1 Introduction

Eighty-three percent of stroke survivors have balance deficits which are related to poor mobility and reduced independence in activities of daily living [1]. Control of balance and foot-placement are closely associated and impairments in these are thought to contribute to stroke survivors' increased risk of falls due to trips, slips and misplaced steps [2,3]. The ability to adapt foot-placement is the most effective means of balance control when walking [4], allowing safe navigation of cluttered and dynamic community environments. Understanding stroke survivors' ability to adjust foot-placement when step adjustments enlarge or reduce base of support (BoS) (and therefore challenge or aide balance) may help identify targets for rehabilitation (e.g practice altering foot-placement and/or compensations to increase BoS).

Foot-placement defines the acceleration of the body at push-off as well as size and shape of future base of support (BoS) [4–6]; making accurate foot-placement control intrinsically important for balance control [5]. Foot-placement that affords a larger BoS (e.g. lengthening/widening) may enhance balance. Whereas adaptations of foot-placement that limit the BoS (e.g. step shortening and narrowing) challenge balance and, possibly as a result, are less accurately achieved by healthy adults [6–8]. Stroke survivors are known to have difficulties adjusting foot-placement [7,8], these impairments have been related to clinical balance assessment scores [9]; underscoring the importance of control of foot-placement in maintenance of balance. However, it is unclear whether stroke survivors' ability to control foot-placement is affected different when step adjustments enlarge the BoS as opposed to reduce it.

Many stroke related impairments including foot deformity, weakness and spasticity (and many others) have the capacity to affect how well stroke survivors will be able to take weight, distribute their CoP in their BoS, aim the paretic limb etc. and therefore will affect both control of foot-placement and balance. One rehabilitation tool commonly prescribed to overcome any one of these impairments is the use of walking aides (crutches); which externally enlarge BoS and provide additional compensatory control of CoM momentum. Indeed one study has shown that balance support from crutches improves stroke survivors' accuracy in narrowing steps from standing [7]. Further, during steady state walking, aides (such as a cane) have been shown to increase stride time, step length and swing time and stroke survivors reduced their step width (and hence BoS) to resemble that of healthy adults

[10]. However, one of the main aims of prescribing walking aides is to promote independent mobility and/or maintenance of balance in cluttered and dynamic environments. Therefore, a greater understanding of the effects of walking aides on walking when step adjustments must be made in response to the environment is needed to determine if these prescribed aides are indeed improving mechanics of walking in the circumstances for which they are intended.

This study examines control of foot-placement when the BoS is reduced (challenging balance) or enlarged (potentially enhancing balance), with and without crutch support, in stroke survivors and healthy young and older adults. We hypothesize that during walking, crutch support will facilitate greater stepping accuracy during adjustments of foot-placement; reducing stepping error, particularly, when step adjustments reduce BoS and for stroke survivors who have greater error in foot-placement than healthy counterparts.

2 Methods

2.1 Participants

Young (18-40 years old), and older (age matched with stroke survivors) healthy adults and stroke survivors participated. For stroke survivors, inclusion criteria were: >6 months post-stroke, 10m walking test >30s, able to walk 10m independently without orthopaedic aids or assistance. Exclusion criteria for both healthy participants and stroke survivors included neuro-musculoskeletal (apart from stroke) conditions affecting walking ability and receptive and/or language problems that could preclude informed consent. In line with research governance policies in the UK, the University ethics committee (HSR1617-27) approved the study and all participants provided written informed consent prior to participation in line with the declaration of Helsinki.

Demographic and anthropometric data collected included: date of birth, date of stroke, hemiparesis side, height, weight. In addition, a clinical assessment of cognition (Montreal cognitive assessment[11]), executive function (trail making test A and B[12]), balance (Berg balance scale[13]), gait adaptability (10-item dynamic gait index[14]), motor recovery (Fugl-Meyer assessment of lower limb[15] and Modified Ashworth Scale [16]), falls history (number of falls in the past 12 months) and visual field (apple cancelation test [17]) deficits was carried out. One stroke survivor showed signs of egocentric and allocentric neglect, but was able to see and step to targets projected on the treadmill and participate safely.

2.2 Apparatus

A treadmill with a single embedded force platform (C-Mill, MotekForcelink, Culemborg, The Netherlands) was used allowing for online gait event detection while projecting visual stepping targets, as previously validated [18]. Two forearm crutches were installed at the side of the treadmill adjacent to handlebars for the supported trials (see figure 1A). Participants walked on the spot at the centre of the treadmill belt so crutches could be fixed in one location. Height of crutches were adjusted for each participant so that forearms rested in the cups at elbow height and handles could be comfortably gripped. Stroke survivors were free to adjust paretic arms for comfort. For both supported and unsupported walking conditions participants were instructed to walk and place their feet as accurately as possible in targets. A six-camera system (Qualisys, Gottenorg, Sweden) was used to track markers on the foot synchronously with measurements obtained through the software of the treadmill, including the centre of pressure (CoP). This synchronisation is established by a trigger pulse sent from the C-Mill to the Qualysis software starting the kinematic data capture. Calibration of the Qualysis system was aligned at the origin of the C-Mill force plate.

2.3 Familiarization

Participants familiarised themselves with walking on the treadmill for roughly 3 minutes. Firstly, SSWS was determined by gradually increasing speed from 1km/h by 0.1km/h increments until participants felt they reached comfortable walking speed, followed by a one minute walking period. Targets were then projected on the treadmill positioned at participants' usual step lengths and widths to allow participants to become acquainted with target stepping (CueForce1, MotekForcelink) for one minute. After this minute of familiarisation with target stepping, participants walked without targets at their SSWS for 30 seconds; in this period, step length and width were calculated to inform future target locations (custom Matlab program based on step lengths and widths recorded by CueForec1).

2.4 Experimental setup and protocol

All participants were asked to complete ten trials of 100 steps each: one stepping trial with targets placed at preferred foot landings, and three trials of each of the following conditions (Figure 1), 1.) No targets; baseline trial, 2.) Unsupported adaptable target stepping without crutches for support, 3.) Supported adaptable target stepping with support of crutches. In these adaptability trials 24 targets were placed to alter preferred foot-placement (6 each of shortening, lengthening, narrowing and widening) interspersed semi-randomly with 76 preferred foot landing targets. All targets were visible 2 steps in advance. Stroke survivors wore a safety harness to prevent a fall, this harness did not provide any support for weight or balance.

These ten walking trials were presented in a randomised order and interspersed with the baseline trials as the 1st, 5th and 10th trial. All trials together accounted for a total of 1000 steps. Whenever the participant requested it, or the researcher deemed it necessary, a rest was taken between trials.

2.5 Measure of stepping performance

Stepping error was used to measure accuracy of foot-placement control. First, the centre of the foot (CoF) was calculated from four foot markers (calcaneus, 1st, and 5th metatarsal head and 2nd distal phalanx head) at midstance. Stepping error was then defined as the distance of the CoF to the centre of target. The error of foot-placement was analysed separately in the medio-lateral (ML) and antero-posterior (AP) directions for all steps. Absolute error was used to compare the magnitude of error, and the average bias (calculated as the average of the signed error magnitude) of foot-placement was used to analyse the direction of error (undershoot vs. overshoot) between the different steps.

The size of the foot was projected around the CoF and overlaid on the target (custom Matlab 2016a). When no part of the foot was in contact with the target, it was considered a missed step. The percentage of missed steps was calculated as the number of misses divided by the total number of steps taken in that condition.

2.6 Statistics

In total four ANCOVAs were carried out individually on each outcome measure; absolute AP error, AP bias, absolute ML error and ML bias. The same model was used in each of these four ANCOVAs with 3 within subject factors: 1) condition (supported and unsupported) 2) steps (BoS enlarging, preferred or BoS limiting) and 3) side (left right for healthy and paretic and non-paretic for stroke) and the between subjects factor of groups (young, older adults and stroke survivors). Self-selected walking speed is known to be related to balance control and accuracy of foot-placement adjustments [9] and so was used as a covariate. Post-hoc comparisons were assessed using Bonferroni test with adjustment for multiple comparisons. A $p<0.05$ was used for statistical significance. The percentage of missed targets was angular transformed to stabilize the variance and reach normal distribution as in [6,19], and compared between different conditions (unsupported and supported) with a repeated measures ANOVA.

3 Results

3.1 Participants

Eleven stroke survivors, 10 older and 13 young healthy adults participated in the study (See table 1 for participant demographics). According to suggested thresholds for SSWS of stroke survivors [20], based on treadmill walking speeds two stroke survivors were non-functional walkers (<0.4m/s), seven limited outdoor walkers (0.4-0.8m/s) and two healthy walkers (>0.8m/s). Three healthy older and one young adult were healthy walkers (0.4-0.8m/s) and the rest of the healthy adults walked at SSWS exceeding the 0.8m/s limit [13]. According to the suggested thresholds for Berg balance scores, one stroke survivor should be walking with an assistive device (score <40), two had higher risk of falls (score <45)[21]. It must be taken into consideration that treadmill walking speeds are slower than over ground walking speeds, likely due to the increased metabolic cost[22] and the fact treadmill walking was set on target stepping speed. Therefore, these functionality assessments may underestimate participants' walking abilities. Based on the TUG five stroke survivors would be at increased risk of falls and four of these five also have an increased falls risk indicated by the 10-item DGI.

3.2 Absolute error

There were no overall group differences for absolute error (means and standard deviations are represented in table 2). There was an overall effect of crutch support which reduced absolute error in both the AP ($F_{(1, 30)}$ =13.518, p=0.001, η_p^2= 0.854, see figure 2A) and ML ($F_{(1, 30)}$ =18.141, p<0.001, η_p^2= 0.377, see figure 2B) directions for all groups (Figure 2). No interaction effects for condition and direction of step or for group was found.

3.3 Bias

An interaction effect of crutch support, side and group was found ($F_{(2,30)}$=3.871, p=0.031, η_p^2= 0.205, see figure 3A). A main effect of crutch support in the AP direction ($F_{(1, 30)}$ = 5.970, p=0.021, η_p^2= 0.166) and a main effect of side ($F_{(1, 30)}$ = 7.655, p=0.010, η_p^2= 0.203) were also found. Additionally, an interaction effect of step direction (shortening, preferred and lengthening) and group ($F_{(4,60)}$=7.238, p<0.001, η_p^2= 0.325, see figure 3B) and a main effect of step direction (($F_{(2, 29)}$ = 117.313, p<0.001, η_p^2= 0.890) were found. indicating lengthening steps are significantly undershot more than preferred steps (p<0.001) and shortening steps or significantly overshot more than preferred steps (P<0.001). However, in stroke survivors all steps were undershot (see figure 3B). No significant differences were found for crutch support or side in ML direction (for means and standard deviations see table 2).

3.4 Percentage of missed targets

A main effect of group was found for the percentage of misses ($F_{(2, 31)}$ =11.091 p=0.001, η_p^2= 0.417). Figure 4 shows that stroke survivors in general missed significantly more targets than healthy young and older adults in both crutch supported and unsupported target stepping conditions. No interaction effect was found for group and crutch support such that crutch support was not seen to reduce targets missed for stroke survivors more than healthy young or older adults (see table 2 for means and standard deviations).

4 Discussion

This is the first study to examine the effects of altering shape and size of BoS and use of crutch support for balance, on foot-placement control during walking in healthy and older adults and stroke survivors. Understanding which aspects of altering foot-placement in response to the environment are difficult for stroke survivors (compared to healthy counterparts), and how any difficulty is altered by use of commonly prescribed crutches is essential to target rehabilitation efforts to achieve better functional mobility outcomes. The method used to evaluate control of foot-placement (support with crutches and foot-placement altering size of BoS in response to the environment) mimics clinical approaches to treatment as well as foot-placement adjustments required in real world. The paradigm therefore allows insight into not only the mechanisms of control of foot-placement for stroke survivors, but also potential benefits of typical treatments. This study shows that crutch support increases foot-placement accuracy, in any direction of step adjustment, in all participant groups during walking. There were no significant differences in foot-placement accuracy between the paretic side and the non-paretic side for stroke survivors and crutch support did not improve accuracy more when changing foot-placement in one direction over another.

Crutch support reduced the magnitude of stepping error (absolute error) in both AP and ML directions; indicating foot-placement accuracy benefits generally from balance support. The benefits of crutch support on foot-placement accuracy also helps all participant groups; even young healthy adults. While not statistically significant (when differences in SSWS are accounted for as a covariate), reductions in error due to balance support are largest for stroke survivors (see figure 2). The effects of speed of walking on accuracy of controlling foot-placement (speed-accuracy trade-offs) are indicated to be non-linear when balance must also be maintained [23]. While speed-accuracy trade-offs have been examined only in discrete individual steps from standing in healthy participants stroke survivors are less successful at avoiding obstacles with less time [19] and slow down to achieve accuracy target stepping tests [9]. Combined with the findings here that crutch support improves accuracy of foot-placement (when speed of walking is accounted for), this suggests that speed of walking is an important underlying factor in ability to accurately place and adjust foot landings and consideration should be taken if rehabilitation goals to increase walking speed may compromise ability to adapt foot-placement to maintain balance in response to the environment.

Our study showed smaller reduction in foot-placement error due to support (about 2.1cm, 6.8cm unsupported, 4.7cm unsupported) than previously reported by Nonnekes et al.,(2010) (about 7cm, 11cm unsupported to 4 cm supported). This may be due to a difference of response time in the 2 training paradigms; 3 seconds [7] or 2 steps. Alternatively, the dynamic nature of foot-placement adaptation in our paradigm vs. the singular step by Nonnekes et al.,(2010) could account for the smaller reduction in stepping error with crutch support in walking (compared to standing). Adjusting foot-placement when making a single step to a standing position requires foot-placement to arrest CoM momentum. Whereas, during gait, foot-placement adjustments need to perpetuate CoM momentum, with control through anticipation of CoM trajectory in subsequent steps [24,25]. Differences in the size of effect of crutch support on foot-placement error between target stepping paradigms during walking and a singular step, suggest that balance differentially affects foot-placement control across these two contexts [4,25].

Step adjustments which limited the BoS (narrowing and shortening steps) were expected to be more challenging (higher error) when unsupported (and with potentially larger reductions in error when balance was supported) than those that enlarged the BoS (widening and lengthening steps). Surprisingly, despite the fact that narrowing steps directly challenge balance when making a single step from a stationary standing position [7,26], narrowing and widening steps were seen to have similar error magnitudes. This indicates that when walking, making accurate step adjustments in any direction may challenge balance equally and, again, influences of balance on stepping accuracy are different according to the context in which balance must be maintained (walking vs singular step).

Stroke survivors missed significantly more targets than healthy young and older counterparts did; indicating stroke survivors have larger errors than just those seen with healthy ageing. Reductions in error with crutch support were too small to translate to significantly fewer targets missed. Stroke survivors in this study missed 9% of targets which is comparable [27] or even higher [9] than reported in previous studies. While only a small number of targets were missed the potential consequences of even inaccurate stepping (error as opposed to completely miss-stepping) in real world environments may be high (e.g. leading to trips, slips and misplaced steps on three-dimensional foot fall locations which are uneven, slippery or insecure). For this reason, simple measures of percentage of targets missed may have insufficient resolution to identify impairments or capture changes due to walking aides or other interventions. Further, if one in every 9-10 step adjustments completely miss a target landing location then this translates to a high number of opportunities for falls in daily life.

In a previous study [6] shortening steps were less accurately achieved than lengthening steps. We also saw this in our group of young healthy adults (see figure 2A). However, older healthy adults and stroke survivors had larger error magnitudes when lengthening steps rather than when shortening steps. Older healthy adults and stroke survivors undershot (mean error) all targets; with lengthening steps undershot more than shortened steps in both, support conditions. This tendency to “fall behind” the targets could be due to difficulty synchronising to the targets i.e. maintaining speed and accuracy of stepping simultaneously. Indeed, stroke survivors have been shown to lag in synchronising foot falls to auditory beats [28]. This is in contrast to young healthy adults who anticipate by foot falls leading beats [28–32]. Collectively, observations of both stroke survivors and older adults undershooting visual targets in this study and lagging auditory targets in other studies indicates difficulty anticipating and synchronising steps in time as well as space may be an effect of ageing as opposed to stroke specific difficulty.

Recruitment was as widely inclusive as possible while still ensuring sufficient mobility to safely take-part in the protocol. As a result participants are, at worst, of moderate mobility impairment. Participants with mild to moderate impairments may offer insight into the aspects of foot-placement control which are most vulnerable after stroke – so one could hypothesize that people with more severe stroke would have even larger errors in foot-placement [1]. Further, mild to moderately impaired participants are likely to be those who are navigating community environments, challenging their mobility with the required foot-placement adaptability [2]. The results of this study provide insight into the mechanics of the difficulties they may face. The use of crutches, while not typical in treadmill walking, was used to understand how when upper body is supported by external stabilization accuracy of foot-placement differs and hence characterisation of the interplay between balance control and processes of controlling aiming of the foot to a target foot fall location. Foot deformities and other weakness and spasticity may all be additional factors which can affect control of foot-placement [7]. These factors all require further investigation in future studies.

The enforced consistency of speed of walking on the treadmill may have made the task of stepping to targets differentially challenging for participant groups. However, participants walked more slowly on the treadmill than over ground, which likely is caused by the increased metabolic cost of treadmill walking [22], and the fact that people slow walking when needing to adjust foot-placements [9,33], possibly due to attentional cost [33,34]. Specifically, healthy young adults may have sufficient control to be accurate without slowing down. However, the inability to slow down (as has been previously seen when stepping to targets over ground [9,34]) on a treadmill may have made the target stepping more challenging for stroke survivors. As both decreased walking speed and impoverished balance [35] are direct impairments of stroke, covarying for SSWS may have equalized the performance of the groups (yielding non-significant differences between groups) by accounting for deficits due to the effect of stroke. Care should be taken when extrapolating results from this study to walking conditions without treadmill walking, as walking speeds are not identical on treadmill and over ground in the recent study.

We expected walking aids (crutch support) to affect accuracy when aiming with the non-paretic leg more than with the paretic, as maintaining paretic stance has been suggested to be the most challenging for balance [36]. However, the results seen here are in line with the results by Nonnekes et al., (2010) showing a bilateral increase of stepping accuracy with crutch support in stroke survivors. This indicates control of foot-placement relies on bilateral organisation to maintain balance and, adjustments have to be made in both the stance and swing legs when aiming with the lower limb [37].

5 Conclusion

Stroke survivors missed more targets than healthy younger and older counterparts but foot-placement errors were similar between paretic and non-paretic legs and in all directions of step adjustments; indicating stroke survivors have overall greater difficulty adjusting foot-placement in response to the environment than healthy older and younger counterparts. Coinciding with this, crutch support reduced stepping error for all groups and in all directions of stepping adjustment. Indicating external balance support improves accuracy of foot aiming even in healthy young participants. Both older healthy adults and stroke survivors undershot targets indicating processes of ageing may reduce ability to synchronize foot placement with external/environmental imperatives. Undershooting errors were greater in magnitude for stroke survivors however, this failed to reach significance when accounting for effects of group differences in walking speed. Overall, these results highlight the importance of walking speed and balance control on the ability to aim the foot to safe footfall locations in the environment. External balance support can improve accurate control of foot placement and consideration and further investigation should be given to the effects of rehabilitation, which aims to increase walking speed if this may cause greater difficulties adjusting foot placement in response to the environment.

[1] S.F. Tyson, M. Hanley, J. Chillala, A. Selley, R.C. Tallis, Balance disability after stroke, Phys Ther. 86 (2006) 30–38. doi:10.1093/ptj/86.1.30.

[2] J.M. Blennerhassett, W. Dite, E.R. Ramage, M.E. Richmond, Changes in balance and walking from stroke rehabilitation to the community: a follow-up observational study, Arch Phys Med Rehabil. 93 (2012) 1782–1787. doi:10.1016/j.apmr.2012.04.005.

[3] J. Smith, A. Forster, J. Young, Use of the “STRATIFY” falls risk assessment in patients recovering from acute stroke, Age Ageing. 35 (2006) 138–143. doi:10.1093/ageing/afj027.

[4] C.D. MacKinnon, D.A. Winter, Control of whole body balance in the frontal plane during human walking, J Biomech. 26 (1993) 633–644. doi:10.1016/0021-9290(93)90027-C.

[5] M.A. Townsend, Biped gait stabilization via foot placement, J Biomech. 18 (1985) 21–38. doi:10.1016/0021-9290(85)90042-9.

[6] W. Hoogkamer, Z. Potocanac, J. Duysens, Quick foot placement adjustments during gait: direction matters, Exp Brain Res. (2015). doi:10.1007/s00221-015-4401-y.

[7] J.H. Nonnekes, P. Talelli, M. de Niet, R.F. Reynolds, V. Weerdesteyn, B.L. Day, Deficits underlying impaired visually triggered step adjustments in mildly affected stroke patients, Neurorehabil Neural Repair. 24 (2010) 393–400. doi:10.1177/1545968309348317.

[8] R.F. Reynolds, B.L. Day, Visual guidance of the human foot during a step, J Physiol. 569 (2005) 677–684. doi:10.1113/jphysiol.2005.095869.

[9] K.L.L. Hollands, T.A.A. Pelton, S. van der Veen, S. Alharbi, M.A.A. Hollands, A novel and simple test of gait adaptability predicts gold standard measures of functional mobility in stroke survivors, Gait Posture. 43 (2016) 170–175. doi:10.1016/j.gaitpost.2015.09.018.

[10] T.S. Kuan, J.Y. Tsou, F.C. Su, C.Z. Hong, Gait analysis of stroke patients with and without walking aids, in: 8th World Congr. Int. Rehabil. Med. Assoc. (Irma Viii), Pts 1-2, 1997: pp. 373–378.

[11] Z.S. Nasreddine, N.A. Phillips, V. Bedirian, S. Charbonneau, V. Whitehead, I. Collin, J.L. Cummings, H. Chertkow, The Montreal Cognitive Assessment, MoCA: a brief screening tool for mild cognitive impairment, J Am Geriatr Soc. 53 (2005) 695–699. doi:10.1111/j.1532-5415.2005.53221.x.

[12] J. Moll, R. de Oliveira-Souza, F.T. Moll, I.E. Bramati, P.A. Andreiuolo, The cerebral correlates of set-shifting: an fMRI study of the trail making test, Arq Neuropsiquiatr. 60 (2002) 900–905. doi:10.1590/S0004-282X2002000600002.

[13] K.O. Berg, S.L. Wood-Dauphinee, J.I. Williams, B. Maki, Measuring balance in the elderly: Validation of an instrument, Can. J. Public Heal. 83 (1992).

[14] J. Jonsdottir, D. Cattaneo, Reliability and validity of the dynamic gait index in persons with chronic stroke, Arch Phys Med Rehabil. 88 (2007) 1410–1415. doi:10.1016/j.apmr.2007.08.109.

[15] D.J. Gladstone, C.J. Danells, S.E. Black, The fugl-meyer assessment of motor recovery after stroke: a critical review of its measurement properties, Neurorehabil Neural Repair. 16 (2002) 232–240. doi:10.1177.

[16] J.M. Gregson, M. Leathley, A.P. Moore, A.K. Sharma, T.L. Smith, C.L. Watkins, Reliability of the Tone Assessment Scale and the modified Ashworth scale as clinical

tools for assessing poststroke spasticity, Arch Phys Med Rehabil. 80 (1999) 1013–1016. doi:10.1016/S0003-9993(99)90053-9.

[17] W.L. Bickerton, D. Samson, J. Williamson, G.W. Humphreys, Separating forms of neglect using the Apples Test: validation and functional prediction in chronic and acute stroke, Neuropsychology. 25 (2011) 567–580. doi:10.1037/a0023501.

[18] S.M. van der Veen, U. Hammerbeck, R.J. Baker, K.L. Hollands, Validation of gait event detection by centre of pressure during target stepping in healthy and paretic gait, J. Biomech. (2018). doi:10.1016/J.JBIOMECH.2018.07.039.

[19] R. van Swigchem, H.J.R.R. Van Duijnhoven, J. Den Boer, A.C. Geurts, V. Weerdesteyn, Deficits in motor response to avoid sudden obstacles during gait in functional walkers poststroke, Neurorehabil Neural Repair. 27 (2013) 230–239. doi:10.1177/1545968312462070.

[20] A. Schmid, P.W. Duncan, S. Studenski, S.M. Lai, L. Richards, S. Perera, S.S. Wu, Improvements in speed-based gait classifications are meaningful, Stroke. 38 (2007) 2096–2100. doi:10.1161/STROKEAHA.106.475921.

[21] K. Berg, S. Wood-Dauphinée, J.I. Williams, D. Gayton, Measuring balance in the elderly: preliminary development of an instrument., Physiother. Canada. (1992). doi:10.1016/j.archger.2009.10.008.

[22] S.A. Fraser, O. Dupuy, P. Pouliot, F. Lesage, L. Bherer, Comparable Cerebral Oxygenation Patterns in Younger and Older Adults during Dual-Task Walking with Increasing Load, Front Aging Neurosci. 8 (2016) 240. doi:10.3389/fnagi.2016.00240.

[23] K. Uemura, T. Oya, Y. Uchiyama, Effects of speed and accuracy strategy on choice step execution in response to the flanker interference task, Hum Mov Sci. 32 (2013) 1393–1403. doi:10.1016/j.humov.2013.07.007.

[24] H.P. von Schroeder, R.D. Coutts, P.D. Lyden, E. Billings, V.L. Nickel, Gait parameters following stroke: a practical assessment., J. Rehabil. Res. Dev. 32 (1995) 25–31.

[25] M.J. Bancroft, B.L. Day, The Throw-and-Catch Model of Human Gait: Evidence from Coupling of Pre-Step Postural Activity and Step Location, Front Hum Neurosci. 10 (2016) 635. doi:10.3389/fnhum.2016.00635.

[26] R. Moraes, F. Allard, A.E. Patla, Validating determinants for an alternate foot placement selection algorithm during human locomotion in cluttered terrain, J Neurophysiol. 98 (2007) 1928–1940. doi:10.1152/jn.00044.2006.

[27] A.R. Den Otter, A.C.H. Geurts, M. de Haart, T. Mulder, J. Duysens, Step

characteristics during obstacle avoidance in hemiplegic stroke , Exp. Brain Res. . 161 (2005) 180–192. doi:10.1007/s00221-004-2057-0.

[28] M. Roerdink, C.J. Lamoth, J. van Kordelaar, P. Elich, M. Konijnenbelt, G. Kwakkel, P.J. Beek, Rhythm perturbations in acoustically paced treadmill walking after stroke, Neurorehabil Neural Repair. 23 (2009) 668–678. doi:10.1177/1545968309332879.

[29] P.J.M. Bank, M. Roerdink, C.E. Peper, Comparing the efficacy of metronome beeps and stepping stones to adjust gait: steps to follow!, Exp Brain Res. 209 (2011) 159–169. doi:10.1007/s00221-010-2531-9.

[30] G.C. McIntosh, S.H. Brown, R.R. Rice, M.H. Thaut, Rhythmic auditory-motor facilitation of gait patterns in patients with Parkinson’s disease, J Neurol Neurosurg Psychiatry. 62 (1997) 22–26. doi:10.1136/jnnp.62.1.22.

[31] M. Roerdink, P.J. Bank, C.L. Peper, P.J. Beek, Walking to the beat of different drums: practical implications for the use of acoustic rhythms in gait rehabilitation, Gait Posture. 33 (2011) 690–694. doi:10.1016/j.gaitpost.2011.03.001.

[32] M. Roerdink, C.J. Lamoth, G. Kwakkel, P.C. van Wieringen, P.J. Beek, Gait coordination after stroke: benefits of acoustically paced treadmill walking, Phys Ther. 87 (2007) 1009–1022. doi:10.2522/ptj.20050394.

[33] M. Mazaheri, M. Roerdink, R.J. Bood, J. Duysens, P.J. Beek, C.L. Peper, Attentional costs of visually guided walking: effects of age, executive function and stepping-task demands, Gait Posture. 40 (2014) 182–186. doi:10.1016/j.gaitpost.2014.03.183.

[34] C.L. Peper, M.J. de Dreu, M. Roerdink, Attuning one’s steps to visual targets reduces comfortable walking speed in both young and older adults , Gait Posture . 41 (2015) 830–834. doi:10.1016/j.gaitpost.2015.02.016.

[35] T. Krasovsky, A. Lamontagne, A.G. Feldman, M.F. Levin, Effects of walking speed on gait stability and interlimb coordination in younger and older adults, Gait Posture. 39 (2014) 378–385. doi:10.1016/j.gaitpost.2013.08.011.

[36] C.M. Said, P.A. Goldie, A.E. Patla, E. Culham, W.A. Sparrow, M.E. Morris, Balance during obstacle crossing following stroke, Gait Posture. 27 (2008) 23–30. doi:10.1016/j.gaitpost.2006.12.009.

[37] R.F. Reynolds, B.L. Day, Rapid visuo-motor processes drive the leg regardless of balance constraints, Curr Biol. 15 (2005) R48-9. doi:10.1016/j.cub.2004.12.051.

Figure 1. A schematic representation of A) the treadmill, with the projector and crutches installed next to the treadmill, B) the target position on the treadmill, white squares are preferred target positions, yellow targets represent the targets requiring adaptations to foot-placement (shortening, lengthening, widening and narrowing) with grey shadow of where the target would have been for preferred foot location. The blue line represents the centre of pressure trajectory of one participant with the red asterisk representing centre of pressure at midstance. The green circle represents the centre of foot at time of midstance with the representation of the foot (larger circle around the centre of foot). AP, adjustments in anterior-posterior direction; ML, adjustments in medio-lateral direction.

Figure 2. Bars represent mean absolute foot-placement error in A) anterio-posterior direction and B) medio-lateral direction. Filled bars represent participants where unsupported and hatched bars represent when crutch support was provided. Error bars represent standard error of the mean. * represents a p value <0.05 and **a value<0.001.

Figure 3. Bars represent bias in anterio-posterior direction, negative values represent undershooting and positive overshooting. A) Represents the interaction effect between group and side, where B) represent the interaction between group and stepping direction. Error bars represent standard error of the mean.

Figure 4. Bars represent the percentage of missed targets. Error bars represent the standard error of the mean, ** represents a p value<0.001.

Table 1. Participant demographics and clinical characteristics represented in mean± SD unless specified differently.

Table 2: Data table, Means and standard error of the mean is represented for the different adaptation steps and conditions per group in cm.