

**EFFECTS OF HANDRAIL USE ON HEALTHY TREADMILL WALKING**

by

Nicole Zahradka

A thesis submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Master of Science in Mechanical Engineering

Fall 2010

Copyright 2010 Nicole Zahradka  
All Rights Reserved

**EFFECTS OF HANDRAIL USE ON HEALTHY TREADMILL WALKING**

by  
Nicole Zahradka

Approved: \_\_\_\_\_  
Jill S. Higginson, Ph.D.  
Professor in charge of thesis on behalf of the Advisory Committee

Approved: \_\_\_\_\_  
Anette M. Karlsson, Ph.D.  
Chair of the Department of Mechanical Engineering

Approved: \_\_\_\_\_  
Michael J. Chajes, Ph.D.  
Dean of the College of Engineering

Approved: \_\_\_\_\_  
Charles G. Riordan, Ph.D.  
Vice Provost for Graduate and Professional Education

## ACKNOWLEDGMENTS

In the fifth grade, I decided that when I grew up I wanted to work in the biomedical field. Watching knee and hip replacement surgeries with my cousin, who was in a biomedical program, held my attention for hours. When applying to undergraduate programs, I had my heart set on attending the University of Delaware, but chose the mechanical engineering program at Villanova University instead. Four years later, I was accepted into the masters program at the University of Delaware and started my research in the biomechanical field.

Fulfilling dreams formed in the fifth grade would not have been possible without several key people. Jill Higginson, my advisor, has provided support and constructive criticism since I've started research. Coming from a mechanical engineering background, she had been patient, shed light on biomechanical concepts, and never criticized my questions. I would not have been as productive without my lab mates, Chris Henderson, Brian Knarr, Andrew Kubinski, John Ramsay, and Elisa Schrank, who kept me sane and helped me trouble shoot any data collection or processing issues I had. Thomas Kepple was an integral part of data processing. He taught me the ins and outs of Visual 3D and was always willing to answer any questions I had. Darcy Reisman, Trisha Kesar, and Erin Helm provided insight into data collections and understanding stroke subjects' behavior. My research at the University of Delaware would not have been as successful without the help of all of these people, as well as my committee members, Darcy Reisman and Kurt Manal.

I'd like to thank my family for all of their support in every decision I've made. They're always there to help celebrate my success and find the silver lining in my failures. To everyone else not mentioned, thank you for influencing my life because it has lead me to this success.

## TABLE OF CONTENTS

|  |      |
|--|------|
| LIST OF TABLES .....   | vii  |
| LIST OF FIGURES .....  | viii |
| ABSTRACT .....   | x    |
| Chapter  |      |
| 1 INTRODUCTION .....   | 1    |
| 1.1 Gait Analysis .....  | 1    |
| 1.2 Instrumented Treadmill .....   | 2    |
| 1.3 Upper Extremity .....  | 3    |
| 1.4 Objectives .....   | 6    |
| 1.4.1 AIM 1: Analyze how the forces are distributed between the ground and the handrails during treadmill walking. ....  | 7    |
| 1.4.2 AIM 2: Compare hip angles, trunk list, and trunk tilt between two conditions: (1) no handrail is used and (2) five percent of body weight (BW) is applied in the vertical direction on the handrail..... | 7    |
| 1.5 References .....   | 8    |
| 2 HANDRAIL USE HAS SIGNIFICANT EFFECT ON HEALTHY KINETICS DURING TREADMILL WALKING .....   | 11   |
| 2.1 Introduction .....   | 11   |
| 2.2 Methods .....  | 13   |
| 2.2.1 Subjects .....   | 13   |
| 2.2.2 Equipment and Experimental Procedure .....   | 14   |
| 2.2.3 Data Analysis .....  | 15   |
| 2.3 Results .....  | 16   |
| 2.4 Discussion.....  | 19   |
| 2.4.1 Limitations/Future Directions .....  | 21   |
| 2.5 Acknowledgements .....   | 22   |

|       |  |    |
|-------|--|----|
| 2.6   | References .....   | 23 |
| 3     | EFFECTS OF CONTROLLED HANDRAIL FORCES ON GAIT<br>KINEMATICS .....  | 25 |
| 3.1   | Introduction .....   | 25 |
| 3.2   | Methods .....  | 27 |
| 3.2.1 | Subjects .....   | 27 |
| 3.2.2 | Equipment and Experimental Procedure .....   | 28 |
| 3.2.3 | Data Analysis .....  | 30 |
| 3.3   | Results .....  | 31 |
| 3.4   | Discussion.....  | 35 |
| 3.4.1 | Limitations/Future Directions .....  | 37 |
| 3.5   | Conclusion.....  | 38 |
| 3.6   | References .....   | 39 |
| 4     | CONCLUSIONS .....  | 41 |
| 4.1   | AIM 1: Analyze how the forces are distributed between the ground<br>and the handrails during treadmill walking .....   | 41 |
| 4.2   | AIM 2: Compare hip angle, trunk tilt, and trunk list between two<br>conditions: (1) no handrail is used and (2) five percent of body<br>weight (BW) is applied in the vertical direction on the handrail. .... | 43 |
| 4.3   | References .....   | 47 |
|       | APPENDIX .....   | 48 |
|       | BIBLIOGRAPHY .....   | 50 |

## LIST OF TABLES

|           |   |    |
|-----------|---|----|
| Table 2.1 | Healthy information of the twenty two subjects. This includes age, gender, height, weight, self-selected walking speed, and handedness (mean $\pm$ standard deviation).....   | 14 |
| Table 2.2 | Peak handrail forces during the four conditions (BHR, LHR, RHR, NHR). These forces were normalized to percent of body weight over one gait cycle and can also be found in Figure 2.2. A one way ANOVA with repeated measures was used to compare the right forces during the RHR and BHR conditions, the left forces during the LHR and BHR conditions, the forces during the RHR and LHR conditions, and the forces during BHR conditions. No significant differences were found. .... | 18 |
| Table 3.1 | Healthy information of the twenty five subjects who participated in the force control study including age, gender, height, weight, self-selected walking speed, and handedness (mean $\pm$ standard deviation).....   | 28 |
| Table 3.2 | These values are the maximum flexion angle, maximum extension angle, and excursion of the left hip under during the three conditions: no hands on the handrail (NHR), left hand applying five percent body weight to the handrail (LHR), and right hand applying five percent body weight to the handrail (RHR) measure in degrees $\pm$ standard deviation .....   | 32 |
| Table 3.3 | The maximum flexion angle, maximum extension angle, and excursion of the right hip under during the three conditions: no hands on the handrail (NHR), left hand applying five percent body weight to the handrail (LHR), and right hand applying five percent body weight to the handrail (RHR) measure in degrees $\pm$ standard deviation .....   | 32 |
| Table 3.4 | The minimum and maximum trunk list in the frontal plane.....  | 33 |
| Table 3.5 | The minimum and maximum forward tilt in the sagittal plane. ....  | 34 |

## LIST OF FIGURES

|            |  |    |
|------------|--|----|
| Figure 2.1 | Right vertical ground reaction forces for the four conditions (BHR, LHR, RHR, NHR) normalized to percent body weight from right heel strike to right heel strike. A one way ANOVA with repeated measures was used to compare the four conditions at the first and second peaks. The first peak has a significant difference between conditions ( $p=.0046$ ) and there was no significant difference in the second peak ( $p=.1442$ ). ..... | 17 |
| Figure 2.2 | Left and right handrail forces in the vertical direction normalized to percent body weight. The handrail forces were plotted from right heel strike to right heel strike for three handrail conditions: both handrails (BHR), right handrail only (RHR), and left handrail only (LHR). The peak values are found in table 2.2.....   | 18 |
| Figure 2.3 | Left and right handrail forces in the vertical direction normalized to percent body weight. These forces were plotted from left heel strike to left heel strike for the three handrail conditions. This figure demonstrates that the handrails are used in a similar pattern during right heel strike (Figure 2.2) and left heel strike (Figure 2.3).....  | 19 |
| Figure 3.1 | The real time feedback display available to the subjects. The blue line represented the handrail force the subject exerted on the left handrail, the white line represented 5% of the subject's BW and the gray area was $\pm 5$ N. The same display was available during the RHR condition.....   | 29 |
| Figure 3.2 | Trunk list in the frontal plane shows significant differences between the conditions where a handrail is used and the no handrail condition. The three conditions, no handrails used (NHR), 5% BW applied to the left handrail (LHR), and 5% BW applied to the right handrail (RHR), were plotted from RON to RON. ....  | 33 |
| Figure 3.3 | Trunk tilt in the sagittal plane demonstrates that subjects lean forward during both conditions where they were applying handrail forces. All trials were plotted from RON to RON. ....  | 34 |

Figure A.1 Right and left ankle moments for three of the subjects, normalized to one gait cycle. The right ankle moments were plotted from RON to RON. The left ankle moments were plotted from LON to LON..... 49

## ABSTRACT

Upper extremity movements and forces have important implications for rehabilitation of pathological gait. Patients with pathological gait often rely on an outside source (such as a cane or handrail) to assist them. It is important to take into consideration if handrail usage significantly alters the kinetic and kinematic data during different walking conditions because the amount of force used on the handrail varies between subjects and may influence these parameters. This study combined treadmill gait analysis with upper extremity force measurement on healthy subjects to provide an understanding of the changes that occur in kinetics and kinematics with handrail use. **Objective:** The objective of this study was to evaluate the effect of handrail use on the kinetics and kinematics during treadmill walking in healthy subjects. We compared (1) magnitudes of handrail and ground reaction forces and (2) hip angles when no handrails were used and when five percent of body weight was applied to the handrail. **Methods:** Twenty five healthy young adults were recruited for this study. Kinematic and kinetic data were collected while all subjects walked on an instrumented split-belt treadmill at their self-selected speed. To determine the effect of handrail usage on walking kinetics, subjects walked under four handrail conditions (no hands on handrail (NHR), both hands on the handrail (BHR), left hand only on the handrail (LHR), and right hand only on the handrail (RHR)). The subjects walked under three additional conditions to quantify changes in kinematics when handrail force was controlled (no hands on handrail, 5% body weight (BW) on right

handrail, and 5% body weight on left handrail). Ground reaction forces and handrail forces as well as hip angles, trunk list and trunk tilt were analyzed. **Results:** The first peak of the ground reaction force was reduced significantly when both handrails were used. This suggests the handrails were used for stability during heel strike. The left and right handrail forces when only one handrail was used showed no significant differences and the right handrail forces in the RHR and BHR conditions also remained the same. During handrail force control conditions, the trunk list, in the frontal plane, and trunk tilt, in the sagittal plane, were significantly different, but the hip angles were not. **Conclusions:** Handrail use needs to be considered because it has an effect on ground reaction forces, which may influence inverse dynamics calculations. Although there were no significant differences in hip angles when a handrail force of 5% BW (or less) is applied to the handrails, there were significant differences in trunk list and trunk tilt, which may be more pronounced with greater handrail forces applied. When a subject uses a handrail during a data collection, different handrail conditions and the amount of force applied to the handrails need to be taken into account.

## **Chapter 1**

### **INTRODUCTION**

Upper extremity movements and forces have important implications for rehabilitation of pathological gait. Patients with pathological gait often rely on an outside source (such as a cane or handrail) to assist them (Jørgensen et al., 1995). For stroke patients, who are often paralyzed on one side of their body, they rely on their non-paretic upper limb for support. When performing research studies with stroke patients on a treadmill, a handrail may be used (Chen et al., 2005, Ada et al., 2010). It is important to take into consideration whether the handrail significantly alters the ground reaction forces because the amount of force used on the handrail varies between subjects and may influence loads on the lower extremity. Furthermore, it is unclear how differences in handrail usage affect ground reaction forces and subsequent inverse dynamics calculations. This study combined treadmill gait analysis with upper extremity force measurement on healthy subjects to provide an understanding of the changes that occur in kinetics and kinematics with handrail use.

#### **1.1 Gait Analysis**

Gait analysis is a useful tool when investigating locomotion in humans. Successful performance of gait relies on the ability to maintain support of body weight during stance phase (Olney et al., 1996, Winter, 1991). When a patient has a stroke, it often leaves them impaired on one side of their body. If the impaired limb is not able to support the body during single support, the stroke patients usually rely on their non-

paretic upper extremity and an outside source (such as a cane or handrail) to provide the stability needed. This stabilization (using an outside source such as handrail and harness) has been found to improve a subset of gait deviations (Chen et al., 2001).

In most gait studies, the upper extremity does not play a significant role. The arms are allowed to swing freely and the measurements are taken from the lower extremity. However, when a stroke patient holds onto something for support, it becomes an important part of gait analysis (Chen et al., 2001, Chen et al., 2005). The upper body and outside source are used for stability and may reduce the loads carried by the paretic limb. This may affect ground reaction forces and joint angles used by stroke patients. One focus of this study is the relationship between upper extremity support and the ground reaction forces. Understanding how the use of the upper extremity changes the kinetics and kinematics of gait is useful to better assess stroke gait during rehabilitation evaluations.

## **1.2 Instrumented Treadmills**

Instrumented treadmills are commonly used in clinical and laboratory studies for several reasons. They provide a consistent environment and limit the amount of space necessary to analyze gait. Spatial-temporal gait parameters can be readily determined on treadmills. Treadmills can have force plates embedded in them, which allows the ground reaction forces and moments to be measured. The free moment and center of pressure can be calculated from this information. Gait events can be determined from the forces (Roerdink et al, 2008) and some parameters that are commonly analyzed are ground reaction forces, stride length, cadence, gait symmetry, lower extremity joint angles, and stride interval dynamics (White et al.,

1998, Stolze et al., 1997, Chen et al., 2005, Alton et al., 1998, Riley et al., 2007, Chang et al., 2009).

When using instrumented treadmills, the ground reaction forces and moments are recorded for multiple gait cycles. Besides quantifying the magnitude of forces exerted over an individual gait cycle, a comparison of how the forces change between consecutive gait cycles can be made. Another benefit is that walking speed is a controlled variable in this setting. In overground walking, speed is difficult to control, may affect dependent measures such as ground reaction forces, and not all overground trials are able to be used (Riley et al., 2007, White et al., 1998). The simultaneous capture of kinetic data on the instrumented treadmill is also a way to verify the kinematic data collected by the motion capture system (Mickelborough et al., 2000).

It has been suggested that walking on a treadmill is not an accurate representation of a subject's normal walking pattern (Murray et al., 1985). Comparisons between treadmill walking and overground walking have been performed and it has been found that instrumented treadmill measuring ground reaction force components were similar to overground gait (Riley et al., 2007, White et al., 1998). Based on the assumption that the subjects respond to treadmill testing equally, the few variables, such as maximum hip flexion angle, stance time, cadence, and hip range of motion, that differ between overground and treadmill walking can be overlooked (Alton et al., 1998).

### **1.3 Upper Extremity**

Although many previous studies have analyzed the lower limb during gait, little is known about the upper extremity's kinetics and kinematics, especially on a

treadmill. Most of the studies that have investigated upper extremity parameters focused on the kinematics of patients who have experienced some sort of trauma. The experiments use “reach tasks” to evaluate the joint angles, such as the elbow angles (Fitoussi et al., 2009, Chang, et al., 2009). The focus is on the top half of the body and how it reacts when given instructions while the lower extremities are stationary. One purpose of measuring upper extremity kinematics is to quantify motor control and to assess rehabilitation (Hingtgen et al., 2006).

Previous studies looking at handrail and no handrail conditions (Chapdelaine et al., 2005, Reeves et al., 2008, Jeka et al., 1994, Siler et al., 1997) did not consider the forces exerted on the handrail. These studies focused on the effects of handrail use on the kinetics during stair locomotion (Chapdelaine et al., 2005, Reeves et al., 2008), the relationship between postural sway and contact forces at the fingers (Jeka et al., 1994), and the kinematics when using a handrail on the treadmill (Reeves et al., 2008). Therefore they were not able to determine if there was a correlation between the magnitude of force applied to the handrail and the magnitude of the ground reaction forces. Another limitation is that they had handrail use as a nominal variable. While they looked at handrail (both) vs. no handrail, previous studies did not investigate the use of a single handrail by either the left or right hands. It is important to study the use of single handrails because certain pathologies, such as stroke, may limit the ability of a patient from supporting themselves with two arms.

Some gait studies that include the upper extremity focus on the effect of arm swing on kinematic parameters. A common method used to evaluate this was to restrict one or both of the arms from its natural swinging motion during overground walking (Ford et al., 2007, Eke-Orkoro et al., 1997, Umberger et al., 2008). This was

accomplished with either bondages or having the subjects cross their arms. The outcomes varied between studies. While some investigators determined that constraining the arms does alter gait patterns, such as stride frequency and velocity, in healthy and patient populations (Ford et al., 2007, Eke-Orkoro et al., 1997), another determined that in healthy subjects, kinematic and kinetic variables are only marginally affected by arm swing in the sagittal direction (Umberger et al., 2008). Although these studies determined if gait parameters changed with arm swing, there were no external forces applied by the upper limbs. The conditions that limited or changed arm swing were done within the subject. There was no outside source provided to support the subject or hinder their movements. Therefore, it would not be valid to assume that the addition of handrail use would result in the same gait parameter values as restricting arm movement.

Stephenson et al. (2009) studied the coordination of the upper and lower limb movement during treadmill walking in healthy and stroke patients. The vertical forces exerted on the handle by healthy subjects during their comfortable speeds were  $2.7 \pm 1.5\%$  body weight. Both groups took shorter and more frequent strides when they weren't using the handles. The effect of speed was taken into consideration and they found that stroke subjects exerted more force on the handles during the fast speed. It was also determined that the arm-leg coordination during walking was unaffected by the use of the moving handles (Stephenson et al., 2009). There was no difference found in arm-leg coordination between the stroke group and the healthy groups, however there was a significant difference in arm-leg coordination seen within each group. They found that arm movements during treadmill walking changed stride characteristics and lower limb muscle activation patterns, but not joint kinematics

(Stephenson et al., 2010). The stroke patients put more weight on the handrails during the second half of the gait cycle of the stationary handle trials compared to the sliding handle trials. The muscle activation patterns and intensity of the activity differed between the no handles condition and the use of handles. The results of this study demonstrate that gait patterns are altered when an external source is present. However, this study was limited to the both handrail condition and evaluation of muscle activation patterns. Our goal is to determine the change in ground reaction forces between no hands on the handrail, one hand on the handrail, and both hands on the handrail. Furthermore, no studies have investigated the kinematics when a controlled force is applied to the handrails.

#### **1.4 Objectives**

The objective of this study was to evaluate the effect of handrail use on kinetics during treadmill walking in healthy subjects. We compared magnitudes of handrail and ground reaction forces to determine if an increase in handrail use is related to a decrease in ground reaction forces. Another goal was to see if the handrail force when only using the right handrail was identical to the handrail force when using only the left handrail. Lastly, the hip angles were compared when no handrails were used and when five percent of body weight was applied to the handrail. The results of these aims provide a basis for comparison of kinetic and kinematic changes in stroke gait when a handrail is used.

**1.4.1 AIM 1: Analyze how the forces are distributed between the ground and the handrails during treadmill walking.**

Hypothesis 1.1: The most ground reaction force is seen when there are no handrails used and this ground reaction force decreases with an increase in the number of handrails used.

Hypothesis 1.2: When both handrails are used, the subject distributes an equal amount of weight on both handrails.

Hypothesis 1.3: When only one handrail is used, the same handrail force is applied by the subject.

**1.4.2 AIM 2: Compare hip angles, trunk list, and trunk tilt between two conditions: (1) no handrail is used and (2) five percent of body weight (BW) is applied in the vertical direction on the handrail.**

Hypothesis 2.1: Hip angles are largest when no handrails are used.

Hypothesis 2.2: In the sagittal plane, the trunk tilt is the same under both conditions.

Hypothesis 2.3: In the frontal plane, the trunk creates a larger angle with the midline during the condition when five percent of body weight is applied to the handrail than under the condition when no handrails are used.

## 1.5 References

Ada, L., Dean, C.M., Vargas, J., Ennis, S., 2010. Mechanically assisted walking with body weight support results in more independent walking than assisted overground walking in non-ambulatory patients early after stroke: a systematic review. *Australian Journal of Physiotherapy* 56, 153-161.

Alton, F., Baldey, L., Caplan, S., Morrissey, M.C., 1998. A kinematic comparison of overground and treadmill walking. *Clinical Biomechanics* 13, 434-440.

Chang, M.D., Shaikh, S., Chau, T., 2009. Effect of treadmill walking on the stride interval dynamics of human gait. *Gait & Posture* 30, 431-435.

Chapdelaine, S., McFadyen, B.J., Nadeau, S., St-Vincent, G., 2005. Instrumented staircase for kinetic analyses of upper- and lower-limb function during stair gait. *Medical & Biological Engineering and Computing* 43, 552-556.

Chen, C.L., Chen, H.C., Wong, M.K., Tang, F.T., Chen, R.S., 2001. Temporal stride and force analysis of cane-assisted gait in people with hemiplegic stroke. *Archives of physical medicine and rehabilitation* 82, 43-48.

Chen, G., Patten, C., Kothari, D.H., Zajac, F.E., 2005. Gait deviations associated with post-stroke hemiparesis: improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold. *Gait & Posture* 22, 57-62.

Eke-Okoro, S.T., Gregoric, M., Larsson, L.E., 1997. Alterations in gait resulting from deliberate changes of arm-swing amplitude and phase. *Clinical Biomechanics* 12, 516-521

Fitoussi, F., Maurel, N., Diop, A., Laassel, E.M., Ilharreborde, B., Presedo, A., Mazda, K., Penneçot, G.F., 2009. Upper extremity kinematics analysis in obstetrical brachial plexus palsy. *Orthopaedics & Traumatology: Surgery & Research* 95, 336-342

Ford, M.P., Wagenaar, R.C., Newell, K.M., 2007. Arm constraint and walking in healthy adults. *Gait & Posture* 26, 135-141.

- Hingtgen, b., McGuire, J.R., Wang, M., Harris, G.F., 2006. An upper extremity kinematic model for evaluation of hemiparetic stroke. *Journal of Biomechanics* 39, 681-688.
- Jeka, J.J., Lackner, J.R., 1994. Fingertip contact influences human postural control. *Experimental Brain Research* 100, 495-502.
- Jørgensen, H.S., Nakayama, H., Raaschou, H.O., Olsen, T.S., 1995. Recovery of walking function in stroke patients: The Copenhagen stroke study. *Archives of Physical Medicine and Rehabilitation* 76, 27-32.
- Mickelborough, J., van der Linden, M.L., Richards, J., Ennos, A.R., 2000. Validity and reliability of a kinematic protocol for determining foot contact events. *Gait and Posture* 11, 32-37.
- Murray, M.P., Spurr, G.B., Sepic, S.B., Gardener, G.M., Mollinger, L.A., 1985. Treadmill vs. floor walking: kinematic, electromyogram, and heart rate. *Journal of Applied Physiology* 59, 87-91.
- Olney, S.J., Richards, C., 1996. Hemiparetic gait following stroke. Part I: Characteristics. *Gait & Posture* 4, 136-148.
- Reeves, N.D., Spanjaard, M., Mohagheghi, A.A., Baltzopoulos, V., Maganaris, C.N., 2008. Influence of light handrail use on the biomechanics of stair negotiation in old age. *Gait & Posture*, doi: 10.1016/j.gaitpost.2008.01.014
- Riley, P.O., Paolini, G., Croce, U.D., Paylo, K.W., Kerrigan, D.C., 2007. A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects. *Gait & Posture* 26, 17-24.
- Roerdink, M., Coolen, B.H., Clairbois, B.H.E., Lamoth, C.J.C., Beek, P.J., 2008. Online gait even detection using a large force platform embedded in a treadmill. *Journal of Biomechanics* 41, 2628-2632.
- Siler, W.L., Jørgensen, A.L., Norris, R.A., 1997. Grasping the handrails during treadmill walking does not alter sagittal plane kinematics of walking. *Archives of Physical Medicine and Rehabilitation* 78, 393-398.
- Stephenson, J.L., Lamontagne, A., De Serres, S.J., 2009. The coordination of upper and lower limb movements during gait in healthy and stroke individuals. *Gait & Posture* 29, 11-16.

Stephenson, J.L., De Serres, S.J., Lamontagne, A., 2010. The effect of arm movements on the lower limb during gait after a stroke. *Gait & Posture* 31, 109-115.

Stolze, H., Kuitz-Buschbeck, J.P., Mondwurf, C., Boczek-Funcke, A., Jöhnk, K., Deuschl, G., Illert, M., 1997. Gait analysis during treadmill and overground locomotion in children and adults. *Electroencephalography and clinical Neurophysiology* 105, 490-497.

Umberger, B.R., 2008. Effects of suppressing arm swing on kinematics, kinetics, and energetic of human walking. *Journal of Biomechanics* 41, 2575-2580.

White, S.C., Tucker, C.A., Lifeso, R.M., 1996. Verbal feedback cues for altering vertical ground reaction forces patterns in gait re-education training. *Gait & Posture* 4, Issue 2, 206.

Winter, D.A.. *The Biomechanics and Motor Control of Human Gait: Normal, Elderly, and Pathological*. Ontario: Waterloo Biomechanics, 1991.

## Chapter 2

### HANDRAIL USE HAS SIGNIFICANT EFFECT ON HEALTHY KINETICS DURING TREADMILL WALKING

#### 2.1 Introduction

Upper extremity movements and forces have important implications for rehabilitation of pathological gait. Patients with pathological gait frequently rely on an outside source (such as a cane or handrail) to assist them (Jørgensen et al., 1995). For stroke patients, who are often paralyzed on one side of their body, they rely on their non-paretic upper limb for support. The upper body and outside source are used to stabilize and reduce the loads carried by the paretic limb. The use of handrails may affect ground reaction forces and joint angles of stroke patients during gait studies. It is unclear how differences in handrail usage affects ground reaction forces and subsequent inverse dynamics calculations.

Instrumented treadmills are commonly used in clinical and laboratory studies because they provide a consistent environment and limit the amount of space necessary to analyze gait. Treadmills allow spatial-temporal gait parameters to be easily determined and they can have force plates embedded in them, which allows the ground reaction forces and moments to be measured (Riley et al., 2007, White et al., 1998). Some common parameters evaluated during gait analysis are ground reaction forces, stride length cadence, gait symmetry, step kinematic variability, lower extremity joint angles, and stride interval dynamics (White et al., 1998, Stolze et al.,

1997, Chen et al., 2005, Alton et al., 1998, Riley et al., 2007, Chang et al., 2009). A concern regarding instrumented treadmills is that walking on a treadmill is not an accurate representation of a subject's normal walking pattern. Comparisons between treadmill walking and overground walking have been performed and it has been found that ground reaction force components measured on an instrumented treadmill were similar to those during overground gait (Riley et al., 2007, White et al., 1998). Based on the assumption that all groups respond to treadmill testing equally, the few variables that differ between overground and treadmill walking can be overlooked (Alton et al., 1998).

An instrumented treadmill, along with instrumented handrails, can provide insight to how walking kinetics change with the inclusion of the upper extremity. Some gait studies that have included the upper extremity focus on the effect of arm swing on kinematic parameters. While some determined that constraining the arms does alter gait patterns in healthy and patient populations (Ford et al., 2007, Eke-Orkoro et al., 1997) such as stride frequency and velocity, another determined that in healthy subjects, kinematic and kinetic variables are only marginally affected by arm swing in the sagittal direction (Umberger et al., 2008). Although these studies investigated whether gait parameters changed with different arm swing, there were no external forces applied by the upper limbs. Two studies performed by Stephenson et al. evaluated some of the gait parameters under three conditions: holding onto stationary handles, holding onto handles that were free to slide in the anterior/posterior direction, and no handles at a self-selected speed and a fast speed (Stephenson et al., 2009, Stephenson et al., 2010). They reported that both healthy and stroke subjects took shorter and more frequent strides when they weren't using the

handles. The effect of speed was taken into consideration and they found that stroke subjects exerted more force on the handles during the fast speed. It was also determined that the arm-leg coordination during walking was unaffected by the use of the moving handles (Stephenson et al., 2009). Although arm movements during treadmill walking changed stride characteristics and lower limb muscle activation patterns, no differences in joint kinematics were observed (Stephenson et al., 2010).

The objective of this study was to evaluate the effect of handrail use on the kinetics during treadmill walking in healthy subjects. We will (1) compare magnitudes of handrail and ground reaction forces, (2) compare left and right handrail forces when only one handrail is used and (3) compare the left and right handrail forces when both handrails are used. For the first objective, we hypothesized that an increase in handrail use would lead to a decrease in ground reaction forces. We hypothesized that handrail force on the right and left handrails would be identical when only using the right handrail, only using the left handrail and using both handrails.

## **2.2 Methods**

### **2.2.1 Subjects**

Twenty five individuals participated in this study to determine if there was a change in their kinetics and kinematics when handrails were used. These subjects were men and women between 18 and 40 years of age with no history of muscle, bone or nervous system disorders. Potential subjects were asked to complete a physical activity readiness questionnaire and sign an informed approved by the Human Subjects Review Board at the University of Delaware consent prior to participating.

The subject's height, weight, handedness, and walking speed were recorded (Table 2.1).

**Table 2.1 Subject Information (mean  $\pm$  st dev)**

|                                    |                   |
|------------------------------------|-------------------|
| Age                                | 24 $\pm$ 4.64     |
| Gender                             | 12 F, 13 M        |
| Height (m)                         | 1.72 $\pm$ 0.10   |
| Weight (kg)                        | 70.34 $\pm$ 13.82 |
| Walking Speed (m*s <sup>-1</sup> ) | 1.27 $\pm$ 0.15   |
| Hand Dominance                     | 23 R, 2 L         |

### 2.2.2 Equipment and Experimental Procedure

An instrumented split-belt treadmill (Bertec Corp., Columbus, OH) was used during this experiment. Two force plates (embedded in the treadmill) and two force transducers (embedded in the handrails) captured the ground reaction forces and handrail forces, respectively, at 1080 Hz. Forces were collected in all three planes, but only the vertical forces were evaluated because the forces in the other direction were much lower. All trials were recorded in Cortex (version 1.0.0.198) by eight cameras (Motion Analysis Corp., Santa Rosa, CA) that capture motion of reflective markers attached to the body segments at 60 Hz.

The subject's self-selected speed was calculated by timing the subjects as they walked overground for 10 meters. This was done twice and their average was considered to be their comfortable speed (self-selected speed) and used throughout all of the trials.

Individual markers were attached to the subject's anatomical landmarks of the upper and lower bodies at the sternum, right scapula, shoulders, upper arms,

elbows, lower arms, wrists, hands, right and left ASIS, sacrum, thighs, medial and lateral knees, shanks, medial and lateral ankles, toes, and heels. Three standing trials were completed before the walking tasks. The subjects were then asked to walk under four conditions (no hands on handrail (NHR), both hands on the handrail (BHR), left hand only on the handrail (LHR), and right hand only on the handrail (RHR)). The order of these trials was randomized and each trial was recorded for 30 seconds.

### **2.2.3 Data Analysis**

The data were checked for marker drop outs and marker accelerations in Cortex. Poor data points were deleted from the data set and the small gaps were connected using pattern recognition, also in Cortex. The files were exported to C3D files.

A model was created in Visual 3D including feet, legs, torso, arms and hands with segments scaled to subject height and weight. The targets were filtered using a Butterworth lowpass filter with a cut off frequency of 6 Hz. The handrail data were imported into Visual 3D and converted to Newtons. The handrail and force plate data were both normalized to body weight. A Butterworth lowpass filter with a cut off frequency of 30 Hz was used on the force plate and handrail data. Gait events were identified (1) when the right foot hits the force plate (RON), (2) when the right foot leaves the force plate (ROFF), (3) when the left foot hits the force plate (LON) and (4) when the left foot leaves the force plate (LOFF). Each subject's trials were averaged over all gait cycles, from RON to RON for the right ground reaction forces, right handrail forces, and left handrail forces. The left ground reaction forces were averaged over all gait cycles from LON to LON. All trials were then normalized to 101 points. The four force profiles (left handrail force, right handrail force, right

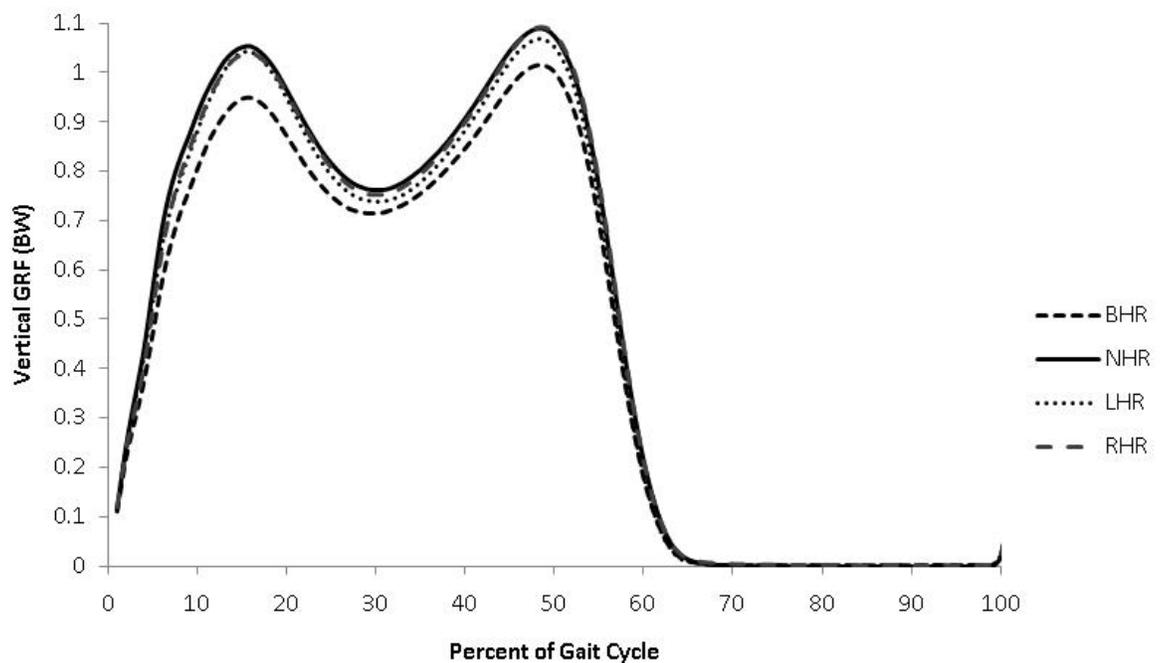
ground reaction force, and left ground reaction force) for each trial were exported to Excel and averaged among subjects for each condition.

Statistics were performed on the peak forces of the ground reaction forces and handrail forces. The first peak ground reaction forces of the right side were analyzed between the four conditions (BHR, NHR, LHR, RHR) using a one way analysis of variance (ANOVA) with repeated measures. The same statistical test was used to determine if there was a significant difference between conditions for the second peak forces. Paired t-tests were used to evaluate the handrail forces between sides and conditions. Handrail forces were compared between the left vs. right in the BHR condition, left vs. right in the LHR and RHR conditions, right handrail forces between BHR vs. RHR conditions, and left handrail forces between BHR vs. LHR conditions. Statistical analyses were performed using SAS Institute (2009) software.

### **2.3 Results**

The first peak force ranged from  $1.075 \pm 0.059$  BW (NHR) to  $0.968 \pm 0.206$  BW (BHR). The LHR and RHR conditions had similar peak ground reaction forces (LHR:  $1.064 \pm 0.061$  BW, RHR:  $1.063 \pm 0.062$  BW). The second peak force also varied between conditions. The BHR condition had a peak force of  $1.025 \pm 0.222$  and the NHR condition had a peak force of  $1.095 \pm 0.071$ . Unlike the first peak force where the LHR and RHR ground reaction forces were very similar, the second peak forces were  $1.077 \pm 0.073$  and  $1.102 \pm 0.073$ , respectively (Figure 2.1). Although the four conditions had varying peaks during the gait cycle, the differences in peak values were low for the NHR, RHR, and LHR conditions. The differences between the first peaks were 0.012 BW (NHR-RHR) and 0.011 BW (NHR-LHR) and the differences between the second peaks were 0.018 BW (NHR-RHR) and 0.007 BW (NHR-LHR).

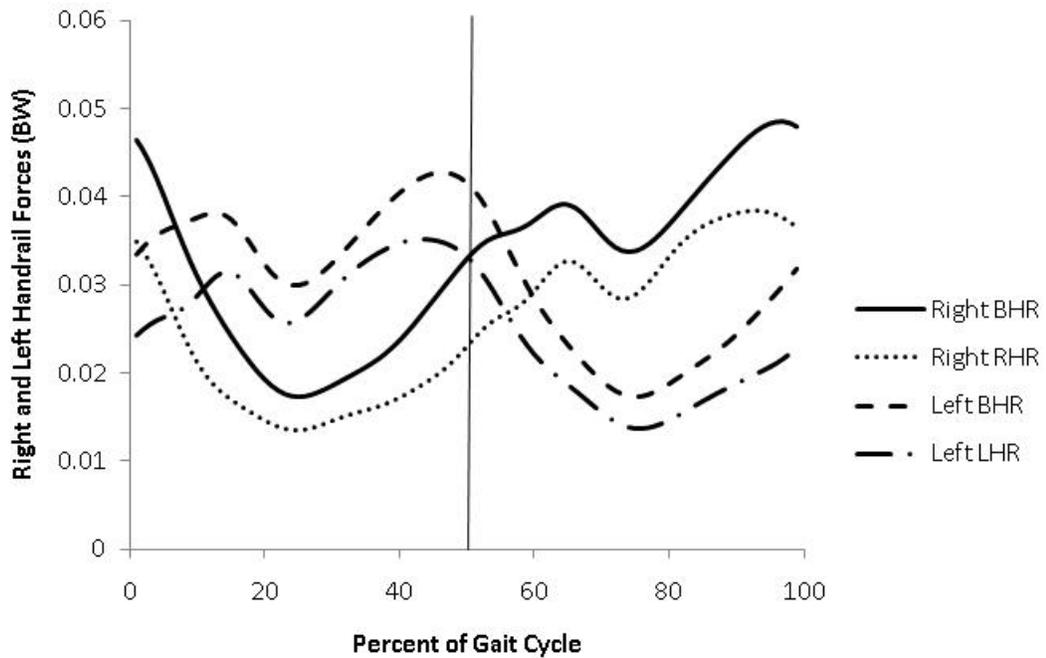
However, there was a larger difference between these three conditions (NHR, RHR, and LHR) and the BHR conditions at both of the peaks (0.107 BW between the first peaks and 0.065 BW between the second peaks). The left ground reaction force followed in a similar pattern. There was a significant difference ( $p=0.005$ ) in the right ground reaction forces under different handrail conditions during the first peaks. The second peaks did not have a significant difference ( $p=0.144$ ) between conditions.



**Figure 2.1** Significant differences noted in the first but not second peaks of the right vertical ground reaction force for four conditions (both handrails (BHR), no handrails (NHR), left handrail only (LHR), and right handrail only (RHR)).

The handrail forces for the BHR condition were greater than the handrail forces when only one of the handrails were held (Table 2.2), however there was not a significant difference between them (left:  $p=0.185$ , right:  $p=0.081$ ). The right arm exerted a higher peak force than the left arm under each comparable condition (Figure

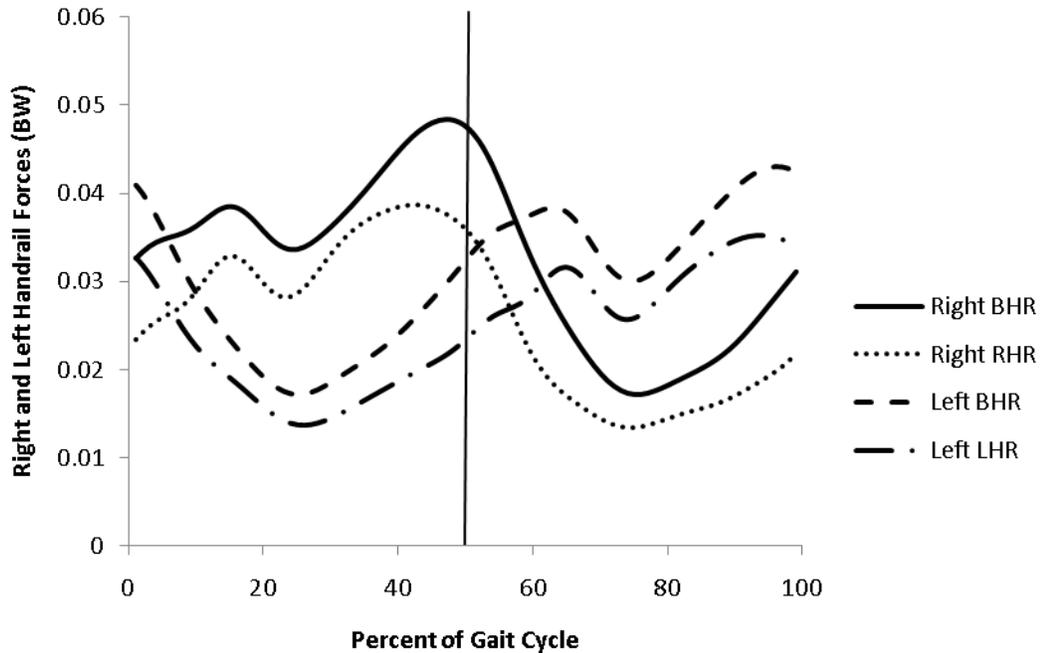
2.2). The right BHR peak was  $0.049 \pm 0.005$  BW while the left BHR peak was  $0.042 \pm 0.004$  BW. There was no significant difference between the right and left side during the BHR ( $p=0.094$ ). Similarly the right RHR peak force was  $0.038 \pm 0.002$  BW and the left LHR peak force was  $0.035 \pm 0.003$  BW. There was no statistical significance between these two conditions ( $p=0.411$ ).



**Figure 2.2** The left and right hands exert different peak forces on the handrails normalized to one gait cycle (RON to RON). The line denotes when the left foot contacts the ground.

**Table 2.2** Peak handrail forces normalized to percent of body weight over one gait cycle.

| Condition | First Peak (% BW) | Second Peak (% BW) |
|-----------|-------------------|--------------------|
| R BHR     | $4.644 \pm 0.822$ | $4.853 \pm 0.505$  |
| R RHR     | $3.490 \pm 0.675$ | $3.845 \pm 0.278$  |
| L BHR     | $3.823 \pm 0.715$ | $4.282 \pm 0.449$  |
| L LHR     | $3.139 \pm 0.610$ | $3.510 \pm 0.282$  |



**Figure 2.3** The left and right handrail forces show similar patterns to Figure 2.2 when plotted from LON to LON. These forces are normalized to one gait cycle. The line denotes when the right foot contacts the ground.

## 2.4 Discussion

Upper extremity forces do influence the ground reaction forces. The significant differences seen in the ground reaction force early in stance were diminished prior to toe-off suggesting that force was applied to the handrail for balance after heel strike and no longer needed once stabilized. The average peak right handrail force for the BHR condition was  $4.644 \pm .822\%$  BW; however it ranged up to 10% of body weight. This amount of force on the handrail caused a significant reduction in the ground reaction forces which will have implications for inverse dynamics calculations (Figure 5.1).

It is not surprising that an increase in handrail use decreased the ground reaction forces. During the LHR and RHR conditions, healthy young adults applied

approximately 5% of body weight on the handrail during contralateral swing while the BHR condition had at least one handrail engaged throughout the entire gait cycle. During the BHR condition, the left handrail force had a greater magnitude during right stance and the right handrail force had a greater magnitude during left stance. The handrail force, for the BHR condition, never fell below 3% of body weight (Figure 2.2). Although there was no significant difference between the amount of force applied to the handrail during the comparative conditions (right handrail force for BHR vs. RHR and left handrail force for BHR vs. LHR), the fact that there was a constant force being applied to the handrails in the BHR condition may explain why there was a greater difference between the ground reaction forces of the BHR condition and NHR condition than there was for the LHR or RHR condition compared to the NHR condition. Several observations were made regarding the magnitude of the handrail forces, but one noticeable trend was the handrail force was out of phase with the ground reaction force, similar to arm swing and consistent with observations by Stephenson et al. (2010).

There was no significant difference between the left and right handrail forces when one handrail was held (LHR/RHR conditions). The subjects applied approximately the same amount of force to the handrail when only one is available which establishes the basis that handrail forces only need to be evaluated for one side of the body. If healthy young adults are only using one handrail, it does not matter which side is used. It is important to note that 23 out of 25 subjects were right handed. We did not determine if hand dominance had an effect on handrail forces, however further investigation may explain the differences, even though insignificant, between the right and left handrail forces.

The effect of speed was not taken into account; however previous studies found that subjects exerted more force on the handrails during fast trials than during their self-selected speed (Stephenson et al., 2009). The average walking speed in this study was  $1.27 \pm 0.15$  m/s. It is likely that these healthy subjects would also exert increased handrail forces when walking at a faster speed. This would affect the ground reaction forces, and further alter the inverse dynamics calculations. Speed should be considered in future studies to evaluate the role it plays during walking with different handrail conditions and how it affects gait parameters.

These differences in gait parameters become important when evaluating the progress of stroke patients during rehabilitation. The amount of force applied to the handrails varies between subjects and may be affected by therapeutic interventions. Comparing gait parameters between subjects is limited without taking into account the change in ground reaction forces since handrail usage alters the subject's kinetic data.

#### **2.4.1 Limitations/Future Directions**

One limitation of this study was the sample size. The left and right handrail forces during the BHR condition and right handrail forces compared between the BHR condition and RHR condition may have approached significance if more subjects were added. The distribution of right handed subjects and left handed subjects was skewed. There were approximately seven times more right handed subjects and this was not taken into account in this study. The height of the handrail used was constant which may have altered the forces applied to the handrail by subjects of varying height. The ground reaction forces and handrail forces were only evaluated in the vertical direction, similarly to Stephenson et al. (2009), because it was

the direction where the greatest forces were applied. However, future studies should evaluate all three directions because there is very little information known about the magnitude of handrail forces. Further studies are needed to determine if these results can be applied to stroke subjects and other patients with pathological gait who rely on handrails. If the amount of handrail force applied by patients with pathology is significantly greater than healthy subjects, it may be useful to quantify what force causes the gait parameters to significantly differ from the no handrail condition.

This study evaluated the effect of handrail use on ground reaction forces and found that handrail use does significantly change the ground reaction forces. It also demonstrated that there was no significant difference between the right and left sides in both (BHR and LHR or RHR) conditions. It is likely that subjects with a pathological gait pattern who use handrails have altered joint loads which should be considered during rehabilitation and other treadmill studies.

## **2.5 Acknowledgements**

We gratefully acknowledge the efforts of Thomas Kepple, Trisha Kesar, and Andrew Kubinski for their assistance in data collection and post-processing. Financial support for this project was provided by NIH R01 HD38582 and NIH P20-RR16458.

## 2.6 References

- Alton, F., Baldey, L., Caplan, S., Morrissey, M.C., 1998. A kinematic comparison of overground and treadmill walking. *Clinical Biomechanics* 13, 434-440.
- Chang, M.D., Shaikh, S., Chau, T., 2009. Effect of treadmill walking on the stride interval dynamics of human gait. *Gait & Posture* 30, 431-435.
- Chen, C.L., Chen, H.C., Wong, M.K., Tang, F.T., Chen, R.S., 2001. Temporal stride and force analysis of cane-assisted gait in people with hemiplegic stroke. *Archives of physical medicine and rehabilitation* 82, 43-48.
- Chen, G., Patten, C., Kothari, D.H., Zajac, F.E., 2005. Gait deviations associated with post-stroke hemiparesis: improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold. *Gait & Posture* 22, 57-62.
- Eke-Okoro, S.T., Gregoric, M., Larsson, L.E., 1997. Alterations in gait resulting from deliberate changes of arm-swing amplitude and phase. *Clinical Biomechanics* 12, 516-521
- Ford, M.P., Wagenaar, R.C., Newell, K.M., 2007. Arm constraint and walking in healthy adults. *Gait & Posture* 26, 135-141.
- Jørgensen, H.S., Nakayama, H., Raaschou, H.O., Olsen, T.S., 1995. Recovery of walking function in stroke patients: The Copenhagen stroke study. *Archives of Physical Medicine and Rehabilitation* 76, 27-32.
- Riley, P.O., Paolini, G., Croce, U.D., Paylo, K.W., Kerrigan, D.C., 2007. A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects. *Gait & Posture* 26, 17-24.
- SAS Institute. 2009. SAS/STAT 9.2 user's guide, 2nd ed. Cary (NC): SAS Institute.
- Stephenson, J.L., Lamontagne, A., De Serres, S.J., 2009. The coordination of upper and lower limb movements during gait in healthy and stroke individuals. *Gait & Posture* 29, 11-16.

Stephenson, J.L., De Serres, S.J., Lamontagne, A., 2010. The effect of arm movements on the lower limb during gait after a stroke. *Gait & Posture* 31, 109-115.

Stolze, H., Kutz-Buschbeck, J.P., Mondwurf, C., Boczek-Funcke, A., Jöhnk, K., Deuschl, G., Illert, M., 1997. Gait analysis during treadmill and overground locomotion in children and adults. *Electroencephalography and clinical Neurophysiology* 105, 490-497.

Umberger, B.R., 2008. Effects of suppressing arm swing on kinematics, kinetics, and energetic of human walking. *Journal of Biomechanics* 41, 2575-2580.

White, S.C., Tucker, C.A., Lifeso, R.M., 1996. Verbal feedback cues for altering vertical ground reaction forces patterns in gait re-education training. *Gait & Posture* 4, Issue 2, 206.

**Chapter 3**  
**EFFECTS OF CONTROLLED HANDRAIL FORCES ON GAIT**  
**KINEMATICS**

**3.1 Introduction**

Gait analysis is a useful approach to investigating the walking pattern of humans. Comparing the changes in kinetics and kinematics between healthy and pathological gait provides insight to improve gait deviations through rehabilitation. Successful performance of gait relies on the ability to maintain support of body weight during stance phase (Olney et al., 1996, Winter et al., 1991). When a patient has a stroke, it often leaves them impaired on one side of their body. If the impaired limb is not able to support the body during single support, stroke patients usually rely on their non-paretic upper extremity and an outside source (such as a cane or handrail) to provide the stability needed (Jørgensen et al., 1995). This stabilization (using an outside source such as handrail and harness) has been found to improve the gait deviations (Chen et al., 2005). However, this stabilization also means that there are external forces being applied by the upper extremity during the gait cycle. It is unclear how forces applied by the upper extremity affect gait parameters.

Previous studies looking at handrail usage (Siler et al., 1997, Chapdelaine et al., 2005, Reeves et al., 2008, Jeka et al., 1994) did not consider the forces exerted on the handrail. In healthy adults, Siler (1997) concluded that holding handrails did

not alter sagittal plane gait kinematics. Stephenson et al. (2009) also determined that the arm-leg coordination during walking was unaffected by the use of the moving handles. In this study, the vertical forces exerted on the handle by healthy subjects during their comfortable speeds were  $2.7 \pm 1.5\%$  body weight, while high functioning stroke subjects used up to  $4.9 \pm 2.9\%$  body weight on the handrails and force magnitude increased with speed. Both groups took shorter and more frequent strides when they weren't using the handles. It appears that arm movement during treadmill walking influences stride characteristics and lower limb muscle activation patterns, but not joint kinematics (Stephenson et al., 2010). Although the presence of handrails does not affect the joint kinematics, the amount of force applied to the handrail may influence these parameters.

Previous studies have suggested that trunk sway may be related to balance and stability (Janssen et al., 2009, Jeka et al., 1994). Trunk list, in the frontal plane, and trunk tilt, in the sagittal plane, may reflect center of mass stability but the effects of holding a single handrail (left handrail only or right handrail only) or applying a controlled amount of force to the handrails have not been studied.

Controlling the amount of force applied to the handrail can allow us to quantify the amount of handrail force required to cause a significant difference in gait parameters. This can provide insight into the maximum amount of handrail force possible before the kinetics and kinematics are significantly different from the no handrail condition. The ability to monitor handrail forces may make it possible to limit the amount of force applied by patients who rely on handrails. Feedback systems have previously been used to improve balance during gait retraining (Rougier, 2004, White et al., 1996, Janssen et al., 2009, Barrios et al., 2010) and rehabilitation of the

upper extremity (Wolf et al., 1983, Bowman et al., 1979). The common goal of these studies was to change the habits of a subject by using a feedback system, using verbal or visual feedback.

The objective of this study was to determine if applying a known force to the handrail causes a change in gait kinematics. We compared (1) the hip angles, (2) the trunk list in the frontal plane, and (3) the trunk tilt in the sagittal plane under three conditions (no hands on handrail (NHR), five percent of the subject's body weight applied to the left handrail (LHR), and five percent of the subject's body weight applied to the right handrail (RHR)). Controlling the amount of force applied to the handrail provides a consistent comparison between these two conditions without having to take into account the variation of handrail force between subjects. This will allow use to determine if there is a significant change in gait parameters when 5% BW is applied to the handrail.

## **3.2 Methods**

### **3.2.1 Subjects**

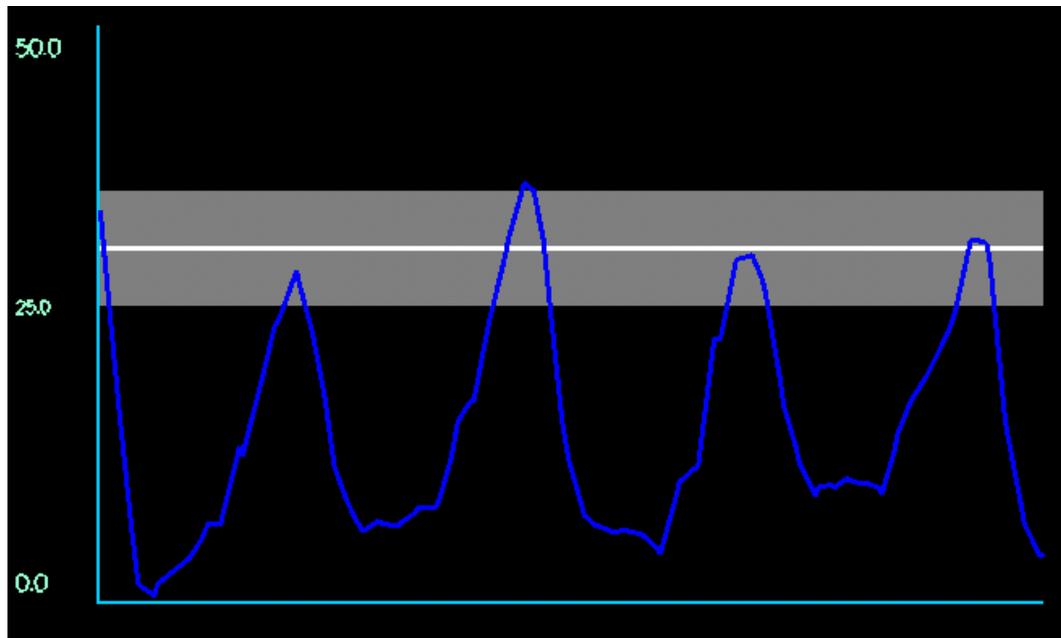
Twenty two individuals participated in this study. These men and women, between 18 and 40 years of age, had no history of muscle, bone or nervous system disorders. Potential subjects were asked to complete a physical activity readiness questionnaire and sign an informed consent approved by the Human Subjects Review Board at the University of Delaware prior to participating. The subject's height, weight, walking speed, and hand dominance were recorded (Table 3.1).

**Table 3.1 Subject Information (mean  $\pm$  st dev)**

|                                    |                   |
|------------------------------------|-------------------|
| Age                                | 24 $\pm$ 4.94     |
| Gender                             | 9 F, 13 M         |
| Height (m)                         | 1.73 $\pm$ 0.11   |
| Weight (kg)                        | 72.75 $\pm$ 12.80 |
| Walking Speed (m*s <sup>-1</sup> ) | 1.26 $\pm$ 0.15   |
| Hand Dominance                     | 20 R, 2 L         |

### **3.2.2 Equipment and Experimental Procedure**

An instrumented split-belt treadmill (Bertec Corp., Columbus, OH) was used during this experiment. Two force plates (embedded in the treadmill) and two force transducers (embedded in the handrails) captured the ground reaction forces and handrail forces, respectively, at 1080 Hz. Forces were collected in all three planes, but only the vertical forces were evaluated. All trials were recorded in Cortex (version 1.0.0.198) by eight cameras (Motion Analysis Corp., Santa Rosa, CA) that capture motion of reflective markers attached to the body segments at 60 Hz. The handrail forces were converted from Volts to Newtons and output onto a screen visible to the subjects using Visual 3D. The graph, displaying the handrail forces, in the vertical direction, had a white line representing five percent of the subject's body weight and a gray box indicating  $\pm$  5 N (Figure 3.1).



**Figure 3.1** The real time feedback display available to the subjects. The blue line represented the handrail force the subject exerted on the left handrail, the white line represented 5% of the subject's body weight and the gray area was  $\pm 5$  N.

The subject's self-selected speed was calculated by timing the subjects as they walked overground for 10 meters. This was done twice and their average was considered to be their self-selected speed and used throughout all of the trials. Individual markers were then attached to the subject's anatomical landmarks of the upper and lower bodies at the sternum, an offset on the back, shoulders, upper arms, elbows, lower arms, wrists, hands, right and left ASIS, sacrum, thighs, medial and lateral knees, shanks, medial and lateral ankles, toes, and heels. Three standing trials were completed before the walking tasks. The subjects were then asked to walk under three conditions (no hands on handrail (NHR), left hand only on the handrail (LHR), and right hand only on the handrail (RHR)). Under the conditions where the handrails were being used, the subject's goal was to apply five percent of their body weight (BW) to it. The subjects were able to monitor the forces they applied to the handrails

with the Real-Time plug in (Visual 3D). These subjects were instructed to “apply enough of your weight to the handrail so that your peak force is as close to the white line as possible.” Each subject was allotted approximately two minutes to become familiar with the amount of force necessary to produce the peak force within the acceptable range. The order of these trials was randomized and each trial was recorded for 30 seconds.

### **3.2.3 Data Analysis**

The data were checked for marker drop outs and marker accelerations in Cortex. This program was also used to delete poor data points from the data set and pattern recognition was used to connect the gaps, before exporting as C3D files.

A model was created in Visual 3D including feet, legs, torso, arms and hands with segments scaled to subject height and weight. The targets were filtered using a Butterworth lowpass filter with a cut off frequency of 6 Hz. The handrail data were imported into Visual 3D and converted to Newtons. The handrail and force plate data were both normalized to body weight. A Butterworth lowpass filter with a cut off frequency of 30 Hz was used on the force plate and handrail data. Gait events were identified (1) when the right foot hits the force plate (RON), (2) when the right foot leaves the force plate (ROFF), (3) when the left foot hits the force plate (LON) and (4) when the left foot leaves the force plate (LOFF). Each subject’s trials were averaged over gait cycles, from RON to RON for the right hip flexion angles, trunk list, in the frontal plane, and trunk tilt, in the sagittal plane. The left hip angles were averaged over the gait cycles from LON to LON. The trunk list and trunk tilt were calculated relative to the virtual coordinate system (created so the subject walked in the y-direction) to the trunk. All trials were then normalized to 101 points. The four

kinematic parameters (left hip angles, right hip angles, trunk list in the frontal place, and trunk tilt in the sagittal plane) for each trial were exported to Excel and averaged among subjects for each condition.

The maximum flexion, maximum extension, and excursion of the right and left hip were analyzed between three conditions (NHR, RHR, and LHR) using a one way analysis of variance (ANOVA) with repeated measures. The maximum and minimum trunk tilt in the frontal plane between (1) RHR vs. NHR and (2) LHR vs. NHR were evaluated using paired t-tests. In the sagittal plane, a one way repeated measures ANOVA was used to compare the trunk tilt of the three conditions (NHR, LHR, and RHR). A paired t-test was also used to determine if there was a difference between the RHR and LHR trunk list in the sagittal plane. Statistical analyses were performed using SAS Institute (2009) software and a significant difference was taken to be  $p < .05$ .

### **3.3 Results**

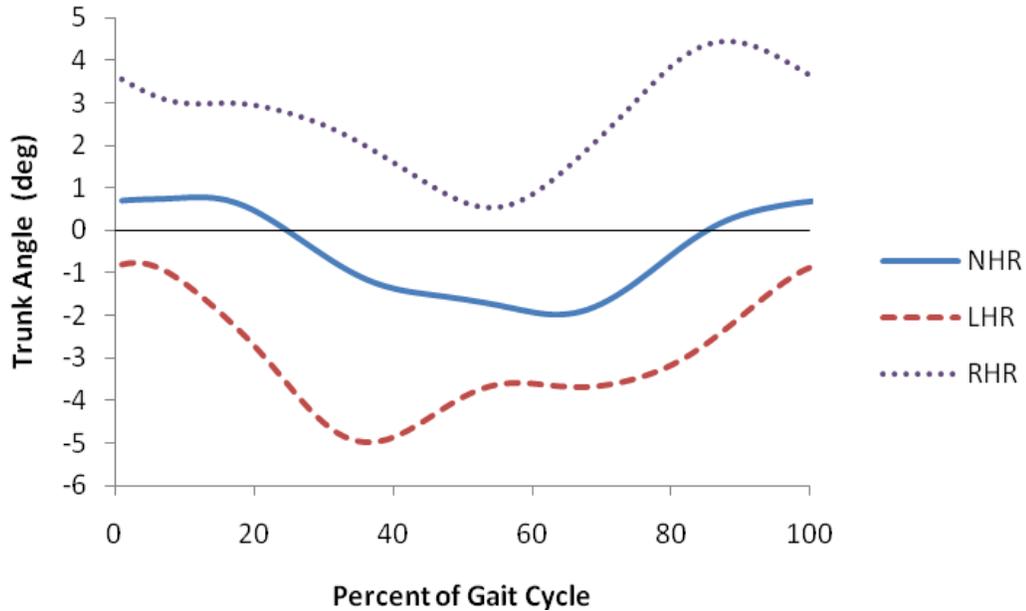
Hip angles for the LHR and RHR conditions, on the left and right side, were within  $\pm 1$  degrees from the NHR condition (Table 3.2 and Table 3.3). There were no significant differences in maximum hip angles between the conditions (left:  $p=0.9043$ , right:  $p=0.8427$ ). Similarly, there were no significance differences between conditions for minimum hip angle (left:  $p=0.5270$ , right:  $p=0.5847$ ) or hip excursion (left:  $p=0.8778$ , right:  $p=0.2918$ ).

**Table 3.2 Left Hip Angle (deg. ± st dev)**

|                       | NHR         | LHR         | RHR         |
|-----------------------|-------------|-------------|-------------|
| Maximum Hip Flexion   | 19.0 ± 6.2  | 19.8 ± 7.2  | 19.2 ± 7.4  |
| Maximum Hip Extension | -18.5 ± 4.8 | -17.2 ± 5.9 | -16.9 ± 5.7 |
| Excursion             | 37.1 ± 4.1  | 37.0 ± 5.5  | 36.1 ± 5.2  |

**Table 3.3 Right Hip Angle (deg. ± st dev)**

|                       | NHR         | LHR         | RHR         |
|-----------------------|-------------|-------------|-------------|
| Maximum Hip Flexion   | 18.4 ± 6.6  | 19.1 ± 7.4  | 19.1 ± 7.7  |
| Maximum Hip Extension | -18.7 ± 4.5 | -17.4 ± 5.0 | -17.0 ± 5.5 |
| Excursion             | 37.0 ± 3.6  | 36.5 ± 4.6  | 36.1 ± 4.7  |



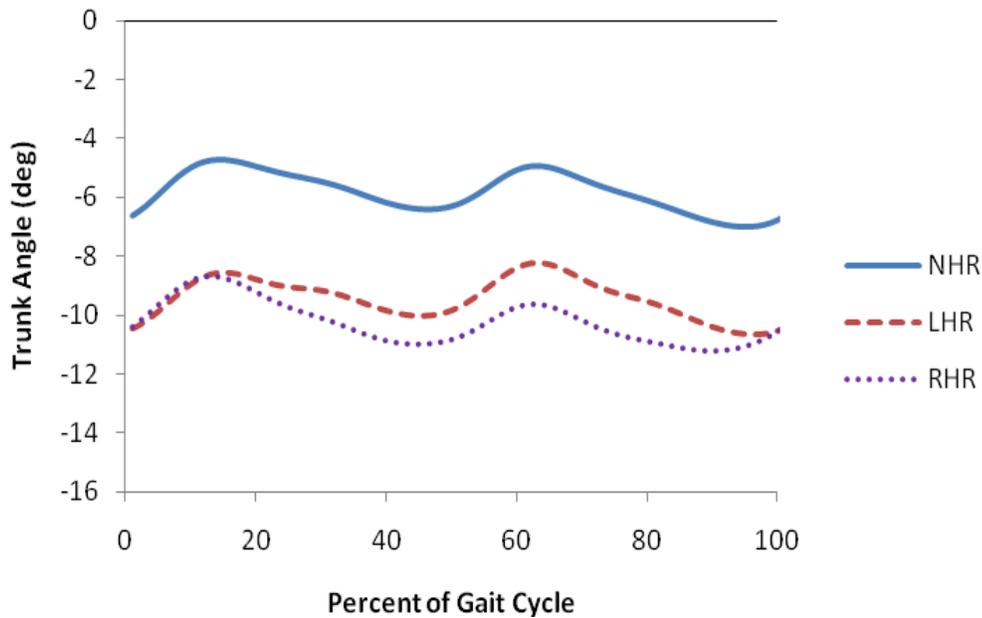
**Figure 3.2** Trunk list in the frontal plane shows significant differences between the conditions where a handrail is used and the no handrail condition. The three conditions, no handrails used (NHR), 5% BW applied to the left handrail (LHR), and 5% BW applied to the right handrail (RHR), were plotted from RON to RON.

**Table 3.4** The minimum and maximum trunk list in the frontal plane.

| Handrail Condition        | Minimum Trunk List (°) | Maximum Trunk List (°) |
|---------------------------|------------------------|------------------------|
| No Handrail (NHR)         | -2.6 ± 1.2             | 1.5 ± 1.8              |
| Left Handrail Only (LHR)  | -5.1 ± 3.3             | -0.6 ± 3.3             |
| Right Handrail Only (RHR) | 0.4 ± 3.7              | 4.5 ± 3.6              |

The trunk list in the frontal plane and trunk tilt in the sagittal plane showed much larger differences than the hip angles. In the frontal plane, we found a significant difference between the maximum trunk list ( $p=0.002$ ) and the minimum trunk list ( $p=0.001$ ) between the RHR and NHR conditions (Figure 3.2, Table 3.4). Similarly, there was a significant difference between the LHR and NHR conditions for both the maximum and minimum trunk list (max:  $p=0.013$ , min:  $p=0.002$ ).

Unlike the frontal plane, the RHR and LHR conditions in the sagittal plane had similar values (Figure 3.3, Table 3.5). There were no significant differences between the maximum and minimum trunk tilt for these two conditions (maximum:  $p=0.052$ , minimum:  $p=0.356$ ). However, we found significant differences for both the maximum and minimum trunk tilt (maximum:  $p=0.042$ , minimum:  $p=0.035$ ) when all three conditions were compared (NHR, LHR, and RHR).



**Figure 3.3** Trunk tilt in the sagittal plane demonstrates that subjects lean forward during both conditions where they were applying handrail forces. All trials were plotted from RON to RON.

**Table 3.5** The minimum and maximum forward tilt in the sagittal plane.

| Handrail Condition        | Minimum Forward Tilt (°) | Maximum Forward Tilt (°) |
|---------------------------|--------------------------|--------------------------|
| No Handrail (NHR)         | $-4.2 \pm 5.0$           | $-7.2 \pm 5.1$           |
| Left Handrail Only (LHR)  | $-7.9 \pm 6.0$           | $-10.9 \pm 5.8$          |
| Right Handrail Only (RHR) | $-8.5 \pm 6.7$           | $-11.7 \pm 6.9$          |

### 3.4 Discussion

Upper extremity forces influence trunk kinematics while having no significant effect on hip angles. The lower limb kinematics remained similar between the different handrail conditions, consistent with the finding of Siler et al. (1997). Determining that there were no significant differences between the NHR condition and the conditions when the handrail was used is important for clinicians because it implies that lower limb kinematics are unaffected when a patient uses a handrail. It is important to note that the hip angles were not changed by the presence of the handrails when a force of five percent of the subject's body weight was applied to the handrail. The lower limb kinematics may be influenced if greater forces are applied by the upper extremity.

Although there were no changes in the hip angles, holding a handrail caused significant differences in trunk list and trunk tilt. In the frontal plane, the subjects altered their trunk list depending on which handrail they were holding (Figure 3.2). When the subjects were holding the right handrail, it caused them to lean towards the right side; similar trends were seen when the left handrail was held. If these changes were seen when a healthy subject applied 5% BW to the handrail, we may expect that stroke subjects who apply a greater force will show a greater change in trunk list. The trunk list, over the gait cycle, does not fluctuate in a similar pattern between conditions. The subjects leaned toward the left handrail during the LHR condition and reached a maximum lean during right stance. Similarly, during the RHR condition, the subjects leaned toward the right handrail and reached their maximum lean during left stance. Although maximum trunk list occurred at different times during the gait cycle, the excursions of all of the conditions were approximately the same (NHR:  $4.0 \pm 1.8^\circ$ , LHR:  $4.5 \pm 1.2^\circ$ , RHR:  $4.5 \pm 1.5^\circ$ ). This suggests that

although the magnitudes of the peaks differ between the conditions, the range remains the same.

The trunk tilt in the sagittal plane also showed significant differences between conditions. In both conditions where a force is applied to the handrail, it caused the subjects to lean forward. The statistics demonstrated that there were no significant differences between the trunk tilt in the LHR and RHR conditions, showing that subjects compensate for the upper extremity force the same way in the sagittal plane, regardless of which side is loaded. The changes of the trunk list and trunk tilt may not influence the hip angle, but they may influence the knee and ankle kinematics and the balance of the subjects. For clinicians, this might mean that their patients may improve their gait with handrail usage but not their stability. Incorporating a method to reduce the reliance on handrails as well as improving their gait parameters should be a future direction for patients who rely on an outside source of support.

Five percent of the subject's body weight was chosen because it was the average of healthy subject's comfortable handrail hold. This was taken from the average handrail forces exerted by the healthy subjects during the first part of the study (Aim 1, Chapter 2). It was important to control the amount of handrail force exerted because this force varied between subjects. When instructed to comfortably hold the handrail, some healthy subjects relied heavily on the handrails, while others lightly touched it. Controlling the handrail force with a feedback system eliminated the handrail force as a variable between subjects. The kinematics evaluated during this study was only influenced by the handrail conditions.

Five percent body weight applied to the handrail may be less than that used by low functioning stroke patients. Chen et al. (2001) determined that when

stroke patients held a cane,  $12.7 \pm 4.7\%$  BW was applied by the cane in the vertical direction. Stephenson et al. (2009) found that only  $2.7 \pm 1.5\%$  body weight was exerted on the handrails by high functioning stroke patients. Future studies should take this into account and evaluate the lower limb kinematics when a larger percentage of body weight is applied to the handrail.

### **3.4.1 Limitations/Future Directions**

One limitation of the current study was that healthy subjects do not depend on the handrails. It is possible that some subjects used their muscle strength to manipulate the handrail forces by making their arms pulse to reach the targeted force, instead of their body weight. Prior to the study, different percentages of body weight were tested, such as 5%, 10%, and 15% BW, for handrail force. The healthy subjects were able to apply the 10% BW condition, but the 15% BW was too difficult for the subjects to perform. Thus, the forces exerted by stroke patients were not able to be replicated by the healthy subjects. In addition, the subjects did not reach 5% BW during every gait cycle. Some gait cycles fell within the  $\pm 5$  N that was supplied as the standard deviation. This standard deviation was generous for the subjects who weighed less than the average. Refining this allowed range may improve the accuracy of the results. Future studies might vary the percentage of body weight applied to the handrails to quantify when the handrail force significantly affects the lower limb kinematics. A feedback system, similar to the one used in this study, may be useful to train patients to apply less force to the handrails throughout their rehabilitation process.

### **3.5 Conclusion**

In conclusion, the hip angles, trunk list, and trunk tilt were evaluated while a controlled force was applied to the handrail in healthy subjects. The right and left hip angles were unaffected by the presence of handrails. Trunk list in the frontal plane and trunk tilt in the sagittal plane, showed significant differences between the conditions where a handrail was present (left or right) and absent. This significant difference is evident when only five percent of the subject's body weight is applied to the handrail. It can be inferred that subjects who apply a greater handrail force, such as stroke subjects, have a greater change in trunk list and trunk tilt. A real-time feedback system can be useful in training stroke patients to reduce the forces they apply on the handrails.

### 3.6 References

- Barrios, J.A., et al., Gait retraining to reduce the knee adduction moment through real-time visual feedback of dynamic knee alignment. *Journal of Biomechanics* (2010), doi:10.1016/j.jbiomech.2010.03.040
- Bowman, B.R., Baker, L.L., Water, R.L., 1979. Positional feedback and electrical stimulation: An automated treatment for the hemiplegic wrist. *Archives of Physical medicine and Rehabilitation*, 497-502.
- Chapdelaine, S., McFadyen, B.J., Nadeau, S., St-Vincent, G., 2005. Instrumented staircase for kinetic analyses of upper- and lower-limb function during stair gait. *Medical & Biological Engineering and Computing* 43, 552-556.
- Chen, C.L., Chen, H.C., Wong, M.K., Tang, F.T., Chen, R.S., 2001. Temporal stride and force analysis of cane-assisted gait in people with hemiplegic stroke. *Archives of physical medicine and rehabilitation* 82, 43-48.
- Chen, G., Patten, C., Dhara, H.K., Zajac, F.E., 2005. Gait deviations associated with post-stroke hemiparesis: improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold. *Gait & Posture* 22, 57-62.
- Janssen, L.J.F., Verhoeff, L.L., Horlings, C.G.C., Allum, J.H.J., 2009. Directional effects of biofeedback on trunk sway during gait tasks in healthy young subjects. *Gait & Posture* 29, 575-581.
- Jeka, J.J., Lackner, J.R., 1994. Fingertip contact influences human postural control. *Experimental Brain Research* 100, 495-502.
- Jørgensen, H.S., Nakayama, H., Raaschou, H.O., Olsen, T.S., 1995. Recovery of walking function in stroke patients: The Copenhagen stroke study. *Archives of Physical Medicine and Rehabilitation* 76, 27-32.
- Moe-Nilssen, R., Helbostad, J.L., 2004. Estimation of gait cycle characteristics by trunk accelerometry. *Journal of Biomechanics* 37, 121-126.
- Olney, S.J., Richards, C., 1996. Hemiparetic gait following stroke. Part I: Characteristics. *Gait & Posture* 4, 136-148.

Reeves, N.D., Spanjaard, M., Mohagheghi, A.A., Baltzopoulos, V., Maganaris, C.N., 2008. Influence of light handrail use on the biomechanics of stair negotiation in old age. *Gait & Posture*, doi: 10.1016/j.gaitpost.2008.01.014

Rougier, P., 2004. Optimising the visual feedback technique for improving upright stance maintenance by delaying its display: behavioural effects on healthy adults. *Gait & Posture* 19, 154-163.

SAS Institute. 2009. SAS/STAT 9.2 user's guide, 2nd ed. Cary (NC): SAS Institute.

Siler, W.L., Jørgensen, A.L., Norris, R.A., 1997. Grasping the handrails during treadmill walking does not alter sagittal plane kinematics of walking. *Archives of Physical Medicine and Rehabilitation* 78, 393-398.

Stephenson, J.L., Lamontagne, A., De Serres, S.J., 2009. The coordination of upper and lower limb movements during gait in healthy and stroke individuals. *Gait & Posture* 29, 11-16.

Stephenson, J.L., De Serres, S.J., Lamontagne, A., 2010. The effect of arm movements on the lower limb during gait after a stroke. *Gait & Posture* 31, 109-115.

White, S.C., Tucker, C.A., Lifeso, R.M., 1996. Verbal feedback cues for altering vertical ground reaction forces patterns in gait re-education training. *Gait & Posture* 4, Issue 2, 206.

Winter, D.A.. *The Biomechanics and Motor Control of Human Gait: Normal, Elderly, and Pathological*. Ontario: Waterloo Biomechanics, 1991.

Wolf, S.L., Binder-Macleod, S.A., 1983. Electromyographic biofeedback applications to the hemiplegic patient. *Physical Therapy: Journal of the American Physical Therapy Association* 63, 1393-1403.

## **Chapter 4**

### **CONCLUSIONS**

The objectives of this thesis were to determine if handrail usage affected the (1) kinetics during the four handrail conditions (NHR, BHR, LHR, and RHR) and (2) kinematics when a handrail force of 5% BW was applied in the vertical direction. We found that handrail use does influence ground reaction forces. There were no significant differences in the handrail forces during the four conditions (NHR, BHR, LHR, and RHR). The hip angles did not change with the presence of a 5% BW handrail force, however trunk tilt and trunk list were significantly different between the conditions (NHR, 5% BW applied to right handrail, and 5% BW applied to left handrail).

#### **4.1 AIM 1: Analyze how the forces are distributed between the ground and the handrails during treadmill walking.**

Hypothesis 1.1: The most ground reaction force is seen when there are no handrails used and this ground reaction force decreases with an increase in the number of handrails used.

Hypothesis 1.2: When both handrails are used, the subject distributes an equal amount of weight on both handrails.

Hypothesis 1.3: When only one handrail is used, the same handrail force is applied by the subject.

As expected, the ground reaction forces during the NHR condition were higher than the conditions where the handrails were present (LHR, RHR, BHR). The differences in ground reaction forces between all conditions were statistically significant during the first peak but not during the second peak. This suggests that the subjects exerted more force on the handrail (when available) during the first peak, potentially for stability, and did not rely on the handrail during the second peak, before toe off. The number of handrails used does significantly influence the ground reaction forces, supporting Hypothesis 1.1. The difference in the ground reaction forces between conditions may lead to inaccuracies in inverse dynamics calculations.

Although there was a significant difference between the ground reaction forces, there were no significant differences in handrail forces between conditions. The average handrail forces for the left and right side, during the BHR condition, were not equal; however the difference was not large enough to show significance, supporting Hypothesis 1.2. The small difference in handrail forces may be due to hand dominance. Twenty three of the subjects were right handed, however handedness was not taken into account when comparing the handrail forces. Similarly to the BHR condition, the force exerted on the right handrail during the condition when only one handrail was held was larger (but not significantly different) than the force exerted on the left handrail. Again, handedness was not taken into account and the difference between these handrail forces did not show significance. When handrails were present, the ground reaction forces were not affected by the side on which the handrail force was applied, supporting Hypothesis 1.3.

The significant difference in kinetics between the handrail conditions should be important to clinicians. This healthy data can provide insight into the

‘typical’ handrail forces applied during different conditions by subjects to give the clinician a standard when looking at the amount of handrail force applied by a subject with pathological gait. Understanding how the handrail conditions influence ground reaction forces can hint at how much force the subject is removing from their lower limbs. By retraining the subjects not to rely on handrails, it will make them apply more force to their lower limbs and may improve symmetry of their ground reaction forces.

**4.2 AIM 2: Compare hip angle, trunk tilt, and trunk list between two conditions: (1) no handrail is used and (2) five percent of body weight (BW) is applied in the vertical direction on the handrail.**

Hypothesis 2.1: Hip angles are largest when no handrails are used.

Hypothesis 2.2: In the sagittal plane, the trunk tilt is the same under both conditions

Hypothesis 2.3: In the frontal plane, the trunk creates a larger angle with the midline during the second condition than the first condition.

It was determined that there was no significant difference between the hip angles during the no handrail condition and the 5% BW condition. This suggests that the other lower limb angles, such as the knee and ankle angles do not differ either. This study provides a validation that lower limb kinematics can be compared between subjects who do not rely on handrails and subjects who use handrails but apply 5% BW or less. In previous studies, it was not possible to determine if there were relationships between the magnitude of the handrail force and ground reaction force or kinematics because the force applied to the handrail was unknown (Chapdelaine et al., 2005, Reeves et al., 2008, Jeka et al., 1994, Siler et al., 1997). For example,

suggesting that everyone apply ‘finger tip touch’ to the handrail and assuming all subjects apply the same force through their fingers may lead to erroneous inverse dynamics calculations. Future studies can use a feedback system to monitor the handrail forces applied by the subjects.

The trunk list in the frontal plane and trunk tilt in the sagittal plane both show significant differences when 5% BW was applied to the handrail, partially supporting Hypothesis 2.2. The change in angles show that healthy subjects lean towards the side that is using the handrail and forward in order to apply the required force, consistent with Hypothesis 2.3. However, hip angles do not change between conditions, suggesting that the stride length remains the same. Hypothesis 2.1 is not supported by this data. If a healthy subject has to lean to produce 5% BW, it can be inferred that a subject who is reliant on the handrail and applies a greater force than 5% BW will lean more. Other assistive devices such as canes and walkers may also influence their trunk list and trunk tilt.

During rehabilitation the most common goal of stroke patients is restoration of independent gait (Bohannon et al., 1988). Many stroke studies evaluate the symmetry of gait and suggest that rehabilitation should focus on the kinematics of the lower limbs. While symmetry is an important factor, posture and stability may be another factor that should be taken into consideration. If the subject felt stable without the use of assistive devices, their reliance on these devices may decrease. When they are retrained to stand up straight and pay more attention to their posture, the symmetry of their gait may follow.

The results for the second aim were collected using instrumented handrails and a visual feedback system. The handrail forces exerted during the trials

were streamed onto a screen visible to the subject in real time. In order to produce a force of 5% BW, the subjects altered their handrail forces according to the screen. They were able to maintain a 5% BW peak handrail force using this feedback system. It provided the subjects with a visual and more tangible idea of how much force they were applying. It supplied a target for them to reach, as well as a sense of how much error was between the force exerted and the target force. Stroke subjects can be taught to apply less force to the handrail using a visual feedback system, which could improve their balance. In unpublished data, we found that stroke subjects applied handrail forces between 7% BW and 20% BW. This range suggests that the severity of the stroke and success of rehabilitation may play a major role in how dependent stroke subjects are on the handrail.

Instrumented handrails combined with a real time feedback system allow the clinician to know how much force the subjects are applying to the handrail. This is important for clinicians who are trying to get the subjects to apply more force to their impaired lower limb. Therefore, this feedback system is not only helpful for the patient, but also provides instant feedback to the clinicians and allows them to give audio instructions as well as the visual feedback received from the monitor.

This study provides a solid basis for future studies involving handrail use and pathological gait. The results demonstrate the importance of taking handrail use into account when analyzing kinetic data and calculating inverse dynamics. It is also necessary to keep the handrail forces under 5% BW, until future studies determine if the differences between the no handrail condition and conditions with a higher handrail force remain insignificant, in order to compare lower limb kinematics between subjects. The 5% BW cut off is subject to change provided that more studies

are performed analyzing the lower limb kinematics when more body weight is applied. Other studies and calculations should be performed to provide support to the inferences made (e.g. the ankle and knee angles) and on subjects with a pathology to determine if the kinetics and kinematics behave in a similar manner.

### 4.3 References

Bohannon, R.W., Andrews, A.W., Smith, M.B., 1988. Rehabilitation goals of patients with hemiplegia. *International Journal of Rehabilitation Research* 11, 181-183

Chapdelaine, S., McFadyen, B.J., Nadeau, S., St-Vincent, G., 2005. Instrumented staircase for kinetic analyses of upper- and lower-limb function during stair gait. *Medical & Biological Engineering and Computing* 43, 552-556.

Jeka, J.J., Lackner, J.R., 1994. Fingertip contact influences human postural control. *Experimental Brain Research* 100, 495-502.

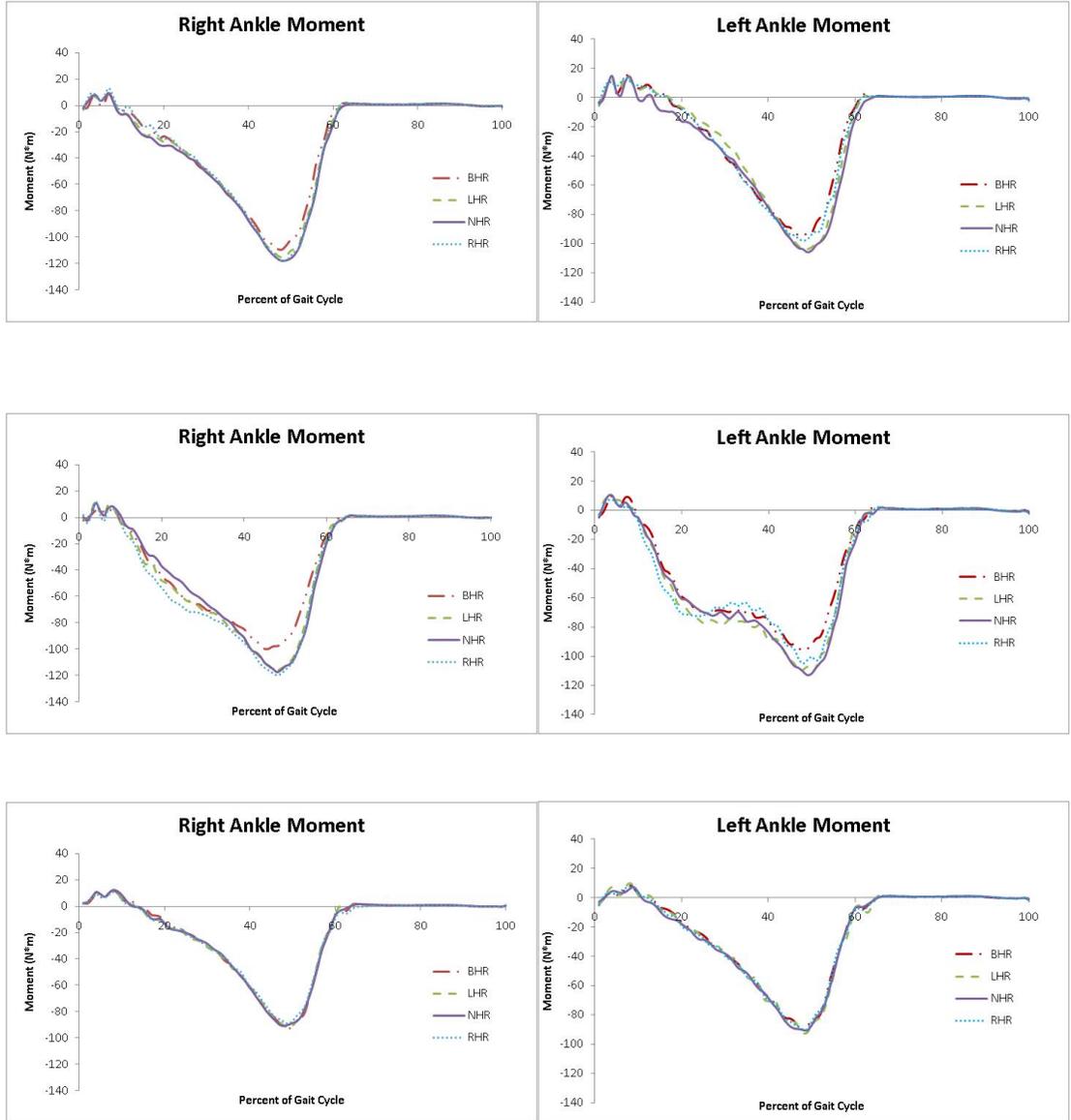
Reeves, N.D., Spanjaard, M., Mohagheghi, A.A., Baltzopoulos, V., Maganaris, C.N., 2008. Influence of light handrail use on the biomechanics of stair negotiation in old age. *Gait & Posture*, doi: 10.1016/j.gaitpost.2008.01.014

Siler, W.L., Jorgensen, A.L., Norris, R.A., 1997. Grasping the handrails during treadmill walking does not alter sagittal plane kinematics of walking.

## APPENDIX

### **Inverse Dynamic Calculations:**

Chapter 2 focused on the differences in the ground reaction forces between four conditions (BHR, LHR, NHR, and RHR). It was shown that there was a significant difference between the first peaks of the right ground reaction forces. These differences in the ground reaction forces may lead to inaccuracies in inverse dynamics calculations. The right and left ankle moments were determined for three of the subjects to show that there were variations in ankle moments between conditions. The ankle moments were calculated in Visual 3D and normalized to one gait cycle. The right and left ankle moments were plotted for each of the three subjects, to highlight the differences between the four conditions. Two of the subjects had variations in the magnitude of their ankle moments. However, the ankle moments followed the same trend for all of the conditions. The purpose of these calculations was to show that the inverse dynamics calculations need to be analyzed further to determine if handrails significantly alter these values. If there are significant differences between the conditions, it is important to take this into account when comparing these parameters between subjects.



**Figure A.1** Right and left ankle moments for three of the subjects normalized to one gait cycle. The right ankle moments were plotted from RON to RON. The left ankle moments were plotted from LON to LON.

## BIBLIOGRAPHY

- Ada, L., Dean, C.M., Vargas, J., Ennis, S., 2010. Mechanically assisted walking with body weight support results in more independent walking than assisted overground walking in non-ambulatory patients early after stroke: a systematic review. *Australian Journal of Physiotherapy* 56, 153-161.
- Alton, F., Baldey, L., Caplan, S., Morrissey, M.C., 1998. A kinematic comparison of overground and treadmill walking. *Clinical Biomechanics* 13, 434-440.
- Barrios, J.A., et al., Gait retraining to reduce the knee adduction moment through real-time visual feedback of dynamic knee alignment. *Journal of Biomechanics* (2010), [doi:10.1016/j.jbiomech.2010.03.040](https://doi.org/10.1016/j.jbiomech.2010.03.040)
- Bohannon, R.W., Andrews, A.W., Smith, M.B., 1988. Rehabilitation goals of patients with hemiplegia. *International Journal of Rehabilitation Research* 11, 181-183
- Bowman, B.R., Baker, L.L., Water, R.L., 1979. Positional feedback and electrical stimulation: An automated treatment for the hemiplegic wrist. *Archives of Physical medicine and Rehabilitation*, 497-502.
- Chang, M.D., Shaikh, S., Chau, T., 2009. Effect of treadmill walking on the stride interval dynamics of human gait. *Gait & Posture* 30, 431-435.
- Chapdelaine, S., McFadyen, B.J., Nadeau, S., St-Vincent, G., 2005. Instrumented staircase for kinetic analyses of upper- and lower-limb function during stair gait. *Medical & Biological Engineering and Computing* 43, 552-556.
- Chen, C.L., Chen, H.C., Wong, M.K., Tang, F.T., Chen, R.S., 2001. Temporal stride and force analysis of cane-assisted gait in people with hemiplegic stroke. *Archives of physical medicine and rehabilitation* 82, 43-48.
- Chen, G., Patten, C., Kothari, D.H., Zajac, F.E., 2005. Gait deviations associated with post-stroke hemiparesis: improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold. *Gait & Posture* 22, 57-62.
- Eke-Okoro, S.T., Gregoric, M., Larsson, L.E., 1997. Alterations in gait resulting from deliberate changes of arm-swing amplitude and phase. *Clinical Biomechanics* 12, 516-521

- Fitoussi, F., Maurel, N., Diop, A., Laassel, E.M., Ilharreborde, B., Presedo, A., Mazda, K., Penneçot, G.F., 2009. Upper extremity kinematics analysis in obstetrical brachial plexus palsy. *Orthopaedics & Traumatology: Surgery & Research* 95, 336-342
- Ford, M.P., Wagenaar, R.C., Newell, K.M., 2007. Arm constraint and walking in healthy adults. *Gait & Posture* 26, 135-141.
- Hingtgen, b., McGuire, J.R., Wang, M., Harris, G.F., 2006. An upper extremity kinematic model for evaluation of hemiparetic stroke. *Journal of Biomechanics* 39, 681-688.
- Janssen, L.J.F., Verhoeff, L.L., Horlings, C.G.C., Allum, J.H.J., 2009. Directional effects of biofeedback on trunk sway during gait tasks in healthy young subjects. *Gait & Posture* 29, 575-581.
- Jeka, J.J., Lackner, J.R., 1994. Fingertip contact influences human postural control. *Experimental Brain Research* 100, 495-502.
- Jørgensen, H.S., Nakayama, H., Raaschou, H.O., Olsen, T.S., 1995. Recovery of walking function in stroke patients: The Copenhagen stroke study. *Archives of Physical Medicine and Rehabilitation* 76, 27-32.
- Mickelborough, J., van der Linden, M.L., Richards, J., Ennos, A.R., 2000. Validity and reliability of a kinematic protocol for determining foot contact events. *Gait and Posture* 11, 32-37.
- Moe-Nilssen, R., Helbostad, J.L., 2004. Estimation of gait cycle characteristics by trunk accelerometry. *Journal of Biomechanics* 37, 121-126.
- Murray, M.P., Spurr, G.B., Sepic, S.B., Gardener, G.M., Mollinger, L.A., 1985. Treadmill vs. floor walking: kinematic, electromyogram, and heart rate. *Journal of Applied Physiology* 59, 87-91.
- Olney, S.J., Richards, C., 1996. Hemiparetic gait following stroke. Part I: Characteristics. *Gait & Posture* 4, 136-148.
- Reeves, N.D., Spanjaard, M., Mohagheghi, A.A., Baltzopoulos, V., Maganaris, C.N., 2008. Influence of light handrail use on the biomechanics of stair negotiation in old age. *Gait & Posture*, doi: 10.1016/j.gaitpost.2008.01.014
- Richards, C.L., Olney, S.J., 1996. Hemiparetic gait following stroke. Part II: Recovery and physical therapy. *Gait & Posture* 4, 149-162.

- Riley, P.O., Paolini, G., Croce, U.D., Paylo, K.W., Kerrigan, D.C., 2007. A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects. *Gait & Posture* 26, 17-24.
- Roerdink, M., Coolen, B.H., Clairbois, B.H.E., Lamoth, C.J.C., Beek, P.J., 2008. Online gait even detection using a large force platform embedded in a treadmill. *Journal of Biomechanics* 41, 2628-2632.
- Rougier, P., 2004. Optimising the visual feedback technique for improving upright stance maintenance by delaying its display: behavioural effects on healthy adults. *Gait & Posture* 19, 154-163.
- SAS Institute. 2009. SAS/STAT 9.2 user's guide, 2<sup>nd</sup> ed. Cary (NC): SAS Institute.
- Siler, W.L., Jørgensen, A.L., Norris, R.A., 1997. Grasping the handrails during treadmill walking does not alter sagittal plane kinematics of walking. *Archives of Physical Medicine and Rehabilitation* 78, 393-398.
- Stephenson, J.L., Lamontagne, A., De Serres, S.J., 2009. The coordination of upper and lower limb movements during gait in healthy and stroke individuals. *Gait & Posture* 29, 11-16.
- Stephenson, J.L., De Serres, S.J., Lamontagne, A., 2010. The effect of arm movements on the lower limb during gait after a stroke. *Gait & Posture* 31, 109-115.
- Stolze, H., Kutz-Buschbeck, J.P., Mondwurf, C., Boczek-Funcke, A., Jöhnk, K., Deuschl, G., Illert, M., 1997. Gait analysis during treadmill and overground locomotion in children and adults. *Electroencephalography and clinical Neurophysiology* 105, 490-497.
- Umberger, B.R., 2008. Effects of suppressing arm swing on kinematics, kinetics, and energetic of human walking. *Journal of Biomechanics* 41, 2575-2580.
- White, S.C., Tucker, C.A., Lifeso, R.M., 1996. Verbal feedback cues for altering vertical ground reaction forces patterns in gait re-education training. *Gait & Posture* 4, Issue 2, 206.
- Winter, D.A.. *The Biomechanics and Motor Control of Human Gait: Normal, Elderly, and Pathological*. Ontario: Waterloo Biomechanics, 1991.
- Wolf, S.L., Binder-Macleod, S.A., 1983. Electromyographic biofeedback applications to the hemiplegic patient. *Physical Therapy: Journal of the American Physical Therapy Association* 63, 1393-1403.