost analogous to the method described by Wagenaar and van Emmerik[37] and Slagter et al.[33] The acceleration signals of these 10 cycles were low-pass filtered (Butterworth, second order, 5 Hz), split in separate cycles, and normalised in amplitude and duration. For each leg phase planes were created, based on the filtered and normalised cycles, and their normalised derivatives. From the four corner points of the phase plane, phase angles were derived. The phase difference between the movements of both legs was calculated by subtracting the phase angles of the left leg from those of the right leg. The mean standard deviation was calculated over corresponding samples of the 10 cycles, for each point separately, which were then averaged. The higher this calculated value, the higher the movement variability.

Heart rate was calculated from the R-R intervals. The heart rate during a specific walking speed period (HRspeed) was derived from the mean heart rate during the last 10 seconds of that period. The resting heart rate (HRrest) was derived from the mean heart rate in the fourteenth minute of the resting period at the beginning of a measurement. The maximum heart rate (HRmax) was determined from the formula: HRmax=220–age,[e.g. 28] and assumed to be constant over time.[5,17,30] From these data the percentage heart rate reserve (%HRR) was calculated. The %HRR is the speed-related heart rate minus the resting heart rate, divided by the maximal heart rate minus the resting heart rate (×100%).[11,26,27,41]

In the same way, the physiological cost index (PCI) was calculated, which is used in literature as a measure of economy.[9,13,16,18,32] The PCI (beats/meter) is defined by HRspeed minus HRrest (beats/min), divided by walking speed (m/min).

Motility

The signals of the four sensors were successively high-pass filtered, rectified and smoothed. The high-pass filtered signal was calculated by subtracting the low-pass filtered derivative (Finite Impulse Response, 0.3 Hz) from the measured signal; smoothing occurred by moving average and downscaling the sample frequency to 1 Hz. The more 'dynamic' an activity is, the more variable the accelerometer signals, and the higher the acceleration energy of these signals. The four motility signals

were added and divided by four, to obtain the 'body' motility signal used in this study.

Statistics

Changes in time within the patient group were tested by the Friedman test. Differences between groups were tested with the Mann-Whitney *U* test. Test-retest reliability between the three sessions was calculated by means of a reliability coefficient (RC): the between-subject variance divided by the between-subject variance and the within-subject variance. From the within-subject variance the standard error of measurement (SEM) was calculated. These variances were calculated by means of ANOVA. The Wilcoxon test was used to examine changes due to increasing walking speed.

Linear regression equations, the standard error of estimate (SEE), explained variance (r^2), and P-values were calculated to determine the relation between motility on the one hand, and %HRR and walking speed on the other. By linear regression with random coefficients the SEE was also calculated. The smaller this SEE in relation to the SEE of the standard linear regression equation, the greater the between-subject variance component within the pooled data. Differences between intercept and slope of regression lines were tested with multiple linear regression. Individual regression equations (linear and quadratic) obtained from the fixed walking speed test of the second session (individual calibration) were used to predict %HRR and walking speed at comfortable speeds at all sessions. For all statistics SPSS for Windows (release 7.5.2) was used; only for the linear regression analysis with random coefficients was performed with SAS 6.2 for Windows. An alpha value of 0.05 was taken as level of significance.

Results

Eight of the 10 patients completed the study; two patients were not measured for the third time. All patients could perform the 0.28 $m.s^{-1}$ and 0.56 $m.s^{-1}$ part of the fixed speed test. Four patients were able to walk at least three speed levels at all sessions.

Detection of walking and climbing stairs

In the persons with an amputation, overall 157 analysed walking periods were analysed. In 149 cases (95%), walking was correctly detected. In one subject (P3) with a TFA and walking with 2 crutches, comfortable walking (comfortable walking 1 and 2), walking fast, and walking at 0.28 and 0.56 $m.s^{-1}$ was partly detected as walking, and partly as 'general movement' at the first session. The same occurred at one speed (0.28 $m.s^{-1}$) in subject P6, also with a TFA and walking with 2 crutches.

In subject P5, walking fast and walking with 0.83 $m.s^{-1}$ was partly detected as walking, and partly as climbing stairs. In the comparison group, overall 294 walking periods were analysed, of which 289 (98%) were correctly detected. The errors concerned one time walking at 0.28 $m.s^{-1}$ as walking/general movement, and 4 times walking fast as running.
The results of climbing stairs were less optimal: in the persons without amputation, climbing stairs was well determined in 48%, partly correct in 44%, and false in 8%. In the persons with amputation, the percentages were 0%, 39%, and 61%, respectively.

Gait quality and physical strain variables
When walking at comfortable speed, velocity, stride frequency, and stride length were significantly lower in the patient group than in the comparison group, while PCI, %HRR, and movement variability were significantly higher (Table 11.2). No changes over time were found within the patient group, although a general trend may be seen towards the comparison group. The test-retest reliability (RC) was the highest for stride length, stride frequency, and walking speed, and the lowest for movement variability and PCI.

Table 11.2. Data on gait quality, physical strain, and motility of the patient group (session 1 (t_1), 2 (t_2), and 3 (t_3)) and the comparison group (t_1), during comfortable speed. Differences between sessions in the patient group, and differences between the patient and comparison group at t_1 were tested (P $t_1/t_2/t_3$ and P t_1, respectively). Data on test-retest reliability are provided by the standard error of measurement (SEM) and the reliability coefficient (RC).

Variable	Patient				Comparison			Patient v. comparison
	t_1	t_2	t_3	P $t_1/t_2/t_3$	t_1	SEM	RC	P t_1
Comfortable								
Speed ($m.s^{-1}$)	0.63	0.66	0.73	0.45	1.31	0.07	0.87	0.001
Stride frequency (/s)	0.61	0.63	0.67	0.14	0.89	0.02	0.86	0.001
Stride length (m)	1.00	1.03	1.08	0.88	1.47	0.05	0.96	0.001
Variability (degrees)	14.6	15.6	13.9	0.42	12.0	1.9	0.53	0.04
HRR (%)	42.5	39.3	31.8	0.22	27.6	3.5	0.83	0.02
PCI (beats/m)	0.89	0.80	0.60	0.05	0.32	0.05	0.58	0.001
Motility (g)	0.53	0.56	0.59	0.20	0.91	0.08	0.87	0.001

When walking at fixed speeds, neither changes were found in stride frequency, stride length, and movement variability within the patient group, nor differences between the groups (Table 11.3). Significant differences were measured between groups with respect to %HRR and PCI. For these variables also no changes in time were found within the patient group. The RCs were generally the highest for stride length and stride frequency and, again, the lowest for PCI and movement variability.

Table 11.3. Data on gait quality, physical strain, and motility of the patient group (session 1 (t_1), 2 (t_2), and 3 (t_3)) and the comparison group (t_1), at fixed speeds. Differences between sessions in the patient group, and differences between the patient and comparison group at t_1 were tested (P $t_1/t_2/t_3$ and P t1, respectively). Data on test-retest reliability are provided by the standard error of measurement (SEM) and the reliability coefficient (RC).

Variable	Patient				Comparison			Patient v. comparison
	t_1	t_2	t_3	P $t_1/t_2/t_3$	t_1	SEM	RC	P t_1
0.28 m.s^{-1}								
Stride frequency (/s)	0.46	0.46	0.48	0.78	0.48	0.04	0.91	0.47
Stride length (m)	0.63	0.62	0.59	0.78	0.63	0.05	0.94	0.47
Variability (degrees)	18.8	17.8	18.6	0.78	16.5	2.5	0.01	0.17
HRR (%)	22.4	26.3	34.2	0.47	7.68	3.9	0.48	0.001
PCI (beats/m)	1.03	1.17	0.88	0.14	0.44	0.20	0.03	0.002
Motility (g)	0.29	0.28	0.31	0.37	0.29	0.04	0.77	0.85
0.56 m.s^{-1}								
Stride frequency (/s)	0.57	0.57	0.58	1.00	0.56	0.03	0.90	0.57
Stride length (m)	0.99	0.98	0.97	1.00	1.05	0.05	0.91	0.57
Variability (degrees)	12.9	13.2	16.0	0.47	14.7	2.3	0.38	0.32
HRR (%)	24.7	30.6	36.9	0.11	10.1	4.3	0.60	0.001
PCI (beats/m)	0.56	0.67	0.50	0.11	0.29	0.12	0.36	0.003
Motility (g)	0.40	0.40	0.40	0.78	0.39	0.03	0.60	0.60
0.83 m.s^{-1}								
Stride frequency (/s)	0.69	0.71	0.69	0.63	0.67	0.03	0.91	0.90
Stride length (m)	1.23	1.19	1.22	0.63	1.27	0.05	0.92	0.90
Variability (degrees)	12.6	13.8	12.3	0.37	10.2	1.6	0.35	0.22
HRR (%)	37.2	40.4	44.6	0.37	13.0	4.5	0.76	0.01
PCI (beats/m)	0.57	0.60	0.46	0.11	0.25	0.09	0.52	0.01
Motility (g)	0.55	0.58	0.54	0.37	0.56	0.06	0.36	0.56

Motility versus %HRR

Increasing walking speed by 0.28 $m.s^{-1}$ (fixed walking speed test) caused changes in %HRR as well as in motility, both in patients and comparison subjects (range P-values 0.005-0.043). At these fixed speeds (0.28, 0.56, and 0.83 $m.s^{-1}$), no differences were found in motility data of the patient and comparison group (Table 11.3), contrary to the physical strain measure %HRR. The pooled data of both groups showed significant correlation coefficients between motility and %HRR (Figure 11.1). Because the patients walked only at the lower speeds, the regression line of the corresponding measurements and data of the comparison group (matched comparison data) was also calculated. The between-subject differences in regression curves were relatively large: SEE decreased from 10.2 to 4.7% in the comparison group, and from 15.7 to 9.2% in the patient group when linear regression analysis with random coefficients was applied. The intercepts of the regression line of the patient data and the reduced comparison data were significantly different (P=0.014).

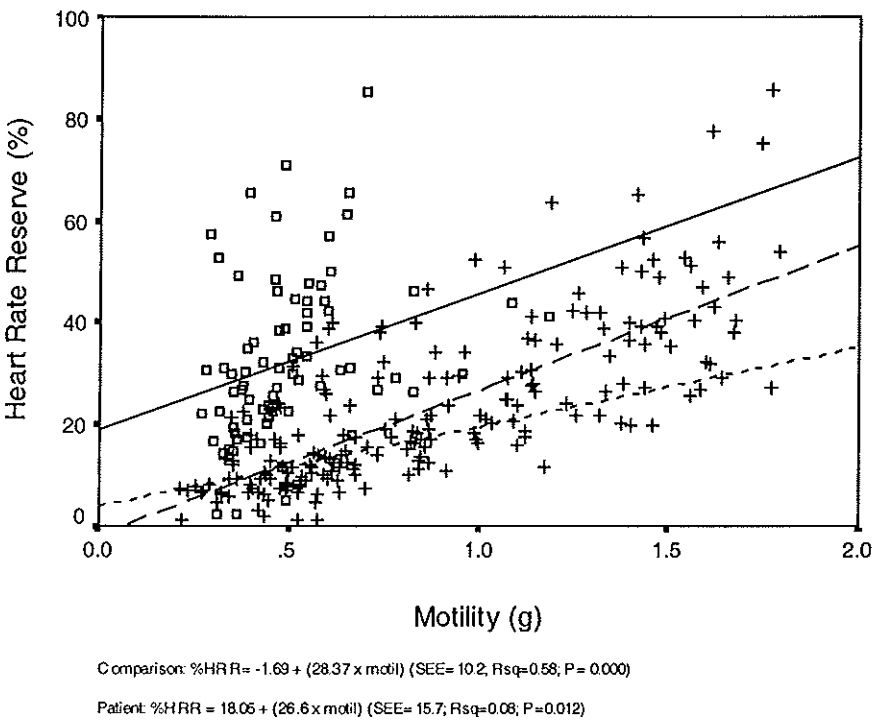

Figure 11.1. Scatter plot (□ patients; + comparisons) and estimated linear curves of motility (motil) against %HRR of the patient data (——), matched comparison data (-----), and all comparison data (— —). The pooled regression equations for the patient group and the complete comparison group, their standard error of estimate (SEE), the explained variance (Rsq), and corresponding P-value are also given.

In the subjects of the comparison group, high individual linear correlations were found: mean r^2 of 0.979 (range 0.911-0.990; SD 0.026). Using a quadratic model for curve estimation resulted in slightly higher r^2 values: mean r^2 of 0.985 (range 0.969-0.999; SD 0.011). From these individual regression equations – obtained from the data of the second session – the %HRR during comfortable walking at different sessions was predicted from motility data. No differences existed between the measured %HRR (mean 25.5%) and predicted %HRR, independent from the type of equation (linear or quadratic) used for prediction. The r^2 values of the relation between calculated %HRR and %HRR predicted by motility and a quadratic model, ranged from 0.04 to 0.83 (Table 11.4). The relation between calculated %HRR and predicted %HRR at comfortable speed, was strongest in the session in which the data for the individual regression lines were obtained (session 2).

Table 11.4. Summary data of predicted %HRR by motility, compared to the %HRR actually measured. The prediction is based on quadratic equations obtained from fixed walking speed test (calibration measurement) in the second session. The relation between actual and predicted %HRR is expressed as explained variance (r^2), P-value, and as the standard error of estimate (SEE). The data concern the %HRR when walking at comfortable speed at the second (t_2), first (t_1), and third (t_3) session.

	t_2			t_1			t_3		
	r^2	P	SEE	r^2	P	SEE	r^2	P	SEE
%HRR by motility									
Comparison	0.42	0.04	5.7	0.26	0.13	7.7	0.08	0.44	10.2
Patient	0.83	0.001	7.4	0.29	0.11	15.2	0.04	0.63	11.8

Motility versus walking speed

The mean motility increased with increasing walking speed. The pooled data showed a strong relationship between motility and walking speed (Figure 11.2). The minor part of the estimation error is due to between-subject differences: SEE decreased from 0.132 g to 0.095 g (comparison group), and from 0.080 g to 0.047 g (patient group) when linear regression analysis with random coefficients was applied. The intercepts and slopes of the regression line of the patient data and the matched comparison data were not significantly different.

In the comparison group, high individual linear correlations were calculated: mean r^2 of 0.979 (range 0.96-1.00, SD=0.012). Estimation of the curve based on a quadratic model slightly resulted in differences: mean r^2 of 0.991 (range 0.98-1.00, SD 0.007).

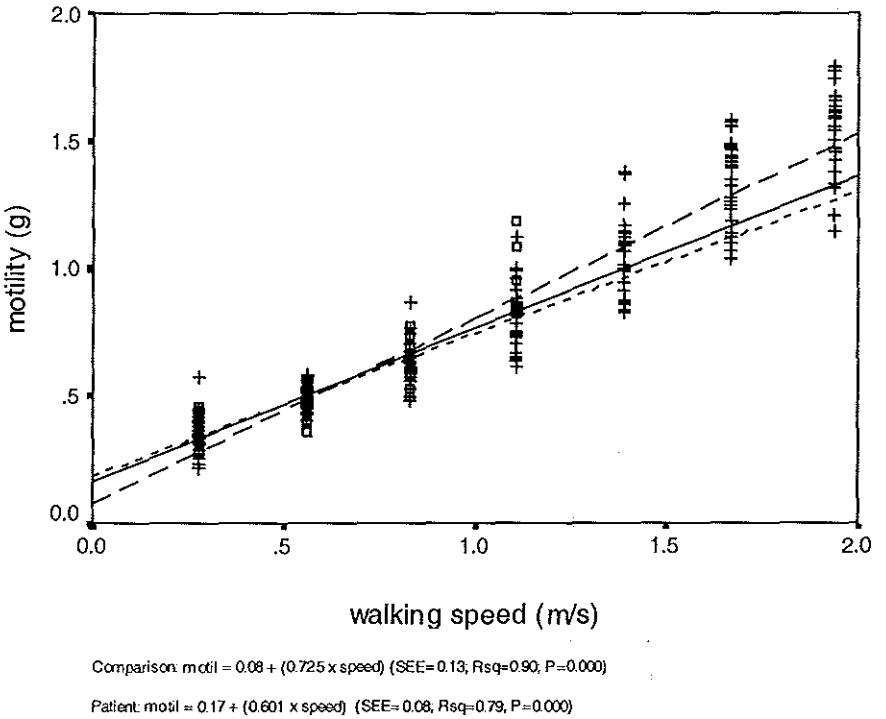

Figure 11.2. Scatter plot (□ patients; + comparisons) and estimated linear curves of motility (motil) against walking speed of the patient data (——), matched comparison data (----), and all comparison data (— —). The pooled regression equations for the patient group and the complete comparison group, their standard error of estimate (SEE), the explained variance (Rsq), and corresponding P-value are also given.

Discussion

Detection of walking and climbing stairs

The results of this study were in agreement with the results found in previous studies (see chapter 7, and van den Berg et al.[2]). In the analysis software, the activity walking is characterised by the level of variability ('motility') of all signals, by cyclicity ('frequency') of the signals, and by low-pass/angular values. It was hypothesised that abnormal types of walking and walking at slow speeds would have

an effect on especially the motility and frequency calculation. In the present study, however, even walking of persons with an amputation at very slow speeds was detected well. Climbing stairs was often detected as walking, although walking was only twice detected as climbing stairs. This is in accordance with previous studies (see chapter 7), in which relatively low sensitivity percentages and high predictive value percentages were found for climbing stairs. The further use of the AM in evaluative and other studies with patients with an amputation seems nevertheless justified, although especially the sensitivity for climbing stairs is relatively low at present.

Gait quality and physical strain variables

When walking at comfortable walking speed, all variables (speed, stride frequency, stride length, %HRR, PCI, movement variability) significantly differed between the patient and comparison group. This is an indicator of the sensitivity of the measurements of these variables, although relatively large mean differences between groups existed.

The differences between persons with and without amputation, with respect to walking speed, stride frequency, stride length, physical strain, and economy, are in correspondence with literature.[3,15,16,18,20,21,23,24,29,32,38,39] However, some authors reported non-significant differences in physical strain at comfortable walking speed, due to lowering of the walking speed.[13,15,16,39] This finding was not supported by the present study.

No significant changes in time were found in the patient group, although tendencies were present at comfortable speed. This may have different reasons. (1) A considerable part of the subjects may not have improved their performance over time, possibly due to the short time interval between measurements. To have a reference for the changes between measurements, questions about ambulation-related performance at the time of measurement, and changes since the preceding measurement session, were asked to patients, their physiotherapists and their rehabilitation specialists. Although a unanimous decline was never found, relatively frequently 'unknown' or 'unchanged' answers were given to questions about changes since the previous measurement, which indicates no or small changes. (2) The relatively large within-subject variance, expressed by the reliability coefficient (RC) and the standard error of measurement (SEM) (see Tables 11.2 and 11.3). Especially for the PCI and movement variability, the RC indicates a large within-subject variance component compared to the between-subject variance. It has to be noted, however, that in our study the between-subject variance was substantial for most variables, contributing to high RCs. Therefore, the SEM values are also provided, as an indicator of the absolute within-subject variance. If these SEM

values are expressed relative to the mean differences between both groups, then the lowest values are found for movement variability, motility, and %HRR. (3) The results on the changes over time in the patient group had low power. Not all patients completed the study and not all patients performed at least three walking speeds at all sessions. Therefore, especially the results on the changes over time from the fixed-speed protocol had lower power. Increasing speed by 0.28 $m.s^{-1}$ (1 $km.hr^{-1}$) appeared to be a large interval for persons with an amputation in the early phase of rehabilitation.

The fixed-speed test showed similar results to the comfortable speed test, that is, no significant changes were found over time in the patient group (even no tendencies), and significant differences were found between both groups with respect to %HRR and PCI. However, gait quality-related variables, such as stride frequency, stride length, and movement variability did not differ between the groups at all three speeds. This suggests that the differences found at comfortable speed were solely a result of differences in walking speed, rather than a result of inherent differences in walking pattern and movement co-ordination between the groups.

Variability of movement co-ordination of subsequent cycles at comfortable speed was significantly higher in the persons with amputation. At the fixed speeds, movement variability showed a U-shaped curve with increasing walking speed in the comparison group. Comfortable walking speed roughly coincided with the lowest point of the curve. Both findings have also been reported in literature.[12,36] At fixed speeds, movement variability did not significantly differ between patients and comparison subjects. The data suggest that the variability of prosthetic gait does not inherently differ from the variability in normal gait, but that the differences found at comfortable speed are due to differences in walking speed.

When the quality- and strain-related variables of this study are used in future studies, the following points have to be taken into account: (1) a considerable part of the patients may not perform better after a 1-month interval; (2) most variables, especially the physical strain-related variables and movement variability, suffer a considerable within-subject variance.

Motility versus %HRR

At comfortable speed, motility, velocity, and %HRR all differed between both groups. At the fixed-speeds, however, motility was not different, contrary to %HRR. This indicates that the relation motility-%HRR is not identical in both groups. This finding was demonstrated in Figure 11.1: the data points of the patients are clearly higher positioned in the graph than the data points of the comparison group. The intercept of the patient curve was significantly higher than the intercept of the comparison regression line.

Motility and %HRR are significantly associated, which is also found by Eston et al.[14] The relationship within the comparison group (with respect to r^2 and SEE) is comparable with the relation found in a previous study (see chapter 10). The regression equations found in these studies are not directly comparable, due to a different scaling of the motility values. In the present study, the relation between motility and %HRR in the patient group was not very strong. This finding may be due to within-subjects variance as well as to between-subject variance. The within-subject variance can be caused by deviations from complete linear or quadratic relationship within measurements, as well as by differences in regression equations between measurements. The data indicate a relatively large between-subject component within both groups: a considerable lowering of the SEE if calculated with linear regression with random coefficients, and high individual r^2 values. This suggests that the variance in Figure 11.1 is to a large extent due to between-subject variance. If motility is used as a strain variable, individual calibration would be necessary. Bouten et al.[4] reported little added value of individual calibration concerning the relation accelerometer output and energy expenditure. In their study, energy expenditure was corrected for sleeping metabolic rate. Between-subject differences in sleeping metabolic rate and, probably more important, larger between-subject differences in economy in the present study, are possible explanations for the discrepancy with the study of Bouten et al.

A strong individual relation should result in a good prediction of the %HRR by motility during walking at comfortable speed. In the session the individual calibration took place (session 2), %HRR was rather well predicted during walking at comfortable speed (see Table 11.4). The prediction of %HRR by motility at other sessions was relatively weak, indicating that the calculated relation between both variables can not be reliably extrapolated to other sessions or days. It has to be noted, however, that the relatively high standard deviation in the patient group at measurements 1 and 3 is strongly influenced by the data of one subject at each measurement (P4 and P7, respectively). Leaving out these data, r^2 changed to 0.68 (SEE 7.5%, P=0.006) and to 0.74 (SEE 6.3%, P=0.13). In these subjects, the resting heart rate data were considerable different from the resting heart rate measured in session 2. Although these differences were also partly present in the task-related heart rate, the %HRR was seriously affected by it. No logical explanations could be found for the changes in heart rate. The most obvious explanation – changes in drug use affecting heart rate, e.g. beta-blockers[1,19,31,35,40] – appeared not to be the reason.

In a previous study (see chapter 10) a rather high level of unreliability was found in the determination and calculation of the %HRR, and the data of the present study suggest the same. The %HRR is based upon the maximal heart rate, the task-related

heart rate, and the resting heart rate. Careful measurement of these components remains important.

Motility versus walking speed

Average motility increased with increasing walking speed. Furthermore, a relatively strong correlation between motility and walking speed was found in both groups, with no significant differences in regression lines. Despite a considerable difference in movement co-ordination and economy the relation between motility and walking speed is not different between the patient and comparison group. This is, in part, in agreement, but also in part in contradiction to the findings described in chapter 10. In that study, walking with decreased economy due to walking with a brace had no or only a small effect on motility, but the effects were significant at higher speeds. Probably, the speeds performed in the present study were too low to show systematic differences in motility. Generally it can be stated, however, that motility is non-sensitive or low-sensitive to changes in economy.

Although the motility-speed relation was studied for the interpretation of the motility-%HRR relationship, additional analysis showed interesting findings. Compared to the motility-%HRR relation, the between-subject variance was relatively small. If motility was used for the prediction of walking speed, rather strong correlations were found between actual and predicted walking speed (range r^2 0.47-0.90, range SEE 0.07-0.12), without a significant session effect. At the individual level high correlation coefficients were found, although not significantly higher than the individual coefficients of the motility-%HRR relationship. The motility-speed relationship was, with respect to the motility-%HRR relation, more stable between patients and comparisons, more stable between subjects, and more stable between sessions.

Conclusion

The present study has provided further insight in the potential of ambulatory accelerometry and heart rate measurement in the rehabilitation of persons with an amputation of the leg. The detection of walking can be regarded as reliable, whereas the detection of climbing stairs is more problematic at this moment, and needs further study. The ambulatory measurement of gait quality and physical stain variables will have to cope with a considerable within-subject and between-subject variance. The results of the examined relationship between motility and %HRR, support the use of motility to predict physical strain. If motility is used as a predictor of physical strain, individual calibration has to take place.

Acknowledgements

We would like to thank P.J. Janssens (*Zuiderziekenhuis*), and B. Kap (*De Hoogstraat)* for their valuable assistance in subject recruitment. We would like to thank also Rehabilitation Centre *De Hoogstraat* for their co-operation and assistance in the measurements of this study.

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12

General discussion and concluding remarks

Relevance of activity monitoring

Within rehabilitation medicine many instruments are used that measure, for example, treatment effects, natural recovery, or decline in physical status. The problem within this field is more likely to be the diversity of available instruments, rather than a lack of instruments. Many of the instruments show a large degree of overlap and are often not validated, or only to a limited extent. Efforts, therefore, should focus on improvement and validation of instruments, instead of focus on the development of new instruments. When new instruments are designed, valid arguments have to be present. The research described in this thesis was, nevertheless, focussed on a new ambulatory instrument that measures mobility-related activities during normal daily life: the Activity Monitor (AM).

In chapter 2 several techniques that are used to measure mobility-related activities were classified, assessed, and discussed. Although chapter 2 was not aimed to advocate the use of activity monitoring, arguments for the development of the AM can be derived from it: the AM measures on activity level, measures aspect(s) of mobility that are not measured by other instruments, and has methodological advantages. Probably the most important characteristic of the AM is that the AM measures objectively what a person really 'does' during daily life. Other instruments usually measure other aspects: what a person wants, can, or what s/he thinks s/he does.

Especially in rehabilitation medicine, however, interest also focuses on 'higher' functional levels, such as complex activities, social participation, and quality of life. The AM measures 'simple activities', as a sort of compromise between practical feasibility and highest functional degree. However, these simple activities are measured during the performance of functional activities in normal daily life; in this way the relevance of the measurements is upgraded. Furthermore, relationships with the higher functional levels are assumed to exist. Future studies will certainly include questions about these relationships. In addition, the activity pattern (or physical lifestyle) of patients is frequently in itself a central issue in treatment programs, and thus a valuable outcome variable.

These arguments should not be confused with the idea that an activity monitor is a new reference method, and that measurement of what a person really does is most important. The AM should be regarded as a relevant and valuable addition to the techniques currently used in rehabilitation medicine. Whether the information provided by the AM is relevant or not in a specific case, depends on the research questions, which in turn depend on clinical issues and questions in rehabilitation. These issues are frequently related to actual behaviour during normal daily life, indicating the added value of the AM in rehabilitation research.

Methodological considerations

From a methodological viewpoint, the AM has some important advantages. Nevertheless, three methodological issues need most consideration.

First, the activity pattern of any given person will differ between days,[4,15,52] as was also shown in chapter 8.[6] For example, workdays or weekend days, day-specific leisure activities, irregular activities, and so on, will cause between-day variability in activities. This variability will increase the number of days during which measurements have to take place to obtain sufficient statistical power. This between-day variability in activity has to be studied before using the AM in intervention studies. The study described in chapter 8, with congestive heart failure patients, is an example of such a study. The between-day variability may also depend on the type of population: subjects with a large physical capacity have a greater range of possible activity levels than subjects with a small physical capacity. The results of the study described in chapter 8 tend to support this assumption. Furthermore, the between-day variability will also depend on the specific variables that are measured. For example, the total duration of lying periods may differ less between days than that of walking periods.

Second, validity of study results will also depend on external factors influencing activity pattern. For example, pre-intervention measurement in winter and post-intervention measurement in summer may elucidate differences in activity pattern that are in fact not due to the intervention.[33] In the design of studies these effects should be considered. The selection of a control group and the planning of measurement will, therefore, be important.

Third, the AM may influence the activity pattern of subjects: patients are aware that they are measured ('reactivity effect' or 'perturbation effect').[20,33] In order to address this issue, several days could be measured, assuming a diminishing reactivity effect (e.g. not using the data of the first day). Furthermore, the instruction given to the patient is of major importance; patients should not be focussed on the measurement of activity level, and the measurement should not be explained as testing the patient's performance.

Practical considerations

A few practical points need consideration. The attachment of the sensors to the body is still not optimal. Our group and others have tested several methods of attachment:[16] e.g. fixing the sensors with tape, elastic bands, elastic underwear, and different types of harnesses. In our experience, taping the sensors is the best method, although problems still exist. We continue to investigate and test different ways of

attaching the sensors; a good method should be comfortable, and allow patients to remove and re-attach the sensors without considerable effect on sensor location and position. The latter issue is important, because the study presented in chapter 9 showed a significant effect on the signal of relatively small changes of sensor position. Furthermore, in the activity detection, especially the detection of climbing stairs will be relatively sensitive to changes in sensor position within a measurement. For other activity categories deviations in sensor location will not immediately affect reliability, but one should aim to have minor deviations within a measurement.

The measurement may cause some discomfort, and patients may find the recorder and the cables somewhat irritating. The system can not be used in a wet environment (e.g. not during bathing or taking a shower). Some find the recorder and/or the cables disturbing while sleeping or (un)dressing, and others dislike being seen wearing the instrument. Again, we are investigating methods to allow patients to easily attach and remove the recorder and sensors themselves. Discomfort will of course decrease when the recorder becomes smaller and lighter.

Generally, the accelerometers functioned properly. However, the mass within the sensor can get jammed, especially after accelerations exceeding the measurement range of the sensor (e.g. after accidentally dropping the sensor).[54] Furthermore, a small change in offset and calibration factors was sometimes noticed. In all measurements described in this thesis, the sensors were calibrated on each measurement day. Due to technological improvement, this procedure needs to be done less frequently nowadays, which enhances the practical feasibility.

Another important aspect of practical feasibility is the costs of the instrument and measurements. The accelerometers, the data recorder, and the analysis software mainly determine the cost of the AM. The production of the RAM system is planned. On the short term, the costs of a complete system will be about the 24,000 Dutch guilders (about US$ 12,000). The penlight batteries mainly determine the additional cost per measurement. Using a PCMCIA flash card instead of a hard disk considerably lengthens the measurement time for each battery set.

Ethical considerations

The AM could be regarded by some as 'Big Brother is watching you', as an invasion of one's privacy. Research with the AM has to be subject to the usual guidelines of medical and ethical committees. Patients can not be forced to participate in studies, and they have to be well informed about the study and the consequences of participating. The results of the measurement may not necessarily have consequences for their treatment. In the instruction it should be clear that it is not the

patient who is tested, but in fact the treatment. Finally, it should be noted that the AM provides a specific set of mobility-related activities. The output is no more and no less than these output categories.

Comparison with other activity monitors

In the general introduction and validation chapters, other systems have been reported and discussed. Generally, comparison of the AM with these systems is difficult. Instruments discriminate different activity categories, are used for different aims, or validity is determined following a different or unknown method. The number of instruments that resemble the characteristics of the AM is small. In literature, some systems are reported which are aimed at a similar set of activities and which are based on accelerometry.

Walker et al.[50] reported a validation study of an activity monitor based on mercury switches and accelerometers. Posture and number and vigour of steps was recorded. Validation was studied in terms of steps counted; validation of body positions was not reported. In the same manuscript, the relationship of activity to disability was explored.

Kiani and colleagues[27,28] and Groeneveld et al.[23] described the AMMA system; an activity monitor with the analysis initially based on a neural network. Although a neural network may be a powerful tool, it has the disadvantages of needing training data and has an extended analysis time. Recently, Kiani et al.[27] proposed and tested a fuzzy logic type of analysis, which comes close to the type of analysis of the AM. Although validation results of their system are presented, the way the results were obtained was not clearly described and therefore difficult to interpret.

Another instrument that is similar to the AM is the Dynaport ADL monitor described by Busser and colleagues.[9] In the Dynaport system the 'trunk' sensors are integrated into the recorder, which is carried in a belt around the waist. The instrument is validated in children; an overall agreement percentage between 76 and 92% is reported. It is not clear how activities are distinguished in the analysis, although the posture detection is probably based on the same principle as used in the AM.

Fahrenberg et al.[21,22] studied the possibilities of accelerometry to detect postures and movement from a psychophysiological viewpoint, also using a Vitaport measurement system. Fahrenberg and colleagues applied acccelerometers and a hydrostatic tube to monitor their subjects' ambulatory activity. In their study, only a small number of standardised activities were performed to determine validity

Quantity of activities

This thesis partly reflects the several phases in the development of an instrument. Not only the final (extended) version of the AM is described, but also the developmental stages and the first version. Although such an approach may have a disadvantage with respect to readability, these phases provided information that is relevant in assessing the functioning of the instrument.
The validity of the AM was studied in several populations: healthy subjects (chapter 4),[12] leg amputees (chapter 4),[12] patients after failed back surgery (chapter 6),[10] students participating in a psychophysiological study (chapter 5),[10] and in patients with heart failure (no part of this thesis).[5] The results indicated a low sensitivity for population-related factors, although each study revealed new knowledge on the functioning of the AM. Distinguishing the category climbing stairs from (especially) walking remains a problem. Currently, we are searching for an optimal procedure to correct; in the analysis, for initial angular deviations of the sensor from the ideal ('in plane') attachment. This procedure may increase the validity of the detection of stair climbing. Probably, unusual types (i.e., not step over step) of climbing stairs may remain difficult to detect. Wheelchair driving was only performed in the master study (see chapter 3). In an ongoing study the detection of driving a wheelchair is the focus of interest; the analysis software of the extended version of the AM has still to be applied on the signals. In future research that includes driving a wheelchair, the use of an extra (arm) sensor will be an option. The output of the AM can also be related to other signals or variables simultaneously measured; for example, the combination with ECG/heart rate, but also the combination with other signals (e.g. blood pressure, electromyography) or other data (e.g. perceived pain, specific events).

Quality of activities

Walking is an important activity in the treatment of many rehabilitation patients. Measurement of gait quality traditionally takes place in gait laboratories. The generalisability of results obtained in this setting to daily life functioning is, however, questionable. It was, therefore, our aim to measure gait quality during daily life, to obtain more valid data on gait quality. In this way walking can be measured for descriptive, explorative, and evaluative purposes. One of the clinically relevant possibilities may be the measurement of changes in gait quality within one day. However, it is not possible to determine whether a change in gait quality results from (external) different conditions (e.g. floor surface, shoes), or whether it results from the person himself (e.g. fatigue). This problem can be managed in different

ways. First, the effect of factors can be quantified, to assess when and, if so, to what extent any given factor influences gait quality. An example of such an approach is the study described in chapter 8. Second, it may be possible to measure these factors simultaneously with the accelerometer signals, e.g. light and sound.[23,27] Third, it may be possible to perform (shorter term) ambulatory measurements under controlled conditions: subjects follow a type of natural track outside the laboratory, disturbances (or factors of influence) are controlled, manipulated, or measured. In this way impairment-orientated gait analysis can change into the direction of disability-orientated gait analysis.[37] For this type of measurement, ambulatory monitoring is very appropriate and promising.
In this thesis, the potential of ambulatory accelerometry to measure quality was studied with respect to walking. Accelerometer signals will also contain information on the quality of other activities. One of the activities currently under study in our group is the transition from sitting to standing and walking. The speed and phasing of this movement are examples of variables aimed to be measured with accelerometry.

Signals from piezo-resistive accelerometers

Accelerometer signals are often difficult to interpret or predict – especially during dynamic activities – in contrast to signals on position and speed. The signals from piezo-resistive accelerometers are particularly complex, because they contain different acceleration components. This is disadvantageous on the one hand, because the interpretation of the signal is hampered by it; on the other hand, this may be advantageous, because different aspects of movement are represented within the same signal. The research described in chapter 9 has provided insight in these different aspects during walking, and has thus enhanced interpretation. The accelerometer signal has been considered inadequate for the measurement of kinematic variables.[47] Nevertheless, accelerometer signals have shown to be of value in the assessment of gait quality. In chapters 10 and 11, accelerometer signals were used to determine stride frequency and variability of movement co-ordination. With respect to movement variability no differences between persons with and without an amputation were found at fixed speeds, and variability of movement was characterised by large test-retest variability. Nevertheless, measurements performed by Slagter et al.,[45] using the same sensors and analysis method, showed differences in variability of movement co-ordination between elderly age groups. In that study, foot switches, and not peaks in the accelerosignal, were used to distinguish separate gait cycles. An exploratory study revealed, however, that at comfortable speed the way of gait cycle determination did not significantly influence the results. At lower

speeds, the positive peak before heel strike is less clear and more difficult to use. Then it is more practical to use alternative types of cycle determination: footswitches or accelerometers radially mounted on the lower leg.
In other studies, accelerosignals were also used for the measurement of movement co-ordination.[17,49] Furthermore, acceleration signals are frequently used to provide quality-related measures not related to movement co-ordination (see chapter 1). Therefore, there is no direct need for additional or different sensors in the measurement of gait quality, although the use of other types of sensors and multi-axial accelerometers, as well as the use of accelerometers at different locations will get further attention. Especially the use of gyroscopic sensors may be promising.[24,35]

Physical strain

Motility, or the variability of the accelerometer signal, depends on the movements of the segment the sensor is attached to. Motility and physical strain are theoretically not identical, but a relation between motility and physical strain is reported in literature.[8, 18,29,32, 33,38,46,51] If so, accelerometer signals may provide data on physical strain simultaneous to the detection of activities. Motility may be used instead of, or in combination with, heart rate measurement. A methodological advantage may be a higher reliability and validity, whereas a practical advantage is – if heart rate does not have to be measured – the simpler and more comfortable sensor configuration. Heart rate is sensitive to mental processes, stress, fear, illness, medication, temperature, body position, and type of movement.[2,31,38,42] The validity of motility may be threatened by external vibrations,[32,38] static work,[8,18,36] non-level walking[33,36] and, more generally by differences in motility-physical strain relations between activities.[8,33]
In the studies described in chapters 10 and 11, motility was characterised by a strong correlation with physical strain variables, relatively small differences within subjects, low test-retest variability; and relatively low sensitivity to changes in or differences in economy. The latter point is a disadvantage when, within a measurement, changes in economy occur that are not due to changes in walking speed. Furthermore, when persons with wide differences in the economy of walking are involved, an individual calibration curve has to be made. However, the results are promising, and further exploration of the potential of motility to measure physical strain is recommended. One of the issues of interest will be the combined measurement of heart rate and motility, possibly in combination with the output of the AM (sitting, standing, walking, etc). The combination of heart rate and motility measures is also done or suggested by others.[30,38], although a limited added value is also reported.[18] Nevertheless, the combination of the AM output, heart rate, and

motility, may be a powerful tool in the assessment of mobility-related activities. Another issue of interest will be the use of motility and the relation between motility and physical strain measures in activities other than walking. Field studies will be necessary to obtain final answers on questions about the feasibility of motility in the assessment of physical strain. This type of studies has also been performed by others.[e.g.32,51]

The %HRR was characterised by relatively large intra-subject and inter-subject differences. In the study presented in chapter 10, the maximum heart rate was based on a maximal bicycle ergometry test, whereas in the study described in chapter 11 the maximum heart rate was based on age. The actual maximum heart rate is difficult to determine in persons with an amputation. First, physical capacity tests are difficult to perform in these subjects. Due to the amputation and poor peripheral conditions, such as weak muscle strength, tests can lead to poor maximal performance without maximal cardiovascular and pulmonary strain.[13,14] Furthermore, due to central vascular problems[1,13,19,26,40,43] the maximal load level of tests is frequently determined by objective or subjective cardiac symptoms.[13,25,34] Cardiac function can also be assessed by stress tests using drugs, such as dobutamine.[39] In future research, the possibilities of several stress tests will be further explored.

Another point of interest is the use and effect of drugs. Persons with an amputation frequently use beta-blockers due to cardiac problems. The effects of beta-blockers are highly dependent on type, but generally a (small) decrease of HRrest and a (greater) decrease of HRmax and HRtask is found.[3,22,41,48,53] This leads to a smaller increment of heart rate with increasing load. Especially changes in drug use of patients during a study may threaten validity.

Applicability of the AM

The AM is extensively validated and, although some aspects need further study, the AM can be applied in research. One explorative study is described in this thesis (chapter 8).[6] In that study, AM variables differed between a group of patients with congestive heart failure, and a comparison group. The validation study presented in chapter 5[10] was designed in view of a psychopharmacological study on the effects of two oral dose levels of alprazolam, and one dose level of lorazepam.[7] In that study, AM variables also significantly differed between different conditions. So in both studies the AM was able to detect differences between groups and conditions, which supports its validity and usability in clinical research.

The AM is currently used in a research project on failed back surgery patients. The relation between activity level (pain behaviour) on the one hand, and other pain

measures and quality of life on the other, will be examined, as well as the effects of different treatment protocols on these measures. The AM will also be used in patients with heart failure receiving exercise training. Similarly, the AM may be used in patients receiving pulmonary rehabilitation, to study the effect of this rehabilitation on activity, and the relation between activity level and measures on the impairment level. The research related to persons with a leg amputation will continue; this research will aim at establishing the effect of different treatment strategies on activity level, but also on the quality, strain, economy and capacity measures as described in this thesis. Recently, we have started investigating the use of arms and hands in patients with sympathetic dystrophy; in this project, acceleration signals from the upper extremities will be studied and be coupled to the AM output.

Besides the prospects in evaluative studies, the AM can also be used for the development of theory within rehabilitation medicine. For example, the AM may be used to study and clarify relationships between the levels function, activity, and role fulfilment. Furthermore, the instrument can be used to explore and study the relationship between the aspects of mobility described in chapter 2: performed, possible, and preferred, from either the patient or the professional viewpoint. Discrepancies between these aspects can provide clinical insight in the issues of interest for individual patients, or in typical characteristics of patient groups.

However, also in other fields the AM will have an extensive set of possibilities, of which some have already been mentioned. For example, the AM may be used to facilitate the interpretation of other signals, such as ECG or blood pressure (which is described in more detail in chapter 5). Finally, the field of ergonomics has to be mentioned: ambulatory accelerometry may replace time-consuming observational techniques currently regularly used.

Concluding remarks

The work described in this thesis concerns studies on feasibility, validation and exploration of the AM in, especially, rehabilitation research. The studies have provided much information about the potential of the AM in research, and have created a base on which other studies are and can be built. Ambulatory monitoring of mobility-related activities is a relevant extension of the range of instruments used in rehabilitation. Ambulatory accelerometry, with the piezo-resistive accelerometers attached at the thighs and trunk, allows valid measurements of a large number of mobility-related activities. The detection of some activities needs further study. It seems that the functioning of the AM does not depend on the research population.

Accelerometer signals contain information on the quality of activities. The study of

this potential has been started with walking, but the measurement of the quality of other activities will also be possible. Measurement of prosthetic gait obtained with the AM of prosthetic gait appeared to be feasible, although one will have to cope with a considerable within-subject and between-subject variance.
Motility, a routine feature of the AM, is a promising variable in the measurement of physical strain during walking. The use of motility in activities other than walking, and the combination with heart rate and AM output, will be the next phase in our ongoing research.

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Summary

Within rehabilitation medicine, as well as in other medical disciplines, there is a need for reliable, sensitive and valid instruments to measure on the level of daily functioning. One of the possible perspectives is to regard daily functioning as a complete range of postures, transitions between postures, and movements, which together are called *mobility-related activities.*

Ambulatory monitoring enables measurements to be performed on persons without being restricted by space due to use of e.g. instruments, cables, etc. Due to latest technological developments, an ambulatory instrument to measure mobility-related activities could be developed: the *Activity Monitor* (AM), which is the thread that runs throughout the thesis. The AM is an instrument based on long-term ambulatory accelerometry, and aimed at the measurement of mobility-related activities. Distinction is made between the aspects *quantity* (which activity is performed, when, how frequent, for how long), *quality* (how is the activity performed), and *physical strain* (the physical reaction of the body due to the performance of an activity). Activities such as walking, climbing stairs, driving a wheelchair, lying, standing, sitting, and the transitions between these body positions are aimed to be distinguished. The thesis is structured to correspond with the three main aspects of mobility-related activities: quantity (chapters 3-8), quality (chapter 9), and physical strain (chapter 10); in chapter 11 all three aspects are studied.

Chapter 1 is a general introduction of the main issues in this thesis. An overview is provided of ambulatory systems for activity monitoring that are described in literature. Furthermore, a brief overview is provided of studies in which accelerometry has been used to study human movement, as well as an overview of the ambulatory measurement of physical strain. The last section of the chapter contains an outline of the thesis.

Mobility is an important construct in rehabilitation medicine, and many instruments have emerged which measure or assess (aspects of) mobility. In the selection or development of an appropriate technique, knowledge about essential characteristics of techniques and fundamental differences between them is necessary. Furthermore, the requirements within the field of rehabilitation should be considered. Thus, *chapter 2,* presents an overview aimed to classify, assess, and discuss current techniques that are or can be used to measure aspects of mobility. Eight techniques (physical science techniques, clinimetry, observation, diaries, questionnaires, actigraphy, physiological techniques, and activity monitors) are studied on level of

outcome variables, the aspect of mobility they measure, and methodological and practical criteria. It is concluded that rehabilitation medicine has a particular need for instruments that enable measurement of outcome variables on the level of activity and role fulfilment. Techniques differ in type and number of mobility aspects they measure. Furthermore, important differences exist based on methodological and practical criteria. It is concluded that the choice of a technique will always depend on a complexity of factors, such as clinical problem, research question, mobility aspect of interest, required methodological strength, costs, and availability.

In *chapter 3* the AM is described in more detail, with focus on the quantity aspect. Requirements for sensors used in ambulatory measurement of activity are discussed. Piezo-resistive accelerometers fulfil these requirements, and their functioning is explained. A feasibility study showed the applicability of piezo-resistive accelerometers and a theoretical detector scheme. Following this feasibility study, a *first version* of the AM analysis software is developed, followed by a more *extended version*. The procedures and settings within the analysis software of both versions are guided by a 'master study', which is described. A more detailed technical description of the extended version of the AM is provided in the last section of the chapter.

Chapters 4, 5, and 6 present three validation studies. Although these validation studies are similar in design, they differ in subjects, kind of activities, and setting: (1) healthy subjects and amputees in measurements with standardised functional activities in an apartment of an occupational therapy department (*chapter 4*); (2) subjects without known diseases or impairments within a psychopharmacological study, performing spontaneous activities and standardised activities in a living room (*chapter 5*); and (3) patients after failed back surgery in measurements with standardised functional activities in and around their own house (*chapter 6*). In these chapters, the validity results are based on the first version of the AM, the output of which is restricted to several static activities (several types of lying, sitting, and standing), all transitions, and dynamic activities as one group. In the three studies, the subjects performed normal daily activities, during which accelerations were measured and videotape recordings were made (reference method). Validity is assessed by calculating agreement scores between AM output and videotape analysis, and by comparing the number of transitions and dynamic periods, and the duration of activities, determined by both methods. In the three studies, the overall agreement between the AM and the videotape analysis was 90, 88, and 87%, respectively. Other agreement scores were generally within the 0-10% error range. The number of transitions and duration of activities is well determined. It is concluded that the measurements with the AM provide valid data on the performed activities.

After the development and validation of the first version of the Activity Monitor analysis program, an extended version is developed as a sequel to that first program. This extended version is based upon a non-hierarchical decision scheme and three input features, and allows also the detection of several dynamic activities. In *chapter 7*, the validity of measurements with this extended version is described. The signals from three previously performed and reported validity studies are used (see chapter 4, 5, and 6). The overall agreement between AM output and videotape analysis for the three studies was 89%, 93%, and 81%, respectively. In the studies with considerable walking periods walking has agreement scores ranging from 67 to 95%. In climbing stairs, especially the sensitivity scores were lower (mean 24% and 76%, respectively, range 0-87%), generally due to the misdetection as walking. Generally, the duration of walking is slightly underestimated (–0.8% in both studies). The number of walking periods is well determined (169 versus 170, and 255 versus 240, respectively). The agreement scores for cycling ranged from 51-100%. It is concluded that the extended version of the AM is a valuable extension of the first AM version, although the detection of some activities (especially climbing stairs and driving a wheelchair) will need further study.

From the validation studies it is concluded that the AM could be used in applied research. One such example is described in *chapter 8*. The aim of the study was to obtain detailed information on everyday physical activity variables measured with the AM (extended version) in chronic congestive heart failure (CHF) patients (n=7), and on between-day variance in physical activity in this group. In addition, results found in the CHF group are compared with results found in a healthy, matched comparison group (n=5). In the CHF group, measurements are performed during 2 consecutive weekdays and during one of these days of the subsequent week; in the healthy group, measurements were performed during 2 consecutive weekdays. The total duration of dynamic activities (as a percentage of the measurement time) was 3.9% (SD 1.5%) in the CHF group and 11.3% (SD 3.0%) in the comparison group (P=0.02). The mean *motility* – a measure on the variability of the acceleration signal, and assumed to be related to the intensity of movement – and the number of walking periods were significantly lower in the CHF group. The total number of transitions and the number of sit-to-stand transitions tended to be lower in the CHF group than in the comparison group. In contrast to our expectation, the between-day variance in the total duration of dynamic activities in the CHF group was significantly smaller for different weekdays (0.8%) than for similar weekdays (5.49%). The between-day variance tended to be higher in the comparison subjects (16.18%) than in the patients (3.94%). The results indicate that several physical activity variables, as measured with the AM, are considerable lower in patients with CHF than in healthy

comparison subjects. Monitoring physical activity in CHF patients during similar days of the week do not reduce the between-day variance.

The input of the AM is provided by thigh-fixed and trunk-fixed uni-axial piezo-resistive accelerometers. The use of the accelerometer signals in the assessment of the quantity and quality of walking is complicated by the fact that the measured signal is composed of different types of acceleration, and that the signal may vary between or within measurements. The study described in *chapter 9* was aimed to obtain insight in the signal from the tangential AM-accelerometer attached at the thigh during walking, and in its components; to investigate its relation with temporal events; and to investigate the influence of subject variability, walking speed, walking surface, and sensor attachment on the signal. Six subjects walked with three different speeds (comfortable, slow, and fast) and under several conditions (different types of walking surface and sensor attachment). Simultaneous measurements were performed with accelerometers, footswitches, and an optoelectronic system. From the optoelectronic system the components of the acceleration signal were calculated. Distinction is made between accelerations due to inclination and inertia. The components of the inertial acceleration – accelerations due to translation, rotation, and deformation – were additionally studied. The results showed that the acceleration signal is generally characterised by relatively high frequency components with regard to the movement frequency, and low amplitudes. Most pronounced is a high positive acceleration peak just before heel strike. The inclination component is characterised by a sinusoidal shape; the inertial component and inclination component have opposing contributions to the signal. The effect of deformation of the thigh on the signal is small and not general. If the hip joint marker is taken as point of reference, both the rotational and the translational acceleration component contribute to the inertial acceleration. The root mean square – used as a measure of agreement between signals – was significantly higher for the factors inter-subject variability, walking speed, and displacing the sensors 2 cm medially. Insight in the acceleration, its components, and its sensitivity to subject variability, walking speed, walking surface, and sensor attachment was obtained. This knowledge will contribute to the further use of these signals in the assessment of gait quality, and in the detection of walking.

The motility signal – derived from the accelerosignals – is possibly an indicator of physical strain. Therefore, *chapter 10* was aimed at the potential of AM-accelerosignals to measure physical strain, compared to the more frequently used measure of heart rate, and in relation to the reference measure oxygen uptake. The study was focused on the feasibility of accelerometry in the evaluation of physical strain in walking at different walking speeds and different levels of economy. Twelve subjects without known diseases or impairments performed a walking test

on a treadmill with increasing walking speed. After a 6-week period these measurements were repeated, whereas additional measurements were performed during perturbed walking with a brace. Motility, oxygen uptake (VO_2), and percentage heart rate reserve (%HRR) were calculated. Motility, VO_2, and %HRR all increased on average with increasing speed, although no complete linearity between the measures existed. The pooled explained variance for the motility-VO_2 and %HRR-VO_2 relation was 0.91 and 0.85, respectively. The motility-VO_2 relation showed small differences between subjects, whereas both relations showed high individual correlation coefficients. The sensitivity to changes in physical strain due to an increase in walking speed was the highest for motility; this measure also had the highest test-retest reliability. The pooled changes in all three variables due to walking with a brace significantly correlated with each other; now the relation between changes in %HRR and VO_2 showed the highest explained variance (0.66 versus 0.31, respectively). The sensitivity to changes due to perturbed walking was generally the highest for VO_2 and the lowest for motility. The relation between motility and VO_2 found during increasing speed could not be applied to the perturbed gait condition. It was concluded that motility appeared to be a feasible alternative to measure physical strain during walking, but that further study is necessary to examine the feasibility in subjects with different levels of economy, such as persons with an amputation.

Some questions arose about the feasibility and validity of measurements with the AM in the early phase of rehabilitation of persons with an amputation. The study described in *chapter 11* was aimed at the validity of the detection of walking and climbing stairs, the reliability of gait quality and physical strain measurements, the sensitivity of these measurements to differences and changes, and the potential of motility to predict physical strain. Ten patients with an amputation of the leg and 10 matched comparison subjects performed, with an interval of one month, the same protocol three times, including comfortable walking, climbing stairs, and a test with fixed walking speeds. Signals from accelerometers and electrocardiography (ECG) were monitored. Walking speed, stride frequency, stride length, movement variability, motility, percentage heart rate reserve (%HRR) and physiological cost index (PCI) were variables of interest. Overall, 98% (comparison group) and 95% (patient group) of the walking periods was correctly determined. The detection of climbing stairs was less successful: 0% correct in the patient group, 48% correct in the comparison group. At comfortable walking speed, all variables differed between groups. At fixed walking speeds only %HRR and PCI differed between groups. No changes over time were found in the patient group. Test-retest reliability (reliability coefficient) was the highest in walking speed, stride frequency, and stride length, and the lowest in PCI and stability. The relation between motility and %HRR found

in the previous study (chapter 10) was confirmed; the motility-%HRR relation of patients differed from that of the comparison group. The accuracy to predict %HRR with motility was highest in the session in which the calibration curve had been made. It was concluded that the detection of walking is reliable. The ambulatory measurement of gait quality and physical strain variables must be able to cope with a considerable within-subject and between-subject variability. When motility is used as an indicator of physical strain, an individual calibration curve will be necessary.
The last chapter (*chapter 12*) is a general discussion with concluding remarks on some of the main issues in this thesis.

Samenvatting

Binnen de revalidatiegeneeskunde bestaat er behoefte aan betrouwbare, sensitieve en valide instrumenten waarmee uitspraken mogelijk zijn over het dagelijks functioneren van patiënten. Het dagelijks functioneren kan worden opgevat als een geheel van bewegingen (*dynamische activiteiten*), houdingen (*statische activiteiten*) en overgangen tussen houdingen (*transities*), die gezamenlijk *mobiliteit-gerelateerde activiteiten* worden genoemd.
Ambulante registratie wil zeggen dat de persoon die wordt gemeten niet door instrumenten, kabels enz. aan plaats en ruimte is gebonden. Hierdoor is deze techniek geschikt voor het meten tijdens het dagelijks functioneren. In dit proefschrift wordt de ontwikkeling, validatie en toepassing beschreven van de Activiteiten Monitor (AM), een instrument waarmee mobiliteit-gerelateerde activiteiten ambulant kunnen worden gemeten. Door middel van langdurig en ambulant geregistreerde signalen van op de huid bevestigde *accelerometers* (versnellingsopnemers) kunnen uitspraken worden gedaan over een verzameling mobiliteit-gerelateerde activiteiten. Tot die verzameling behoren liggen, zitten, staan, lopen, traplopen, rolstoel rijden, fietsen, algemeen bewegen en transities. Binnen het proefschrift wordt onderscheid gemaakt in drie aspecten van mobiliteit-gerelateerde activiteiten: *kwantiteit* (welke activiteit wordt uitgevoerd, hoe vaak, hoe lang, wanneer), *kwaliteit* (hoe wordt de activiteit uitgevoerd), en *fysieke belasting* (de fysieke reactie van het lichaam op het uitvoeren van activiteiten). In hoofdstuk 3 tot en met 8 staat het kwantitatieve aspect centraal, in hoofdstuk 9 het kwalitatieve aspect en in hoofdstuk 10 de fysieke belasting. Het onderzoek dat in hoofdstuk 11 is beschreven omvat alle drie aspecten.

Hoofdstuk 1 begint met een algemene inleiding over het onderwerp van het proefschrift, waarna een overzicht wordt gegeven van in de literatuur beschreven systemen van ambulante registratie van activiteiten. Vervolgens wordt ingegaan op het gebruik van accelerometrie binnen de analyse van het menselijk bewegen en op het ambulant meten van fysieke belasting.
Mobiliteit is een belangrijk begrip binnen de revalidatiegeneeskunde en er bestaan veel instrumenten die gericht zijn op het meten van (aspecten van) mobiliteit. Kennis van belangrijke eigenschappen van meettechnieken en wezenlijke verschillen tussen die technieken is van groot belang. Verder is het essentieel dat wordt stilgestaan bij de vraag aan welk type instrumenten nu feitelijk behoefte is binnen de revalidatiegeneeskunde. Daarom is *hoofdstuk 2* gericht op het classificeren,

beoordelen en bediscussiëren van beschikbare technieken voor het meten van mobiliteit. Acht technieken, te weten natuurwetenschappelijke technieken, clinimetrie, observatie, dagboeken, vragenlijsten, actigrafie, fysiologische technieken en activiteiten monitoren worden besproken. De technieken worden met elkaar vergeleken voor wat betreft het niveau van de uitkomstmaat, het aspect van mobiliteit dat zij meten en methodologische en praktische eigenschappen. Geconcludeerd wordt dat er binnen de revalidatiegeneeskunde vooral behoefte bestaat aan instrumenten op het niveau van activiteiten en rolvervulling/sociale participatie. Technieken blijken van elkaar te verschillen in soort en aantal aspecten van mobiliteit dat zij meten. Verder wordt geconstateerd dat er belangrijke verschillen in methodologische en praktische eigenschappen bestaan. Benadrukt wordt dat de keuze voor een bepaalde techniek altijd wordt bepaald door een complex van factoren, zoals de klinische vraag, de onderzoeksvraag, het aspect van mobiliteit waarin men is geïnteresseerd, vereiste methodologische sterkte, kosten en beschikbaarheid.

In *hoofdstuk 3* is de AM meer in detail beschreven, met de nadruk op het aspect kwantiteit. Allereerst wordt ingegaan op eisen die aan sensoren worden gesteld bij ambulante metingen. Piezo-resistieve accelerometers blijken het best aan deze eisen te voldoen, en hun werking wordt toegelicht. Vervolgens wordt een haalbaarheidsonderzoek beschreven dat de mogelijkheden van deze sensoren en een theoretisch detectieschema heeft aantoonde. Na dit haalbaarheidsonderzoek is een *eerste versie* van de AM analyse software ontwikkeld, later gevolgd door een *uitgebreide versie*. De procedures en instellingen binnen de analyseprogrammatuur zijn gestuurd door zogenaamde leermetingen, waarvan de opzet en resultaten zijn beschreven. Het hoofdstuk eindigt met een gedetailleerde beschrijving van de uitgebreide versie van de AM.

In *hoofdstuk 4, 5 en 6* wordt verslag gedaan van drie validatie-onderzoeken. Deze onderzoeken zijn overeenkomstig in opzet, maar verschillen in onderzoeksgroep, uitgevoerde activiteiten en setting. Het eerste onderzoek betreft gezonde proefpersonen en personen met een onderbeenprothese tijdens het uitvoeren van gestandaardiseerde functionele activiteiten in een 'appartement' op een afdeling ergotherapie (*hoofdstuk 4*). Het tweede onderzoek betreft gezonde deelnemers aan een psychopharmacologisch onderzoek tijdens het uitvoeren van spontane en gestandaardiseerde activiteiten in een ingerichte kamer (*hoofdstuk 5*). Het derde validatie-onderzoek betreft patiënten na failed back surgery, tijdens het uitvoeren van gestandaardiseerde functionele activiteiten in hun eigen woonomgeving (*hoofdstuk 6*). De onderzoeksresultaten in deze hoofdstukken zijn gebaseerd op de eerste versie van de AM. De verzameling te detecteren activiteiten van deze versie beperkt zich tot verschillende statische activiteiten (verschillende vormen van

liggen, zitten en staan), alle transities en dynamische activiteiten als één groep. Tijdens de metingen zijn versnellingen geregistreerd en zijn simultaan video-opnames gemaakt als referentiemethode. Validiteit van de metingen met de AM is bepaald met behulp van overeenstemmingsmaten tussen AM-output en video-analyse. Verder is met beide methoden het aantal transities, het aantal dynamische periodes en de duur van alle activiteiten bepaald. De totale overeenstemming tussen AM-output en video-analyse is respectievelijk 90%, 88% en 87%. De grootte van fouten van andere overeenstemmingsmaten ligt in het algemeen binnen de 0 tot 10%. Het aantal transities en de duur van activiteiten worden goed bepaald.
Recent zijn in de software algoritmes geïmplementeerd om dynamische activiteiten te onderscheiden en is de structuur van de analyseprogrammatuur aangepast. Dit heeft geleid tot de uitgebreide versie van de AM (beschreven in hoofdstuk 3). Deze uitgebreide analyse versie van de AM is gevalideerd met behulp van de signalen van de drie reeds beschreven validatie-onderzoeken (*hoofdstuk 7*). De totale overeenstemming tussen AM output en video-analyse is nu respectievelijk 89%, 93% en 81%. In de metingen waarin loopperiodes geregeld voorkomen varieren de sensitiviteit voor lopen en predictieve waarde van lopen van 67 tot 92%. Van traplopen is de predictieve waarde hoog (van 61 tot 100%), maar de sensitiviteit lager (van 0 tot 87%). De sensitiviteit voor fietsen varieert van 84 tot 100%, terwijl de predictieve waarde van fietsen varieert van 51 tot 100%. De duur van lopen wordt in het algemeen enigszins onderschat (gemiddelde afwijking –0.8%). Het aantal loopperioden (169 versus 170 en 255 versus 240) wordt goed bepaald. Geconcludeerd wordt dat de uitgebreide AM versie het lopen (duur, aantal perioden) en fietsen (duur) goed bepaalt, naast de reeds eerder aangetoonde goed detectie van statische activiteiten en transities. Traplopen wordt vaak als lopen gedetecteerd, terwijl voor de detectie van rolstoel rijden aanvullend onderzoek wordt voorgesteld.
Gezien de resultaten van de validatie-onderzoeken wordt geconcludeerd dat de AM gebruikt kan worden binnen toegepast onderzoek. In *hoofdstuk 8* is een voorbeeld van een dergelijke toepassing beschreven. De AM wordt in dat onderzoek gebruikt voor de meting van dagelijkse activiteiten van patiënten met hartfalen (congestive heart failure, CHF). Het doel van het onderzoek is gedetailleerde informatie te krijgen over de dagelijkse activiteiten van CHF-patiënten (n=7), gemeten met de AM, en over de tussen-dag-variantie in activiteiten. De in deze groep gevonden resultaten worden vergeleken met die van een gezonde, gematchte vergelijkingsgroep (n=5). In de CHF-groep zijn metingen gedaan op twee achtereenvolgende weekdagen en gedurende één van deze dagen een week later. In de vergelijkingsgroep hebben de metingen alleen op 2 achtereenvolgende weekdagen plaatsgevonden. De duur van dynamische activiteiten (uitgedrukt als percentage van de meettijd) was 3.9% (SD 1.5%) in de CHF-groep en 11.3% (SD

3.0%) in de vergelijkingsgroep (P=0.02). De gemiddelde *motiliteit* – een van de versnellingssignalen afgeleide maat die kan worden opgevat als een maat voor de mate waarin en de intensiteit waarmee wordt bewogen – en het aantal loopperioden zijn significant lager in de CHF-groep. Het aantal transities en het aantal zit-tot-stand-transities vertonen een tendens om lager te zijn in de CHF-groep. In tegenstelling tot de verwachting is de tussen-dag-variantie in duur van dynamische activiteiten in de CHF-groep significant kleiner tussen verschillende weekdagen dan tussen dezelfde weekdagen. De tussen-dag-variantie vertoont een tendens om hoger te zijn in de vergelijkingsgroep (16.18%) dan in de patiëntengroep (3.94%). Het onderzoek toont aan dat de AM gedetailleerde en relevante informatie kan geven over het dagelijks functioneren van mensen. Zoals verwacht, dient er wel rekening gehouden te worden met een aanzienlijke tussen-dag-variantie.

De input van de AM wordt geleverd door uni-axiale piezo-resistieve accelerometers die bevestigd zijn op de bovenbenen en de romp. Het gebruik van versnellingssignalen voor de detectie van activiteiten en het meten van de kwaliteit van activiteiten is bemoeilijkt door het feit dat het signaal uit verschillende versnellingscomponenten bestaat. Verder is het signaal mogelijk gevoelig voor factoren als variatie binnen en tussen personen, loopsnelheid, ondergrond en sensorbevestiging. Het onderzoek dat is beschreven in *hoofdstuk 9* is gericht op het verkrijgen van inzicht in (de componenten van) het signaal van de AM-accelerometer die bevestigd is op het bovenbeen. Verder is het onderzoek gericht op de relatie van het versnellingssignaal met specifieke momenten van de schrede en op de invloed van de genoemde factoren. Zes proefpersonen liepen met drie verschillende snelheden en onder verschillende condities een parcours. Gelijktijdige metingen zijn verricht met accelerometers, voetcontactschakelaars, en een opto-electronisch systeem. Met dit laatste systeem zijn de verschillende versnellingscomponenten berekend. Onderscheid wordt gemaakt tussen versnellingen door *inclinatie* (hoekpositie) en *inertie* (beweging). De inertiële versnelling is vervolgens verder ontleed in versnellingen ten gevolge van translatie, rotatie en vervorming van het bovenbeen. Het versnellingssignaal tijdens lopen wordt gekarakteriseerd door hoog-frequente componenten in verhouding tot de bewegingsfrequentie en geringe amplitudes. Het meest opvallend is een forse positieve versnellingspiek net voor hielcontact. De analyse van de verschillende componenten toont aan dat de inclinatiecomponent globaal sinusvormig is. De inertiecomponent is ruwweg tegengesteld aan de inclinatiecomponent. Beide curves vertonen daardoor de tendens elkaar uit te doven. Versnellingen door vervorming van het bovenbeen blijken geen essentiële bijdrage aan het signaal te leveren. De wortel van het gekwadrateerde verschil (rms waarde), die is gebruikt als maat voor overeenstemming tussen signalen, is significant hoger voor de factoren tussen-

subject-variabiliteit, loopsnelheid en het 2 cm mediaal verplaatsen van de sensor (getoetst ten opzichte van de rms waarden van de test-hertest meting).
Het van de versnellingssignalen afgeleide motiliteitssignaal is mogelijk geschikt als een ambulant te meten maat voor fysieke belasting. In hoofdstuk 10 wordt een onderzoek beschreven naar het vermogen van het motiliteitssignaal om als maat van fysieke belasting te dienen. Deze geschiktheid is bepaald door een vergelijking te maken met een andere, algemeen gebruikte, belastingsmaat: de hartslagfrekwentie, en door deze beide maten te relateren aan de referentiemaat zuurstofopname. Loopsnelheid en loopefficiëntie worden gemanipuleerd. Twaalf gezonde proefpersonen voeren een looptest met toenemende loopsnelheid uit op een loopband. Na een periode van 6 weken wordt deze test herhaald. In deze tweede sessie wordt eveneens een loopprotocol uitgevoerd met een door een kniebrace verstoord looppatroon. Motiliteit, zuurstofopname (VO_2) en het percentage hartslag reserve (percentage heart rate reserve, %HRR) worden berekend. Alle drie variabelen nemen toe met toenemende loopsnelheid, hoewel hun onderlinge relatie niet volledig lineair is. De verklaarde variantie voor alle data samen is voor de motiliteit-VO_2 relatie 0.91 en voor de %HRR-VO_2 relatie 0.85. De motiliteit-VO_2 relatie vertoont ten opzichte van de %HRR-VO_2 relatie geringere inter-individuele verschillen, terwijl beide relaties hoge individuele correlaties vertonen. Voor motiliteit is de sensitiviteit voor veranderingen in fysieke belasting door toenemende loopsnelheid het hoogst. Verder heeft motiliteit ook de hoogste test-hertest betrouwbaarheid. De veranderingen in de variabelen door het lopen met een brace correleren met elkaar. De %HRR-VO_2 relatie vertoont de hoogste verklaarde variantie (0.66 versus 0.31 van de motiliteit-VO_2 relatie). De sensitiviteit voor veranderingen door het lopen met een brace is over het algemeen het hoogst voor VO_2 en het laagst voor motiliteit. De relatie tussen motiliteit en VO_2 tijdens toenemende loopsnelheid is niet valide voor het lopen met een brace. Geconcludeerd wordt dat motiliteit het vermogen heeft om fysieke belasting tijdens lopen te meten, maar dat aanvullend onderzoek nodig is bij personen met een afwijkende bewegingsefficiëntie, zoals personen met een beenamputatie.
Onder andere naar aanleiding van het onderzoek beschreven in hoofdstuk 10, ontstaan vragen over de mogelijkheden en validiteit van metingen met de AM in de vroege revalidatiefase van patiënten met een beenamputatie. Het onderzoek zoals beschreven in *hoofdstuk 11* is gericht op de detectie van lopen en traplopen, de betrouwbaarheid van metingen van loopkwaliteit en fysieke belasting, de sensitiviteit van deze metingen voor verschillen en veranderingen, en het vermogen van motiliteit om fysieke belasting te voorspellen. Tien patiënten met een beenamputatie en tien gematchte vergelijkingspersonen voeren drie maal, met een interval van een maand, een protocol uit, bestaande uit comfortabel lopen, traplopen,

en lopen op diverse vaste snelheden. Signalen van accelerometers en electrocardiografische signalen (ECG) worden geregistreerd. Variabelen zijn loopsnelheid, schredefrequentie, schredelengte, variabiliteit van beweging, motiliteit, %HRR en de efficiëntiemaat physiological cost index (PCI). In totaal werd 98% (vergelijkingsgroep) en 95% (patiëntengroep) van de loopperioden volledig correct gedetecteerd. De detectie van traplopen was minder goed: 0% volledig correct in de patiëntengroep, 48% correct in de vergelijkingsgroep. Op comfortabele loopsnelheid verschillen alle variabelen tussen de twee groepen. Bij het lopen op vaste snelheden verschillen alleen %HRR en PCI. Er worden geen veranderingen in de tijd gevonden in de patiëntengroep. Test-hertest betrouwbaarheid (gemeten in de vergelijkingsgroep) is het hoogst voor loopsnelheid, schredefrequentie en schredelengte, en het laagst voor PCI en variabiliteit van beweging. De relatie tussen motiliteit en %HRR gevonden in eerder onderzoek (hoofdstuk 10) wordt bevestigd en blijkt significant verschillend tussen beide groepen. Geconcludeerd wordt dat de detectie van lopen betrouwbaar is. Bij de ambulante meting van loopkwaliteit en fysieke belasting zal rekening gehouden moeten worden met een aanzienlijke variantie binnen-persoon en tussen-persoon. Indien motiliteit als een voorspeller van fysieke belasting wordt gebruikt, zal een individuele ijking noodzakelijk zijn.

Het laatste hoofdstuk van het proefschrift (*hoofdstuk 12*) bestaat uit een algemene discussie en afsluitende opmerkingen.

Dankwoord

De strofe aan het begin van het proefschrift is zowel van toepassing op het onderwerp van dit proefschrift als op de wijze waarop dit proefschrift tot stand is gekomen. Het heeft wat langer geduurd en wat meer moeite gekost dan ik aanvankelijk dacht, maar de tijd zal dit doen vergeten. Wat minder snel zal vervagen is de sfeer waarin ik mijn werk heb gedaan, de uitdaging die mijn werk is geweest en het plezier dat ik in mijn werk heb gehad. Daarom gaat mijn zeer grote dank uit naar Prof. dr. Henk Stam, mijn promotor en hoofd van Revalidatie AZR&EUR. Henk, de vriendschappelijke manier van omgaan met elkaar, je enthousiasme over en geloof in de onderzoekslijn, het vertrouwen dat je mij heeft gegeven, je creatieve ideeën, de ondersteuning die het project en ik hebben gekregen en zeker ook alle informele activiteiten hebben mede de afgelopen jaren tot mooie jaren gemaakt. Ik hoop dat we nog lange tijd zullen samenwerken.

Alle medewerkers van Revalidatie AZR&EUR van de afgelopen jaren wil ik bedanken voor hun bijdrage geleverd aan het onderzoek. Laten we blijven streven naar een verdere integratie tussen afdeling en instituut; we kunnen elkaar niet missen. Een aantal mensen wil ik speciaal noemen. Armande Slagter en Yvonne Alofs-van Boxel, bedankt voor jullie bijdrage aan mijn onderzoek en aan de gezelligheid op het werk. Rita van den Berg, bedankt voor alle inhoudelijke en praktische hulp die je mij hebt gegeven gedurende het laatste anderhalf jaar van mijn onderzoek. Léonie, bedankt voor al je hand- en spandiensten. Fabiënne en Ruud, bedankt voor jullie begrip voor mijn drukte. Met alle energie gaan we verder werken aan jullie proefschrift.

De tijdsinvestering die de onderzoeken die in dit proefschrift zijn beschreven hebben gevergd is niet te onderschatten. Het analyseren van video-opnames, het 'knippen' van cycli, het doen van metingen bij patiënten thuis of op de Hoogstraat – ik geef nu zomaar wat voorbeelden – had ik nooit allemaal zelf kunnen doen in dit tijdsbestek. Daarom gaat mijn dank uit naar alle studenten die hun wetenschappelijke onderzoeksstage bij mij hebben gevolgd: Petra, Birgit, Carola, Jerry, Berna, Fleur, Yvette, Eefje, Sonia, Ivo, Diederik, Ciska, en Léonie. Als blijvend aandenken zijn alle papieren en bestanden nog steeds op jullie namen geordend.

De onderzoekslijn 'Ambulante registratie' van Revalidatie AZR&EUR is geen geïsoleerde lijn. Vele mensen van 'buiten' hebben bijgedragen aan de ontwikkeling van de Activiteiten Monitor, waarvoor ik ze veel dank verschuldigd ben. Van het begin af aan zijn Peter Veltink (Universiteit Twente) en Wim Martens (Phyvision) betrokken geweest bij het project. Zij hebben beiden een belangrijke rol gespeeld in

de totstandkoming van de Activiteiten Monitor en daarmee ook van dit proefschrift. In dit verband wil ik ook Rob van Lummel (McRoberts) noemen, een derde 'pionier van het eerste uur'. TEMEC Instruments, en met name Harrie Timmers, is belangrijk geweest voor de recente ontwikkelingen op het gebied van de recorders en sensoren.
Veel personen en onderzoeksgroepen zijn enthousiast geworden over activiteiten registratie, hetgeen voor mij een grote stimulans is geweest. Voor hun enthousiasme wil ik met name Joke Tulen (Psychiatrie AZR/EUR), Peer Neeleman (Pijn Kennis Centrum AZR), en Robert Wagenaar (voorheen Fysiotherapie VU) dank zeggen. Ik hoop met allen de samenwerking voort te zetten.
Verder wil ik een aantal interne afdelingen danken die op verschillende wijze bij mijn onderzoek betrokken zijn geweest: de Centrale Instrumentele Dienst, de afdeling Epidemiologie en Biostatistiek (in het bijzonder dr. Th. Stijnen) en de afdeling Biomedische Natuurkunde en Technologie.
Via verschillende wegen is mijn onderzoek direct of indirect financieel gesteund. Daarvoor mijn dank aan het Algesiologisch Instituut, Jaeger Toennies Benelux, het Ministerie van Economische Zaken, Pijn Kennis Centrum AZR, de Rotterdamse Stichting voor Cardiologische Revalidatie, en TEMEC Instruments.
Al met al zijn bij de metingen heel wat patiënten en niet-patiënten betrokken geweest. Zonder hun enthousiaste medewerking was het onderzoek niet mogelijk geweest. Ik wil voor hun hulp bij het werven van 'externe' patiënten de revalidatie-artsen Bert Kap en Peter Janssens van harte danken. Ook dank aan revalidatiecentrum 'de Hoogstraat' voor hun toestemming voor en medewerking bij het doen van metingen in het centrum.
Voor hun belangstelling en hulp wil ik mijn (schoon)familie en vrienden van harte bedanken. Ik ben niet vergeten dat zij belangrijker zijn dan werk.
Veel trotser dan op het product dat voor u ligt ben ik op drie andere 'producten' die in dezelfde periode tot stand zijn gekomen: mijn drie kinderen Joep, Bart en, sinds kort, Marieke. Natuurlijk, ze hebben me heel veel tijd 'gekost', maar wat hebben ze me ook veel gegeven.
Via mijn kinderen kom ik vanzelfsprekend uit bij Annemieke. Annemiek, dankzij jou heb ik op een bepaald moment gekozen voor datgene waar mijn hart lag, het wetenschappelijk onderzoek. Ik zal je daarvoor altijd dankbaar blijven. We hebben een drukke tijd met jouw en mijn 'bevalling' toch maar weer goed doorstaan. Dank je wel voor je aanwezigheid en je liefde.

About the author

Hans (Johannes Bernardus Josephus) Bussmann was born on February 28, 1961 in Etten-Leur (the Netherlands). After attending the *VWO* at the *Scholengemeenschap* Etten-Leur, he graduated at the Academy of Physical Therapy 'West-Brabant' (Breda) in 1983. From 1983 to 1985 he did his alternative (*vervangende*) military service at the Rehabilitation Centre Breda. In 1986 he started his study at the Faculty of Human Movement Sciences at the *Vrije Universiteit* in Amsterdam; before and during this study he worked as physiotherapist in several practices. He graduated *cum laude* in 1991 with an MSc degree, majoring in Functional Anatomy and with minors in Exercise Physiology and Psychology. In the same period he also completed a teacher course at the *Vrije Universiteit*. In 1992 he became faculty member at the Institute of Rehabilitation Medicine of the Erasmus University Rotterdam, closely connected with the Rehabilitation department of the University Hospital-Dijkzigt. Together with Prof. dr Henk J. Stam, he started in that year the research project 'Ambulatory monitoring of mobility-related activities'. From 1992, he has worked on the research presented in this thesis, was supervisor of students and researchers, was involved in the training of medical students and rehabilitation specialists and in various management tasks. After completion of this thesis, he will continue his work on ambulatory monitoring and related activities.
Hans Bussmann currently lives in Breda, is married to Annemieke Lagerberg and they have three children: Joep, Bart, and Marieke.