Perturbing stroke patients

Quantifying dynamic walking stability based on the response to perturbations in order to discriminate between healthy controls and stroke patients with and without falls

Lilja Kristín Siemelink

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by

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This thesis is confidential and cannot be made public until April 24, 2017.

Preface

Hereby I would like to thank everybody who assisted me during this project. First of all, I would like to thank dr. ir. H. Vallery and dr. K. Meijer for the guidance of the project along the whole trajectory. Additionally, I would like to thank Sanne Roeles for the substantive discussions and support on all levels ranging from control software development, technical measurement issues, data analysis until reporting. In my private environment, I would like to thank Tim Holtermann for his support, encouragement and patience during the whole project. The complete project was designed and performed by myself. The control software for my protocol was also created by myself, using an existing environment. I would like to thank the Ethical Committee board of the Sichuan Bayi hospital for approving my protocol. All the measurements were performed at the CAREN Center at Bayi Rehabilitation hospital in Chengdu. Both patients and healthy subjects were recruited at this hospital. Therefore, I would like to explicitly thank dr. Shao, the director of the hospital, for facilitating the recruitment. Additionally, I would like to thank the neurological department leaders for their cooperation and neurological doctor Mike Sun for his assistance in patient recruitment. The recordings of the measurements were performed by myself, assisted by CAREN center employees. I would like to thank physical therapist and CAREN center employee Chao Yu for the translation of the instructions during the measurements, performing the physical therapy measures and the assistance in subject recruitment. Additionally, I would like to thank CAREN center employee Rainsong Liu for his translations and assistance during the measurements and in subject recruitment. The data analysis, statistical analysis and writings were all performed by myself took place at Sichuan Bayi hospital and at the Motek office in Amsterdam. Last but not least, I would like to thank the commissioning party Motek for the realisation of the whole project.

In this report are two experimental studies comprised. In chapter one the overarching topic of the studies is introduced. Chapter two consists of a pilot study. The aim of this study was decision making for a substantiated protocol, as existing literature on this topic was limited. Most importantly, the perturbation type and intensity were decided based on this pilot study. In chapter three is the stroke study presented. Using perturbations based on the pilot study, the responses of stroke patients were compared with healthy subjects. Additionally, the relation between the perturbation response and clinically assessed fall risk was analysed. The main aim was to analyse whether the response to perturbations could discriminate stroke and healthy persons and faller and non-faller stroke patients. A final conclusion can be found in Chapter 4. The Appendix consist of additional rationale and information for different methodological decisions like the amount of perturbations, number of incorporated steps, inclusion of the first perturbation of a trial. Also, the platform trajectory and relation with gait cycle are discussed. Furthermore, questionnaires and materials of the study can be found in the appendix.

Lilja Kristín Siemelink
Den Bosch, April 2017

Contents

1
Introduction

Falls are a big problem in the society (Rubenstein et al., 2006; Callisaya et al., 2011). Most falls occur during walking, as it is the most common loco-motor activity (Callisaya et al., 2011; Bruijn et al., 2013). To identify the fall risk during walking, the dynamic walking stability should be quantified. However, gait is very complex and is studied using many different variables (Verlinden et al., 2013). There is no general consensus on how walking stability is defined. Bruijn et al. (2013) define stable gait as "gait that does not lead to falls in spite of perturbations". A more general definition of stability is to 'avoid falling' comprised by the viability kernel of Wieber (Hobbelen et al., 2007). Full et al. (2002) define stability as the capacity of a system to respond to perturbations. Based on these definitions walking stability in this study was defined as "Gait that does not lead to falls in spite of perturbations due to the capacity to respond to perturbations".

In order to record the subjects response to external perturbations, instantaneous platform perturbations were applied to the walking surface. As not much was known yet about suitable perturbation types and intensities, a pilot study was first conducted to provide insights into different perturbation types and responses of healthy persons. This pilot study is presented in chapter 2. Based on the information of the pilot study, a study was performed including stroke patients and healthy subjects, aiming to discriminate between these groups and to show a relation with clinically assessed fall risk. This study can be found in chapter 3.

2

Pilot study

2.1. Abstract

Introduction Falls are a primary cause of accidental deaths, serious injuries, dependency and society costs. The response to perturbations could identify how well persons are able to reject perturbations. The aim of the study was to compare four different medio-lateral swing perturbations in order to select the most appropriate perturbation type and intensity for further perturbations studies.
Methods Five healthy subjects aged between 18 and 40 were included. Baseline walking on an CAREN was measured, followed by four perturbation trials in which the similar perturbation was twelve times repeated. Subjects were exposed to two different perturbation types; contra-lateral and ipsi-lateral perturbations and two different perturbation intensities; 0.05m displacement in 1.77s and 0.035m displacement in 1.62s. The perturbation response was quantified using the gait sensitivity norm and observational analysis.
Results The lager intensity of 0.05 m showed an increased response to contra-lateral perturbations compared to the lower intensity 0.035 m (p=0.02). Contra-lateral perturbations tend to result in a larger response compared to ipsi-lateral perturbations. Subjects showed opposite responses to contra- and ipsi-lateral perturbations. Following contra-lateral perturbations, subjects decreased MOS and step width in the fist two step following the perturbation and increased step length and step time. In response to ipsi-lateral perturbations, subjects increased their MOS and step width, but decreased step length and step time.
Conclusion The contra-lateral perturbation of 0.05 m intensity and the described protocol were recommended for further studies in order to discriminate fall-prone subjects.

2.2. Introduction

In the population above 65 years in the United States, falls are the primary cause of accidental death. Of persons older than 75 years, falls even account for 70% of accidental deaths (Fuller et al., 2000). Furthermore, falls result in serious injuries, loss of confidence, dependency and high society costs (Rubenstein et al., 2006; Callisaya et al., 2011). Therefore, it is important to identify fall prone individuals with the aim to reduce their fall risk in training programs.

Most falls occur due to a trip, slip, misplaced steps or a push (Leavy et al., 2015). These unexpected perturbations could result in a fall when a person is not able to respond adequately to such events. As walking is defined stable when it does not lead to falls despite perturbations (Bruijn et al., 2013) due to the systems capacity to respond to these perturbations (Full et al., 2002), it appears essential to study a subjects response to external perturbations when quantifying the walking stability. By analysing the response to a perturbation, it might be possible to quantify dynamic walking stability with a closer relationship to actual falling compared to steady state walking (McAndrew et al., 2011; Bruin et al., 2013; Bruijn et al., 2011;Yang et al., 2014).

Bauby and Kuo (2000) proposed lateral stability to be more important than anterio-posterior stability in human walking, as active control is needed to ensure lateral stability while for anterio-posterior stability passive dynamic properties were utilised. In accordance, McAndrew et al. (2010) showed that subjects were more sensitive to medial-lateral perturbations compared to anterior-posterior perturbations. This indicates that perturbing a subject in medio-lateral direction would require a more evident response to remain walking stability. A medio-lateral sway perturbation of the movable surface most closely represents a push or loss of balance in daily life.

Previous studies perturbed subjects during walking in order to study their response. Slip and trip perturbations were applied in several study designs. In many of these cases, the experimental setup consisted of a long walkway in which a stumble or slip mimicking device suddenly applied a perturbation to the subject (Yang et al., 2012;2014; Grabiner et al., 1993). The methods in which these responses to perturbations were analysed varied widely. Often, there was focused on a learning or adaptation effect (Quintern et al., 1985; Heiden et al., 2006; Yang et al., 2014; McCrum et al., 2014). Several studies used perturbations during continuous walking (McAndrew et al., 2010; Hak et al., 2012; O'Connor and Kuo, 2009), these perturbations were applied as pseudo-random oscillations and the subject's continuous adaptations were analysed while walking on moving surfaces.

Instantaneous perturbations were not often used in order to mimic external perturbations that could result in falls. Therefore, not much is known yet about suitable intensities, perturbation courses and response analysing methods. Although a possible learning effect might be essential and promising when developing treatment programs in order to increase the dynamic walking stability, the question still remains how to identify fall prone adults in order to expose the right subjects to fall preventive training, resulting in a reduced amount of falls. Essential for discrimination between fallers and non-fallers is to find the characteristics contributing to decreased dynamic stability. With the use of currently available technology, it is possible to perturb individuals in a safe and standardized manner while continuously walking and to study their responses. This method differs from gangway walking in which only intermittent walking can be performed and the walking speed cannot be standardised.

The aim of the study was to compare four different medio-lateral swing perturbations in order to select the most appropriate perturbation type and intensity for further perturbations studies. Additionally, the subjects response to perturbations and the method to analyse this response will be evaluated. Several healthy subjects will be subjected to different perturbation types while continuously treadmill walking. Their response in the four steps following the perturbations will captured in one encompassing measure that represents the magnitude of the response compared to baseline. Based on the findings of this pilot study, a more substantiated research protocol can be designed with the aim to identify individuals with decreased walking stability. It is hypothesised that a larger perturbation intensity results in larger responses and that a contra-lateral perturbation results in an a larger perturbation response compared to the ipsi-lateral perturbation, as the contra-lateral perturbation moves the (extrapolated) center of mass closer to the borders of the base of support, which decreases the margins of stability.

2.3. Methods

2.3.1. Subjects

Five healthy subjects aged between 18 and 40 were recruited (table 2.1). Participants showed no motor or sensory impairments and no history of lower extremity injury or surgery. Ethical approval of the protocol was obtained from the Human Research Ethics committee of Sichuan 81 Rehabilitation hospital. All subjects signed informed consent before participation.

Table 2.1: Subject characteristics

Subject	**Age (years)**	**Gender (M/F)**	**Walking speed (m/s)**	**Leg length (m)**
1	27	Male	0.7	0.77
2	31	Male	0.7	0.75
3	32	Female	0.7	0.75
4	25	Male	0.7	0.77
5	21	Female	0.7	0.80
mean (SD)	27.20 (SD 4.49)	3 M/2 F	0.7 (SD <0.1)	0.77 (SD 0.02)

2.3.2. Materials

The measurements were performed using a Computer Assisted Rehabilitation Environment (Figure 2.1, CAREN; Motekforce Link, Amsterdam, at Sichuan Bayi Rehabilitation Hospital, Chengdu) consisting of a 6 degrees of freedom movable platform (E2M Technologies, Amsterdam), a dual-belt instrumented treadmill and a spherical projection screen with a virtual reality. 47 reflective markers were applied to the subject following the HBM full body marker model (Van den Bogert et al., 2013; Motekforce Link, Amsterdam). The marker coordinates were captured using 12 infra-red motion capture cameras with a sample frequency of 100 Hz (Vicon, UK). Force plate data (1000 Hz) were down-sampled to synchronise motion capture data. Platform movements were recorded using 3 reflective markers on the platform pane. All data was filtered using a unidirectional Lowpass Butterworth filter with a cutoff frequency of 6 Hz.

Figure 2.1: Computer Assisted Rehabilitation Environment (CAREN) at Bayi Hospital, China

2.3.3. Measurement protocol

Familiarization

The experimental trial started with a familiarization walk of three minutes on the treadmill at a dimensionless speed of $v_d = 0.25$, which is a slow walking speed (Hof et al., 1996; McAndrew et al., 2010; Schwartz et al., 2008). A low walking speed was chosen so that the findings would be applicable in further studies with patient populations walking at a low speed. The actual walking speed v in m/s was calculated using equation 2.1, in which l is the vertical leg length in meters from the lateral malleolus to the trochanter major. This fixed walking speed was used during all trials.

$$v = v_d \cdot \sqrt{(9.81\,\mathrm{m/s^2} \cdot l))} \qquad (2.1)$$

Experimental protocol

After the familiarization, two minutes of baseline walking were recorded. Following this, four medio-lateral sway perturbation trials were executed in a randomized order. The four perturbations types consisted of two different intensities and two different onset directions; contra- and ipsi-lateral. The high intensity was characterised by a displacement of 0.05 m in 1.77 s with maximum velocity of 0.11 m/s and a maximum acceleration of 0.92 m/s^2. The lower intensity had a displacement of 0.035 m in 1.62 s with a maximum velocity and acceleration of respectively 0.08 m/s and 0.71 m/s^2. During each trial, the subjects were exposed to 12 medio-lateral sway perturbations of similar intensity and similar direction. The left and right order were randomised. All perturbations were initiated at initial contact of the gait cycle and peak accelerations were perceived around mid stance. A random interval of 10-15 strides was used between perturbations. The patients were instructed to keep walking during the whole trial and to look forward to the virtual environment with visual flow. The appendix A.4.2 can be consulted for a more extended description of the platform trajectory, velocities, accelerations, delays and gait cycles.

2.3.4. Data analysis

Gait events

The gait events 'initial contact' (IC) and 'toe-off' (TO) were calculated following the method of Zeni et al. (2008) (Figure 2.2). Initial contact was determined at the maximal distance between the sacral marker coordinate (Z_s) and the heel marker coordinate (Z_h) in the longitudinal direction of the treadmill during a gait cycle. Toe-off was found at the minimal distance between the sacral marker coordinate (Z_s) and the toe marker coordinate (Z_t). In Appendix A.3 the rationale for this decision can be found.

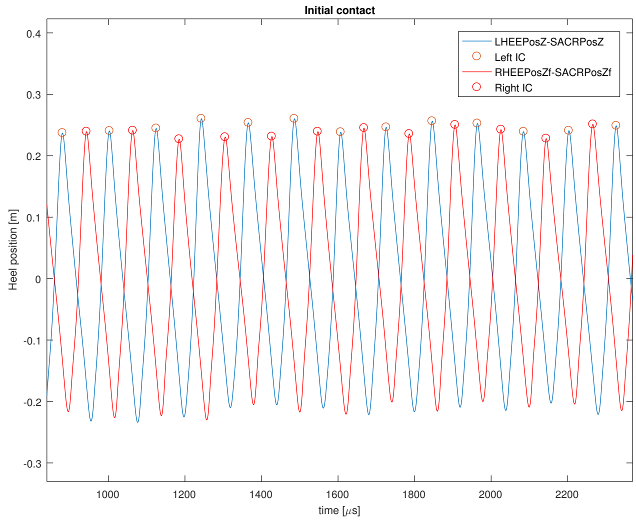

Figure 2.2: The trajectories of the heel marker with respect to the sacrum are displayed in the longitudinal direction of the belt (z) for the left (blue) and right (red) foot. Each determined initial contact instance is marked with a circle

Spatio-temporal parameters

Step length (SL), step widt (SW) and step time (ST) were calculated using the IC gait event. SL and ST were defined as the forward distance, corrected for the traveled belt distance, and time respectively of the heel marker at two successive IC gait events. SW was defined as the medio-lateral distance between the right and left heel marker at the IC event (Hak et al., 2012). An example of subsequent step lengths following perturbations is presented in figure 2.3.

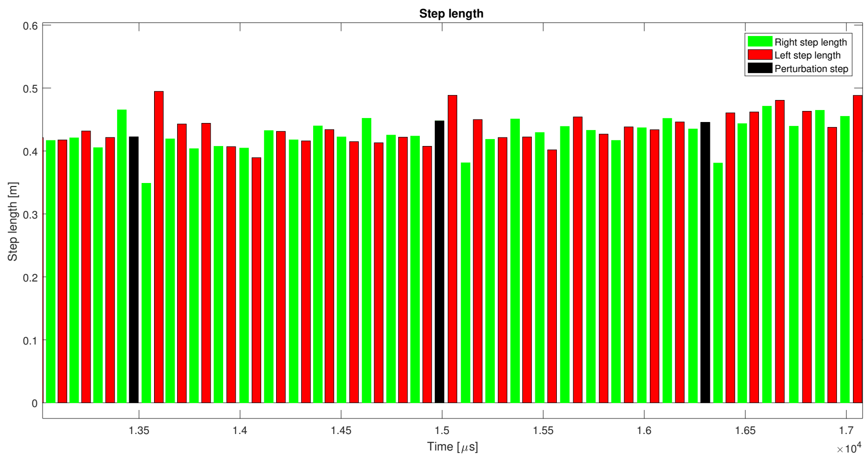

Figure 2.3: The magnitude of subsequent left (red) and right (green) step lengths are presented in bars next to each other, with perturbation steps (black) in between.

Margin of stability

The margins of stability (MOS) in the mediolateral direction were calculated as proposed by Hof et al. (2005) using the extrapolated centre of mass (XCOM) (equation 2.2). The XCOM was calculated by adding the velocity component $\left(\frac{V_{CoM}}{\omega_0}\right)$ to the center of mass (COM). In which ω_0 the eigenfrequency was, represented by $\omega_0 = \sqrt{\frac{g}{l}}$. This included the gravitational acceleration ($g = 9.81\,\mathrm{m/s^2}$) and the leg length l in meters.

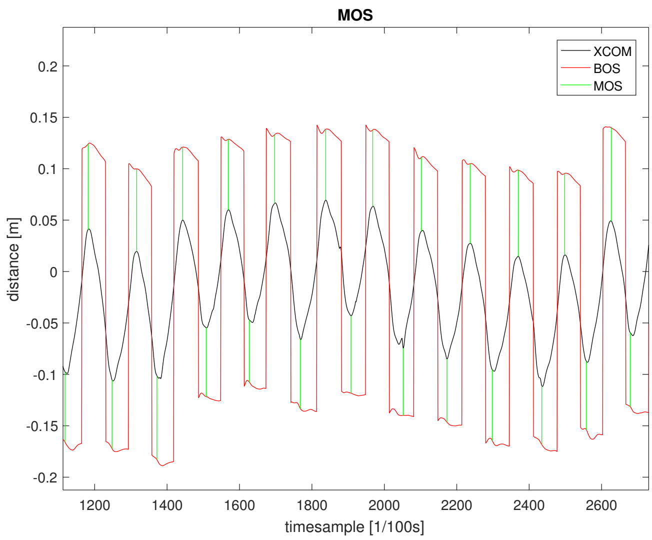

Figure 2.4: The margins of stability (green lines) are calculated as the minimum distance between the borders of the base of support (red) and the extrapolated center of mass (black) and the green line is the MOS

$$XCOM = COM + \frac{V_{COM}}{\omega_0} \tag{2.2}$$

The margins of stability (MOS) were defined as the minimal distance between xCOM and the base of support (BOS) following equation 2.3 for each step. The BOS was calculated using the markers of the lateral malleoli on the feet (Hak et al., 2012), which represent the lateral borders of the covert ground area by the feet.

$$MOS = BOS - XCOM \tag{2.3}$$

Baseline walking

Baseline walking outcomes were calculated as mean over 40 steps for all parameters. The first 10 strides were disregarded to provide some stabilizing time within the trail. The baseline outcomes were; step length, step width, step time and MOS.

Gait sensitivity norm

In order to define dynamic walking stability in humans, the ability to reject external perturbations was calculated using the Gait Sensitivity Norm (GSN) (Hobbelen et al., 2007). The equation of Hobbelen et al. (2007) was slightly adjusted to be applicable to humans, using a fixed amount of steps and a single perturbation intensity per trial (Equation 2.4). u represents the amount of gait indicators and v the number of steps after the perturbation, which was fixed to 4. The gait indicators (h) are: SL, SW, ST and MOS. Baseline values of the gait indicators (*) were subtracted from each k-th step of the indicator. The GSN outcome represents the size of the dynamic response, in which a lower values corresponds to better disturbance rejection (Hobbelen et al., 2007). In the appendix A.6 is explained why 4 steps were incorporated.

$$GSN = \sqrt{\sum_{i=1}^{u} \sum_{k=0}^{v} \left(h_i(k) - h_i^*\right)^2} \tag{2.4}$$

To account for the amount of variance in baseline walking, a corrected gait sensitivity norm (GSN_c) (Aarts et al. [in submission]) was also calculated (Equation 2.5). The gait indicator outcomes were divided by their baseline variance (σ).

$$GSN_{corr} = \sqrt{\sum_{i=1}^{u}\sum_{k=0}^{v}\left(\frac{h_i(k)-h_i^*}{\sigma_i^*}\right)^2} \tag{2.5}$$

GSN per variable

The GSN summed the outcomes of all gait indicators. Additionally, the GSN was calculated for each gait indicator following a similar approach (equations 2.6, 2.7, 2.8 and 2.9).

$$GSN_{SL} = \sqrt{\left(\sum_{k=1}^{k=4}(SL(k)-SL^*)\right)^2} \tag{2.6}$$

$$GSN_{SW} = \sqrt{\left(\sum_{k=1}^{k=4}(SW(k)-SW^*)\right)^2} \tag{2.7}$$

$$GSN_{ST} = \sqrt{\left(\sum_{k=1}^{k=4}(ST(k)-ST^*)\right)^2} \tag{2.8}$$

$$GSN_{MOS} = \sqrt{\left(\sum_{k=1}^{k=4}(MOS(k)-MOS^*)\right)^2} \tag{2.9}$$

2.3.5. Statistical analysis

In order to statistically analyse the within group differences of the GSN, GSN$_c$ and GSN per variables, two null hypothesis were conceived:

- $H0_1$: No difference exists between the outcome measures of 0.05 m and 0.035 m perturbation intensities.
- $H0_2$: No difference exists between the outcome measures of contra-lateral and ipsi-lateral perturbations.

A within subject design was compiled since the same participants were subjected to different perturbation intensities and sides. The mean over 10 perturbations was taken for each condition and for all outcome measures: GSN, GSN$_c$, GSN$_{SL}$, GSN$_{SW}$, GSN$_{ST}$, GSN$_{MOS}$. When normally distributed, group differences were statistically compared using paired sample t-test. In case the data sets were not normally distributed, the non-parametric Wilcoxon signed-rank test was performed. A Bonferroni correction was applied to reduce likelihood of incorrectly rejecting the null hypotheses (a type I error), because of two null hypothesis. Therefore, the original $\alpha = 0.05$ was divided by the number of hypothesis $\alpha_{corr} = 0.05/2 = 0.025$. This means that the null hypothesis was only be rejected when the p-value was below α_{corr}.

2.3.6. Observational analysis

Observation analysis was performed on the individual step outcomes following a perturbation. This workflow is presented in figure 2.5. The observational analysis was performed for further interpretations of the findings, as data was be averaged in the comprising GSN measures.

2.4. Results

2.4.1. Statistical analysis

The mean outcome values and standard deviations of the GSN per variable, GSN and GSN$_c$ are presented in Table 2.2. In figures 2.6 and 2.7 the differences in mean (SD) values are shown for GSN and GSN_c. Statistical results can be found in Tables 2.3 and 2.4.1. The GSN and the GSN_c showed significantly larger responses for the higher intensity (0.05m) compared to the lower intensity (0.035m) for contra-lateral perturbations (p = 0.02). These differences were not seen for the ipsi-lateral perturbations or for any of the GSN values based on only one gait indicator.

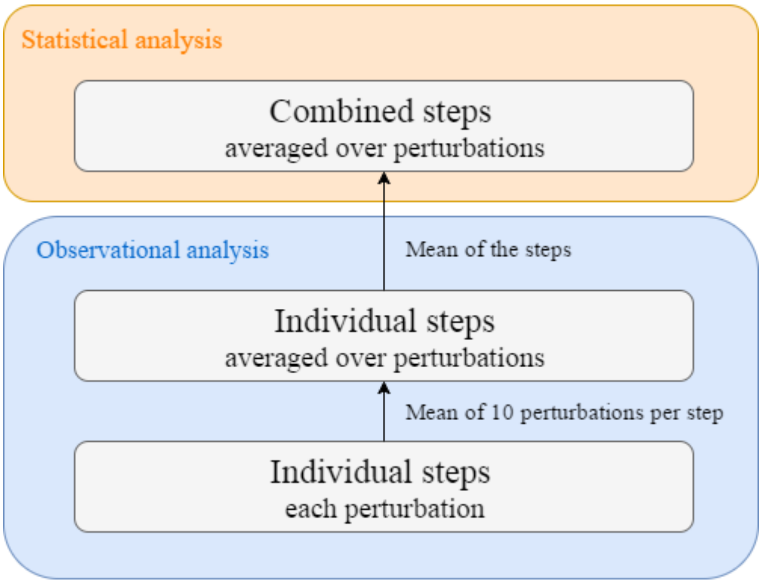

Figure 2.5: Statistical and observational analysis

No significant differences were found between the two perturbation types: contra-lateral and ipsi-lateral perturbations (Table 2.4.1). Trends with larger mean values for contra-lateral perturbations compared to ipsi-lateral appeared for all GSN outcomes at high intensity. At lower intensity these trends were less profound. Contra-lateral perturbations resulted in larger mean value trends for GSN_{SL}, GSN_{SW}, GSN and GSN_c compared to ipsi-lateral perturbations.

Table 2.2: Mean values and standard deviations of the GSN outcomes of the four coditions; contra-lateral 0.05m, ipsi-lateral 0.05m, contra-lateral 0.035m and ipsi-lateral 0.035m

	Contra 0.05m		**Ipsi 0.05m**		**Contra 0.035m**		**Ipsi 0.035m**	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
GSN Step length [m]	0.148	0.081	0.140	0.053	0.121	0.063	0.121	0.062
GSN Step width [m]	0.181	0.118	0.126	0.058	0.122	0.065	0.114	0.043
GSN Step time [s]	0.143	0.087	0.136	0.047	0.116	0.060	0.125	0.045
GSN MOS [m]	0.078	0.054	0.065	0.025	0.053	0.025	0.057	0.023
GSN	0.184	0.091	0.154	0.036	0.137	0.046	0.131	0.034
GSN_c	1.270	0.563	1.073	0.270	0.929	0.267	0.907	0.222

Table 2.3: Intensity comparison of 0.05m with 0.035m - Mean differences and p-values of the paired sample T-test. Significance was reached for p values below $\alpha = 0.025$

Intensity differences (0.05 vs 0.035)	**Contra**		**Ipsi**	
	Mean difference	**p-value**	**Mean difference**	**p-value**
GSN step length [m]	0.026	0.03	0.018	0.42
GSN step width [m]	0.059	0.03	0.013	0.45
GSN step time [s]	0.027	0.16	0.011	0.18
GSN MOS [m]	0.024	0.08	0.008	0.06
GSN	0.048	**0.02**	0.023	0.07
GSNcorr	0.340	**0.02**	0.166	0.06

2.4.2. Observational analysis

In figure 2.8 are the mean outcomes for each step visualised in response to all four perturbation types. In red is the confidence interval presented during normal walking. The statistical findings and trends on the combined steps are corresponding to the findings based on the mean steps. A larger response is shown for

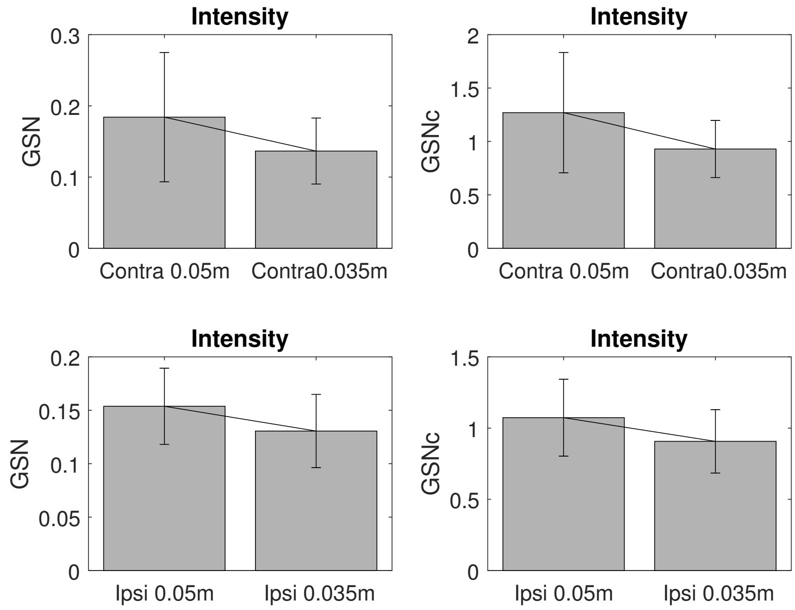

Figure 2.6: Differences in magnitude of the response to perturbation for two ipsi-lateral intensities (0.05m and 0.035m). The outcome values present the magnitude of the normalised combined steps.

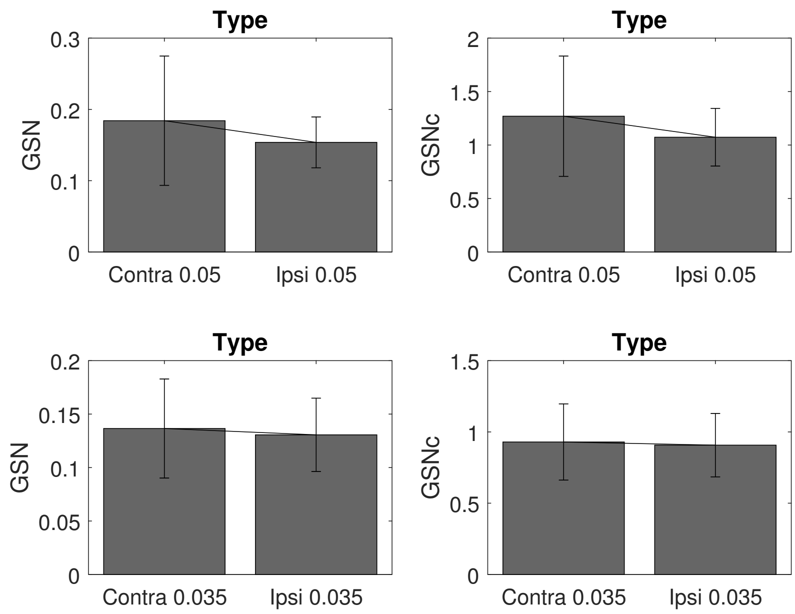

Figure 2.7: Differences in the corrected GSN for two different conditions and intensities.

Table 2.4: Perturbation onset side comparison contra-lateral and ipsi-lateral - Mean differences and p-values of the paired sample T-test. Significance was reached for p values below $\alpha = 0.025$

Type differences (contra vs ipsi)	**0.05m**		**0.035m**	
	Mean difference	**p-value**	**Mean difference**	**p-value**
GSN step length [m]	0.008	0.28	-0.001	1.00
GSN step width [m]	0.055	0.10	0.008	0.72
GSN step time [s]	0.008	0.73	-0.009	0.58
GSN MOS [m]	0.013	0.29	-0.003	0.53
GSN complete	0.030	0.14	0.006	0.57
GSNcorr complete	0.120	0.15	0.022	0.75

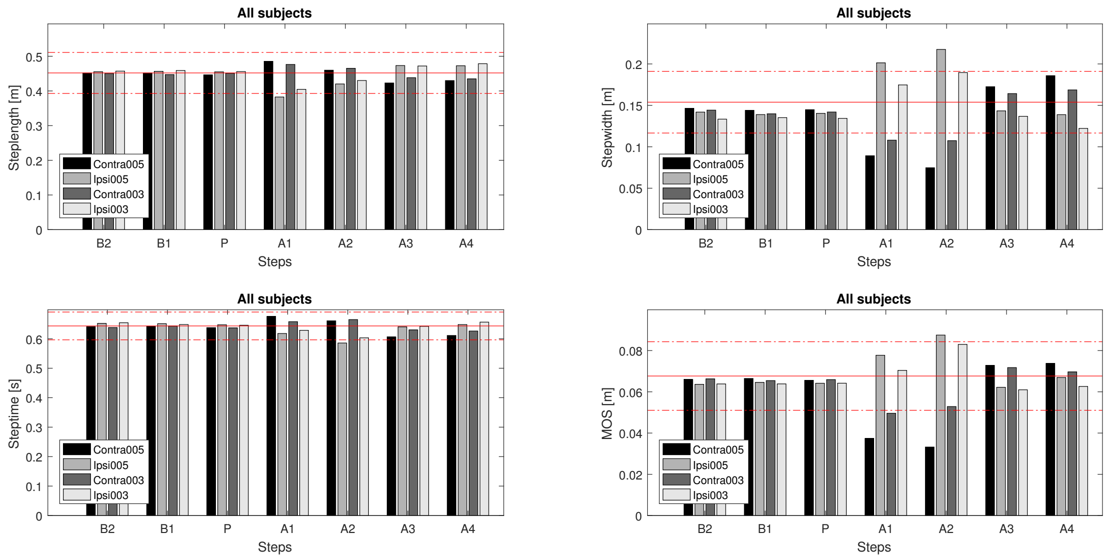

Figure 2.8: Seven sequential steps are displayed for 4 different perturbation types. P was the perturbed step, the responses of this perturbation are visible in the steps after perturbation (from A1 to A4). Two steps before the perturbation were shown as baseline (B1 and B2). Black and dark grey are the contra-lateral perturbations with 0.05m and 0.035m intensity respectively. Light grey and white are the ipsi-lateral perturbations with 0.05m and 0.035m intensity respectively. The confidence interval of normal walking is presented in red as reference.

the larger intensity. This is especially visible at the first two steps following the perturbation, A1 and A2, for all variables. The magnitude seemed larger for the contra-lateral perturbations. Furthermore, opposite responses were seen in contra-lateral and ipsi-lateral perturbations. Following a contra-lateral perturbation, the margin of stability and the step width decrease at A1 and A2. After an ipsi-lateral perturbation, the opposite occurred as the margin of stability and step width increased at A1 and A2. In addition, the step length and step time seemed to increase in the first steps after a contra-lateral perturbation and decrease following a ipsi-lateral perturbation.

2.5. Discussion

The aim of this study was to analyse the responses of healthy participants to four medio-lateral perturbation types in order to select the most suitable perturbation for future study protocols including patients. Five healthy participants were exposed to the perturbations types to identify which perturbation intensity and type were most appropriate.

The GSN is a comprising measure which expressed the absolute magnitude of the response to a perturbation. Significantly larger responses were shown for the higher intensity (0.05m) compared to the low intensity (0.035m) in contra-lateral perturbations (p = 0.02). This means that a significantly better disturbance rejection was shown for the lower intensity compared to the higher intensity (Hobbelen et al., 2007). As the platform moved a larger distance in approximately the same time, higher velocities and accelerations were

reached. The intensity of the perturbation conceived by the subject was larger, which resulted in increased anticipation in order to respond adequately to this perturbation. This is in accordance with the study of Lars et al. [in submission], who deployed different perturbation intensities and found larger GSN outcomes for increasing intensity using both sway and acceleration/deceleration perturbations. In the study of Hobbelen et al. (2007) was a 2D limit cycle walker and the response to four varying perturbation intensities modeled and measured. The perturbation intensities were defined by the height of step-downs in the floor. Larger step height disturbance only resulted in higher GSN values or worse disturbance rejection in the highest compared to the second highest intensity. Therefore, a larger intensity does not necessarily have to result in a larger response. Since both the perturbation and the gait indicators were chosen differently, we cannot directly relate their outcomes to our outcomes.
Although not significant, the response to the contra-lateral perturbation side tend to be larger than the ipsi-lateral perturbation response for the higher intensity. For this reason it makes sense that with low statistical power only the differences in intensity for the contra-lateral perturbation appeared with significance. When including a larger sample group, more substantiated statements could be derived. In this study, the GSN and GSN_c revealed the largest response to a the contra-lateral 0.05 m perturbation, meaning that persons showed larger adjustments to overcome the perturbation. Therefore, this is the most appropriate perturbation to deploy in further studies.

The GSN and GSN_c showed significant differences, while the GSN per variable did not. As the GSN comprises the response of many different variables in one outcome, it represents the absolute magnitude of the response of gait indicators combined resulting in a larger difference compared to individual measures. This is beneficial for statistical analysis and interpretation. However, individual trades are masked. Therefore, the GSN per variable and observational analysis were used to further explain the manner in which subjects responded to perturbations.

The largest responses to the perturbations were seen in the two steps directly following the perturbation. Following contra-lateral perturbations, the step width and MOS decreased, while the step length and step time increased. The ipsi-lateral perturbation resulted in an opposite effect with an increased step width and MOS and a decreased step length and step time. When a contra-lateral perturbation was applied, the platform moved for example to the right on a left step, causing the center of mass (and it's velocity) to move outwards. Following this, a step was performed with decreased step width. When decreasing the step width, it is likely for the MOS to decrease as well as the border of the base of support is decreased and the extrapolated center of mass is closer to the border of the base of support. The opposite occurred following an ipsi-lateral step after which the step width was increased. In the second step after the perturbation, the same effects were seen as in the first step. In the steps following, an increased step width and MOS for the contra-lateral perturbation and a decreased step width and MOS following the ipsi-lateral perturbation. This might be a compensatory effect of the third and fourth steps compared to the first two steps for the either decreased or increased MOS. A recent study (Punt et al., 2017) also investigated contra-lateral and ipsi-lateral sway perturbations. They however applied these perturbations to a stroke population. Similar step response patterns were found in response to the ipsi-lateral perturbations for step width, step length and step time. For the contra-lateral perturbations, lower responses were shown in their study. However, based on this data we cannot draw conclusions on whether these differences are a result of the different populations or different test conditions (i.e. perturbation intensity).

Step length and step time also showed opposite responses to contra- and ipsi-lateral perturbations, although these responses were less apparent with respect to the confidence interval. Following a contra-lateral perturbation, step length and time increased for the first two steps while they decreased following ipsi-lateral perturbations. This was in accordance with the response seen performed with the paralysed limb in the study of Punt et al. (2017). The responses with the non-paralyzed limb did not show this pattern. In our study, a compensatory response was seen for the third and fourth step. The increased step length is related to the increased step time when walking speed remained unchanged, as there is more time to place the foot further away (Signleton et al., 1992). Additionally, a relation between the reduced MOS and increased step time could exist. Elderly with reduced stability showed larger step times compared to young persons (Lord et al., 1999). As the MOS showed that stability is reduced, increased time could be required to ensure save foot placement.

The GSN_c scaled the outcome with respect to their standard deviation (σ) during normal walking. On the one hand, this seemed useful as steady state walking variability was eliminated from the perturbation response (Aarts et al., [in submission]). On the other hand, this might neglect important aspects in the walking response. When recording the disturbance rejection, we would like to quantify the response that was provoked by the disturbance and not variability that was already present during steady state walking. However, as increased variability in steady state walking was found in elderly and patients (Maki et al., 1997; Hausdorff et al., 2001; Sosnoff et al., 2012; Flegel et al., 2012), this might be related to the dynamic walking stability. Compensating for this steady state variability might result in unjustified lower values for disturbance rejection in persons with high baseline variability. Hobbelen et al. (2007) tried to reduced steady state variability by increasing the number of trials and therefore reducing the confidence interval. Both GSN and GSN$_c$ methods contain advantages and disadvantages and give therefore useful insights.

Based on this pilot study, some recommendations for further studies can be drawn. In order to select the intensity and perturbation type resulting in the largest response, it is recommended to choose the intensity of 0.05 m. Additionally, it is recommended to choose the contra-lateral perturbation type as the response appeared with larger magnitude and is therefore more likely to highlight differences in inadequate responses as the required responses are higher. Based on this pilot study, it is recommended to infer a larger study involving a vulnerable group, for example elderly fallers or a patient group to employ specific characteristics contributing to decreased dynamic walking stability in order to indentify fall-prone subjects. Furthermore, for further studies it is recommended to apply the perturbation type and analysis as described in this study.

2.6. Conclusion

The lager intensity of 0.05 m showed an increased response to contra-lateral perturbations compared to the lower intensity 0.035 m. Contra-lateral perturbations tend to result in a larger response compared to ipsilateral perturbations. Subjects showed opposite responses to contra- and ipsi-lateral perturbations. Following contra-lateral perturbations, subjects decreased MOS and step width in the first two step following the perturbation and increased step length and step time. In response to ipsi-lateral perturbations, subjects increased their MOS and step width, but decreased step length and step time. The contra-lateral perturbation of 0.05 m intensity and the described protocol were recommended for further studies in order to discriminate fall-prone subjects.

3

Study with stroke survivors

3.1. Abstract

Introduction Around 40% of stroke patients show residual walking disabilities that increase fall risk. Therefore, it is important to identify patient specific responses related to fall risk. The aim of this study was to determine whether the ability to recover from external perturbations on the walking surface could discriminate patients from healthy controls and discriminate fallers from non-fallers in the stroke patient group. In addition, relations were studied between clinical fall assessments and perturbation responses.
Methods 14 stroke patients and 15 healthy controls were included in the study. Baseline walking on a CAREN was measured, followed by a trail with ten contra-lateral perturbations with an 0.05m displacement in 1.77s. The trials were performed in fixed speed and self-paced walking. The perturbation response was quantified using the gait sensitivity norm and observational analysis. Group differences in perturbation response were tested between stroke and healthy subjects and within the stroke patient group between fallers and non-fallers. Clinical fall assessments were correlated to GSN outcomes.
Results Stroke patients showed a larger perturbation response based on the gait sensitivity norm compared to healthy controls ($p = 0.04$) in fixed speed walking. In self-paced walking, stroke patients showed a larger gait sensitivity norm response (corrected)($p=0.04$) and a larger step time ($=0.02$) and MOS response ($p=0.03$). Stroke patients showed reduced step width response ($p=0.03$). No differences in perturbation responses were found between stroke patients with and without fall history. Positive correlations were found between the timed up and go (TUG) score and GSN outcomes ($p=0.03$, $p=<0.01$, $p=<0.01$, $p=<0.01$).
Conclusion The lacking identification of fall prone stroke patients can be related to the inconsistency of current clinical fall risk assessments. The correlation between TUG and GSN can be explained because both are performance measures instead of subjective assessments. As stroke patients compensate for deficits in functionality, the higher GSN outcomes might indicate a less efficient way to cope with the perturbation compared to healthy controls. Although the GSN does not specifically indicate which gait indicator showed an enlarged response and how stroke patients and healthy persons reacted differently on a step basis, it does give a discriminative overall response between stroke patients and healthy subjects. Therefore, it might be an effective way to quantify the response to perturbations.

3.2. Introduction

Stroke is a major cause of chronic impairment and disability (Chee et al., 2014). Around 40% of post-stroke patients show residual walking disabilities that increase fall risk (Chee et al., 2014). The fall incidence in post-stroke patients is 73% (Nott et al., 2014) and in general, 70% of falls lead to serious injuries and even death (Fuller et al., 2000). Additionally, falls result in loss of confidence, dependency and high society costs (Rubenstein et al., 2006; Callisaya et al., 2011). Therefore, it is important to identify fall prone stroke patients with the aim to reduce their fall risk in a training program.

Falls will occur when persons are not able to respond adequately to unexpected perturbations. Most falls occur as result of a trip, slip, misplaced steps, a push (Leavy et al., 2015) or loss of balance (Hyndman et al. 2002). Walking is defined stable when it does not lead to falls despite perturbations (Bruijn et al., 2013) due to the system's capacity to respond to these perturbations (Full et al., 2002). Therefore, it appears essential to study a subjects response to external perturbations. Using the response to a perturbation, it might be possible to quantify dynamic walking stability with a close relationship to actual falling and to identify characteristics of fall-prone adults.

Previously in a gait-slip experiment, stroke patients showed a backwards falling movement followed by stepping strategies to regain the stability (Kajrolkar et al., 2014). Krasovsky et al. (2013) showed that stroke patient used more strides to recover from a leg arrest perturbation compared to healthy subjects. Furthermore, stroke patients showed reduced ability to make visually triggered step adjustments (Nonnekes et al., 2010). As walking is usually quantified during steady state walking, deficits in perturbation responses remain undetected. In a recent study of Punt et al. (2017) the responses on instantaneous perturbations during continuous treadmill walking were measured in stroke patients with and without fall history. Although interesting findings were conceived, no significant differences were found between fallers and non-faller. No comparison was however made with healthy subjects responding to the same perturbations. These comparisons could identify patient specific or lacking responses to perturbations. It was also shown that multiple slip-perturbation improved the ability to use feedback control to improve slip outcomes, improving the patient's ability to recover from perturbations. Accordingly, healthy persons needed less steps to regain baseline margin of stability level after repeated perturbations (McCrum et al., 2014). These two studies showed an adaptation or learning effect based on perturbation training and demonstrated the relevance of mimicking environmental perturbations to incorporate these in training programs. However, first it is of high importance to quantify walking stability in a manner that could discriminate between fall-prone adults and non-fall-prone adults based on perturbation responses.

Treadmill walking was shown to be effective in allowing for continuous gait recording and thereby reducing the variability compared to intermittent over ground walking (Paterson et al., 2009). However, some contradictions are present in literature about the differences between treadmill walking and over ground walking and the clinical relevance of this difference. Alton et al. (1998) did find some differences in kinematics and kinetics, while Riley et al. (2006) reported that the kinematic and kinetic data are quantitatively and qualitatively similar. They stated that the differences were within variability of normal walking and the magnitude of the difference is therefore clinically irrelevant. However, possible existing differences could be explained due to the inability to adjust the walking speed in fixed speed treadmill walking compared to overground walking. Based on this paradox, self-paced treadmill walking was developed in which the subjects were able to adjust the belt speed in order to more closely match overground walking (Plotnik et al., 2015; Sloot et al., 2014). Although there still some essential differences with overground walking, subjects do have the ability to adjust the walking speed.

The aim of this study was to determine whether the ability to recover from external perturbations on the walking surface could discriminate patients from healthy controls and discriminate fallers from non-fallers in the stroke patient group. Furthermore, the influence of self-paced walking on the response to perturbations and the relation between the response and clinically assessed fall risk were analysed. Unilateral stroke patients and healthy controls will be subjected to repeated perturbations of the same type and intensity while continuously walking on a treadmill on a fixed speed and on self-paced walking. It is hypothesised that stroke patients will show a larger response to the perturbation compared to healthy controls. Furthermore, it is expected that the outcomes can to discriminate fallers from non-fallers in the stroke patient group and that relations exist between the outcome measures and clinically assessed fall risk.

3.3. Methods

3.3.1. Subjects

Ethical approval of the protocol was obtained from the Human Research Ethics committee of Sichuan 81 Rehabilitation hospital. All subjects signed informed consent before participation.

Stroke patients were included when the following criteria were met: chronic phase post-stroke >3 months after the incident, lower extremity Fugl-Meyer assessment score ≥ 19, ability to walk independent for 20min without assisting devices (reported by patient), no other neurological, muscular-skeletal, cardiovascular disorders or co-morbidity's, no history of lower extremity injury, surgery or low bone density, cognitive function >20 on the short orientation–memory–concentration test (SOMT).

Control subjects were included when: aged between 40 - 80 years, did not show walking disorders, neurological, muscular-skeletal, cardiovascular or co-morbidity's, did not use medicines, sored above 20 on the SOMT.

Fourteen hemi-paretic stroke patients and fifteen healthy controls participated in the study (Table 3.1). The groups showed no differences in age, gender and cognitive function measured with the SOMT (Table 3.2).

Table 3.1: Subject characteristics (mean SD)

	Stroke	**Healthy control**
Subjects (#)	14	15
Males (#)	10	9
Females (#)	4	6
Age (years)	53.21 ±15.10	49.80 ±7.84
Weight (kg)	63.67 ±8.17	64.43 ±10.40
Leg length (m)	0.75 ±0.03	0.74 ±0.03
Left affected / non-dominant (#)	7	15
Right affected / non-dominant (#)	7	0
Time post stroke (months)	14.3 ±19.68	-
Haemorrage (#)	7	-
Ischemic (#)	7	-
Fugl-Meyer score	29.67 ±3.29	-
BBS score	50.13 ±5.13	-

3.3.2. Materials

The same experimental setup was used as described in section 2.3.2. Briefly, a Computed Assisted Rehabilitation Environment (CAREN; Motekforce Link, Amsterdam) at Bayi Hopital (Chengdu, Sichuan) was used which contained a dual-belt instrumented treadmill (Motekforce Link, Amsterdam), a 6 DOF motion base (E2M Technologies, Amsterdam), 12 100 Hz motion capture cameras (Vicon, Oxford) and a virtual reality.

3.3.3. Measurement protocol

The experiments started with a familiarization walk of three minutes on the treadmill at a dimensionless speed coefficient (v_d) of 0.25 or the fastest speed possible (see paragraph 2.1). This speed remained constant during the fixed speed (DS) trials. After the familiarization, the subject walked for two minutes to determine the baseline measures. This was followed by a perturbation trial, in which the subjects were exposed to ten medio-lateral sway perturbations, five initiated at the left and five initiated at the right step in a random order. All perturbations were applied at heel strike, receiving the largest platform accelerations around mid stance. Subsequent perturbations were executed with a random amount of ten to fifteen strides in between perturbations. The perturbation type and intensity were chosen based on the findings of the healthy subject study 2.6 and contained contra-lateral perturbations with an intensity of 0.05 m displacement in 1.77 s, with maximum velocity of 0.11 m/s and maximum acceleration of 0.92 m/s^2. These three trials were repeated on a self-paced (SP) comfortable walking speed, including familiarization, baseline walking and the perturbation trial.

3.3.4. Data collection

Clinimetrics including the Berg Balance Scale (BBS), the timed up and go (TUG), the fall efficacy scale (FES) and self-reported retrospective falls and circumstances (Berg et al., 1997) were examined by a physical therapist prior to the experimental trials. During the experimental trials, marker data of the 47 marker full body HBM model (Van den Bogert et al., 2013; Motekforce Link, Amsterdam) and force plate data were captured. Before the experiment and after each measured trial, participants were asked to rate their tiredness, pain and nervousness on a VAS scale ranging from zero to ten, with ten indicating unbearable tiredness or pain or and 0 lack of any of these. When tiredness or pain exceeding a score of 8, a break was performed. In addition, the participants were asked to rate the difficulty of the two perturbation trials. After 9.6 (SD 0.6) months, a follow-up was performed in which all patients were called to quantify the prospective number of falls after the measurements.

3.3.5. Data-analysis

Data analysis was performed as described in the healthy subject study in 2.3.4. Briefly, gait events were calculated based on kinematic data (Zeni et al., 2008). Outcome measures that were calculated included spatio-temporal parameters (Hak et al., 2012) (see paragraph 2.3.4.2), margins of stability (MOS) based on the extrapolated center of mass (equation 2.2 and 2.3); Full et al., 2005; Hak et al., 2012), gait sensitivity norm (GSN) and the GSN corrected for baseline variability(GSN_c) (equation 2.4 and 2.5) (Hobbelen et al., 2007) and the gait sensitivity norm per gait indicator (equation 2.6, 2.7, 2.8 and 2.9). The GSN gave a single value for the absolute response to each perturbation by incorporating the gait indicators in 4 steps following the perturbation. A lower GSN value indicated better disturbance rejection (Hobbelen et al., 2007).

The outcomes of the normal walking trial were used as a baseline by taking the mean over 40 steps for all parameters. The first 10 strides were ignored to provide some stabilizing time within the trail. The analysis was performed over the subsequent 40 steps.

Clinically tested fall risk

The categorisation of ’fallers’ and ’non-fallers’ was done using several methods; based on retrospective falls, the FES, the TUG and prospective falls. Using retrospective falls, a subject was identified as faller when one or more falls occurred during the last year (after the stroke incidence). A fall was defined as unintentional loss of balance with the result of lying on the ground which was not a result of a seizure, stroke or other occurrence. Furthermore, persons were categorised as faller; when exceeding 23 points on the FES (Delbaere et al., 2010), when exceeding the duration of 14 seconds in the TUG (Shumway-Cook et al., 2000) or when prospective falls occurred, retrieved from a follow-up in which patients self-reported the amount of falls after the measurement.

3.3.6. Statistical analysis

Group differences between stroke and healthy controls were tested using independent t-tests (IBM SPSS statistics v24, Armonk, NY) for fixed speed and self-paced walking. When the data was not normally distributed, a non-parametric Mann-Whitney U test was performed. The differences between fallers and non-fallers were compared within the stroke patient group using independent t-tests. Additionally, the relations between the outcome measures and the clinical fall risk measures; retrospective number of falls, FES, TUG and prospective falls were analysed using Pearson’s correlation or Superman’s correlation when data was not normally distributed.

3.3.7. Observational analysis

Observational analysis on step basis were performed as described in section 2.3.6.

3.4. Results

3.4.1. Clinimetrics

Stroke patients showed significantly higher mean scores on the FES ($p = <0.01$), the TUG ($p = <0.01$), tiredness DS ($p = <0.02$) and experienced difficulty ($p = <0.01$) (Table 3.2). Stroke patients walked significantly slower in both fixed speed and self-paced walking. Stroke patients reported more tiredness in the fixed speed condition. Furthermore, stroke patients rated the perturbation trial as significantly more difficult than healthy subjects.

3.4.2. Baseline walking

During baseline walking, significant differences were seen between stroke patients and healthy controls (Tabel 3.3). Stroke patients walked with shorter step length and larger step time during fixed speed and self-paced walking. At self-paced walking, stroke patients additionally increased step width and MOS.

Table 3.2: Group differences incuding fixed speed (DS) and self-paced (SP) walking trials

	Stroke		**Healthy controls**		**p-value**
	Mean	**SD**	**Mean**	**SD**	
SOMT	27.43	0.94	27.72	0.59	0.30
FES	24.79	5.73	17.27	1.94	**<0.01**
TUG	26.08	15.24	8.05	1.32	**<0.01**
Walking speed DS	0.36	0.19	0.68	0.17	**<0.01**
Walking speed SP	0.41	0.30	1.17	0.17	**<0.01**
Tiredness DS	3.32	1.59	1.73	1.84	**0.02**
Tiredness SP	3.96	1.99	2.59	2.08	0.06
Difficulty DS	4.00	1.70	1.10	1.00	**<0.01**
Difficulty SP	4.04	2.11	1.50	1.61	**<0.01**

Table 3.3: Baseline walking - Group difference stroke and healthy

	Stroke		**Healthy controls**		**p-value**
Normal walking	**Mean**	**SD**	**Mean**	**SD**	
Step length DS [m]	0.219	0.085	0.390	0.096	**<0.01**
Step width DS [m]	0.183	0.063	0.157	0.054	0.24
Step time DS [s]	0.680	0.151	0.564	0.060	**0.02**
MOS DS [m]	0.081	0.152	0.075	0.014	0.36
Step length SP [m]	0.250	0.121	0.539	0.147	**<0.01**
Step width SP [m]	0.180	0.066	0.147	0.040	**<0.01**
Step time SP [s]	0.739	0.217	0.539	0.041	**<0.01**
MOS SP [m]	0.079	0.016	0.074	0.011	**<0.01**

3.4.3. Stroke versus healthy

Fixed speed walking

Significantly higher perturbation responses were seen in stroke patients compared to healthy controls (Table 3.4 and Figure 3.1), based on the GSN and the GSN_c, containing all gait variables. No significant differences were shown for the GSN outcomes per variable (Table 3.4 and Figure 3.2).

Mean values per step after the perturbation can be observed in Figure 3.4 and 3.5. Healthy subjects increased the step length, followed by decrease in steps A2 and A3, while stroke patients only slightly increased the step length, followed by a larger decrease after the perturbation. Additionally, healthy subjects tended to decrease the step width at two steps following perturbation (A1, A2) followed by an increase (A3,A4) whereas stroke patients only increased their step width. The response in step time seems similar with a slight increase in A1 and A2 and a slight decrease in A3 and A4 and both the healthy and stroke patients decreased MOS at A1, followed by an increase in MOS. This increase was larger in stroke patients at A2.

Self-paced walking

During self-paced walking the GSN_c showed a significantly larger outcome for stroke patients compared to healthy controls (Table 3.5 and Figure 3.1). Although not significant, the GSN showed a similar trend. Additionally, stroke patients had a significantly decreased step width, increased step time and increased MOS response to perturbations compared to healthy subjects (Table 3.5 and Figure 3.3).

Healthy subjects decreased step width during A1, showed baseline values at A2 and overcompensated at A3, based on observational analysis (Figure 3.6). Stroke patients did not decrease the step width and only showed increases at A2 and A3 (Figure 3.7). In step time and MOS similar patterns were shown, only slightly larger in stroke patients.

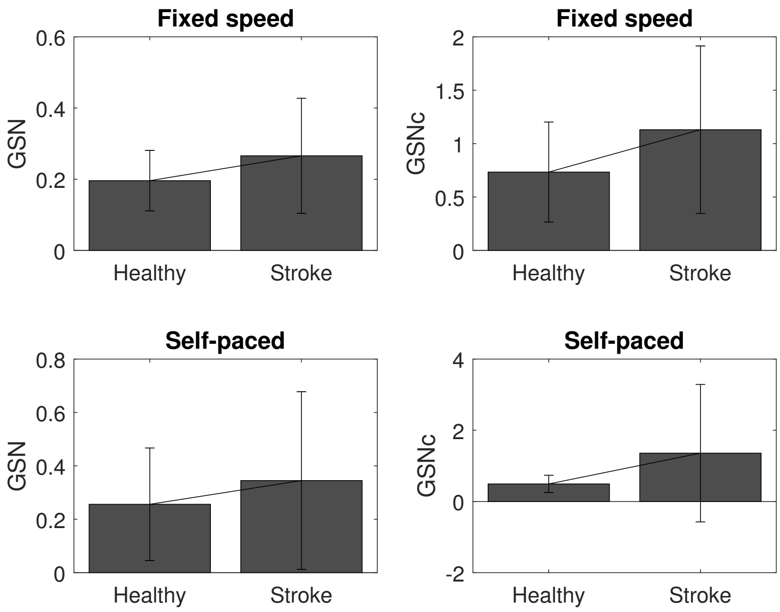

Figure 3.1: Group differences for GSN and GSN$_c$ at fixed and self-paced speed

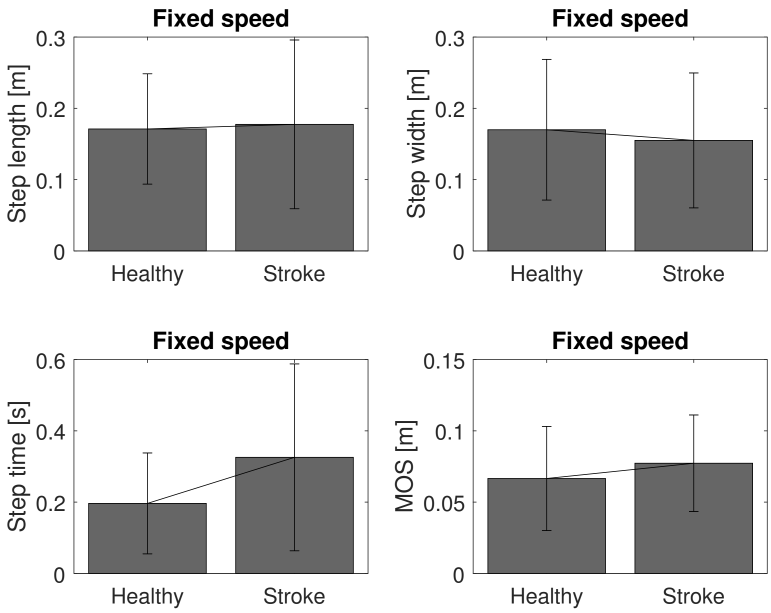

Figure 3.2: Group differences for GSN per variables step length, step width, step time and MOS at a fixed walking speed

3.4.4. Outcomes related to clinically tested falls

In Tabel 3.6 can be seen which stroke patients were identified as fallers based on the previously defined thresholds for retrospective falls, FES, TUG and prospective falls. In the healthy control group no subjects were identified as fall-prone subjects, using any of the above definitions. No significant differences in outcomes were found between the group of stroke patients that actually fell retrospectively or prospectively and the non-faller stroke patients (Table 3.7). Correlations were found between the GSN outcomes (corrected, uncorrected, fixed and self-paced speed) and the TUG score (Table 3.8). A positive correlation was shown

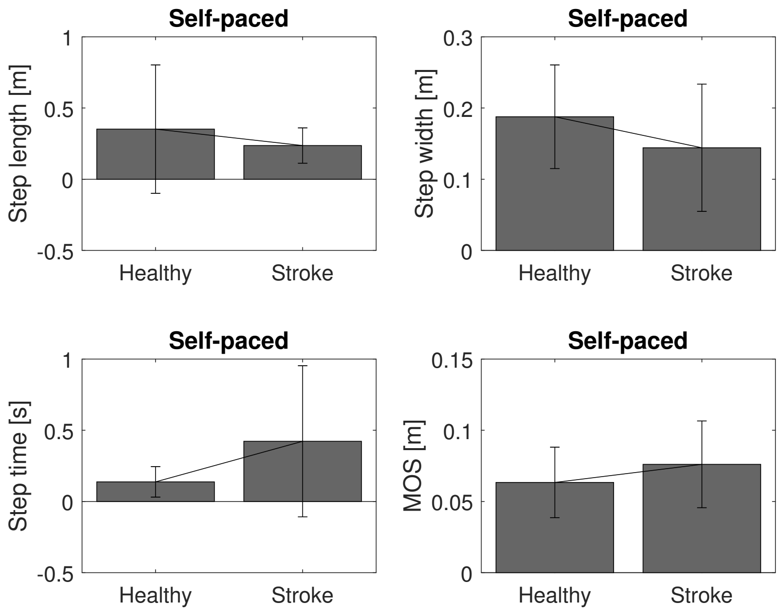

Figure 3.3: Group differences for GSN per variables step length, step width, step time and MOS at a self-paced walking speed

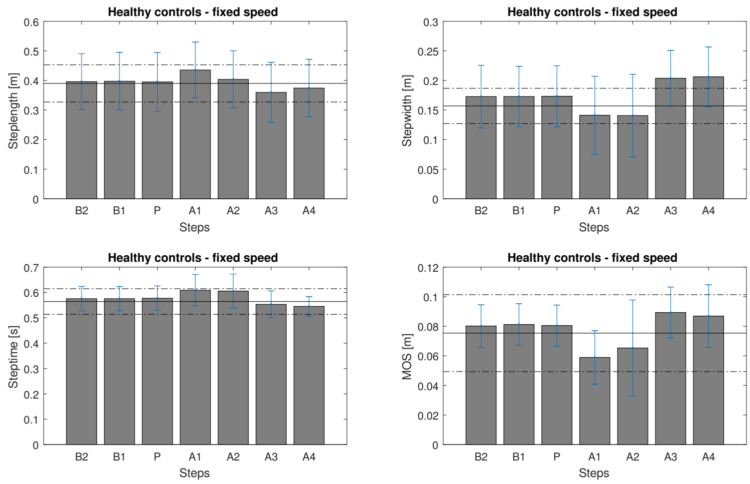

Figure 3.4: Mean steps healthy subjects fixed speed

Table 3.4: GSN, GSN$_c$ and GSN per variable outcomes in response to perturbations at fixed speed walking

	Stroke		**Healthy controls**		**Independent t-test**
	Mean	**SD**	**Mean**	**SD**	**P-value** $\alpha = 0.05$
Step length	0.178	0.118	0.171	0.077	0.79
Step width	0.155	0.095	0.170	0.099	0.52
Step time	0.326	0.262	0.196	0.142	0.09
MOS	0.077	0.034	0.067	0.037	0.12
GSN	0.266	0.162	0.196	0.085	**0.04**
GSNcorr	1.130	0.784	0.734	0.468	**0.04**

indicating a higher GSN outcome score was related to a longer TUG duration. Other faller identification methods did not show any significant correlations with the GSN outcomes.

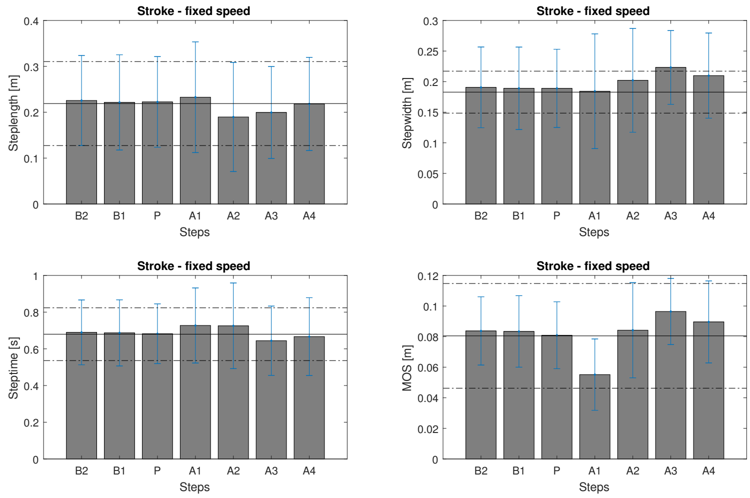

Figure 3.5: Mean steps stroke subjects fixed speed

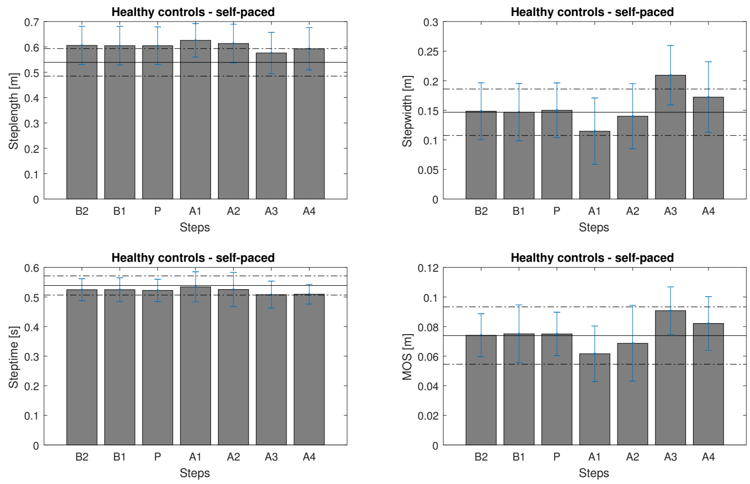

Figure 3.6: Mean steps healthy subjects self paced

Table 3.5: GSN, GSN_c and GSN per variable outcomes in response to perturbations at self-paced speed walking

	Stroke		**Healthy controls**		**Independent sample t-test**
	Mean	**SD**	**Mean**	**SD**	**P-value**
Step length	0.237	0.124	0.352	0.451	0.98
Step width	0.144	0.089	0.188	0.073	**0.03**
Step time	0.423	0.530	0.138	0.107	**0.02**
MOS	0.076	0.031	0.063	0.025	**0.03**
GSN	0.345	0.333	0.256	0.211	0.09
GSNcorr	1.357	1.930	0.493	0.242	**0.04**

3.5. Discussion

The aim of this study was to determine whether the ability to recover from external perturbations on the walking surface could discriminate patients from healthy controls and fallers from non-faller within stroke patients in fixed speed walking and self-paced speed. Additionally, it was aimed to reveal patient specific

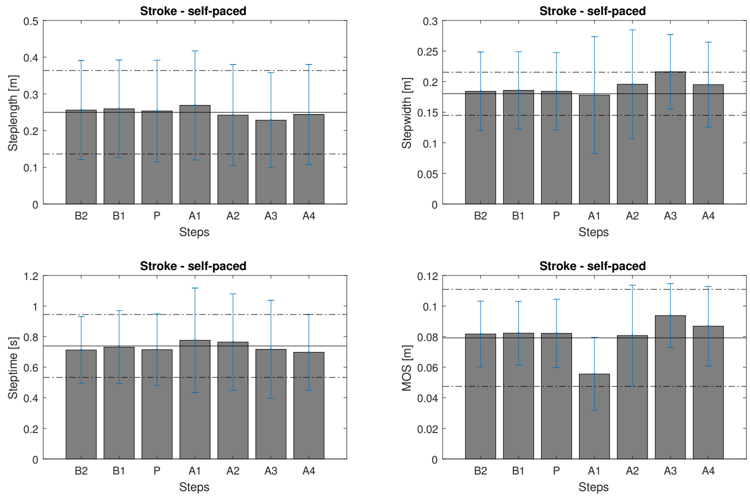

Figure 3.7: Mean steps stroke subjects self paced

Table 3.6: Faller categorizations based on different definitions

	Retrospective falls	**FES faller**	**TUG faller**	**Prospective falls**
Stroke 1	x	Faller	Faller	x
Stroke 2	x	x	Faller	Faller
Stroke 3	x	Faller	Faller	x
Stroke 4	x	Faller	Faller	x
Stroke 5	x	x	Faller	Faller
Stroke 6	Faller	Faller	x	Faller
Stroke 7	Faller	x	x	x
Stroke 8	x	Faller	x	x
Stroke 9	Faller	x	Faller	x
Stroke 10	Faller	Faller	Faller	Faller
Stroke 11	Faller	Faller	Faller	x
Stroke 12	x	Faller	Faller	x
Stroke 13	x	Faller	x	x
Stroke 14	x	x	x	x

responses based on observational step analysis and whether the response is related to clinically assessed fall risk. The walking stability of stroke patients and healthy controls was challenged by contra-lateral perturbations with an displacement of 0.05 m in 1.77 s during continuous treadmill walking.

The GSN and GSN$_c$ both resemble the absolute magnitude of the response over four steps to the perturbations. As expected, stroke patients showed a larger response with respect to their own baseline compared to healthy subjects. By the definition of Hobbelen et al. (2007), a larger GSN outcome represents a larger dynamic response to the perturbation of a system due to larger variability. Therefore, a lower values corresponds to better disturbance rejection. In previous studies performed during steady state walking, a relationship was shown between increased variability and prospective falls (Hausdorff et al., 2001; Maki et al., 1997). Regarding this, the higher GSN outcome for stroke patients would make sense as more variability was shown in disturbance rejection, indicating a higher fall risk and less dynamic walking stability. This was even the case when compensating for baseline variability in both groups using the GSN$_c$. However, previous studies showed that adequately responding to perturbations is very important in order to maintain stability or not to fall (Bruijn et al., 2013; Young & Dingwell, 2012; Li et al., 2005; Hak et al., 2012). In these studies it was claimed that certain responses would reduce the likeliness of falling. Therefore an increased variability represented as adjustments would make a subject better capable of rejecting perturbations. In this light, both healthy and stroke patients used adjustments to overcome the perturbation. The GSN outcomes however showed

Table 3.7: Differences in stroke patients between fallers (prospective + retrospective) and non-fallers

	Fixed speed p-value	**Self-paced** p-value
Step length	0.75	0.81
Step width	0.52	0.33
Step time	0.23	0.22
MOS	0.54	0.70
GSN	0.22	0.18
GSN_{corr}	0.38	0.17

Table 3.8: Correlations

	GSN_DS	GSNcorr_DS	GSN_SP	GSNcorr_SP	MOS_DS	MOS_SP
Restrospective falls	0.35	0.72	0.57	0.86	0.43	0.37
FES	0.94	0.33	0.41	0.64	0.09	0.61
TUG	**0.03**	**<0.01**	**<0.01**	**<0.01**	0.40	0.12
Prospective falls	0.33	0.37	0.29	0.34	0.59	<0.01

that stroke patients used more walking adjustments to overcome the perturbations. Therefore, the GSN outcomes indicated that healthy subjects were capable of returning to baseline faster or needed less adjustments to cope with the external perturbations.

The GSN and GSN_c are able to quantify the response to the perturbation in one comprising outcome indicating whether the population is performing better or worse. In addition, it has several advantages with respect to the power in statistic analysis, group comparisons and calculation time (Hobbelen et al., 2008). The individual contribution of each of the four gait indicators was however disregarded in the outcome. In fixed speed walking, only significant differences were shown in the GSN and GSN_c values and not in the GSN per variable. This indicates that the GSN and GSN_c contained a more distinctive character due to the accumulation of the different gait indicator responses than in the gait indicators solely.

This distinctive character was however not seen in self-paced walking, where significant differences were seen for GSN_c and GSN_{SW}, GSN_{ST} and GSN_{MOS}. Based on the GSN per variable, the stroke patients changed their step width significantly less than healthy controls. However, they showed more response in step time and MOS. Although not significant, in fixed speed walking similar trends were seen for step width, step time and MOS. This could indicate that comparable responses occurred during fixed speed as self-paced walking, whilst being more profound during self-paced walking. Multiple reasons can underlie this difference. First of all, in fixed speed walking the patient is not able to influence the belt speed. Therefore, a walking rhythm could be externally imposed, which might inhibit flexibility and fluctuations in the walking pattern (Sloot et al., 2014). In self-paced walking the patient is able to influence the belt speed based on the anterio-posterior position on the belt, resulting in long term stride fluctuations resembling over ground walking (Sloot et al., 2014). When a subject is walking more in the front, the belt will accelerate and in the back it will decelerate. This mimics overground walking more closely, as a person is also able to change his walking speed required by the environment (Sloot et al., 2014; Plotnik et al., 2015). Walking speed is another factor which might affect the outcomes. In fixed speed walking, the healthy subjects were asked to walk slow at a dimensionless speed of 0.25 to observe their behaviour more closely to stroke subjects. During self-paced walking, the subjects were instructed to find their comfortable pace for normal walking. Although the walking speeds between stroke patients and healthy subjects differed significantly in both fixed speed and self-paced, the mean difference in walking speed was larger for self-paced walking. This might have an influence on the outcomes, as the healthy subject has a shorter time range to respond to perturbations and more forward propulsion.

The GSN_c considers the standard deviation in order to eliminate the effects of fluctuations due to baseline variability. However, the findings of the GSN and GSN_c are comparable. This indicates that the group differences in response to the perturbation were likely the result of the perturbation rather than differences in steady state variability. For a further discussion about GSN_c chapter 2.5 can be consulted.

The observational analysis showed a roughly similar response pattern in the stroke subjects group and the healthy controls. However, healthy subjects decreased their step width in the first step following perturbation, resulting in a concurrent decrease in MOS. This is in accordance with Hof et al. (2007), who showed a relation between an increased step width and an increased medio-lateral MOS. Stroke patients also showed a decrease in MOS, interestingly however they did not decrease their step width. The decrease in MOS without the decrease in step width indicates that the XCOM reached a closer distance to the border of the base of support (Hof et al., 2005). This could be explained by an increased medio-lateral body sway, which was also seen in stroke patients in previous studies (Hak et al., 2013; Chen et al., 2005). During the second step, stroke patients increased the MOS compared to the first step, while healthy subjects increased MOS less. Since the first and second steps contained one left and one right step, this might indicate that the XCOM of the healthy subject traveled further to the base of support of the next step while in stroke patients XCOM remained close to the base of support of the previous step. Although stroke patients already walked with increased step width during baseline walking, in response to the perturbation they did not to reduce the step width like healthy subjects did. Healthy subjects also showed this reduced step width response in the pilot study (2.5). Accordingly, in the study of Hak et al. (2013) stroke patients walked with larger step width during normal walking. However, they were able to increase step width and MOS a similar amount as healthy controls in response to continuous perturbations. As the instantaneous contra-lateral perturbation evoked a smaller step width, it appeared that the stroke patient group was not able to decrease the step width. Possibly because that might further decrease their MOS. It is likely that stroke patients used a larger step width to compensate for other deficiencies (Hak et al., 2013) or as a compensation for the larger medio-lateral body sway (Chen et al., 2005). Accordingly, stroke patients did also not reduce step width following contra-lateral sway perturbations in a recent study of Punt et al. (2017). In contrast with our study, stroke patients also not show a reduced MOS following the perturbation. This could be related to less medio-lateral upper body sway.

Healthy subjects walked with a larger step length during baseline walking compared to stroke patients, which was in accordance with Hak et al. (2013). In response to the perturbation, this difference remained and the groups did not respond differently, although healthy controls increased step length slightly further than stroke patients. It might be possible that stroke patients are unable to further increase their step length due to lack of propulsion as a result of muscle disfunction (i.e. spasticity) (Balasubramanian et al., 2007; Roerdink Beek, 2011) or by trying to maintain backwards margin of stability (Espy et al., 2010). While stroke patients already walked with increased step time during baseline walking compared to healthy subjects, stroke patients further increased step time in response to the perturbation. This might be a result of the increased difficulty that stroke patients experienced based on their self reported difficulty from 1-10. When difficulty or demands are increased, walking patterns are less automated and more concentration and cognitive function is required (Lajoie et al., 1993). When more cognitive function is needed, increasing the step time might give more time to position the steps following the perturbation. Punt et al. (2017) also showed increased step length and step time in response to the contra-lateral perturbation.

When considering the GSN value as efficiency of perturbation response, a person responded adequately when his response was large enough to prevent from falling, but as efficient as possible. With a lower value, healthy subjects showed a more efficient response. Stroke patients did not decrease the step width as healthy subjects did and therefore their compensation in the other gait indicators resulted in a larger, less efficient response. This is in line with the findings of Hak et al. (2014), where was stated that adaptability is required to remain balanced. A person who is more 'adaptable' might use the most efficient way to respond to the perturbation and afterwards return faster to baseline level. For this reason, training programs incorporating perturbations could be very effective and train the subjects response to perturbations with the aim to respond as efficient and adequate as possible. For this reason, it is advised to provoke responses to perturbations in order to generate a more appropriate response and therefore 'overcome' the perturbations faster and go earlier back to 'limit cycle walking'.

The four clinical fall risk assessments showed inconsistent categorizations of fallers and non-fallers. This emphasizes the necessity for a specific and all comprising measure. Although these four measures showed internal validity (Dewan MacDermid, 2014; Podsiadlo Richardson, 1991; Bruijn et al., 2014), with these inconsistencies it is not clear which measure contained predictive validity. Therefore, the identification of fall risk based on these clinically fall risk assessments is poor. This could be an explanation for the lack in significant differences between the faller and non-faller groups based on retrospective and prospective falls. It could however also be a result of a the limited sample size. Punt et al. (2017) also did not find differences in perturbation response between stroke patients with and without prospective falls. This could also be explained by the fact that the definition of the faller group based on a fall incidence might not be a good discriminator.

It is difficult to draw conclusions based on GSN outcomes and clinically assessed fall risk. Retrospective falls described the fall history, however, the likelihood of falling reduced when persons are inactive, although the fall risk might be equally high. This could also apply to prospective falls. In addition, as the incidence of falling is instantaneous, a fall prone subject could fall right outside of the perceived time scope of the study. Furthermore, the falls are subjectively reported. That could decrease the reliability as subjects tend to only remember very serious falls (Hauer et al., 2006). The FES is a measure in which the patients subjectively rate their fear of falling during different tasks. This is also subjective as people rate their own experience differently. Previously, it was shown that there might exist a relation between fear of falling and fall risk (Delbaere et al., 2010), however, this might not be the case for each individual. Some persons experience more fear than is necessary while others might overestimate their abilities. The last clinical measure that was used was the TUG. The TUG was a functional measure, so in contrast to the other subjective measures it was reported objectively. The TUG measured the time duration in which patients could stand up from a chair and walk 3 meters forward and back. Therefore, this measure might not have the closest relationship to actual falling, but it does represent some functional activity level. That could be the reason that the GSN and GSN_c outcomes showed significant correlations with the TUG. This indicates that the subjects with reduced functional ability and therefore a longer TUG time showed higher response to perturbations in GSN and GSN_c outcomes. In order to discriminate fallers from non-fallers in stroke patients, perturbation based gait assessment might be more sensitive in quantifying the individual differences in walking stability compared to subjective clinical fall assessments and hence might provide for a better fall risk assessment.

For further studies, it is recommended to follow subjects over an elongated prospective period of time, so that the results could be analysed with respect to actual falls over a long period of time. In the development of a new study, it would even be better to utilise a fall-tracking device to eliminate the dependency of the participants subjective reporting. Also, a larger group of stroke patients could be incorporated to study differences and distinctions in walking stability within one patient group. Additionally, this approach could be repeated using perturbations of different intensities, of different kinds and even with more gait indicators. Furthermore, training programs can be developed and it could be studied whether patients are capable of increasing the efficiency in the response and reducing the magnitude of the outcome, while still responding appropriately in order not to fall.

3.6. Conclusion

Overall, stroke patients showed a larger response in the four steps following the perturbation based on the four gait variables; step length, step width, step time and MOS. In fixed speed walking, GSN was larger in stroke compared to healthy controls ($p = 0.04$). During self-paced walking, a significant larger step length and step time response was found in stroke patients. They however showed a decreased step width response. Although the GSN does not specifically indicate which gait indicator showed an enlarged response and how stroke patients and healthy persons reacted differently on a step basis, it does give a discriminative overall response between stroke patients and healthy subjects. Therefore, it might be an effective way to quantify the response to perturbations in one easily interpretable measure, indicating the magnitude to the response. No differences were found between stroke patients with and without fall history, which might be due to incorrect identification of fall risk based on current clinical fall risk assessments. In response to contra-lateral perturbations, MOS seemed similarly affected in both the stroke and healthy control groups, while healthy controls reduced step width and stroke patients did not. To compensate for this, stroke patients did increase the step length and step time. As stroke patients compensate for deficits in functionality, the higher GSN outcomes might indicate a less efficient way to cope with the perturbation. Therefore, it is advised to develop training programs in order to learn the efficient strategies healthy control use to cope with perturbations and which will enhance the dynamic walking stability.

4

Conclusion

In conclusion, both studies showed a similar behaviour following the contra-lateral perturbations in healthy subjects. As stroke patients showed a significantly larger response of the GSN compared to healthy subjects, this measure might be useful to discriminate between the groups. When more knowledge is desired on the exact behaviour following perturbations, the response on step basis can be consulted. As clinically assessed fall risk is very arbitrary and inconsistent, it is not dissolving the paradox by relating the GSN outcome to clinically assessed fall risk. The GSN measure could be used to further evaluate the response to perturbations in order to quantify walking stability. For these subjects, trainings could be developed in which is learned to overcome perturbations.

5 Bibliography

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A

Appendix: Pilot study

A.1. Chosen perturbations

Four different perturbations were analysed. These were combinations of two different intensities and two different platform directions. The two intensities were chosen based on trials and previous published articles (McAndrew et al., 2010; Hak et al., 2012). These studies used similar hardware and applied a continuous multisine signal which resulted in a platform movement with a maximum displacement of respectively 0.6 m, a maximum velocity of 1.13 m/s and a maximum acceleration of 1.78 m/s^2. These perturbations were not instantaneous, however, it gave an impression of the possible range. In order to perturb the subject during walking, an instantaneous platform movement was required during one step. For this reason, different perturbation intensities were tested on a healthy subject accompanied by a physiotherapist to find perturbations shorter than one step, big enough to give a noticeable response but still appropriate for patient groups. Based on these experiments, two perturbation intensities were chosen to compare the effects on several subjects more objectively. The first perturbation was controlled with a reference displacement of 0.05 m and the second perturbation with a reference displacement of 0.035 m using a customised control application written in programming language Lua within D-Flow software (Figure C.4; Motek, Amsterdam).

The two types that were tested were different platform directions; the contra-lateral and ipsi-lateral direction. A contra-lateral perturbation represents a platform movement in the opposite direction of the stance foot. For instance; the platform moved to the right during a left foot stance. For the ipsi-lateral condition, the platform moved to the same side as the stance foot, meaning a left platform movement during the left stance phase. Although only distinctions were made between contra- and ipsi-lateral perturbations and no distinction was made in the analysis between left and right perturbations, the perturbations sides were randomised to eliminate prior anticipation to a certain side. The aim was to conceive the largest effect of the perturbation during stance phase. Therefore, the perturbations were triggered at initial foot contact.

A.2. Number of perturbations

Previous studies varied in the numbers of repeated perturbations. As an impression of studies using a walkway; Ferber et al. (2003) used 12 repeated perturbations, Yamaguchi et al. (2016) repeated 5 slip perturbations, Schillings et al. (1996) executed 5 stumbling perturbations, Yang et al. (2011) repeated 8 slip perturbations, Quintern et al. (1985) used 35 perturbations of different intensities and repeated this protocol on 6 different days, Bhatt et al. (2012) performed three times perturbation sessions, the first two consisted of 8 perturbations of a similar types and later on 15 perturbations of different types.

To ensure a number of perturbations that is large enough to represent a generalised perturbation response, but not so large that it results in fatigue effects, the number of repeated perturbations was set to 10 perturbations. This means that the subjects underwent 10 perturbations of each kind. Since there were 4 different perturbation types tested, subjects received 40 perturbations in total.

A.3. Gait events

Initial contact (IC) and toe-off (TO) were calculated. A golden standard method to determine gait events was based on force plate data. However, by incorporating platform movements, a kinematic method to determine gait events is favorable as the force plate data is affected by inertial forces. Furthermore, the determination of gait events based on kinematic data was shown to be more accurate and easier to implement compared to force plate algorithms (O'Connor et al., 2007). For these reasons, it was chosen to implement the kinematic method for determining gait events following the method of Zeni et al. (2008) (see paragraph A.3). The time instance of initial contact (t_{IC}) or toe off (t_{TO}) were found by the maximum distance between the sacral coordinates (Z_s) and the heel coordinates (Z_h) or the minimal distance between the sacral coordinates (Z_s) and the toe coordinates (Z_t) in the longitudinal direction of the treadmill were calculated.

A.4. Platform trajectory

A.4.1. Methods

The intensity of the two different perturbations was characterised by a medio-lateral sway displacement of 0.05 m and 0.035 m. The order of the displacement side was randomised. After the platform displacement, the platform did not move back to the original position but remained in the end position and utilised that as the new starting position.

For platform movement analysis, the delay between the trigger and the actual platform movement was calculated. The trigger was a represented as a binary value created by the control software 'D-Flow'. The actual onset of the platform movement was defined when a difference in platform position was detected which was larger than the minimal noticeable difference, which was set to 0.5 mm. This threshold was chosen as it represents the measure accuracy of translations using the Vicon motion capture system (http://www.vicon.com/products/camera-systems/bonita). Accordingly, Eichelberger et al. (2016) showed the measured trueness, represented by the mean of the marker distance error, during subject experiments was within the range [-0.38, 0.38]mm, which is below 0.5 mm. The duration of the platform movement was calculated using the time between onset and end of the platform movement, in which these time instances were also defined using the same the minimal noticeable difference.

A.4.2. Results

Platform displacement course

An example of the trajectory of the platform with an intensity of 0.05 m can be seen in Figure A.1 and A.2. These trajectories are different for each subject.

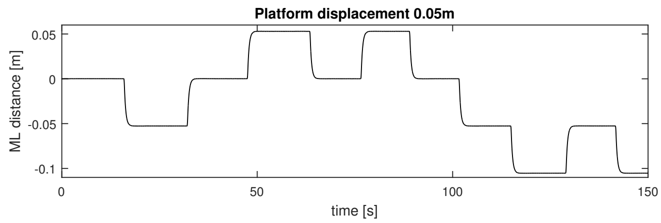

Figure A.1: Example of platform displacement trajectory of perturbation intensity 0.05m

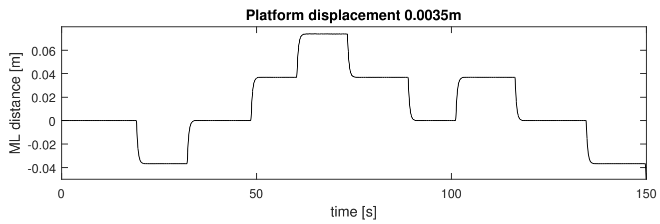

Figure A.2: Example of platform displacement trajectory of perturbation intensity 0.035m

The perturbation type with a displacement of 0.05 m, had a maximum velocity of 0.11 m/s and a maximum acceleration of 0.92 m/s^2. The second perturbation had a displacement of 0.035 m with a maximum velocity of 0.08 m/s and a maximum acceleration of 0.71 m. These profiles can be seen in Figure A.3 and A.4. The velocities and accelerations are lower than in previous studies. Since they need to be applicable to frail persons and patient groups and are applied instantaneously at a random moment in time, the peak velocities and accelerations can not be too high. By observing the figures A.3 and A.4, the highest velocities and accelerations are reached during the first half of the displacement.

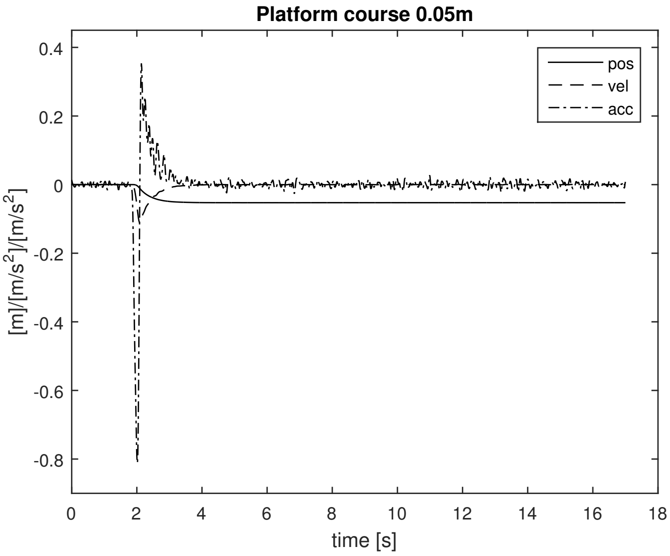

Figure A.3: Platform displacement, velocity and acceleration during one perturbation of an intensity 0.05m

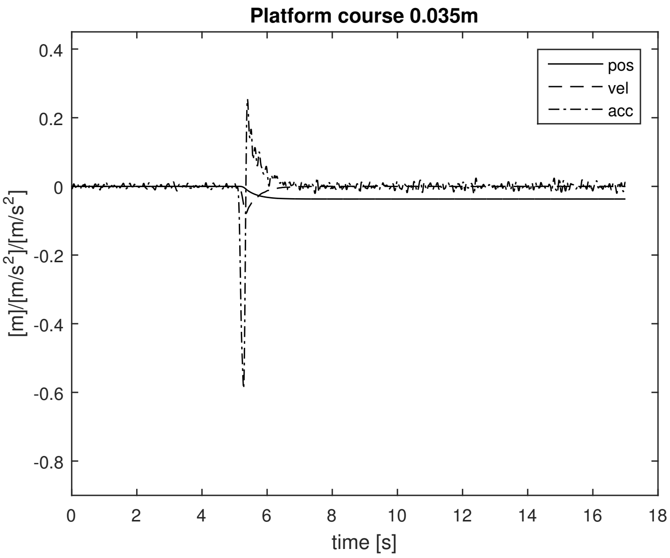

Figure A.4: Platform displacement, velocity and acceleration during one perturbation of an intensity 0.035m

Platform delay

In Figure A.5 and A.6 are examples displayed of the trigger (circle) against the platform movement with different scales (solid line). The mean and standard deviations of the delay between the trigger and the movement onset and the duration of the platform movement are presented in Table A.1.

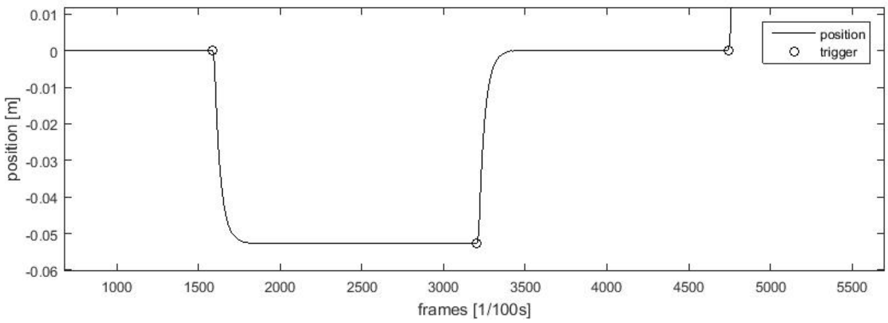

Figure A.5: The platform displacement of 0.05 plotted against the trigger signal

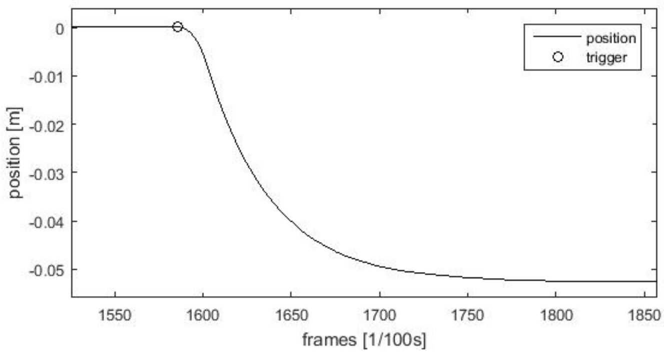

Figure A.6: A closer view of the trigger signal against the platform displacement

Table A.1: Delay and duration of platform movement

	Mean delay (s)	**SD delay (s)**	**Mean duration (s)**	**SD duration (s)**
Perturbation 0.05m	0.078	0.017	1.768	0.067
Perturbation 0.035m	0.083	0.018	1.618	0.037

A.4.3. Discussion

Based on the platform analysis, the platform delay appeared to be below 0.1 seconds and the duration of the platform movement was below 2 seconds. These values seem quite reasonable for a real-time controlled system. More importantly is how the delay and displacement were related to the gait pattern as it was desirable to perturb the subject during one gait cycle. The intensity of the perturbation were highest during the first half of the perturbation, as the velocities and acceleration were highest. The first half of the displacement occurred completely during a single stance phase for both conditions. Therefore, these perturbations seem appropriate. However, the walking speed and the step frequency might influence the gait cycle percentages. When the walking speed or the step frequency will be increased, the duration of the stance time might decrease. To ensure the complete perturbation during one gait cycle for a decreased walking speed or an increased frequency, a shorter perturbation duration could be considered. However, in a shorter duration, the platform could cover less distance resulting in a smaller intensity or the acceleration is higher in which the impact of the perturbation might not be suitable anymore for patient groups. Although the walking speed selected in this study was a low walking speed, it is likely that a patient groups would even walk on a lower walking speed. Therefore, the delay and duration of the perturbations used in this study seem applicable for further studies.

A.5. Gait cycle

A.5.1. Methods

In order to analyse if the perturbation was applied within the duration of one step and during which events of the gait exactly, the percentages of the gait cycle were calculated during the onset, middle and end of the platform movement. A full gait cycle was calculated as 100% from initial contact to the next initial contact. The onset, middle and the end of the perturbation were defined with the same minimum noticeable difference set to 0.5 mm.

A.5.2. Results

The perturbation is characterised by the platform movement. Since this is not an instantaneous event but rather a course in time, it is relevant to analyse the platform movement with respect to the gait cycle. The platform movement was divided into 3 stages for each perturbation; the beginning, the middle and the end of the platform movement. The corresponding percentages of the gait cycle were calculated and are represented in table A.2.

When comparing the percentages from table A.2 with the gait cycle percentages of normal human walking in figure A.7, the gait cycle phases during the three stages of the perturbation can be observed. For both perturbation intensities the beginning of the perturbation occurred directly after loading response, in mid stance. The middle of the perturbation occurred for the high intensity perturbation around 45% and for the low intensity perturbation around 38% of the gait cycle. In figure A.3 and A.4 can be seen that the largest velocities and accelerations of the platform occur during the first half of the platform course. For this reason the effect of the perturbation can be conceived as largest during the first half of the displacement. The results indicated that the largest effects of the perturbations occurred right after loading response till terminal stance. At the end of the platform movement, the percentages of the gait cycles of the two intensities were both within the same gait cycle.

Table A.2: Percentage of gait cycle at begin, mid and end of the perturbation

	begin % gait cycle mean	**begin % gait cycle std**	**mid % gait cycle mean**	**mid % gait cycle std**	**end % gait cycle mean**	**end % gait cycle std**
Pert 0.05m	11.26	0.88	45.10	3.19	78.94	5.52
Pert 0.035m	11.14	0.58	38.19	1.65	65.23	2.78

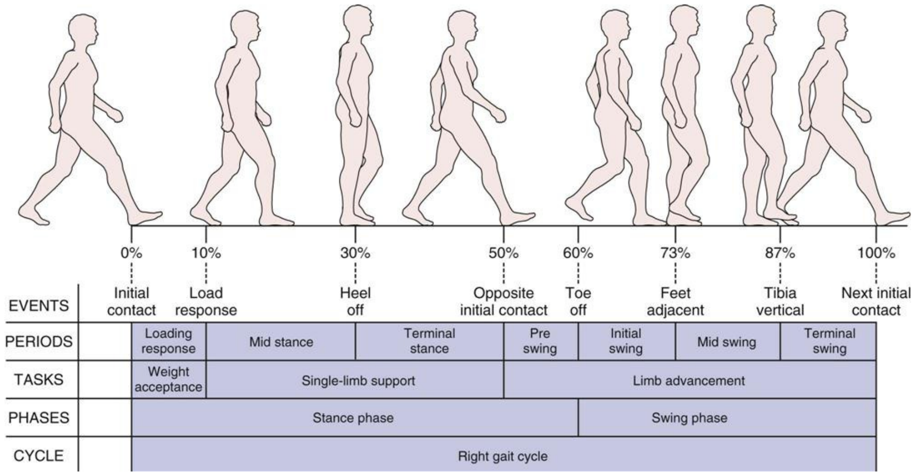

Figure A.7: The percentages of a gait cycle for normal walking (Neumann, D., 2013).

A.6. Number of steps after perturbation

A.6.1. Methods

In order to choose the number of steps after the perturbation that need to be incorporated in the data processing, an observational analysis was made. Not many articles have been published considering this matter or are not directly translatable to this study. When consulting the thesis of Hobbelen et al. (2008) an amount of 1-20 steps was recommended for physical robotic prototypes. During data collection, the amount of strides in between perturbations was set to a random number between 10 and 15 strides. Since the effects of the perturbations were expected to be largest in the several steps directly following the perturbation, a chance of averaging exists when incorporating too many steps. Another possibility when analysing too many steps is that proactive adjustment prior to the next perturbation will be considered as reactive response of the previous step. However, when not enough steps are taken into consideration, there exists a chance that a part of the effects of the perturbation are neglected. Therefore, it is relevant to find the optimum amount of perturbations. To this extend, an observational analysis was performed comprising the individual perturbations and mean perturbations of six step after the perturbation compared with the normal gait and the confidence interval.

A.6.2. Results

In order to decide for the amount of steps to incorporate in the analysis, an observational analysis was performed using the mean perturbations per step and the individual perturbations per step. In figure A.8, the mean step perturbations and the mean confidence intervals (red) are presented. All four perturbation intensities were presented side by side, with the darker colors representing the contra-lateral perturbations and the lighter colors representing the ipsi-lateral perturbations. The largest response to the perturbations was visible in the step width, where the outcomes for three out of four perturbation types are outside the confidence interval at A1 and A2. Additionally, a large response was seen in margins of stability at A1 and A2. Step length and step time are less affected by the perturbation, although some small adjustments were seen from A1 till A4. Furthermore, deviating responses are visible in A3 and A4 compared to B2 and B1. In step width A4, the contra 0.05m perturbation almost exceeds the confidence interval.

The individual perturbations per step were displayed in the figures A.9, A.10, A.11 and A.12. In figure A.9 are the contra-lateral perturbations displayed with two intensities and in figure A.10 are the ipsi-lateral perturbations displayed with two intensities. In figure A.11 and A.12 were the two sides of the perturbations displayed.The step length was mostly affected at A1 and A2, slightly at A3 and A4 and unaffected at A5 and A6. The step time is mostly affected at A1 and A2 and in figure A.10 also at A3 and A4. The response in step width is mostly presented at A1 and A2 for all figures and in figure A.11 and A.12 also at A3 and A4. The margins of stability show the largest response at A1 and A2. At A5 and A6, the values seem comparable to B2 and B1 for all outcomes.

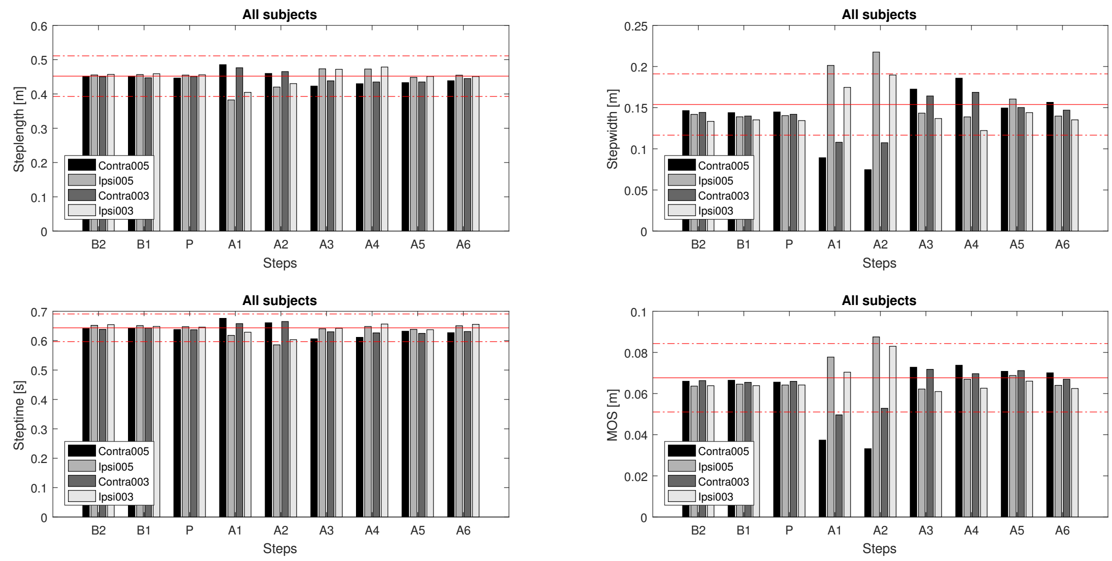

Figure A.8: All subjects, the mean of each step of all perturbations presented in one bar. P is the physical perturbation and A1 untill A6 are the 6 steps afterwards. Black and dark grey are the contra-lateral perturbations with 0.05m and 0.035m intensity respectively. Light grey and white are the ipsi-lateral perturbations with 0.05m and 0.035m intensity respectively. The confidence interval of normal walking is presented in red as reference.

In order to choose the optimum amount of steps after the perturbation for further processing, on the one hand enough steps should be included to capture the complete response and on the other hand not too many steps should be included that average the outcome. Based on these observations, an optimal amount of steps following the perturbation was considered 4.

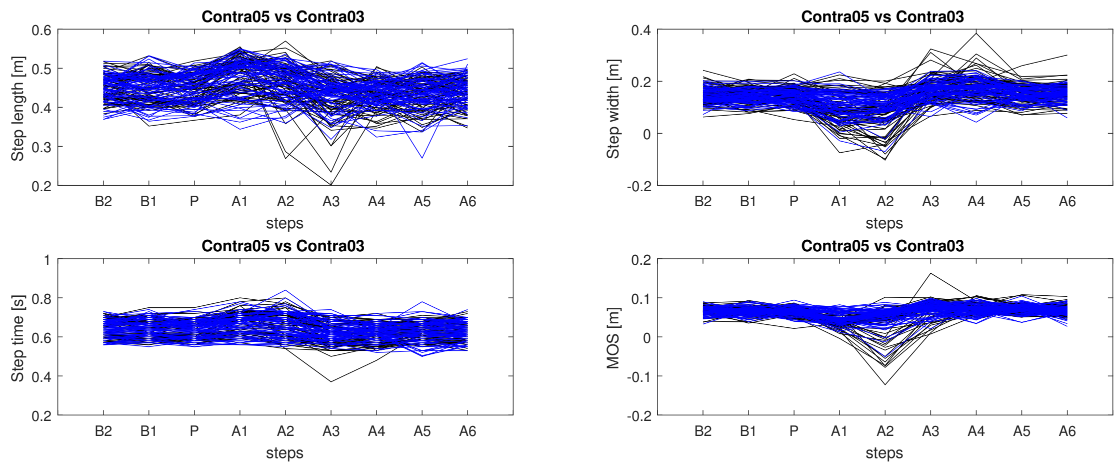

Figure A.9: The step responses of two contra-lateral platform perturbation intensities expressed in step length, step width, step time and MOS. P is the physical perturbation, B are the step before the perturbation and A are the step after the perturbation. Black = 0.05 m, Blue = 0.035m

A.6.3. Discussion

The optimum number of steps that was selected in the study following a perturbation was 4. The largest effects were seen in all gait indicators at the first and second step. However, to ensure that also compensation following the first and second step are observed, the optimum amount of steps was set to 4. In the steps 3 and 4, more deviations from the mean were seen compared to baseline walking prior to the perturbation. The amount of 4 steps was within the amount of steps that were recommended for the GSN in physical prototypes

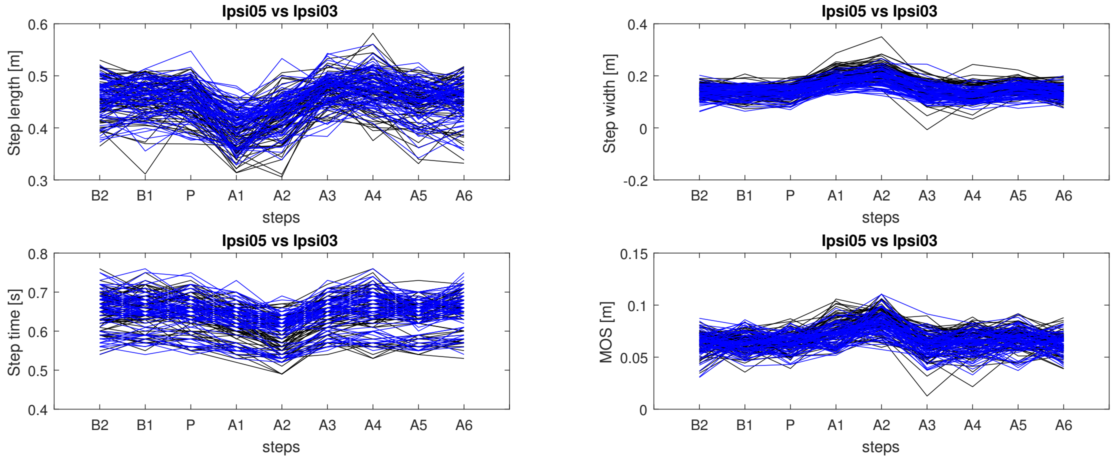

Figure A.10: The step responses of two ipsi-lateral platform perturbation intensities expressed in step length, step width, step time and MOS. P is the physical perturbation, B are the step before the perturbation and A are the step after the perturbation. Black = 0.05m, Blue = 0.035m

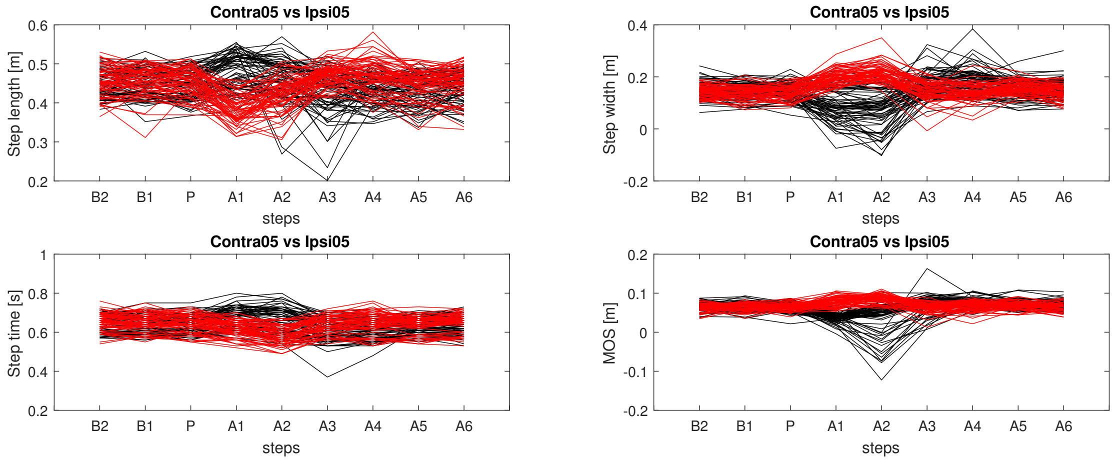

Figure A.11: The step responses of two platform perturbation directions with 0.05m intensity expressed in step length, step width, step time and MOS. P is the physical perturbation, B are the step before the perturbation and A are the step after the perturbation. Black = Contra, Red = Ipsi

of robots by Hobbelen et al. (2008). However, they recommended a number of steps between 1 and 20, which is a wide range. When involving more steps, the effects could be averaged and might therefore appear less profound. By including fewer steps, the possibility exists that some important aspects of the response to the perturbation were neglected.

A.7. Incorporation first perturbation of trial

A.7.1. Methods

In order to analyse the response to the perturbations, the magnitude and the variability were quantified. The variability over successive perturbations was used to quantify repeatability. This is of interest since a learning or adaptive effect might be present while repeating the same perturbation type. Although a learning effect might be beneficial for training purposes, when aiming to quantify dynamic walking stability, consistent outcomes and minor fluctuations between the responses of repeated perturbations are rather desirable. Concerning this matter, a previous study suggested to ignore the first perturbation as this gave a significantly different outcome compared to the following steps (Bierbaum et al., 2010). When considering the repeatability of the perturbations, two issues were addressed. Firstly, it was considered whether neglecting the first

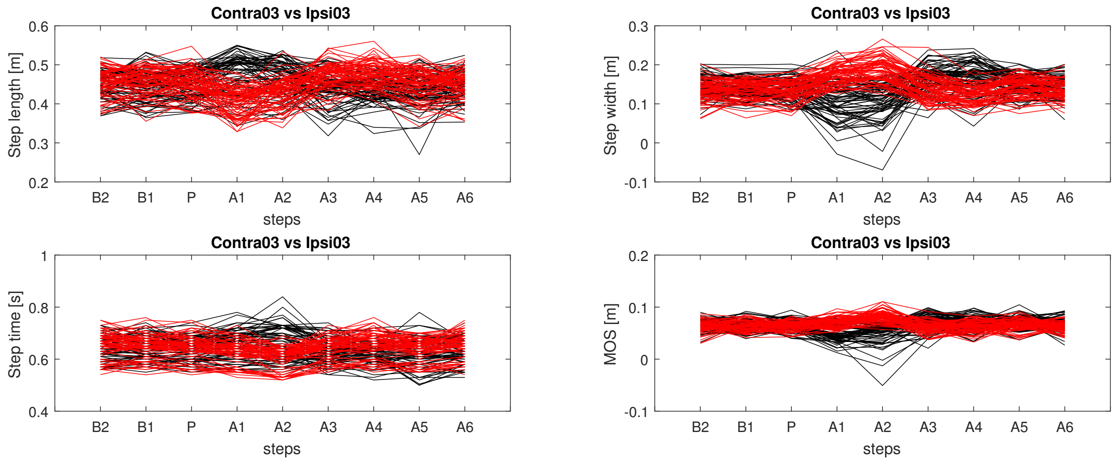

Figure A.12: The step responses of two platform perturbation directions with 0.03m intensity expressed in step length, step width, step time and MOS. P is the physical perturbation, B are the step before the perturbation and A are the step after the perturbation. Black = Contra, Red = Ipsi

or first several perturbations of a trial would eliminate the largest learning effect and decrease the variance hence increasing the repeatability. Furthermore, it was analysed whether there existed a difference in the variance between the contra-lateral and ipsi-lateral perturbations. The standard deviation (σ) was calculated over five perturbations. When the first or first several perturbations would comprise some outliers in the response, this would increase the standard deviation. Therefore, the sum of the standard deviation of four steps after perturbation were calculated for 5 successive perturbations. The response of the first till fifth perturbation was compared to the response of the second till sixth, third till seventh and fourth till eighth perturbation. When a large adaptive or learning effect exists for the first few perturbations, a large standard deviation is expected over the first few perturbations. When the variability between perturbation responses would decrease and the data points would be closer together, the standard deviation would decrease. A repeated measures Anova was performed with two within factors. The first factor was 'repeats' indicating the different perturbations and the other factor was the 'side' consisting of 'contra-lateral' or 'ipsi-lateral'. For a visual impression, the outcomes of individual perturbations per subject were plotted with the first three lines coloured differently to observe whether the the subjects response was different in the first three perturbation compared to the rest.

A.7.2. Results

In figure A.13 and A.14 are the standard deviations presented for 5 subsequent perturbations. When the first, or first few perturbations would deviate from the following perturbations, a decline in the slope would be expected for the standard deviation in the figures. However, more or less horizontal lines with small varieties in inclines or declines were observed. No systematic inclines or declines were observed for both intensities and no remarkable differences were seen between contra (black) and ipsi (red) lateral perturbations. This indicates that the first of first several perturbations did not diverge from the following outcomes.

When consulting the figures A.15 and A.16, examples are given of the individual perturbations of two subjects, one following a contra-lateral perturbation of 0.05m and one following an ipsi-lateral perturbation of 0.035m. The sequential first three perturbations were colored differently. The pattern following the first perturbation (red) was not specifically deviating from the other perturbations. In some cases, the second (blue) or the third (green) perturbation showed a more aberrant outcome. No consistent deviations were seen in the first or first perturbations for both intensities and sides. Other individuals showed similar inconsistencies.

A.7.3. Discussion

The analysis of the first perturbation did not show outlying responses compared to the subsequent perturbations of a trial. Although Bierbaum et al. (2010) recommended to disregard the first perturbation, no differences in SD were found involving the first and following perturbations. Based on the standard deviations (figure A.13) involving different perturbations, the repeatability is quite consistent in term of variance be-

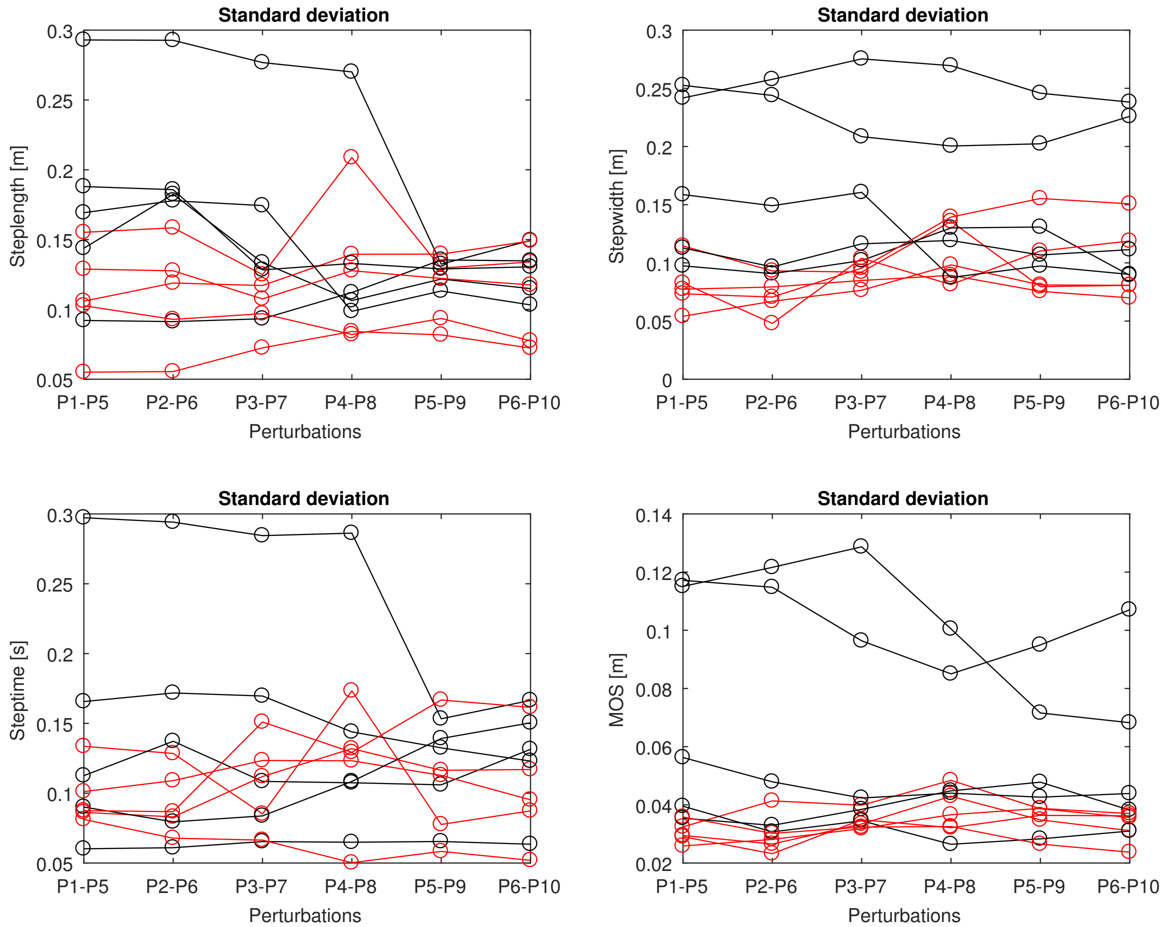

Figure A.13: The Standard deviation over 5 subsequent perturbations of all 5 subjects(each subject one line) for the contra-lateral perturbation (black) and the ipsi-lateral perturbation (red) with a 0.05m intensity

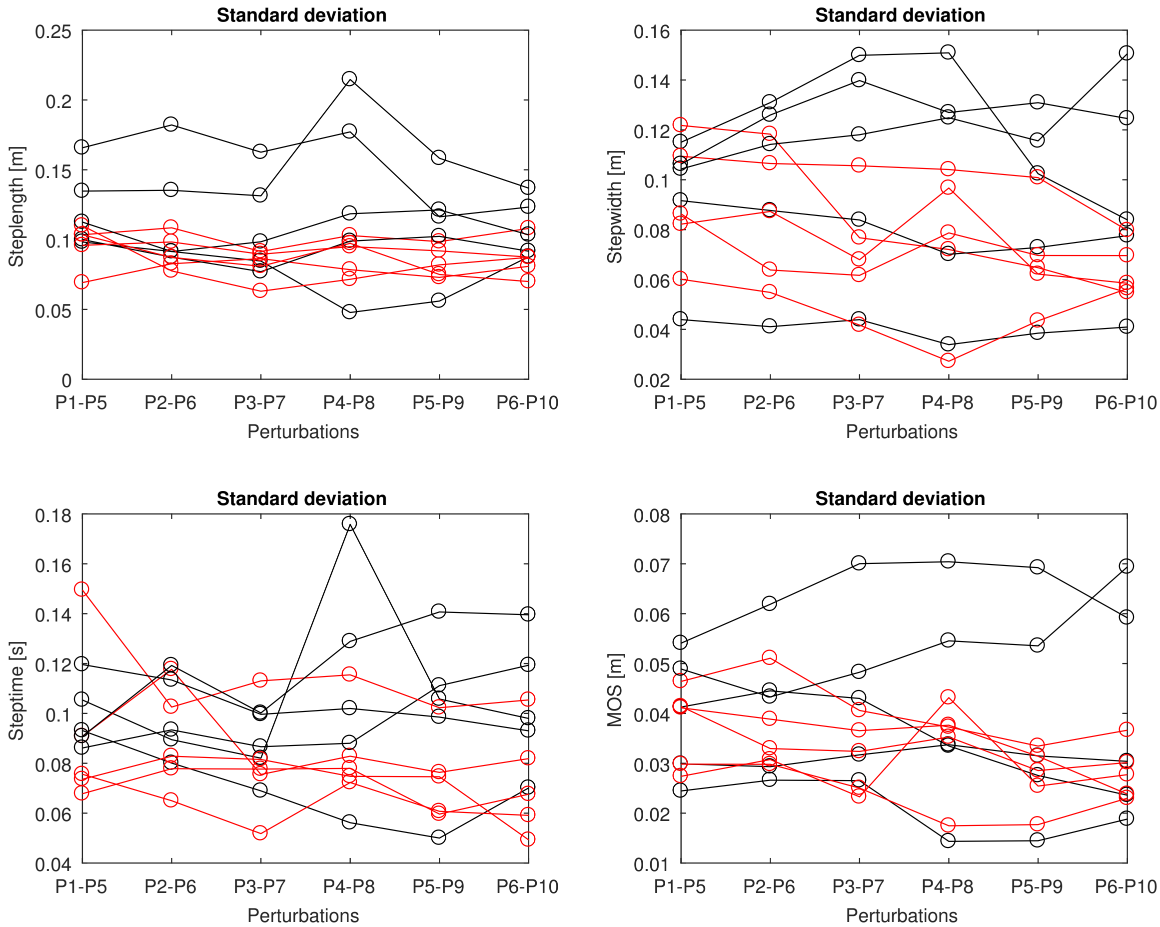

Figure A.14: The Standard deviation over 5 subsequent perturbations of all 5 subjects(each subject one line) for the contra-lateral perturbation (black) and the ipsi-lateral perturbation (red) with a 0.035m intensity

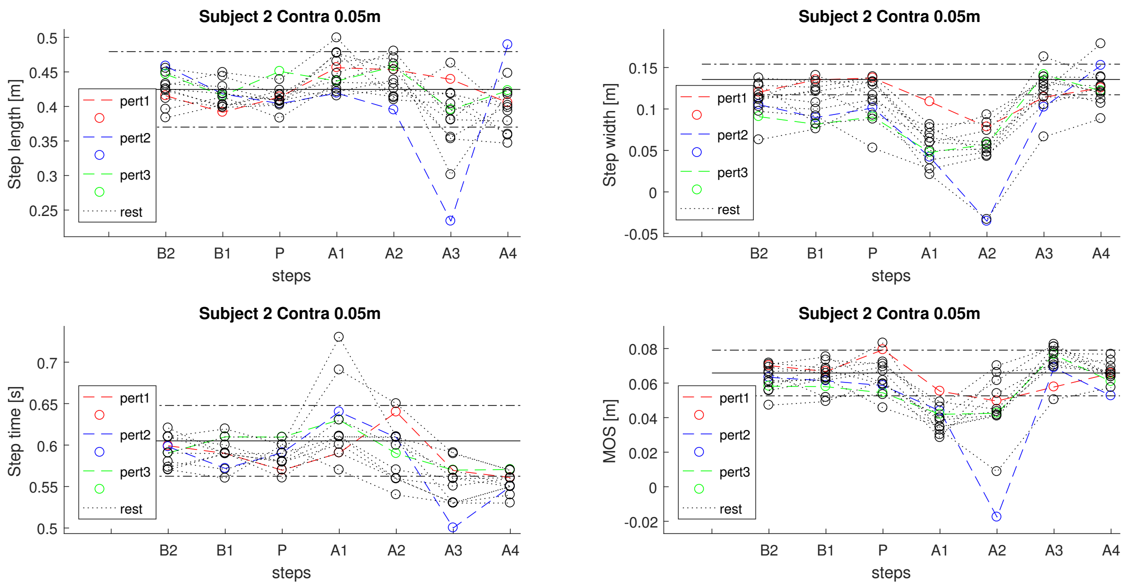

Figure A.15: The Standard deviation of all 5 subjects for the contra-lateral perturbation (black) and the ipsi-lateral perturbation (red) over 5 different perturbations

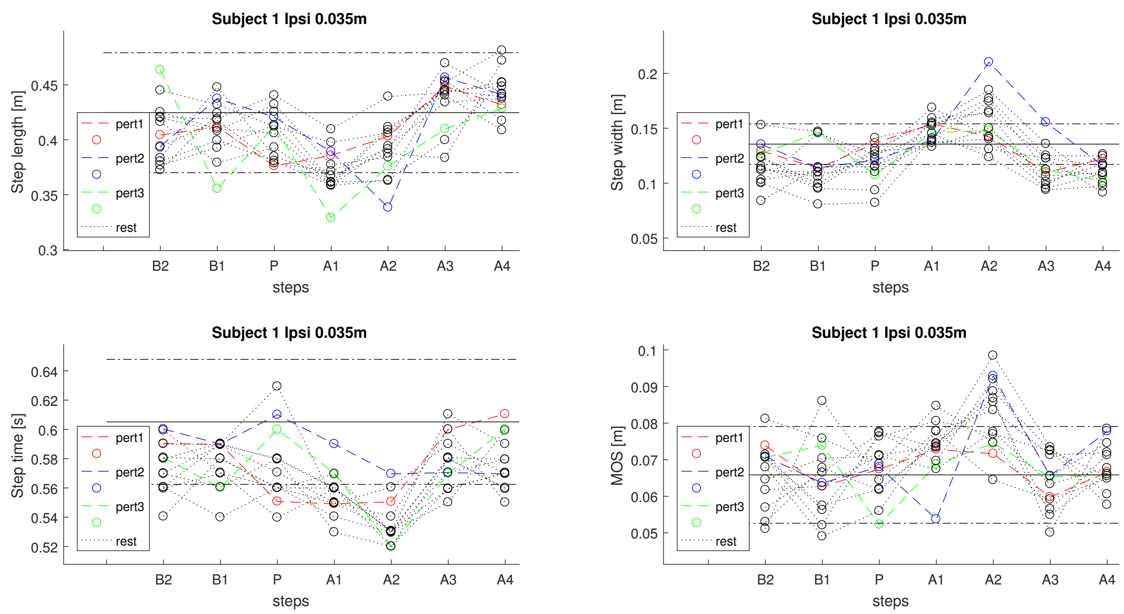

Figure A.16: The Standard deviation of all 5 subjects for the contra-lateral perturbation (black) and the ipsi-lateral perturbation (red) over 5 different perturbations tween the first five perturbations or five different successive perturbations. Therefore, neglecting the first or first few perturbations will not result in an increased repeatability (lower standard deviation) and is not conceived as a necessity. However, as the variance did not tend to be different for the first perturbation, excluding the first perturbation would result in similar findings as well. For this reason, by both in- or excluding the first perturbation a similar outcome is expected. However, it was chosen to include all perturbations in the analysis, as that will increase the amount of data.

A.8. Conclusion

The considered perturbations were valid as they were mainly applied during a single stance phase. The intensity of the perturbation was large enough to detect responses in gait outcomes, yet not so large that it results in dangerous situations. Four steps following the perturbation should be incorporated in the GSN measure and the first perturbation of a trail does not need to be neglected. The contra-lateral perturbation of 0.05m intensity and the described protocol were recommended for further studies in order to discriminate fall-prone subjects.

B

Appendix: Stroke study

B.1. Affected vs Unaffected side

B.1.1. Aim

The aim was to quantify whether the ability to recover from external perturbations on the walking surface could identify patient specific responses in relation to the affected and unaffected side.

B.1.2. Methods

The GSN gave a single value for the response to each perturbation in which the absolute response of the gait indicators compared to baseline in 4 steps following the perturbation were summed. The outcomes were differentiated between affected (non-dominant) side and unaffected (dominant) side responses. The effects of the perturbations to the affected side of the stroke patient were compared to the perturbations of the non-dominant side of the healthy subject and the unaffected side was compared to the healthy dominant side using independent t-tests. These comparisons were also performed for the normal walking condition to determine baseline differences.

B.1.3. Results

Normal walking

There were no significant differences in outcomes between the dominant and non-dominant legs of healthy participants, except for step width during fixed speed walking where the step width in the direction of the dominant leg was 0.00029m larger. Since the mean difference was so small, it cannot be considered clinically relevant. Stroke patients showed a significantly larger step length (fixed and self-paced) and MOS (fixed speed) for their affected leg compared to their non-affected leg.

Table B.1: Normal walking - Group differences affected stroke-healthy and unaffected stroke-healthy

	Affected vs Non-dominant					**Unaffected vs Dominant**				
	Stroke		Healthy-age matched			Stroke		Healthy-age matched		
	Mean	SD	Mean	SD	p-value	Mean	SD	Mean	SD	p-value
Step length DS [m]	0.247	0.076	0.394	0.096	**<0.01**	0.194	0.098	0.388	0.099	**<0.01**
Step width DS [m]	0.182	0.007	0.157	0.052	0.26	0.182	0.065	0.157	0.052	0.26
Step time DS [s]	0.685	0.159	0.563	0.061	**0.01**	0.687	0.191	0.572	0.058	**0.01**
MOS DS [m]	0.086	0.024	0.075	0.016	0.18	0.073	0.014	0.075	0.015	0.18
Step length SP [m]	0.278	0.120	0.552	0.144	0.12	0.234	0.129	0.555	0.137	**<0.01**
Step width SP [m]	0.177	0.066	0.144	0.039	**<0.01**	0.177	0.066	0.144	0.039	0.11
Step time SP [s]	0.744	0.213	0.532	0.037	**<0.01**	0.728	0.231	0.536	0.039	**0.04**
MOS SP [m]	0.086	0.026	0.074	0.012	0.10	0.071	0.018	0.073	0.012	0.65

Perturbations during fixed speed walking
When subjects walked at fixed speed, the response to the perturbation triggered on the affected side did not show any significant outcomes. However, the perturbations triggered on the unaffected side showed an increased step time, MOS, GSN and GSNcorr in stroke patients compared to healthy (Table B.2).

When analysing the mean steps, a similar pattern is seen in the response following affected and the unaffected perturbation in fixed speed. However, in fixed speed the response following the unaffected perturbation showed an increased step time response at A1. Also, the response in MOS seems slightly larger than in unaffected.

Perturbations during self-paced walking
When walking at self-paced, significant differences were shown for all parameters except for the step length. Subjects showed similar responses to perturbations triggered at the affected leg and the unaffected leg. Stroke patients showed a smaller step width, a larger step time and an increased MOS, GSN and GSN corrected in response to perturbations compared to healthy subjects (Table B.3).

When consulting the mean steps of self-paced walking, a larger difference was seen between baseline values of stroke and healthy controls. The pattern per step for affected and unaffected legs were comparable with the mean result of both legs. Healthy subjects decreased step width at A1 and A2, where stroke subjects only increased step width at A2. Healthy controls increased step length at A1, where stroke only slightly increased step length followed by decreases. Stroke increased step time more, especially at A1. MOS were more decreased at A1 compared to healthy, but at A2 stroke patients increased MOS, while healthy remained it decreased.

Notice: The mean and standard deviation lines in the figures are the mean of affected and unaffected together, solely as indication. In the calculations was the mean of the right leg subtracted from right leg values and similar for the left leg.

Table B.2: Independent t-test fixed speed walking. Mean 10 pert affected-healthy and unaffected-healhty

	Affected vs Non-dominant				
	Stroke		Healthy-age matched		
	Mean	SD	Mean	SD	p-value
Step length DS [m]	0.173	0.113	0.17	0.042	0.38
Step width DS [m]	0.128	0.063	0.177	0.075	0.07
Step time DS [s]	0.309	0.204	0.198	0.126	0.14
MOS DS [m]	0.070	0.025	0.069	0.023	0.86
GSN DS	0.249	0.125	0.201	0.061	0.34
GSNcorr DS	1.059	0.504	0.753	0.410	0.12
	Unaffected vs Dominant				
	Stroke		Healthy-age matched		
	Mean	SD	Mean	SD	p-value
Step length DS [m]	0.182	0.074	0.172	0.046	0.65
Step width DS [m]	0.182	0.075	0.163	0.077	0.49
Step time DS [s]	0.343	0.216	0.194	0.126	**0.05**
MOS DS [m]	0.084	0.017	0.064	0.023	**0.01**
GSN DS	0.283	0.121	0.191	0.067	**0.03**
GSNcorr DS	1.202	0.655	0.714	0.419	**0.02**

B.1.4. Discussion
By comparing the responses to perturbations exposed to the unaffected or affected leg with healthy controls, no significant differences were found between the affected stroke side compared to healthy controls non-dominant side in fixed speed walking. However, differences were seen for perturbations exposed to the stroke's unaffected leg compared tot the controls dominant leg. In the first step after the perturbation, an

Table B.3: Independent t-test selfpaced speed walking. Mean 10 pert affected-healthy and unaffected-healhty

	Affected vs Non-dominant				
	Stroke		Healthy-age matched		
	Mean	SD	Mean	SD	p-value
Step length SP [m]	0.229	0.129	0.340	0.453	0.92
Step width SP [m]	0.128	0.093	0.188	0.075	**<0.01**
Step time SP [s]	0.431	0.563	0.136	0.113	**<0.01**
MOS SP [m]	0.076	0.034	0.064	0.027	**0.02**
GSN SP	0.342	0.341	0.251	0.211	**<0.01**
GSNcorr SP	1.361	2.043	0.486	0.252	**<0.01**
	Unaffected vs Dominant				
	Stroke		Healthy-age matched		
	Mean	SD	Mean	SD	p-value
Step length SP [m]	0.244	0.120	0.364	0.454	0.77
Step width SP [m]	0.160	0.084	0.188	0.072	**0.02**
Step time SP [s]	0.414	0.504	0.139	0.102	**<0.01**
MOS SP [m]	0.076	0.027	0.063	0.023	**<0.01**
GSN SP	0.348	0.329	0.262	0.213	**<0.01**
GSNcorr SP	1.353	1.839	0.500	0.235	**<0.01**

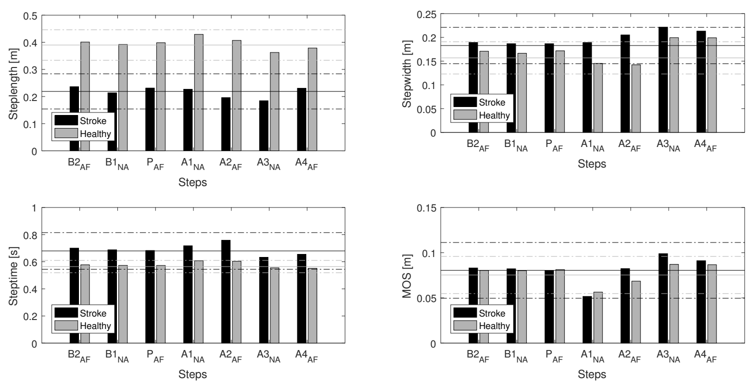

Figure B.1: Perturbation onset on affected/non-dominant leg - fixed speed

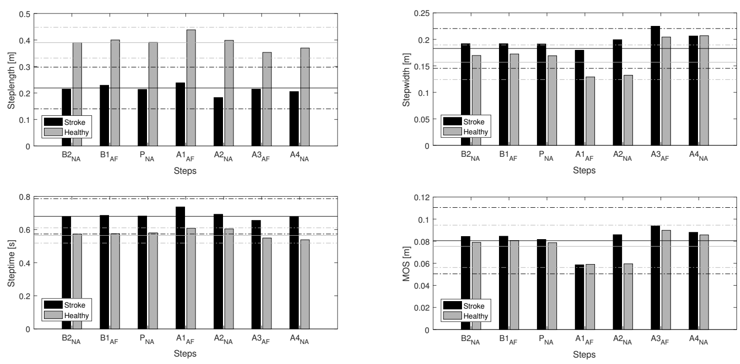

Figure B.2: Perturbation onset on unaffected/dominant leg - fixed speed

adequate response is needed and this has therefore a large contribution in the outcome. To a perturbation exposed to the unaffected side should be responded with the affected side. This might explain the significant difference, as the affected leg might be unable to respond adequately. From the observational analysis, this difference is however not very apparent and the steps follow a similar pattern as the overall response.

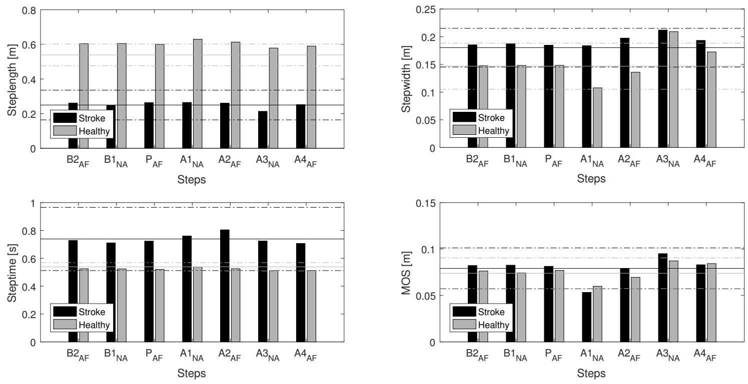

Figure B.3: Perturbation onset on affected/non-dominant leg - self paced

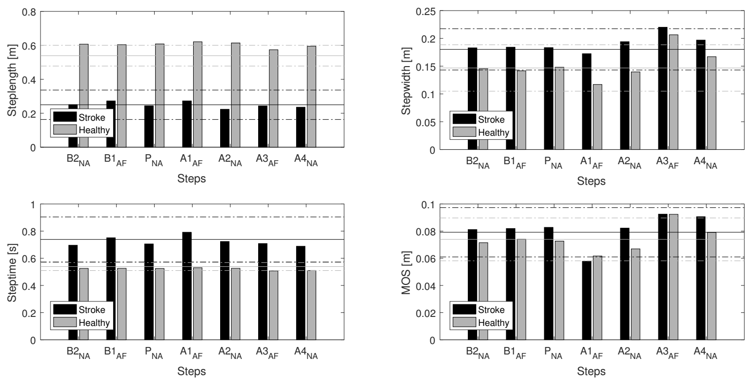

Figure B.4: Perturbation onset on unaffected/dominant leg - self paced

In self-paced walking were significant differences shown for both the response to the affected leg and to the unaffected leg compared to healthy. The exposed affected and unaffected perturbation sides showed a similar order of magnitude of the outcome and similar significant differences. The mean step values showed comparable trends in the response to perturbations following affected and unaffected perturbations. These trends were also resembling a similar pattern as the mean values of both perturbations (without making the differentiation between affected and unaffected). Therefore, in self-paced walking, both the unaffected and the affected side seem to respond similarly.

Conclusion

Overall, it can be concluded that stroke patients showed a larger response in the four steps following the perturbation based on the four gait indicators; step length, step width, step time and MOS. The differences are more dominant for the perturbations applied to the unaffected leg, where the first responsive step is performed with the patients affected leg.

C
Additional Materials

C.1. D-flow control application

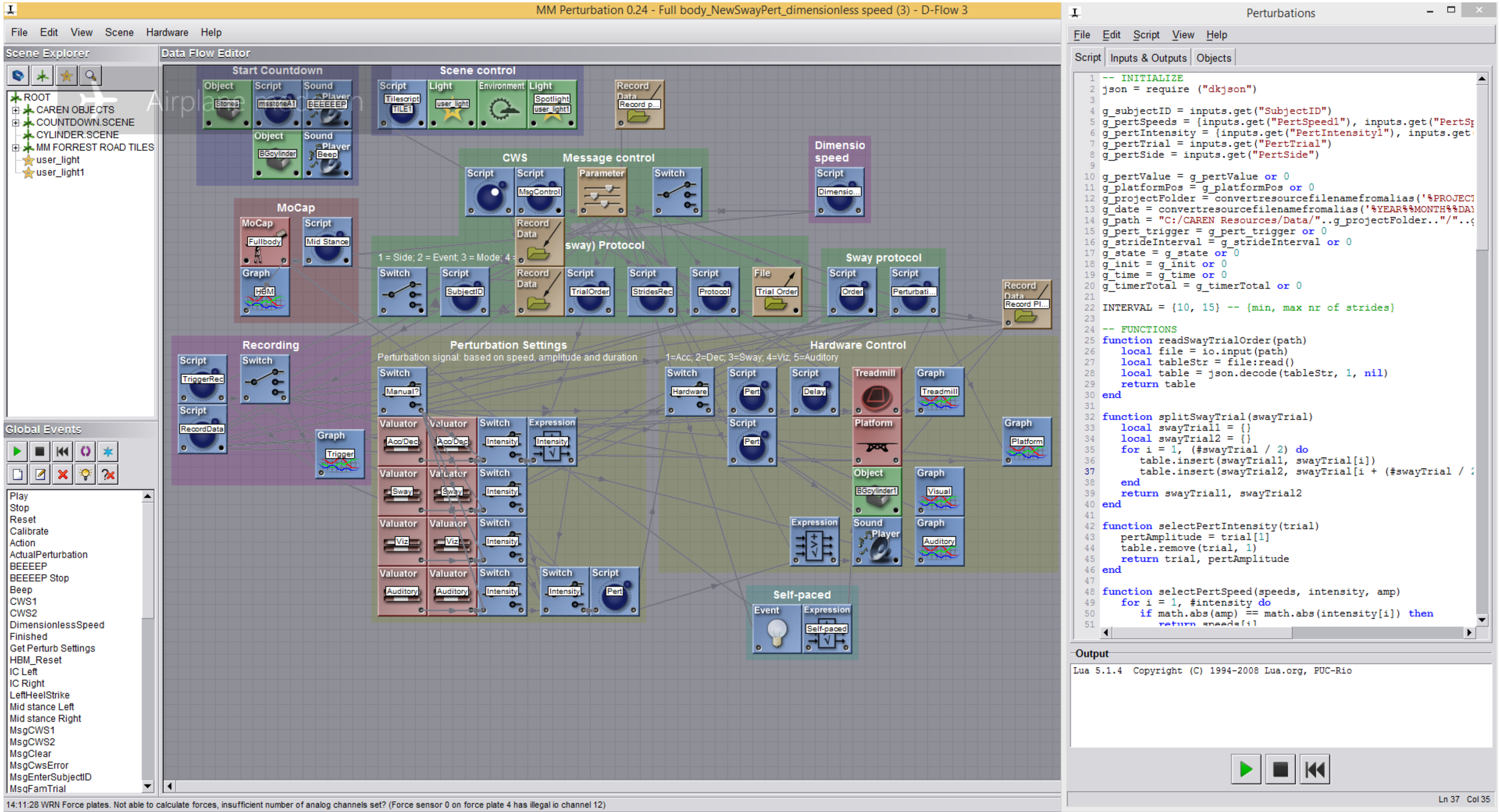

Figure C.1: D-flow application created for the research project controlling the virtual environment, platform excursions and recordings

C.2. Introduction Video

Figure C.2: Explanation and measurement

Figure C.3: Marker placement and preparations

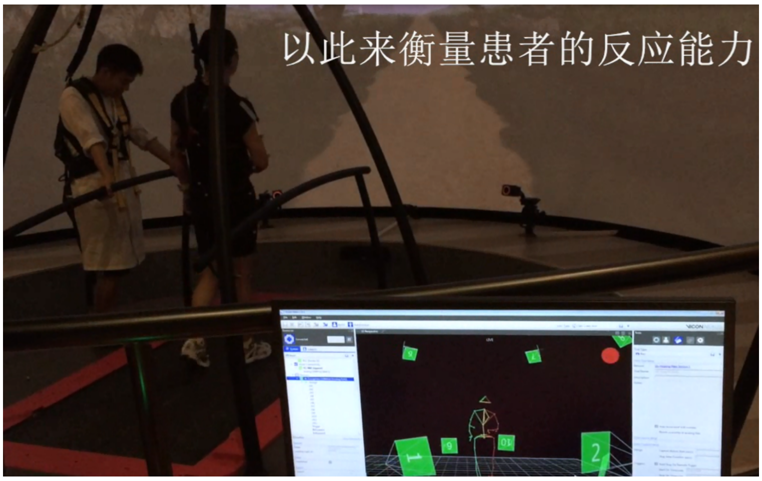

Figure C.4: Perturbation trial and recordings

C.3. Questionnaires

Stroke perturbation study

CAREN Center, Sichuan BaYi Rehabilitation Hospital

Research

Falls are a big problem in post-stroke hemi paretic patients. Therefore it is very important to identify patients with fall risk. To identify fallers, we have to quantify the walking stability. Many measures have been used to describe walking stability, but they often fail to have a relation to actual falling. A possible explanation is that the walking stability was only measured during unperturbed steady state walking. Persons are well capable of adapting for their function loss during steady state walking. It is expected that a poor walking stability is more profound when a person has to deal with disturbances instead of steady state walking. To find a way to identify stroke patients with fall risk, patient will be disturbed during walking with an external sway perturbation. A new measure is used with the aim to find fall prone stroke patients.

The measurements take part on the CAREN (Computer assisted rehabilitation environment). A few walking trials will be recorded in a safe and controlled environment in which there is no actual fall risk for the patient. The measurements will take 1-1.5 hour including preparations. The participant is allowed to have some rest or stop the measurements. Further instructions will be given during each step of the procedures.

Inclusion criteria

- ☐ Hemi paretic post stroke patients, chronic phase 6-18 months after stroke incidence
  - ☐ Patient fell down during the last year (after their stroke incidence) [Circumstances of falling; Appendix 4]
  - ☐ Patient that did not fall during the last year or after their stroke incidence
- ☐ Patient reported to be able to walk for 20 min without assistive device
- ☐ The patient should be able to walk on a slow walking speed (dimensionless speed 0.25 → approximately 0.7m/s
- ☐ No other neurological, musculoskeletal or cardiovascular disorders or co-morbidities
- ☐ ≥19 on lower extremity motor performance of Fugl Meyer Assessment [Appendix 1] Outcome: __________
- ☐ >20 on the short orientation–memory–concentration test [Appendix 2] Outcome: __________

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纳入标准

- ☐ 中风后偏瘫病人，中风后 6-18 个月的慢性期
  - ☐ 去年 15 名中风后跌倒过的病人（附件 4：跌倒情况问卷调查）
  - ☐ 去年或中风后 15 名未跌倒过的病人
- ☐ 无辅具下能走 20 分钟
- ☐ 能在 0.7m/s 速度下慢速走路，无因次参数速度 0.25
- ☐ 无其他神经性、肌肉骨骼、心血管疾病或并发症
- ☐ Fugl Meyer 下肢运动功能量表≥19 （附件 1）　结果：
- ☐ 短时定向力、记忆力、注意力量表≥20 （附件 2）　结果：

Patient information

Name:______________________ Date:______________________

Date of birth	
Phone number	
Date stroke	
Stroke characteristics	
Affected side	
Doctor	
Duration hospital	

Preparations before experiment [included patients only]

- ☐ Fall efficacy scale [Appendix 3]
- ☐ Circumstances of falling [Appendix 4]
- ☐ Berg Balance score [Appendix 5]　Outcome: __________
- ☐ Timed up and go score [Appendix 6]　Time: _____________

CAREN preparations

Knee width	
Ankle width	
Walking speed	
Folder	
ID	
Date measurement	

- Inform procedure
  Dimensionless speed
  - Familiarization (3min)
  - Normal walking (2min)
  - Perturbation trial 1 (3min)

  Self-paced
  - Familiarization (2min)
  - Normal walking (2min)
  - Perturbation trial 1 (3min)

- Explain VAS scale (tiredness & pain) & speed questions

During experiment [included patients only]

	Before	Normal walking	Perturb 1	SP walking	Perturb 2
VAS tiredness [0-10]					
VAS pain [0-10]					
Difficulty [0-5]					
Speed [1:slow, 2: normal, 3: fast]					
Nervous [0-10]					

Inclusion patients

Appendix 1: Fugl Meyer Assessment

Name:________________________ Date: ___________________________

下肢	**1 反射活动（仰卧位）**			
	（1）跟腱反射（2）	0 分□：无反射活动		2 分□：有反射活动
	（2）（髌）膝腱反射（2）	0 分□：无反射活动		2 分□：有反射活动
	2 联带运动（仰卧位）			
	屈肌联带运动			
	（1）髋关节屈曲（2）	0 分□：不能进行	1 分□：部分进行	2 分□：充分进行
	（2）膝关节屈曲（2）	0 分□：不能进行	1 分□：部分进行	2 分□：充分进行
	（3）踝关节背屈（2）	0 分□：不能进行	1 分□：部分进行	2 分□：充分进行
	伸肌联带运动			
	（4）髋关节伸展（2）	0 分□：没有运动	1 分□：微弱运动	2 分□;几乎与对侧相同
	（5）髋关节内收（2）	0 分□：没有运动	1 分□：微弱运动	2 分□;几乎与对侧相同
	（6）膝关节伸展（2）	0 分□：没有运动	1 分□：微弱运动	2 分□;几乎与对侧相同
	（7）踝关节跖屈（2）	0 分□：没有运动	1 分□：微弱运动	2 分□;几乎与对侧相同
	3 伴有联带运动的活动（坐位）			
	（1）膝关节屈曲大于 90 度（2）	0 分□：无主动运动	1 分□：膝关节能从微伸位屈曲，但<90 度	2 分□：屈曲>90 度
	（2）踝背屈（2）	0 分□：不能主动背屈	1 分□：主动背屈不完全	2 分□：正常背屈
	4 分离运动（髋关节 0 度）站位）			
	（1）膝关节屈曲（2）	0 分□：在髋关节伸展位时不能屈膝	1 分□：髋关节不屈曲的情况下，膝能屈曲，但<90 度，或在进行时髋关节屈曲	2 分□：能自如运动
	（2）踝背屈（2）	0 分□：不能主动活动	1 分□：能部分背屈	2 分□：能充分背屈
	5 正常反射（坐位）（2）只有第 4 阶段得 4 分，本项目评分才计入总分			
	膝部屈肌、膝反射、跟腱反射	0 分□：2~3 个明显亢进	1 分□：1 个反射亢进或 2 个反射活跃	2 分□：活跃的反射≤1 个
	6 协调/速度：跟膝胫试验（连续重复 5 次）			
	（1）震颤（2）	0 分□：明显震颤	1 分□：轻度震颤	2 分□：无震颤
	（2）辨距障碍（2）	0 分□：明显的不规则的辨距障碍	1 分□：轻度的规则的辨距障碍	2 分□：无辨距障碍
	（3）速度（2）	0 分□：比健侧长 6 秒	1 分□：比健侧长 2~5 秒	2 分□：比健侧长 2 秒

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Inclusion patients

Appendix 2: Short Orientation Memory Test

Name:_______________________ Date: __________________________

短时定向力、
记忆力、
注意力量表

病人姓名：
评定者：
时间：

编号	问题	最大错误量	分数*比重
1	今年是几几年？	1	-------*4 =-----------
2	现在是几月份？	1	-------*3 =-----------
	重复这个短句 李小花 永宁镇 八一路 81 号 或者： 李小花 金牛区 大天路 5 号		
3	这个是几点？（一个小时 内）	1	----------*3 =-------------
4	倒数 20 到 1	2	---------*2 =-------------
5	以倒序说月份-	2	---------*2 =-------------
6	重复刚才给的那句话	5	---------*2 =-------------

总共错误得分 =----------/28

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Appendix 3: Fall efficacy scale

Name:_______________________ Date: __________________________

国际跌倒效能量表

我将问一些问题，关于你担心跌倒的可能性。以下有各类活动，请选出你做这项活动时最接近你担心程度的选项。请回想一下你通常做这些活动时的想法，如果你没做过（例如：有人帮你购物），请联想如果由你自己做你的担心程度。

	一点不担心	些许担心	相当担心	非常担心
1、 打扫房间（例如 打扫、吸尘）				
2、穿脱衣服				
3、准备简单的饭				
4、洗澡或淋浴				
5、去购物				
6、从椅子上起来或坐下				
7、上楼梯或下楼梯				
8、在附近逛				
9、拿高于头顶或地上的东西				
10、在铃声停止前接电话				
11、在滑的路面上走（例如湿的、冰的）				
12、拜访朋友或亲戚				
13、在人群中走				
14、在不平的路上走（如石子路、维护不善的路）				
15、上斜坡或下斜坡				
16、外出参加社交活动（如做礼拜、家庭聚会				
小计				
计				/64

Appendix 4: Circumstances of falling

Name:________________________ Date: ___________________________

跌倒情况问卷调查

通过以下问题我们将了解更多关于导致跌倒发生的情况。请回想在过去一年发生的跌倒。如果需要，可多选。

1. 最近 12 个月内跌倒过多少次

____

备注：跌倒是指因为无意识的失去平衡导致摔倒在地，并且不是由癫痫、中风或其他疾病发作导致。

2. 跌倒的原因

☐绊倒
☐滑倒
☐错位步伐
☐失去平衡
☐腿屈曲
☐碰倒
☐失去支撑
☐其他___________________

3、跌倒的地方？

☐在家，室外
☐在家，室内
☐不在家，熟悉的地方
☐不在家，陌生的地方

4、跌倒时在做什么?

☐ 在水平路面或地面走路
☐在不平路面或地面走路
☐急着完成工作
☐在花园或院子工作
☐搬重物或庞大的物体
☐上楼梯
☐下楼梯
☐站立时四处观望或转身
☐运动
☐其他_______________

5、跌倒导致的损伤?

____
____

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Preparations

Appendix 5: Berg Balance score

Name:________________________ Date: ___________________________

Berg 平衡量表

姓名： 性别： 年龄： 住院号： 评定者： 评定日期:

1、从坐位站起	4 分□：不用扶手能够独立地站起并保持稳定 3 分□：用手扶着能够独立地站立 2 分□：几次尝试后自己用手扶着站起 1 分□：需要他人小量的帮助才能站起或保持稳定 0 分□：需要他人中等或大量的帮助才能站起或保持稳定
2、无支持站立	4 分□：能够安全站立 2 分钟 3 分□：在监护下能够站 2 分钟 2 分□：在无支持的条件下能够站立 30 秒 1 分□：需要若干次尝试才能无支持地站立达 30 秒 0 分：无帮助时不能站立 30 秒
3、无靠背坐位，但双脚着地或放在一个凳子上	4 分□：能够安全地保持坐位 2 分钟 3 分□：在监护下能够保持坐位 2 分钟 2 分□：能坐 30 秒 1 分□：能坐 10 秒 0 分□：没有靠背支持不能坐 10 秒
4、从站立位坐下	4 分□：最小量用手帮助安全地坐下 3 分□：借助于双手能够控制身体的下降 2 分□：用小腿的后部顶住椅子来控制身体的下降 1 分□：独立地坐，但不能控制身体下降 0 分□：需要他人帮助坐下
5、转移	4 分□：稍用手扶就能够安全地转移 3 分□：绝对需要用手扶着才能够安全地转移 2 分□：需要口头提示或监视才能够转移 1 分□：需要一个人的帮助 0 分□：为了安全，需要两个人的帮助或监视
6、无支持闭目站立	4 分□：能够安全地站 10 秒 3 分□：监视下能够安全地站 10 秒 2 分□：能站 3 秒 1 分□：闭眼不能达 3 秒钟，但站立稳定 0 分□：为了不摔倒而需要两个人的帮助
7、双脚并拢无支持站立	4 分□：能够独立地将双脚并拢并安全站立 1 分钟 3 分□：能够独立地将双脚并拢并在监视下站立 1 分钟 2 分□：能够独立地将双脚并拢，但不能保持 30 秒 1 分□：需要别人帮助将双脚并拢，但能够双脚并拢站保持 15 秒 0 分□：需要别人帮助将双脚并拢，双脚并拢站立不能保持 15 秒
8、站立时上肢向前伸展并向前移动	4 分□：能够向前伸出＞25 cm 3 分□：能够安全地向前伸出＞12 cm 2 分□：能够安全地向前伸出＞5 cm 1 分□：上肢可以向前伸出，但需要监视 0 分□：在向前伸展时失去平衡或需要外部支持
	上肢向前伸展达水平位，检查者将一把尺子放在之间末端，手指不能触及尺子。测量的距离是被检查者身体从垂直位到最大前倾位时手指向前移动的距离。如可能，要求被检查者伸出双臂以避免躯干的旋转。
9、站立位时从地面捡起物品	4 分□：能够轻易地且安全地将鞋捡起 3 分□：能够将鞋捡起，但需要监视 2 分□：伸手向下达 2-5cm 且独立地保持平衡，但不能将鞋捡起 1 分□：试着做伸手向下捡鞋的动作时需要监视，但仍不能将鞋捡起 0 分□：不能试着做伸手向下捡鞋的动作，或需帮助免于失去平衡或摔倒
10、站立位转身向后看	4 分□：从左右侧向后看，体重转移良好 3 分□：仅从一侧向后看，另一侧体重转移较差 2 分□：仅能转向侧面，但身体的平衡可以维持 1 分□：转身时需监视 0 分：需要帮助以防止失去平衡或摔倒
11、转身 360°	4 分□：在≤4 秒的时间内安全地转身 360° 3 分□：在≤4 秒的时间内仅能从一个方向安全地转身 360° 2 分□：能够安全地转身 360° 但动作缓慢 1 分□：需要密切监视或口头提示 0 分：转身时需要帮助
12、无支持站立时将一只脚放在台阶或凳子上	4 分□：能够安全且独立地站，在 20 秒的时间内完成 8 次 3 分□：能够独立地站，完成 8 次的时间＞20 秒 2 分□：无需辅助具在监视下能够完成 4 次 1 分□：需要少量帮助能够完成＞2 次 0 分：需要帮助以防止摔倒或完全不能做
13、一只脚在前无支持站立	4 分□：能够独立地将双脚一前一后地排列（无间距）并保持 30 秒 3 分□：能够独立地将一只脚放在另一只脚的前方（有间距）并保持 30 秒 2 分□：能够独立地迈一小步并保持 30 秒 1 分□：向前迈步需帮助，但能够保持 15 秒 0 分□：迈步或站立时失去平衡
14、单腿站立	4 分□：能够独立抬腿并保持时间>10 秒 3 分□：能够独立抬腿并保持 5-10 秒 2 分□：能够独立抬腿并保持时间≥3 秒 1 分□：试图抬腿，不能保持 3 秒，但可维持独立站立 0 分□：不能抬腿或需要帮助以防止摔到

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Appendix 6: Timed up and go test

Timed Up and Go (TUG) Test

Name:____________________________ MR: _______________________ Date:________

1. Equipment: arm chair, tape measure, tape, stop watch.
2. Begin the test with the subject sitting correctly (hips all of the way to the back of the seat) in a chair with arm rests. The chair should be stable and positioned such that it will not move when the subject moves from sit to stand. The subject is allowed to use the arm rests during the sit – stand and stand – sit movements.
3. Place a piece of tape or other marker on the floor 3 meters away from the chair so that it is easily seen by the subject.
4. Instructions: "On the word GO you will stand up, walk to the line on the floor, turn around and walk back to the chair and sit down. Walk at your regular pace.
5. Start timing on the word "GO" and stop timing when the subject is seated again correctly in the chair with their back resting on the back of the chair.
6. The subject wears their regular footwear, may use any gait aid that they normally use during ambulation, but may not be assisted by another person. There is no time limit. They may stop and rest (but not sit down) if they need to.
7. Normal healthy elderly usually complete the task in ten seconds or less. Very frail or weak elderly with poor mobility may take 2 minutes or more.
8. The subject should be given a practice trial that is not timed before testing.
9. Results correlate with gait speed, balance, functional level, the ability to go out, and can follow change over time.

Normative Reference Values by Age

Age Group	Time in Seconds (95% Confidence Interval)	
60 – 69 years	8.1	(7.1 – 9.0)
70 – 79 years	9.2	(8.2 – 10.2)
80 – 99 years	11.3	(10.0 – 12.7)

Cut-off Values Predictive of Falls by

Group	Time in Seconds
Community Dwelling Frail Older Adults	> 14 associated with high fall risk
Post-op hip fracture patients at time of discharge[3]	> 24 predictive of falls within 6 months after hip fracture
Frail older adults	$\geq$ 30 predictive of requiring assistive device for ambulation and being dependent in ADLs

Date	Time	Date	Time	Date	Time	Date	Time