WALKING**AFTER**STROKE

Co-ordination Patterns & Functional Recovery

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WALKING AFTER STROKE
CO-ORDINATION PATTERNS & FUNCTIONAL RECOVERY

PROEFSCHRIFT

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aan de Universiteit Twente,
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volgens besluit van het College voor promoties
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CHAPTER 1

Introduction and outline of the thesis

Introduction

Worldwide hundreds of thousand cases of stroke are reported each year. The incidence of stroke in The Netherlands is approximately 30,000 per year and the prevalence is 120,000 patients [1] on a total population of 16.280.000.
The consequences of a stroke can be seen as a massive disturbance in information processing, causing not only motor deficits but also cognitive impairments and behavioural problems [2]. Although neuro-psyscological deficits and personal changes are the hardest to cope with, regaining the ability to walk is an important goal for the stroke patient [3]. Of those who survive the acute phase about 11% are unable to walk and 55% have moderate to severe walking disabilities. Hence relearning to walk is relevant for a large group of the total stroke population. For most of the patients, in the early stage, movements cannot be performed or the co-ordination of movements has lost its smoothness. It is thought that after several weeks novel co-ordination patterns will evolve. Although most of the therapies used in clinical practice are based on the assumption that improvement of functional abilities is facilitated through the improvement of muscle co-ordination little is known whether these co-ordination patterns indeed do change, let alone the speed at which this would happen and the type of patterns that come with it [4-7]. Treatment of stroke patients may benefit greatly when more insight would exist in the recovery process and the mechanisms that may influence recovery.
The lack of basic knowledge about the nature of the newly emerging but often pathological coordination patterns and the speed at which this happens, originated in more than nine different treatment methods that are used in today’s clinical practice [8-16]. Among these, often contradicting, methods Brunnstrom, Johnstone, Priopriocceptive Neuromuscular Facilitation (PNF), Motor Relearning Programme (MRP) and Neuro Developmental Treatment (NDT) are the most well known. Although it is assumed that every method has a positive effect on functional recovery of walking no scientific evidence exists [17-28] that one method is superior to the other [29].
This fact raises an important question. How come that different and often contradicting methods lead to the same end result? Most studies evaluate outcome using clinical measures. These measures explore one or more items on "function" and or "ability level" such as: symmetry, synergies, muscle force, muscle tone, walking and Activities of daily living. Although most of the treatment methods assume that improvement of functional abilities is reached by

improving function (e.g. muscle tone, muscle force, muscle coordination), only one study looked at the actual changes in muscle activation during recovery on a more fundamental level and the supposed relation between change in muscle activation patterns and improvement of functional abilities[6]. Kwakkel [30] already stated in 1998 that the training of ADL implies that we need to know more about the nature of co-ordination deficits in functional tasks. In other words we need to know more about the natural laws of co-ordination and control in the performance of such tasks. Thus far, assessment is focused on skills [31] rather than the quality of co-ordination patterns, although the latter forms an integral part of neurologically oriented methods like NDT and MRP. Hence the scientific basis of the existing and new treatment methods remains poor.
For the development of an evidence based rehabilitation medicine it is necessary that the employed treatment methods are firmly based on scientific principles.

Outline of the thesis.

The aim of this thesis is to provide a better understanding of the development of co-ordination patterns after stroke, their relation with functional performance and to what extend they can be manipulated.
In order to quantify muscle co-ordination surface electromyography (SEMG) is used. A frequently employed method to present the SEMG signal during gait is the SEMG profile, being the average SEMG smooth rectified pattern during one gait cycle. A disadvantage of this method is that it omits the step to step variability of the timing of the muscle activation patterns; information that might be relevant as a measure of performance of motor control and balance.

In **chapter 2**, a method is introduced in which every step in the gait cycle can be analyzed with respect to the timing of the muscle activation. For this purpose the approximated generalized likelihood (AGLR) algorithm [32] has been implemented and tested. In this way timing parameters can be calculated from a SEMG recording during gait and a measure for co-ordination is extracted. This method is used in **chapter 3, 4 and 5** in order to quantify changes in muscle co-ordination during recovery after stroke, and when walking is manipulated.

Chapter 3 describes the changes that occur during recovery after stroke. Although most of the treatment methods like for example NDT, Brunnström and

PNF presume that improvement of functional abilities is reached by improving function (e.g. muscle tone, muscle force, muscle coordination), none of these methods really investigated the actual changes in muscle activation patterns. Changes in so-called synergies almost always are evaluated with the Fugl-Meyer score. Literature describing muscle co-ordination patterns after stroke is scarce [4-7]. In **chapter 3** strong indications were found that muscle co-ordination patterns do not change over time. In the next two chapters attention is given to the manipulation of muscle activation patterns by external forces (walking aids and surgical intervention).

The experiment reported in **chapter 4** explores the changes in muscle activation patterns that occur when subjects with stroke have to walk with and without a walking aid. This chapter addresses the topic whether it is possible to manipulate co-ordination patterns by the use of a conventional rehabilitation assistive device. Furthermore from a clinical point of view the outcome of this study may help therapists in their decision making whether or not patients should use a cane or quad stick while walking.

Chapter 5 describes the changes in muscle activation patterns during walking in patients with cerebral palsy (CP), before and after hamstring lengthening. In rehabilitation, motor behaviour of CP children often is thought to be comparable to that of adult stroke patients. Since surgical interventions are rather common in the rehabilitation of these patients (in contrast to the rehabilitation of stroke patients) this group is used to evaluate whether a surgical intervention like hamstring lengthening changes muscle coordination patterns. It is assumed that when this is the case in CP children the same holds for adult stroke patients.

Chapter 6 addresses the use of an ankle-foot orthosis. The aim of this study was to investigate the effect of an AFO on walking ability in chronic stroke patients. In contrast to the other chapters this chapter focuses on the functional performance of an often used rehabilitation aid. The prevalence of stroke survivors with a dropped foot is approximately 18,000 [33]. Twenty chronic stroke-patients, wearing an AFO for at least six months were tested with and without their ankle foot orthosis, using a cross-over design with randomisation for the interventions. Although the topic of clinical relevance was also addressed in **chapter 2, 3 and 4** in this study this topic takes a rather prominent place. Based on scientific literature clinically relevant differences were a priori defined for walking speed and the Timed Up and Go test.

Chapter 7 addresses the knowledge transfer between a general hospital and an expert in another country in order to determine the suitability of stroke patients for a specialist surgical procedure: the split anterior tibial tendon transfer [34]. Gait analysis data were discussed with the expert using personal computers, an ISDN connection (128 kbit/s) and TCP/IP-based communication tools. The key issue in this study was whether the staff in the general hospital became better able to determine suitability for surgery.

Finally the thesis concludes with a summary and a general discussion in **chapter 8**.

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CHAPTER 2

Surface electromyography analysis for variable gait

D Roetenberg, JH Buurke, PH Veltink, A Forner Cordero and HJ Hermens

The surface electromyographic (SEMG) signal obtained during gait is often presented as the SEMG profile, the average SEMG activation pattern during one gait cycle. A disadvantage of this method is that it omits the step-to-step variability of the timing of the muscle activation patterns that might be relevant information as a performance measure of motor control and balance. In this paper, a method was used in which every step in the gait cycle could be analysed with respect to the timing of the muscle activation. For this purpose, the approximated generalised likelihood (AGLR) algorithm was implemented and tested. Results of the simulations show that the AGLR was much more accurate than a standard threshold criterion. Timing parameters could be calculated from a SEMG recording during gait and a measure for symmetry and coordination could be extracted. The amplitude distribution within and outside defined bursts is also presented to avoid the less precise classification into on and off patterns.

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1. Introduction

Surface electromyography (SEMG) has been shown to be an efficient tool in the analysis of pathological gait [1–4]. Often, the clinician is interested in the presence or absence of a particular muscle activity during a selected part of the cycle. The shape of the smooth rectified SEMG (SRE), averaged over several gait cycles, may be used to see if a particular muscle activity is abnormal in its time versus amplitude behaviour during its burst of activity. Applications where timing of muscle activation is important include studies of hemiparetic gait after stroke [5–7] and surgical planning in cerebral palsy [8,2]. Little is known about the muscle activation patterns arising during the different phases of recovery after stroke. Assessment of parameters such as the onset of SEMG, duration of activity and the cycle-to-cycle variability of these parameters may provide more insight into a patient's recovery after a stroke. Orthopedic surgery is often carried out to improve walking performance in cerebral palsy. Surface EMG is often used before planning tendon transfer surgery. The underlying assumption is that after a tendon transfer, the muscle activation pattern will not change very much [1].

Up to now, the timing of muscular action is usually assessed by visual inspection of the patterns or by studying the smooth rectified EMG (SRE) profiles. Timing parameters are derived from the profiles by applying a threshold to detect onset and offset of burst activity. However, there is no consensus in literature about the selection of parameters used to detect this [9].

Moreover, these methods introduce large systematic errors [10,11,4] and by averaging timing parameters over a number of steps, information about step to step coordination, consistency and variability is lost, which may be essential in analyzing motor control aspects, such as balance. Although good burst detectors are available [10,12,13], they are not commonly used in clinical applications. There are many studies in the recent literature that use simple threshold criteria [14-24]. It is possible that the mathematical complexity and extensive computational time to use burst detectors has prevented their more wide spread use in clinical practice.

In this study, we developed a method where the timing and amplitude of individual steps, their distribution and the relation between muscles can be analyzed. For this purpose, we implemented an objective motor onset detector based on the AGLR principle developed by Staude and Wolf [11]. In this paper, the background of the AGLR test will be considered and the performances of this test compared with a standard threshold criterion. Finally, by using a clinical example, we will show how the test can used to quantify parameters during gait.

2. Methods

2.1. Background theory

The general theory of the generalized likelihood ratio test has been well documented (for e.g. Refs. [25,26]). Suppose a series of SEMG samples y_1, y_2,…, y_k and a set of hypotheses ***H*** describing the statistical properties in terms of probability density functions (PDF) of these observations are available. Detecting the contraction onset can be presented as a testing problem between two hypotheses. The null hypothesis H_0 would be related to the muscle in relaxed state at time $1 \leq i \leq k$ with no change in statistical properties of the sequence, and hypothesis H_1 would be related to the contracted state with a change in statistical properties at some unknown change time $1 \leq j \leq k$. The log-likelihood ratio test which compares the logarithm of the ratio between the two parametric joint density function $p(y1, y2,…,yk|H_0)$ and $p(y1,y2,…,yk|H_1)$ for either hypothesis with a threshold *h* each time *k* a new sample is available:

$$g(k) = \ln \frac{p(y_1, y_2,\ldots, y_k|H_1)}{p(y_1, y_2,\ldots, y_k|H_0)} \gtrless h$$

If the ratio exceeds threshold *h*, hypothesis H_1 is more likely than hypothesis H_0, so contraction can be assumed. The implementation of the test depends upon the knowledge about the mean and variance describing the probability density functions before and after change. When the parameters remain unknown, the generalized likelihood ratio (GLR) test can be applied.

The GLR test is complex and computationally expensive, since the maximum likelihood of the unknown parameters must be computed for each possible change time $1 \leq j \leq k$, each time a new sample is available. In the approximated GLR (AGLR) test, a sliding window of fixed size *L* is continuously shifted along the data sequence y_1, y_2,…, y_k. For every location of the window, the maximum likelihood of the unknown parameters is determined from *L* data points covered by the window and the corresponding log-likelihood ratio *g*(*k*) is computed which is compared to a threshold (Figure 1, middle graph). As soon as a change has been indicated (alarm time t_a), the precise change time is estimated by maximizing the likelihood function [27] (Figure 1, lower graph).

The procedure is repeated with the previous change time replaced by the new change time until the end of the sequence is reached. The size of window *L*

should be chosen to be bigger than the shortest event to be detected. It is generally accepted that a muscle activation shorter than 30 ms has no effect in controlling the joint motion during gait [28], so *L* should be chosen >30 ms.
Two types of computer simulations have compared the performances of a standard threshold criterion and the AGLR test. The first simulation implied the detection of a discrete onset of SEMG activity. The SEMG has been synthesized by applying a step function with a variance of 50 in addition to a noise signal with a variance of 1 at a known change time. The assumed sample rate is f_s=1000 Hz. As is the case in real SEMG recordings, the onset of muscle activity is unlikely to behave as a step function, so the second simulation study involved the detection of a ramp of SEMG activity (see Fig. 2). At a known change time *t*=200, the variance linearly increased from 1 to 40 in 400 ms.

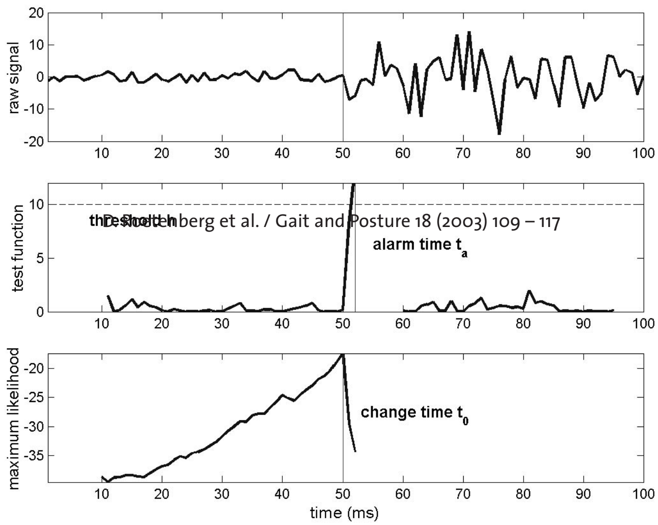

Fig. 1.
Upper: raw signal at *t* = 50, a change in variance.
Middle: when the test function exceeds threshold *h* , an alarm time is given at time instant t_a.
Lower: after an alarm has been given, the maximum likelihood of the test function is the estimated change time t_0 .

For the threshold criterion, the SEMG was rectified and filtered recursively with second order Butterworth filter with a cut-off frequency of 25 Hz. The threshold was based on the primary noise level of the signal. We used two thresholds namely: two and three times the S.D. of the primary noise level [29, 9]. For the AGRL test, the test window size *L* was set at 50 ms and detection threshold, *h*, at 15. For 1000 realisations of the burst-modulated noise described above, the onset times were estimated for the AGLR and threshold algorithm. The estimation errors in onset detection time of the burst $\epsilon = t_{\text{exact}} - t_{\text{estimated}}$ were computed for each method and realization and plotted in a histogram.

2.2. Parameter extraction

The procedure of burst detection was performed for the SEMG sequences of all recorded muscles during a gait trial. All detected on- and offset times were then normalized in time using the stride length starting from the related heel strike. In many cases of neurologically impaired subjects, there is a continuous muscle activity throughout the complete gait-cycle with superimposed bursts of activity. In these cases, the simple classification of activity into 'on' and 'off' is of limited value. Adding the description of amplitudes in- and outside the burst together with the maximum amplitude for each step will provide more valid data. Muscles do not act separately.

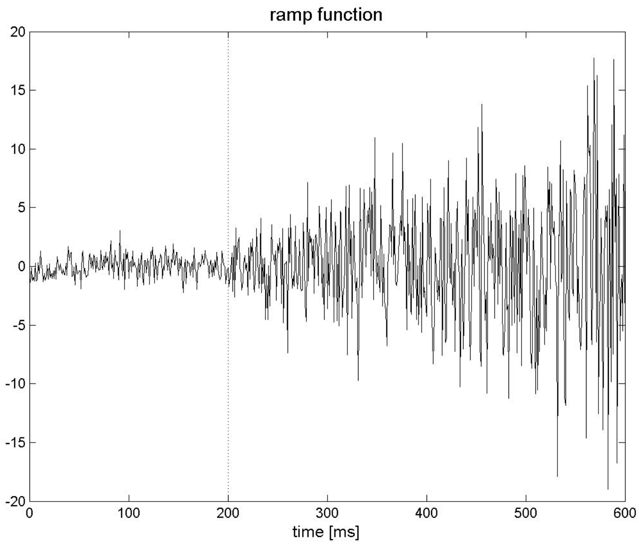

Fig. 2.
Realization of a computer generated noisy ramp signal. At t=200, the variance linearly increases from 1 to 40 in 400 ms.

The contraction of one muscle is always related to at least one other muscle. One of the most common representations of this is the co-contraction [18]. However, not only antagonist muscles are related. Additional co-activation of different muscles of the lower extremity (synergies) is very common in central nervous disorders. The relation between on- and offset times of muscles can be obtained by plotting the absolute on- and offset times with respect to the related heel strike for each step in a scatter diagram and calculating the confidence ellipse. Assuming a Gaussian distribution, the probability that the variables will fall within the area marked by the ellipse is determined by the value of the coefficient that defines the ellipse (e.g. 95%). The orientation of this ellipse is determined by the sign of the linear correlation between the two variables (the longer axis of the ellipse is superimposed on the regression line). If the ellipse is small, the timing of the bursts is consistent over the different steps. If the ellipse is long-drawn, the on-and offset times are highly correlated; e.g. if a burst starts late, it ends late.

2.3. SEMG recordings

In the clinical trials, surface EMGs of four muscles (left and right gastrocnemius medialis and tibialis anterior) were recorded during walking using a wireless 16-channel Glonner Biotel 99 EMG amplifier with a cut-off frequency of 600 Hz/–3 db and a first order 16 Hz high-pass filter. Raw SEMG signals were digitized at 1000 Hz sample rate with 12 bits resolution and stored on a VICON 370 system. The SEMG signals were filtered off-line (third order Butterworth high pass filter –3 dB at 20 Hz) using Matlab (The Mathworks Inc.). 'Meditrace pellet No. 1801 graphics control' electrodes were used. Electrode size, inter electrode distance and electrode placement and skin preparation were according to the 'SENIAM' protocol [30]. Footswitches were used to determine stance and swing phase. Walking speed was measured over a distance of 7.5 m using light gates to detect start and stop. The SEMG processing and parameter extraction was performed with Matlab

3. Results

Fig. 3 shows the results of the simulation of the step function. With the selected detection parameter settings, both detectors were able to detect the onset of the activity. However, the accuracy was quite different. The threshold criterion provided biased estimates if the threshold was not chosen properly (upper graph). When the threshold was better tuned for the specific signal to noise ratio (middle graph), the results were more accurate but the estimates were still quite variable. The estimation error shown in the lower graph demonstrates the advantage of using the AGLR algorithm. The accuracy of detection had improved substantially. The results of the ramp function simulations in Fig. 4 show that onset detection with the threshold criterion became even more variable and biased. Note that the distribution of the onset error for threshold h=3 was much larger than of threshold h=2, because of the increasing variance in time. The onset of the ramp with the AGLR was well detected. The estimation error was obviously more variable than the detection of a step. Decreasing the threshold h made the algorithm more sensitive for small changes in the sequence.

3.1. Clinical examples

The possibilities and advantages of the new method in analysing muscle activation patterns are illustrated using the recorded gait patterns of different neurologically impaired patients.

3.1.1. Timing

In Fig. 5, an example of burst detection with the AGLR is given. The graph shows 15 s of recorded SEMG of the gastrocnemius of a stroke patient with a left hemiparesis.

The on- and offset times have been estimated for ten consecutive strides and normalized in time using the stride length starting from the related heel strike. The muscle activation patterns are presented in a new format (Fig. 6) that shows not only the ensembled average profiles of the left and right tibialis anterior and gastrocnemius together with their S.D. but also the timing information along the x-axis. The median on and off time is presented in a black solid bar connecting the median on time with the median off time. The smaller little grey bars indicate the 25th and 75th percentiles of the median on and off times.

The dotted vertical lines, representing average toe-off, clearly indicate the asymmetry between the left and right leg. When considering the timing of the muscles, the most striking feature is the early onset of the burst of the left gastrocnemius muscle (around heel strike) showing a large amount of peak activity, probably caused by a decreased threshold for stretch reflexes [31]. A second characteristic is that the offset of the burst is too early. The main difference in timing between left and right of the tibialis anterior muscle is seen in the length of the burst after heel strike. The burst of the non-affected side is much longer and ends at ≈25% of the gait cycle, whereas the activity of the affected left tibialis anterior ends at ≈7% of the gait cycle.

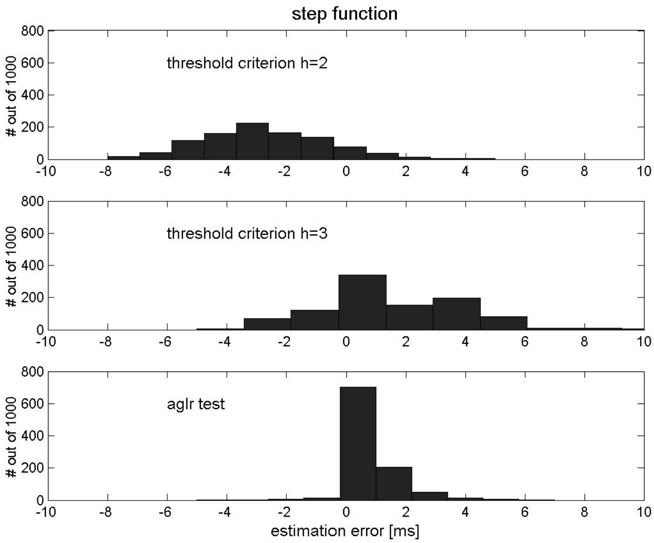

Fig. 3
Estimation error in ms of 1000 realizations of step modulated noise with a threshold criteria with *h*=2 (upper), *h*=3 (middle) and with the approximate generalized likelihood ratio test (lower).

3.1.2. Amplitude

In Fig. 7(left side), the ensembled average profile together with the S.D. and timing parameters of the tibialis anterior of a CP patient are presented. On the right side, the median amplitude outside the burst (off) and inside the burst (on) together with median maximum amplitude (max) are presented using box plots. The ratio between activity inside and outside the bursts provides a relevant indication of the motor control capacities of the patient. Muscle activity outside the bursts may be considered beyond the voluntary control of the movements generated by the subject.

In this particular case, the median amplitude outside the burst was ≈20 uV, which is an indication that the tibialis anterior is active and could contribute to a possible varus position in the ankle during stance.

3.1.3. Coordination

In Fig. 8, the on and off times of the left and right gastrocnemius are plotted together with their 95% confidence intervals. The recording is from the same subject as the examples in Fig. 5 and Fig. 6. It can be seen that most of the onsets of the left side occur before initial contact. Note that the surface area of the ellipse of the healthy right side is larger compared to the ellipse of the affected left side. An explanation can be that the affected side of the brain has lost part of is adaptive ability and can only control the muscle in a rather fixed pattern. An increase in overlap of the left and right ellipsoids indicates more symmetry in the muscle activation patterns.

In this example, the on and off times of the left and right gastrocnemius muscles are plotted against each other. Another possibility is to plot on times of two different muscles against each other and calculate the confidence intervals. In this way, a measure for inter muscular coordination of the same leg is obtained.

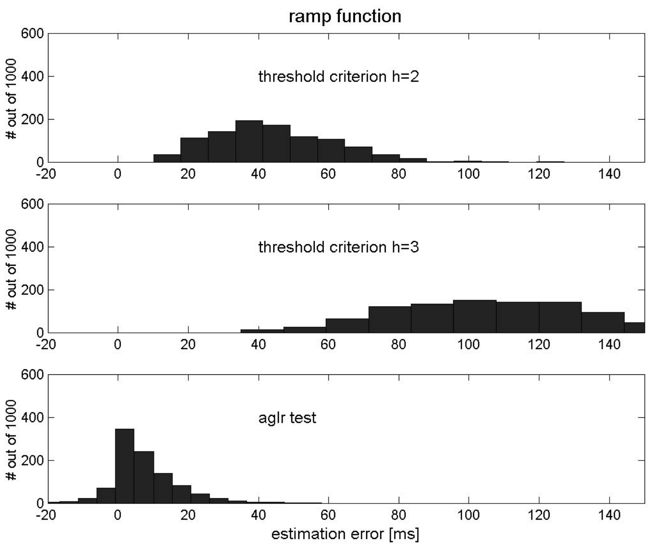

Fig. 4
Estimation error in ms of 1000 realizations of ramp modulated noise with a threshold criteria with *h*=2 (upper), *h*=3 (middle) and with the approximate generalized likelihood ratio test (lower).

4. Discussion

We have presented a method to analyse individual steps to obtain information with respect to the timing of muscle activation. For this type of analysis, a simple threshold is often used, which is applied to the smooth rectified SEMG pattern. We investigated the use of the AGLR test to quantify the timing parameters for

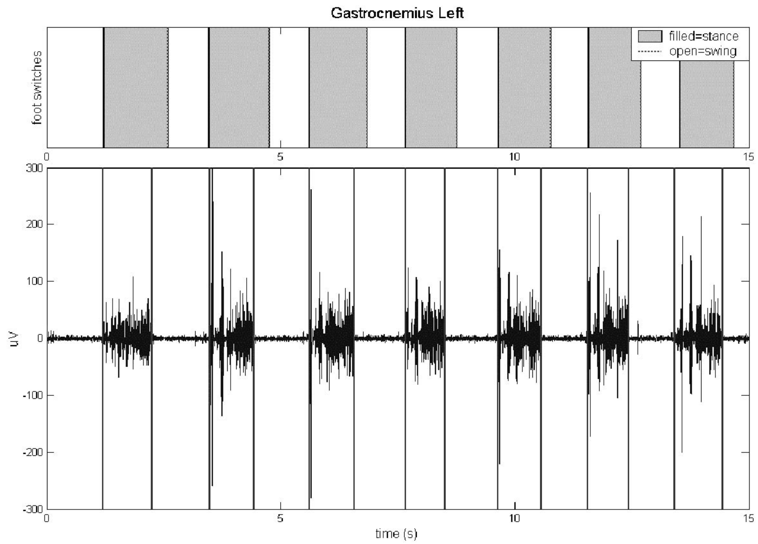

Fig. 5
Recording of the left gastrocnemius for 15 s of a stroke patient.
Upper: footswitch contacts: filled area: stance phase, open area: swing phase.
Lower: SEMG segmentation in seven bursts of activity with the AGLR test. The walking speed is 0.48 m/s.

each step. From the present computer simulations and from the literature [27], it can be seen that estimating change times with the AGLR test is much more accurate than with a threshold method. This was seen especially in the detection of the ramp function where the threshold criterion introduced highly biased and variable estimates. If, for example, a threshold of 20 μV [18] had been used in the clinical examples of Fig. 6, some steps would have shown no activity, while it can be seen that there is a clear burst of activity. Staude [27] also studied the performance dependence on detection thresholds. The detection error for the AGLR is not very sensitive for deviations in threshold *h*, although the detection rate decreases slightly when the choice of *h* is not optimal. Staude and Wolf [11] recommended the use of an adaptive whitening filter in addition to the AGLR test. We have tested a few trials with and without the use of such a filter, but only found differences within a few ms, which we considered not worth the computational effort. Each burst detector has a trade-off between a robust but selective detection process. It should be emphasized that tuning of parameters of the algorithm can involve subjective criteria [11]. Current methods to analyse the muscle activation patterns during gait often utilize the muscle activation pattern, expressed as the smooth rectified EMG pattern, averaged over a number of steps.

Although this method provides useful information on the course of the muscle activation, it does not show information on the variability of the timing, which is an important aspect of performance of motor control and balance. Kleissen et al. [1] consider that ensemble averaging is not appropriate where the assumption of repeatable muscle activation cycles is violated, which is often the case where there is impaired neuromuscular control.

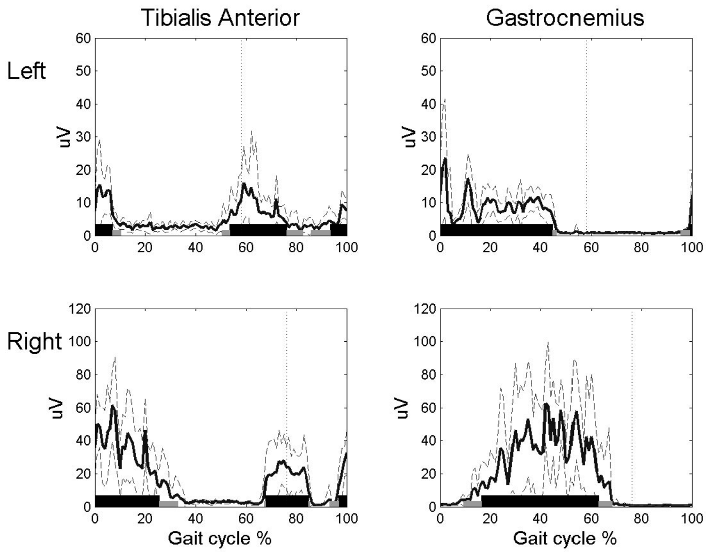

Fig. 6
SRE profiles and S.D. of the left and right tibialis anterior and gastrocnemius. The dotted vertical line shows the toe-off point. Along the x-axis, the timing is illustrated in black and gray lines. The start of the gray line is the 25th percentile of the on-times. The start and end of the black lines denote the median on and off times, respectively. The end of the gray line is the 75th percentile of the off-times.

Commonly, the envelope of the myoelectric signal obtained during gait is used to study muscle activation. We propose a format in which the timing is presented along the x-axis of the linear envelope. The on- and offset timing distribution gives relevant additional information, which cannot be obtained from the ensemble average profiles. From the estimated on- and offset times, other clinical relevant parameters, such as co-contraction and stability of the muscle activation pattern, can be derived. Confidence intervals of related on and off times are a measure for symmetry and coordination. The approximate classification into on and off patterns can be partly overcome by considering amplitudes in and outside the defined bursts.

The advantage of the described method is an increased objectivity in the analysis of on- and offset times and related coordination patterns using a fairly simple algorithm. The automated software provides reduced the time required to perform the analysis. Differences between successive measurements in, for example, the recovery after stroke or evaluation of surgery on CP patients can be observed and analysed statistically. Future research will aim at expanding the presented method to produce a clinically relevant and comprehensive description of muscle synergies possible.

In conclusion, the proposed method for burst detection is considered to be more accurate then a standard threshold method. It offers a new representation of EMG profiles containing additional relevant information on the variability of timing of muscle activation. It also offers quantitative ways to assess muscle activation levels on intra and inter muscle coordination due to its high reliability.

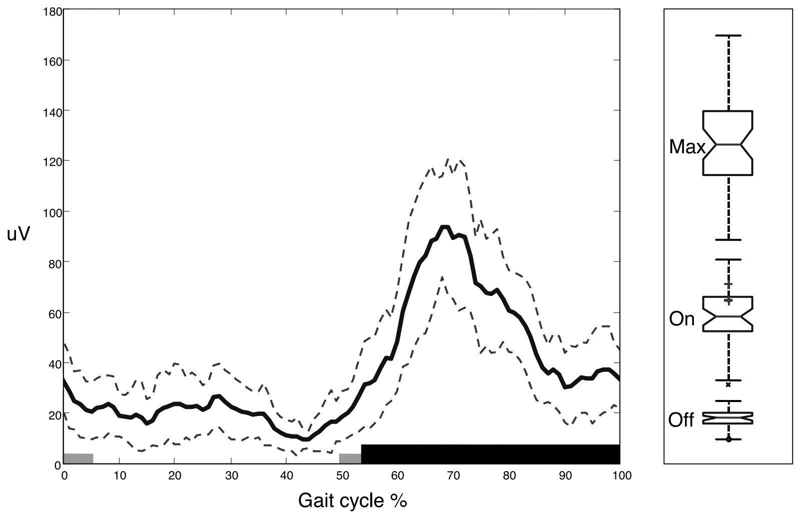

Fig. 7.
Distribution of the average amplitudes for ten steps inside and outside the burst and the maximum amplitude within the burst of the tibialis anterior of a CP patient.

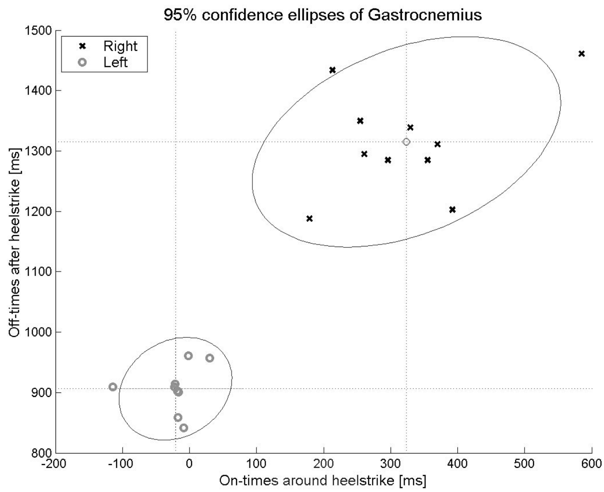

Fig. 8
On and off times of the left affected (*o*) and right unaffected (*x*) gastrocnemius with their 95% confi-dence
intervals of a stroke patient. Timing of activation and deactivation are plotted in absolute time with respect to the heelstrike of the related side.

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CHAPTER 3

Recovery of walking after stroke. What really changes?

JH Buurke, HJ Hermens, AV Nene, CV Erren-Wolters, G Zilvold

Most of the methods used in clinical practice assume that improvement of functional abilities is facilitated through the improvement of muscle co-ordination. However little is known whether and how these co-ordination patterns change, the speed at which they change and the relation with functional recovery. The present study focuses on the longitudinal description of changes in the neuromuscular co-ordination during the recovery of walking after stroke and its relationship to functional recovery. Thirteen patients diagnosed with a first unilateral ischemic stroke participated in the study. Functional recovery of walking was measured using the Rivermead Mobility Index (RMI), Functional Ambulation Categories (FAC), Barthel Index (BI), Trunc Control Test (TCT), Motricity Index (MI) and comfortable walking speed. Surface electromyography (SEMG) of the Erector Spinae, Gluteus Maximus, Gluteus Medius, Rectus Femoris, Vastus Lateralis, Semitendinosus, Gastrocnemius and Tibialis Anterior of both legs were used to quantify co-ordination patterns. Assessment took place during walking at 3, 6, 9, 12 and 24 weeks post stroke. Timing parameters of the SEMG patterns were calculated, using an objective burst detection algorithm and together with the measurements of functional recovery statistically analysed.
All functional measures, except the TCT, improved over time and showed statistically significant differences. In contrast SEMG patterns did not change over time.
The constant SEMG patterns over time suggest, that the functional improvement of gait is related to other mechanisms than to restoration of co-ordination patterns of the affected leg or compensating changes in co-ordination patterns of the unaffected leg.
Whereas most treatment methods assume that functional improvement is facilitated by the improvement of muscle co-ordination, this assumption could not be confirmed in this study.

Submitted for publication

1. Introduction

Worldwide hundreds of thousand cases of stroke are reported each year. The incidence of stroke in The Netherlands is approximately 30,000 per year and the prevalence is 120,000 patients [1] in a total population of 16.280.000.
The consequences of a stroke could be seen as a massive disturbance in information processing, causing not only motor deficits but also cognitive impairments and behavioural problems [2]. Although neuro-psycological deficits and personal changes are the hardest to cope, regaining the ability to walk is one of the most important goals for the stroke patient [3]. Of those who survive the acute phase about 11% are unable to walk and 55% have moderate to severe walking disabilities. Hence relearning to walk is an important aspect of treatment. In the early stage most patients are not able to perform movements or these movements are largely uncoordinated. It is believed that after several weeks new co-ordination patterns evolve. Although most of the therapies used in clinical practice are based on the assumption that improvement of functional abilities is facilitated through the improvement of muscle co-ordination very little is known whether the original co-ordination patterns change, the speed at which the change occurs and the type of patterns that come with the change [4-7]. This lack of fundamental knowledge has originated in more than nine different methods that are used in today’s clinical practice [8-16]. Although it is reported that every single method has a positive effect on functional recovery of stroke patients there is no clear evidence in scientific literature [17-28] that puts one method above the other [29].
Most studies evaluate outcome using clinical measures. These measures explore one or more items on "function" and/or "ability level" such as: symmetry, synergies, muscle force, muscle tone, walking and Activities of daily living. Only one study, until now, was focused on the actual changes in muscle co-ordination during recovery and the presumed relation between the change in muscle co-ordination and improvement of functional abilities [6]. Hence the scientific basis of the existing treatment methods remains poor.
Treatment would benefit substantially when more insight exists in the recovery process and the mechanisms that may influence the recovery. Therefore this study focuses on the description of longitudinal changes in the neuromuscular co-ordination of leg and trunk muscles during the recovery of walking after stroke and its relationship to functional recovery.

2. Methods

2.1. Subjects

Stroke patients recruited from the Rehabilitation Centre 'Het Roessingh' in Enschede, The Netherlands were included in the study if they had a first unilateral ischemic stroke and sufficient cognitive abilities (Mini-mental State Examination (MMSE $\geq$ 22) [30]. Patients were excluded if they suffered more than one stroke or had other physical conditions adversely influencing walking ability. The study was approved by the medical ethics committee of the Rehabilitation Centre. All subjects signed an informed consent before participating in the study.

2.2. Experimental set-up and procedures

Assessment of patients took place at 3, 6, 9, 12 and 24 weeks after stroke. Patients were included in the study as soon as they were referred to the rehabilitation centre.

Functional recovery was measured using the Rivermead Mobility Index (RMI) [31], Functional Ambulation Categories (FAC) [32], Barthel Index (BI) [33], Trunc Control Test (TCT) [34], Motricity Index (MI) [35] and comfortable walking speed [7, 36]. In addition clinical measurements such as Range of Motion (ROM) of hip, knee and ankle, as well as hamstrings length, clonus of calf muscles, sensation and Ashworth scores [37] of calf muscles were performed.

Assessment of the muscle co-ordination was carried out using surface electromyography (SEMG) of the Erector Spinae, Gluteus Maximus, Gluteus Medius, Rectus Femoris, Vastus Lateralis, Semitendinosus, Gastrocnemius, and Tibialis Anterior muscles of both legs during walking. 'Meditrace pellet # 1801 graphics control' electrodes were used. Electrode size (1 cm2), inter electrode distance (2 cm), electrode placement and skin preparation were according to the 'SENIAM' protocol [38]. A wireless 16-channel Glonner Biotel 99 EMG amplifier with a cut-off frequency of 600hz / -3db and a first order 16 Hz high-pass filter. Raw SEMG signals were digitized at 1000Hz sample rate with 12 bits resolution and stored on a VICON 370 system. The raw SEMG signals were rectified and filtered off-line (third order Butterworth high pass filter -3dB at 20 Hz) using Matlab (The Mathworks Inc.).

Subjects walked along a 10 m-walkway with the middle 7.5 m delineated for data collection with photo electric beams. Simultaneous recordings were made of foot-floor-contacts as subjects walked over the walkway at a comfortable self-selected speed with customary assistive devices and shoes. The footswitches consisted of an aluminium conductive sheet covering the sole of the shoe in conjunction with a conductive rubber mat [39]. The SEMG processing and parameter extraction was performed using Matlab. For each subject, during each session, at least 10 gait cycles were recorded and stored for further analysis.

2.3. Data reduction

An objective method was implemented to automatically analyze the muscle activation patterns [40]. This method consists of an automatic burst on and off detector of SEMG signal based on the approximated generalized likelihood ratio (AGLR) principle described by Staude and Wolf [41]. This algorithm was used to analyze the raw SEMG signals of every stride in the gait cycle separately with respect to the on- and off-times of muscle activation. Subsequently, all detected on- and off-times were normalized in time using the stride time starting from the related heel strike (fig.2).

2.4. Statistical analysis

Improvement in functional recovery and clinical examination were analyzed using the Friedman test. On- and off-times of the selected muscles were analyzed separately. Since the gathered data were not normally distributed the median on- and off-times in percentage of the gait cycle (together with the 25 and 75 percentile) were calculated for each subject. Differences in median on- and off-times were statistically analyzed using the Friedman test. If appropriate the Wilcoxon signed ranks test was used for post hoc testing.

The level of statistical significance for multiple testing was set according to Simes [41]. This is a modification to the Bonferroni procedure which is more powerful whilst still ensuring that the overall significance level does not exceed α for positively correlated variables [42]. All statistical analysis was carried out using SPSS version 11.5.

3. Results

3.1. Study population

During eighteen months fifteen patients were included (table 1). Two patients dropped out: one suffered from an ulcer under the foot and one patient was lost to follow up (stopped participation due to early discharge). Thirteen patients, six

men and seven women participated in the study. The mean age ± standard deviation (SD) was 56.0±13.5 years (range 36 – 75 years). Eight patients suffered a right hemispheric stroke and five patients suffered a left hemispheric stroke. All patients had sufficient cognitive abilities to participate in the study. Median MMSE was 28 (range 25 – 29).
Referral form the hospital to the rehabilitation centre was not always reached within 3 weeks post stroke. Therefore the start of clinical measures and gait analysis differed from patient to patient. This resulted in 8, 4 and 1 patients that were referred to the rehabilitation centre within 3, 6 and 9 weeks post stroke respectively. Most of the patients were not able to walk especially during first measurement. This resulted in 4, 7 and 2 patients having the first gait analysis at 3, 6 and 9 weeks post stroke respectively. The three different situations were statistically analyzed separately.

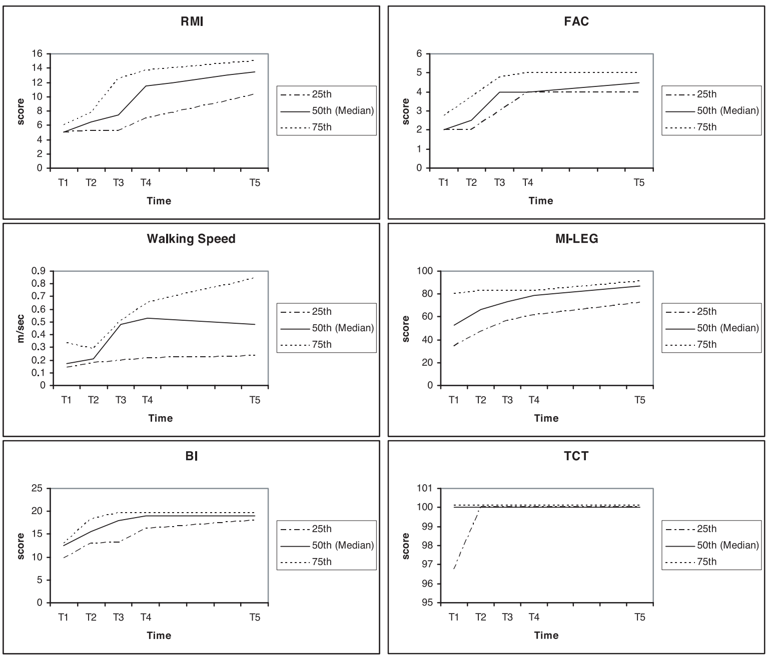

Fig. 1.
Clinical measures over time.
T1= 3 weeks after stroke, T2= 6 weeks after stroke, T3= 9 weeks after stroke, T4= 12 weeks after stroke, T5= 24 weeks after stroke. RMI= Rivermead Mobility Index, FAC= Functional Ambulation Categories, MI-Leg= Motricity Index leg, BI= Barthel Index, TCT= Trunc Control Test.

3.2. Functional recovery

Although small individual changes in muscle tone, clonus and muscle length over time were observed, clinical examination of ROM, hamstring length, clonus and Ashworth scores of calf muscles did not reveal any statistical significant difference. Conversely all functional measures (fig.1), except the TCT, improved and showed statistical significant differences; RMI ($p < 0.000$); FAC ($p < 0.002$); walking speed ($p < 0.003$); MI-leg ($p < 0.000$); BI ($p < 0.001$); TCT from 8 subjects at 3, 6, 9, 12 and 24 weeks ($p = 0.144$); TCT from 12 subjects at 6, 9, 12 and 24 weeks ($p = 0.194$); TCT from 13 subjects at 9, 12 and 24 weeks ($p = 0.135$).

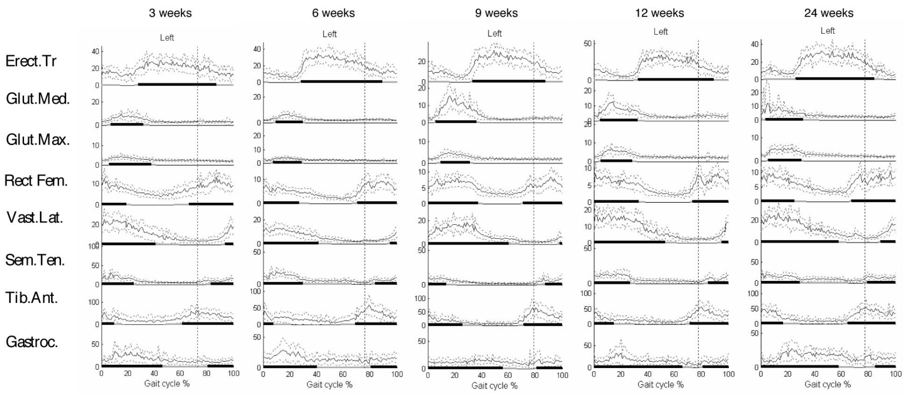

Fig. 2.
Typical example affected leg:
Stride normalized Smooth Rectified EMG (SRE) of the eight muscles of the affected leg is presented 3, 6, 9, 12 and 24 weeks post stroke.
Erect.Tr.=Erector Trunci, Glut.Med.=Gluteus Medius, Glut.Max.=Gluteus Maximus, Rect. Fem.=Rectus Femoris, Vast.Lat.=Vastus Lateralis, Sem.Ten=Semitendinosus, T.b.Ant.=Tibialis Anterior, Gastroc.=Gastrocnemius.
Along the y-axis the amplitude of the SRE in microvolts is presented. Along the x-axis of the stride normalized SRE graphs, the timing as derived from the burst detection algorithm is shown in black and gray lines. The median on and off time is presented in a black solid bar connecting the median on time with the median off time. The somewhat smaller little grey bars indicate the 25 and 75 percentiles of the median on and off times. The dashed vertical line represents toe off.

3.3. Muscle co-ordination

Figure 2 shows a typical example of the muscle activation patterns of the affected side of a left sided hemiplegic patient at 3, 6, 9, 12 and 24 weeks post stroke. Although amplitude of the SEMG of the different muscles varies from time to time no clear differences in timing can be detected.

Calculation of timing parameters in the total population showed clearly delayed median off-times in Gluteus Medius (35% of gait cycle), Gluteus Maximus (35% of gait cycle), Vastus Lateralis (44% of gait cycle) and Semitendinosus (30% of gait cycle). Even more prominent abnormalities on the affected side are found in the Erector Spinae and Rectus Femoris. None of the patients showed a SEMG pattern that came close to the normal two phasic pattern of the erector spinae [43]. And only 6 out of the 13 patients showed a second burst during initial swing in the Rectus Femoris. Median on-times off all muscles except the Gastrocnemius (0.05 % of the gait cycle) were close to normal.

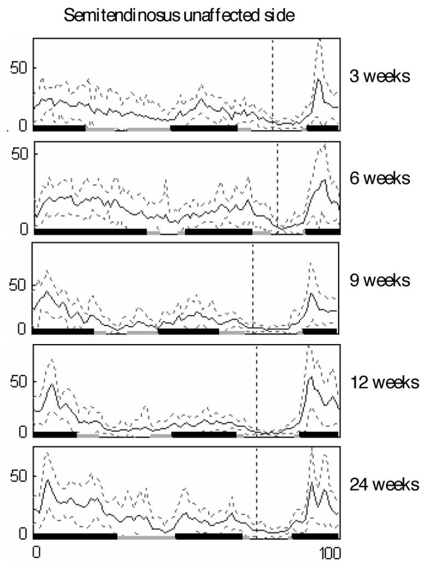

Fig. 3.
The extra but consistent burst of the semitendinosus of the unaffected leg during stance measured at 3, 6, 9, 12 and 24 weeks post stroke.
Along the y-axis the amplitude of the SRE in microvolts is presented. Along the x-axis of the stride normalized SRE graphs, the timing is shown in black and gray lines. The median on and off time is presented in a black solid bar connecting the median on time with the median off time. The somewhat smaller little grey bars indicate the 25 and 75 percentiles of the median on and off times. The dashed vertical line represents toe off.

Statistical analysis of median on- and off-times for the eight muscles in the affected leg of all patients showed no significant differences. Although Friedman tests of the affected leg for the on-times of the Gluteus Medius, Gluteus Maximus, Rectus Femoris and Semitendinosus and off-times of the Gluteus Maximus and Rectus Femoris showed p-values < 0.05, after correction of α according to Simes [44] no significant differences were found

Not only the affected leg but also the unaffected leg showed SEMG patterns that are considered abnormal. Calculation of timing parameters in the total population showed clearly delayed off-times of the Gluteus Medius (53% of gait-cycle), Gluteus Maximus (42% of gait-cycle), Rectus Femoris (53% of gait-cycle), Vastus Lateralis (58% of gait-cycle), Semitendinosus (32% of gait-cycle) and Tibialis Anterior (37% of gait-cycle). The most prominent abnormality in the unaffected leg however was the extra burst of the Semitendinosus during stance (fig. 3). Eight of the thirteen patients showed this abnormal extra burst. On-times off all muscles except the Gastrocnemius and Tibialis Anterior were close to normal. The median on-time of the Gastrocnemius was to early (12% of the gait-cycle) and the on-time of the Tibialis Anterior was delayed (76% of the gait cycle) as compared to normal.

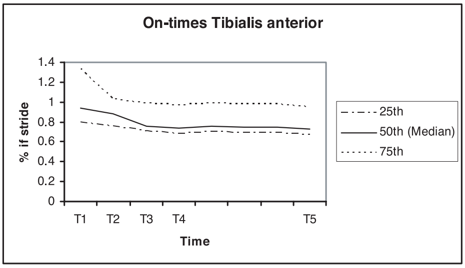

Fig. 4:
On-times of Tibialis anterior 3 (T1), 6 (t2), 9 (T3), 12(T4) and 24 (T5) weeks after stroke.

Statistical analysis of the muscles of the non-affected leg showed *p*-values < 0.05 for the on-times of the Gluteus Maximus, Vastus Lateralis, Semitendinosus, Tibialis anterior and the off-times of the Erector Trunci, Gluteus medius and Gastrocnemius. After correction of α according to Simes [44] only a significant decrease in on-times over time was found for the Tibialis Anterior (fig. 4).
Post hoc analysis using the Wilcoxon signed ranks test revealed significant differences when T2 was compared to T3 (p=0.004), T4 (p=0.003) and T5 (p=0.003) and T3 was compared to T4 (p=0.013) and T5 (p=0.019). This means that the decrease in on-times took place during the first 9 weeks.

4. Discussion

Literature on automated detection of timing parameters from SEMG in order to study changes in muscle co-ordination during recovery of gait after stroke is scarce. In our study changes over time in muscle co-ordination have been investigated by calculating on- and off-times from SEMG signals using the approximated generalized likelihood ratio principle described by Staude and Wolf [41]. This method is considered to be more accurate then a standard threshold method [40]. As far as we know this is the first time that an automated burst detector is used to detect changes in muscle activation timing.

Knutsson [4] visually evaluated the SEMG and only looked at one moment in time. This study of Knutsson resulted in the description of four different EMG patterns. Changes from one pattern to another however can not be confirmed in this way. Shiavi [5] also visually evaluated SEMG signals recorded in the early (1 to 10 weeks post stroke) and in late recovery period (6 to 24 months post stroke) and used the classification of Knutsson to describe changes in synergy patterns over time. Richards [6] used SEMG to study the relationship of gait speed to clinical measures and muscle activation. She however looked at the changes in amplitude of the SEMG in pre-selected intervals of especially the Triceps Surae and Tibialis Anterior rather than timing parameters. Mulroy [45] described differences in muscle activation patterns between groups of patients that were classified using non-hierarchical cluster analysis of temporal-spatial and kinematic parameters of walking. Timing parameters from SEMG however were calculated using a simple threshold.

During recovery walking speed of the stroke subjects in this study increased (fig. 1). Hoff (2002) [46] showed that walking speed in healthy subjects does not influence timing parameters. Whether or not this holds for stroke patients is unclear. Experiments concerning changes in walking speed with stroke patients during this longitudinal study however did not reveal any difference in timing of the muscles measured.

The SEMG patterns measured during the longitudinal study in both affected and unaffected leg showed abnormal SEMG patterns. This was also reported before by Shiavi [5]. Most prominent abnormalities on the affected side were shown in the Erector Spinae and Rectus Femoris. The abnormal activity of the Erector Spinae during stance phase of the affected leg (fig. 2) should be judged as a compensation for lost stability in trunk and affected leg [47]. The missing second

burst in the Rectus Femoris can be associated with the absence of a second burst due to the low walking speed as reported by Nene [48] or abnormal activation patterns with delayed off-times thus incorporating the second burst.
The most prominent abnormality on the unaffected side, the second burst of the semitendinosus during stance, has also been reported by Shiavi [5] who stated: "Whether this abnormality is due the stroke itself or biomechanical compensation is not clear". This extra burst however is also seen in amputee patients [49] and can be associated with a diminished push-off of the affected leg in pre-swing. According to Czerniecki [50] the increase in mechanical work of the hipextensors and ankle plantarflexors may partially compensate for the reduced push-off function in the prosthetic limb. In stroke patients, contraction of the hip extensors (including semitendinousus) of the unaffected leg causes an extension of the unaffected hip; thereby facilitating the swing of the affected leg thus compensating for the diminished swing caused by the lack of push-off and reduced force of hip flexors on the affected side.

Over time no change in clinical measures like ROM and muscle tone was found. From the start of inclusion hip flexion, hip extension, knee flexion, hamstring length and length of Triceps Surae were less as compared to normal. Though preservation of ROM and normalisation of muscle tone is part of daily routine physical therapy no change in clinical measures over time in ROM and muscle tone over time was found. This might be an indication that ROM and muscle tone are not easily influenced by normal daily routine physical therapy.
As described by Twitchell [51] restoration of motor function proceeds through a stepwise sequence: (1) The areflexic flaccid paralysis (2) return of reflexes (3) increase of muscle tone and development of spasticity (4) appearance of voluntary movements as part of synergies (5) performance of movement outside synergies and finally stage (6) in which muscle tone is normalized and occurrence of normal movements. Though individual changes in tone were present, no significant change in muscle tone and muscle activation patterns over time for the group could be detected. Changes in force and function however were statistically significant. This would suggest a contrast with the steps described by Twitchell or as we did not find such changes most of the steps already occurred before inclusion. The latter would be in accordance to Richards [7] who stated that recovery lasts only up to 6 weeks after stroke.
The contrast in progress in functionality represented by increasing scores on RMI,

FAC, MI, BI and walking speed, with the largely constant SEMG patterns over time is striking. Whereas most treatment methods assume that improvement of functionality is facilitated by the improvement of muscle co-ordination, in this study this assumption cannot be confirmed. The constant SEMG patterns over time suggest, that the functional improvement of gait is related to other mechanisms than to the restoration of co-ordination patterns of unaffected and affected leg. This conclusion is in accordance to de Haart [52] who stated that "The clear lack of normalization for measures reflecting static and dynamic aspects of postural asymmetry suggests that the functional improvements in balance and gait must be more related to other mechanisms than to the restoration of support functions and equilibrium reactions of the paretic leg".
The results do not seem to match with results from a recent study by Miyai et al [53] involving functional neuroimaging techniques, showing that functional recovery may be associated with cortical reorganization of the affected and unaffected hemisphere. In their study functional outcome was measured using Fugl-Meyer scores, cadence and swing phase. No measures involving kinematics or muscle co-ordination were performed. Increasing activation of the sensorimotor cortex and premotor cortex, as shown by Miyai, however are no clear evidence that quality of locomotion improves. Increasing muscle force, as demonstrated by increasing MI scores in our study, improved balance characterized by a reduction in postural sway and a reduction in visual dependency as shown by de Haart [52] and the development of novel adaptive strategies following therapy and learning as reported by Mulder [54], are more likely to have a positive influence on the improvement in functionality and increasing involvement of the sensorimotor cortex and premotor cortex. This coincides with the increasing evidence that treatment should be functional and goal oriented [55].

5. Conclusion

Recovery of walking after stroke is characterized by functional improvement as measured by RMI, FAC, BI, MI and walking speed. Whereas most treatment methods assume that functional improvement is facilitated by the improvement of muscle co-ordination this assumption could not be confirmed in this study. The largely constant SEMG patterns over time suggest, that the functional improvement of gait is related to other mechanisms than to the restoration of co-ordination patterns of both unaffected and affected leg.

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CHAPTER 4

The effect of walking aids on muscle activation patterns during walking in stroke patients

JH Buurke, HJ Hermens, CV Erren-Wolters and AV Nene

The purpose of this study was to investigate changes in muscle activation patterns with respect to timing and amplitude that occur when subjects with stroke walk with and without a walking aid. This knowledge could help therapists in deciding whether or not patients should use a cane or quad stick while walking.
Thirteen patients suffering from a first unilateral ischemic stroke participated in the study. Surface electromyography (SEMG) of the erector spinae, gluteus maximus, gluteus medius, vastus lateralis, semitendinosus, gastrocnemius and tibialis anterior of the affected side were measured during three different conditions: (1) walking without a walking aid, (2) walking with a cane and (3) walking with a quad stick. Timing and amplitude parameters of the activation patterns were quantified using an objective burst detection algorithm and statistically evaluated.
Results showed a statistically significant and clinically relevant decrease in burst duration of both erector spinae and tibialis anterior when walking with a cane. The amplitude of the vastus lateralis and tibialis anterior dropped when patients walked with a cane and quad stick.
The use of a cane should be considered when therapy is given to stroke patients to achieve normal muscle activation patterns.

Gait & Posture. Article in Press.

1. Introduction

Regaining the ability to walk is a major goal during the rehabilitation of stroke patients [1]. Factors important to reach this goal are early, functional, goal oriented intensive training [2]. In clinical practice direct assistance during exercise is intentionally restricted and strong personal involvement of the patient is encouraged. Walking aids are often used to maintain safety and increase independence during gait training [3].

Different opinions exist on the effect of walking aids on the gait pattern of stroke patients. Kuan et al. [3] stated that the use of walking aids increases stability, reduces the chance of falling and improves independent walking. Lennon et al. [4], on the other hand, mentions that the use of walking aids might hinder the training of a symmetrical walking pattern, and Davies [5] considered that a walking aid should only be given to a stroke patient when he or she is able to walk without one. Very little evidence exists to support these assumptions. Only six studies investigated the effects of a walking aid on gait [3, 6-10]. Results suggest that the use of a cane had positive effects on stride length and walking speed [3]. Other studies [8] and [10] did not measure significant change in walking speed when using a walking aid. Furthermore, neither significant effects on the symmetry of the walking pattern [9, 10] nor on the symmetrical distribution of weight over the legs were demonstrated. The study of Tyson [9] presumes that walking aids had more effect on the lateral sway of the trunk rather than fore-aft sway.

No clear consensus exists of effects of walking aids on symmetry of gait, walking speed and weight bearing. Furthermore, there is a distinct paucity of detailed studies on muscle activation patterns. The Neuro Developmental Treatment (NDT) concept [5] is one of the most widely used approaches in stroke rehabilitation within Europe [10] and aims to improve recovery of the hemiplegic side by focussing on normalizing muscle function and symmetry [5]. However, in only one study were muscle activation patterns recorded [10]. Results showed that walking with a cane did not differ from walking without an aid with regards to the timing of muscle activation patterns and mean amplitude of activity. Muscle activation patterns were qualitatively rated and mean amplitude was only computed and statistically tested in selected intervals. Based on the study of Hof and van den Berg [11], who described that the amplitude of the electromyography (EMG) of muscles depends on the muscle force produced, one would expect to record lower amplitudes in EMG of anti gravity muscles when walking with a walking aid.

The objective of this study was to investigate changes in muscle activation timing and amplitude that occur when walking with and without a walking aid, using an objective automatic burst on and off detector [12].

2. Methods

2.1. Subjects

Stroke patients recruited from the Rehabilitation Centre 'Het Roessingh' in Enschede, The Netherlands were included in the study if they were aged between 40 and 75 years, had a first unilateral ischemic stroke and were able to walk without physical assistance (Functional Ambulation Categories (FAC) ≥3 [13]). They were required to walk approximately100 m with and without a walking aid for the measurements. Furthermore, sufficient cognitive abilities (Mini-mental State Examination (MMSE) ≥22) [14]) were required to participate. Patients were excluded if they have had more than one stroke or had other physical conditions adversely influencing walking ability. The medical ethics committee of the Rehabilitation Centre approved the study. All subjects signed an informed consent before participating in the experiment.

2.2. Experimental set-up and procedures

Selected patients were tested in three different conditions in randomized order: (1) walking without a walking aid, (2) walking with a cane and (3) walking with a quad stick.

All patients wore their own preferred shoes. Five patients used an ankle foot orthosis (AFO), three wore a plastic AFO in their shoes and two wore a double bar brace attached to the shoe. The canes used were normal height walking sticks with a normal grip. Normal height was defined as 'at the level of the radial styloid of the sound wrist' with the arm straight hanging down [15].

2.3. Muscle activation patterns

Activation patterns of erector spinae, gluteus maximus, gluteus medius, vastus lateralis, semitendinosus, gastrocnemius and tibialis anterior muscles on the affected side were assessed using surface electromyography. SEMG of these seven muscles was recorded during walking using an 8-channel K-lab KL-100 EMG amplifier. 'Meditrace pellet #1801 graphics control' electrodes were used. Electrode size (1 cm^2), inter electrode distance (2 cm), electrode placement and

skin preparation were according to the 'SENIAM' protocol [16]. The SEMG signals were band pass filtered (third order Butterworth high-pass filter –3 dB at 20 Hz, first order low-pass filter –3 dB at 500 Hz) and amplified. The SEMG signals were digitised at 1000 Hz sample rate with 12 bits resolution and stored by a VICON 370 system. The SEMG processing and parameter extraction was performed using MATLAB. At least 10 gait cycles were recorded for each subject with each condition and stored for further analysis.
Footswitches were used to determine stance and swing phase of both legs. The footswitches consisted of an aluminium conductive sheet covering the sole of the shoe in conjunction with a conductive rubber mat [17, 18]. The SEMG and footswitch signals were collected within one session to make an accurate as possible within subject comparison. Walking speed was measured over a distance of 7.5 m using light gates to detect start and stop.

2.4. Data reduction

An objective method was implemented to automatically analyze the muscle activation patterns [12]. This method consisted of an automatic burst on and off detector of SEMG signal based on the approximated generalized likelihood ratio (AGLR) principle described by Staude and Wolf [19]. This algorithm was used to analyze the raw SEMG signals of every stride in the gait cycle separately with respect to the on- and off-times of muscle activation. Subsequently, all detected on- and off-times were normalized in time using the stride time starting from the related heel strike. SEMG was rectified and 25 Hz low-pass filtered and plotted together with the timing information along the x-axis (Fig. 1). Additionally the mean amplitude within the detected burst was calculated from this smoothed rectified EMG (SRE).

2.5. Statistical analysis

Data of the selected muscles were analyzed separately. Since the gathered data were not normally distributed the median on- and off-times in percentage of the gait cycle (together with the 25th and 75th percentiles) and the total burst duration (off-time minus on-time in percentage of gait cycle) were calculated for each subject. Differences in walking speed, median on- and off-times, burst duration and mean amplitude within the burst, between the interventions were analyzed statistically using the Friedman test in SPSS. The level of statistical significance was set at $p < 0.05$. Comparisons that showed significant differences were analysed post hoc using the Wilcoxon Signed Ranks test.

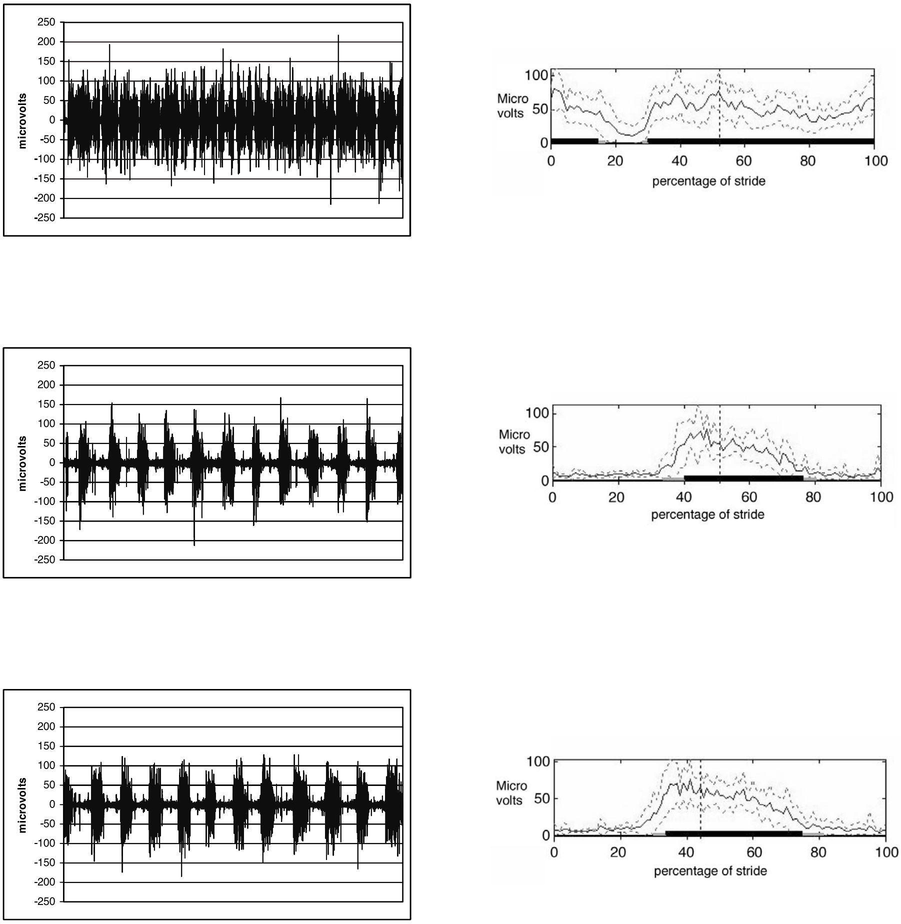

Fig. 1.
Both raw EMG and stride normalized smooth rectified EMG (SRE) of the erector spinae is presented for the three different walking conditions. (a) The SEMG of the erector spinae when walking without an aid, (b) when walking with a cane and (c) when walking with a quad stick. Along the y-axis the amplitude of both raw SEMG and SRE in microvolts is presented. Along the x-axis of the stride normalized SRE graphs, the timing as derived from the burst detection algorithm is shown in black and gray lines. The median on- and off-time is presented in a black solid bar connecting the median on-time with the median off-time. The somewhat smaller little grey bars indicate the 25th and 75th percentiles of the median on- and off-times. The dashed vertical line represents toe off.

3. Results

3.1. Study population

Ten men and three women (mean age 63 years, range 50–72 years) participated in the study. They had a mean time since an ischaemic stroke of 205 days (range 47–385 days). All selected patients used a cane for walking during normal every-day activities. Ten patients had a left hemiplegia and three a right hemiplegia. The median FAC-score was 4 (range 3–5). All patients had sufficient cognitive abilities to participate in the study. Median MMSE was 28 (range 22–29).

3.2. Walking speed

Differences in walking speed during the three interventions are presented in Table 1. Statistical analysis of walking speed using the Friedman test showed no significant differences (p = 0.193). Post hoc testing using the Wilcoxon Signed Ranks test, how-ever, revealed a statistical significantly lower walking speed (p = 0.045) when walking with a quad stick was compared to walking with a cane.

Walking speed	N	Mean	Std. Deviation	Percentiles		
		m/sec		25th	50th (Median)	75th
Without aid	13	.45	.19	.27	.42	.67
Cane	13	.44	.19	.23	.47	.61
Quad stick	13	.39	.14	.23	.40	.54

Table 1.
Descriptive statistics of walking speed during the three interventions

3.3. Muscle activation patterns

Fig. 1 shows a typical example of the muscle activation patterns of the erector spinae of one stroke patient when walking without a walking aid (Fig. 1a), with a cane (Fig. 1b) and when walking with a quad stick (Fig. 1c). Walking without an aid showed clear differences in timing compared to walking with a cane or a quad stick. When walking without an aid the erector spinae was almost continuously active throughout the gait cycle and a clear phasic activity was seen when walking with an aid. Differences between walking with a cane and walking with a quad stick were small and walking with a quad stick showed an earlier on-time compared to walking with a cane. The average amplitude did not differ much between the three different walking conditions.

3.4. Timing

Statistical analysis of the timing parameters measured during the three different walking conditions, using the Friedman test, showed significant differences for the burst duration of the erector spinae (p = 0.006), the on-time of the vastus lateralis (p = 0.009), the on-time of the tibialis anterior (p = 0.008) and the burst duration of the tibialis anterior (p = 0.023). Post hoc testing using the Wilcoxon Signed Ranks test (Table 2) showed a statistically significant decrease in burst duration of the erector spinae, when walking with a cane was compared to

Timing	Compare	Negative ranks	Positive ranks	Ties	Total	Z	Asymp. Sig. (2-tailed)
Erector spinae Burst duration	Cane – without aid	11	2	0	13	-2.062	.039 *
	Quad stick – without aid	11	2	0	13	-1.852	.064
	Quad stick - cane	4	9	0	13	-1.503	.133
Vastus Lateralis On-time	Cane – without aid	7	6	0	13	-.175	.861
	Quad stick – without aid	2	11	0	13	-.1.992	.046 *
	Quad stick - cane	2	11	0	13	-2.551	.011 *
Tibialis Anterior On-time	Cane – without aid	5	8	0	13	-.314	.753
	Quad stick – without aid	2	11	0	13	-1.712	.087
	Quad stick - cane	2	11	0	13	-2.691	.007 *
Tibialis Anterior Burst duration	Cane – without aid	11	2	0	13	-2.411	.016 *
	Quad stick – without aid	10	3	0	13	-2.062	.039 *
	Quad stick - cane	8	5	0	13	-7.34	.463

Table 2.
Results of the post hoc testing of the difference in timing parameters, using the Wilcoxon Signed Ranks test
* significant difference.

walking without an aid. Although not statistically significant, a similar tendency was observed when walking with a quad stick. The on-times of the vastus lateralis were significantly later during the stride when comparing walking with a quad stick and a cane and between walking with a quad stick and walking without an aid. The on-times of the tibialis anterior were significantly later during the stride when comparing walking with a quad stick and walking with a cane. Burst duration of the tibialis anterior decreased significantly while walking with a cane or with a quad stick compared to walking without an aid.

3.5. Amplitude

Statistical analysis, using the Friedman test, showed significant differences in the average amplitude of the burst of erector spinae ($p = 0.028$), gluteus maximus ($p = 0.004$), gluteus medius ($p = 0.004$), vastus lateralis ($p < 0.001$) and tibialis anterior ($p = 0.020$) measured during the three different walking conditions.

Post hoc testing using the Wilcoxon Signed Ranks test (Table 3) showed a statistically significant decrease in average amplitude of the erector spinae when comparing walking with a quad stick and walking without an aid, and between walking with a quad stick and walking with a cane. The decrease in the amplitude of gluteus maximus, gluteus medius and vastus lateralis was statistically significant when walking without an aid was compared to walking with a cane and walking with a quad stick. The decrease in amplitude of the tibialis anterior was statistically significant when walking with a quad stick was compared to walking without an aid and walking with a cane.

3.6. Clinical relevance

The 25th, 50th and 75th percentiles of the timing parameters were used to consider the clinical relevance of the described changes. Fig. 2 clearly shows the asymmetric distribution of the data. The 25th, 50th and 75th percentiles of the on-times of the vastus lateralis were close to normal [20, 21]. The differences in on-times of the vastus lateralis between the three walking conditions were only about 2.5% of the total stride time.

The on-times of the tibialis anterior show a major increase in the 25th, 50th and 75th percentiles when walking with a quad stick. Compared to normal this on-time was increasingly delayed and thus abnormal. Walking without an aid and walking with a cane showed on-times close to normal.

The 25th, 50th and 75th percentiles of the burst duration of the erector spinae and tibialis anterior drops and shifted towards more normal values when walking with a cane or quad stick.

The 25th, 50th and 75th percentiles of the amplitudes (Fig. 3) of the different muscles during three different walking conditions showed only small differences in the erector spinae, gluteus maximus and gluteus medius muscles. Larger changes were found in vastus lateralis and tibialis anterior.

Amplitude	Compare	Negative ranks	Positive ranks	Ties	Total	Z	Asymp. Sig. (2-tailed)
Erector Spinae	Cane – without aid	8	4	0	12	-1.412	.158
	Quad stick – without aid	10	2	0	12	-2.432	.015 *
	Quad stick - cane	10	3	0	13	-2.201	.028 *
Gluteus Maximus	Cane – without aid	10	2	0	12	-2.040	.041 *
	Quad stick – without aid	11	1	0	12	-2.275	.023 *
	Quad stick - cane	9	4	0	13	-1.433	.152
Gluteus Medius	Cane – without aid	10	2	0	12	-2.275	.023 *
	Quad stick – without aid	11	1	0	12	-2.981	.003 *
	Quad stick - cane	8	5	0	13	-1.153	.249
Vastus Lateralis	Cane – without aid	11	1	0	12	-2.981	.003 *
	Quad stick – without aid	12	0	0	12	-3.059	.002 *
	Quad stick - cane	9	4	0	13	-1.363	.173
Tibialis Anterior	Cane – without aid	8	4	0	12	-1.726	.084
	Quad stick – without aid	10	1	0	11	-2.756	.006 *
	Quad stick - cane	9	3	0	12	-2.275	.023 *

Table 3.
Results of the post hoc testing of the difference in amplitude using the Wilcoxon Signed Ranks test
* significant difference.

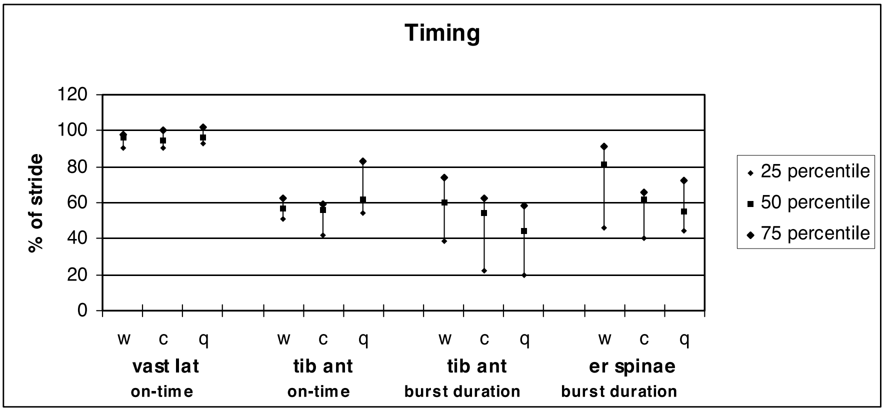

Fig. 2.
The 25th, 50th and 75th percentiles of the difference in timing parameters. w: without aid; c: cane; q: quad stick. er spinae: erector spinae; glut med: gluteus medius; glut max: gluteus maximus; vast lat: vastus lateralis; tib ant: tibialis anterior.

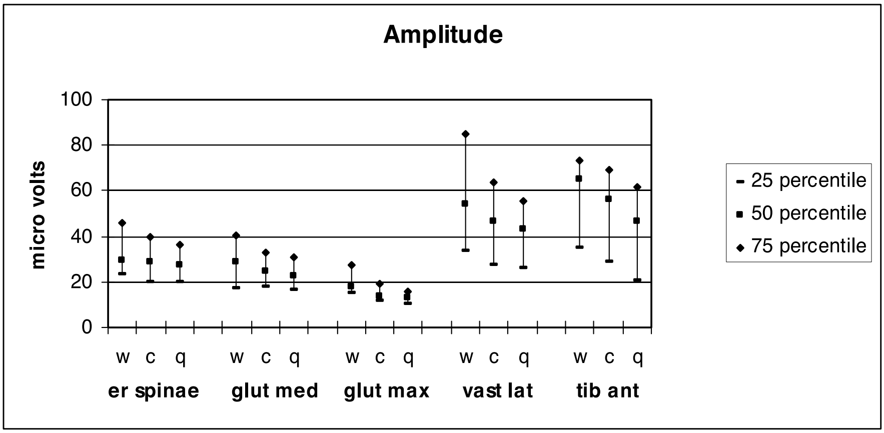

Fig. 3.
The 25th, 50th and 75th percentiles of the difference in amplitude within the burst. w: without aid; c: cane; q: quad stick. er spinae: erector spinae; glut med: gluteus medius; glut max: gluteus maximus; vast lat: vastus lateralis; tib ant: tibialis anterior

4. Discussion

In the past the timing of muscular action was usually assessed by visual inspection of the raw EMG patterns or by studying the smooth rectified EMG profiles. In this study we used an objective motor onset detector based on the approximated generalized likelihood ratio principle developed by Staude and Wolf [19] and applied for burst detection in SEMG by Roetenberg et al. [12]. A considerable advantage of using such a burst detection algorithm is that it enables SEMG signals to be analysed objectively and automatically.

In contrast to the findings of Hesse et al. [10], we found significant changes in the muscle activation patterns and average amplitudes within the burst. Hesse et al. did not find any difference in walking with an aid when compared to walking without an aid. This difference in outcome might be because Hesse et al. calculated the differences in amplitude during pre-selected intervals, whereas in our study the total burst was taken into account. Differences in timing might be due to the method used. The automated burst detector might be more sensitive to change compared to visual observation of the SEMG signal.

Most important changes in timing were found in the burst duration of the tibialis anterior and erector spinae. Part of the decrease in burst duration of the tibialis anterior might have been due to the delayed on-time when walking with a quad stick. The reason for this might be related to an increased lateral sway when using a quad stick, enabling foot-clearance without using the tibialis anterior muscle. The decrease in burst duration of the erector spinae seems to be related to the support of a walking aid. The activity of the erector spinae decreased during the period that weight was taken on the walking aid.

The reduced amplitudes of the anti gravity muscles (erector spinae, gluteus maximus, gluteus medius, vastus lateralis) were statistically different in this experiment but were not reported by Hesse et al. [10]. The reduction is in agreement with Hof and van den Berg [11] who stated that the amplitude of the EMG of muscles depends on the muscle force produced. When walking with a cane less muscle force is needed, because weight is taken on the cane, and thus less EMG activity is generated.

The median amplitudes (Fig. 3) of the different muscles during the three different walking conditions show that the most important changes were found in vastus lateralis and tibialis anterior. Changes in gluteus maximus, gluteus medius and erector spinae are less clear. The significant difference in amplitude in the tibialis anterior was unexpected. Walking without a walking aid induces higher amplitudes compared to walking with a cane or quad stick. Similarly, Hwang et al. [22] suggested that rail support during treadmill walking in hemiparetic patients reduced EMG linear envelope variability of the tibialis anterior. In stroke patients higher amplitudes of the tibialis together with an elongated burst duration is likely to cause a varus of the foot during stance and/or swing.

Hesse et al. [10] showed that the use of an AFO decreased the amplitude of the SEMG of the tibialis anterior during walking. In our experiment five out of 13 patients used an AFO. Comparisons between the three patients walking with a plastic AFO to those walking without an AFO did not reveal significant differences. The two patients wearing a double bar brace did differ from the other patients. Both burst duration and amplitude of the tibialis anterior increased when walking with a walking aid.

The abnormal activation pattern of the tibialis anterior was seen alongside an abnormal activation pattern of the erector spinae when walking without a walking aid. This observation might explain the rationale behind NDT therapists believing that trunk stability is an important prerequisite for normal arm and leg movement. Overexertion may provoke abnormal tone and stereotypical mass patterns of the affected side [23].

Changes in muscle activation patterns were only measured in the affected leg and comparisons were made to normal because Shiavi et al. [24] showed that both the affected and unaffected leg in stroke patients can demonstrate abnormal muscle activation patterns.

An influence of walking speed on SEMG amplitude in the experiments cannot be excluded. Statistical analysis of walking speed revealed a significant difference when comparing walking with a quad stick with a cane. Table 3 shows that differences in amplitude of the erector spinae and tibialis anterior were found when comparing walking with a quad stick to walking with a cane. No significant differences in amplitude of gluteus maximus, gluteus medius and vastus lateralis were found when comparing walking with a cane and walking with a quad stick. However, differences in amplitude were statistically significant when walking without an aid was compared to walking with a cane and walking with a quad stick (Table 3). These findings underline that the differences in amplitude of these muscles are related to the support of the aid. Although the reported changes were statistically significant there is still the question whether these changes are beneficial and thus clinically relevant. Since most therapies such as NDT [5] focus on relearning normal movements, changes in timing are thought to be clinically relevant when they become more normal as defined according to Perry [20] and Shiavi et al. [21]. Fig. 2 shows that the burst duration of the erector spinae and tibialis anterior dropped and shifted towards more normal values when walking with a cane or quad stick. Therefore these statistically significant changes can be considered clinically relevant. The 25th, 50th and 75th percentiles of the on-times of the vastus lateralis were close to normal. Differences in the 25th, 50th and 75th percentiles of the on-time of the vastus lateralis between the three walking conditions were only about 2.5% of the total stride time. Changes in the on-times of the vastus lateralis were so small that they would not be considered clinically relevant [25]. Walking with a quad stick induced abnormal on-times of the tibialis anterior whereas walking without an aid and walking with a cane show on-times close to normal.

5. Conclusion

The use of a cane resulted in less muscular effort, particularly of the vastus lateralis and was associated with a normalisation of muscle activation timing of the erector spinae and tibialis anterior. The use of a cane should be considered in the rehabilitation of stroke patients when therapy aims at normalisation of muscle activation patterns.

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CHAPTER 5

Influence of hamstring lengthening on muscle activation timing

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The purpose of this study was to describe the changes in muscle activation patterns using surface electromyography (sEMG) during walking in patients with cerebral palsy (CP), before and after hamstring lengthening. In the current clinical use of sEMG during walking in CP for pre-operative planning, various authors have observed that timing of muscle activation patterns hardly changes after surgical intervention. This observation is based on tendon transfer studies and visual interpretation of raw EMG signals. Little is known about the effect of muscle lengthening on muscle activation patterns of the lengthened muscles and their antagonists. Fifteen children with CP comprising a total of 23 hamstring lengthenings were included in this study. Surface EMG of semitendinosus and vastus lateralis was measured before and after surgery. Timing parameters of the sEMG patterns were quantified, using an objective burst detection algorithm and statistically evaluated. Results showed that hamstring lengthening causes statistically significant differences in timing of both the semitendinosus and vastus lateralis. It is concluded that timing parameters of operated muscles and their antagonists after surgery do change. The delayed off-time of the semitendinosus and the decreased burst duration of the antagonist (the vastus lateralis) after surgery were the most important changes.

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1. Introduction

Cerebral palsied-children (CP-children) often are confronted with a deterioration of their gait because of hypertonic muscles, deregulated muscle activation patterns, muscle contractures and disturbed growth and bony deformities [1]. Soft tissue surgery addresses muscle function and is a powerful method to correct structural changes. When an intensive rehabilitation program follows surgery, it may stop the deterioration of gait and even improve the gait pattern and consequently the functional level of walking [2, 3]. Thorough clinical examination, visual observation of the gait pattern and quantitative movement analysis are generally used to support clinical decision making for the selection of tendons or muscles and surgical procedures to improve walking [4, 5]. Interventions on muscles require good understanding of them in movement. A serious limitation in understanding muscle function is the fact that the underlying complex muscle activation patterns cannot be observed by the naked eye alone, nor can these uniquely be concluded from kinematics and kinetics of the joints they activate. Detailed information about muscle activation patterns and more insight inthe hindrances that muscles might cause during walking can be derived from electromyographic data. At present the muscle activation patterns during gait can be measured using surface electromyography (sEMG) [6] and although the added value of sEMG in determining interventions is not often appreciated, sEMG has been shown to be a reliable tool in the analysis of pathological gait [6-9]. Various authors have observed that the timing of muscle activation patterns during walking in CP children hardly changes through surgical intervention [10-14]. This knowledge is used in some existing clinical pre-operative protocols to set the proper indication for surgery [10, 12, 13, 15].

Significant limitations in the value of these findings in literature are twofold. Firstly, subjective visual interpretation of raw EMG signals was used. Secondly, conclusions of most of the currently known studies are limited to tendon transfers. Little is known about the effect of muscle lengthening on muscle activation timing of operated muscles and their antagonists [11]. The main goal of the present study therefore was to investigate the influence of hamstring lengthening on muscle activation timing using sEMG during walking in CP, quantified by using an automatic and objective burst detection algorithm.

2. Methods

2.1. Subjects

The study that was approved by the medical ethics committee board of the Roessingh Rehabilitation Center, Enschede (The Netherlands) included 15 subjects (9 males and 6 females) comprising a total of 23 hamstring lengthenings (Table 1). The mean age at date of surgery was 14.5 years (S.D. 5.9). Patient data were recorded at two different locations: location A was the Orthopedic Department of the University Hospital Münster, Münster (Germany) and location B was the Department of Rehabilitation Medicine of the VU University Medical Center in Amsterdam (The Netherlands). Data were analyzed in Enschede (The Netherlands).

Loc	Code	M/F	Age	Ham	Ach	Pso	Add	Rec	Other
A	4	M	17.9	*					
	5	M	18.6	*	*				*
	6	M	14.9	*	*		B		
	7	F	12.9	B			B		
	10	M	12.9	*	B		B		
	11	M	9.4	B	B		B		
	12	F	15.4	B	B				
B	59	M	4.5	B		B	B	B	
	34	F	16.2	B	B	B		B	*
	29	F	19.1	*		*	*	*	*
	19	F	1.1	B		B			*
	02	M	28.1	B		B		B	*
	26	M	7.6	*					*
	99	F	8.6	B					*
	06	M	20	*	B	*	B		*

Table 1.
Performed surgery
Type of surgery performed per centre per subject (B: both sides, *: one side). Loc: location; M/F: male/female; Age: years at surgery; Ham: hamstrings; Ach: Achilles; Pso: psoas; Add: adductor; Rec: rectus femoris.

The indications for surgery that always included hamstring lengthening, was done by clinical examination, observational movement analysis (video) and sEMG measurements during walking. Additional inclusion criteria were: age between 6 and 35 years and ability to walk independently on bare feet 20 m, the distance required for data collection. Subjects with a history of other diseases that could interfere with gait were excluded.

2.2. Experimental set-up and procedures

Semitendinosus and vastus lateralis were evaluated for change in muscle activation patterns after hamstring lengthening. sEMG of selected muscles was measured just before and 6 months after surgery. During the pre- and post-surgery, measurements subjects were instructed to walk at their own natural speed.

The protocols used for electrode placement and skin preparation in the two different movement laboratories were based on the SENIAM protocol [16]. sEMG was recorded using bipolar surface electrodes (size: 10 mm–10 mm). Inter-electrode distance was set at 20 mm (center to center). Pairs of electrodes were placed longitudinally on the selected muscles and their antagonists. sEMG was bandpass filtered to remove residual movement or cable artefacts (high pass filter: –3 dB at 20 Hz, low pass filter: –3 dB at 1500 Hz). Subsequently sEMG signals were full-wave rectified and low-pass filtered at 2 Hz (second-order Butterworth) and stored for further analysis. The protocols for recording initial foot contact were different in the two movement laboratories. At location "A" initial foot contact was recorded using two hand switches, which were pressed by the investigator during the sEMG measurement. Although this method has not been validated yet, it appeared to be quite successful in daily clinical routine. In this way over 20 gait cycles were identified and stored for further analysis. In location "B" initial foot contact was identified offline, through frame-by-frame analysis of the simultaneously recorded video, using a time code synchronization [17]. In this way at least three gait cycles were identified and stored for further analysis.

2.3. Data reduction

An objective method was implemented to automatically analyze the muscle activation patterns [18]. This method consists of an automatic burst on and off detector of sEMG signal based on the approximated generalized likelihood ratio (AGLR) principle developed by Staude and Wolf [19]. This algorithm was used to analyze the sEMG signals of every stride in the gait cycle separately with respect to the on- and off-times of muscle activation. Subsequently, all detected on- and off-times are normalized in time using the stride time starting from the related heel strike.

2.4. Statistical analysis

Data of semitendinosus and vastus lateralis muscles before and after surgery were analyzed separately. For each subject the median on- and off-times in percentage of the gait cycle (together with the 25 and 75 percentile) and the total burst duration (off-time minus on-time in percentage of gait cycle) of both muscles were calculated. Distribution of the data proved to be normal. Consequently differences, in the median on- and off-times and the total burst duration, between pre- and post-surgery measurements were analyzed statistically usinga paired t-test in SPSS. The level of statistical significance was set at a value of *P*<0.05.

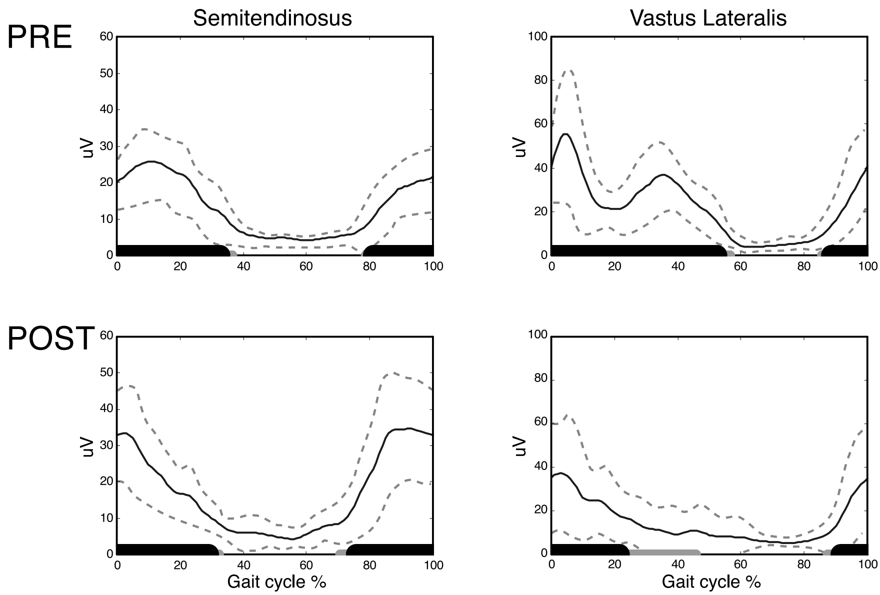

Fig. 1.
Examples of SRE profiles and standard deviations together with the timing parameters of pre-measurement (upper graph) and post-measurement (lower graph) of semitendinosus (left) and vastus lateralis (right). Along the Y-axis the amplitude in uV is presented. Along the X-axis, the timing is illustrated in black and gray lines. The start of the gray line is the 25 percentile of the on-times. The start and end of the black lines denote the median on- and off-times, respectively.
The end of the gray line is the 75 percentile of the off-times.

3. Results

Fig. 1 is a typical example of a pre- and post-surgery measurement of one subject showing a clear decrease in burst duration of the vastus lateralis after surgery. Along the X-axis, the timing derived from the burst detection algorithm is shown in black and gray lines.

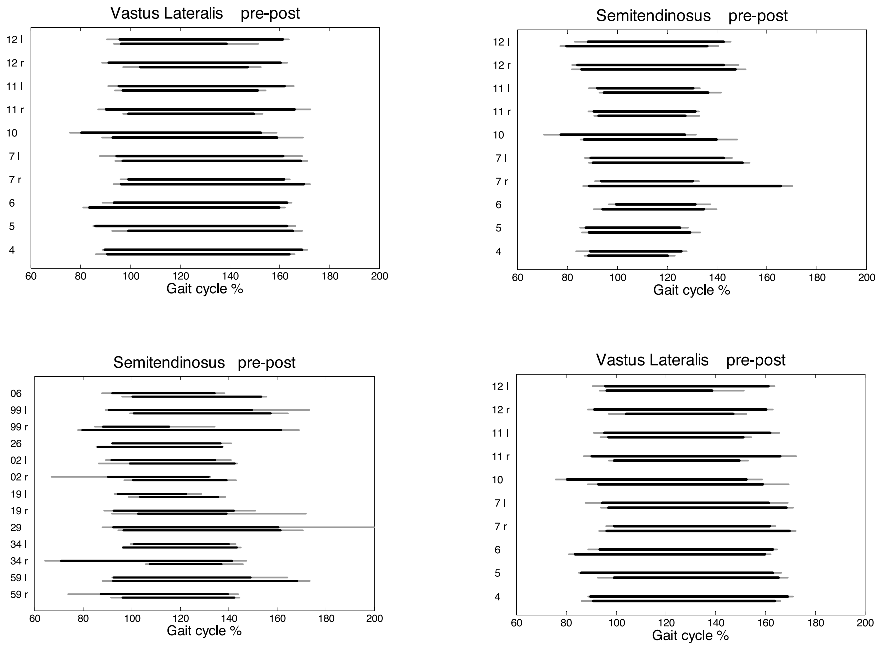

Fig. 2.
(a) Timing parameters of hamstring tendon lengthening before and after surgery in location A. One hundred percent gait cycle represents heel strike. Results of semitendinosus are presented in the left graph. The right graph shows the results of the vastus lateralis. For all subjects, the timing before and after surgery is plotted next to one other. The upper bar is related to the pre-measurement, the lower bar to the post-measurement. l: left leg, r: right leg. (b) Timing parameters of hamstring tendon lengthening before and after surgery in location B. One hundred percent gait cycle represents heel strike. Results of semitendinosus are presented in the left graph. The right graph shows the results of the vastus lateralis. For all subjects, the timing before and after surgery is plotted next to one other. The upper bar is related to the pre-measurement, the lower bar to the post-measurement.
l: left leg, r: right leg.

Results of pre- and post-surgery measurements per subject are presented in Fig. 2. Since both semitendinosus and vastus lateralis are active around heel strike, timing parameters of these muscles are presented in a different layout [8]. In these graphs 100% gait cycle represents heel strike. This way of presenting on-

and off-times and burst duration enables one to easily assess the differences in the activation patterns before and after surgery of the individual subjects.
In general the recordings of the vastus lateralis and semitendinosus in both locations showed the same differences between pre- and post-surgery measurements. The only differences between the two locations are the larger gray bars of the data measured in location B, representing the 25 and 75 percentile of the on- and off-times, respectively, and the later start of the activation patterns in location B, as related to initial contact. There was only one clear exception. The recording of the right leg of subject 7 shows a noticeable big increase in off-time after surgery. This difference, however, coincided with the clinical observation that the child was walking very insecurely after surgery and needed a lot of support. The functional outcome after surgery was judged to be very poor. Table 2 summarizes the results of the statistical evaluation of the 23 datasets.

	Paired differences					t	df	Sig.(2-tailed)
				95% Confidence Interval of the difference				
	Mean	Std. Deviation	Std. Error Mean	lower	upper			
semitendinosus								
Pre_off - Post_off	-0.088	0.123	0.025	-0.141	-0.034	-3.404	22	.003
Pre_on - Post_on	-0.043	0.097	0.020	-0.085	-0.00086	-2.116	22	.046
Burst Pre – Burst Post	-0.045	0.184	0.038	-0.124	0.035	-1.164	22	2.57
vastus lateralis								
Pre_off - Post_off	0.062	0.132	0.027	0.004	0.119	2.254	22	.035
Pre_on - Post_on	-0.029	0.060	0.012	-0.055	-0.002	-2.275	22	.033
Burst Pre – Burst Post	0.091	0.127	0.026	0.036	0.146	3.424	22	.002

Table 2.
Means, standard deviations, standard errors and the 95% confidence intervals of the differences together with the P-values of the "on and off" times and burst duration pre- and post-surgery of the semitendinosus and vastus lateralis

The statistical results showed significant differences for both the on- and off-times of the semitendinosus before and after surgery. Activation of the semitendinosus started later in the gait cycle and lasted longer. Burst duration therefore increased as well, although the mean change of 4.5% was not statistically significant.
The activation of the vastus lateralis showed statistically significant differences for both on- and off-times as well as burst duration. Activation of the vastus lateralis started later in the gait cycle and stopped earlier. The mean decrease in burst duration was 9.13% of the gait cycle.

4. Discussion

Up to now, the timing of muscular action has usually been assessed by visual inspection of the raw EMG patterns or by studying the smooth rectified EMG (SRE) profiles. Timing parameters are then derived from the profiles by applying a threshold to detect onset and offset of burst activity. However, there is no consensus in literature about the selection of parameters used for this purpose [20]. Moreover, these methods introduce large systematic errors [9, 19, 21] and by averaging timing parameters over a number of steps, information about step-to-step coordination, consistency and variability is lost.

In this study we have implemented an objective motor onset detector based on the AGLR principle developed by Staude and Wolf [19]. This automatic burst detection algorithm was shown to be both a reliable and objective tool for detecting on- and off-times in EMG signals [18, 19, 22]. A considerable advantage of using such a burst detection algorithm is that it enables one to analyze sEMG signals in an objective and automated way after which the changes in muscle activation patterns before and after surgery can be analyzed statistically. Although the burst detector was originally designed to be used on raw sEMG signals it appeared to be applicable on 2 Hz low pass filtered sEMG as well. As far as we know this is the first time that an automated burst detector has been used to detect changes in muscle activation timing before and after surgery.

The results measured in the different locations showed two differences (Fig. 2). Firstly, the larger gray bars of the data measured in location B represent the 25 and 75 percentile of the on- and off-times, respectively, and secondly, the later start of the activation patterns in location B, as related to initial contact. These increased deviations may be because the number of steps measured in location B was much lower than in location A. The later start of the activation patterns in location B may be explained by the way in which initial contact was detected. The different methods used for the detection of initial contact may introduce some error in the exact timing. These errors, in comparison to the differences found, were thought to be relatively small and therefore of little influence to the final results, especially because in each subject the differences between pre- and post-surgery were determined in the same laboratory with the same staff. In general the recordings of both locations show the same changes in post- versus pre-surgery measurements. These changes are in contrast with Gueth et al.'s [11], who stated that 'the muscular activity reacts as a kind of fingerprint', meaning that there was no difference between pre- and post-operative EMG. Our results

showed a change in on-times (4.3% later in the gait cycle) and off-times (8.8% later in the gait cycle) of muscle activation patterns in the lengthened semitendinosus after surgery. Fifteen out of the 23 datasets showed later on-times after surgery and nineteen datasets showed later off-times. Although these changes were statistically significant there is still the question as to their clinical relevance In the literature no unambiguous criteria for clinical relevance have been described. The only reference, which might be related to clinical relevance, is from Perry [8] who described a period of 5% of the gait cycle as "meaningful muscle action". This indicates that the change in off-times is clearly clinically relevant whereas the statistically significant change in on-times just comes close to clinical relevance. This later off-time might be caused by an increased anterior pelvic tilt due to the hamstring lengthening [23]. In this case the anterior pelvic tilt causes the semitendinosus, which also acts as a hip extensor [24], to be active longer throughout the gait cycle to extend the hip.
As compared to the semitendinosus the changes seen in the vastus lateralis (antagonist) were clearer, and all statistically significant. Burst duration of the vastus lateralis became 9.1% shorter in the gait cycle because of the on-times, starting 2.9% later and off-times, starting 6.2% earlier. Eighteen out of 23 datasets showed a shorter burst duration after surgery.
Using the earlier cited criterion of "meaningful muscle action", earlier off-time and decreased burst duration were clearly clinical relevant. The earlier off-time and decreased burst duration indicated a shift to more normal activation patterns [25]. A similar decrease in burst duration of the antagonist has also been reported by Hesse et al. [26], who described a mean decrease in burst duration of 17.6% in the rectus femoris after injecting the hamstrings with botulinum toxin A. Perry in 1976 [27] reported changes in muscle activity of non-operated muscles. This decrease in burst duration was very likely due to the increased knee extension during stance phase as reported by many authors [3, 26, 28, 29, 30].
The inter-individual differences in detected timing parameters after surgery of the subjects in both centers were considerable, especially in the semitendinosus. The additional interventions such as Achilles tendon and psoas lengthening that were performed in many subjects were likely to have caused additional changes in the individual biomechanics and consequently to the inter individual variability of the muscle activation patterns.
Since walking speed was not included in the routine clinical measurement no unambiguous judgement about the influence of walking speed on the detected differences in timing can be made. Subjective observations of the pre- and

post-measurements in most of the cases did not reveal large differences in walking speed. In healthy subjects Hof et al. [31] described the amplitude of the averaged EMG patterns differs but timing of EMG activity remained invariant at different walking speeds. In pathological gait, the influence of different walking speeds has not yet been investigated systematically.
Current clinical practice in most centres is based on the assumption that EMG activity does not change following soft tissue surgery. This study provides evidence that this assumption is not safe.

5. Conclusion

The main goal of the project was to investigate the influence of hamstring lengthening on muscle activation timing using sEMG during walking in CP. The underlying motivation for this research question is that knowledge of changes in muscle coordination patterns after surgery can be used for surgical indications of hamstring lengthening in CP-children.
Our results showed that there was a clear change in off-time of the semitendinosus muscle which was significantly later in the gait cycle after surgery and judged to be clinically relevant.
Differences in the activation of the antagonist, vastus lateralis, were more pronounced. The burst duration after surgery was significantly shorter due to the significantly later on-time and earlier off-time.
We concluded that timing parameters of semitendinosus and vastus lateralis after hamstring lengthening do change in CP-children. These changes are likely to be related to the biomechanical changes after hamstring lengthening and the response of the vastus lateralis.

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CHAPTER 6

The effect of an ankle-foot orthosis on walking ability in chronic stroke patients

DCM de Wit, JH Buurke, JMM Nijlant, MJ IJzerman and HJ Hermens

Regaining walking ability is a major goal during the rehabilitation of stroke patients. To support this process an ankle-foot orthosis (AFO) is often prescribed. The aim of this study is to investigate the effect of an AFO on walking ability in chronic stroke patients.
Twenty chronic stroke patients, wearing an AFO for at least six months, were included. Walking ability was operationalized as comfortable walking speed, scores on the timed up and go (TUG) test and stairs test. Patients were measured with and without their AFO, the sequence of which was randomized. Additionally, subjective impressions of self-confidence and difficulty of the tasks were scored. Clinically relevant differences based on literature were defined for walking speed (20 cm/s), the TUG test (10 s). Gathered data were statistically analysed using a paired t-test. The mean difference in favour of the AFO in walking speed was 4.8 cm/s (95% CI 0.85–8.7), in the TUG test 3.6 s (95% CI 2.4–4.8) and in the stairs test 8.6 s (95% CI 3.1–14.1). Sixty-five per cent of the patients experienced less difficulty and 70% of the patients felt more self-confident while wearing the AFO. The effect of an AFO on walking ability is statistically significant, but compared with the a priori defined differences it is too small to be clinically relevant. The effect on self-confidence suggests that other factors might play an important role in the motivation to use an AFO.

Clinical Rehabilitation 2004; 18: 550-557

1. introduction

Walking ability after stroke is often disturbed because of muscle weakness, spasticity, impaired sensorimotor control or loss of cognitive functions. Yet regaining the ability to walk is a major goal during the rehabilitation of stroke patients [1]. Often ankle-foot orthoses (AFO) are used to support this goal. An AFO is generally prescribed to provide mediolateral stability at the ankle in stance phase, facilitating toe clearance in swing phase and promoting heel strike [2]. In practice in the Netherlands often different types of plastic, nonarticulated, off the shelf AFOs and articulated metal double-bar braces are used.
Literature reveals only one randomized controlled trial concerning the effect of an AFO on walking in stroke patients [3]. In this study no beneficial effects of an AFO on walking ability were found, neither statistically significant nor clinically relevant. Other studies, however, suggest improvement in walking speed [4-11], gait pattern [4,10,12] and stride length [6-9].
Although AFOs are thought to have beneficial effects on functional walking ability, results reported in the literature are scarce and inconsistent [2,3,11]. Only one study assessed the effect of an AFO on functional outcome [11], suggesting an improvement of functional walking ability using the Functional Ambulation Categories [13]. In asystematic review Leung and Moseley [2] reported a dominance of positive studies suggesting improvement of walking speed. The significance of these changes on daily functioning (clinical relevance) and implications for the wider population however remain unresolved.
The literature is not conclusive and especially the effect of an AFO on daily practice remains unknown. The aim of this study, therefore, was to investigate the effect size and clinical relevance of an AFO on walking ability in chronic stroke patients already wearing an AFO.

2. Methods

2.1. Subjects

Patients were recruited from the Rehabilitation Centre ‘The Roessingh’ in Enschede and from the Department of Rehabilitation of the ‘Twenteborg’ hospital in Almelo, the Netherlands.
Stroke patients were eligible for the study if they were between 40 and 75 years old and had a first unilateral ischaemic or haemorrhagic stroke from the middle

cerebral artery. Patients had to be at least six months post stroke and had to wear a plastic, nonarticulated ankle-foot orthosis daily for at least six months. Patients had to be able to walk independently with shoes with and without orthosis (walking aids were permitted). Furthermore sufficient communication (measured with Utrechts Communication Examination, UCO) [14], cognitive abilities (measured with Mini-mental State Examination, MMSE) [15] and a satisfactory condition were required for full participation in the study.
Patients were excluded if they had more than one stroke, had other diseases negatively influencing walking ability or were unable to walk independently.
Ethical approval was obtained from the medical ethics committee board of 'The Roessingh' Rehabilitation Centre and the 'Twenteborg' hospital. All subjects signed an informed consent before participating.

2.2. Ankle-foot orthoses

Selected patients used plastic, nonarticulated ankle-foot orthoses. Due to small individual differences between the patients the AFOs were of three different types: (1) an AFO with a small posterior steel (Distrac or Dynafo; Ortho Medico, Herzele, Belgium); (2) an AFO with a big posterior steel, sometimes individually made (Camp; Basco Healthcare, Zaandam, the Netherlands), and (3) an AFO with two crossed posterior steels and an open heel (Ottobock; Ottobock, Son en Breugel, the Netherlands). The main function of these AFOs is to facilitate foot clearance during swing phase and improve initial contact.

2.3. Outcome measures

The Motricity Index of the affected leg [16] and the Functional Ambulation Category (FAC score) [13] were used to characterize the population.

Walking ability was operationalized using:

1) Comfortable walking speed
2) Timed up and go test
3) Stairs test.

Patients were measured walking with and without their AFO while wearing shoes. The sequence of walking with or without the AFO first was randomized.

1) Comfortable walking speed was measured in the gait laboratory on a 10 m walkway. Walking speed was automatically measured with two infra-red beams over a distance of 7.5 m. Patients walked the 10-m walkway three times. The average walking speed was calculated and used for further analysis.
2) Timed up and go test (TUG). In this test the patient is timed while he or she rises from a chair, walks 3 m, turns, walks back and sits down again. Patients were permitted to use a walking aid, but no physical help. The same standard chair with seat height 47 cm and arm height 67 cm was used for all the patients. Patients were allowed to practise the test once before being timed. The TUG is a reliable and valid test for quantifying functional walking ability in elderly people with different kind of medical history, including strokes [17]. In stroke patients the TUG might be expected to be highly responsive as long as the gait speed is not too close to normal values [18]. The TUG was measured three times. The average time needed was calculated and used for further analysis.
3) Stairs test , being an extended version of the TUG, including the same tasks from the TUG, with addition of two tasks: stair ascent and descent [18]. Due to practical circumstances the test was slightly modified. Patients were timed while they rose from a chair, walked 1.18 m, ascended a flight of 12 stairs, walked 1.64 m, touched the wall, turned around, descended the stairs, walked back to the chair and sat down. They were allowed to use a walking aid and the rail of the stairs, if necessary. The same chair was used as in the TUG and the stairs used had 12 steps 25 cm deep and 16.5 cm high (the rise percentage is 66%).

   The stairs were 1 m wide, with a rail on both sides. The instructions before the test and the procedure were the same as in the TUG. Richards [18] found that this test can detect improvements in patients with a moderate to normal walking speed. The stairs test was measured three times. The average time needed was calculated and used for further analysis.

Additionally, after the TUG and the stairs test, with and without the AFO, patients were asked to express their subjective impressions considering self-confidence and difficulty of the tests using two five-point rating scales:

- **Self-confidence:** 1 – very insecure, 2 – insecure, 3 – neither insecure, nor secure, – secure and 5 – very secure.
- **Difficulty:** 1 – very difficult, 2 – difficult, 3 – neither difficult, nor easy, 4 – easy and 5 – very easy.

2.4. Data analyses

Walking speed, TUG and stairs test were statistically analysed using a paired t -test in SPSS. Subjective impressions of the patients concerning self-confidence and difficulty were analysed by means of the Wilcoxon signed ranks test in SPSS. The main interest of this study concerned the significance of the observed changes for daily functioning (clinical relevance). Hence clinically relevant effect sizes were defined prior to the study.

1) Walking speed: Based on the study of Perry et al. [19] a difference of 20 cm/s in walking speed with and without the AFO was defined as clinically relevant. Perry assigned the walking speed to six categories of functional walking ability, the modified Hoffer Functional Ambulation Scale. Taking the inclusion criteria into account (able to walk independently), the last three categories of this scale (most-limited, least-limited community and community walkers) were judged as the most important. Between these categories a mean difference in walking speed of approximately 20 cm/s was observed.

2) Timed up and go test: For this test a difference of 10 seconds was defined as clinically relevant. This was based on the study of Podsiadlo and Richardson [17]. Podsiadlo described significant differences in Barthel scores (and independency) with a TUG score of less than 20, 20–30 and more than 30 seconds.

3) Stairs test: No suggestion for clinically relevant differences could be extracted from the literature. Based on the defined clinical relevant differences of the TUG and taking into account the stairs test takes more time to perform, a clinical relevant difference of 15 seconds was defined by the authors.

4) Subjective impressions of the patients concerning self-confidence and difficulty: Authors defined any improvement of one point on these scales as clinically relevant.

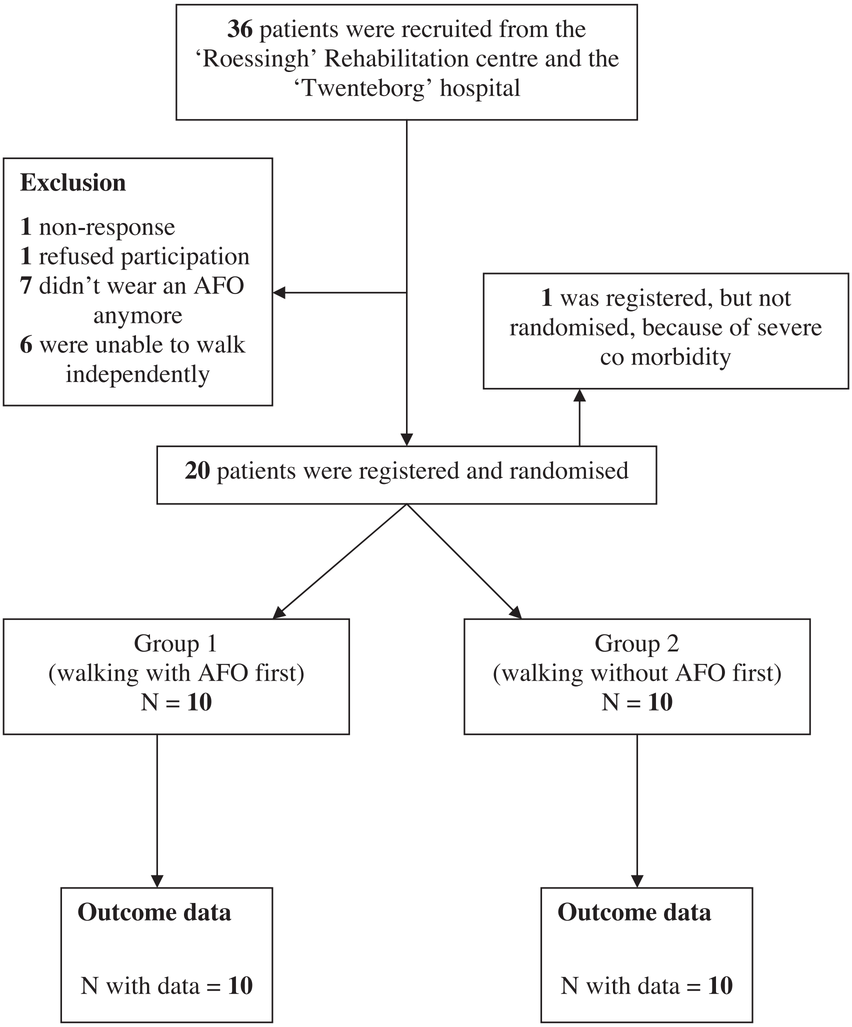

Figure 1.
Flowchart of patients through the study.

3. RESULTS

3.1. Study population

Between February 2002 and November 2002, 20 stroke patients were included in the study. Figure 1 shows a flowchart describing the flow of the patients throughout the study.

Twelve men and eight women participated in the study with a mean age of 61.2 years old (range 41-73 years) and a mean time since stroke of 25.6 months (range 8–48 months). Mean time since start of using an AFO was 20.9 months (range 6-44 months). Eighteen patients had an ischaemic stroke and two a haemorrhagic stroke; 11 patients had a left hemiplegia and nine a right hemiplegia. The median Motricity Index for the affected leg was 58.0 (interquartile range (IQR) 27.0) and the median FAC score was 4.5 (IQR 1.0). All patients had sufficient communicative (median UCO 6.0 and IQR 0.8) and cognitive abilities (median MMSE 26.0 and IQR 2.8) to participate in the study.

Table 1 Baseline characteristics in the two randomization groups

	Group 1 (AFO-without AFO)	Group 2 (Without AFO-AFO)
Age years (range)	61.1 (51 - 73)	61.2 (41 - 70)
Time since stroke, months (range)	26.9 (8 - 42)	24.2 (8 - 48)
Time wearing AFO, months (range)	20.8 (6 - 39)	20.9 (6 - 44)
Ischaemic/heamorrhagic strokes	9/1	9/1
Left/right hemisphere	7/3	4/6
Median UCO (IQR)	6 (0.3)	6 (1.3)
Median MMSE (IQR)	25.5 (3.8)	26.5 (3.5)
Median Motricity Index, affected leg (IQR)	59.5 (26)	53.0 (28.3)
Median FAC score (IQR)	4.5 (1)	4.5 (1)

IQR, interquartile range; UCO, Utrechts Communication Examination.

Table 1 shows the baseline characteristics for the two randomized groups; group 1 walked with the AFO first and group 2 without the AFO first. The two groups were comparable for baseline characteristics. Consequently the results of the two groups were pooled.

Table 2 Descriptive statistics and results of paired *t*-test (*p*-values)

	Velocity (cm/s) with AFO	Velocity (cm/s) without AFO	Velocity (cm/s) difference	Timed Up and Go (s) with AFO	Timed Up and Go (s) without AFO	Timed Up and Go (s) difference	Stairs (s) with AFO	Stairs (s) without AFO	Stairs (s) difference
Mean (SD)	49.6 (24.3)	44.9 (24.0)	4.8 (8.4)	25.6 (11.7)	29.2 (12.9)	3.6 (2.5)	73.0 (37.8)	81.6 (44.4)	8.6 (11.8)
Median	44.2	34.3	4.2	23.3	29.3	3.2	68.6	74.5	3.4
Minimum	18.0	18.0	-17.0	10.3	10.5	0.2	27.4	27.8	-3.3
Maximum	93.0	93.0	24.0	49.7	56.7	8.1	157.9	181.8	34.1
Interquartile range	37.7	43.9	0.1	19.6	23.8	4.9	59.2	68.1	15.9
p-value *t*-test			0.020			0.000			0.004

3.2. Effect on walking ability

Table 2 shows the descriptive statistics and the results of the paired t -test (p-values) for all tests. The mean difference in favour of the AFO for walking speed was 4.8 cm/s (95% CI 0.85-8.7), for the TUG test 3.6 s (95% CI 2.4-4.8) and for the stairs test 8.6 s (95% CI 3.1-14.1). Results show a statistically significant effect in favour of the AFO in walking speed, TUG and stair test. However, when taken into account the a priori defined values none of these effects can be described as clinically relevant (Figure 2).

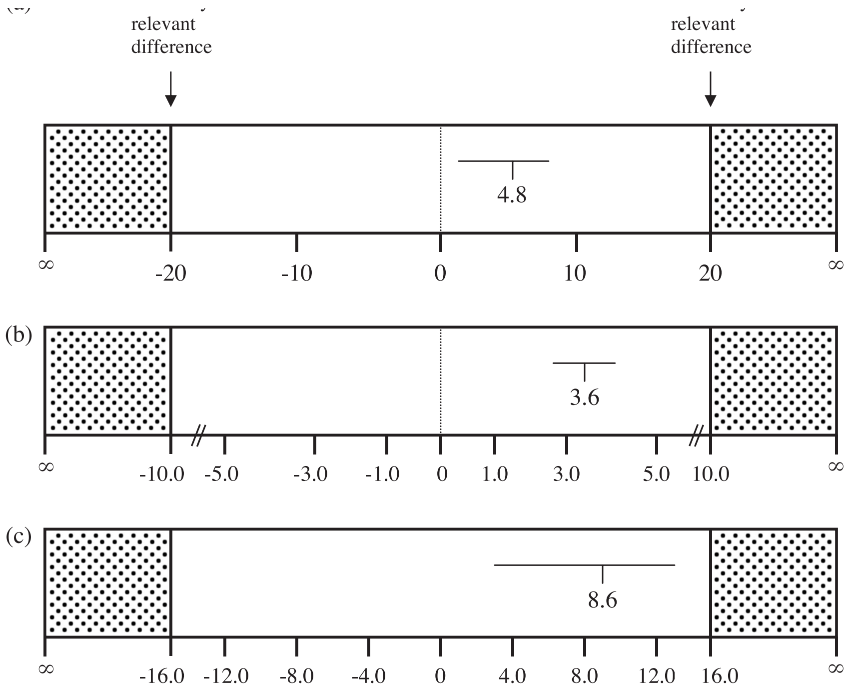

Figure 2.
Effect AFO on walking ability. (a) Walking speed (cm/s), mean difference and 95% CI. (b) Timed up and go test (s), mean difference and 95% CI. (c) Stairs test (s), mean difference and 95% CI.

3.3. Effect on subjective impressions

Of all patients 65% (13/20) experienced less difficulties (p-value=0.001) and 70% (14/20) felt more self-confident (p-value=0.005) while wearing the AFO ≥ 1 point change on the scales). In both difficulty and self-confidence 25% (5/20) of the patients scored no effect of the AFO. In contrast to the tests for walking ability these differences were judged as clinically relevant.

DISCUSSION

The aim of this study was to investigate the clinically relevant effect of an AFO on walking ability in chronic stroke patients already wearing an AFO. Results show that the effect of an AFO on walking ability and subjective impressions are statistically significant. Compared to the a priori defined values however, these differences are too small to be considered clinically relevant, which is in agreement with earlier findings of Beckerman et al [3].

The subjective impressions obtained with the questions about difficulty and self-confidence are clearly in favour of the AFO and judged as clinically relevant. We acknowledge the procedure to ask for self-confidence and difficulty with a five-point scale has no well established validity and reliability, but was considered an important outcome that had to be highlighted in this study. Most patients felt more self-confident and experienced less difficulty during the measurements when wearing an AFO. This is comparable to the findings in the study of Tyson [11].

In the present study 14 out of 20 patients mentioned the beneficial effects of the AFO. Five patients mentioned no effect performing the tasks with or without the AFO. One patient mentioned even a worse performance of the tasks with the AFO. In trying to find an explanation for this difference, baseline characteristics of the patients were evaluated. All but one patient experiencing no effect of the AFO had a Motricity Index score for the affected leg $\geq$ 60 points. Most of the patients mentioning beneficial effects of the AFO had a Motricity Index for the leg $\leq$ 60 points. The only patient mentioning worse effects of the AFO also had a Motricity Index score for the leg $\geq$ 60 points. It appears that patients with a better function of their leg experience less beneficial effects of an AFO compared with the more affected patients.

Although the number of patients included in the study was relatively small, the effects were statistically significant. Including more patients theoretically will only change the width of the confidence interval, but would not change the overall conclusion. The relatively small number of patients included is inherent to the aim of the study (clinical relevance of the nonarticulated AFO for daily practice). Homogeneity of the study population was considered to be very important. The population therefore was highly selected, which of course may limit the generalization of the results.

The mean difference in walking speed in favour of the AFO is 4.8 cm/s (95% CI 0.85-8.7). This finding is comparable with findings from literature; Leung and

Moseley [2] reported on four studies comparing walking with AFO versus walking with shoes (as in this study). The mean differences in walking speed in these four studies varied between 0 and 12.0 cm/s with a moderate degree of variability in 95% confidence interval.
The clinically relevant differences for walking speed and TUG test were based on the studies of Perry et al [19] and Podsiadlo and Richardson [17] respectively. When using the computed differences in walking speed for each individual patient, walking with and without an AFO, only four subjects changed to a higher category in Perry's classification. Based on the outcomes of the TUG only five subjects changed to a higher score on the Barthel Index. These results are in contrast with the findings of Tyson and Thornton [11]. They concluded that most of the patients improved in functional walking ability using the FAC (median FAC without AFO is 2 and with AFO is 4). Differences in outcome may be due to differences in population; Tyson included acute stroke patients whereas in this study only chronic stroke patients, at least six months post stroke and walking with an AFO for at least six months, were included.
The a priori defined, clinically relevant differences chosen for the TUG were perhaps arbitrary and too high. In the studies of Geiger et al [20] and Ahmed et al [21] smaller differences in TUG score were found, respectively after physical therapy (mean difference 8.46 seconds) and during recovery after stroke (mean difference 9 seconds after five weeks and 0 seconds from five weeks to three months). On the other hand the a priori chosen differences were related to functional scores, the modified Hoffer Functional Ambulation Scale and the Barthel Index. Perhaps one has to accept that the effect of an AFO is too small to be clinically relevant. Lehmann [22] already stated that 'one must recognize that these orthoses are worn by many patients who can walk without them but who cannot walk safely'. If this is true, safety should be the prime motivation for prescribing an AFO rather then walking ability.

In conclusion, for patients using an AFO for everyday activities post stroke, the AFO is beneficial for their walking ability. Differences in walking speed, TUG test and stairs test are statistically significant in favour of the AFO. Hence most patients feel more self-confident and experience less difficulty performing the measurements while wearing an AFO. Clinical relevance for daily functioning, however, remains unclear.

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CHAPTER 7

A feasibility study of remote consultation to determine suitability for surgery in stroke rehabilitation

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We studied knowledge transfer for the determination of the suitability of stroke patients for a specialist surgical procedure (split anterior tibial tendon transfer). Gait analysis data from patients at a general hospital were discussed with an expert in another country using personal computers, an ISDN connection (128 kbit/s) and TCP/IP-based communication tools. The key issue was whether the staff in the general hospital became better able to determine suitability for surgery. Twelve patients were studied. In three of the first four cases the advice of the remote expert changed the plan for surgery. After that the treatment plans did notchange after consultation. After eight cases the local clinicians did not need to ask for further advice. There was a rapid increase in skill in determining suitability for surgery. The experience and skills of the local clinicians were thought to increase more rapidly than would have been the case without the consultations with a remote expert.

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1. Introduction

It is difficult to describe human movement unambiguously in a letter, fax or email message. This represents a fundamental difficulty in communication about patients with movement disorders. Quantitative recordings of human movement patterns [1, 2] provide a better and less ambiguous description. In the mid-1990s it was realized that such recordings could be used to overcome distance, via telecommunications [3, 4]. This offers the possibility of applying telemedicine techniques to rehabilitation of motor disability. Experiments have shown that recordings of human movement are sufficiently unambiguous for a remote expert to form a clinical opinion [5, 6]. Discussions about movement disorders on the Internet have also proved feasible [7].
Although not mentioned in assessment studies of telemedicine [8–11], learning and teaching effects of remote consultation have been reported [12–18]. Indeed, Curry et al [19] concluded that every project which included teleconsultation contained an educational component. These findings suggest that new treatment strategies and good practice in health-care may be disseminated more readily with the appropriate use of remote consultation with experts. The traditional cognitive apprenticeship model (CAM) for medical education [20–22]is applicable to remote expert consultation [23]. This model suggests that there are three phases to the process of learning:

(1) the learner acquires new skills from the expert;
(2) the learner's experience and skills increase under supervision of the expert;
(3) the collaborative learning process ends when the learner has become independent from the expert.

This approach promises better initial treatment results, less risk for patients, more confidence and a steeper learning curve for the local clinician, and consequently a much faster way to implement new treatment strategies in the daily routine of patient care.

1.1. Tendon transfer

We studied the use of teleconsultation to determine the suitability of stroke patients for split anterior tibial tendon transfer. The objective of this surgical technique is to centralize the pull of the tibialis anterior muscle and thereby improve dorsiflexion during swing. The operation corrects a varus deformity of the foot and circumvents the need for traditionally prescribed orthoses or orthopaedic shoes. Despite being described in the literature [24, 25] and used in other countries, this surgical procedure has not been adopted in the rehabilitation of stroke patients in The Netherlands. As a result there was a lack of experience in performing the operation and in determining the suitability of patients for surgery, as well as in their postoperative management.

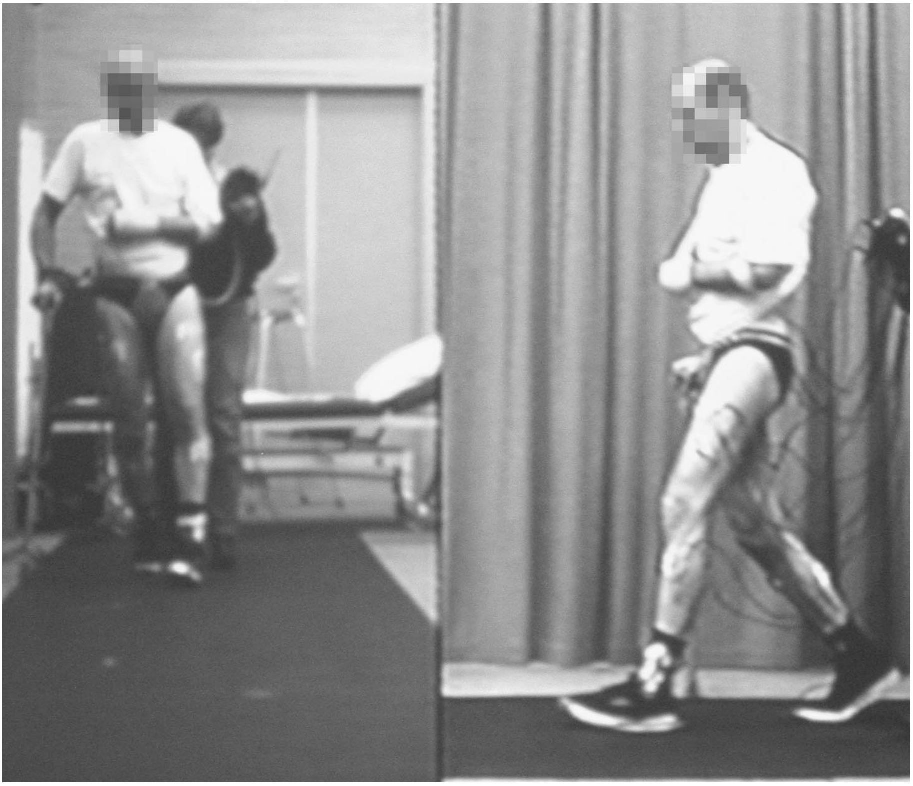

Fig 1
Split-screen video-recording (frontal and sagittal plane) during walking.

2. Methods

Ethics permission was granted by the appropriate committee and informed consent was obtained from the patients. The following inclusion criteria were imposed: the patient had had a stroke at least six months earlier; the stroke had

resulted in hemiplegia; there was insufficient dorsiflexion or varus during swing of the hemi-paretic foot, which caused an abnormal initial contact of the lateral side of the foot. Subjects were excluded if they had any medical condition that prevented surgery.
Suitability for surgery was determined using gait analysis. A split-screen video-recording (Fig 1) was used to register the swing phase varus of the foot. Surface electromyography (EMG) of the lower leg muscles was used to detect whether or not the observed swing phase varus of the foot was caused by the unopposed pull of the anterior tibial muscle. Clinical examination was used to score the muscle force of the anterior tibial muscle (see Appendix) and to make sure that no other major impairments were present around the ankle, knee or hip.
Data from the clinical examination and gait analysis were stored on a server. Stored data were accessible for the different users at any time for asynchronous analysis using personal computers, an ISDN connection (128 kbit/s) and TCP/IP-based communication tools. The video-recordings were captured at 25 frames/s and a spatial resolution of 720 x 540 pixels. The average file size of the video-recordings was 8 MByte and their average duration was 8 s. In order to reduce the transfer time, the video-recordings were compressed using the MPEG1 algorithm, which resulted in an average file size of 1 MByte.
The key issue was the local staff's ability to determine patients' suitability for surgery, initially under supervision of the remote expert. In addition, the communication tools used and the quality of communication were evaluated. These items were verified with the local clinicians and patients in an interview. No questionnaire was used but specific questions were raised in a face-to-face conversation. Patients were asked:

(1) Would you have been willing to be operated on without the advice of the expert?
(2) Are you satisfied with the outcome of the surgery?

The local clinicians were asked:

(1) Would you have taken the same decision to operate without the advice of the remote expert?
(2) Are you satisfied with the results?
(3) After how many patients did you feel secure in determining suitability for surgery?

(4) Does the written information (e.g. notes on the clinical examination, questions asked and discussion) serve your needs in terms of increasing your experience in determining suitability for surgery?
(5) What about the quality of the visual information and surface EMG signals?

The first phase of the CAM, involving the learning of new skills from the (remote) expert, started with a two-day training session at the Orthopaedic University Clinic in Heidelberg (Germany) for two rehabilitation specialists, an orthopaedic surgeon, a physical therapist and a biomedical engineer from Enschede in The Netherlands. The training comprised learning of the surgical technique and the protocol for determining the suitability of patients. Special attention was given to the protocol and procedures for communication.
After the training, the rehabilitation specialists in Enschede selected patients who might benefit from split anterior tibial tendon transfer. They were then sent to the movement laboratory in Enschede for movement analysis in order to complete the data-set needed for determination of their suitability for the operation. This data-set consisted of a clinical history, the results of the clinical examination, and a splits-screen video-recording of the gait pattern in the sagittal and frontal plane, together with surface EMG of the tibialis anterior and gastrocnemius.
After obtaining the patient's consent the multimedia data-set was made available for the remote expert in Heidelberg, using TCP/IP-based communication tools. In order to meet privacy regulations a password protected server was used. Email was used to notify the expert of the addition to the server of a new case for consideration.

3. Results

Twelve patients (five men, seven women) participated in the experiment. Depending on the advice given and the agreement reached between the remote expert and the local clinicians, patients were either operated on or prescribed conventional therapy and/or an orthosis. To illustrate the procedure and communication, a typical case is presented in detail in the Appendix.
The results concerning an increase in experience and skills (phase 2 of the CAM) and attaining independence (phase 3) are presented in Table 1. In three of the

first four cases the advice of the remote expert changed the plan for surgery. After that treatment plans did not change after the consultation. After the first eight cases the local clinicians no longer felt they needed to ask for advice.

3.1 Responses to the questions

One of the patients was very clear in her statement. She was willing to undergo the surgery only on the advice from the remote expert and with his presence during surgery. The other patients were less explicit. The first three patients to undergo the surgery (those for whom there was sufficient follow-up) were satisfied. They said that the result matched their expectations. All were capable of walking barefoot and experienced considerable benefit during their everyday activities.

	Improving experience and skills (phase 2)			Independence (phase 3)
Subject	Suitable for SPLATT? y/n	Advice requested? y/n	Advice changed about suitability for surgery? y/n	Tele-consultation necessary? y/n
1	y	y	y	y
2	y	y	n	y
3	y	y	y	y
4	y	y	y	y
5	n	y	n	y
6	y	y	n	y
7	n	y	n	y
8	n	y	n	y
9	y	n	-	n
10	n	n	-	n
11	n	n	-	n
12	y	n	-	n

Table 1
Summary of the learning process for the local practitioners in determining the suitability for surgery of the first 12 subjects

The local team was satisfied with the performance of the patients after surgery. They felt that the possibility of discussing the findings with an expert quickly increased their ability to determine suitability for surgery. However, complete independence was not reached and the expertise of the remote expert was still needed, especially in the more difficult cases.

The visual quality of the digitized video-recording of the patient was judged to be poor or just sufficient, probably owing to the MPEG compression. In two cases the expert asked for a higher-quality video-recording before he could give his opinion on the movement pattern. Information from the clinical examination and the surface EMG was judged as good.

4. Discussion

Even though the experiment took place in a laboratory setting with experienced staff, skill was felt to have grown more rapidly than would have been the case without the possibility of consulting a remote expert, although this could not be measured objectively. Having the possibility of asking questions and discussing findings with a (remote) expert is likely to increase learners' skills more rapidly compared with the situation where learners have to solve things on their own, making the same mistakes that have been made before by others. Teleconsultation may therefore become a strategic tool in cost control in health-care.

After patient data were made available on the private server, communication was mainly via email and sometimes by ordinary telephone. Videoconferencing was not used at all. The main advantage of email is that it is offline communication, so that it does not interfere with daily clinical routine. The written information and the email exchanges that followed were found to be useful in improving skill at determining suitability for surgery. Although objective recordings of human movement have been used successfully in teleconsultation before [5, 6], this is probably the first time that the transfer of knowledge has been the principal object of study. The results showed a rapid increase in knowledge. This knowledge was thought to have increased more rapidly than would have been the case if consultations with a remote expert had not been possible.

5. Appendix. Case study

ML was a 60-year-old married woman with no relevant medical history. She had suffered an ischaemic stroke in the right thalamus area, which caused a mild paresis of her left leg. A few years after the stroke she complained of increased difficulties during walking. Clinical observation revealed a swing phase varus that caused problems with foot clearance and an abnormal initial contact.

5.1. Movement analysis

Observational gait analysis using split-screen video-recording revealed only slight deviations from normal in the knee and foot when ML was walking barefoot. During the swing phase the left fore- and hind-foot were in varus, the first toe was extended and the lateral toes were clawed. Close-up views of the ankle and foot during swing phase clearly showed the tendon of the tibialis posterior just below the medial malleolus, indicating tibialis posterior activity.

Clinical examination identified a slightly reduced dorsiflexion of the left ankle (10° with the knee flexed at 90° and 5° with the knee extended) and a reduced range of motion in valgus of the ankle (0°). There was no increase in tone or clonus of the triceps surae or any other muscle. Muscle strength of left hip extensors, left hip flexors, left knee extensors, left plantar flexors and left dorsiflexors was scored as 4 on the Medical Research Council (MRC) scale [26]. This scale ranges from 0 (paralysis) to 5 (normal strength). Invertors of the left ankle were scored as MRC 5 and evertors as MRC 3. The strength of all other muscles of the left and right leg was scored as MRC 5.

A surface EMG recording revealed a normal timing of the gastrocnemius during walking. The muscle activation timing of the tibialis anterior was prolonged [27]. Unfortunately, it was impossible to measure the activity of the tibialis posterior with surface electrodes. The close-up view from the video-recording of the foot during walking revealed reversed tibialis posterior activity, which indicated that the tibialis posterior was active during swing phase and absent during stance.

5.2. Discussion with the remote expert

After analysing the data in Enschede it was concluded that the tibialis anterior caused the swing phase varus. However, the local team was in doubt about the contribution of the tibialis posterior and the strength of the tibialis anterior. The doubts and questions were formulated as follows: would split anterior tibial tendon transfer alone be sufficient to solve the problem of swing phase varus or should a tibialis posterior transfer procedure be considered as well, in order to overcome the possible contribution of this muscle to the varus and to solve the problem of possible tibialis anterior weakness?

The case was made available to the remote expert in Germany, after which the patient data and conclusions were discussed. According to the remote expert, ML was an ideal case for split anterior tibial tendon transfer. Because of the extension of the first toe, dorsiflexor weakness could be overcome by transferring the extensor digitorum longus to the dorsum of the foot, thus contributing to dorsiflexion during swing and leaving the first toe in a neutral position. The local team was only partly satisfied with this suggestion, because there was no answer regarding the possible involvement of the tibialis posterior. Therefore the remote expert was asked again to give his opinion on this matter. In his answer he agreed with the observation. The tendon was clearly visible and likely to contribute to the varus. However, since the varus occurred only in swing phase, there was no real need to perform a split tibialis posterior transfer. In conclusion it was decided that split anterior tibial tendon transfer together with transfer of the extensor hallucis longus to the dorsum of the foot would be sufficient to solve the problem.

5.3. Intervention

Surgery was planned and carried out by the local orthopaedic surgeon in collaboration with the expert. Postoperative treatment consisted of two weeks of non weight-bearing plaster cast followed by three weeks of weight-bearing plaster cast. After removing the plaster the subject had to use an ankle foot orthosis during walking and a night splint for half a year. Physical therapy was prescribed for six weeks, once a week, in order to strengthen the weakened dorsiflexors and to guide the restoration of the walking pattern.

5.4. Postoperative status

The first evaluation took place about six months after surgery. It revealed a nicely balanced foot during swing phase resulting in adequate foot clearance and an improved initial contact. The dorsiflexor strength was sufficient (MRC 4). As expected, the range of motion of the ankle remained unchanged.
A second evaluation 18 months after surgery showed the same walking pattern. ML was able to walk without any difficulty barefoot indoors and outdoors in normal shoes. She felt very satisfied with the result.

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CHAPTER 8

General discussion

GENERAL DISCUSSION

As discussed in the introduction of this thesis, most therapies used in clinical practice assume that improvement of functional abilities is facilitated through the improvement in muscle co-ordination. Restoration or development of new co-ordination patterns is thought to be a prerequisite in the improvement of functional abilities. Kwakkel [1] stated that "the training of ADL implies that we need to know more about the nature of co-ordination deficits in functional tasks, and thus, about the natural laws of co-ordination and control in the performance of such tasks". However, until now most stroke studies have been focused on assessment of skills [2] and did not assess changes in co-ordination patterns.
The aim of this thesis therefore was to provide a better understanding of the development of the co-ordination patterns during walking, after stroke, their possible relation with functional performance and to what extend they can be manipulated.

SEMG processing

In this thesis timing parameters derived from the SEMG of 16 muscles were used to quantify muscle co-ordination. These timing parameters were shown to be considerably less variable then amplitude parameters in the analysis of dynamic SEMG [3]. In literature timing of muscle activation is usually assessed by means of visual inspection of the unprocessed SEMG patterns or by studying the smooth rectified EMG (SRE) profiles [4, 5, 6, 7]. A disadvantage of this method is that it omits the step to step variability of the timing of the muscle activation patterns; information that might be relevant as a measure of performance of motor control and balance.
Chapter 2 of this thesis describes the development and application of an objective muscle on and off-set detector based on the approximated generalized likelihood ratio (AGLR) principle developed by Staude and Wolf [8]. This automatic burst detection algorithm was shown to be a reliable tool for detecting on- and off-times in EMG signals [8, 9, 10]. A considerable advantage of using such a burst detection algorithm is that it enables the analysis of SEMG signals of single steps in an objective and automated way. This enables one to analyze statistically the changes in muscle activation patterns during recovery (chapter 3) and before and after different interventions (chapter 4 and 5).

Without such automated quantification of SEMG signals the processing of large amounts of data is very cumbersome and probably the reason that there is a very limited amount of literature on changes in muscle co-ordination during recovery of gait after stroke. Some of these concern the studies of Richards [4] who used SEMG to study the relationship between gait speed, clinical measures and muscle activation and Mulroy [5] who described differences in muscle activation patterns in early and late recovery phase. In addition, an SEMG study by Knutsson [6] resulted in the description of four different EMG patterns. Shiavi [7] recorded the SEMG patterns in the early (1 to 10 weeks post stroke) and late recovery period (6 to 24 months post stroke) and used the classification of Knutsson to describe changes in synergy patterns over time.

The studies of Richards, Mulroy, and Shiavi all reported changes in muscle synergies over time. These changes could not be confirmed in the study described in chapter 3 of this thesis. Possible causes for this discrepancy are twofold. Firstly, the study population: Neither Mulroy nor Shiavi presented a thorough description of the functional level of the included stroke patients. Consequently the study population might have been different from that of the studies presented in this thesis.

Secondly, the actual quantification of SEMG patterns: Richards studied the changes in amplitude of the SEMG, especially of the Triceps Surae and Tibialis Anterior in pre-selected intervals rather than using timing parameters. Mulroy used a simple threshold criterion and Knutsson and Shiavi studied the SEMG by visual inspection.

Although results presented in chapter 2 show that the AGLR provides more accurate timing parameters than visual inspection or a standard threshold criterion, quantification of SEMG needs further development. In the first place little is known about the relevance of a certain change in timing of muscle activation. The only reference on this topic is by Perry [11] describing a period of 5% of the gait cycle as "meaningful muscle action". Secondly, the timing parameters are used to describe activation of single muscles. By doing so, subtle changes in synergies might be missed. Quantification of SEMG therefore needs to be expanded. A more sophisticated method should not only quantify the timing of single muscles but should also be able to quantify synergies.

Measuring motor recovery

In many other studies involving motor recovery of both upper and lower extremities co-ordination is measured using the Fugl-Meyer Assessment (FMA) [12]. In clinical practice as well as in research the FMA is considered as one of the most comprehensive quantitative measures of motor impairment following stroke [13] and as such it is widely used to evaluate motor recovery. Identifying literature in PubMED with the keywords: "Stroke AND Fugl-Meyer" produces 147 citations. The FMA is designed specifically as a measure to enable assessment of recovery in the post stroke hemiplegic patient involving a set of different tests with good interrater and intrarater reliability [13]. However, the use of FMA scores to quantify changes in muscle co-ordination during walking is questionable. Firstly the test is a qualitative assessment in which the judgement of the performed movements by the examiner is very important. Secondly the test is done in different positions (lying, sitting and standing) and not during functional movements like walking. Thirdly starting point of the test is a stepwise sequence of the restoration of motor function as described by Twitchell [14]. The existence of such sequence however could not be confirmed in this thesis (chapter 3). Although functional abilities improved significantly, muscle co-ordination, measured with SEMG, did not change. This raises an interesting topic for discussion. Does the Fugl-Meyer assessment really measure motor recovery? Increasing muscle force during recovery, as measured by the Motricity Index [15] in chapter 3, might be a possible confounder. The increasing muscle force is likely to cause changes in moments around the different joints, thereby causing stronger or other movements as recovery proceeds.

Patient population

Selected patients in the different studies of this thesis were all recruited from the rehabilitation population being only a small part of the total population. Generally, in the Netherlands, 60- 65% of the stroke population returns to their homes directly after dismissal from the hospital, 25% is referred to nursing homes and only 10-15% of the total stroke population is referred to a rehabilitation centre[16]. Generalisation of reported results therefore is not obvious. The more affected stroke patients that need intensive therapy are referred to a rehabilitation centre. Therefore it remains questionable whether

reported results hold true for the total population or not. For instance in constraint induced movement therapy [17] the less affected patients show better results compared to severe affected patients.
In studies on stroke, it is often tried to involve a homogeneous group of patients. Patients often are classified in different categories describing the cause of the stroke. About 80 % of the strokes are caused by a first ever ischemic infarct [18]. Hence a lot of research is done on this part of the stroke population. However during the patient inclusion of the longitudinal study (chapter 3) it became clear that the stroke population of the rehabilitation centre differed from the total population, as ischemic infarcts were not that numerous. Other causes like haemorrhagic infarcts, multiple infarcts and combinations of left and right affected hemispheres were relatively more present as compared to the total stroke population.
In order to get a better view on the population treated in a rehabilitation centre one should consider paying more attention to other stroke categories which apparently are more present in a rehabilitation population.

Striving for normal motor behaviour

Normal motor behaviour often is used as a reference to describe abnormal motor behaviour, thus identifying abnormalities that need to be treated. In clinical practice treatment concepts also use normal motor behaviour as the goal to be reached. In a lot of less serious disorders like recovery from a fracture, muscle rupture or even mild stroke this makes sense. However in stroke victims, needing intensive rehabilitation, this way of thinking does not seem to be valid. Parts of the brain are damaged, and although the plasticity of the brain may assist in partially overcoming this, it is not likely that these stroke victims will fully recover. For these patients it seems invalid to strive for normal motor behaviour after rehabilitation. Instead goals should be directed at optimisation of existent or recovered motor behaviour. This requires two things. On one hand knowledge about the effect of treatment related to specific motor patterns and on the other hand quantitative assessment of motor patterns and functional abilities in individual patients. Since muscle co-ordination just as kinetics and 3D kinematics are hard to observe one should consider implementing movement analysis techniques in routine clinical practice in order to be able to provide optimal in individual patient care [19].

Results of the longitudinal study during recovery phase (chapter 3) show that although the functional abilities of the patients improve, muscle co-ordination patterns after stroke during walking do not change. Kinematics therefore are not likely to change either. However, this does not imply that muscle co-ordination patterns cannot be changed. Hesse [20] showed that facilitation techniques, as propagated by Davies [21], have a positive short term effect on muscle co-ordination. As soon as the facilitation (or support) given by the therapist stops, the effect on muscle co-ordination disappears. The use of walking aids (chapter 4) was shown to cause similar results. Walking with a walking aid induced a positive effect on muscle co-ordination which disappeared when walking without an aid. Soft tissue surgery (chapter 5) showed positive changes in muscle co-ordination which were permanent.
When rehabilitation goals are set to improve muscle co-ordination and thus kinematics, conventional therapy is not likely to be successful. Muscle co-ordination hardly changes. However, kinematics can be changed using this paradigm. In Cerebral Palsied (CP) children it has been rather well accepted that assessment of muscle co-ordination patterns is useful for the indication of soft tissue surgery [22, 23, 24]. Jacquelin Perry and Leo Doederlein among others used this assessment to perform soft tissue surgery in adult stroke patients. The underlying assumption is that the abnormal muscle co-ordination does not change due to such surgery and as a consequence outcome of tendon transfers can be predicted. Preliminary results of a study on the Split Anterior Tibial Tendon Transfer (chapter 7) show very good results. Varus of the foot diminishes and walking without orthosis is possible for most of the patients.

Future developments

Expanding these thoughts and combining them with what is accepted as general daily clinical routine in CP-children leads to new thoughts about the improvement of kinematics after stroke. Restoring the complex co-ordination of many muscles working closely together with conventional therapies seems not feasible after stroke. The recovery after stroke is characterized by fixed synergies causing a stereotype walking pattern without much variation. In the development of new treatment options, lost functions should be supported using rehabilitation aids or should be restored through for instance soft tissue surgery. Hyperactive muscles should be counteracted, preferably during the period in the gait cycle in which this hyperactivity is harmful (for instance: Gastrocnemius activity during early stance), but not during the period in which its activity is beneficial (for instance: Gastrocnemius activity during push-off). Single interventions though are not likely to be successful for this complex problem. Synergies will have to be treated as a whole. Therefore new treatment protocols have to be defined based on the fundamental knowledge of pathological gait after stroke and taking individual characteristics into account. In this way, in the future, treatment of motor disorders will consist of a combination of different interventions combining surgery with for instance botox injections, electrical stimulation, rehabilitation aids or other new treatment options. In the preceding diagnostic process extensive gait analysis will be indispensable.

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Summary

Stroke is one of the most frequently occurring disabling diseases in the western world. The incidence of stroke in The Netherlands is approximately 30,000 per year and the prevalence is 120,000 patients in a total population of 16.280.000. Regaining the ability to walk is an important goal for many stroke patients.
Although most of the treatment methods assume that improvement of functional abilities is achieved by improving muscle function (e.g. muscle co-ordination), only Richards (1992) investigated the assumed relation between change in muscle activation patterns and improvement of functional abilities.
The aim of this thesis was to provide quantitative data and a better understanding of the development of co-ordination patterns after stroke, their possible correlation with functional performance and to what extend these co-ordination patterns can be manipulated. In order to quantify muscle co-ordination surface electromyography (sEMG) was used.

The SEMG signal obtained during gait is often presented as the SEMG profile, the average SEMG activation pattern during one gait cycle. A disadvantage of this method is that it omits the step-to-step variability of the timing of the muscle activation patterns that might be relevant as a performance measure of motor control and balance.

In **chapter 2**, a method is described with which every step in the gait cycle could be analysed with respect to the timing of the muscle activation. The method describes an approximated generalised likelihood (AGLR) algorithm that was implemented and tested. Simulations show that the AGLR is much more accurate than a standard threshold criterion. Timing parameters are calculated from a SEMG recording during gait and a measure for symmetry and co-ordination is extracted. The amplitude distribution within and outside defined bursts is also presented to avoid the less precise classification into on and off patterns.
This method was used in studies described in chapters 3, 4 and 5. The studies were carried out in order to quantify changes in muscle co-ordination during recovery after stroke, and when walking is manipulated.

Chapter 3 is focused on the longitudinal description of changes in the neuromuscular co-ordination during the recovery of walking after stroke and its

relationship to functional recovery.
Thirteen patients diagnosed with a first unilateral ischemic stroke participated in the study. Functional recovery was measured using the Rivermead Mobility Index (RMI), Functional Ambulation Categories (FAC), Barthel Index (BI), Trunc Control Test (TCT), Motricity Index (MI) and comfortable walking speed. Surface Electro Myo Graphy (SEMG) of the erector spinae, gluteus maximus, gluteus medius, rectus femoris, vastus lateralis, semitendinosus, gastrocnemius and tibialis anterior of both legs were used to quantify co-ordination patterns. Assessment took place at 3, 6, 9, 12 and 24 weeks post stroke. Timing parameters of the surface EMG patterns were quantified, using the burst detection algorithm described in the chapter 2 and statistically evaluated together with the measurements of functional recovery.
All functional measures, except the TCT, improved over time and showed statistically significant differences. In contrast, SEMG patterns did not change over time.
The largely constant SEMG patterns over time suggest, that the functional improvement of gait is related to other mechanisms than to the restoration of co-ordination patterns in the affected leg.
Whereas most treatment methods assume that functional improvement is facilitated by the improvement of muscle co-ordination, this assumption could not be confirmed in this study.

Chapter 4 focuses on changes in muscle activation patterns that occur when subjects with stroke walk with and without a walking aid. This knowledge could help therapists in deciding whether or not patients should use a cane or quad stick while walking.
Thirteen patients suffering from a first unilateral ischemic stroke participated in the study. SEMG of the erector spinae, gluteus maximus, gluteus medius, vastus lateralis, semitendinosus, gastrocnemius and tibialis anterior of the affected side were measured during three different conditions: (1) walking without a walking aid, (2) walking with a cane and (3) walking with a quad stick. Timing and amplitude parameters of the activation patterns were quantified using the method described in chapter two.
Results showed a statistically significant and clinically relevant decrease in burst duration of both erector spinae and tibialis anterior when walking with a cane. The amplitude of the vastus lateralis and tibialis anterior decreased when

patients walked with a cane and quad stick. It was concluded that the use of a cane should be considered in the rehabilitation of stroke patients when therapy aims at normalisation of muscle activation patterns.

Chapter 5 describes the changes in muscle activation patterns during walking in patients with cerebral palsy (CP), before and after hamstring lengthening.
In rehabilitation motor behaviour of CP children often is thought to be comparable to that of adult stroke patients. Since surgical interventions are rather common in these patients, in contrast to the rehabilitation of stroke patients this group was used to evaluate whether a surgical intervention like hamstring lengthening changes muscle co-ordination patterns. It was assumed that when this is the case in CP children the same holds true for adult stroke patients.
Various authors have observed that timing of muscle activation patterns hardly changes after surgical intervention. This observation is based on visual interpretation of raw EMG signals in tendon transfer studies. Little is known about the effect of muscle lengthening on muscle activation patterns of the lengthened muscles and their antagonists. Fifteen children with CP comprising a total of 23 hamstring lengthenings were included in this study. SEMG of semitendinosus and vastus lateralis was measured before and after surgery. Timing parameters of the SEMG patterns were quantified, using the method described in chapter two. Results showed that hamstring lengthening caused statistically significant differences in timing of both the semitendinosus and vastus lateralis.

Chapter 6 described the effect of an AFO on walking ability in chronic stroke patients.
Twenty chronic stroke patients, wearing an AFO for at least six months, were included. Walking ability was operationalized as comfortable walking speed, scores on the timed up and go (TUG) test and stairs test. Patients were measured with and without their AFO using a cross-over design with randomization for the interventions. Additionally, subjective impressions of self-confidence and difficulty of the tasks were scored. Clinically relevant differences based on literature were a priori defined for walking speed (0.20 m/s) and TUG test (10 sec). Gathered data were statistically analysed using a paired t-test.

Results showed a mean difference in favour of the AFO in walking speed of 0.048 m/s, in the TUG test of 3.6 s and in the stairs test of 8.6 s. Sixty-five percent of the patients experienced less difficulty and 70% of the patients felt more self-confident while wearing the AFO.
It was concluded that the effect of an AFO on walking ability in this study was statistically significant, but compared with the a priori defined differences too small to be clinically relevant. The effect on self-confidence suggested that other factors might play an important role in the motivation to use an AFO.

Chapter 7 focuses on the implementation of a new treatment method. Rightfully the importance of dissemination and moreover implementation of results in clinical practice is emphasized in grant submissions and asked for by the potential users. This chapter addresses the knowledge transfer from an expert orthopaedic surgeon to the staff of a movement analysis laboratory in rehabilitation centre in order to determine the suitability of stroke patients for a specialist surgical procedure: the split anterior tibial tendon transfer. Gait analysis data from patients at the rehabilitation centre were discussed with an expert in another country using personal computers, an ISDN connection (128 kbit/s) and TCP/IP-based communication tools. The key issue was whether the staff in the general hospital became better able to determine suitability for surgery.
Twelve patients were studied. In three of the first four cases the advice of the remote expert changed the plan for surgery. After that the treatment plans did not change after consultation. After eight cases the local clinicians did not need to ask for further advice. There was a rapid increase in skill in determining suitability for surgery. The experience and skills of the local clinicians were thought to increase more rapidly than would have been the case without the consultations with a remote expert.

Samenvatting

Een beroerte of Cerebro Vasculair Accident (CVA) is één van de meest voorkomende invaliderende aandoeningen in de westerse wereld. Er wordt geschat dat er op dit moment in Nederland ca. 120.000 mensen met de gevolgen van een beroerte leven en elk jaar weer worden er in Nederland ca. 30.000 mensen voor de eerste keer getroffen door een beroerte. Het herwinnen van loopvaardigheid is voor veel patiënten een belangrijk doel tijdens de revalidatie. Alhoewel de meeste behandelmethoden er van uit gaan dat vooruitgang van dagelijkse vaardigheden wordt bereikt door een verbeterde spier coördinatie is dit tot nu toe alleen in beperkte mate door Richards (1992) onderzocht. Het grote belang van kennis over deze herstelprocessen voor de behandeling vormden de aanleiding voor het nationaal programma ‘herstel van lopen na CVA’. Als onderdeel van dit nationale programma richt dit proefschrift zich op (1) het herstel van de spier coördinatie na een beroerte, (2) de mogelijke relatie met dagelijkse vaardigheden en (3) in welke mate de spier coördinatie beïnvloed kan worden. Om spier coördinatie te kwantificeren is gebruik gemaakt van oppervlakte elektromyografie (EMG).

In **Hoofdstuk 2** wordt de analyse van het oppervlakte EMG besproken. Het EMG signaal gemeten tijdens het lopen wordt vaak gepresenteerd als een EMG profiel, het gemiddelde EMG signaal over één schrede. Een nadeel van deze methode is dat de variabiliteit tussen de stappen van het spieractivatie patroon niet weergeeft. Informatie die relevant kan zijn als maat voor coördinatie en balans. In dit hoofdstuk is een methode beschreven waarmee de timing van de spieractivatie patronen tijdens elke stap van het lopen geanalyseerd kan worden. Om dit mogelijk te maken werd het zogenaamde "approximated generalised likelihood Ratio" (AGLR) algoritme geïmplementeerd en getest. Simulaties laten zien dat dit algoritme veel nauwkeuriger is dan het gebruik van standaard drempelwaarden. "Timing" parameters, zijnde de momenten waarop de spier actief en inactief wordt, worden berekend uit de ruwe EMG signalen. Om de in pathologische situaties soms aanwezige continue spieraanspanning buiten de gedefinieerde perioden van activiteit beter te kunnen beschrijven, kan de gemiddelde amplitude van het EMG signaal zowel binnen als buiten de gedefinieerde periode van activiteit berekend en gepresenteerd worden.

Hoofdstuk 3 beschrijft de veranderingen in spiercoördinatie en de relatie van deze veranderingen met het functioneel herstel na een beroerte. Voor dit onderzoek werden dertien patiënten met een eerste unilateraal ischaemisch herseninfarct geselecteerd. Functioneel herstel werd gemeten met de Rivermead Mobility Index (RMI), Functional Ambulation Categories (FAC), Barthel Index (BI), Trunc Control Test (TCT), Motricity Index (MI) en comfortabele loopsnelheid. Spier coördinatiepatronen werden gekwantificeerd door het oppervlakte EMG van de erector spinae, gluteus maximus, gluteus medius, rectus femoris, vastus lateralis, semitendinosus, gastrocnemius en tibialis anterior van beide benen te meten. De onderzoeken werden 3, 6, 9, 12 en 24 weken na de beroerte uitgevoerd. Timing parameters uit het oppervlakte EMG werden gekwantificeerd met het AGLR algoritme, zoals beschreven in hoofdstuk 2, en samen met de testen betreffende het functioneel herstel statistisch geanalyseerd. Alle functionele maten, behalve de TCT, verbeterden in de loop van de tijd en lieten statistisch significante verschillen zien. De oppervlakte EMG patronen daarentegen veranderden niet. Dit suggereert dat het herstel van lopen gerelateerd is aan andere mechanismen dan het herstel van spiercoördinatie patronen.

Hoofdstuk 4 beschrijft de veranderingen in spiercoördinatie die zich voordoen als CVA patiënten met een loophulpmiddel lopen. Deze kennis kan behandelaars helpen bij de keuze of een CVA patiënt al of niet een wandelstok of vierpoot zou moeten gebruiken. Aan dit onderzoek deden dertien patiënten met een eerste unilateraal ischaemisch herseninfarct mee. Het oppervlakte EMG van de erector spinae, gluteus maximus, gluteus medius, vastus lateralis, semitendinosus, gastrocnemius en tibialis anterior van het aangedane been werd gemeten tijdens drie verschillende condities: (1) lopen zonder loophulpmiddel, (2) lopen met een wandelstok en (3) lopen met een vierpoot. Timing en amplitude van de spieractivatie patronen werden gekwantificeerd met de in hoofdstuk 2 beschreven methode. Resultaten laten een statistisch significante en klinisch relevante afname in activatie duur van de erector spinae en de tibialis anterior zien tijdens het lopen met een stok. Bovendien neemt de amplitude van met name de vastus lateralis en tibialis anterior af wanneer patiënten met een wandelstok of vierpoot lopen. De conclusie van dit onderzoek is dat, wanneer de therapie van CVA patiënten gericht is op normalisatie van spieractivatie patronen, het gebruik van een wandelstok overwogen zou moeten worden.

Hoofdstuk 5 beschrijft de veranderingen in spieractivatie patronen, tijdens het lopen van kinderen met een cerebrale parese (CP), voor en na een hamstrings verlenging. Binnen de revalidatie wordt de aansturing van de motoriek van CP kinderen vaak vergeleken met die van volwassen CVA patiënten. Aangezien chirurgische ingrepen, in tegenstelling tot CVA patiënten, relatief vaak voorkomen bij CP kinderen, werd een groep CP kinderen geselecteerd om te onderzoeken of een hamstring verlenging veranderingen in spiercoördinatie teweegbrengt. Aangenomen werd dat wanneer dit het geval was, dit ook voor CVA patiënten zou gelden. Verschillende auteurs hebben gerapporteerd dat, als gevolg van een chirurgische ingreep, de timing van de spieractivatie niet of nauwelijks verandert. Deze observaties zijn echter vooral gebaseerd op de visuele interpretatie van de EMG signalen. Drieëntwintig hamstring verlengingen bij in totaal vijftien CP kinderen werden in deze studie onderzocht. Het oppervlakte EMG van de semitendinosus en vastus lateralis werd gemeten vlak voor en 6 maanden na de chirurgische ingreep. Timing parameters van de spieractivatie patronen werden gekwantificeerd met de in hoofdstuk 2 beschreven methode. Analyse van de resultaten toont aan dat de chirurgische ingreep statistisch significante verschillen veroorzaakt in de timing van zowel de semitendinosus als de vastus lateralis.

In **Hoofdstuk 6** wordt aandacht besteed het effect van de enkel voet orthese (EVO) op de loopvaardigheid van CVA patiënten in de chronische fase. Twintig CVA patiënten, die tenminste 6 maanden een EVO gebruikten, werden voor deze studie geselecteerd. Loopvaardigheid werd vastgelegd door middel van de Timed Up & Go Test (TUG), de "stairs" test (trap op en af lopen) en de comfortabele loopsnelheid. Daarnaast werden subjectieve ervaringen betreffende de zelfverzekerdheid en de moeilijkheid van de verschillende taken met en zonder EVO gescoord. In dit onderzoek werd gebruik gemaakt van een cross-over design waarin de volgorde van de interventie (lopen met en zonder EVO) was gerandomiseerd. Op basis van literatuurgegevens werden a-priori klinisch relevante verschillen gedefinieerd voor de comfortabele loopsnelheid (0.20 m/sec) en de TUG test (10 sec). De verzamelde data werden statistisch geanalyseerd met behulp van een gepaarde t-toets. Dit resulteerde in een positief gemiddeld verschil ten faveure van de EVO voor de loopsnelheid (0.048 m/sec), de TUG (3.6 sec) en de "stairs"test (8.6 sec). Vijfenzestig procent van de patiënten ondervonden minder moeilijkheden en 70% van de patiënten voelden zich zelfverzekerder wanneer ze tijdens de testen de EVO droegen. Geconcludeerd werd dat het effect van een EVO op loopvaardigheid statistisch significant is maar dat, vergeleken met de a-priori gedefinieerde waarden, de verschillen te klein zijn om klinisch relevant te zijn. Het positieve effect op de zelfverzekerdheid en ervaren moeilijkheid suggereert dat andere factoren een belangrijke rol spelen bij de motivatie om een EVO te gebruiken.

Hoofdstuk 7 richt zich op de implementatie van een nieuwe behandelmethode. Het belang van het uitdragen van kennis en meer nog de implementatie van deze kennis in de klinische praktijk is groot. Hierdoor wordt er nadrukkelijk rekening gehouden met de potentiële gebruikers van de resultaten van wetenschappelijk onderzoek.
Dit onderzoek richtte zich op de overdracht van specialistische kennis van een orthopedisch chirurg naar personeel van een bewegingsanalyse laboratorium in een revalidatiecentrum teneinde de geschiktheid van CVA patiënten voor een specifieke chirurgische ingreep, de "split anterior tibial tendon transfer" (SPLATT), te bepalen. Resultaten van lichamelijk onderzoek en gangbeeld analyse (video en EMG) van patiënten uit het revalidatiecentrum werden besproken met de expert in een ander land. Hierbij werd gebruik gemaakt van personal computers, een ISDN verbinding (128kbit/s) en op TCP/IP gebaseerde communicatie software. Het centrale punt in dit experiment was of het personeel van het bewegingsanalyse laboratorium op termijn beter in staat zou zijn de goede indicatie voor de SPLATT te stellen. Twaalf patiënten werden op de beschreven manier gemeten en besproken. Bij de eerste drie van vier patiënten veranderde het operatievoorstel op advies van de expert. Daarna gaven de besprekingen geen aanleiding meer om de operatievoorstellen te veranderen. Na acht besproken patiënten was het niet meer nodig om advies omtrent operatievoorstellen in te winnen. Er was sprake van een snelle groei in de vaardigheid om de indicatie voor operatie te stellen. De conclusie is dat de vaardigheid op deze manier sneller groeit dan zonder de besprekingen met de expert.

Nawoord

Het is klaar. Mijn proefschrift is af. Het begon allemaal in de zomer van 1983. Na twee maanden als ‘waarnemer’ te hebben gewerkt op de afdeling Fysiotherapie Volwassenen van Het Roessingh kon ik als dienstweigeraar aan de slag op de afdeling Research & Innovatie. Ik had geen flauw idee waar ik eigenlijk aan begon, maar ik was klaar voor het avontuur. Een avontuur dat na iets meer dan 21 jaar geresulteerd heeft in dit proefschrift. Eenentwintig jaar om als onderzoeker volwassen te worden. Een tijd waarin veel is gebeurd.

Tijdens die periode op Het Roessingh ontspon zich een bijna ideaal scenario. Het werk als fysiotherapeut kon gecombineerd worden met het werk op een onderzoeksafdeling. Ervaringen vanuit de praktijk konden meegenomen worden in het onderzoek en kennis uit het onderzoek kon direct in de praktijk ingezet worden. Het leverde me een schat aan kennis en ervaring op. Die kennis en ervaring resulteerden in het midden van de jaren negentig in een idee over hersteltheorieën bij CVA patiënten. Een patiëntengroep die al vanaf de stageperiode mijn specifieke belangstelling had. Na vele en lange discussies gedurende de daarop volgende jaren begonnen de ideeën en de daarbij horende onderzoeksvragen zich steeds meer uit te kristalliseren. Ideeën en vragen die uiteindelijk geformuleerd werden in een onderzoeksaanvraag. Nadat de onderzoeksaanvraag begin 1999 gehonoreerd werd begon ik op 1 maart 1999 met mijn promotieonderzoek. Nu, bijna 6 jaar later, is het af. Al dat werk heb ik natuurlijk niet alleen gedaan. Er zijn ontzettend veel mensen, die ik dank verschuldigd ben. Zonder hen zou dit niet mogelijk geweest zijn. Een aantal wil ik graag bedanken in dit nawoord.

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Rob, zonder jou was ik nooit de onderzoeker geworden die ik nu ben. Niet alleen tijdens het begin van mijn promotie onderzoek, maar vooral in al die jaren daarvoor ben jij degene geweest die me begeleid en voor een groot deel gevormd heeft. Jij hebt me enthousiast gemaakt voor de bewegingsanalyse. De vele uren die we samen op de kamer hebben doorgebracht, de projecten, de discussies, de reizen en het lief en het leed hebben een hechte band gesmeed.
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ervaringen opgedaan om als clinicus met een ingenieur samen te werken. Later tijdens de dagelijkse begeleiding van het promotieonderzoek hebben je betrokkenheid, je kennis en inzicht en je vrolijke en positieve aard een grote indruk op me gemaakt.

Gerrit Zilvold was met name belangrijk in zijn faciliterende rol voor het onderzoek. Gerrit, samen met Hermie heb jij me destijds aangenomen als 'dienstweigeraar' op de afdeling Research & Innovatie. Zonder jou zou er geen avontuur geweest zijn.
Te vaak nog is de klinische epidemiologie een ondergeschoven kindje op het terrein van het bewegingsanalyse onderzoek. Maarten IJzerman wees me op deze beperkingen en heeft me enig inzicht in deze materie gegeven. Maarten, het vertrouwen dat jij in mij stelde als onderzoeker heeft me goed gedaan.
Vanuit de universiteit Twente was er de deskundige inbreng van Peter Veltink. Peter, de samenwerking met jou en je open manier van communiceren heb ik als zeer prettig ervaren.

Het in dit proefschrift beschreven onderzoek was een deel van een het project 'CVA lopen', waarvan Theo Mulder de projectleider was. Theo, als één van de aanvragers van het project heb jij de basis gelegd voor dit onderzoek en als projectleider heb jij de verschillende onderzoeken bij elkaar gehouden en tot een geheel gemaakt.
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Een groep mensen zonder wie dit onderzoek nooit plaats had kunnen vinden, zijn de patiënten die aan de diverse experimenten deelnamen. Zonder uitzondering zijn ze van harte bereid geweest om de vele testen te ondergaan, zonder dat ze daar zelf beter van werden. "Ze deden het voor de ander", hun lotgenoten, die na hen zouden revalideren.

De ervaring die ik opdeed als fysiotherapeut is een wezenlijk onderdeel van dit proefschrift. Ad Moolhuyzen was degene die me destijds benaderde om als invalkracht op de afdeling Fysiotherapie van Het Roessingh te komen werken. Toen na bijna 2 jaar een einde kwam aan het werk als dienstweigeraar kon ik gelukkig weer terecht op diezelfde afdeling Fysiotherapie. Ik heb daar echter niet lang fulltime gewerkt. Al snel kreeg ik een aantal uren tot mijn beschikking om onderzoek te doen. Ad, jouw vooruitziende blik, in een tijd waarin wetenschappelijk onderzoek binnen de fysiotherapie nog schaars was, is van groot belang geweest voor mijn persoonlijke ontwikkeling als onderzoeker.
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resultaten uit het onderzoek in de praktijk werd daardoor een stuk makkelijker. De rol die Martin Tenniglo in die implementatie speelde was zeer belangrijk. Martin, zonder jou zou de implementatie van de resultaten niet zo succesvol zijn geweest.

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CURRICULUM VITAE

Jaap Buurke werd geboren op 8 september 1958 in Musselkanaal. Na het volgen van de middelbare school in Aalten studeerde hij aansluitend Fysiotherapie op de Academie voor Fysiotherapie te Enschede. Hij rondde zijn studie af in 1982, vervulde als onderzoeksmedewerker zijn vervangende dienstplicht op de afdeling Research & Innovatie van Het Roessingh en begon in 1985 als fysiotherapeut op de afdeling Fysiotherapie. Vanaf 1991 werd hij voor 50% van de tijd gedetacheerd naar het huidige Roessingh Research & Development. Daar kreeg hij de mogelijkheid om zich als wetenschappelijk onderzoeker te scholen en zich te verdiepen in de bewegingsanalyse. Binnen het ZonMw programma startte hij in 1999 met zijn promotieonderzoek "Herstel van lopen na een CVA".

Momenteel werkt hij als fysiotherapeut bij het Revalidatiecentrum Het Roessingh en is hij als onderzoeker werkzaam bij Roessingh Research & Development binnen het cluster Non-Invasive Assessment of Neuromuscular Function(ing). Hier werkt hij als senior onderzoeker aan de projecten, "Weke delen chirurgie bij CVA patiënten", "FES for Footdrop in Hemiparesis" (NIH, USA) en, in samenwerking met de Universiteit Twente, aan het project "Effect van revalidatie hulpmiddelen op het evenwicht bij CVA patiënten" (ZonMw). Daarnaast maakt hij deel uit van de projectgroep "Innovatiecentrum voor Revalidatietechnologie".