Effects of Assistive Device Use on Lower Limb Kinematics during Gait in Individuals with Huntington's Disease

A Senior Honors Thesis

Presented in Partial Fulfillment of the Requirements for graduation *with research distinction in* Biology in the undergraduate colleges of The Ohio State University

by
Yi Ding

The Ohio State University

Abstract

Physical therapists typically prescribe ambulatory assistive devices (ADs) for people with Huntington's Disease (HD) to improve their stability and prevent falls. Little is known, however, about how using these ADs affect gait patterns. This study was conducted to determine the effects of using ADs on lower extremity joint kinematics (i.e., peak angular excursions) during gait. Nineteen individuals with HD (mean age=49.8±11) who were able to ambulate 10 meters without assistance participated in the study. Subjects with reflective silver markers placed on body landmarks were videotaped while they ambulated 20 ft for 4 trials under each of six conditions using: 1) No AD; 2) a cane; 3) a standard walker; 4) a two-wheeled walker (2WW); 5) a three-wheeled walker (3WW); and 6) a 4-wheeled walker (4WW). The film was analyzed using Peak Motus motion analysis system and peak angular excursions were calculated for the swing and stance phases at the knee and hip. Data was analyzed using repeated measures ANOVA and post hoc pairwise comparisons. The devices did not have a significant impact on the knee peak angular excursions during stance nor swing phase. At the hip during the stance phase, peak excursions were significantly increased for the 3WW compared to no AD. The 4WW also had a larger peak excursion at the hip, however the improvements were not significant compared to no AD. The standard walker in all cases had a significantly decreased excursion compared to no AD. Across all phases for both the knee and the hip, the 3WW and 4WW performed significantly better than the standard walker. Our results indicate that the use of the 3WW and 4WW produced greater knee and hip motions during walking. These findings may explain improved velocity and stride length measures with use of the 3WW and 4WW in the same subjects that were previously reported by our lab. Due to its greater stability and better maneuverability, we recommend that clinicians prescribe the 4WW over a 3WW and standard walker for fall prevention in individuals with HD.

Introduction

Huntington's disease (HD), also known as Huntington's chorea, is a neurodegenerative disease that is inherited as an autosomal dominant disease. It affects roughly 30,000 people in the United States, with another 150,000 to 200,000 who are at risk for inheriting the disease (Family Caregiver Alliance). Disease onset typically occurs in the late 30s to early 50s, which means that affected individuals will have a high chance of already having children before finding out about the disease (Klug et al., 2009). The children of individuals with HD will have a 50% chance of inheriting the gene that causes the disease, which further keeps the disease prevalent in the population.

HD affects the motor neurons in the basal ganglia of the brain, which causes gradual reduction of muscle control and motor coordination, and results in the random involuntary jerking movements called chorea. Patients also show signs of decreased cognitive abilities and spatial skills, and may have problems controlling their emotions (Warby et al., 2010). In normal conditions the huntingtin (Htt) protein controls the processes that are essential to the survival of neurons in the brain (Klug et al., 2009). However in patients with HD, the genes encoding the huntingtin protein (in chromosome 4) can carry many more trinucleotide CAG repeats that will result in the addition of polyglutamines to the protein (Lin et al., 2001). The onset of the disease occurs much earlier with more trinucleotide repeats in the genome, and the disease will have a faster progression (Delval et al., 2006).

One of the most visible changes in individuals with HD is in their walking (i.e., gait) patterns. The degradation of voluntary muscle control leads to shorter and slower steps, and these stride patterns increasingly vary between each phase of the gait cycle as

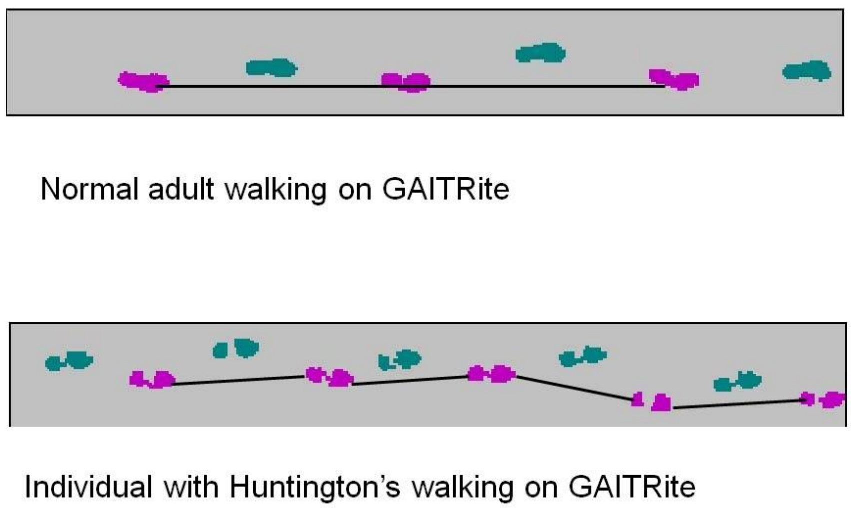

Figure 1: Comparison of gait patterns between a healthy adult (top panel) and an individual with HD (bottom panel). Notice the path deviation to the right and the variability in step lengths of the individual with HD.

the severity of the disease increases (Reynolds et al., 1999; Hausdorff, 2007). Figure 1 shows the difference between a normal gait pattern and a pattern belonging to an adult with HD. The footfalls fail to remain on a straight line as they do for the normal gait. More importantly, these gait deviations along with balance problems contribute to an increased rate of falls in individuals with HD compared to age-matched healthy controls (Grimbergen et al., 2008; Busse et al., 2009). Falls are very dangerous to people with HD because they can lead to serious injuries and hospitalization (Hausdorff, 2007; Grimbergen et al., 2008). Studies have found that a history of falls and the fear of falling causes people to take fewer steps and walk slower as precautions to prevent falls, thereby exacerbating gait problems caused by HD (England, 2007; Busse et al., 2009). As balance confidence levels drop, people with HD will become less active (England, 2007; Busse et al., 2009). Thus, an important aspect of the management of gait disorders in individuals with HD is to promote gait stability by prescribing ambulatory assistive devices (ADs) to help them maintain balance and allow safe and efficient walking.

Physical therapists may prescribe ambulatory ADs such as canes and walkers for individuals with HD to prevent fall-related injuries, motor coordination losses, and deconditioning from inactivity. Overall, very little is known about how using these assistive devices affect the gait patterns of individuals with HD. Therefore, this study was conducted to determine the effects of assistive devices commonly used by individuals with HD on lower extremity joint kinematics (i.e., angular excursions). It was hypothesized that the lower extremity joint kinematics would be better or more normal when individuals ambulated with devices that were easy to handle and provided greater amounts of stability (i.e., wheeled walkers) compared to ambulation with canes, standard walkers without wheels, or no assistive device. Information obtained from this study will help physical therapists make decisions regarding which assistive devices should be prescribed for individuals with HD.

Methodology

Subjects

Nineteen subjects who attended the Ohio State University Movement Disorders Clinic volunteered to participate in the study. There were 14 females and 5 males in the sample group and the participants' ages varied from 24 to 66 (Table 1). These participants had no prior diagnoses that would affect their lower extremities, and they were capable of ambulating a distance of at least 10 meters without other assistance. The sample group had scores on the motor section of the Unified Huntington's Disease Rating Scale (UHDRS) that ranged from 11 to 58. The UHDRS is a scale that quantifies the severity of disease symptoms by analyzing the motor, cognitive, behavioral, and functional abilities of the person with HD. The motor section, specifically, assesses eye movements, tongue protrusion, chorea, rapid alternating movements, dystonia, bradykinesia, gait, and tandem walking. Each of the 31 items on the motor section is scored from 0-4 with 0 being no impairment and 4 being severe impairment; the maximum possible score is 124 (Rao et al., 2009). Thus, the higher the motor score, the more severe the disease symptoms are (Huntington Study Group, 1996). Prior to any testing, an informed consent was obtained from each subject.

Table 1: Subject Characteristics

Subject Characteristics (n=19)	Mean (SD)	Range
Age (yrs)	49.84 (10.96)	24-66
Years Post Diagnosis	4.7 (3.9)	1.0-14
UHDRS Motor Scale Score	38.37 (13.57)	11.0-58

Instruments

The participants used five different assistive devices throughout the study. They were a standard straight cane, a standard walker with no wheels, a two-wheeled walker (2WW) with two fixed front wheels (Figure 2), a three-wheeled walker (3WW) in a tricycle configuration and one front swivel caster, and a four-wheeled walker (4WW) with two front swivel-casters (Figure 3). The assistive devices (except 3WW) were all adjustable to fit the height needs of the participants to better achieve a normal walking posture.

Figure 2: Two wheeled walker

Figure 3: Four wheeled walker

Kinematic data (i.e., lower extremity joint movements in the sagittal view) were obtained by videotaping the subjects with a single camera while walking on the GAITRite walkway (i.e., a carpet that measured 16 ft long by 3 ft wide containing embedded pressure sensors that detect footfalls during ambulation). The data were then analyzed using a Peak Modus motion analysis computer system (version 7.2.6).

Procedure

The motor section of the UHDRS was administered and demographic information (i.e., age, sex, number of years since symptom onset, and medications) was obtained for each subject by one of the researchers. Silver reflective tapes were placed on the right side of the body at the temple, humeral head, the greater trochanter of the hip, the lateral femoral condyle, and the lateral malleolus prior to videotaping (Kelleher et al., 2010). After the silver tape had been applied, subjects were asked to walk at a normal pace across a GAITRite walkway that collected spatial and temporal gait parameter data for another part of this study. White tapes were placed 5 feet before and at the end of the walkway to control for acceleration and deceleration of gait speeds when the subjects walked across the carpet. A camera was set up on the side (perpendicular to) the carpet about 10 feet away to capture each subject's movement across the middle region of the carpet which was approximately 10-15 steps.

The subject was first asked to walk the entire length of the walkway without any device (no AD) for four trials, the second and fourth trials were filmed. The first trial was designed as a practice run to acquaint the subject to the research setup. The subjects was taught by a physical therapy student or physical therapist how to use each device and were allowed to practice with the device in the open area of the room until they were comfortable with each device. Next the subjects walked across the walkway using each assistive device for four trials each with rests between trials as needed. The sequence of the use of assistive devices was random throughout the study to eliminate the influence of fatigue on outcome.

Data Analysis

The video was uploaded into the Peak Motus motion analysis system and the markers were digitized. The markers allowed the computer to calculate the body angles at the hip, and the knee throughout each gait cycle (Figure 4). Normal gait cycle has eight phases: initial contact, loading response (yield), midstance, terminal stance, preswing, initial swing, midswing, and terminal swing. Only initial contact, midstance, and liftoff were marked on the computer.

Angular excursions (i.e., amount of joint angle movement in degrees) of the knee and hip were calculated using two methods. The first method calculated angular excursions for Initial Contact (IC) to Midstance (MS), MS to Liftoff (LO) and LO to IC phases of the gait cycle by subtracting the angle value of the next gait phase from the previous one and taking an absolute value of it. This set of data provided information about joint angle changes relative to specific phases of the gait cycle to determine whether using the devices have affected joint movements. The second set of data analyzed were the peak angular excursions achieved at certain stages of gait. Peak excursions were calculated by identifying the maximum and minimum angle values throughout the gait cycle and subtracting one from another and taking an absolute value of the difference. This is a more precise calculation of angular excursions compared to gait phase excursions because peak values are based on the maximum flexion or extension that is achieved by the joint. The angle values included Initial Contact Maximum Extension (ICME), Yield Flexion (YF), Mid-stance Maximum Extension (MSE), and Lift off Maximum Flexion (LOMF) for the knee; Initial Contact Maximum Flexion (ICMF) and Lift Off Maximum Extension (LOME) for the hip. Between the two methods of analyzing excursion, the peak angular excursion, although similar to the excursion values from the phases of gait, is much more precise. It's widely used by others as a measurement of gait (Wang et al, 2009; Kelleher et al, 2010). Only the peak angular excursion data is presented in the results section. The results for the gait phase angular excursion data can be found in Appendix 1.

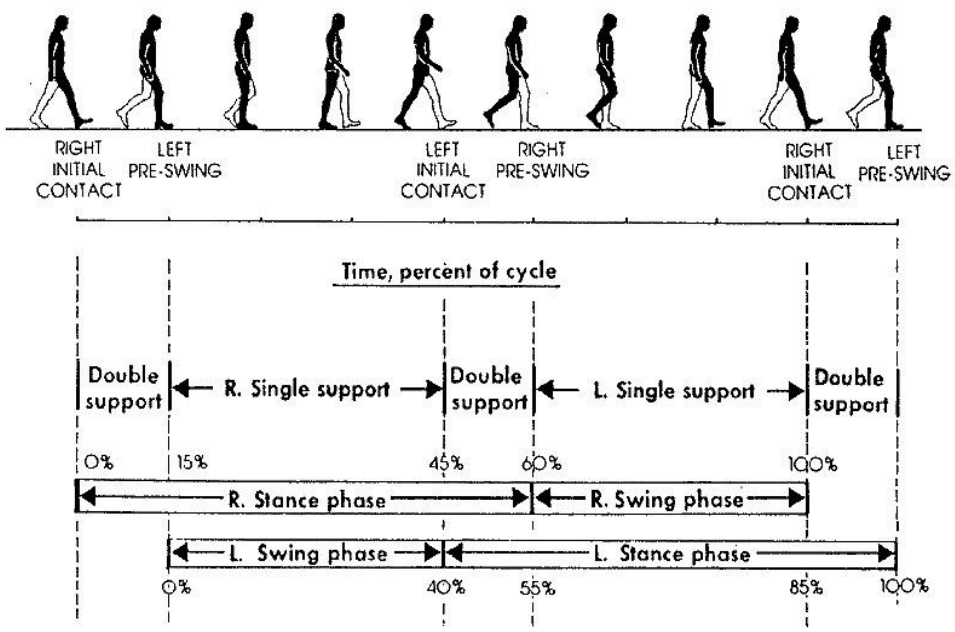

Figure 4: Phases of the Gait Cycle. The gait cycle is divided into stance and swing phases.

SPSS (Version 17) was used for all statistical analyses. Mean angular excursion and peak angular excursion data were compared across all conditions using SPSS one-way repeated measures ANOVA followed by post hoc pairwise comparisons tests. For all comparisons, a two-sided $p \leq 0.05$ level of significance was used. (Due to a missing 3WW trial, there is an insignificant difference between the actual 3WW peak angular excursions value and the comparison value between 3WW and other devices).

Results

Knee Peak Angular Excursions during early Stance phase

During the early portion of the stance phase (ICME-YF), the knee showed an average peak angular excursion of 9.959° ±1.566° for the no AD condition (Table 2). This represents the brief flexion that occurs at the knee when the foot first lands after initial contact, also known as loading response (Figure 4). The amount of peak flexion at the knee was decreased using the standard walker (-4.311° ±1.538°, $P \leq 0.012$) compared to no AD. The peak excursion for all other devices was not significantly different than the no AD condition. The 2WW (2.273° ±0.835°, $P \leq 0.014$), the 3WW (2.292° ±0.991°, $P \leq 0.027$), and the 4WW (3.335° ±0.049°, $P \leq 0.005$) had all showed a larger angular excursion compared to the standard walker (Figure 5).

Table 2: Knee early Stance Phase Maximum Yield Flexion (Loading Response)

Condition	Mean ±Standard Deviation (SD)
No Assistive Device (AD)	9.959 ± 1.566
Cane	8.091 ± 1.110
Standard Walker	5.648 ± 0.485
2 Wheeled Walker	7.921 ± 1.026
3 Wheeled Walker	9.189 ± 1.492
4 Wheeled Walker	8.983 ± 1.198

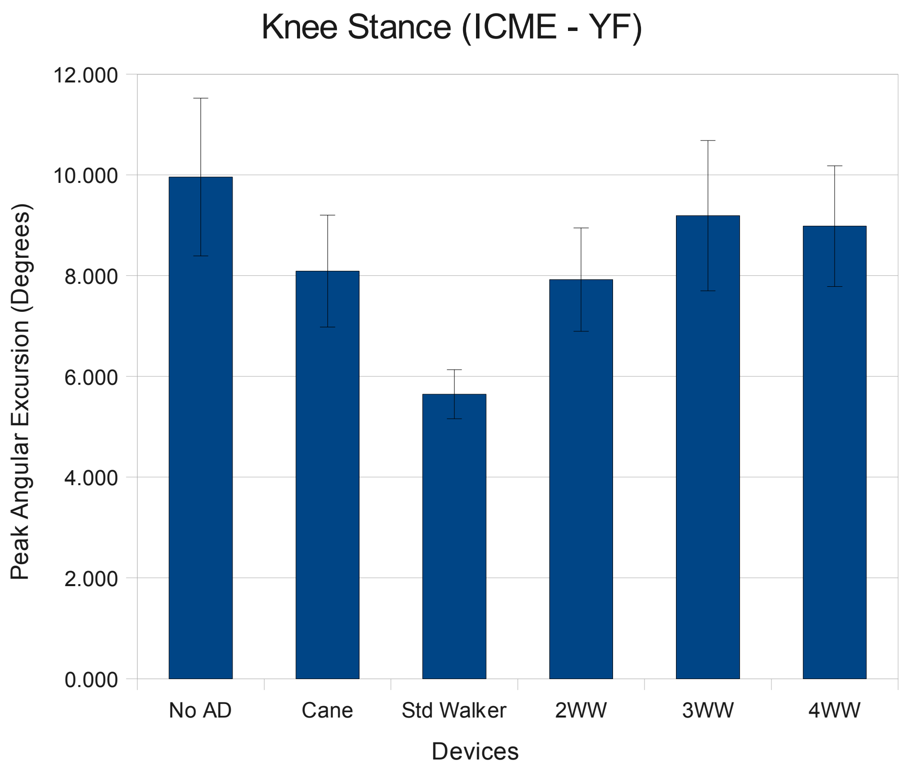

Figure 5: Knee Peak Angular Excursion at early Stance Phase (ICME-YF)

Knee Peak Angular Excursions during Swing phase

During the swing phase (LOMF-ICME), the standard walker had a significantly decreased peak angular excursion (-5.926° ±2.656°, $P \leq 0.039$) compared to the no AD condition (Table 3). This means that the subjects had a dramatically decreased amount of extension at the knee through the swing phase when using the standard walker. In contrast, the 4WW (6.860° ±2.178°, $P \leq 0.006$) and the 3WW (6.932° ±2.353°, $P \leq 0.009$) both showed improvements in peak excursion compared to the standard walker condition (Figure 6). No other condition comparisons were significantly different.

Table 3: Knee Swing Phase Maximum Extension	
Condition	**Mean ±Standard Deviation (SD)**
No Assistive Device (AD)	61.328 ± 2.935
Cane	59.714 ± 2.625
Standard Walker	55.402 ± 2.751
2 Wheeled Walker	59.929 ± 2.295
3 Wheeled Walker	63.661 ± 2.647
4 Wheeled Walker	62.262 ± 1.938

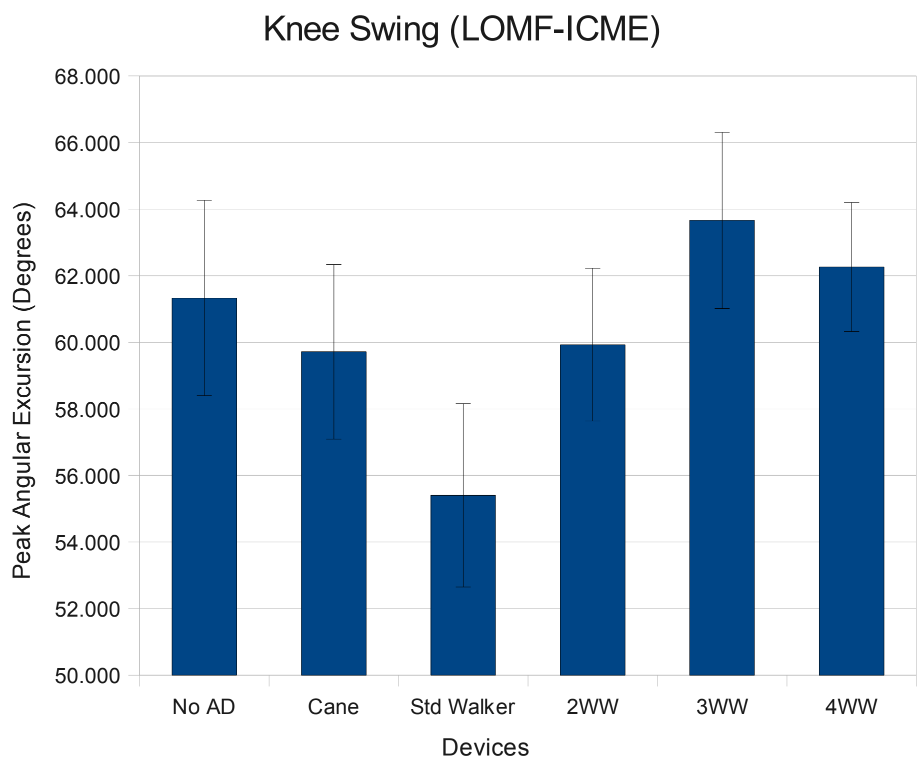

Figure 6: Knee Peak Angular Excursion at Swing Phase

Hip Peak Angular Excursion during Stance phase

Compared to the no AD condition, the standard walker had a significant decrease in excursion (-3.916° ±1.451°, $P \leq 0.015$) (Table 4). The 3WW had a larger excursion

(2.292° ±0.991°, $P \leq 0.033$) compared to no AD condition (Figure 7). Compared to the standard walker, 2WW (4.100° ±1.318°, $P \leq 0.006$), 3WW (6.207° ±1.205°, $P \leq 0.000$), and 4WW (5.526° ±1.235°, $P \leq 0.000$) all had significantly larger excursion. The 3WW also performed better than the cane (2.598° ±1.077°, $P \leq 0.027$).

Table 4: Peak Angular Excursion for the Hip During Stance Phase (ICMF-LOME)

Condition	Mean ±Standard Deviation (SD)
No Assistive Device (AD)	38.050 ± 1.601
Cane	37.745 ± 1.587
Standard Walker	34.134 ± 1.517
2 Wheeled Walker	38.234 ± 2.022
3 Wheeled Walker	40.662 ± 1.420
4 Wheeled Walker	39.660 ± 1.698

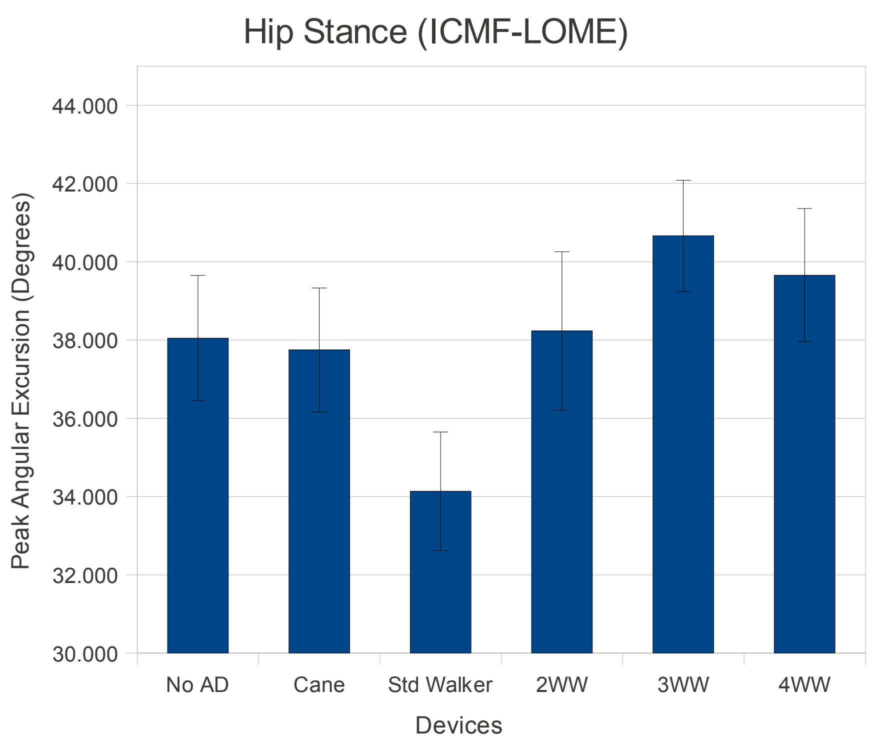

Figure 7: Hip Peak Angular Excursion at Stance Phase

Hip Peak Angular Excursion during Swing Phase

The peak angular excursion data at the hip during swing phase are shown in Table 5. Between LOME-ICMF, the swing phase, the peak angular excursion for the standard walker (Figure 8) was significantly decreased (-4.120° ±1.355°, $P \leq 0.007$) compared to no AD. The 4WW and 3WW both performed better than the standard walker (5.290° ±1.028°, $P \leq 0.000$ and 5.673° ±1.099°, $P \leq 0.000$, respectively) and cane (1.891° ±0.746°, $P \leq 0.021$and 2.327° ±0.743°, $P \leq 0.006$, respectively). The 2WW also performed better than the standard walker (3.983° ±1.030°, $P \leq 0.001$).

Table 5: Hip Peak Flexion during Swing Phase (LOME-ICMF)

Condition	**Mean ±Standard Deviation (SD)**
No Assistive Device (AD)	36.549 ± 1.597
Cane	35.827 ± 1.454
Standard Walker	32.429 ± 1.384
2 Wheeled Walker	36.411 ± 1.744
3 Wheeled Walker	38.479 ± 1.495
4 Wheeled Walker	37.718 ± 1.555

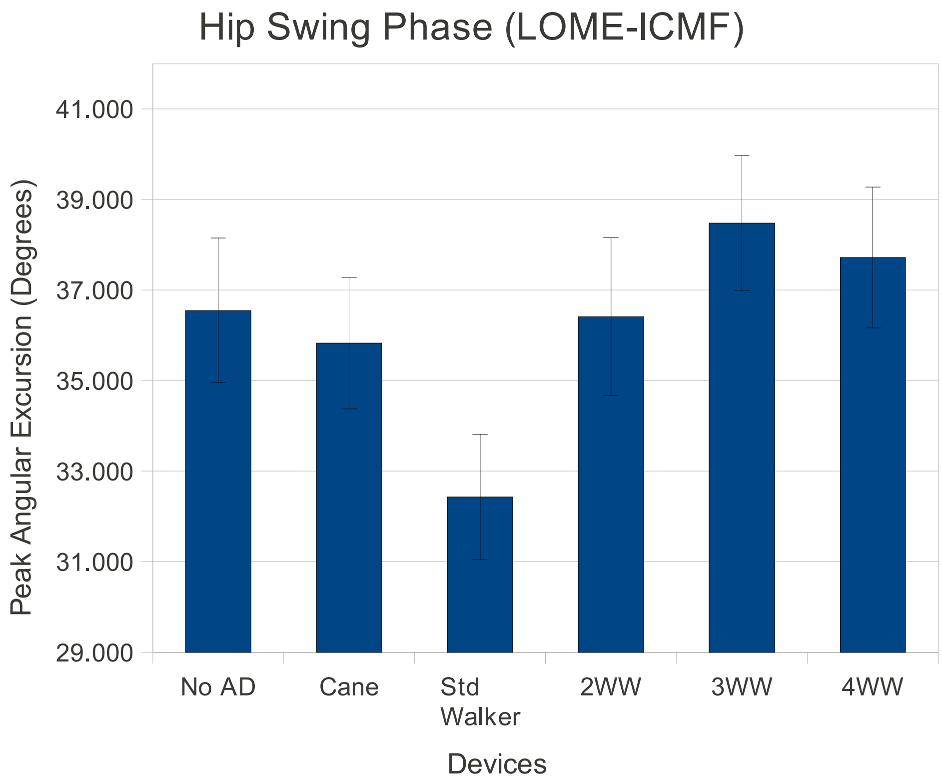

Figure 8: Hip Peak Angular Excursion during Swing Phase

Irregularity Count

During data analysis some abnormal patterns of joint kinematics were observed at both the knee and the hip. Often subjects would display no YF (Figure 9), indicating that the lower limb was held abnormally stiffly in extension instead of slightly flexing after initial contact. Yield phase was absent with the highest per subject frequency during gait with the standard walker. Yield was present with the greatest frequency when using the 4WW, even more so than in the no AD condition.

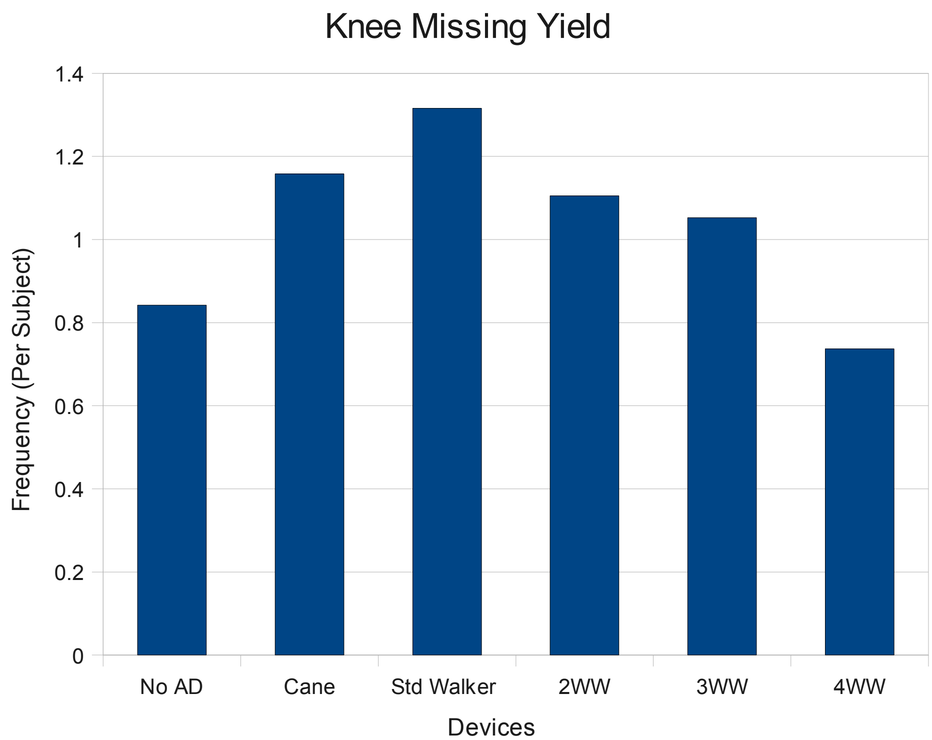

Figure 9: Frequency of Missing Yield at the Knee using Specific Devices

In addition, multiple peaks instead of one peak in knee or hip flexion were found between IC and MS, which indicated that the knee or hip was not stable and exhibited instability during the initial loading of the limb. The standard walker again displays the highest frequency of instability both at the knee (Figure 10) and the hip (Figure 11).

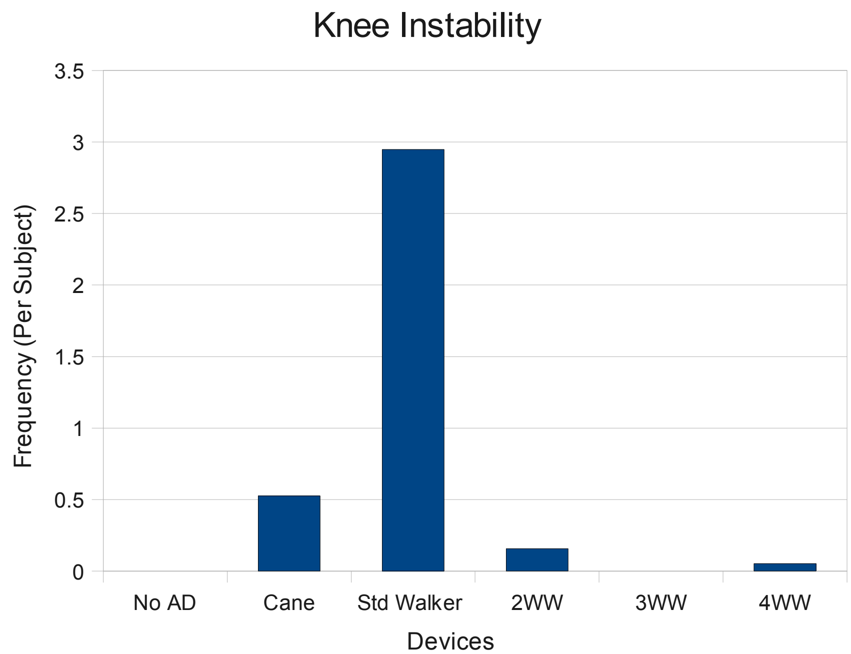

Figure 10: Frequency of Buckling at the Knee across Devices

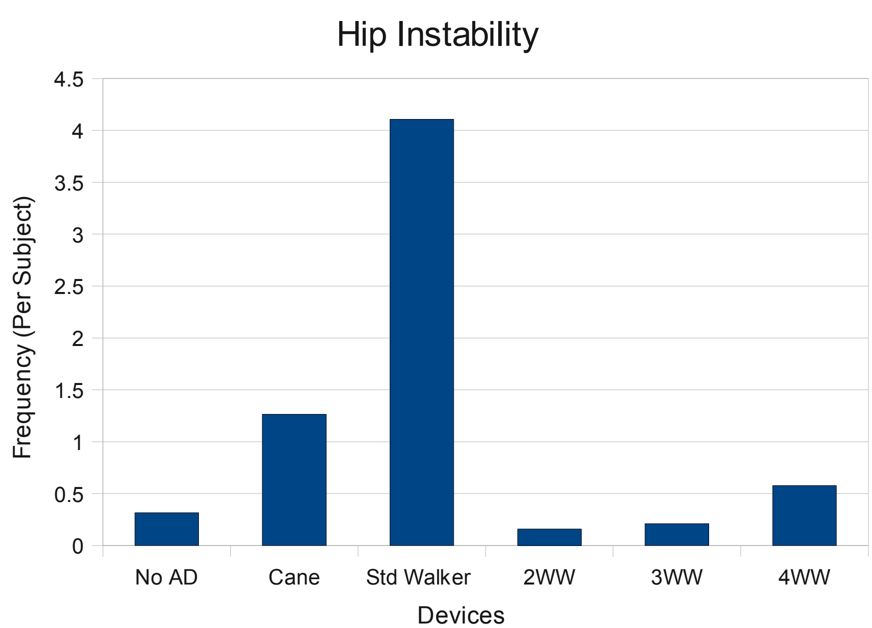

Figure 11: Frequency of Buckling at the Hip across Devices

Discussion

The loss of motor coordination is a serious issue in individuals with HD and could lead to falls. This study showed that the 4WW produced the safest and most efficient gait pattern as measured through joint peak angular excursions. These results were possibly due to the 4WW providing better stability and ease of mobility than other commonly used assistive devices. Thus, the gait of individuals with HD could be improved with the use of the 4WW and might lead to a reduction in the numbers of falls.

The peak angular excursion data largely supported the data from excursion at the phases of gait. The hip had improved amplitudes of movements during the both the stance and swing phases using the 4WW and 3WW compared to the no AD condition, although the differences did not reach significance (except for the 3WW during stance phase). The differences between the standard walker and the 4WW/3WW at the hip during swing were, however, significant. These findings suggest that individuals with HD may have improved foot clearance during ambulation with 4WW/3WW devices, which may decrease the incidences of stumbles, trips, and falls. More importantly, the normal person has a peak hip excursion during swing of 43.3° ±6.6° (Perry, 2010). Thus compared to the no AD (36.549° ±1.597°), the 3WW (38.479° ±1.495°) and 4WW (37.718° ±1.555°) both had a larger hip peak excursion. This means that the 3WW and 4WW were able to improve the hip excursion more toward the normal value compared to all other devices. At the knee, both 3WW and 4WW also had a larger peak excursion compared to no AD during the swing phases, but they were also not significant. Similar to the hip, the standard walker was again out performed by the 3WW and the 4WW. The normal person has a peak knee excursion during swing of 59.6° ±5.2° (Perry, 2010). The 3WW (63.661° ±2.647°) and 4WW (62.262° ±1.938°) both fall into this range, which means the subjects knee during swing is normal. The slight larger excursion than the no AD condition (61.328° ±2.935°) implies that the 3WW and 4WW improves the range of motion even more, which leads to a better gait pattern. These findings suggest that the 3WW and 4WW walkers are safer and should be recommended more often to patients with HD than the standard walker. Interestingly, during loading response (ICME-YF) at the knee, the normal value of yield is 18.8° ±5.2° (Perry, 2010). The no AD condition had a ICME-YF excursion of 9.959° ±1.566° and the only device capable of reaching that was the 3WW (9.189° ±1.492°). Since the standard walker yielded the least amount of excursion, it can be concluded that the subjects were exhibiting a stiff gait at the knee to improve stability. Despite the devices not being able to improve their knee flexion during loading response, the stability that's provided by a stiffer knee is better than an unstable knee.

Overall the 3WW and the 4WW produced the least numbers of abnormalities in joint kinematics (i.e., "buckling" of the knee or hip or missing YF), whereas the standard walker produced the greatest numbers of abnormalities. When analyzed, the presence of abnormalities did not correlate with UHDRS motor scores, suggesting that they could not be attributed to greater disease severity. An explanation for these findings is that the 3WW and the 4WW provided greater stability and ease of use than the other devices and therefore subjects needed less motor control to maintain lower extremity joint stability during initial limb loading and weight bearing. However, when using the standard walker, the subjects were required to pick up the walker in order to advance it, which dramatically decreased their base of support. Due to the greater demands for joint stability, they may have locked their knee in a stiff extended position during limb loading (loading response) to prevent knee collapse. Knee and hip instability, or “buckling”, may have been due to difficulties with dual tasking (i.e., coordinating the advancement of the walker with walking) that people with HD are known to have. In a study analyzing the relationship between visual cues and gait patterns, it was concluded that frequent training with visual cues helps people with Parkinson's Disease to improve their gait (Sidaway et al., 2006). When compared to simple visual cues, however, walking with a standard walker is a much more complex task that may cause gait to worsen. In addition, some people with HD have dystonias that cause abnormal knee flexion during stance which may have been exacerbated by the complex demands of using the standard walker (Louis et al, 1999).

The 3WW was excellent in its larger wheels, which could be better when walking over rougher terrain. However, given its tricycle design, people with longer limbs are more likely to take a smaller step in order to not hit the device while walking. Walking with the 2WW did not improve hip movements to the extent that the 3WW and 4WW did. One explanation for this finding is that the back legs of the 2WW did not articulate smoothly with the ground and kept tripping over uneven surfaces. This made it harder for the subjects to keep a steady forward progression. Because the 4WW provided a bigger base of support and better mobility, the 4WW is ultimately more favorable.

Conclusions

Taken together our findings indicate that ambulation with the 4WW produced the most normal and safe hip and knee joint kinematics compared to other commonly prescribed devices, and in some instances it improved the joint kinematics over the no AD condition. And despite the differences in excursion between 4WW and no AD being non-significant, the improvement in stability is undeniable. Based on our findings, physical therapists should prescribe the 4WW more often than other devices for people with HD who are at risk or are falling.

Acknowledgements

I would like to thank my two advisors Dr. Anne D. Kloos and Dr. Deborah A. Kegelmeyer for their guidance and support. I would also like to thank the Vaughan family for their financial support of this project.

References

- Huntington's Disease Facts. Retrieved on March 8 2009. http://www.caregiver.org/caregiver/jsp/content_node.jsp?nodeid=574
- Huntington's Disease: Hope Through Research. Retrieved on April 30 2009.

http://www.ninds.nih.gov/disorders/huntington/detail_huntington.htm

- Huntington Study Group. Retrived on April 21, 2010. http://www.huntington-study-group.org/Education/tabid/91/Default.aspx
- Busse, M. E., Wiles, C. M., and Rosser, A. E. (2009). Mobility and falls in people with Huntington's disease. *Journal of Neurology. Neurosurgery. Psychiatry*, 80: 88-90.
- Delval, A., Krystkowiak, P., Blatt, JL., Labyt, E., Dujardin, K., Destee, A., Derambure, P., and Defebvre, L. (2006). Role of hypokinesia and bradykinesia in gait disturbances in Huntington's disease: A biomechanical study. *Journal of Neurology*, 253(1): 73-80.
- England, S. A., Granata, K. P. (2007). The influence of gait speed on local dynamic stability of walking. *Gait & Posture*, 25: 172-178.
- Grimbergen, Y. A. M., Knol, M. J., Bloem, B. R., Kremer, B. P. H., Roos, R. A. C., and Munneke, M. (2008). Falls and gait disturbances in Huntington's disease. *Movement Disorders*, 23(7): 970-976.
- Hausdorff, J. M. (2007). Gait dynamics, fractals and falls: finding meaning in the stride-to-stride fluctuations of human walking. *Human Movement Science*, 26: 555-589.
- Huntington Study Group (1996). Unified Huntington’s Disease Rating Scale: Reliability and Consistency. *Movement Disorders,* 11(2), 136-142.
- Kelleher KJ, et al. The effect of textured insoles on gait patterns of people with multiple sclerosis. Gait Posture (2010), doi: 1016/j.gaitpost.2010.03.008

- Kloos, AD., Kegelmeyer, DK., Kostyk, SK. The effects of assistive devices on gait measures in Huntington's disease. Neurotherapeutics 2009; 6 (1): 209-210.
- Klug, W., Cummings, M., Spencer, C., and Palladina., M. (2009). *Concepts of Genetics* 9th edition. San Francisco CA: Pearson Benjamin Cummings.
- Lin, CH., Tallaksen-Greene, S., Chien, WM., Cearley, J. A., Jackson, WS., Crouse, A. B., Ren, S., Li, XJ., Albin, R. L., and Detloff, P. J. (2001). Neurological abnormalities in a knock-in mouse model of Huntington's disease. *Human Molecular Genetics*, 10(2): 137-144.
- Louis, E., Lee, P., Quinn, L., Marder, K. (1999). Dystonia in Huntington's disease: prevalence and clinical characteristics. *Mov Disord,* 14: 95-101.
- Perry, J., Burnfield, J. M. (2010). Gait Analysis: Normal and Pathological Function (2nd ed.). California: SLACK Incorporated.
- Reynolds, N. C., Myklebust, J. B., Prieto, T. E., and Myklebust, B. M. (1999). Analysis of gait abnormalities in Huntington disease. *Academy of Physical Medicine and Rehabilitation*, 80: 59-65.
- Rao, A. K., Muratori, L., Louis, E. D., Moskowitz, C. B., Marder, K. S. (2009). Clinical measurement of mobility and balance impairments in Huntington's disease : Validity and responsiveness. *Gait & Posture*, 29: 433-436.
- Sidaway, B., Anderson, J., Danielson, G., Martin, L., Smith, G. (2006). Effects of Long-Term Gait Training Using Visual Cues in an Individual With Parkinson Disease, 86(2) : 186-194.
- Warby, S. C., Graham, R. K., Hayden, M. R. (2010) Huntington Disease.

http://www.ncbi.nlm.nih.gov/bookshelf/br.fcgi?book=gene&part=huntington

Appendix 1

Gait Phase Angular Excursions (IC – MS - LO)

No significant differences were found for the knee gait phase angular excursions between the no AD condition and the other devices. The standard walker performed significantly better than the cane (3.768° ±1.659°, $P \leq 0.036$) and the 3WW performed better than the cane (3.754° ±1.295°, $P \leq 0.010$).

The most significant finding occurred at the hip (Table 1), between lift off and initial contact, which is considered the swing phase. Both the 4WW (2.683° ±1.060°, $P \leq 0.021$) and the 3WW (2.271° ±0.934°, $P \leq 0.026$) conditions showed significantly larger angular excursions at the hip compared to the no AD condition. In contrast, the hip angular excursion with the standard walker was significantly smaller (-5.373° ±1.487°, $P \leq 0.002$) than the no AD condition. Hip angular excursions using the cane and 2WW were not different than in the no AD condition. The 4WW performed better (Figure 1) than the cane (2.723° ±0.921°, $P \leq 0.008$) and the standard walker (8.056° ±1.261°, $P \leq 0.000$). This is also true with the 3WW against the cane (2.155° ±0.814°, $P \leq 0.017$) and standard walker (7.205° ±1.455°, $P \leq 0.000$).

Table 1: Gait phase Angular Excursion for the Hip During Swing Phase (LO - IC)

Condition	Mean ±Standard Deviation (SD)
No Assistive Device (AD)	20.568 ± 1.223
Cane	20.528 ± 1.274
Standard Walker	15.194 ± 1.395
2 Wheeled Walker	21.975 ± 1.217
3 Wheeled Walker	22.946 ± 1.035
4 Wheeled Walker	23.25 ± 1.199

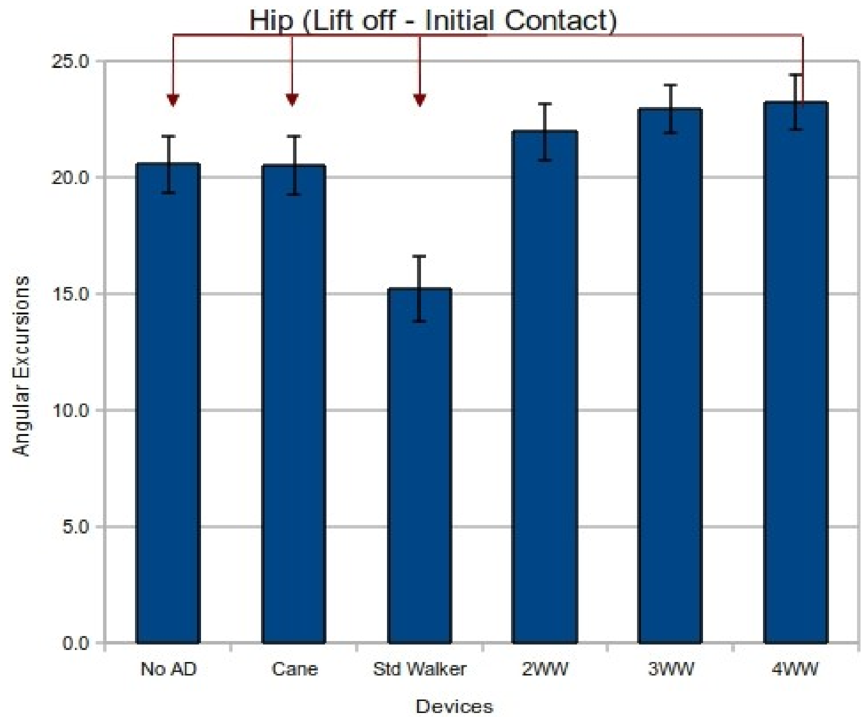

Figure 1: Hip Angular Excursion (degrees) during Swing Phase. Significance are marked in red for the 4WW.

Our findings from the gait phase excursion data show that the gait phase angular excursions of the hip were the greatest when using the 4WW compared to all of the other devices and the no AD condition. The subjects who used the 4WW or the 3WW exhibited an increase in the amount of hip flexion during the swing phase. These findings were supported by the improved velocity and stride length measurements obtained with the GAITRite when the subjects used the 3WW and 4WW. Increased hip movement likely indicates that the subjects have more confidence in their stability and are willing to take a bigger step.