AN ABSTRACT OF THE THESIS OF

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Title: The Accuracy of Pedometers for Adults with Down Syndrome during Controlled and Free-Walking Conditions.

Abstract approved:

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It has been demonstrated that walking is the most common form of physical activity for adults with intellectual disabilities, including Down syndrome (DS). The pedometer is common measurement tool to quantify steps walked, yet there is little evidence of the psychometric properties for individuals with intellectual disabilities, particularly DS. Thus, this work was to provide reliability and validity evidence to determine if pedometers can be used for adults with DS. The first study addressed the accuracy of spring-levered and piezoelectric pedometers under controlled conditions for adults with and without DS. It was determined that there were significant differences in measurement error between adults with and without DS for both pedometer models. Additionally, piezoelectric pedometers were found to be more

accurate than spring-levered pedometers, particularly at slower walking speeds. These differences between pedometer models were also explained by increasing waist-to-hip ratios of individual participants. Absolute error rates for adults with DS were higher than the control group, but may still be acceptable for future use. The second study addressed the reliability and sources of variance in spring-levered and piezoelectric pedometer measurements under free-walking conditions for adults with and without DS. This was conducted using Generalizability (G) theory. Adults with DS demonstrated greater intra-individual variability during the while the control group, conversely, had greater residuals, or unexplained errors. The spring-levered pedometer showed problems with inter-unit variability through substantive variance components. The piezoelectric pedometer demonstrated little systematic error. Additionally, reliability coefficients were calculated for each group and model combination. The piezoelectric pedometer demonstrated higher reliability than the spring-levered based on moderate to high reliability coefficients. Collectively these studies provide evidence that piezoelectric pedometers are more accurate and more reliable than spring-levered pedometers for both adults with and without DS. For future studies measuring walking activity of adults with DS, the use of piezoelectric pedometers is recommended.

The Accuracy of Pedometers for Adults with Down Syndrome during Controlled and Free-Walking Conditions

by

Edward Andrew Pitchford

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I understand that my thesis will become part of the permanent collection of Oregon State University libraries. My signature below authorizes release of my thesis to any reader upon request.

Edward Andrew Pitchford, Author

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CONTRIBUTION OF AUTHORS

Dr. Joonkoo Yun was involved with the conceptualization of the study and research design, data analysis, interpretation of results, and reviewing of the thesis.

Dr. Jeff McCubbin provided useful feedback of the study design and assisted through facilitating the process of recruiting participants.

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THE ACCURACY OF PEDOMETERS FOR ADULTS WITH DOWN SYNDROME DURING CONTROLLED AND FREE-WALKING CONDITIONS

Chapter 1: General Introduction

Down syndrome (DS), a disorder resulting from a chromosomal abnormality on chromosome 21, translocation or mosaicism of chromosomes affects 0.92 per 1,000 births (Roizen, 2002). DS is a symptom complex that is characterized by an intellectual disability, but also by unique body and facial features including obesity and growth stature, muscle hypotonia, joint laxity and a variety of medical conditions including congenital heart disease and atlantoaxial instability (Latash, 2000; Roizen, 2002). Physical activity promotion is an area of need for this population to limit secondary health conditions and other preventable health disparities (Stanish, Temple & Frey, 2006; U.S. Department of Health and Human Services [USDHHS], 2002). There is evidence that individuals with DS may not be at risk for cardiovascular disease to the extent that adults with intellectual disabilities without DS (Draheim, McCubbin & Williams, 2002), but interventions to increase physical activity behaviors are still of need. Walking is considered the most common form of physical activity for individuals with intellectual disabilities (Draheim, Williams & McCubbin, 2002; Stanish & Draheim, 2005a, 2005b; Temple, Anderson & Walkley, 2000). Given that the physical activity habits of individuals with DS are relatively unknown compared to the general population, it is important to have accurate, reliable, and objective measures of walking activity.

The pedometer is a common instrument used to quantify walking activity for practical and research purposes. It is particularly practical for use among populations

with intellectual disabilities as it is a simple, user-friendly, unobtrusive, and relatively inexpensive device (Bassett et al., 1996; Crouter, Schneider, Karabulut, & Bassett, 2003; Le Masurier, Lee, & Tudor-Locke, 2004; Le Masurier & Tudor-Locke, 2003; Manns, Orchard, & Warren, 2007; Schneider, Crouter, & Bassett, 2004; Tudor-Locke & Myers, 2001). There are three types of pedometers, each utilizing different mechanisms. The most common pedometer mechanism is a spring-suspended level arm that records steps as it moves up and down from vertical movement at the hip by connecting an electrical circuit (Crouter et al., 2003; Schneider, Crouter, Lukajic, & Bassett, 2003). An improvement upon this mechanism is a glass-enclosed magnetic reed proximity switch that uses a similar level arm with a magnet that triggers a switch when vertical movement occurs (Crouter et al., 2003; Schneider et al., 2003). The third mechanism is similar to a uni-axial accelerometer that uses a horizontal beam and a piezoelectric-crystal and records steps based on the number of zero-crossings of the instantaneous acceleration versus time curve (Crouter et al., 2003; Schneider et al., 2003). The two most common commercially available pedometer mechanisms are the spring-levered arm and piezoelectric pedometers.

Numerous studies have demonstrated that certain brands and models are more accurate than others (Bassett et al., 1996; Crouter et al., 2003; Schneider et al., 2003, 2004). In general, most pedometers have been found to have acceptable accuracy for measuring steps, and to a lesser degree distance, time and calories expended. However, to author's knowledge, the pedometer has yet to be validated specifically for the DS population.

A study by Stanish (2004) is the closest to providing evidence for adults with Down syndrome. In this study, Stanish found that a spring-levered pedometer was accurate for individuals with intellectual disabilities (ID). The sample of adults with ID included individuals with DS. The results found that the intraclass correlation between pedometer and criterion measured steps was greater than 0.95, indicating high criterion validity.

In addition to the lack of direct evidence of the psychometric properties of pedometers for adults with DS, there are common characteristics of an adult with DS that may create potential measurement errors when using pedometers: a unique gait pattern, slow walking speed, adipose tissue distribution, and other systematic errors.

The gait pattern typically observed among individuals with Down syndrome is described as a shuffling pattern that includes a wider base and increased side to side movement. This gait pattern is most likely the result of the individual overcoming joint laxity and muscle hypotonia through muscle co-contractions and stiffness (Kubo & Ulrich, 2006; Smith, Kubo, Black, Holt, & Ulrich, 2007; Smith & Ulrich, 2008; Ulrich, Haehl, Buzzi, Kubo, & Holt, 2004). Empirically, this gait is characterized by increased variability of the center of mass (COM) in the medio-lateral direction (Agiovlasitis, 2007; Kubo & Ulrich, 2006). This means that the body moves from side to side more during the gait pattern than it would in the general population. This could potentially have an adverse effect on the accuracy of pedometers in two ways. First, since spring-levered pedometers measure vertical displacement at the hip, the increased medio-lateral variability of the DS gait may adversely affect the mechanical functioning of the instrument. Second, many of the piezoelectric pedometers utilize a

sensitivity filter to reduce the counting of "non-steps" (Crouter et al., 2003; Le Masurier et al., 2004; Schneider et al., 2003, 2004). Given the shuffling nature of the gait, there may be insufficient reaction force or vertical acceleration for a piezoelectric pedometer to record steps, thus underestimating walking activity. This could be further amplified by the muscle stiffness and co-contraction that is common in the DS gait. Studies focusing on pedometer accuracy for individuals with walking disabilities have found that gait variability has a significant effect and is negatively correlated with accuracy (Manns, Orchard & Warren, 2007). It is conceivable that the unique gait pattern of adults with DS could have similar effects.

Walking speed has consistently been determined to be a significant source of error in pedometers for the general population (Bassett et al., 1996; Crouter, et al., 2003; Le Masurier & Tudor-Locke, 2003; Le Masurier et al., 2004; Melanson et al., 2004; Schneider, et al., 2003, 2004). While many pedometers have been found to have very high levels of accuracy at speeds of 80 m/min^{-1} and above, most pedometers demonstrate lower, at times significantly lower, levels of accuracy at slower speeds, particularly at 54 m/min^{-1}. This source of error is also due to the lack of vertical hip displacement during slow gait speeds. Studies examining self-paced walking of preadolescents and older adults with DS have reported walking speeds ranging from approximately 40 to 60 m/min^{-1}(Smith et al., 2007; Smith & Ulrich, 2008; Ulrich et al., 2004), compared to the average walking speed for the general population of 96.5 m/min^{-1} (Schneider et al., 2003). Given the characteristically slow walking speed for adults with DS, it is certainly possible that walking activity for individuals with DS is being underestimated due to walking speed.

Individuals with Down syndrome have also been shown to have a body composition characterized by additional adipose tissue in the torso area, in a distinctive pattern that is unique to this population (Roizen, 2002). Studies on pedometer accuracy conducted in the general population on overweight and obese individuals have found that while BMI is not significantly associated with pedometer accuracy, the angle of pedometer tilt is (Crouter, Schneider, & Bassett, 2005; Melanson et al., 2004; Schneider et al., 2003; Swartz, Bassett, Moore, Thompson, & Strath, 2003). The rationale for this error is that for a spring-levered pedometer to work properly, it must be positioned vertically in order for the level arm to move up and down in accordance with vertical movement. A piezoelectric pedometer has been shown to be more resistance to this source of error, but could still be affected if the pedometer tilt is greater than 15 degrees in either direction (Crouter et al., 2005). The body composition of individuals with DS is conducive with creating pedometer tilt, thus creating a source of pedometer error.

Finally, there may also be systematic sources of measurement error that are specific to adults with DS. In all measurement, there are both systematic and random sources of error. For the potential reasons discussed previously, as well as other sources of error, it is possible that any measurement error associated with pedometers and adults with DS is systematic in nature. Furthermore, there may also be systematic errors that are inherent to pedometers, both spring-levered and piezoelectric. It is important to understand and identify these sources that may otherwise be viewed as completely random. This information may be useful for improving measurement with pedometers for adults with and without Down syndrome in the future.

Thus, it is necessary to gather empirical evidence on the accuracy and validity of both spring-levered and piezoelectric pedometers for adults with Down syndrome under both controlled and free-walking conditions so that the most appropriate instrument can be used for the measurement of walking activity. Given the common sources of error that have been shown in the general population, the DS population may present additional and unique potential sources of error.

Statement of the Problem

The purpose of this study was to establish if spring-levered arm and piezoelectric crystal pedometers are accurate measures of walking activity (steps taken) in individuals with Down syndrome. This has been done through: 1) examining the absolute error associated with spring-levered and piezoelectric pedometer measurements under controlled conditions, 2) determining the effects of speed and anthropometric body characteristics on pedometer accuracy, 3) establishing any differences in pedometer accuracy between individuals with and without Down syndrome, 4) examining the systematic and random sources of variance in pedometer measurement under free-walking conditions, and 5) determining the reliability of spring-levered and piezoelectric pedometers for adults with and without Down syndrome.

Research Questions

1. Were there significant differences in absolute pedometer error between the measurements of spring-levered and piezoelectric pedometers?

2. Were there significant differences in absolute pedometer error at faster and slower walking speeds for spring-levered and piezoelectric pedometers?
3. Were there significant difference in absolute pedometer error for spring-levered and piezoelectric pedometers between individuals with Down syndrome and the general population?
4. Was there a significant influence of waist-to-hip ratio on the absolute pedometer error for spring-levered and piezoelectric pedometers?
5. What were the systematic sources of variance in pedometer measurement unique to groups of participants with and without Down syndrome?
6. Did the spring-levered and piezoelectric pedometer demonstrate acceptable levels of reliability for both individuals with and without Down syndrome?

Delimitations

The study was delimited to the following:

1. Participants from small cities in a Pacific Northwest state that were independently ambulatory and did use an assistive walking device including:
    a. Twenty adults with Down syndrome, ages 18 to 65, with as close to an equal proportion of males and females as possible.
    b. Twenty four adults, without a disability of any kind, ages 18 to 65, with as close to an equal proportion of males and females.
2. The use of the Yamax Digiwalker SW-200 and Omron HJ-112 pedometers to represent spring-levered arm and piezoelectric crystal pedometers respectively.

Assumptions

In this study the following assumptions were made:

1. The gait pattern of the accompanying researcher did not affect the gait pattern of the participant during controlled trials or free-walking period.
2. The participant took an even number of steps during each 2 minute trial, as the number of strides observed was multiplied by 2 to determine total steps.
3. Hip-to-waist ratio is a reasonable substitute for pedometer tilt angle and represents the same physical condition.
4. Participant's self-selected speed during controlled conditions trials was representative of normal pace during daily walking.
5. Contact between piezoelectric pedometers within the pocket during the free-walking conditions did not result in miscounting of steps or additional errors.

Limitations

The study was limited to the following:

1. The use of a convenience sample may not necessarily represent adults with Down syndrome or the general population, thus limiting the ability to generalize results.
2. The between-subjects variance was larger than the within-subjects variance, due to a heterogeneous sample of adults with Down syndrome. This may have affected the analyses examining differences between groups.
3. The study failed to adequately create a controlled condition on the fast pace trial as the actual speed walked differed between groups. This was due to the

inability of participants in the DS group to walk at the desired 4.0 mph without transitioning into a running pattern. This may have affected the analyses examining differences between groups and speeds.

4. The study was conducted at three locations, resulting in controlled courses of three different dimensions and total walking lengths. While there were no statistical differences on absolute error rates between locations, the lack of control within the experimental conditions is still a limitation.
5. The 20 minute walking period may not have been sufficient time for participants to distinguish individual walking patterns, thus limiting the total variance.
6. There were multiple facets and interactions with negative estimated variance components. While relatively small in magnitude, this could indicate data issues.

Operational Definitions

1. Pedometer step count. The direct and objective measure of the number of steps accrued over a set period time as recorded by the spring-levered and/or piezoelectric pedometer when displaying the total steps on the output screen.
2. Observed step count. The subjective, yet criterion measure of the number of steps accrued over a set period of time as recorded by a researcher using a hand-tally counter device based on direct observation during the walking trial. An observed step is defined as each time the lead foot touches the ground. The total observed step count is the number of lead foot touches multiplied by two.

3. Absolute error. The degree of measurement error between the pedometer step count and observed step count. The absolute error is calculated as: (|Observed steps – Pedometer| / Observed steps) (Zhu & Lee, 2008).
4. Waist-to-hip ratio (WHR). The difference in proportions of waist and hip circumference is used to determine the pattern of body fat distribution. The WHR is the waist circumference divided by the hip circumference (ACSM, 2000). The WHR was employed to represent the physical condition causing pedometer tilt.

CHAPTER 2

The Accuracy of Pedometers for Adults with Down Syndrome

E. Andrew Pitchford and Joonkoo Yun

ABSTRACT

The purpose of this study was to examine the accuracy of spring-levered and piezoelectric pedometers for adults with and without Down syndrome (DS). Twenty adults with DS and 24 adults without a disability walked for periods of two minutes on a predetermined indoor course for three trials at a self-selected, slower and faster pace. During each trial, the number of steps taken was measured by two types of pedometer, spring-levered and piezoelectric. A criterion step count of observed steps, participant walking speed, and waist-to-hip ratio were also measured. The pedometer recorded and observed steps were compared to determine pedometer error. There was a significant interaction between pedometer model and walking speed. Piezoelectric pedometers demonstrated significantly less measurement error than spring-levered pedometers, particularly at slower walking speeds. These differences were further explained by increasing waist-to-hip ratios of participants. There were also significant differences in pedometer error between adults with and without DS. The study concludes that there are significant differences in pedometer measurement error between adults with and without DS and recommends that piezoelectric pedometers be used in the future to measure walking activity.

(183 words)

Keywords: *disability, pedometer, measurement, validity*

The physical activity habits of adults with intellectual disabilities (ID), particularly those with Down syndrome (DS), are relatively unknown. The current literature does indicate that the vast majority of individuals with ID are not sufficiently active, yet the methodology under which these behaviors have been examined is questionable (Temple, Frey & Stanish, 2006). Individuals with DS typically exhibit characteristics unlike other forms of ID including unique body and facial features, obesity and growth stature differences, muscle hypotonia, joint laxity and a variety of medical conditions including congenital heart disease (Latash, 2000; Roizen, 2002). There is also evidence that individuals with DS walk with a unique gait pattern, marked by additional medio-lateral variability (Agiovlasitis, 2007; Kubo & Ulrich, 2006). Although individuals with DS may not be at risk for cardiovascular disease to the extent that adults with intellectual disabilities without DS (Draheim, McCubbin & Williams, 2002), this population still experiences preventable health disparities that could be remedied by increased access to physical activity (Frey, Stanish & Temple, 2008; Stanish, Temple, & Frey, 2006; U.S. Department of Health and Human Services [USDHHS], 2002). Of what little evidence is available, it appears walking is the most common form of physical activity for individuals with ID, both with and without DS (Draheim, Williams & McCubbin, 2002; Stanish & Draheim, 2005a, 2005b; Temple, Anderson & Walkley, 2000). Thus, efforts should be made to maximize the accuracy, reliability and efficiency of methodology used to measure this behavior.

A few studies have examined the walking activity of individuals with and without DS. These studies utilized a 10,000 step/day criteria consistent with many public health standards (Tudor-Locke & Bassett, 2004). Stanish (2004) measured the

walking activity of individuals with mild ID without DS. Walking activity was measured over 7 days and found that adults without DS walked approximately 11,800 steps/day compared to 5,600 to 8,800 steps/day for adults with DS, indicating a significant difference between subgroups. These findings have not been directly substantiated by subsequent studies (Peterson, Janz & Lowe, 2008; Stanish & Draheim, 2005a, 2005b, 2007). While these studies reported no significant differences in steps counted between groups, the Stanish and Drahiem (2005a, 2005b, 2007) studies showed that a higher percentage of participants with DS recorded fewer than 7,500 steps/day. The inconsistencies in statistical differences between adults with and without DS, or at the very least a trend of adults with DS having lower steps counts regardless of statistical significance, could represent differences between different samples, but could also reflect a source of measurement error.

Many previous studies examining physical activity have utilized pedometers. It is a simple, user-friendly, unobtrusive and relatively inexpensive device (Bassett et al., 1996; Crouter, Schneider, Karabulut, & Bassett, 2003; Le Masurier & Tudor-Locke, 2003; Schneider, Crouter, & Bassett, 2004; Tudor-Locke & Myers, 2001) and may be particularly practical for use among populations with intellectual disabilities. There are three types of pedometers, each utilizing different mechanisms including spring-levered arm, magnetic reed, and piezoelectric crystal. The two most common commercially available pedometer mechanisms are the traditional spring-levered arm and newer piezoelectric pedometers. The spring-suspended level arm mechanism records steps as an internal horizontal lever arm moves up and down from vertical movement at the hip (Crouter et al., 2003; Schneider, Crouter, Lukajic, & Bassett,

2003). The piezoelectric pedometer mechanism is similar to a uni-axial accelerometer that uses a horizontal beam and a piezoelectric crystal to record steps based on the number of zero-crossings of the instantaneous acceleration versus time curve (Crouter et al., 2003; Schneider et al., 2003).

All pedometer types have been extensively validated in the general population, and numerous studies have demonstrated that certain brands and models are more accurate than others (Bassett et al., 1996; Crouter et al., 2003; Schneider et al., 2003, 2004). In general, most pedometers have been found to have acceptable accuracy for measuring steps, and to a lesser degree distance, time and calories expended. However, the pedometer has undergone less validation within disability populations, particularly individuals with DS. Of the limited evidence available, Stanish (2004) found intraclass correlations for spring-levered arm pedometers for adults with intellectual disabilities to be very high across speed, location and surface (all above ICC = 0.95). Additional studies have identified unique sources of measurement error, yet provided moderate support for youth with developmental disabilities, youth with visual impairments, and adults with neurological disabilities (Beets, Combs, Pitetti, Morgan, Bryan & Foley, 2007; Beets, Foley, Tindall, & Lieberman, 2007; Manns, Orchard & Warren, 2007). However, to the best of current knowledge, no known studies have examined accuracy of piezoelectric pedometers for individuals with DS.

All of these studies however, are limited by the lack of stratified analyses that may show important differences between subgroups as opposed to using broadly collected and innately heterogenous groups. Stratified designs have been recommended to better account for sub-group differences, particularly with samples

including individuals with DS (Temple, Frey & Stanish, 2006). The previously discussed studies all employed "whole group" analyses, which despite moderate to high levels of accuracy could still miss significant group differences. Furthermore, none of these validation studies employed any control group, so while these studies have provided initial evidence of pedometer accuracy in certain populations, there is no indication on how this level of accuracy compares between adults with and without disabilities. This is important to determine if the body of literature on pedometers in the general population can be generalized to specific disability groups.

In the Stanish (2004) study, the author suggests that while the "validity data on pedometers is likely applicable to adults with MR, it is still important to gather evidence of accuracy when using instruments for research purposes" (p.168). The same sentiment applies to individuals with Down syndrome, although there are additional reasons why pedometers may not be as accurate in this population.

First, individuals with DS typically walk with unique gait pattern described as a shuffling pattern that includes a wider base and increased medio-lateral variability to overcome joint laxity and muscle hypotonia through muscle co-contractions and stiffness (Agiovlasitis, 2007; Kubo & Ulrich, 2006; Smith, Kubo, Black, Holt, & Ulrich, 2007; Smith & Ulrich, 2008; Ulrich, Haehl, Buzzi, Kubo, & Holt, 2004). Gait variability has been identified as a significant source of pedometer error (Manns, Orchard & Warren, 2007), so this gait pattern could present problems for pedometer measurement. Second, individuals with DS typically walk at slower self-selected speeds, notably under 70 m/min^{-1} (Smith & Ulrich, 2008; Smith et al., 2007; Ulrich et al., 2004). Pedometers have been shown to become less accurate at speeds less than 80

m/min^{-1}, particularly at 54 m/min^{-1} (Bassett et al., 1996; Crouter, et al., 2003; Le Masurier & Tudor-Locke, 2003; Le Masurier, Lee & Tudor-Locke, 2004; Melanson et al., 2004; Schneider et al., 2003, 2004). It is certainly possible that walking activity for individuals with DS is being underestimated due to this slower walking speed. Third, individuals with DS have been shown to have a body composition characterized by additional adipose tissue in the torso area, in a pattern that is unique to this population (Roizen, 2002). Studies in the general population have found that pedometer tilt, caused by adipose abdominal tissue in overweight and obese individuals can cause pedometer error by negatively influencing the devices ability to utilize its mechanism for measurement (Crouter, Schneider, & Bassett, 2005). The body composition of individuals with DS is conducive with creating pedometer tilt, thus creating a source of pedometer error. Due to these potential sources of error, certain pedometer mechanisms may be more appropriate for adults with DS.

It is necessary to gather empirical evidence on the accuracy and validity of both spring-levered and piezoelectric pedometers for adults with Down syndrome so that the most appropriate instrument can be used for the measurement of walking activity. Given the common sources of error that have been shown in the general population, the DS population may present more potential sources of error. Thus, the purpose of this study was to examine the magnitude of errors and differences in accuracy for spring-levered and piezoelectric pedometers with adults with Down syndrome and adults without a disability. Additionally, this study examined the effect of the walking speed traveled and the anthropometric characteristics of the individual on absolute measurement error for the pedometer models and DS groups.

Method

Participants

Convenience samples of twenty adults with Down syndrome (12 female, 8 male) aged 18-61 years (mean age 29.25 years) and 24 adults without a disability (14 female, 10 male) aged 22-60 years (mean age 32.08 years) participated in the study. All participants were recruited from small cities in the northwest region of the United States and were independently ambulatory without assistive devices. Age was self-reported by participants and height, weight, waist and circumferences were measured without shoes in light clothing to the nearest 0.1 cm and 0.1kg respectively. The demographic and anthropometric characteristics of participants are included in Table 2.1. Independent sample t-tests showed participants without DS had significantly greater body mass (weight; $p < 0.05$), and height ($p < 0.05$). There were no significant differences between participants with and without DS for age, BMI, waist circumference, hip circumference and waist-to-hip ratio (all $p > 0.05$).

Table 2.1. Physical Characteristics of Participants: Descriptive Statistics by Group

Variable	Down Syndrome (N=20)	Control Group (N=24)	All (N=44)
Gender (female/male)	12 / 8	14 / 10	26 / 18
Age (yr)	29.25 ± 12.45	32.08 ± 13.10	30.79 ± 12.74
Height (cm)*	148.96 ± 8.13	170.33 ± 8.22	160.62 ± 13.47
Weight (kg)*	64.16 ± 11.59	76.15 ±17.12	70.70 ± 15.89
BMI ($kg \cdot m^{-2}$)	29.09 ± 5.76	26.14 ± 5.23	27.48 ± 5.61
Waist circumf. (cm)	87.97 ± 12.86	84.93 ± 14.51	86.31 ± 13.71
Hip circumf. (cm)	103.74 ± 11.39	103.60 ± 10.69	103.67 ± 10.88
Waist-to-hip ratio	0.85 ± 0.07	0.82 ± 0.09	0.83 ± 0.08

Note: *significant differences between groups, $p < 0.05$

Written consent was obtained from all participants prior to participation in the study in accordance with Institutional Review Board approval. Informed consent documents for participants that required assistance in completing consent and demographic questionnaire documents were also signed by the assisting parent or legal caregiver. Medical diagnosis of Down syndrome was self reported by participants or reported with assistance of a parent or legal caregiver.

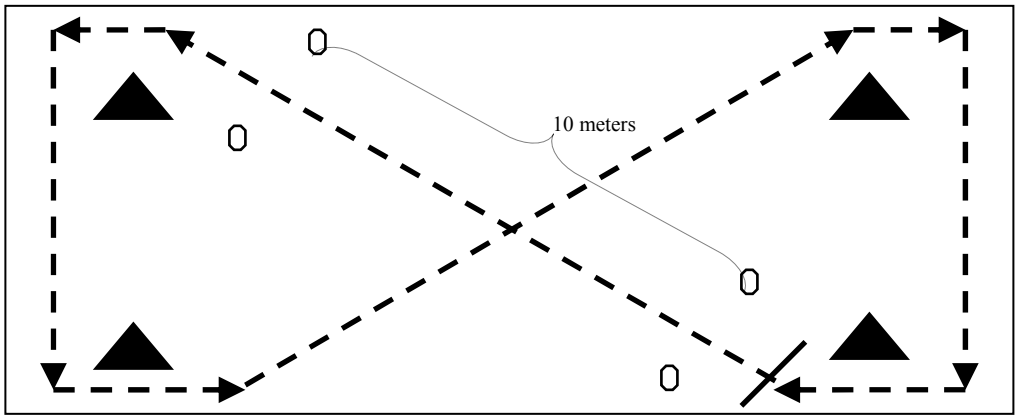
Instruments

Omron HJ-112 (Omron Healthcare, Vernon Hills, IL) and Yamax Digiwalker SW-200 (Yamax Inc., Tokyo, Japan) pedometers were used to represent piezoelectric and spring-levered arm pedometer mechanisms respectively and measure walking steps taken. Both models were selected due to established levels of accuracy and validity (Bassett et al., 1996; Crouter et al., 2003; Doyle, Green, Corona, Simone & Dennison, 2007; Hasson, Haller, Pober, Staudenmayer & Freedson, 2009; Hasson, Prober & Freedson, 2004; Schneider et al., 2003, 2004). All instruments were checked for calibration before the start of the study using a 100 count modified version of a "shake test" (Vincent & Sidman, 2003). Pedometers that demonstrated errors of 1% or less were used in the study. During testing, each participant wore four pedometers, two of each model, on the right waistband at the midline of thigh using an elastic belt. All pedometers were positioned as close to manufacturer's recommendation as was physically possible for the participant. At the start of each trial, all pedometers were reset to zero. At the end of each trial, the number of steps measured was recorded for each instrument.

Testing procedures

The accuracy of pedometers using piezoelectric (Omron HJ-112) and spring-levered arm mechanisms (Yamax Digiwalker SW-200) were tested at three different walking speeds: self-selected, slow, and fast. Each walking trial occurred on a "figure 8" (see Figure 2.1) walking course and lasted 2 minutes each. This walking bout is similar to protocols used in previous treadmill based studies (Crouter et al., 2005; Swartz, Bassett, Moore, Thompson & Strath, 2003). Data were collected in three separate sites. Due to physical constraints of the space at each site, the total course walking distance ranged from 38 meters to 68 meters. All versions of the course had cross tangents of at least 10 meters and were on hard indoor gymnasium surfaces. The 10 meter cross tangent was marked with two cones on each side.

Figure 2.1. Diagram of controlled course layout



Throughout the testing procedure, the participant walked with a researcher. During the self-selected pace trial, the researcher walked behind the participants as to not affect the walking pace. The researcher encouraged the participant and gave simple verbal instructions to ensure that the walking course was followed properly. During the slow and fast paced trials, the researcher walked side by side with the

participants using a calibrated measuring wheel with an attached CatEye (Kuwazu, Japan) speedometer and encouraged the participant to walk at the paced speed throughout each trial. Additionally, a researcher measured the time it took the participant to walk across the marked 10 meter cross tangent. The researcher began timing when the first foot of the participant crossed the start line and stopped timing when the first foot crossed the end line. The time required for the participants to walk the 10 meter distance was averaged over all measurements within a trial and divided by 10 to determine the walking speed in meters per second. This speed was then converted to miles per hour.

During the slow paced trial, the researcher set a pace of 2mph. This speed was selected as it was the slowest speed that could be consistently set using the available CatEye technology. At this pace, the average speed walked was 1.97 mph (SD = 0.45) for participants with DS and 2.17 mph (SD = 0.15) for participants without DS. These slower speeds correspond with the slowest speeds (54 m/min^{-1} and 67 m/min^{-1}) used in pedometer studies on treadmills that have been shown to decrease accuracy (Basset et al., 1996; Crouter et al., 2003, 2005; Le Masurier & Tudor-Locke, 2003; Le Masurier et al., 2004; Swartz, Bassett, Moore, Thompson & Strath, 2003).

During the fast paced trial, the researcher set a pace for the participant at the fastest walking speed possible and/or a maximum of 4 mph. This speed differed between individuals to account for different transition speeds from walking to running and ensured that all participants walked the course. On the fast trials, the average walking speeds were 3.49 mph (SD = 0.67) and 3.93 mph (SD = 0.31) for participants with and without DS respectively. These faster speeds correspond with the fastest

speeds (94 m/min^{-1} and 107 m/min^{-1}) used in multiple treadmill based studies (Bassett et al., 1996; Crouter et al., 2003, 2005; Le Masurier & Tudor-Locke, 2003; Le Masurier et al., 2004; Swartz et al., 2003). The average self-paced walking speed for participants with and without Down syndrome were 2.62 mph (SD = 0.70) and 3.19 mph (0.41) respectively.

During each walking trial, another researcher observed the participant and recorded the number of steps taken using a hand-held tally counter. The researcher counted the number of foot-strikes by the lead foot. This number was doubled to represent the actual number of steps observed and is used as the criterion step count in all analyses. Each participant repeated one trial for test-retest reliability. Intraclass correlations of pedometer error between trials were high across all speed conditions and models (ICC (2,2) >.89).

Statistical Analysis

To examine the accuracy of pedometers, absolute error scores were calculated. Each absolute error score was determined for each pedometer model at three speeds of walking (self-paced, slow and fast paced) by the equation: (|Observed steps – Pedometer| / Observed steps). This error score is used as the dependent variable in subsequent analyses and represents the absolute degree of error between the pedometer recorded steps and the actual steps taken (Lee, Zhu, Yang, Bendis, & Hernandez, 2007; Zhu & Lee, 2008). Additionally, intraclass correlation coefficients (ICC) were calculated to examine the level of conformity between observed and pedometer recorded steps.

A 2 x 2 x 3 (group x model x speed) repeated measures ANOVAs was used to examine differences in absolute error scores between groups with and without DS, piezoelectric and spring-levered pedometers, and self-selected, slow, and fast paced speeds. Post-hoc comparisons to determine significant differences of speed when an interaction was present were examined through one-way (speed) repeated measures ANOVA for each pedometer model. Additionally, a follow-up analysis to determine the influence of waist-to-hip ratio on absolute error was performed using a 2 x 2 x 3 (group x model x speed) repeated measures ANCOVA with the covariate of waist-to-hip ratio. When assumption of sphercity was violated, Huynh-Feldt corrections were employed. Alpha of 0.05 was used to indicate statistical significance.

Bland-Altman (Bland & Altman, 1986) plots were created to show the distribution of pedometer error scores around 0. Each plot represents the individual pedometer error scores for the particular speed and pedometer model for all participants, visually distinguished by DS group affiliation. Solid lines represent the 95% confidence interval (1.96 x SD) for the whole sample at the particular speed and model. The scales of each axis have been standardized for easy comparison.

Results

Absolute percent error across speeds for adults with DS ranged from 11.40% to 22.39% for the spring-levered pedometer and from 7.57% to 8.02% for the piezoelectric. Similarly, absolute percent error for participants without DS ranged from 2.87% to 16.44% for the spring-levered and from 1.06% to 2.96% for the

piezoelectric pedometer. The absolute error scores are presented as percentages in Table 2.2.

Table 2.2. Absolute percent error in number of steps during trials at three speeds

	Self Paced Trial	Slow Paced Trial	Fast Paced Trial
Down Syndrome			
Digiwalker	11.40 ± 19.21	22.39 ± 17.71	11.89 ± 21.13
Omron	7.90 ± 13.08	7.57 ± 13.76	8.02 ± 9.03
Control Group			
Digiwalker	6.49 ± 12.16	16.44 ± 11.66	2.87 ± 6.51
Omron	1.06 ± 1.47	2.96 ± 2.94	1.05 ±2.17

Note: Values are mean absolute percent errors
(|Observed Steps – Pedometer| / Observed Steps x 100).

The 2 x 2 x 3 (group by model by speed) repeated measures ANOVA on absolute percent error scores revealed the following results. There was a significant model by speed interaction, $F(2,84) = 13.14$, $p<0.001$, $\eta^2 = .24$. For the spring-levered model (Digiwalker SW-200) there was a simple main effect for speed, $F(2,86) = 14.01$, $p<0.001$, $\eta^2 = .25$. Simple contrasts revealed the absolute error of the spring-levered model at the self-paced and fast speed were significantly different than the slow speed ($p<0.001$), but not significantly different from each other ($p>.04$). For the piezoelectric model (Omron HJ-112) there was no simple main effect for speed, $F(2,86) = 0.17$, $p>0.8$, indicating absolute error was consistent across walking speeds.

There were also significant group differences on absolute error between participants with and without DS, $F(1,42) = 9.06$, $p<0.01$, $\eta^2 = .18$. Interactions for model by group, $F(1,12) = 0.19$, $p>.8$, speed by group, $F(2,84) = 0.32$, $p>.7$, and model by speed by group, $F(2,84) = 0.40$, $p>.6$, were all not statistically significant.

When the covariate of waist-to-hip ratio was added to the 2 x 2 x 3 (group x model x speed) repeated measures ANCOVA the results change. The group differences between adults with and without DS remained significant, $F(1,41) = 7.35$, $p<0.05$, $\eta^2 = .15$. However, both the main effects for speed, $F(1.82,74.75) = 2.93$, $p>0.1$, $\eta^2 = .04$, and model, $F(1,41) = 1.98$, $p>0.1$, $\eta^2 = .05$, were not significant. All possible interactions were also not statistically significant ($p>0.1$).

Intraclass correlation coefficients are presented in Table 2.3. ICC (2,2) results were variable across model, group and speed. High ICC levels were observed for the Omron pedometers ranging from 0.89 to 0.97 across all speeds for the control group and from 0.87 to 0.90 for participants with DS at self-paced and slow speeds. However, the Omron pedometer demonstrated moderate ICC of 0.66 for participants with DS at the fast speed. Moderate levels of ICC were observed for the Digiwalker pedometers at self-paced and fast speeds for participants in control group (0.67 to 0.70) and with DS (0.76 to 0.79). Very low levels were observed for both groups at slow speeds (ICC>0.52).

Table 2.3. Intraclass Correlation Coefficients between Pedometer and Observed Steps

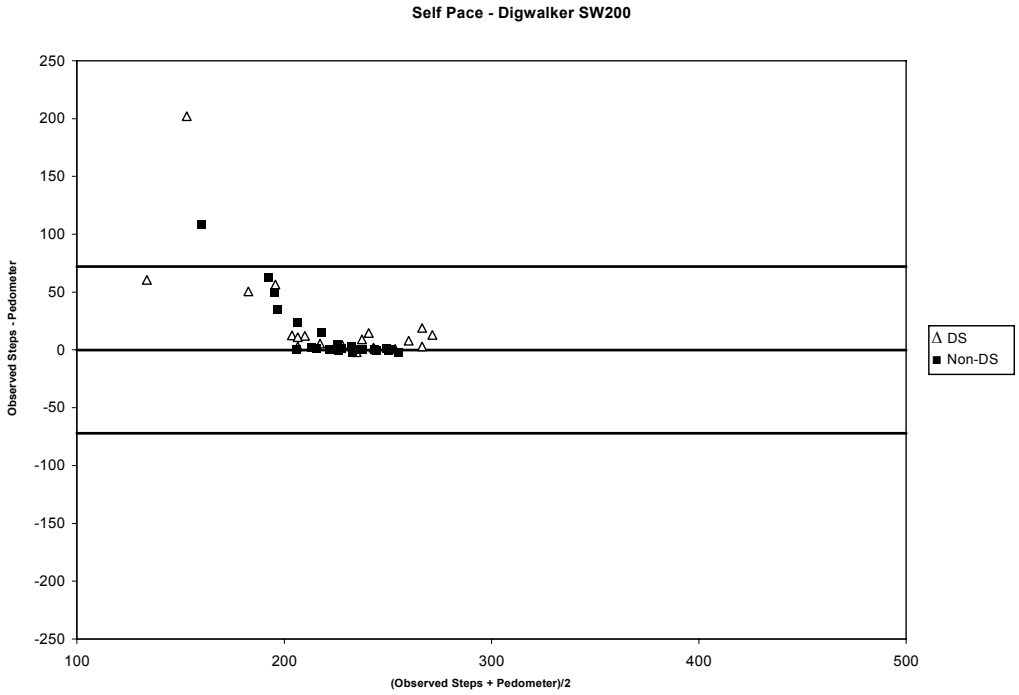
	Control Group		Down Syndrome	
	Digiwalker	Omron	Digiwalker	Omron
Normal	0.67	0.97	0.76	0.87
Slow	0.52	0.89	0.41	0.90
Fast	0.70	0.92	0.79	0.66

Note: ICC(2,2)

Finally, graphical Bland-Altman plots of the distribution of errors by model and speed are presented in Figures 2.2 – 2.7. To assist in comparison, both the x and y-axes are standardized. The 95% confidence interval displayed on each graph is from

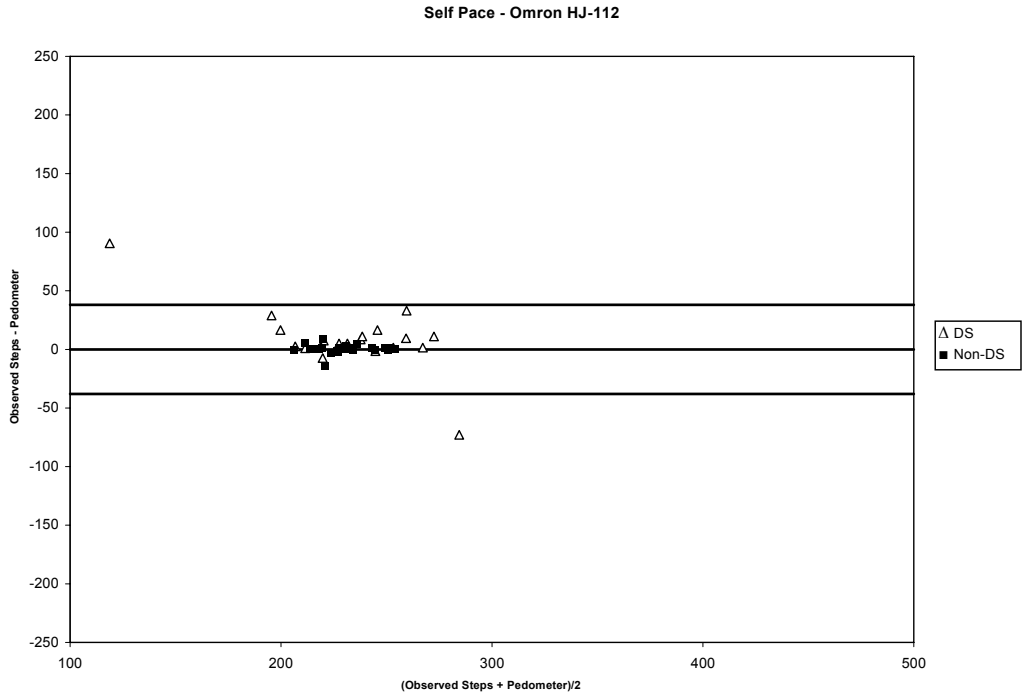
the pooled standard deviation of both groups. For adults with DS, the Omron showed good accuracy at all three speeds with 95% confidence intervals within ± 60 steps. The Digiwalker demonstrated moderate accuracy at self-paced and slow speeds with 95% confidence intervals within ± 100 steps, but poor accuracy at the fast speed (± 149 steps). For adults without a disability, the Omron demonstrated exceptional accuracy at all three speeds with 95% confidence intervals within ± 20 steps while the Digiwalker showed good accuracy (± 60 steps).

Figure 2.2. Bland-Altman plot of Digiwalker SW-200 at self-paced speed



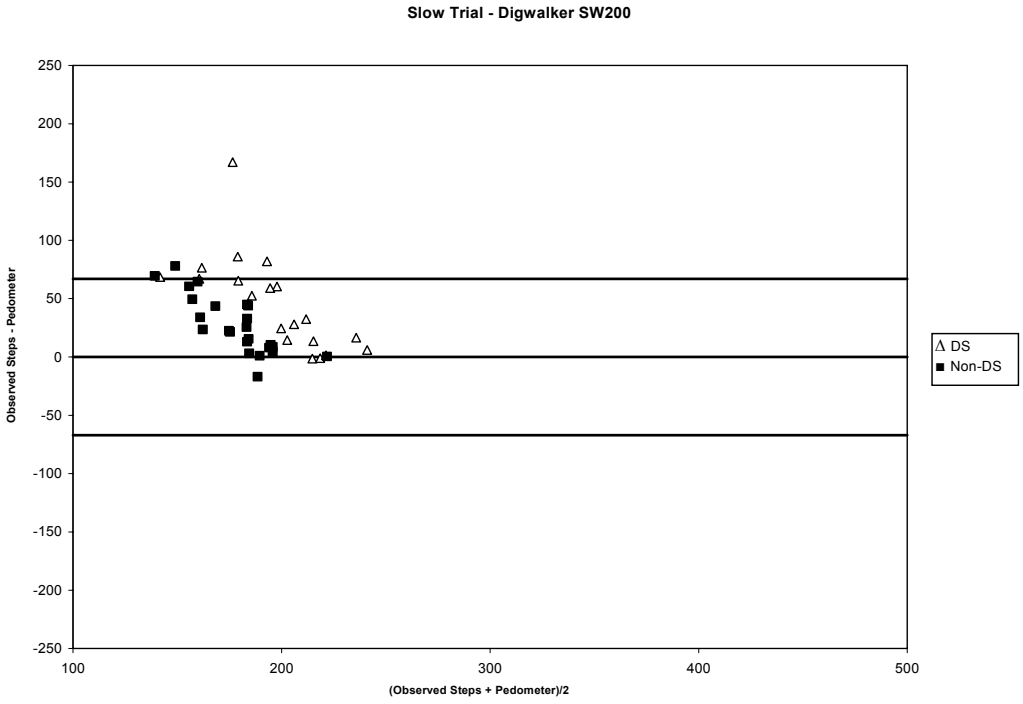
Note: Pooled 95% confidence interval: ±72; DS: ±89; Control: ±52.

Figure 2.3. Bland-Altman plot of Omron HJ-112 at self-paced speed



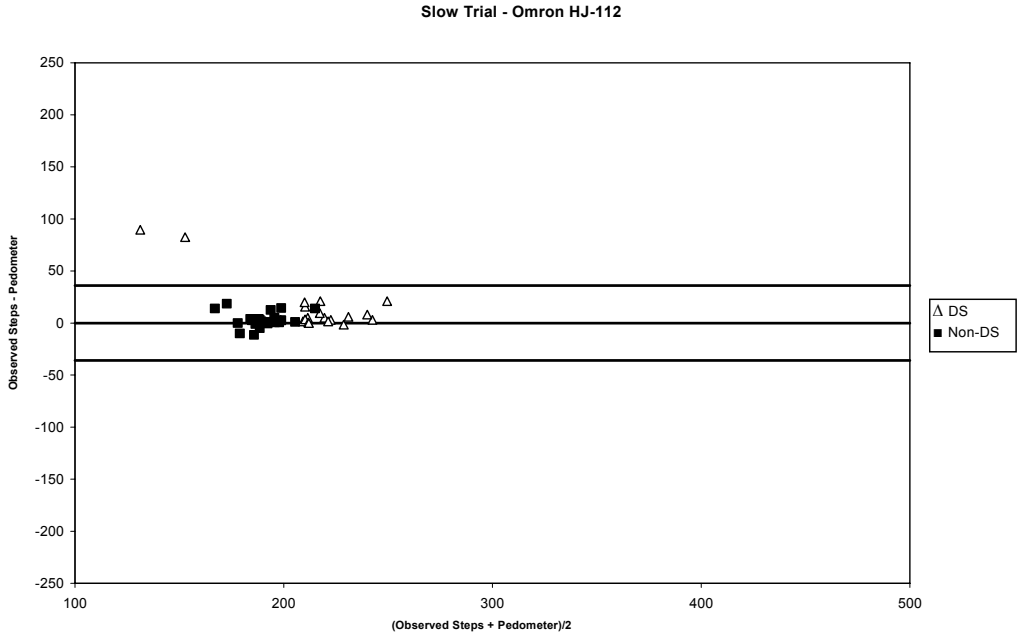
Note: Pooled 95% confidence interval: ±38; DS: ±55; Control: ±7.

Figure 2.4. Bland-Altman plot of Digiwalker SW-200 at slow speed



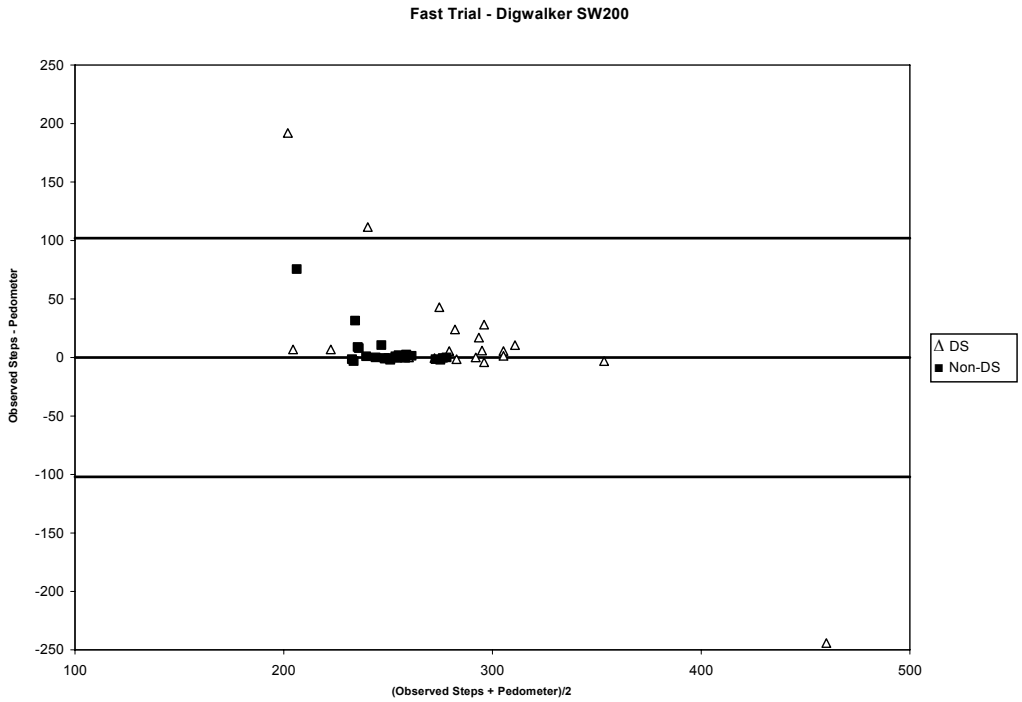
Note: Pooled 95% confidence interval: ±67; DS: ±80; Control: ±48.

Figure 2.5. Bland-Altman plot of Omron HJ-112 at slow speed



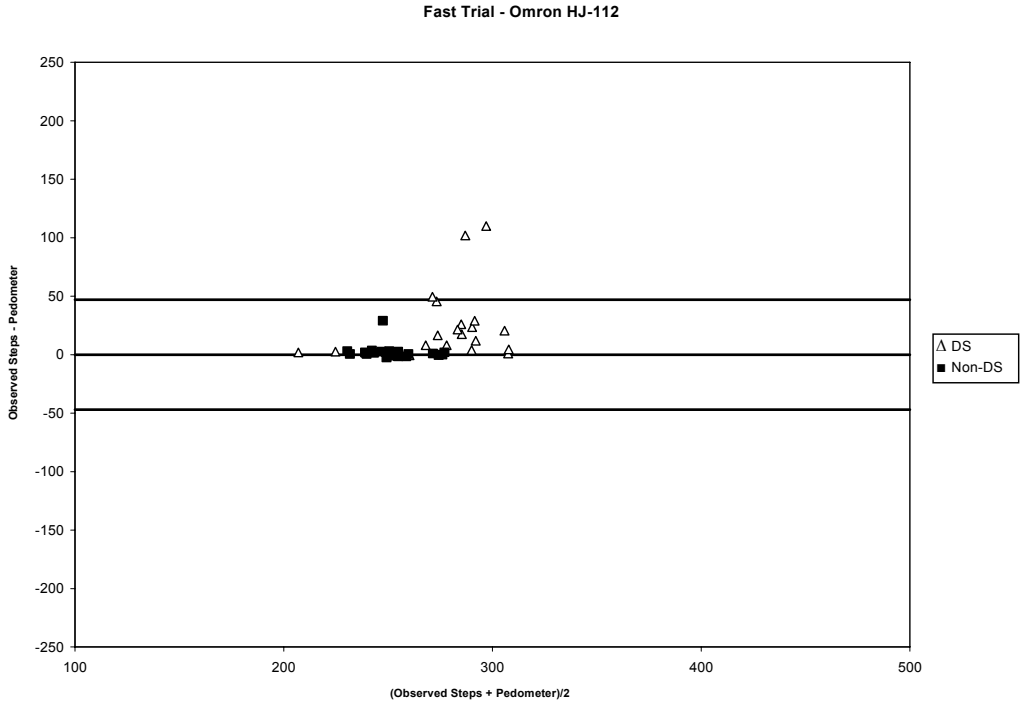
Note: Pooled 95% confidence interval: ±36; DS: ±49; Control: ±14.

Figure 2.6. Bland-Altman plot of Digiwalker SW-200 at fast speed



Note: Pooled 95% confidence interval: ±102: DS: ±149; Control: ±32.

Figure 2.7. Bland-Altman plot of Omron HJ-112 at fast speed



Note: Pooled 95% confidence interval: ±47; DS: ±60; Control: ±11.

Discussion

The pedometer is a widely used measurement tool with many models capable of monitoring the steps an individual takes during daily ambulation in addition to other indices of walking activity including distance walked, time in aerobic walking activity and calories expended. Pedometers have been used in research to measure walking and physical activity behaviors in the general (Tudor-Locke & Myers, 2001) and disability (Peterson et al., 2008; Stanish, 2004; Stanish & Draheim, 2005a, 2005b, 2007) populations. There are also a growing number of individuals that use pedometers recreationally as part of their own lifestyles.

The present study sought to examine if the accuracy of two different types of pedometers differed between adults with Down syndrome and with no disability.

These differences were addressed between self-paced, slower, and faster speeds as well as controlling for waist-to-hip ratio.

The main finding is that pedometers utilizing different mechanisms, particularly spring-levered and piezoelectric pedometers, can have different levels of measurement error. The results indicate that there is a significant main effect related to model. The absolute errors for the spring-levered pedometer ranged from approximately 11% to 22% for adults with DS and 3% to 16% for adults in the control group. Conversely, absolute error for the piezoelectric model ranged from approximately 7% to 8% for adults with DS and 1% to 3% for adults in the control group. The support for wide variability between mechanisms and consumer models has been well documented in the literature. Multiple studies (Bassett et al., 1996; Crouter et al., 2003; Schneider et al., 2003, 2004) have examined numerous models of pedometers under a range of conditions (treadmill, walking track, free-living). The general consensus is that due to the variability amongst models, not all pedometers are accurate. In studies examining numerous models, pedometers using a piezoelectric mechanism were consistently among the most accurate and most reliable (Crouter et al., 2003; Schneider et al., 2003, 2004). Furthermore, when piezoelectric and spring-levered pedometers have been directly compared, the piezoelectric has been found superior, particularly during adverse measurement conditions (Crouter et al., 2005; Melanson et al., 2004). Our findings show that piezoelectric pedometers are dependably more accurate for both adults with and without DS.

A second finding is the significant differences between adults with and without DS on absolute error rates. When used by adults with DS, the pedometer error rate

was consistently higher for both models and at all speeds than individuals in the control group. For example, during the self-paced walking trial the average absolute error rate among adults with DS was approximately 11% for the spring-levered pedometer and 8% for the piezoelectric as opposed to approximately 6% and 1% for the control group respectively. The lack of an interaction between either model or speed and the syndrome groups also indicates that inaccuracy demonstrated for adults with DS is not isolated to one mechanism or one speed.

In addition to these significant differences between pedometer models and participant groups, the results also confirm two sources of measurement error. The first is walking speed. The current study provides additional evidence that the accuracy of pedometers for each model is moderated by the speed at which the mover travels. The spring-levered model was more affected by speed, as demonstrated by additional error reported at the slower walking pace. The piezoelectric pedometer had decreases in accuracy with slower speeds, but this difference was not significantly or substantively different. These results are consistent with much of the recent literature in pedometer accuracy. Melanson et al. (2004) showed that regardless of speed, piezoelectric pedometers were more accurate than spring-levered pedometers, including speeds of 1.0 mph. While the current study used approximately 2.0 mph as the slowest speed, the differences in accuracy for the spring-levered pedometer as opposed to the lack of differences observed in the piezoelectric models represent a comparable and similar trend. This shows that the piezoelectric pedometer is more resistant to speed related errors for both groups.

The remainder of the literature has consistently shown this trend that pedometers become less accurate at speeds under 80 $m \cdot min^{-1}$ (2.98 mph; Bassett et al., 1996; Crouter et al., 2003; Le Masurier et al., 2004; Le Masurier & Tudor-Locke, 2003; Schneider et al., 2003, 2004). In the current sample, adults without a disability demonstrated the usual pattern of accuracy decreasing with speed. Percent error were under 3% at the fastest speed, under 6.5% at self-selected speed, and under 16% at the slowest speed for both pedometer models. However, adults with DS showed a slightly different pattern. With the spring-levered pedometer percent error was under 22% at the slowest speed and under 11.5% at the self-selected speed, but under 12% at the fastest speed. The theory behind walking speed being a cause of error for pedometers is that as speed decreases there is less vertical movement at the hip for the device to detect (Crouter et al., 2003). However, it appears that this phenomenon is not a strictly linear relationship between speed and accuracy. The slightly higher error rate at the fastest speed could represent a more variable gait pattern for adults with DS as they near the threshold for transition for running (Agiovlasitis, Yun, Pavol, McCubbin & Kim, 2008). Despite no significant differences between self-paced and fast speeds, as compared to the significant differences between the slow pace and the two faster speeds, the results indicate that not only dampened, but also excessive hip displacement may results in error. Despite these slight fluctuations from the usual trend, the results clearly indicate that at slower speeds, accuracy is compromised.

The second source of measurement error was pedometer tilt caused by increasing waist-to-hip ratio. There have been contradicting results related to body composition and pedometers. While some studies have found that BMI has no

statistically significant effect on pedometer accuracy (Melanson et al, 2004; Swartz et al., 2003), Crouter et al. (2005) found that higher BMI, waist circumference and pedometer tilt resulted in lower levels of accuracy. Specifically, pedometer tilt was found to be strongest factor affecting steps counted. The study also found that these factors significantly contributed to the significant underestimation of steps by the Digiwalker SW-200, while having a non-significant effect on the New Lifestyles NL-2000, a piezoelectric pedometer. This study found similar results. Waist-to-hip ratio was measured to represent the physical conditions that cause pedometer tilt. An actual measure of pedometer tilt was not used because it is possible that the angle of tilt could change throughout the gait cycle as well as between individual strides. The initial results found an interaction between the two models and the three speed conditions. However, once the covariate of waist to hip ratio was added, all of these effects became non-significant. This indicates that the differences observed on error rate between the spring-levered and piezoelectric pedometer can be explained by the influence of waist-to-hip ratio. This is a particularly important factor for addressing the accuracy of pedometers for individuals with DS, as the population is prone to abdominal obesity (Roizen, 2002). It should be noted however, that in the current sample, there was no statistical difference between groups related to any obesity related measure.

While significant differences between participant groups, pedometer models, and walking speeds are important, the magnitude of error or precision of accuracy must also be taken into account. For adults with DS, the magnitude of pedometer accuracy poses a problem. The spring-levered arm pedometer consistently

demonstrated accuracy levels under 90%. Furthermore, intraclass correlation coefficients were less than .80 making it difficulty to associate moderate levels of agreement with observed steps (Baumgartner, Jackson, Mahar, & Rowe, 2007). The piezoelectric pedometer, despite demonstrating more error than in previous studies (Schneider et al., 2004), may still be acceptable for use among adults with DS. Absolute error rates were consistently around 8% and ICC coefficients were greater than .80 except for at the fast paced speed demonstrating moderate agreement. Beets et al. (2007) concluded that a moderate level of agreement was acceptable for pedometers. Furthermore, other studies have deemed "fair" accuracy to be within 10% of actual steps taken (Schneider et al., 2004; Crouter et al., 2005).

An underlying problem still remains, in that the source of error resulting in significant differences among adults with and without Down syndrome has not been precisely identified by this study. The original hypothesis was that pedometers would have more error for individuals with DS due to unique gait patterns, slow walking speed and stereotypical abdominal obesity. The presence of additional errors has been supported by the results; however the cause of this error is still unclear. When controlling for the pedometer model, walking speed, and waist-to-hip ratio, there remained a significant group difference. There was also no interaction between syndrome group and any other factor, indicating that the influences of model, speed, or body composition are not unique to individuals with or without DS. This signifies that an additional factor(s) specific to individuals with DS, possibly gait pattern, is causing additional pedometer error. There is evidence that gait variability can significantly impact the accuracy of pedometers (Manns, Orchard & Warren, 2007).

Individuals with DS have been shown to walk with a gait pattern with increased variability in the medio-lateral direction (Agiovlasitis, 2007; Kubo & Ulrich, 2006) which could result in the underestimation of steps. This study did not measure gait characteristics, so this factor can not be directly addressed.

An additional issue is the results of this study do not support the previous literature in pedometer accuracy for adults with intellectual disabilities. In the Stanish (2004) study, the Digiwalker SW-200 was used and demonstrated very high accuracy with intraclass correlations greater than 0.95 across different walking surfaces, speeds and sides of the body. The present study found wide ranging intraclass correlations for adults with DS, ranging from 0.41 to 0.79 for the Digiwalker and 0.66 to 0.90 for the Omron. The difference in accuracy between studies using the same pedometer model can be explained in two ways. First, the Stanish study included adults with intellectual disabilities with and without DS, while the current study focused on adults with DS. Given the results of a significant difference between groups, the pedometer error for adults with DS may be systematically different than groups without DS, regardless of ID classification. Second, the experimental conditions were very different between the studies. In the current study, the controlled conditions included a "figure-8" walking course with four discrete turns per lap, whereas Stanish used a 400 meter walk with presumably wider turns. The use of turns and changes in direction are more realistic to daily walking patterns, but could also add to pedometer error. Regardless, the established differences between groups of adults with and without DS support the use of stratified sampling designs, unlike the "whole group" analyses used by Stanish (2004) and Beets et al. (2007). Disability groups that are

inherently heterogeneous should be analyzed carefully as within-subjects variability may be very high.

Based on the findings of this study, the authors would recommend that piezoelectric pedometers be used in future research when pedometers are selected as a measurement tool. Researchers intending to measure the walking activity of individuals with DS should strongly consider other modes of physical activity measurement, such as accelerometers, but could use pedometers if the research questions and study conditions (time, budget, sample size) warrant their use. The use of spring-levered pedometers is not as advised due to the high levels error associated with walking speed and pedometer tilt. Despite the recommendation that piezoelectric pedometers may be used for measuring walking activity in adults with DS, future research should note that there is still a fair amount (<10%) of error associated with pedometers and find ways to address and limit these errors in study methodology. It may also be useful to address pedometer data in terms of 95% confidence intervals rather than traditional group averages to better account for variability.

A limitation of the current study is that during certain analyses, the between-subjects variance was larger than the within-subjects variance. This is due to the variable nature of a heterogeneous group such as adults with DS and is common among disability research, but limits the validity of those analyses. A second limitation was the controlled nature of the testing conditions. When participants were paced on the fast trial, each participant was limited by their maximum walking speed before transitioning to running. Thus, unlike the slow speed, there were differences between groups on the speed walked during that condition. These differences may

have been problematic to the results. Additional limitations to the study include the use of a convenience sample of volunteers and the use of waist-to-hip ratio instead of pedometer tilt angle. The use of volunteers in a population of individuals with intellectual disability may result in a higher functioning group of participants that may not fully represent the true population. Finally, while the waist-to-hip ratio is representative of pedometer tilt angle, the actual angle of tilt was not measured. This could limit the efficacy of the effects due to waist-to-hip ratio.

Pedometers are a widely used tool for measuring walking activity. The present study examined the accuracy of spring-levered and piezoelectric pedometers for adults with Down syndrome as well as determined the effects of walking speed and waist-to-hip ratio on absolute pedometer error. Results indicate that errors in steps recorded by a pedometer are significantly higher for adults with DS, for models with spring-levered mechanisms, at slower speeds, and with greater abdominal obesity. Given these results it is recommended that research utilizing pedometers use piezoelectric pedometers to minimize the effects of walking speed and body composition on measurement error, particularly for samples of individuals with DS. Future research should further examine the sources of pedometer error for adults with DS to determine if gait characteristics are responsible for unexplained errors.

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CHAPTER 3

Sources of Variation in Pedometer Measurement for Adults with and without Down Syndrome: A Generalizability Study

E. Andrew Pitchford and Joonkoo Yun

ABSTRACT

The purpose of this study was to examine the sources of variance and reliability of spring-levered and piezoelectric pedometers for adults with and without Down syndrome (DS) during a free-walking bout. Eighteen adults with DS and twenty three adults without a disability walked continuously for a period of twenty minutes wearing two types of pedometers, spring-levered and piezoelectric. The step counts were analyzed using Generalizability theory to partition and quantify variance components. Adults with DS demonstrated greater intra-individual variability during the walking trial. Individuals without DS conversely had greater residuals, or unexplained errors. The spring-levered pedometer showed problems with inter-unit variability through substantial variance components from the unit facet and subject by unit interaction. The piezoelectric pedometer demonstrated little systematic error. Additionally, reliability coefficients for relative and absolute decisions were calculated for each group and model combination. The piezoelectric pedometer demonstrated higher reliability than the spring-levered. This study provides evidence for moderate to high reliability of the piezoelectric pedometer and suggests that this type of pedometer be used in future research for both adults with and without DS.
(178 words)

Keywords: *disability, pedometer, reliability, Generalizability theory*

Pedometers are a common, widely-used and effective tool for measuring physical activity for practical and research purposes (Tudor-Locke & Myers, 2001). For individuals with Down syndrome and other intellectual disabilities, walking is the most common form of physical activity (Draheim, Williams & McCubbin, 2002; Stanish & Draheim, 2005a, 2005b; Temple, Anderson & Walkley, 2000). Given the ongoing public health trend emphasizing the importance of lifelong physical activity (USDHHS, 1996, 2000, 2002), the ability to measure walking activity is important. The pedometer is a particularly useful tool due to this propensity for walking and is also simple, inexpensive and relatively easy to use and understand.

The accuracy of pedometers has been widely established in the literature. The general consensus is that most pedometers are accurate in measuring steps, and to a lesser extent distance walked, energy expended and physical activity time (Bassett et al., 1996; Crouter, Schneider, & Bassett, 2005; Crouter, Schneider, Karabulut, & Bassett, 2003; Hasson, Haller, Pober, Staudenmayer, & Freedson, 2009; Le Masurier & Tudor-Locke, 2003; Melanson et al., 2004; Schneider, Crouter, Lukajic, & Bassett, 2003; Swartz, Bassett, Moore, Thompson, & Strath, 2003). These studies have provided strong validity evidence for the accuracy of pedometers, but have also demonstrated that all brands and models are not created equal and that significant error can occur. It is important to note that many of these studies were conducted within highly control settings, including treadmills and 400 meter tracks. A limited number of studies have examined accuracy in free-living conditions (Schneider, Crouter & Bassett, 2004; Le Masurier, Lee & Tudor-Locke, 2004; Welk et al., 2000) but more evidence is needed for reliability under natural walking conditions.

Accuracy of pedometers has also extended into disability populations, providing initial evidence of pedometer accuracy for adults with intellectual disabilities (Pitchford, 2009; Stanish, 2004), neurological disabilities (Manns, Orchard & Warren, 2007), and youth with developmental (Beets et al., 2007a) and visual disabilities (Beets, Foley, Tindall, & Lieberman, 2007b). However, these studies provided a single coefficient to make a simplistic judgment on the accuracy of a pedometer, but did not specifically address how to improve measurement practice.

In addition, most studies that examined psychometric properties of pedometers have used methods rooted in Classical Test Theory and thus delimit any measurement error as random. For example, Schneider et al. (2003) used Cronbach's Alpha to examine intramodel reliability while Crouter et al. (2003) used intraclass correlation coefficients to examine reliability between devices on right and left sides of the body. Both of these methods provide a single coefficient from which reliability is interpreted and overlook any systematic trends within the measurement error observed. Conversely, Kim and Yun (2009) employed Generalizability (G) theory to examine the reliability of pedometers when used by youth with developmental disabilities; however the focus of study was determining the minimum number of days to capture walking behavior of youth with developmental disabilities rather than establishing reliability evidence of specific pedometers. As a statistical analysis, G-theory not only provides reliability coefficients of pedometer measurements, but also identifies systematic sources of variance.

The ability to identify sources of variance is particularly important for improving measurement practice. Measurement error, despite the basic tenets of

Classical Test Theory, is not entirely random. There may be systematic sources of error that are being overlooked. This is particularly troublesome with a sample of individuals with disabilities, like Down syndrome, that may present unique characteristics that could cause systematic measurement error. By examining the systematic sources of variance, in addition to random measurement, the variance components resulting in the greatest error can be identified and improved to increase reliability in the future.

An alternative statistical approach to examine reliability is G Theory (Cronbach, Gleser, Nanda, & Rajaratnam, 1972). G Theory has the potential to answer the previously described gap in the pedometer reliability literature as it provides information on both systematic and random error within the measurement. The framework of G theory enables the total variance of a model to be partitioned and the contribution of each potential source of error or source interactions to be estimated (Morrow, 1989; Shavelson & Webb, 1991). In other words, "G theory attempts to identify and estimate the magnitude of the potentially important sources of error in a measurement." (p. 923; Shavelson, Webb, & Rowley, 1989). A second purpose of G theory is to determine the reliability and/or dependability of the observed scores. Similar to Classical Test Theory, it is purported that there is a true or universe score. This is the true score of a measurement without measurement error. G theory can be used to assess the reliability of measurements and determine if observed scores can be generalized to the true universe score (Shavelson & Webb, 1991). This approach has been used in exercise science to examine sources and proportions of variance in pedometers (Kim & Yun, 2009), accelerometers (Coleman & Epstein, 1998; Kim &

Yun, 2009; Welk, Schaben, & Morrow, 2004), self-report measures (Coleman & Epstein, 1998; Crocker, Bailey, Faulkner, Kowalski, & McGrath, 1997), and systematic observation tools (Taylor & Yun, 2006) as well as to determine the optimal number of raters, trials, days, etc, needed for measurement.

This study presents the application of Generalizability theory for evaluating the systematic sources of error in pedometer measurement. The intended purpose was to determine the sources of variance in pedometer measurement and reliability evidence for use with groups of adults with Down syndrome and adults without a disability during a quasi-free-living walking bout. The study employed two pedometers (Omron HJ-112 and Yamax Digiwalker SW-200) and used multiple units and locations to examine the reproducibility of measurements. These multiple measurements over a single trial enable the individual sources of measurement error to be partitioned, quantified and interpreted as a means for understanding the systematic mechanism of variance. These variance components and reliability coefficients will provide evidence to support the reliability of different pedometers when used by adults with Down syndrome to accompany the current literature on pedometer accuracy.

Methods

Participants

In total, 20 adults with Down syndrome and 24 participants without DS were recruited in a convenience sample. Three participants (2 with DS, 1 without) were excluded due to missing data. The final sample included 18 adults with DS (10 female, 8 male) and 23 adults without a disability (13 female, 10 male) between the ages of 18

and 61 years. Average ages were 29.88±12.99 and 32.43±13.28 years for DS and control groups respectively. All participants with DS self-reported the medical diagnosis of Down syndrome. When necessary, this was confirmed by a parent or legal caregiver. Participation was limited to ambulatory individuals that did not require the use of physical assistance or assistive technology. Anthropometric measurements including height, weight, and waist/hip circumference were taken to the nearest 1 cm and 0.1kg respectively. Descriptive statistics (mean and SD) were calculated for each of the demographic and anthropometric measures and are reported in Table 3.1. The study was approved by Institutional Review Board and all participants provided written informed consent before the start of data collection. When additional assistance was required to complete the informed consent documents, the forms were also signed by the assisting parent or legal caregiver.

Table 3.1: Physical Characteristics of Participants: Descriptive Statistics by Group

Variable	**Down Syndrome (N=18)**	**Control Group (N=23)**	**All (N=41)**
Gender (female/male)	10 / 8	13 / 10	23 / 18
Age (yr)	29.88±12.99	32.43±13.28	31.32±13.05
Height (cm)**	148.97 ± 8.58	171.09 ± 7.51	161.38 ± 13.63
Weight (kg)*	64.93 ± 11.78	76.81 ±17.19	71.59 ± 16.03
BMI ($kg \cdot m^{-2}$)	29.45 ± 5.86	26.15 ± 5.35	27.59 ± 5.75
Waist circumf. (cm)	86.28 ± 14.06	84.77 ± 14.82	86.28 ± 14.06
Hip circumference (cm)	104.73 ± 11.47	104.08 ± 10.66	104.36 ± 10.89
Waist-to-hip ratio	0.84 ± 0.07	0.81 ± 0.09	0.82 ± 0.08

Note. Values are means ± SD; BMI: body mass index; *significant differences between groups, $p < 0.05$; ** significant differences between groups, $p < 0.01$;

Instruments

Two models of pedometers, Omron HJ-112 and Yamax Digiwalker SW-200 were used to measure the walking activity of participants. Each represents a different mechanism used by pedometers to measure steps walked.

The Omron HJ-112 is a piezoelectric pedometer that utilizes two piezoelectric sensors to count steps. In general, a piezoelectric pedometer works like a uni-axial accelerometer (Crouter, Schneider, & Bassett, 2005). This mechanism determines the number of zero-crossings of acceleration versus time curve and can be highly sensitive to the magnitude of strain on the element necessary to count steps (Crouter et. al. 2005; Hasson et. al., 2009). The Omron HJ-112 in particular utilizes a two sensor design and can determine which sensor to use when counting steps. This means that the pedometer can record steps placed at a wider range of angles from vertical than the traditional spring-levered pedometer. Thus, the manufacturer's recommendation includes potential for use at various locations including the waist/belt line, shirt pocket, around neck, in purse/bag, and in pants pocket.

The Omron HJ-112 has been empirically validated and found to be highly accurate (less than 5% random error) across a variety of speeds, positions, and user characteristics (Hasson et al., 2009; Hasson, Pober & Freedson, 2004). Studies of similar models (HF-100, HJ-700IT, BI) and models of other brands using similar technology (New Lifestyles NL2000) have also shown high accuracy (Crouter et al., 2005; Doyle et al., 2007; Lee, Kim & Zhu, 2006; Melanson, et al., 2004). All of these studies have concluded that piezoelectric pedometers are more accurate than spring-

levered pedometers, particularly when participants utilize slower walking speeds or have higher abdominal obesity.

The Yamax Digiwalker SW-200 is a spring-levered pedometer that employs a spring-suspended lever arm. This mechanism uses a pendulum design with the lever arm being displaced by vertical motion at the hip. A step is counted when the lever arm makes contact with a metal surface, which opens and closes an electrical circuit (Crouter et al., 2003; Schneider et al., 2003). Each opening of the circuit records a step, however various models use different means for conducting this movement into an electrical signal. The manufacturer's recommendation for positioning is to secure the unit on the waist line of the pants at a position equal to the mid-line of the thigh.

The empirical evidence for the accuracy of the Yamax Digiwalker SW-200 has been good, particularly under ideal conditions. Numerous studies have found the Digiwalker SW-200 to be among the most accurate pedometers, despite significantly lower levels of accuracy at slower speeds (Le Masurier et al., 2004; Le Masurier & Tudor-Locke, 2003; Swartz et al., 2003; Welk et al., 2000). Similar models from Yamax (DW-500, SW-701) that use the same spring-levered mechanism have shown similarly high levels of accuracy (Bassett et al., 1996; Crouter et al., 2003; Schneider et al., 2003, 2004). The SW-200 has even been used by studies as the criterion measure to validate other pedometers (Schneider et al., 2004), and is widely considered the best spring-levered pedometer available.

All instruments were checked for accuracy before the start of the study, with pedometers demonstrating 1% or less error in a 100 count modified version of a "shake test" (Vincent & Sidman, 2003) included for use in the study.

Procedures

All participants wore a total of eight pedometers on an elastic belt placed at the waistband of the pants. On the right and left sides of the body, four pedometers, two of each model, were positioned as close to the mid-line of thigh as was physically possible for the participant. Additionally, two pedometers, both Omron HJ-112, were placed in each of the front pockets for a total of four. Only Omron HJ-112 pedometers were used in the front pockets because the manufacturer's recommendation for the Yamax Digiwalker SW-200 only includes placement at the waistline. Before the start of the walking trial, all pedometers were manually reset to zero.

Each participant completed one trial consisting of free walking for 20 minutes representative of a short free-living walking bout. This protocol is similar to Lee et al. (2006) and Zhu and Lee (2008) which used a 15 minute guided walking course. In the present student, all participants were accompanied by a researcher throughout the trial. The participant dictated the walking pace and general path throughout the 20 minute period and was free to encounter different surfaces, elevation changes or stairs. The researcher walked to the side and slightly behind the participant to ensure that the pace and path were self-determined. The researcher only intervened when it was time to go back to the starting point or when personal safety was an issue (crossing street). At the end of the 20 minute period, the readings of all 12 pedometers were recorded.

Data Analysis

Descriptive statistics were calculated for each model and location on steps recorded. Independent sample t-tests were used to examine differences between

groups. Differences were also tested for anthropometric and demographic variables previously discussed. Due to established differences in pedometer error between adults with DS and adults without a disability and pedometer models (Pitchford, 2009), all analyses were conducted as four separate, independent groups for each participant group and pedometer model combination.

Generalizability (G) theory was employed to quantitatively determine the sources of variance in pedometer measurement and estimate reliability among groups of adults with and without DS. For the spring-levered pedometer, a two-facet fully crossed design (placement location (2) by inter-unit (2)) was used while for the piezoelectric pedometer, a placement location (4) by inter-unit (2) fully crossed design was employed. Both designs examined the sources of variance in steps recorded during the 20 minute walk. Each variance component was calculated using SAS statistical software with the VARCOMP command. These variance components include subject (σ^2_s), placement location (σ^2_p), inter-unit (σ^2_u), as well as three two-way interactions and the residual component (three-way interaction plus error). The percentage of variance associated with each component from the total variance was then calculated. When a variance component was reported as a negative value, it was reset to zero for all calculations (Morrow, 1989).

Reliability coefficients were calculated in two ways for the spring-levered and piezoelectric pedometers for both groups with and without DS. These two coefficients represent differences between relative and absolute decisions made from the results. All reliability coefficients within G theory are used to determine if observed scores (steps recorded by pedometers) can be generalized to an individual's true score or

universe score (actual walking activity; Shavelson & Webb, 1991). Φ coefficients were calculated to make absolute decisions and include all sources of variance. The Φ coefficient was calculated using the following equation derived from Shavelson and Webb (1991):

$$\mathbf{\Phi} = \frac{\boldsymbol{\sigma^2 s}}{\boldsymbol{\sigma^2 s} + \frac{\boldsymbol{\sigma^2 p}}{\boldsymbol{n_p}} + \frac{\boldsymbol{\sigma^2 u}}{\boldsymbol{n_u}} + \frac{\boldsymbol{\sigma^2 sp}}{\boldsymbol{n_p}} + \frac{\boldsymbol{\sigma^2 su}}{\boldsymbol{n_u}} + \frac{\boldsymbol{\sigma^2 pu}}{\boldsymbol{n_p n_u}} + \frac{\boldsymbol{\sigma^2 spu, e}}{\boldsymbol{n_p n_u}}}$$

G coefficients were calculated to make relative decisions where interpretations are made between individual standings and include all interaction terms between the facets and the object of measurement (σ^2_{sp}, σ^2_{su}). The G coefficient was calculated using the following equation derived from Shavelson and Webb (1991):

$$\mathbf{G} = \frac{\boldsymbol{\sigma^2 s}}{\boldsymbol{\sigma^2 s} + \frac{\boldsymbol{\sigma^2 sp}}{\boldsymbol{n_p}} + \frac{\boldsymbol{\sigma^2 su}}{\boldsymbol{n_u}} + \frac{\boldsymbol{\sigma^2 spu, e}}{\boldsymbol{n_p n_u}}}$$

Both coefficients are utilized to determine the influence of subject variance on reliability for each group.

Finally, Bland-Altman plots (Bland & Altman, 1986) were created to provide additional convergent validity and show the distribution of step differences between the Digiwalker and Omron pedometers. This plot will also visually demonstrate any systematic overestimation or underestimation of a particular pedometer model relative to the other. Each plot is for a specific group, placement location and unit. Solid lines represent the 95% confidence interval (1.96 x SD) for the sample. The scales of each axis have been standardized for easy comparison.

Results

For the descriptive and anthropometric results, there were significant differences between groups on height and weight. Differences neared significance for body mass index. All other variables including age, waist circumference, hip circumference and waist-to-hip ratio were not significantly different.

During the 20 minute walking trial, the participants walked an average of 2014 ± 358 steps. When examined by groups, adults with DS walked 2025 ± 413 steps while the control adults walked 2004 ± 314 steps. Descriptive statistics of steps walked by model and location are reported in Table 3.2. Differences between groups at each model and location were not statistically significant (p>.1).

Table 3.2: Pedometer recorded steps during 20 minute walk

		Down Syndrome (N=18)	Control Group (N=23)	All (N=41)
Yamax Digiwalker SW-200				
Right Hip	1	1993 ± 591	1969 ± 365	1980 ± 471
	2	1898 ± 480	1903 ± 327	1901 ± 396
Left Hip	1	2064 ± 399	2047 ± 302	2054 ± 343
	2	1970 ± 453	1962 ± 321	1966 ± 379
Omron HJ-112				
Right Hip	1	2057 ± 353	2058 ± 221	2058 ± 283
	2	2012 ± 456	2072 ± 255	2046 ± 354
Left Hip	1	2097 ± 343	2037 ± 261	2063 ± 297
	2	2076 ± 363	2061 ± 235	2067 ± 294
Right Pocket	1	2077 ± 307	1992 ± 345	2029 ± 327
	2	2064 ± 302	1995 ± 360	2025 ± 311
Left Pocket	1	1931 ± 586	1915 ± 511	1922 ± 538
	2	2066 ± 325	2040 ± 300	2052 ± 307

Note. Values are means ± SD.

One participant with Down syndrome was determined to be a severe outlier on multiple data points resulting in skewed results. This was observed through large differences between steps recorded by Digiwalker and Omron pedometers at the same location. For example, one of pedometer only recorded 7 steps while other pedometers at that location recorded upwards of 2080 steps. Differences at specific data points were 1.5 to 5.7 times greater than the second greatest variance at that location. All subsequent analyses have been conducted with this participant removed, for a sample of 17 in the DS group.

Adults with Down Syndrome

The variance components among measurements for adults with Down syndrome differed between pedometer models. The variance components and relative magnitude for each model are presented in Tables 3.3 and 3.4.

*Table 3.3: Variance Component Estimates and Relative Magnitudes of Digiwalker SW-200 for adults with Down Syndrome****

Source of Variation	Sum of squares	Df	Mean squares	Estimated variance components	Relative magnitude**
Digiwalker SW-200					
Subjects (s)	8113361	16	507085	114633.7	75.11%
Placement (p)	7062.49	1	7062.49	0*	0%
Inter-unit (u)	225400	1	225400	5619.4	3.68%
s x p	434823	16	27176	5487.6	3.60%
s x u	601200	16	37575	10686.9	7.00%
p x u	12966	1	12966	0*	0%
Residual (s x m x p x u, e)	259220	16	16201	16201.2	10.61%
Total	9654032	67	--	152628.8	100%

*Table 3.4: Variance Component Estimates and Relative Magnitudes of Omron HJ-112 for adults with Down Syndrome****

Source of Variation	Sum of squares	Df	Mean squares	Estimated variance components	Relative magnitude**
Omron HJ-112					
Subjects (s)	14399515	16	899970	110605	91.01%
Placement (p)	20020	3	6673.39	0*	0%
Inter-unit (u)	70.62	1	70.62	0*	0%
s x p	663475	48	13822	3222.3	2.65%
s x u	138957	16	8684.82	326.74	0.27%
p x u	11705	3	3901.70	0*	0%
Residual (s x m x p x u, e)	354137	48	7377.86	7377.9	6.07%
Total	15587880	135	--	121531.94	100%

Note. *Negative variance components were set to zero;
** Relative magnitude was calculated using readjusted estimated variance divided by total variance.
***One outlier participant removed, N=17.

For the spring-levered arm pedometer (Digiwalker SW-200) the largest source of variance in steps taken during the 20 minute walking trial was the subject variable (75.11%). This is to be anticipated in most generalizability studies since individual differences are to be expected. The second largest source of variation was the residual term (10.61%) indicating a substantial amount of random error that is still unexplained. The interaction of subject-by-unit was also substantial with approximately 7% of total variance. This indicates that inter-unit reliability varied between participants. Additionally, the inter-unit facet and the subject by placement interaction accounted for an additional 3.68% and 3.60% of total variance respectively. The remaining facets and interactions demonstrated little measurement error. For the piezoelectric

pedometer (Omron HJ-112), the largest variance component was also the subject facet, accounting for 91.01% of total variance. The only other component with substantive magnitude was the residual term (6.07%). This indicates that for adults with DS, over 97% of the variance is attributable to individual differences and random error.

Adults without Down syndrome

Interestingly, the control group demonstrated higher unexplained errors than the adults with DS. For the spring-levered pedometer, the largest source was still from subjects (55.73%), but a considerably high amount of variance was from the residual term (29.65%). Additionally, 10.59% can be attributed to the subject by unit interaction. Similarly for the piezoelectric pedometers, the subject (55.40%) and residual (38.70%) explained most of the variance. A small amount of variance is attributable to the subject by placement interaction (3.80%). The remaining facets and interactions for both pedometer models demonstrated minimal levels of variability. The variance components and relative magnitude for each model are summarized in Table 3.5 and 3.6 for the group of adults without DS.

Table 3.5: Variance Component Estimates and Relative Magnitudes of Digiwalker SW-200 for adults without Down Syndrome

Source of Variation	Sum of squares	Df	Mean squares	Estimated variance components	Relative magnitude**
Digiwalker SW-200					
Subjects (s)	6856364	22	311653	64364.8	55.73 %
Placement (p)	106965	1	106965	2384.2	2.06%
Inter-unit (u)	130955	1	130955	2276.0	1.97%
s x p	654134	22	29733	0*	0%
s x u	1291468	22	58703	12230.1	10.59%
p x u	1800.53	1	1800.53	0*	0%
Residual (s x m x p x u, e)	753342	22	34243	34242.8	29.65%
Total	9795029	91	--	115497.9	100%

Table 3.6: Variance Component Estimates and Relative Magnitudes of Omron HJ-112 for adults without Down Syndrome

Source of Variation	Sum of squares	Df	Mean squares	Estimated variance components	Relative magnitude**
Omron HJ-112					
Subjects (s)	11116365	22	505289	56771.9	55.40 %
Placement (p)	246029	3	82010	810.28	0.79%
Inter-unit (u)	79390	1	79390	421.2	0.41%
s x p	3130579	66	47433	3889.3	3.80%
s x u	953382	22	43336	920.3	0.90%
p x u	110875	3	36958	0*	0%
Residual (s x m x p x u, e)	2617186	66	39654	39654.3	38.70%
Total	18253805	183	--	102467.28	100%

*Negative variance components were set to zero;
** Relative magnitude was calculated using readjusted estimated variance divided by total variance.

Reliability and Validity

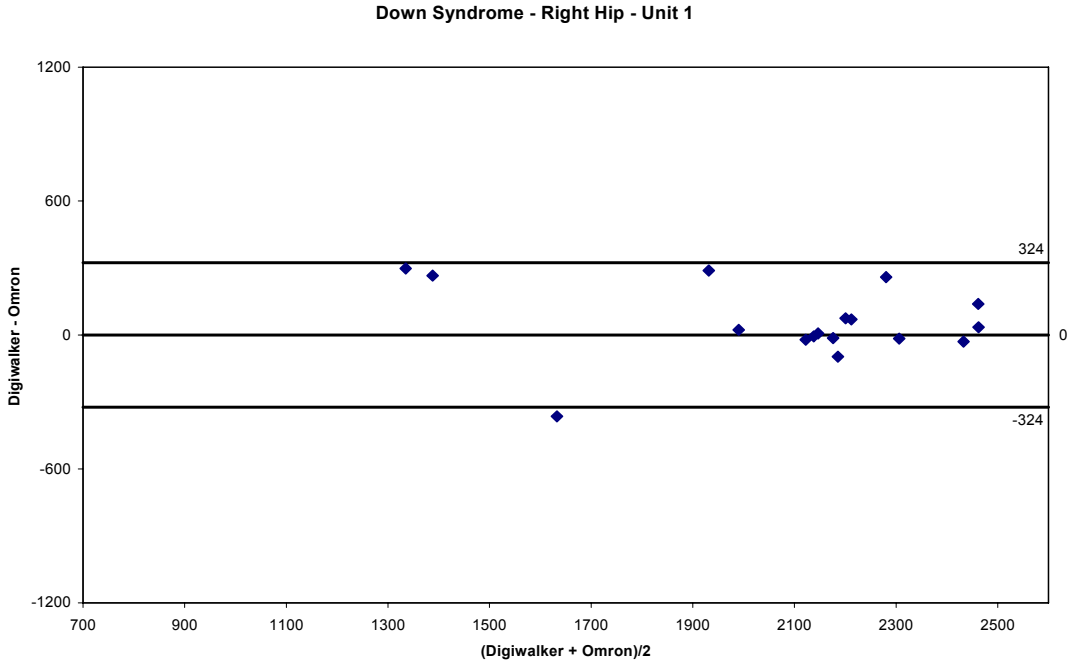
Reliability coefficients differed between groups and pedometer models, but largely between generalizability decisions. The reliability coefficients are reported in Table 3.7. The phi (Φ) coefficient is used for making absolute decisions about the reliability of step counts. For the spring-levered pedometer, Φ was 0.61 for the DS group and 0.57 for the control group. For the piezoelectric pedometer, Φ was 0.79 for the DS group and 0.73 for the control group. The G coefficient (G) is used for making decisions based on relative standing. For the spring-levered pedometer, G was 0.90 for the DS group and 0.81 for the control group. For the piezoelectric pedometer, G was 0.98 for DS group and 0.90 for the control group. These coefficients provide moderate to strong evidence that pedometer measurements using the piezoelectric pedometer can reliable for both groups and can be generalized to the universe score. The reliability evidence for the spring-levered is particularly weak when making absolute decisions, but can be considered moderate to good when making relative decisions. In general, these coefficients favor the reliability of the piezoelectric over the spring-levered pedometer.

Table 3.7: Reliability Coefficients

	Phi Coefficient (Φ)	**G Coefficient (G)**
Down Syndrome Group		
Digiwalker SW-200	0.61	0.90
Omron HJ-112	0.79	0.98
Control Group		
Digiwalker SW-200	0.57	0.81
Omron HJ-112	0.73	0.90

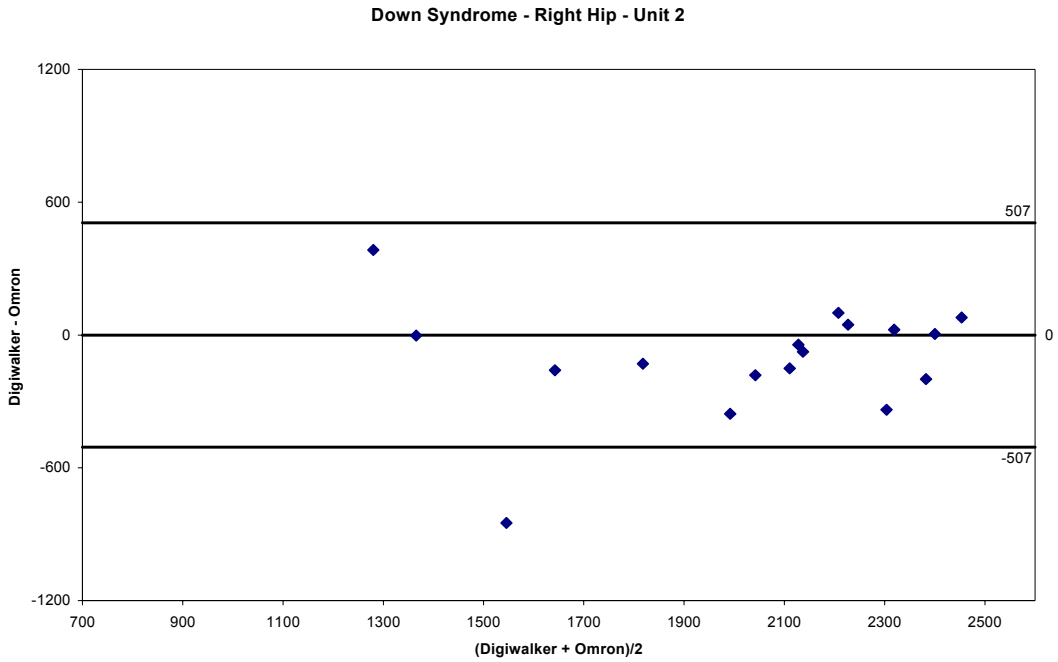
Additionally, the Bland-Altman plots, presented in Figures 3.1 – 3.8, show the systematic nature of measurements between spring-levered and piezoelectric pedometers. Each plot shows the difference between spring-levered and piezoelectric recorded steps by the average of those steps recorded for a specific group, placement location, and unit.

Figure 3.1. Bland-Altman plot of Right Hip, Unit 1 for Down syndrome group



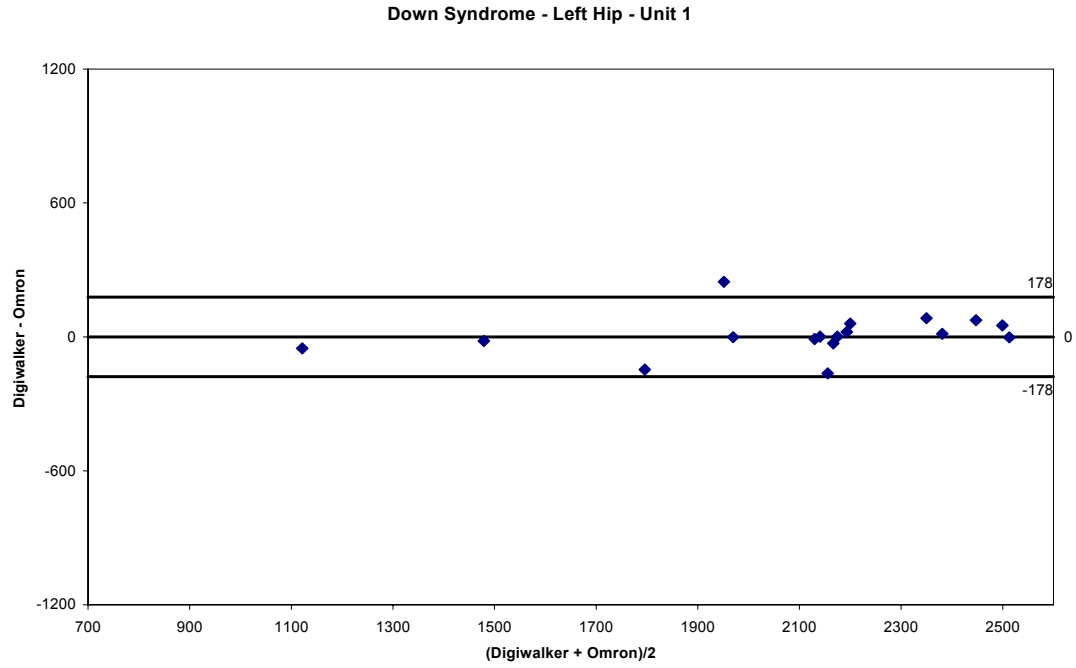
Note: 95% confidence interval: ± 324 steps.

Figure 3.2: Bland-Altman plot of Right Hip, Unit 2 for Down syndrome group



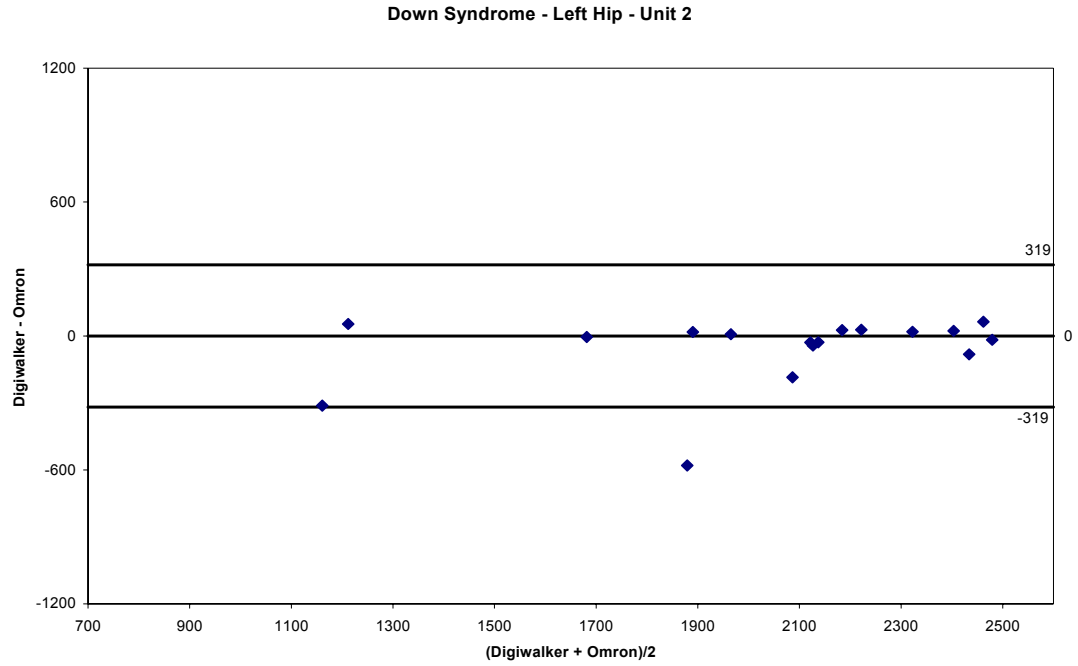
Note: 95% confidence interval: ± 507 steps.

Figure 3.3: Bland-Altman plot of Left Hip, Unit 1 for Down syndrome group



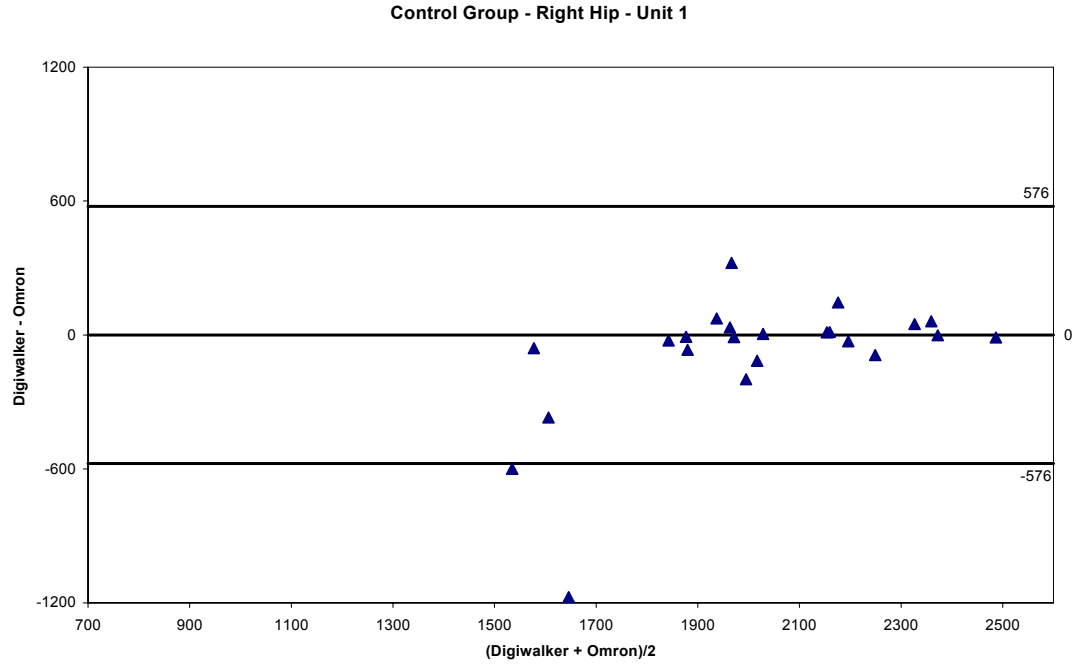
Note: 95% confidence interval: ± 178 steps.

Figure 3.4: Bland-Altman plot of Left Hip, Unit 2 for Down syndrome group



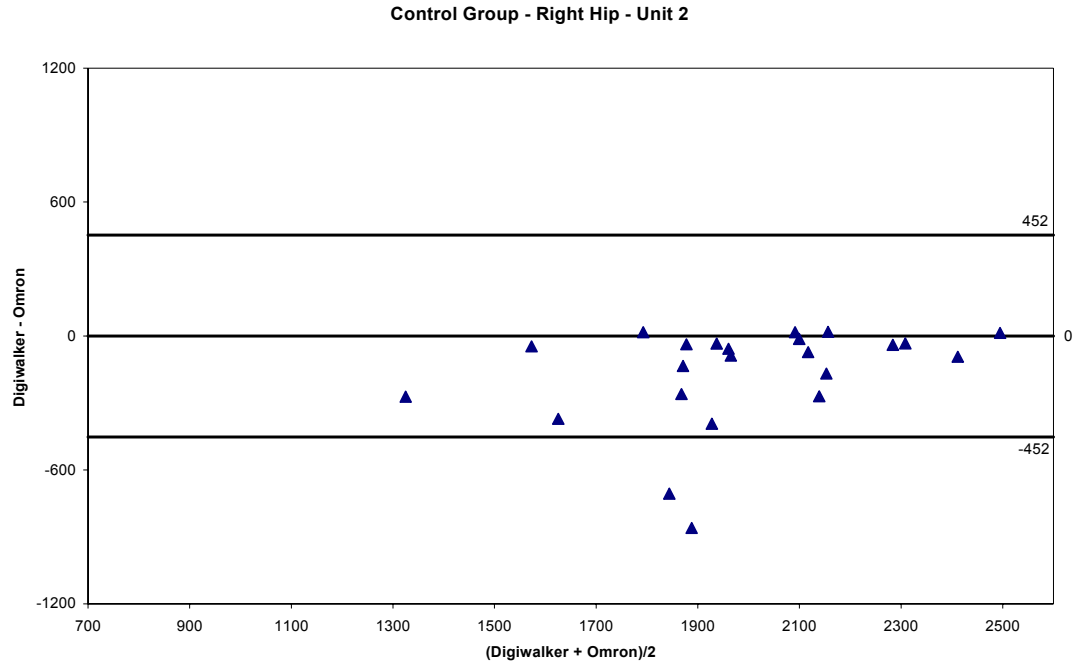
Note: 95% confidence interval: ± 319 steps.

Figure 3.5: Bland-Altman plot of Right Hip, Unit 1 for without DS group



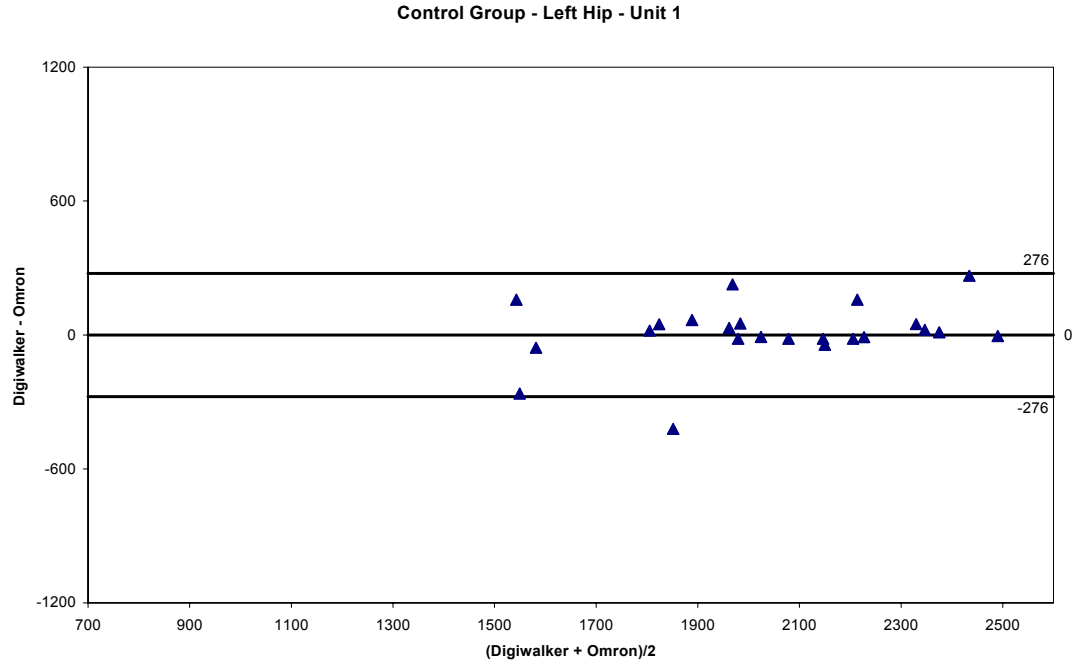
Note: 95% confidence interval: ± 576 steps.

Figure 3.6: Bland-Altman plot of Right Hip, Unit 2 for without DS group



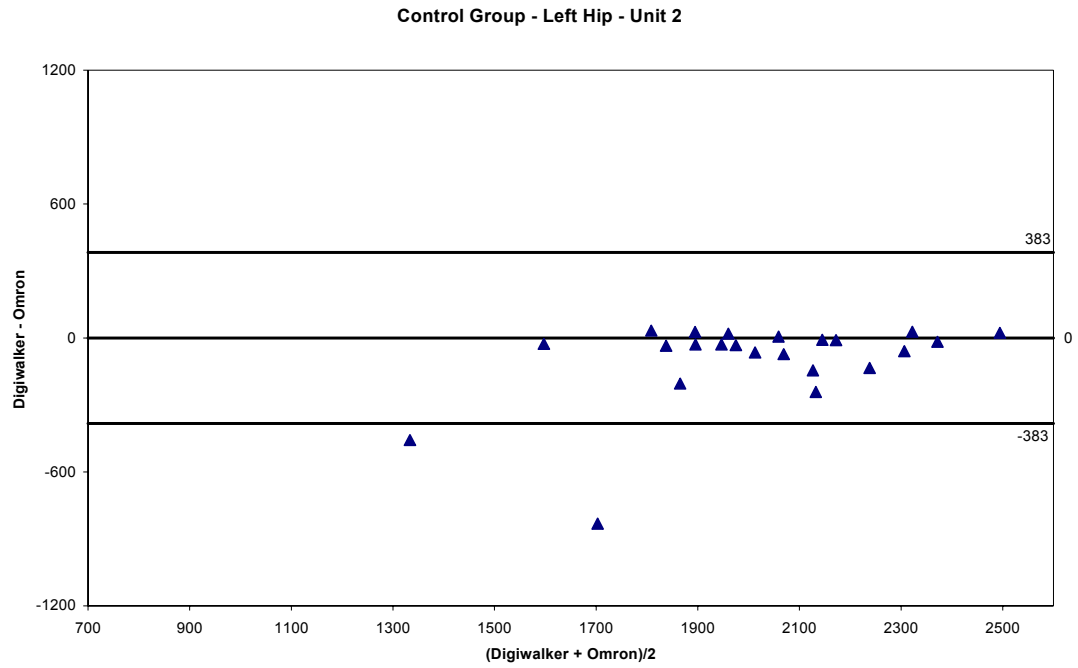
Note: 95% confidence interval: ± 452 steps.

Figure 3.7: Bland-Altman plot of Left Hip, Unit 1 for without DS group



Note: 95% confidence interval: ± 276 steps.

Figure 3.8: Bland-Altman plot of Left Hip, Unit 2 for without DS group



Note: 95% confidence interval: ± 383 steps.

Overall, the plots show that differences between models are fairly consistent across locations and between groups. A possible form of systematic error that is represented is that when differences occur, the spring-levered pedometer tends to underestimate steps compared to the piezoelectric pedometer.

Discussion

The present study sought to determine the sources of variance in the measurement of steps taken over a 20 minute bout of free walking. The substantive sources of variance that have been identified through the G study analysis provide insight into the systematic nature of measurement error. Substantive sources of variance were considered to be any component with a relative magnitude of 3% or

more. The results also provide information to ascertain the reliability of measurements for adults with and without Down syndrome.

For the sample of adults with Down syndrome, variance due to the subject represented the largest source of variation with 75.11% and 91.01% for the spring-levered and piezoelectric pedometer respectively. This is expected due to the differences in steps walked between individuals. The remainder of variance components differed in magnitude between the pedometer models.

For the spring-levered arm pedometer, there were three other sources of variance providing information about measurement error. The three substantive sources of variance included the subject-by-unit term (7%), the inter-unit facet (3.68%), and the subject-by-placement term (3.60%). The first source of variance indicates that for certain subjects, there were issues with inter-unit reliability. The additional 3.68% associated with the inter-unit facet further demonstrates the small but substantive source of measurement error. For adults with DS, there does appear to be an issue with inter-instrument reliability. The subject-by-placement term indicates that for different subjects there were differences between the four pedometer placement locations. Since the spring-levered pedometer was only used at two locations, the right and left hip, this variance component indicates side differences were present for some participants.

For the piezoelectric pedometer, there were no other substantive sources of variance for adults with DS. This indicates that the piezoelectric technology has less systematic error and greater inter-unit reliability than the spring-levered arm pedometer. This lack of systematic error has also been reflected in the accuracy and

validity of piezoelectric pedometer, which have largely been found to be highly accurate (Crouter et al., 2005; Hasson et al., 2009; Melanson et al., 2004).

Among the group of adults without a disability, the largest source of variation was also the subject facet, accounting for 55.73% and 55.40% of total variance for spring-levered and piezoelectric pedometers respectively. Similar to the group with DS, the remaining substantive sources of variance differed between models. For the spring-levered arm model, the other substantive source of variance was the subject by unit interaction (10.59%). This further illustrates that inter-unit reliability is an issue for the spring-levered pedometer, as unit variability was demonstrated in both groups. The Yamax Digiwalker SW-200 and similar models have previously shown high inter-model reliability, however other models using similar mechanisms have been far less reliable (Crouter et al., 2003; Schneider et al., 2004). For the piezoelectric model, there was a small variance component associated with the subject by placement interaction (3.80%). This demonstrates that pedometer readings differed between placement locations for certain subjects. It is possible that this source of variance is the result of the inclusion of the front pants pocket location for pedometer placement with the piezoelectric model. The methodological design of this study would indicate that use of pedometers in the front pockets may be a possible explanation for this variance component. Despite manufacturer recommendation that the piezoelectric pedometer can be used in the front pockets, the accuracy at this location has not been substantiated thoroughly in empirical research (Lee et al., 2006; Zhu & Lee, 2008).

An interesting finding from the results is the large magnitude of the residual term for the control group with both models (29.65% and 38.70% respectively)

compared the group with DS (10.61% and 6.07%). First, this indicates that there is random error that is still not explained by the facets that have been included. For the control group, this random error is large. While the results have added to the understanding of systematic sources of variance, it is clear that the measurement of walking activity using pedometers has inherent systematic and random errors. Given the assumption in empirical research and many statistical analyses that measurement is without error, the combination of systematic error explained and random error that remains may be problematic. Second, this residual component, as well as the other variance components, should be viewed in light of total measurement variance. The total variance was larger for the group with DS compared to the control group, as was measurement with the spring-levered pedometer compared to the piezoelectric. While there may be larger proportions in the magnitude of random error, in reality there less total variance for the control group.

Comparatively, the distribution of variance components observed demonstrates different formations of variance indicating systematic differences not only internal to the measurement, but also between pedometer models and adults with and without DS. The major differences between adults with and without Down syndrome were the relative magnitude of the subjects and residual terms. First, for both groups and models the subject term resulted in the largest explained portion of variance. The DS group had higher relative magnitudes (75.11% and 91.01%) compared to the control group (55.73% and 55.40%). Conversely, the residual term was much larger for the control group (29.65% and 38.70%) compared to the group with DS (10.61% and 6.07%). The subject facet differences are understandable as the sample of adults with

DS was highly heterogeneous and had more within-subjects variance. Another possibility to explain these differences is that the 20 minute walking bout was not appropriate for all participants. While this period appears to be appropriate for adults with DS, the lower proportion of subject variance for the control group could indicate that 20 minutes was not enough time for participants without DS to differentiate themselves from one another. Across all measurements, the participants with DS walked an average of 2025 steps (SD=413), while the control group walked an average of 2004 steps (SD=314). The small amount of variance, particularly for the control group, may have limited findings.

More importantly, once the percent of variance associated with subjects and the residual are accounted for, the remainder of systematic error is relatively similar between groups, but not pedometer models. The spring-levered pedometer appears to have issues with inter-unit reliability with similar variance components for groups with and without DS for the subject by unit interaction (7% and 10.59%), and the inter-unit facet (1.97% and 3.68%). The only difference is that the subject by placement interaction had substantial variance (3.60%) for adults with DS, but not for the control group (0%). The piezoelectric pedometer appears to be relatively free of systematic error with only the subject by placement interaction having a variance component of 2.65% for adults with DS and 3.8% for the control group. The lack of systematic error with the piezoelectric pedometer is similar to the results of Kim and Yun (2009) that found negligible to no variance related to the instrument or subject by instrument interaction when using the Omron HJ-112 with youth with developmental disabilities.

Similar to the differences observed in the sources of error, the reliability differed for the spring-levered and piezoelectric pedometers for adults with and without Down syndrome. These coefficients are highly influenced by the differences in variance associated with subjects and variance concurrent with random error. For the spring-levered pedometer, the reliability coefficients were $\Phi = 0.61$ and $G = 0.90$ for adults with DS, and $\Phi = 0.57$ and $G = 0.81$ for the control group. For the piezoelectric pedometer, the reliability coefficients were $\Phi = 0.79$ and $G = 0.98$ for adults with DS, and $\Phi = 0.73$ and $G = 0.90$ for the control group. These are coefficients are very similar to intraclass correlation coefficients and represent high levels of reliability/dependability when greater than 0.80. In the present study, absolute decisions of reliability are most important since the absolute level of performance is of interest. Thus, Φ coefficient, referred to as the index of dependability (Shavelson & Webb, 1991), will be interpreted as it defines all variance components, except the subject, as error. The Φ coefficients for both groups represent very low levels of reliability for the spring-levered pedometer but moderately high levels of reliability for the piezoelectric pedometer. This provides evidence that step counts from the piezoelectric model can be generalized to the individual's universe score.

While the methodology of G theory does not address the absolute error of a single pedometer measurement (ex. pedometer measurement – observed count), the Bland-Altman plots do provide some additional validity evidence. The reliability and variance components are clearly different between the spring-levered and piezoelectric pedometers; however the Bland-Altman plots show that the distribution of differences

between the two measurements is generally good. When differences do occur between models, the spring-levered pedometer tends to underestimate steps compared to the piezoelectric. This shows that while accuracy, in this case through convergent validity, between two models is good, there may still be issues related to systematic sources of error and different levels of reliability. This supports the notion that both reliability and validity need to be considered when assessing the psychometric properties of instruments.

The purpose of the study was to determine the sources of systematic error in pedometer measurement for adults with and without Down syndrome to gain evidence of the reliability of pedometers to improve future research. The reliability evidence suggests that the piezoelectric pedometer measurement can be generalized to the universe score of walking activity. This recommendation for piezoelectric pedometers can be extended to both adults with and without DS due to moderate Φ and high G coefficients. Pitchford (2009) found that piezoelectric pedometers are more accurate for adults with DS than spring-levered arm pedometers, particularly at slower speeds. The large difference in reliability coefficients and trend of spring-levered underestimation relative to piezoelectric in the current study also signify that the pedometer models may not be equally reliable for adults with DS. Furthermore, given the current rate of obesity and overweight status, piezoelectric pedometers may be more effective for all individuals as they are more resistant to error with increasing abdominal obesity (Crouter et al., 2005). For future research, the use of piezoelectric pedometers is recommended when pedometers are utilized in measuring walking activity in adults with and without DS.

The current study is limited by the convenience sample employed as it may not adequately represent either Down syndrome or general population which limits the generalizability of the results. Furthermore, the use of a 20 min walking bout where all participants were continuously walking may not have generated enough variance, as evidenced by the relatively low magnitude of variance explained uniquely by subjects in the control group, also limits the results. Despite these limitations, the findings of this study expand the reliability evidence on pedometers, particularly among adults with DS. There still remains paucity in the body of evidence for the accuracy and reliability of pedometers for adults with disabilities, particularly Down syndrome. However, there is now initial evidence for acceptable reliability in piezoelectric pedometry.

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CHAPTER 4: CONCLUSIONS

The following summary includes 1) overall research conclusions from the two studies presented including future research directions and 2) research conclusions to specifically address each research question presented in Chapter 1.

Overall Conclusions

The primary purpose of the two studies presented was to determine which type of pedometer mechanism, spring-levered arm or piezoelectric, was most appropriate for use in research for adults with Down syndrome. This was examined from both accuracy and reliability viewpoints. Overall, the results of both studies indicate that piezoelectric pedometers are better suited to measure walking activity for both adults with and without Down syndrome.

The piezoelectric pedometer was found to be more accurate, particularly at slower walking speeds than the spring-levered pedometer for both adults with and without Down syndrome. The ability to accurately measure walking activity at slower speeds is particularly useful as adults with DS tend to walk at speeds slower than the general population. However, it should be noted there remains approximately 7% to 8% measurement error in steps recorded by piezoelectric pedometers when used by adults with DS. This proportion of measurement error should be taken into account in future studies.

The piezoelectric pedometer was also found to be highly reliable for both adults with and without DS. Compared to the spring-levered pedometer, there were less systematic sources of variance and higher reliability coefficients. The spring-levered arm pedometer appears to have some issues with inter-unit reliability, despite using one of the most accurate and reliable brands and models available (Yamax Digiwalker SW-200). The piezoelectric pedometer is recommended for future use for adults with and without Down syndrome due to the lack of systematic error and acceptable reliability.

These studies determined that piezoelectric pedometers should be used when measuring walking activity of adults with Down syndrome compared to spring-levered pedometers. Additional areas for future research include determining if piezoelectric pedometers are accurate for youth with Down syndrome, as well as both adults and youth with intellectual disabilities. Specific evidence related to adults with Down syndrome has been presented, however despite being a subpopulation of intellectual disability, these results cannot be generalized beyond DS.

A final area for future research is determining if the same recommendations for walking activity can be applied to this population. The standard recommendation for daily walking activity continues to be 10,000 steps per day (Tudor-Locke, Hatano, Pangrazi & Kang, 2008). However, it has been shown that adults with Down syndrome walk with a less efficient gait, trading efficiency for stability (Agiovlasitis, McCubbin, Yun, Pavol & Widdick, 2009). Therefore, the daily recommendation for walking activity may need to be adjusted to account for these differences. The current study has shown that a pedometer can accurately and reliably measure walking

activity, so determining how much an individual with Down syndrome needs to walk to achieve health benefits as part of a pedometer intervention is the next logical step.

Specific Research Conclusions

1. Were there significant differences in absolute pedometer error between the measurements of spring-levered and piezoelectric pedometers?

A 2 x 2 x 3 (group by model by speed) repeated measures ANOVA on absolute percent error scores revealed that there was a significant interaction between the pedometer model and walking speed, $F(2,84) = 13.14$, $p<0.001$, $\eta^2 = .24$, including significant main effects for pedometer model, $F(1,42) = 16.87$, $p<0.001$, $\eta^2 = .29$. The average absolute error rate ranged from 1.05% to 8.02% for the Omron HJ-112 compared with 2.87% to 22.39% for the Yamax Digiwalker SW-200. This indicates that there are significant differences in absolute error between pedometer models. The piezoelectric pedometer appears to have significantly less error than spring-levered model.

2. Were there significant differences in absolute pedometer error at faster and slower walking speeds for spring-levered and piezoelectric pedometers?

A 2 x 2 x 3 (group by model by speed) repeated measures ANOVA on absolute percent error scores revealed that there was a significant interaction between the pedometer model and walking speed, $F(2,84) = 13.14$, $p<0.001$, $\eta^2 = .24$, including significant main effects for walking speed, $F(1.73, 77.13) = 7.56$, $p<0.01$, $\eta^2 = .15$. Post hoc analyses through one-way (speed) repeated measures ANOVA for each

pedometer model revealed that there was a simple main effect for speed with the spring-levered pedometer, $F(2,86) = 14.01$, $p<0.001$, $\eta^2 = .25$, but not the piezoelectric pedometer, $F(2,86) = 0.17$, $p>0.8$. Simple contrasts showed that the self-paced and fast speeds were significantly different than the slow speed ($p<0.001$), but were not significantly different from each other ($p>0.4$). This indicates that walking speed was a significant source of error when using the spring-levered pedometer, but not with piezoelectric pedometer. Furthermore, the influence of speed on spring-levered pedometer accuracy results in additional measurement error when walking at speeds of approximately 2 mph.

3. Were there significant difference in absolute pedometer error for spring-levered and piezoelectric pedometers between individuals with Down syndrome and the general population?

A 2 x 2 x 3 (group by model by speed) repeated measures ANOVA on absolute percent error scores revealed that there was a significant main effect for differences between adults with and without Down syndrome, $F(1,42) = 9.06$, $p<0.01$, $\eta^2 = .18$. This indicates that pedometer error was significantly different between the two groups. Average absolute error ranged from 1.05% to 16.44% for the control group compared with 8.02% to 22.39% for adults with DS indicating that pedometer measurements had more error with the DS group. However, there was not a significant group by pedometer model, $F(1,12) = 0.19$, $p>.8$, group by speed, $F(2,84) = 0.32$, $p>.7$, or group by model by speed interaction, $F(2,84) = 0.40$, $p>.6$, indicating that there differences between groups are not restricted to a single pedometer model or walking speed.

4. Was there a significant influence of waist-to-hip ratio on the absolute pedometer error for spring-levered and piezoelectric pedometers?

A 2 x 2 x 3 (group x model x speed) repeated measures ANCOVA with a covariate of waist-to-hip ratio on absolute percent error scores revealed that the inclusion of waist-to-hip ratio explained the differences between spring-levered and piezoelectric pedometers. The main effects for model and speed, as well as subsequent interactions were not significant ($p>0.1$). However, the main effect for groups of adults with and without DS remained significant, $F(1,41) = 7.35$, $p<0.05$, $\eta^2 = .15$. This indicates that differences in individual waist-to-hip ratio explain the additional errors demonstrated by spring-levered pedometers and slower walking speeds. The differences between adults with and without DS remain significant, so there are factors other than pedometer model, walking speed, and waist-to-hip ratio causing additional pedometer error for adults with DS.

5. What were the systematic sources of variance in pedometer measurement unique to groups of participants with and without Down syndrome?

Four separate two-facet (placement location by inter-unit) fully crossed G studies were conducted for groups of adults with and without Down syndrome and the spring-levered and piezoelectric pedometers. The differences in variance components between adults with and without DS were largely restricted to the relative magnitude of variance for the subjects and residual terms. For adults with DS, the subjects term accounted for 75.11% and 91.01% of the total variance versus 55.73% and 55.40% for the control group. For adults with DS, the residual term accounted for 10.61% and

6.07% of total variance versus 29.65% and 38.70% for the control group. Systematic sources of variance were very similar between groups, but not between models. The spring-levered arm pedometer showed problems with inter-unit reliability through substantive variance components for the subject by unit interaction (7% and 10.59%), and the inter-unit facet (1.97% and 3.68%). There was also a small variance component within the DS group for the subject by placement interaction (3.60%). Conversely, the piezoelectric pedometer was relatively free of systematic error with only the subject by placement interaction demonstrating a small variance component (2.65% and 3.8%). These results indicate that there were more walking behavior differences within the DS group and more unexplained random error in the control group. The spring-levered pedometer may have reliability issues related to differences between individual units placed at the same location, but these differences are also moderated by individual participant. The piezoelectric model appears to have little systematic error, with only small amounts of variance coming from pocket locations.

6. Did the spring-levered and piezoelectric pedometer demonstrate acceptable levels of reliability for both individuals with and without Down syndrome?

Reliability coefficients were calculated for both absolute (Φ) and relative (G) decisions. The Φ coefficients for the group of adults with DS were 0.61 for the spring-levered pedometer and 0.79 for the piezoelectric. The Φ coefficients for the control group were 0.57 for the spring-levered pedometer and 0.73 for the piezoelectric. The G coefficients for the group of adults with DS were 0.90 for the spring-levered pedometer and 0.98 for the piezoelectric. The G coefficients for the

control group were 0.81 for the spring-levered pedometer and 0.90 for the piezoelectric. These results show that the reliability coefficients were higher for adults with DS than the control group across both models; however these differences did not change the interpretation of reliability. The piezoelectric pedometer demonstrated better reliability than the spring-levered pedometer for both adults with and without DS, particularly when making relative decisions about reliability. From these coefficients, it appears that the step counts from piezoelectric pedometers can be generalized to the true universe score.

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APPENDICES

APPENDIX A: REVIEW OF LITERATURE

As the health benefits of physical activity become better understood, there is now greater emphasis on the measurement and promotion of physical activity. This emphasis within groups with disabilities has substantially increased due to growing evidence of increased secondary health conditions due to inactivity in these populations. A common method for measuring physical activity is the pedometer, an instrument that counts the number of steps an individual takes. This instrument, however, has not been extensively validated in many disability populations, including Down syndrome (DS). The accuracy of pedometers for individuals with DS can be deemed as questionable given what is known about the difference in mechanisms used in pedometers, the gait and body characteristics of individuals with DS, and their physical activity habits. The literature related to these topics will be discussed in this appendix. For organizational purposes, the literature will be presented under the following topics: (1) Overview of Pedometers, (2) Sources of Error in Pedometry, (3) Pedometers and Disability, (4) Gait Analyses in Down Syndrome, (5) Walking Activity, (6) Physical Activity, (7) Generalizability Theory, and (8) Application of Generalizability Theory in Exercise Science.

Overview of Pedometers

The pedometer is a simple motion sensing device that can be used to assess physical activity. However the primary measurement of the pedometer is walking activity. There are three basic mechanisms used in pedometers for recording steps

(Crouter, Schneider, Karabulut, & Bassett, 2003; Schneider, Crouter, Lukajic, & Bassett, 2003). The original and most common mechanism is a spring-suspended level arm that records steps as it moves up and down from vertical movement (displacement) at the hip. An electrical circuit is opened and closed by a lever arm moving up and down in response to hip displacement. Each time the circuit is connected, a step is recorded (Crouter et al., 2003; Schneider et al., 2003). Commercially available pedometers that utilize this mechanism include the Yamax Digiwalker SW-200,500,701, and 2000 models, Freestyle Pacer Pro, Sportline 330 and 345 models, Step Keeper HSB-SKM, and Walk4Life LS 2500 and 2525 models. The second mechanism is a glass-enclosed magnetic reed proximity switch. This mechanism is similar to the lever arm, as it utilizes a magnet on the end of a horizontal lever arm. As the arm moves in response to vertical movement (acceleration) at the hip, the magnetic field triggers a proximity switch encased in a glass cylinder, thus recording steps (Crouter et al., 2003; Schneider et al., 2003). Examples of pedometers utilizing this mechanism include the Omron HJ-105, Oregon Scientific PE316CA, and Yamax Skeletone. The third mechanism is similar to a uni-axial accelerometer. This mechanism uses a horizontal beam and a piezoelectric crystal and records steps based on the number of zero-crossings of the instantaneous acceleration versus time curve (Crouter et al., 2003; Schneider et al., 2003). Many new pedometers models are using this piezoelectric mechanism including the Omron HF-100, HJ-104, 105, 112, and BI models, Kenz Lifecorder, and New Lifestyles NL-2000.

There have been numerous studies conducted to determine the psychometric properties of pedometers. Many of these studies have compared specific brands and

models to determine which pedometers are the most accurate and best suited to for research settings. Two major trends have emerged in the literature. First, all pedometers are not created equal. There is a great deal of variability between pedometer brands, models and at times between individual units (Bassett et al., 1996; Crouter, et al., 2003; Melanson et al., 2004; Schneider, et al., 2003; Schneider, Crouter & Bassett, 2004). Pedometers that are manufactured in Japan have been consistently shown to be the most accurate. These brands have demonstrated on average a 3% margin of error, many of which with less than 1% error. (Crouter et al., 2003; Schneider et al., 2003, 2004). This may be due to the industrial standard of that country, which dictates that pedometers miscount less than 3 steps out of 100 (3%) (Hatano, 1993). Second, piezoelectric pedometers typically demonstrate more resistance to error than spring-lever arm pedometers, particularly at lower speeds and for obese or overweight individuals. (Melanson et al., 2004; Crouter, Schneider & Bassett, 2005).

While there were initial studies in the 1970s and 1980s to determine the accuracy of the original, purely mechanical pedometers, much of the recent research has examined multiple brands/models of pedometers. Bassett et al. (1996) studied five electronic pedometers and had participants walk on a variety of surfaces and speeds. Across the various aspects of the study there were differences between the pedometer models, however the Yamax Digiwalker SW-500 was found to be the most accurate, particularly at lower speeds. This Digiwalker model demonstrated an average percent error 0.7%.

Subsequent multi-brand/multi-model studies have followed as the technology of pedometers has grown and improved. A few studies conducted similar studies examining accuracy and reliability of the functions from 10 electronic pedometers. First, Crouter et al. (2003) examined the validity of the 10 pedometers while walking on a treadmill for measuring steps, distance and energy cost. On steps, the direct measure of the pedometer, the results show that six pedometers were accurate within 1% at a speed of 80 $m \cdot min^{-1}$ (Yamax Skeletone EM-180, Omron HJ-105, Yamax Digiwalker SW-701 (DW), New Lifestyles NL-2000 (NL), Kenz Lifecorder (KZ), and Walk4Life LS 2525 (WL). Overall, pedometers were found to be less accurate at lower speeds. The DW, NL, WL and KZ were acceptable at 54 $m \cdot min^{-1}$ as demonstrated by the difference between actual and recorded steps not being statistically significant, however only the Digiwalker demonstrated acceptable accuracy at all speeds. Second, Schneider et al. (2003) examined the accuracy and reliability of the same 10 pedometer models over a 400m overground walk. The Digiwalker, Kenz Lifecorder, and New Lifestyle pedometers were found to be the most accurate, all within 3% of actual steps. Surprisingly, the Omron pedometer, which was one of the most accurate in the previous study, was consistently one of the least accurate during this study. Initially these two results would indicate differences between individual pedometers. However, the Omron pedometers also demonstrated high intraclass correlations and intermodal reliability of 0.83 and 0.99 respectively in the two studies which means that manufacturing quality is not necessarily the issue.

Sources of Error in Pedometry

There have been a number of identified sources of error on the steps measured by pedometers in the general population. The two most common sources of error reported in the literature are gait speed and pedometer tilt. Each of these potential sources of error are important to the Down syndrome population, as gait speed tends to be slower and common body characteristics are indicative of causing pedometer tilt.

Gait Speed. The walking speed of the individual has been shown to be the main source of error in pedometer measurements throughout the use of the instrument. It is clear that at slower walking speeds pedometers in general are less accurate and typically underestimate steps, particularly for lever-arm mechanism models (Bassett et al., 1996; Crouter et al., 2003). Some studies have shown that the newer piezoelectric pedometers are more accurate at these lower speeds (Crouter et al., 2003; Melanson et al., 2004). This indicates that at slower walking speeds, there may not be enough vertical displacement at the hip to register a step using the traditional mechanism. Despite this evidence, the Digiwalker models have shown consistent accuracy across all speeds in many studies (Bassett et al., 1996; Crouter, et al., 2003; Le Masurier & Tudor-Locke, 2003; Le Masurier, Lee & Tudor-Locke, 2004; Schneider, et al., 2003, 2004).

Pedometer tilt. In the traditional pedometer mechanism, the unit must be positioned vertically in order for the lever-arm to move correctly in response to vertical motion. If the pedometer is placed at a tilted angle away from vertical, the pedometer will not function correctly. This is problematic, especially for measuring walking activity in individuals that are overweight or obese. Studies have mentioned

pedometer tilt as possible cause of inaccurate measurements (Schneider et al., 2003; Swartz, Bassett, Moore, Thompson & Strath, 2003). For example, Swartz et al. (2003) investigated the effects of body mass index (BMI) on the accuracy of pedometers. The results indicate that there were no significant differences in accuracy based on BMI groupings, however the authors note that pedometer tilt, which was not measured could explain some of the errors seen, particularly at speeds less than 80 m/min^{-1}. Despite the notions that pedometer tilt could be large source of error, only Crouter et al (2005) have measured pedometer tilt and included it in the analysis.

The results of the Crouter et al. (2005) study show that pedometer tilt, regardless of the direction of that tilt, had the largest effect on influencing pedometer accuracy. In the study, overweight and obese individuals walked on a treadmill at a variety of speeds with a Digiwalker SW-200 and a New Lifestyles NL-2000. The NL-2000 (piezoelectric) was more accurate than the SW-200(mechanical level arm) at all speeds, particularly slower ones. While the NL-2000 was unaffected by any anthropometric variables, the SW-200 was more affected by pedometer tilt. This means that if a spring-lever arm pedometer is tilted too far, it will underestimate step counts, while a piezoelectric will be largely unaffected.

Anthropometric measurements such as BMI, waist circumference and hip to waist ratio have also been examined as potential sources of error. Despite inconclusive results, these variables are most likely a mediator of pedometer tilt (Schneider et al., 2003; Swartz, et al., 2003).

Pedometers and Disability

While there is an extensive body of research examining the psychometric properties of pedometers in the general population, there is relatively little research devoted to pedometers among populations with disabilities. The rationale for examining pedometer accuracy in the Down syndrome population is that gait abnormalities, side dominance and the common sources of variability in pedometers discussed previously may be more common in this population. While pedometer accuracy has not been specifically addressed in this population, a few studies have examined pedometer accuracy for adults with intellectual disabilities and neurological disabilities as well as youth with developmental disabilities and visual impairments.

Stanish (2004) examined the accuracy of pedometers in a population of individuals with intellectual disabilities, including those with and without Down syndrome. A sample of 11 adults with intellectual disabilities without DS (8 females, 3 males) and 9 adults with DS (4 females, 5 males) each walked a total of four - 400m trials with two Yamax Digiwalker SW-500 pedometers on each side o the body. Trials included walking on an indoor gym surface and outdoor gravel surface at a normal pace and an unspecified fast pace. Intraclass correlation coefficients were calculated from pedometer-recorded steps and actual steps taken to determine consistency, however this was done for all 20 participants without an attempt to examine any differences between individuals with and without DS. The results show very high consistency between pedometer-recording and actual steps during all situations and on both side placements. All coefficients were 0.95 or greater and there were no significant differences between pedometer placement, walking surface, or

walking speed. This is consistent with Yamax Digiwalker results in the general population with an ICC of 0.98 (Crouter et al., 2003). Johnson (2008) used pedometers as an objective measure of physical activity and reports a similar ICC of 0.96. Overall, the results support pedometers as being accurate for individuals with ID.

Two studies that address the accuracy of pedometers for individuals relative to the effects of gait characteristics for individuals with walking related disabilities are Manns and Orchard (2006) and Manns, Orchard and Warren (2007). The first study (Manns & Orchard, 2006), presented as an ACSM abstract, included a sample of 46 adults (18 females, 28 males) with neurological conditions (stroke, multiple sclerosis, cerebral palsy, Parkinson disease, acquired brain injury) walked at a self-paced speed for 100m with a Yamax Digiwalker SW-200 pedometer placed on each side. The percent accuracy was calculated and stratified by walking speed and gait variability. The results indicate that the mean percent accuracy for the group was 89%, which is low, but potentially acceptable. Bassett et al. (1996) deemed pedometers with error scores within 11% as accurate, while Schneider et al. (2003) defined pedometers with error scores within 20% to have moderate accuracy. More importantly, 42% of the variance in error scores can be accounted for by gait speed alone and 49% from gait variability and gait speed together.

The second study by Manns, Orchard and Warren (2007) also investigated the accuracy of pedometers for forty five (18 female, 27 male) ambulatory adults with neurological disorders. The results of the study indicate that step length variability and gait speed accounted for 8% and 41% of the variance in error scores respectively. As a whole, the Yamax Digiwalker SW-200 demonstrated an average error rate of -

11.2%, meaning the pedometer underestimated steps. This is much higher than the reported error rates for the SW-200 in the general population which is typically 3% error or less (Bassett et al., 1996; Crouter et al., 2003; Schneider et al., 2003). The reliability of error scores between sides was high, ICC = 0.87 and for the various subgroups (stroke, multiple sclerosis, acquired brain injury, etc.) the interclass correlations ranged from ICC = 0.78 to 0.99, indicating that asymmetrical gait is not a concern.

Another study on the subject examined the accuracy of pedometers for youth with disabilities (Beets, Combs, Pitetti, Morgan, Bryan & Foley, 2007). A group of 18 elementary and middle school youth (11 female, 7 male) with disabilities, many with intellectual disabilities completed six self-paced walking trials with five Walk4Life pedometers attached at various locations. The results of the study do show that this pedometer has a moderate level of accuracy, with an ICC of 0.86 for the front right position; however other positions had significantly more error as demonstrated by an ICC ranging from 0.46 to 0.69. The common cutoff for acceptable accuracy based on intraclass reliability coefficients is 0.80 (Crouter et al., 2003; Schneider et al., 2003). Possibly the most interesting aspect of this study is the detail on an outlier in the data. A female with a very high BMI and very low walking speed had percent error levels that were deemed unacceptable. Interestingly, the participants walking speed was not the slowest of the group, thus indicating that other variables, potentially pedometer tilt angle, may be more influential. Finally, the authors attribute the higher percent error at the front left placement to variability within manufactured products. Furthermore, the percent error for activity time was not as great as for steps at that location. An

alternative view could be that the side of the body is relevant, and the criterion in the pedometer to record activity time is low enough to still record adequate activity time even when steps are not being accurately recorded.

A similar study examined the accuracy of voice-announcement pedometers for youth with visual impairment (Beets, Foley, Tindall & Lieberman, 2007). In this study 35 youth (13 girls, 22 boys) with visual impairments completed four 100m trials while wearing three voice announcement pedometers (Sportline, Centrios, Talking) and an New Lifestyles NL-2000. The particularly interesting aspect of this study is that the analysis using intraclass correlation coefficients on the steps recorded by pedometers on each side showed very low correlation between the two sides. All three pedometers demonstrated an ICC lower than 0.79, indicating low agreement and show that recorded steps by sides are considerably different. The NL-2000 was unaffected by location. These differences were attempted to be explained further by examining differences between participants that walked with assistance versus non-assisted. Interestingly, the Sportline and Centrios pedometers had greater absolute percent error scores for assisted than unassisted (11.5% vs. 7.3% and 8.5% vs. 4.5%) while the Talking brand showed the opposite trend (9.6% vs. 13.5%). What this study does expose is that since the NL-2000 (piezoelectric) was unaffected by side, there may be some variable of gait that affects the three VA pedometers (all mechanical lever arm).

This research has generally shown that pedometers can be used in populations with disabilities, unless there are specific gait abnormalities resulting in side preference or consistently slow walking speeds. A major issue with the pedometer validation studies in disability populations that have been done is that specific

disabilities (except visual impairment) have not been examined independently. All of the other studies have grouped participants together, albeit by similar disability classifications, however it is possible that individual disability effects on pedometer accuracy have been lost in the process.

Gait Analyses in Down Syndrome

Individuals with Down syndrome have historically been characterized as having an abnormal gait. Anecdotally, this gait has been described as "wobbling", "waddling" or "clumsy" (Buzzi & Ulrich, 2004; Latash, 2000). In laymen's terms this gait pattern can be described as a shuffling movement consisting of wider base and increased side to side movement. A posited explanation for this gait pattern has been that muscle stiffness and/or co-contractions of muscles are utilized to overcome hypotonia and joint laxity, two conditions commonly associated with DS (Buzzi & Ulrich, 2004; Kubo and Ulrich, 2004; Latash, 2000; Smith & Ulrich, 2008; Ulrich, Haehl, Buzzi, Kubo & Holt, 2004). Given these common conditions in muscular and joint function, there are a number of gait related variables they may effect the accuracy of pedometers in this population.

A study by Ulrich et al (2004) examined a number of gait parameters in addition to leg stiffness, angular impulse, and segmental range of motion (ROM) during overground and treadmill walking in preadolescents with and without DS. While the results indicate that in the sagittal-plane gait kinematics are not significantly different for youth with and without DS, Ulrich et al suggested that children with DS are significantly different on gait speed, step length and step width than their typically

developing peers. Children with DS walk at a slower speed with shorter step length and wider step width. However, when these three variables were examined using dimensionless values, which controls for leg length, only step width remained significantly different. Additionally, during treadmill walking children with DS had significantly higher stride frequencies due to shorter stride lengths. These results could indicate a fundamental difference in gait pattern, but could also be the result of inexperience on a treadmill. Stiffness and angular impulse were also significantly higher for the DS group than their typically developing peers, and increased with speed. Finally, ROM in the three segments of the leg was significantly lower in DS youth than in typically developing youth. Given the mechanism of most pedometers, a lack of range of motion, particularly in the thigh segment may result in underestimation of steps. It could also explain why pedometer accuracy increases with speed, as movements become more pronounced.

Buzzi and Ulrich (2004) also found that there are significantly higher fluctuations within each gait cycle and lower dynamic stability across the gait pattern for youth with DS than typically developing (TD) youth. This indicates that there is greater irregularity in the gait cycle for children with DS which may create a greater potential for error in pedometer measurement.

These gait parameters may also influence the direction of movement differences between populations. An analysis of the anterior-posterior and medio-lateral movements of the pelvis and head, arms and trunk (HAT) further explained these tendencies by showing that differences predominantly occur in the medio-lateral direction (Kubo & Ulrich, 2006). Specifically, amplitude of the pelvis in the medio-

lateral direction for participants with DS was larger than their typically developing peers, most likely due from a larger step width.

However some of these differences have been explained by treadmill practice. Smith, Kubo, Black, Holt & Ulrich (2007) investigated how stiffness and angular impulse change over time with treadmill walking practice. This study indicates that while initially stiffness and angular impulse for DS may be higher than TD, over practice these variables decrease and become more similar to TD. These variables were shown to decrease with practice across both groups, with the DS group following the practice intervention, demonstrating stiffness and angular impulse levels similar to the TD group at pre-test. Although stiffness and angular impulse subsided with practice, the DS group still displayed unique gait characteristics. The results show that preadolescents with DS have a wider step width when walking overground and on a treadmill, even when controlling for leg length. In addition, the DS group had a shorter stride length and higher stride frequency than their TD peers.

Similar results have been replicated in older adults. Smith and Ulrich (2008) found in older adults with DS that these individuals walk with a slower gait speed, shorter step length and wider step width. Furthermore, these variables in addition to spending more time in stance and double support position are to overcome postural instabilities and are exemplified by potential perturbations in walking cycle.

Agiovlasitis (2007) examined the center of mass (COM) of adults (19-44 years old) with and without Down syndrome across walking speeds and found that while individuals with DS had greater range of medio-lateral COM, greater variability between strides in medio-lateral and vertical COM, and greater step width variability.

There was no significant difference on step width between groups. Furthermore, the results indicate that adults with DS walked with a shorter stride length only at the slowest speed and with a shorter step time, indicating a faster cadence than adults without DS. These results are similar to those reported by Smith et al. (2007) but are contrary to other studies (Kubo & Ulrich, 2006; Smith & Ulrich, 2008; Ulrich et al., 2004) that have shown a slower gait speed and wider step width, but could reflect a difference in operationalized definitions.

While the majority of research has focused on the COM and segmental ROM of the upper leg, similar results have been seen at the ankle. Ciono, Cocilovo, Rossi, Paci, and Valle (2001) found that the kinetics of the ankle movement during walking gait is dysfunctional as seen by a reduced energy absorption stage, a reduction in plantar-flexor moment at higher velocities and a longer phase of energy generation at lower velocities. Essentially, the ankle kinetics creates a more stable base for the contralateral foot. The authors suggest that individuals with DS overcome this through increased hip flexor power or hip extensor power. Either solution would explain increased COM variable in the medio-lateral direction discussed.

There clearly are abnormalities in the gait pattern of individuals with DS. One problem is that the Smith and Ulrich (2008), Agiovlasitis (2007) and Cioni et al. (2001) are the only studies to empirically address the DS gait patterns in individuals over age 10. Smith and Ulrich (2008) involved older adults while the Agiovlasitis (2007) and Cioni et al. (2001) study samples ranged from age 19 to 44 and 8 to 36 respectively. There is still little known about the gait patterns of adults with DS;

however it is clear that there is a gait abnormality, particularly related to increased variability in the medio-lateral direction.

Walking Activity: Down Syndrome and Intellectual Disabilities

As previously mentioned, pedometers are purported as a measure of physical activity; however the direct measure of a pedometer is walking activity. There has been limited research conducted on the walking activity patterns of individuals with intellectual disabilities with and without Down syndrome. Within this limited evidence there are conflicting results as to whether individuals with Down syndrome have lower levels of walking activity than individuals with intellectual disabilities without Down syndrome (IDwoDS) and if individuals with intellectual disabilities (ID) meet the guidelines for daily walking activity.

For instance, Stanish (2004) examined the differences in walking activity among 20 individuals with intellectual disabilities (9 with DS, 11 IDwoDS) using the guideline of 10,000 steps per day as the recommendation of daily walking activity. To date, it is the only study known to the author to analyze this behavior between the two segmented groups. Results showed that the number of steps taken per day and the average distance walked per day were significantly lower in the DS group. Average steps were analyzed by condition and gender. Women and men with DS in the sample walked an average of 8815.6 ± 4094.1 and 5449.8 ± 2316.3 steps respectively. Women and men without DS in the sample walked an average of 11809.4 ± 4652.4 and 11885 ± 5645.9 steps respectively. While there were no significant differences between genders, there was a significant difference between DS and IDwoDS groups

demonstrating that individuals with DS accumulate significantly fewer steps. The results also show that the group without DS met the recommended guidelines for physical activity, however it should be noted that this study was conducted in a rural community without public transportation, so walking as a form of personal transportation may be higher than other communities with a small sample size resulting in limited generalizability.

These results are further drawn into question due to the results of the follow up study (Stanish & Draheim, 2005a, 2005b, 2007). This study included 19 adults with DS (9 females, 10 males) and 84 adults with IDwoDS (29 females, 55 males) and used pedometers and the NHANES III Physical Activity Survey. Preliminary analysis on the sample found no significant differences between gender and condition classification, so weekly steps and physical activity levels were analyzed as a total sample. In respect to guidelines for physical and walking activity, the results indicate that only 21.4% accumulated an average of 10,000 per day and 17.5% reported engaging in 30 minutes of MVPA at least 5 days per week. The average steps per day are reported as approximately 7,700. While the follow-up study did not find a significant difference between groups, it also showed that most individuals with ID do not meet recommended guidelines for walking and physical activity, however the steps accumulated by this sample with intellectual disabilities is similar to the steps per week reported in the non-disabled population (Tudor-Locke & Bassett, 2004).

An additional study, Peterson, Janz and Lowe (2008) examined physical activity behaviors of adults with intellectual disabilities in community living settings. This study also found no significant differences gender or DS groups, but did find

significant differences based on level of intellectual disability and age. The primary results of this study is the average walking activity of participants was 6508 ± 3296 steps/day. Only 14.1% of the sample demonstrated an average of 10,000 steps/day. Furthermore, it was determined that individuals with mild ID walked more than individuals with moderate ID, and that more activity took place on weekdays than weekends and during the morning and afternoon than in the evening. These findings point out that weekends and evenings are consistently periods of inactive leisure.

These five published articles represent a paucity of information on the walking activity of individuals with intellectual disabilities, and particularly Down syndrome. Since there have only been two formal studies conducted on this subject (the three later articles are all from the same data) the results must be considered inconclusive and further research is required.

Physical Activity: Down Syndrome and Intellectual Disabilities

Due to the lack of research specifically on walking activity using pedometers in these populations, further evidence must be derived from physical activity research. There has been very little research in this area that specifically examines individuals with DS. While the focus has predominantly been on individuals with intellectual disabilities, it is within reason to assume that the samples of adults and children with ID included those with DS. Also, given the previous research that has demonstrated that individuals with DS engage in similar or less physical activity than individuals with ID, it is appropriate to take into consideration on physical activity patterns of adults and youth with intellectual disabilities.

Adults. Temple, Anderson and Walkley (2000) investigated physical activity in six individuals with intellectual disabilities using direct observation and accelerometers. Over a period of seven consecutive days, only one individual engaged in more than 30 minutes of moderate-vigorous physical activity. Another individual met the guideline of more than 30 minutes of MVPA on the days the individual was not ill. Interestingly, three participants that demonstrated engagement in minimal minutes of MVPA, walked for an average over an hour per day for personal transport. This indicates that while the time spent walking would more than meet the PA guidelines, the walking speed employed is not sufficient to gain health benefits. While this study demonstrates that most of the individuals studied are not adequately physically active, the results indicate that individuals with intellectual disabilities may not be as physically inactive as previous studies had indicated.

A follow-up to this study by Temple and Walkley (2003) utilized accelerometers and proxy reports completed by direct care staff to measure the physical activity and estimates of energy expenditure of 37 adults with mild to moderate MR living in group home facilities. Only 32% of the individuals met the guidelines of 30 minutes of moderate-vigorous physical activity, as compared to 57% of the general community (Armstrong, Bauman & Davies, 2000).

In 2004, Frey conducted a study to directly compare physical activity levels between adults with mental retardation, and active and sedentary controls (adults without MR). All participants wore accelerometers for seven consecutive days. Results indicate that 89% of active controls, 47% of sedentary control and 28% of adults with MR met the guideline of 30 minutes of moderate PA per day. However,

individuals with mental retardation and sedentary controls were not significantly different and on average did not meet recommended guidelines. This suggests that despite a large difference in the percentage of participants that met the PA guidelines, the PA of adults with MR and sedentary adults without MR are similar.

Youth. Not unlike the adult population, there are varying results on the differences in physical activity levels between children with intellectual disabilities and those without. Much of this research has taken place in school settings, but has also included time outside of school. The general consensus is the children with intellectual disabilities engage in less physical activity than their non-disabled peers however results are largely inconclusive (Frey, Stanish & Temple, 2008).

Foley, Bryan and McCubbin (2008) examined the physical activity levels of elementary school children with and without mental retardation using accelerometers. The results indicate that students with MR were significantly less active during recess and inclusive physical education than their peers without MR. The significantly lower physical activity levels were also demonstrated after-school and on weekends.

These results are consistent with other findings in inclusive physical education settings. Temple and Walkley (1999) found that in physical education settings, students with mild intellectual disabilities were active for the same amount of time as their non-disabled peers, but were 40% less engaged at a motor appropriate level. During the course of this study this equated to approximately 15% of physical education time (as little as 5 min.) where students with mild intellectual disabilities were active and appropriately motor engaged. This demonstrates that even when

students are physically active for similar amounts of time as their peers, they may not be engaged to same level.

However, the results of Foley et al. (2008) contradict the findings of Lorenzi, Horvat, and Pellegrini (1999). In this study, the physical activity of children with and without MR was examined during inclusive recess. The results actually show that children with MR were more active than peers based on heart rate and accelerometers; however these results were not seen in direct observation. Additionally, Horvat and Franklin (2001) found no significant differences between inclusive and non-inclusive recess settings for children with MR, although accelerometer activity counts were higher in the non-inclusive setting.

Furthermore, Faison-Hodge and Porretta (2004) found that students with mild MR were similar in MVPA to same age peers with low cardiorespiratory fitness during recess and physical education. Both students with MR and students with LCRF were significantly less active during both of these settings than students that demonstrated high levels of fitness. The study also demonstrated that all students were significantly more active during recess settings than physical education. These results are also consistent with Luke et al. (1994) that showed that there were no differences in daily physical activity, using doubly-labeled water, between children with Down syndrome and control subjects.

Few studies have objectively measured physical activity patterns in children with Down syndrome. Whitt-Glover, O'Neill and Stettler (2006), compared children with DS and their "unaffected" siblings' physical activity levels using accelerometers over a seven day period. The results indicated that the children with DS spent

significantly less time in vigorous physical activity and had short bouts of VPA; however there were not significant difference between groups of moderate PA, low-intensity PA and inactivity. In fact, the average minutes of moderate physical activity for each group was 153.1 minutes and 154.6 minutes. The average vigorous PA for each group was 49.5 and 68.6 minutes. This indicates that both groups met the daily guidelines for moderate-vigorous activity, and in some cases more than doubled it. The study demonstrates that children with DS are capable of achieving adequate amounts of MPA, but may need to engage in more minutes and longer bouts of VPA to experience health benefits. The other study to compare physical activity in children with Down syndrome and there siblings without DS, Sharav and Bowman (1992), however, found children with DS engaged in significantly less physical activity.

Generalizability Theory

Generalizability (G) theory is framework that enables the total variance of a model to be partitioned and subsequently, the contribution of each potential source of error or source interactions to be estimated (Morrow, 1989; Shavelson & Webb, 1991). In other words, "G theory attempts to identify and estimate the magnitude of the potentially important sources of error in a measurement." (p. 923; Shavelson, Webb, & Rowley, 1989). This enables researchers to determine the percentage of error associated with each potential source and to determine the dependability of scores for both relative and absolute interpretations (Shavelson & Webb, 1991).

Within G theory there are generalizability (G) studies and decision (D) studies. A G study is used to determine which sources of variance contribute most to

measurement error (Morrow, 1989; Shavelson, Webb, & Rowley, 1989; Shavelson & Webb, 1991). The steps for conducting a G study, as set forth by Morrow (1989) include (1) selecting the facets and design, (2) estimating the variance components through the ANOVA model, (3) calculating the expected mean squares values for each source of variation, (4) calculating the mean square for each source, (5) calculating the variance components for each source, (6) determining the percentage of variance associated with each source.

A D study uses the information obtained through the G study, most notably the G and phi coefficients, to make decisions about the measurement protocol or to design measurement error for a particular purpose (Morrow, 1989; Shavelson, Webb, & Rowley, 1989). According to Shavelson and Webb (1991) the planning of a D study includes (1) the defining of the universe of generalization, described as the facets that will generalized over, (2) specifying the intention of the interpretation; this includes relative decisions, the standing of individuals compared to others, and absolute decisions, individuals addressed based on absolute performance, (3) using the magnitude of variability from the G study to evaluate the effectiveness of alternative designs. This final step is achieved through calculating the G coefficients for the universes of interest and determining the best protocol by using the percentages of variance, G coefficients, indices of reliability and standard errors of measurement (Morrow, 1989).

A G theory study can be set up in a variety of designs depending on the facets selected and the nature of the model. These variations are described in Morrow (1989). First, a facet is dimension from which measurements can be taken (times, raters) and

can be random or fixed. A random facet is a dimension that has multiple levels and is considered a representative of that facet. A fixed facet has only one level and cannot be generalized beyond that level. There are also a number of models that can be used in G theory. A crossed model entails that all of the facets are random and all subjects are crossed with all other facets. A mixed model is when some facets are random and others are fixed. A nested model is when some facets are fixed within other facets.

Generalizability Theory in Exercise Science

Since the development of Generalizability (G) theory, it has been used, albeit sparingly, in the field of exercise science. This is particularly useful in physical activity research because there are numerous sources of variability in physical activity participation and in the way it is measured. These studies have predominately used G-theory to determine the reliability of an instrument or test, or to determine the number of observations needed to measure physical activity.

Many G-theory studies in this field have been used for estimating the reliability of various tests and instruments. Stamm and Moore (1980) used G-theory to assess the reliability of motor performance test that used first-ball scores to measure performance of beginning college bowlers utilizing a three facet (2 sexes, 10 trials, 10 days) design. The results indicate that trials and days accounted for 0% and 0.2% of the total variance respectively, however 90% of the variance was accounted for by the residual. Since the G-coefficients were all greater than 0.80, at 0.93, 0.92, and 0.84 respectively, this first-ball motor performance test can be considered reliable when generalized over sex, trials, days and when estimated within sex but generalized over

trials and days, but clearly the model is missing an important source of error or the four-way interaction of these facets is responsible for a great deal of error.

Ulrich and Wise (1984) used a two facet (20 rater, 2 occasions) G-theory design to investigate the consistency of scores from the Objectives-Based Motor Skill Assessment Instrument (OBMSAI, would later become the TGMD) across raters and times. Results indicated that differences between participants on 11 skills (out of 12) accounted for 92% of the variance. Interestingly, the rater-by-subject interaction accounted for the majority of the variance (26%) for the run skill indicating that on this skill, raters did not have effectively used the same criteria across participants. Finally, G-coefficients for all skills were above 0.80 except for the run and the two D-studies indicated there were similar G-coefficients when 10 or 20 raters are used.

Morrow, Fridye, and Monaghen (1986) used G-theory to estimate the variance in skinfold measurement as part of AAHPERD Health Related Physical Fitness Test. The study utilized a three-facet (3 testers, 3 instruments, 3 trials) design to determine the reliability across testers performing the skinfold measurement. The results found that less than 4% of the total variance was associated with the tester. On the other hand, almost 15% of the variance was associated with the caliper instrument used. This indicates that while variance between testers is relatively low, there is more error related to the specific caliper device used. However, all of the G coefficients were greater than 0.91 indicating a high level of reliability.

Similarly, Yun, Garcelon, and Ulrich (1997) utilized a three facet (3 raters, 3 times, 2 occasions) crossed design for novice and experienced raters on skinfold measurements. The G-coefficients for novice raters were all very low, and deemed

unacceptable, at 0.64, 0.76, and 0.87 for the triceps, suprailliac, and thigh respectively. G-coefficients for experienced raters were 0.76 and 0.87 for the suprailliac and thigh respectively. The triceps G-coefficient for experienced raters of 0.64 was also deemed unacceptable. Although more detailed results were not included in the published abstract, these coefficients would indicate that a high amount of variance could be associated with the rater, particularly for those with little experience and at the triceps.

Taylor and Yun (2006) also used G-theory to determine the sources of measurement error for the System for Observing Fitness Instruction Time (SOFIT) and the Children's Activity Rating Scale (CARS). The study utilized a two facet (3 raters, 2 times) design for both the SOFIT and CARS instruments. The G-coefficient for the SOFIT was 0.98 indicating high reliability. Ninety four percent of the variance was associated with differences between participants, which is to be expected. Error variances due to trial, rater and the trial-by-rater interaction account for 0.30%, 0.67%, and 0.44% respectively. The G-coefficient for CARS was 0.75 indicating moderate reliability. For this instrument, 49.65% was associated with participants, while error due to raters and the participant-by-rater interaction accounted for 31.49% and 15.41%. This means that there were systematic differences between raters using the CARS.

Finally, and most notable to the current study, Welk, Schaben, and Morrow (2004) used G-theory to assess the reliability of four types of pedometers. The study employed a two-facet (4 monitors, 3 trials) crossed design with the intention of determining sources of error in measuring activity bouts. Possibly the most important result of this study is the variance associated with variability across monitoring units, with 0.9% (CSA/MTI), 9.4% (Biotrainer Pro), 9.6% (Actical), and 11.6% (Tritrac) for

each model. This shows a range of inter-unit errors within brands and represents differing manufacturer quality. The trial-by-subject interaction accounted for 14.4% (Tritrac), 14.5% (CSA/MTI), 19.6 (Actical), and 21% (Biotrainer) of the variance. Finally, the monitors-by-trial-by-subject interaction accounted for 17.8% (Biotrainer), 20.1% (CSA/MTI), 22.5 (Tritrac), and 23.7 (Actical) of the variance. Both of these results indicate that while there is some variation across trials and monitors, these differences vary across participants. While this study does show there are differences between accelerometer brands, all of the G-coefficients (from 0.432 to 0.640) are far less than the acceptable level of 0.80.

The other common type of G-theory studies have used D-studies (many of the above studies did as well) to determine the number of observations, scores or events needed to measured in order for an instrument to be reliable. First, Crocker, Bailey, Faulkner, Kowalski, and McGrath (1997) conducted a G-theory study to determine the number of scores from the Physical Activity Questionnaire for Older Children over the course of the year was needed to obtain a reliable estimate of a child's physical activity. The study employed a one facet (time) design, although the authors report it as being two-facet (person by time). The results indicate that participants accounted for 67.8% of the variance and the person-by-time interaction and time component accounted for 27.5% and 4.7% of the variance respectively. Furthermore, the G-coefficient when using three scores (times) was 0.88 and 0.83 when only using two. This indicates that using two scores instead of three would still result in acceptable reliability. These results were also confirmed when the sample was split into older and younger groups.

Second, Coleman and Epstein (1998) conducted a similar study, but used G-theory to determine the number of days of accelerometry and self report diary are needed to measure physical activity in men who are not regularly active. The results indicate that when using accelerometer vector magnitude data, 4 days are necessary (G-coefficient=0.82) while 6 days (0.81) is needed for using accelerometer MET values. For self report diary MET values, 8 days (0.80) are needed when participants are compliant, although this still includes 65% residual variance. This study utilized G-theory to determine the number of days needed to measure physical activity in this population and determined the accelerometers require less days than self report diaries.

Third, Kim and Yun (2009) used G-theory to determine the number of days needed for measuring physical activity of youth with developmental disabilities using accelerometers and pedometers. The study employed a two-facet (instrument by day) design independently for pedometers and accelerometers. From this G-study design, the largest sources of variances were due to persons, days and the person by day interaction. There was very little evidence of systematic error due to either instrument. Since the study sought to determine the number of days needed to measure PA using weekday, weekend, or both types of days, there were numerous reliability coefficients calculated. The study found the when using pedometers, four weekdays, six weekends, and/or eight weekday and weekend days were needed to measure PA. When using accelerometers, four days of measurement were needed for all three of the time periods. These results provide reliability evidence for piezoelectric pedometers and accelerometers for use among youth with developmental disability and also provide information on the number of days needed to adequately measure physical activity.

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APPENDIX B

IRB Approval and Informed Consent Documents

OSU Oregon State University

Institutional Review Board • Office of Research Integrity
Oregon State University, 312 Kerr Administration Building, Corvallis, Oregon 97331-2140
Tel 541-737-4933 | Fax 541-737-3093 | http://oregonstate.edu/research/ori/humans/subjects.htm
IRB@oregonstate.edu

Amended to Add Study Number

TO: Joonkoo Yun
Nutrition and Exercise Sciences

IRB #4171: The Accuracy of Pedometers for Adults with Down Syndrome in Controlled and Free-Living Conditions (Student Researcher: E. Andrew Pitchford)

Level of Review: Expedited

Expiration Date: 1-12-10

Approved Number of Participants: 80

The referenced project was reviewed under the guidelines of Oregon State University's Institutional Review Board (IRB). The IRB has **approved** the:

(X) Initial Application ( ) Continuing Review ( ) Project Revision
with a (if applicable): ( ) Waiver of documentation of Informed Consent ( ) Waiver of Consent

A copy of this information will be provided to the full IRB committee.

- **CONSENT FORM:** All participants must receive the IRB-stamped informed consent document. If the consent is in a format that could not have stamp placement (i.e. web site language, email language, etc), then the language must be **exactly** as the IRB approved it.
- **PROJECT REVISION REQUEST:** Any changes to the approved protocol (e.g. protocol, informed consent form(s), testing instrument(s), research staff, recruitment material, or increase in the number of participants) must be submitted for approval before implementation.
- **ADVERSE EVENTS:** Must be reported within three days of occurrence. This includes any outcome that is not expected, routine and that result in bodily injury and/or psychological, emotional, or physical harm or stress.
- **CONTINUING REVIEW:** A courtesy notice will be sent to remind researchers to complete the continuing review form to renew this project, however – it is the researcher's responsibility to ensure that continuing review occurs prior to the expiration date. Material must be submitted with adequate time for the office to process paperwork. If there is a lapse in approval, suspension of all activity including data analysis, will occur.
- **DEVIATION/EXCEPTIONS:** Any departure from the approved protocol must be reported within 10 business days of occurrence or when discovered.

Forms are available at: http://oregonstate.edu/research/osprc/rc/humansubjects.htm.

If you have any questions, please contact the IRB Human Protections Administrator at IRB@oregonstate.edu or by phone at (541) 737-8008.

Elisa Espinoza

Date: 1-13-09

Elisa Espinoza
IRB Human Protections Administrator

OSU Oregon State University

Nutrition and Exercise Sciences
Oregon State University, 101 Milam Hall, Corvallis, Oregon 97331
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INFORMED CONSENT DOCUMENT

Project Title: **The Accuracy of Pedometers for Adults with Down Syndrome in Controlled and Free-Living Conditions**
Principal Investigator: **Joonkoo Yun**, Department of Nutrition and Exercise Sciences
Student Investigator: **E. Andrew Pitchford**, Department of Nutrition and Exercise Sciences

WHAT IS THE PURPOSE OF THIS STUDY?

You are being invited to take part in a research study designed to determine the accuracy of pedometers for adults with and without Down syndrome. A pedometer is a small instrument that measures and records the number of steps taken. The study will investigate the difference between the steps measured by a pedometer and the actual number of steps walked for two different types of pedometers under controlled and free-living conditions. The intended outcome of the study is to determine if pedometers are accurate at an acceptable level for adults with Down syndrome for future use in research and physical activity programs. We believe that both types of pedometers will be accurate for adults without disabilities, but may not be accurate for adults with Down syndrome. The results of this study will be used as part of the Master's thesis of the student investigator. It is also intended for professional presentation and to be published in a research journal.

We are studying this because walking is the most common form of physical activity for adults with intellectual impairments. While pedometers have been examined for adults without disabilities, there is little evidence of the accuracy of pedometers for individuals with disabilities, particularly Down syndrome. There is reason to believe that pedometers may not work properly for these individuals. It is important to understand if this instrument can be used to measure walking activity as to better understand physical activity patterns.

WHAT IS THE PURPOSE OF THIS FORM?

This consent form gives you the information you will need to help you decide whether to be in the study or not. Please read the form carefully. You may ask any questions about the research, the possible risks and benefits, your rights as a volunteer, and anything else that is not clear. When all of your questions have been answered, you can decide if you want to be in this study or not.

WHY AM I BEING INVITED TO TAKE PART IN THIS STUDY?

You are being invited to take part in this study because you are an adult between the ages of 18 and 64. You are ambulatory, which means that you are able to walk. In order to participate in this study you must be able to walk for approximately 30 minutes.

Criteria for Participation

You are being invited to participate in this study because you meet the following criteria:

1. Age between 18 and 64 years old — YES _____ NO _____
2. Are ambulatory, with the ability to walk for 30 minutes — YES _____ NO _____
3. I do not have any intellectual disability — YES _____ NO _____

WHAT WILL HAPPEN DURING THIS STUDY AND HOW LONG WILL IT TAKE?

All assessment and study procedures will take place at an open space such as a gymnasium in your convenient location. The researcher will meet with you and will discuss this document. If you elect to participate in the study, you will sign and date this form. The researcher will then have you complete a short demographic questionnaire that will include items related to your age, gender, and a diagnosis of Down syndrome. This portion of the study should take approximately 10 minutes. Following the paperwork, the research study will involve three different portions.

Body Composition Measurements
The researcher will conduct five separate measures of body composition. First, height and weight will be measured with you dressed in lightweight clothing. This information will be used to calculate your Body Mass Index (BMI). Second, the researcher will measure your hip circumference and waist circumference. This information will be used to calculate your hip-to-waist ratio. Finally, your belt circumference will be measured. All of these body composition measurements will be repeated for a total of three times to increase accuracy. In total, these measurements should take approximately 15 minutes.

Controlled Conditions Walking Trials
Following the body composition measurements, you will have a total of 12 pedometers placed on your body. Four will be placed at the waist on each side of the body and 2 will be placed in each of your front pockets. There will be a 100 meter course set up in the gym space in the pattern of a figure 8. You will be asked to walk this course for a period of 2 minutes on 3 separate trials. On the first trial you will walk at your own self-directed pace. During this trial, a researcher will count the number of steps you take and measure the speed at which you are walking. In the second and third trial, you will be paced by a researcher at either a faster or slower walking speed. During these trials, the number of steps taken will also be counted. These trials should take between 15 and 30 minutes to complete.

Free-Living Walking Trial
Following the controlled conditions walking trial, you will be asked to go for a walk for a period of 20 minutes. You will wear the same pedometers as the first three walking trials. During this 20 minute walk you will be free to walk anywhere that you see fit. A researcher will walk with you. Although you may walk anywhere that you would like, if the researcher asks you must comply. This researcher will only intervene if your intended walking path is potentially unsafe. This trial should take a total of 25 to 30 minutes to complete.

If you agree to take part in this study, your involvement will last for no more than a total of 90 minutes.

WHAT ARE THE RISKS OF THIS STUDY?

There are minimal risks for participating in this study. You have the right to take breaks between the 2 minute trials and to walk for as much of the 20 minutes as you feel comfortable. You also have the right to take a break, stop the trial, or withdraw from the study at any time. The potential for risk or discomfort in the study is no greater than that of normal, every day movement. In the unlikely event of research related injury, compensation and medical treatment is not provided by Oregon State University or the researchers of this study.

WHAT ARE THE BENEFITS OF THIS STUDY?

You will not benefit from being in this study. However, we hope that, in the future, other people might benefit from this study because by having better information on the accuracy of pedometers, researchers will be able to more effectively understand physical activity patterns and create more and better interventions and programs.

WILL I BE PAID FOR PARTICIPATING?

You will be compensated for being in this research study. By participating in this study, you will receive at $5 gift card to Fred Meyers or Target. All compensation will be in the form of gift cards. No direct monetary compensation will be given.

WHO WILL SEE THE INFORMATION I GIVE?

The information you provide during this research study will be kept confidential to the extent permitted by law. To help protect your confidentiality, we will utilize a coding system to identify you throughout this research study. All forms used including the background questionnaire, data recording forms, and lab logs will use this identification code. The only form that your name and personal information will appear on is this consent document. This consent document will be stored in a locked lab room and will only be accessible and viewed by the researchers of this study.

If the results of this project are published your identity will not be made public.

DO I HAVE A CHOICE TO BE IN THE STUDY?

If you decide to take part in the study, it should be because you really want to volunteer. You will not lose any benefits or rights you would normally have if you choose not to volunteer. You can stop at any time during the study and still keep the benefits and rights you had before volunteering. If you decide not to take part in this study, your decision will have no effect on your and any of your dependents participation or ability to receive services through the organizations through which you have been recruited.

You will not be treated differently if you decide to stop taking part in the study. On the background questionnaire, you are free to skip any items that you do not want to answer. You are also free to stop the walking trials or withdraw from the study at any time. If you choose to withdraw from this project before it ends, the researchers may keep information collected about you and this information may be included in study reports.

WHAT IF I HAVE QUESTIONS?

If you have any questions about this research project, please contact Andrew Pitchford by email at pitchfoe@onid.orst.edu or by phone at (541) 737-5927. You may also contact Dr. Joonkoo Yun by email at jk.yun@oregonstate.edu or by phone at (541) 737-8584.

If you have questions about your rights as a participant, please contact the Oregon State University Institutional Review Board (IRB) Human Protections Administrator, at (541) 737-4933 or by email at IRB@oregonstate.edu.

Oregon State University • IRB Study #:4171 Approval Date: 01/13/09 Expiration Date: 01/12/10

Your signature indicates that this research study has been explained to you, that your questions have been answered, and that you agree to take part in this study. You will receive a copy of this form.

Participant's Name (printed): _____

_____ _____
(Signature of Participant) (Date)

_____ _____
(Signature of Researcher) (Date)

Oregon State University • IRB Study #:4171 Approval Date: 01/13/09 Expiration Date: 01/12/10

Nutrition and Exercise Sciences
Oregon State University, 101 Milam Hall, Corvallis, Oregon 97331
Tel 541-737-2843 | Fax 541-737-6914 | http://www.hhs.oregonstate.edu/research/index.html

INFORMED CONSENT DOCUMENT

Project Title: **The Accuracy of Pedometers for Adults with Down Syndrome in Controlled and Free-Living Conditions**
Principal Investigator: **Joonkoo Yun**, Department of Nutrition and Exercise Sciences
Student Investigator: **E. Andrew Pitchford**, Department of Nutrition and Exercise Sciences

We are doing a research study. We are trying to learn about pedometers. Pedometers are tools that count the number of steps you take. We want to see if pedometers will work right for people like you.

This form is about the study, so you can learn about the study and decide if you want to be in the study or not. You can ask questions about the study. After all of your questions have been answered you can decide if you would like to be in this study or not. Please read carefully.

I am being asked to be in this study because:

I am between 18 and 64 years old.	YES _____ NO _____
I am able to walk. I can walk for at least 30 minutes.	YES _____ NO _____
I have Down syndrome.	YES _____ NO _____

I understand the following:

1. I am being asked to participate in a study.
2. The researcher wants to find out if pedometers will count the number of steps that I take.
3. If I need help answering questions, my support staff, parent or caregiver can help me.
4. I cannot participate in this study if I am unable to walk.
5. I will be asked to walk for a total of 30 minutes.
6. If I get tired, I can take a break.
7. I will need to wear 8 pedometers on my waist and 4 in my front pants pockets.
8. During the study, I will have my weight, height and waist measurements taken.
9. I will be asked to answer questions about my age, gender and condition.
10. If I am hurt during this study, Oregon State University will not pay for care.
11. My name will not be used in any part of this study.
12. If I complete the study, I will be given a $5 gift card to Fred Meyer or Target.
13. I want to take part in this study.
14. I can stop taking part in this study at any time. Nothing will happen to me if I do stop.

You do not have to be in this study. It is up to you. If you say okay now, but you want to stop later, that is okay too. All you have to do is tell us and we will stop.

If you have any questions about this research project, please contact Andrew Pitchford by email at pitchfoe@onid.orst.edu or by phone at (541) 737-5927. You may also contact Dr. Joonkoo Yun by email at jk.yun@oregonstate.edu or by phone at (541) 737-8584.

If you have questions about your rights as a participant, please contact the Oregon State University Institutional Review Board (IRB) Human Protections Administrator, at (541) 737-4933 or by email at IRB@oregonstate.edu.

If you want to be in this study, please sign your name.

Participant's Name (printed): _____

_____ _____
(Signature of Participant) (Date)

If the participant requires assistance from a parent, guardian, primary giver or authorized legal representative to complete this informed consent document, the background questionnaire or participate in the study, this individual must also sign this informed consent document.

Your signature indicates that this research study has been explained to you and the participant, that your questions have been answered, and that you agree to allow the participant to take part in this study. You will also receive a copy of this form.

_____ _____
(Name of Parent/Guardian/Primary Caregiver, or Legal Representative – printed) (Relationship to Participant - printed)

_____ _____
(Signature of Parent/Guardian/Primary Caregiver, or Legal Representative) (Date)

_____ _____
(Signature of Researcher) (Date)

Oregon State University • IRB Study #:4171 Approval Date: 01/13/09 Expiration Date: 01/12/10

APPENDIX C

SPSS Output of ANOVA results for Manuscript 1

```
GLM PaceRHD SlowRHD FastRHD PaceRHO SlowRHO FastRHO BY syndrome
  /WSFACTOR=model 2 Polynomial speed 3 Polynomial
  /METHOD=SSTYPE(3)
  /EMMEANS=TABLES(syndrome) COMPARE ADJ(BONFERRONI)
  /EMMEANS=TABLES(model) COMPARE ADJ(BONFERRONI)
  /EMMEANS=TABLES(speed) COMPARE ADJ(BONFERRONI)
  /EMMEANS=TABLES(model*speed)
  /PRINT=ETASQ
  /CRITERIA=ALPHA(.05)
  /WSDESIGN=model speed model*speed
  /DESIGN=syndrome.

[DataSet1] \\onid-fs\pitchfoe\DS THESIS DATA\DSData0509.sav
```

General Linear Model

2 x 2 x 3 RM ANOVA
Group x Model x Speed
on absolute error

Within-Subjects Factors

Measure:MEASURE_1

model	speed	Dependent Variable
1	1	PaceRHD
	2	SlowRHD
	3	FastRHD
2	1	PaceRHO
	2	SlowRHO
	3	FastRHO

Between-Subjects Factors

		Value Label	N
syndrome	.00	nonDS	24
	1.00	DS	20

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
model	Pillai's Trace	.287	16.870[a]	1.000	42.000	.000	.287
	Wilks' Lambda	.713	16.870[a]	1.000	42.000	.000	.287
	Hotelling's Trace	.402	16.870[a]	1.000	42.000	.000	.287
	Roy's Largest Root	.402	16.870[a]	1.000	42.000	.000	.287
model * syndrome	Pillai's Trace	.000	.019[a]	1.000	42.000	.890	.000
	Wilks' Lambda	1.000	.019[a]	1.000	42.000	.890	.000
	Hotelling's Trace	.000	.019[a]	1.000	42.000	.890	.000
	Roy's Largest Root	.000	.019[a]	1.000	42.000	.890	.000
speed	Pillai's Trace	.279	7.926[a]	2.000	41.000	.001	.279
	Wilks' Lambda	.721	7.926[a]	2.000	41.000	.001	.279
	Hotelling's Trace	.387	7.926[a]	2.000	41.000	.001	.279
	Roy's Largest Root	.387	7.926[a]	2.000	41.000	.001	.279
speed * syndrome	Pillai's Trace	.011	.222[a]	2.000	41.000	.802	.011
	Wilks' Lambda	.989	.222[a]	2.000	41.000	.802	.011
	Hotelling's Trace	.011	.222[a]	2.000	41.000	.802	.011
	Roy's Largest Root	.011	.222[a]	2.000	41.000	.802	.011
model * speed	Pillai's Trace	.341	10.595[a]	2.000	41.000	.000	.341
	Wilks' Lambda	.659	10.595[a]	2.000	41.000	.000	.341

	Hotelling's Trace	.517	10.595[a]	2.000	41.000	.000	.341
	Roy's Largest Root	.517	10.595[a]	2.000	41.000	.000	.341
model * speed * syndrome	Pillai's Trace	.023	.476[a]	2.000	41.000	.624	.023
	Wilks' Lambda	.977	.476[a]	2.000	41.000	.624	.023
	Hotelling's Trace	.023	.476[a]	2.000	41.000	.624	.023
	Roy's Largest Root	.023	.476[a]	2.000	41.000	.624	.023

a. Exact statistic

b. Design: Intercept + syndrome

Within Subjects Design: model + speed + model * speed

Mauchly's Test of Sphericity[b]

Measure:MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[a]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
model	1.000	.000	0	.	1.000	1.000	1.000
speed	.841	7.079	2	.029	.863	.918	.500
model * speed	.948	2.185	2	.335	.951	1.000	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b. Design: Intercept + syndrome Within Subjects Design: model + speed + model * speed

Tests of Within-Subjects Effects

Measure:MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
model	Sphericity Assumed	.335	1	.335	16.870	.000	.287
	Greenhouse-Geisser	.335	1.000	.335	16.870	.000	.287
	Huynh-Feldt	.335	1.000	.335	16.870	.000	.287
	Lower-bound	.335	1.000	.335	16.870	.000	.287
model * syndrome	Sphericity Assumed	.000	1	.000	.019	.890	.000
	Greenhouse-Geisser	.000	1.000	.000	.019	.890	.000
	Huynh-Feldt	.000	1.000	.000	.019	.890	.000
	Lower-bound	.000	1.000	.000	.019	.890	.000
Error(model)	Sphericity Assumed	.833	42	.020			
	Greenhouse-Geisser	.833	42.000	.020			
	Huynh-Feldt	.833	42.000	.020			
	Lower-bound	.833	42.000	.020			
speed	Sphericity Assumed	.212	2	.106	7.559	.001	.153
	Greenhouse-Geisser	.212	1.726	.123	7.559	.002	.153
	Huynh-Feldt	.212	1.836	.116	7.559	.001	.153
	Lower-bound	.212	1.000	.212	7.559	.009	.153

speed * syndrome	Sphericity Assumed	.009	2	.004	.316	.730	.007
	Greenhouse-Geisser	.009	1.726	.005	.316	.698	.007
	Huynh-Feldt	.009	1.836	.005	.316	.711	.007
	Lower-bound	.009	1.000	.009	.316	.577	.007
Error(speed)	Sphericity Assumed	1.180	84	.014			
	Greenhouse-Geisser	1.180	72.503	.016			
	Huynh-Feldt	1.180	77.125	.015			
	Lower-bound	1.180	42.000	.028			
model * speed	Sphericity Assumed	.163	2	.081	13.143	.000	.238
	Greenhouse-Geisser	.163	1.901	.086	13.143	.000	.238
	Huynh-Feldt	.163	2.000	.081	13.143	.000	.238
	Lower-bound	.163	1.000	.163	13.143	.001	.238
model * speed * syndrome	Sphericity Assumed	.005	2	.002	.397	.674	.009
	Greenhouse-Geisser	.005	1.901	.003	.397	.663	.009
	Huynh-Feldt	.005	2.000	.002	.397	.674	.009
	Lower-bound	.005	1.000	.005	.397	.532	.009
Error(model*speed)	Sphericity Assumed	.521	84	.006			
	Greenhouse-Geisser	.521	79.857	.007			
	Huynh-Feldt	.521	84.000	.006			
	Lower-bound	.521	42.000	.012			

Tests of Within-Subjects Contrasts

Measure:MEASURE_1

Source		speed	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
model	Linear		.335	1	.335	16.870	.000	.287
model * syndrome	Linear		.000	1	.000	.019	.890	.000
Error(model)	Linear		.833	42	.020			
speed		Linear	.003	1	.003	.181	.673	.004
		Quadratic	.210	1	.210	14.786	.000	.260
speed * syndrome		Linear	.005	1	.005	.351	.557	.008
		Quadratic	.004	1	.004	.282	.598	.007
Error(speed)		Linear	.584	42	.014			
		Quadratic	.596	42	.014			
model * speed	Linear	Linear	.003	1	.003	.595	.445	.014
		Quadratic	.160	1	.160	21.039	.000	.334
model * speed * syndrome	Linear	Linear	.004	1	.004	.904	.347	.021
		Quadratic	.001	1	.001	.077	.782	.002
Error(model*speed)	Linear	Linear	.201	42	.005			
		Quadratic	.320	42	.008			

Tests of Between-Subjects Effects

Measure:MEASURE_1

Transformed Variable:Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	1.819	1	1.819	61.789	.000	.595
syndrome	.267	1	.267	9.061	.004	.177
Error	1.236	42	.029			

Estimated Marginal Means **1. SYNDROME**

Estimates

syndrome	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
nonDS	.051	.014	.023	.080
DS	.115	.016	.084	.147

Pairwise Comparisons

(I) syndrome	(J) syndrome	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
nonDS	DS	-.064*	.021	.004	-.107	-.021
DS	nonDS	.064*	.021	.004	.021	.107

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

Univariate Tests

	Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Contrast	.044	1	.044	9.061	.004	.177
Error	.206	42	.005			

The F tests the effect of syndrome. This test is based on the linearly independent pairwise comparisons among the estimated marginal means.

Estimated Marginal Means **2. MODEL**

Estimates

Measure:MEASURE_1

model	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	.119	.018	.083	.155
2	.048	.008	.032	.063

Pairwise Comparisons

Measure:MEASURE_1

(I) model	(J) model	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.071*	.017	.000	.036	.107
2	1	-.071*	.017	.000	-.107	-.036

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

Multivariate Tests

	Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Pillai's trace	.287	16.870[a]	1.000	42.000	.000	.287
Wilks' lambda	.713	16.870[a]	1.000	42.000	.000	.287
Hotelling's trace	.402	16.870[a]	1.000	42.000	.000	.287
Roy's largest root	.402	16.870[a]	1.000	42.000	.000	.287

Each F tests the multivariate effect of model. These tests are based on the linearly independent pairwise comparisons among the estimated marginal means.

a. Exact statistic

Estimated Marginal Means

3. SPEED

Estimates

Measure:MEASURE_1

speed	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	.067	.015	.036	.098
2	.123	.015	.093	.154
3	.060	.014	.031	.088

Pairwise Comparisons

(I) speed	(J) speed	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.056*	.015	.001	-.093	-.020
	3	.008	.018	1.000	-.037	.052
2	1	.056*	.015	.001	.020	.093
	3	.064*	.021	.011	.012	.116
3	1	-.008	.018	1.000	-.052	.037
	2	-.064*	.021	.011	-.116	-.012

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

Multivariate Tests

	Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
Pillai's trace	.279	7.926[a]	2.000	41.000	.001	.279
Wilks' lambda	.721	7.926[a]	2.000	41.000	.001	.279
Hotelling's trace	.387	7.926[a]	2.000	41.000	.001	.279
Roy's largest root	.387	7.926[a]	2.000	41.000	.001	.279

Each F tests the multivariate effect of speed. These tests are based on the linearly independent pairwise comparisons among the estimated marginal means.

a. Exact statistic

General Linear Model

Post Hoc – One-way ANOVA (speed) for Digiwalker on absolute error

```
GLM PaceRHD SlowRHD FastRHD
  /WSFACTOR=speed 3 Simple
  /METHOD=SSTYPE(3)
  /PRINT=ETASQ
  /CRITERIA=ALPHA(.05)
  /WSDESIGN=speed.
```

[DataSet1] \\onid-fs\pitchfoe\DS THESIS DATA\DSData0509.sav

Within-Subjects Factors

Measure:MEASURE_1

speed	Dependent Variable
1	PaceRHD
2	SlowRHD
3	FastRHD

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
speed	Pillai's Trace	.363	11.961[a]	2.000	42.000	.000	.363
	Wilks' Lambda	.637	11.961[a]	2.000	42.000	.000	.363
	Hotelling's Trace	.570	11.961[a]	2.000	42.000	.000	.363
	Roy's Largest Root	.570	11.961[a]	2.000	42.000	.000	.363

a. Exact statistic

b. Design: Intercept

Within Subjects Design: speed

Mauchly's Test of Sphericity[b]

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[a]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
speed	.918	3.611	2	.164	.924	.964	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b. Design: Intercept Within Subjects Design: speed

Tests of Within-Subjects Effects

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
speed	Sphericity Assumed	.381	2	.191	14.008	.000	.246
	Greenhouse-Geisser	.381	1.848	.206	14.008	.000	.246
	Huynh-Feldt	.381	1.927	.198	14.008	.000	.246
	Lower-bound	.381	1.000	.381	14.008	.001	.246
Error(speed)	Sphericity Assumed	1.170	86	.014			
	Greenhouse-Geisser	1.170	79.454	.015			
	Huynh-Feldt	1.170	82.863	.014			
	Lower-bound	1.170	43.000	.027			

Tests of Within-Subjects Contrasts

Source	speed	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
speed	Level 1 vs. Level 3	.014	1	.014	.575	.452	.013
	Level 2 vs. Level 3	.652	1	.652	18.624	.000	.302
speed	Level 2 vs. Level 1	.478	1	.478	20.683	.000	.325
	Level 3 vs. Level 1	.014	1	.014	.575	.452	.013
Error(speed)	Level 1 vs. Level 3	1.011	43	.024			
	Level 2 vs. Level 3	1.506	43	.035			
Error(speed)	Level 2 vs. Level 1	.993	43	.023			
	Level 3 vs. Level 1	1.011	43	.024			

** NOTE: Level 1 = Self Paced Speed Level 2 = Slow Speed Level 3 = Fast Speed

Tests of Between-Subjects Effects

Transformed Variable:Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	.593	1	.593	40.720	.000	.486
Error	.626	43	.015			

General Linear Model

Post Hoc – One-way ANOVA (speed) for Omron on absolute error

```
GLM PaceRHO SlowRHO FastRHO
  /WSFACTOR=speed 3 Simple
  /METHOD=SSTYPE(3)
  /PRINT=ETASQ
  /CRITERIA=ALPHA(.05)
  /WSDESIGN=speed.
```

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Within-Subjects Factors

Measure:MEASURE_1

speed	Dependent Variable
1	PaceRHO
2	SlowRHO
3	FastRHO

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
speed	Pillai's Trace	.010	.207[a]	2.000	42.000	.814	.010
	Wilks' Lambda	.990	.207[a]	2.000	42.000	.814	.010
	Hotelling's Trace	.010	.207[a]	2.000	42.000	.814	.010
	Roy's Largest Root	.010	.207[a]	2.000	42.000	.814	.010

a. Exact statistic

b. Design: Intercept Within Subjects Design: speed

Mauchly's Test of Sphericity[b]

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[a]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
speed	.894	4.729	2	.094	.904	.941	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b. Design: Intercept Within Subjects Design: speed

Tests of Within-Subjects Effects

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
speed	Sphericity Assumed	.002	2	.001	.174	.841	.004
	Greenhouse-Geisser	.002	1.808	.001	.174	.819	.004
	Huynh-Feldt	.002	1.882	.001	.174	.828	.004
	Lower-bound	.002	1.000	.002	.174	.679	.004
Error(speed)	Sphericity Assumed	.545	86	.006			
	Greenhouse-Geisser	.545	77.724	.007			
	Huynh-Feldt	.545	80.934	.007			
	Lower-bound	.545	43.000	.013			

Tests of Within-Subjects Contrasts

Measure:MEASURE_1

Source	speed	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
speed	Level 1 vs. Level 3	8.378E-6	1	8.378E-6	.001	.980	.000
	Level 2 vs. Level 3	.003	1	.003	.199	.658	.005
Error(speed)	Level 1 vs. Level 3	.578	43	.013			
	Level 2 vs. Level 3	.679	43	.016			

Tests of Between-Subjects Effects

Measure:MEASURE_1

Transformed Variable:Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	.088	1	.088	24.890	.000	.367
Error	.153	43	.004			

General Linear Model

2 x 2 x 3 RM ANCOVA
Group x Model x Speed
With waist-to-hip ratio covariate
on absolute error

```
GLM PaceRHD SlowRHD FastRHD PaceRHO SlowRHO FastRHO BY syndrome WITH HWratio
  /WSFACTOR=model 2 Polynomial speed 3 Polynomial
  /METHOD=SSTYPE(3)
  /PRINT=ETASQ
  /CRITERIA=ALPHA(.05)
  /WSDESIGN=model speed model*speed
  /DESIGN=HWratio syndrome.
```

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Within-Subjects Factors

Measure:MEASURE_1

model	speed	Dependent Variable
1	1	PaceRHD
	2	SlowRHD
	3	FastRHD
2	1	PaceRHO
	2	SlowRHO
	3	FastRHO

Between-Subjects Factors

		Value Label	N
syndrome	.00	nonDS	24
	1.00	DS	20

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
model	Pillai's Trace	.046	1.978[a]	1.000	41.000	.167	.046
	Wilks' Lambda	.954	1.978[a]	1.000	41.000	.167	.046
	Hotelling's Trace	.048	1.978[a]	1.000	41.000	.167	.046
	Roy's Largest Root	.048	1.978[a]	1.000	41.000	.167	.046
model * HWratio	Pillai's Trace	.075	3.338[a]	1.000	41.000	.075	.075
	Wilks' Lambda	.925	3.338[a]	1.000	41.000	.075	.075
	Hotelling's Trace	.081	3.338[a]	1.000	41.000	.075	.075
	Roy's Largest Root	.081	3.338[a]	1.000	41.000	.075	.075
model * syndrome	Pillai's Trace	.001	.034[a]	1.000	41.000	.856	.001
	Wilks' Lambda	.999	.034[a]	1.000	41.000	.856	.001
	Hotelling's Trace	.001	.034[a]	1.000	41.000	.856	.001
	Roy's Largest Root	.001	.034[a]	1.000	41.000	.856	.001
speed	Pillai's Trace	.127	2.917[a]	2.000	40.000	.066	.127
	Wilks' Lambda	.873	2.917[a]	2.000	40.000	.066	.127
	Hotelling's Trace	.146	2.917[a]	2.000	40.000	.066	.127
	Roy's Largest Root	.146	2.917[a]	2.000	40.000	.066	.127
speed * HWratio	Pillai's Trace	.104	2.313[a]	2.000	40.000	.112	.104
	Wilks' Lambda	.896	2.313[a]	2.000	40.000	.112	.104

	Hotelling's Trace	.116	2.313[a]	2.000	40.000	.112	.104
	Roy's Largest Root	.116	2.313[a]	2.000	40.000	.112	.104
speed * syndrome	Pillai's Trace	.018	.357[a]	2.000	40.000	.702	.018
	Wilks' Lambda	.982	.357[a]	2.000	40.000	.702	.018
	Hotelling's Trace	.018	.357[a]	2.000	40.000	.702	.018
	Roy's Largest Root	.018	.357[a]	2.000	40.000	.702	.018
model * speed	Pillai's Trace	.123	2.797[a]	2.000	40.000	.073	.123
	Wilks' Lambda	.877	2.797[a]	2.000	40.000	.073	.123
	Hotelling's Trace	.140	2.797[a]	2.000	40.000	.073	.123
	Roy's Largest Root	.140	2.797[a]	2.000	40.000	.073	.123
model * speed * HWratio	Pillai's Trace	.098	2.174[a]	2.000	40.000	.127	.098
	Wilks' Lambda	.902	2.174[a]	2.000	40.000	.127	.098
	Hotelling's Trace	.109	2.174[a]	2.000	40.000	.127	.098
	Roy's Largest Root	.109	2.174[a]	2.000	40.000	.127	.098
model * speed * syndrome	Pillai's Trace	.042	.884[a]	2.000	40.000	.421	.042
	Wilks' Lambda	.958	.884[a]	2.000	40.000	.421	.042
	Hotelling's Trace	.044	.884[a]	2.000	40.000	.421	.042
	Roy's Largest Root	.044	.884[a]	2.000	40.000	.421	.042

a. Exact statistic

b. Design: Intercept + HWratio + syndrome

Within Subjects Design: model + speed + model * speed

Mauchly's Test of Sphericity[b]

Measure:MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[a]		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
model	1.000	.000	0	.	1.000	1.000	1.000
speed	.806	8.631	2	.013	.837	.912	.500
model * speed	.946	2.237	2	.327	.948	1.000	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b. Design: Intercept + HWratio + syndrome Within Subjects Design: model + speed + model * speed

Tests of Within-Subjects Effects

Measure:MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
model	Sphericity Assumed	.037	1	.037	1.978	.167	.046
	Greenhouse-Geisser	.037	1.000	.037	1.978	.167	.046
	Huynh-Feldt	.037	1.000	.037	1.978	.167	.046
	Lower-bound	.037	1.000	.037	1.978	.167	.046

model * HWratio	Sphericity Assumed	.063	1	.063	3.338	.075	.075
	Greenhouse-Geisser	.063	1.000	.063	3.338	.075	.075
	Huynh-Feldt	.063	1.000	.063	3.338	.075	.075
	Lower-bound	.063	1.000	.063	3.338	.075	.075
model * syndrome	Sphericity Assumed	.001	1	.001	.034	.856	.001
	Greenhouse-Geisser	.001	1.000	.001	.034	.856	.001
	Huynh-Feldt	.001	1.000	.001	.034	.856	.001
	Lower-bound	.001	1.000	.001	.034	.856	.001
Error(model)	Sphericity Assumed	.770	41	.019			
	Greenhouse-Geisser	.770	41.000	.019			
	Huynh-Feldt	.770	41.000	.019			
	Lower-bound	.770	41.000	.019			
speed	Sphericity Assumed	.047	2	.023	1.677	.193	.039
	Greenhouse-Geisser	.047	1.675	.028	1.677	.198	.039
	Huynh-Feldt	.047	1.823	.026	1.677	.196	.039
	Lower-bound	.047	1.000	.047	1.677	.203	.039
speed * HWratio	Sphericity Assumed	.039	2	.019	1.383	.257	.033
	Greenhouse-Geisser	.039	1.675	.023	1.383	.256	.033
	Huynh-Feldt	.039	1.823	.021	1.383	.257	.033
	Lower-bound	.039	1.000	.039	1.383	.246	.033

speed * syndrome	Sphericity Assumed	.012	2	.006	.431	.652	.010
	Greenhouse-Geisser	.012	1.675	.007	.431	.616	.010
	Huynh-Feldt	.012	1.823	.007	.431	.633	.010
	Lower-bound	.012	1.000	.012	.431	.515	.010
Error(speed)	Sphericity Assumed	1.142	82	.014			
	Greenhouse-Geisser	1.142	68.672	.017			
	Huynh-Feldt	1.142	74.750	.015			
	Lower-bound	1.142	41.000	.028			
model * speed	Sphericity Assumed	.035	2	.018	2.927	.059	.067
	Greenhouse-Geisser	.035	1.897	.019	2.927	.062	.067
	Huynh-Feldt	.035	2.000	.018	2.927	.059	.067
	Lower-bound	.035	1.000	.035	2.927	.095	.067
model * speed * HWratio	Sphericity Assumed	.026	2	.013	2.114	.127	.049
	Greenhouse-Geisser	.026	1.897	.013	2.114	.130	.049
	Huynh-Feldt	.026	2.000	.013	2.114	.127	.049
	Lower-bound	.026	1.000	.026	2.114	.154	.049
model * speed * syndrome	Sphericity Assumed	.009	2	.004	.741	.480	.018
	Greenhouse-Geisser	.009	1.897	.005	.741	.473	.018
	Huynh-Feldt	.009	2.000	.004	.741	.480	.018
	Lower-bound	.009	1.000	.009	.741	.394	.018

Error(model*speed)	Sphericity Assumed	.495	82	.006			
	Greenhouse-Geisser	.495	77.771	.006			
	Huynh-Feldt	.495	82.000	.006			
	Lower-bound	.495	41.000	.012			

Tests of Within-Subjects Contrasts

Source	speed	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
model	Linear	.037	1	.037	1.978	.167	.046
model * HWratio	Linear	.063	1	.063	3.338	.075	.075
model * syndrome	Linear	.001	1	.001	.034	.856	.001
Error(model)	Linear	.770	41	.019			
speed	Linear	.028	1	.028	2.071	.158	.048
	Quadratic	.019	1	.019	1.306	.260	.031
speed * HWratio	Linear	.030	1	.030	2.215	.144	.051
	Quadratic	.009	1	.009	.598	.444	.014
speed * syndrome	Linear	.010	1	.010	.732	.397	.018
	Quadratic	.002	1	.002	.147	.704	.004
Error(speed)	Linear	.554	41	.014			
	Quadratic	.588	41	.014			

model * speed	Linear	Linear	.009	1	.009	1.990	.166	.046
		Quadratic	.026	1	.026	3.514	.068	.079
model * speed * HWratio	Linear	Linear	.010	1	.010	2.232	.143	.052
		Quadratic	.015	1	.015	2.041	.161	.047
model * speed * syndrome	Linear	Linear	.007	1	.007	1.474	.232	.035
		Quadratic	.002	1	.002	.281	.599	.007
Error(model*speed)	Linear	Linear	.191	41	.005			
		Quadratic	.304	41	.007			

Tests of Between-Subjects Effects

Measure:MEASURE_1

Transformed Variable:Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	.040	1	.040	1.451	.235	.034
HWratio	.111	1	.111	4.033	.051	.090
syndrome	.202	1	.202	7.352	.010	.152
Error	1.126	41	.027			

APPENDIX D

SAS Statistics and Formulas for Manuscript 2

```
Control Group -- Digiwalker

options ps = 55;
options ls = 79;
pageno = 1;
data run;
input
        sub p1u1 p1u2 p2u1 p2u2;
        place=1;unit=1;score=p1u1;output;
        place=1;unit=2;score=p1u2;output;
        place=2;unit=1;score=p2u1;output;
        place=2;unit=2;score=p2u2;output;

cards;
101   2249  1731  2293  1287
102   1059  2093  2222  2011
103   2481  2502  2488  2506
104   2351  2264  2567  2336
105   1235  1189  1622  1105
106   2390  2292  2358  2277
107   1847  1859  1922  1881
108   2372  2365  2381  2363
109   1896  2069  2070  2033
110   1421  1440  1418  1763
111   1873  1804  2082  1908
112   1974  1932  1977  1970
113   1980  1920  2009  1932
114   2031  2004  2020  1980
115   2204  1458  2354  2171
116   1830  1738  1848  1820
118   2182  1491  2197  2054
119   2128  1801  1815  1825
120   1959  2100  1641  2062
121   2166  2166  2128  2167
122   1966  1921  1971  1959
123   1548  1550  1553  1584
124   2160  2081  2138  2141
;

Proc varcomp method=type1;
        class sub place unit;
        model score=sub place unit
                                sub*place
                                sub*unit
                                place*unit;
Proc glm;
        class sub place unit;
        model score=sub place unit
                                sub*place
                                sub*unit
                                place*unit;
run;
```

```
Control Group -- Omron

options ps = 55;
options ls = 79;
pageno = 1;
data run;
input
        sub p1u1 p1u2 p2u1 p2u2 p3u1 p3u2 p4u1 p4u2;
        place=1;unit=1;score=p1u1;output;
        place=1;unit=2;score=p1u2;output;
        place=2;unit=1;score=p2u1;output;
        place=2;unit=2;score=p2u2;output;
        place=3;unit=1;score=p3u1;output;
        place=3;unit=2;score=p3u2;output;
        place=4;unit=1;score=p4u1;output;
        place=4;unit=2;score=p4u2;output;

cards;
101   2103  2124  2135  2119  2107  2109  2127  2150
102   2233  2106  2232  2253  2262  2250  2214  2236
103   2492  2488  2493  2483  2465  2476  2442  2488
104   2302  2304  2302  2308  2220  2291  2272  2248
105   1834  1461  1464  1562  1425  1502  0981  1043
106   2329  2325  2335  2336  2260  2198  2320  2309
107   1913  1896  1855  1910  1966  2040  1993  1853
108   2373  2458  2369  2380  2408  2386  2391  2375
109   2094  2237  2087  2105  2034  2141  2108  2098
110   1791  1811  1681  1967  1889  1772  1977  1892
111   1881  1938  1855  1881  1832  1874  1882  2003
112   1900  1989  1946  1950  1981  2005  1968  1974
113   1946  1954  1958  1961  0893  1874  1500  1931
114   2026  2274  2029  2045  2118  2094  2150  2242
115   2294  2318  2305  2305  2303  2265  0196  2275
116   1855  1998  1800  1855  1934  1858  1908  1869
118   2210  2197  2214  2199  2168  1018  2201  2205
119   1805  1784  1795  1792  1683  1759  1310  1770
120   2075  2083  2062  2056  2003  2073  2120  2135
121   2153  2147  2172  2177  2141  2169  2178  2117
122   1976  2009  1988  1991  1960  1977  2004  2001
123   1607  1596  1610  1610  1610  1613  1625  1617
124   2148  2153  2155  2149  2157  2146  2168  2093
;

Proc varcomp method=type1;
        class sub place unit;
        model score=sub place unit
                                sub*place
                                sub*unit
                                place*unit;
Proc glm;
        class sub place unit;
        model score=sub place unit
                                sub*place
                                sub*unit
                                place*unit;
```

```
run;

Down Syndrome -- Digiwalker

options ps = 55;
options ls = 79;
pageno = 1;
data run;
input
        sub p1u1 p1u2 p2u1 p2u2;
        place=1;unit=1;score=p1u1;output;
        place=1;unit=2;score=p1u2;output;
        place=2;unit=1;score=p2u1;output;
        place=2;unit=2;score=p2u2;output;

cards;
1     1484  1472  1470  1239
2     1521  1364  1096  1005
3     2076  1563  2075  1679
4     2480  2493  2485  2494
6     2137  2250  2074  2236
7     2238  2099  2230  2105
9     2418  2283  2525  2393
10    2531  2135  2512  2471
11    2298  2331  2392  2332
12    2410  2403  2388  2415
13    2247  2258  2204  2197
14    2150  2106  2141  2123
15    2169  2035  2152  1994
16    2135  1814  2175  1589
17    2002  1121  1969  1969
18    1450  1753  1722  1899
20    2112  1951  2125  2107
;

Proc varcomp method=type1;
        class sub place unit;
        model score=sub place unit
                                sub*place
                                sub*unit
                                place*unit;
Proc glm;
        class sub place unit;
        model score=sub place unit
                                sub*place
                                sub*unit
                                place*unit;
run;
```

```
Down Syndrome -- Omron

options ps = 55;
options ls = 79;
pageno = 1;
data run;
input
        sub p1u1 p1u2 p2u1 p2u2 p3u1 p3u2 p4u1 p4u2;
        place=1;unit=1;score=p1u1;output;
        place=1;unit=2;score=p1u2;output;
        place=2;unit=1;score=p2u1;output;
        place=2;unit=2;score=p2u2;output;
        place=3;unit=1;score=p3u1;output;
        place=3;unit=2;score=p3u2;output;
        place=4;unit=1;score=p4u1;output;
        place=4;unit=2;score=p4u2;output;

cards;
1     1186  1088  1488  1185  1274  1288  1307  1197
2     1255  1367  1147  1317  1557  1513  1622  1561
3     1787  1722  1828  1684  1961  2003  1624  1817
4     2445  2414  2410  2430  2385  2415  2393  2463
6     2234  2204  2237  2208  2248  2255  2277  2265
7     2163  2175  2170  2148  1994  2155  2085  2122
9     2448  2482  2474  2475  2429  2477  2481  2481
10    2392  2473  2514  2488  2407  2316  2467  2376
11    2314  2307  2308  2313  2313  2297  2310  2318
12    2151  2398  2374  2392  2336  1971  2334  2213
13    2177  2158  2182  2171  2162  2124  2140  2175
14    2143  2150  2140  2151  2207  2150  2138  2169
15    2183  2186  2181  2179  1890  2024  1484  2095
16    2141  2170  2174  2169  2160  2198  2200  2199
17    1979  1970  1970  1961  1835  1904  1930  1950
18    1815  1883  1869  1881  1897  1800  1833  1851
20    2133  2133  2135  2136  2082  1979  2096  1814
;

Proc varcomp method=type1;
        class sub place unit;
        model score=sub place unit
                                sub*place
                                sub*unit
                                place*unit;
Proc glm;
        class sub place unit;
        model score=sub place unit
                                sub*place
                                sub*unit
                                place*unit;
run;
```

FORMULAS

Relative Magnitude:

$$\text{Relative Magnitude (x\%)} = \frac{\boldsymbol{\sigma_X}}{\boldsymbol{\Sigma\sigma}} \text{ x } 100$$

Phi (Φ) Coefficient:

$$\boldsymbol{\Phi} = \frac{\boldsymbol{\sigma^2 s}}{\boldsymbol{\sigma^2 s} + \frac{\boldsymbol{\sigma^2 p}}{\boldsymbol{n_p}} + \frac{\boldsymbol{\sigma^2 u}}{\boldsymbol{n_u}} + \frac{\boldsymbol{\sigma^2 sp}}{\boldsymbol{n_p}} + \frac{\boldsymbol{\sigma^2 su}}{\boldsymbol{n_u}} + \frac{\boldsymbol{\sigma^2 pu}}{\boldsymbol{n_p n_u}} + \frac{\boldsymbol{\sigma^2 spu, e}}{\boldsymbol{n_p n_u}}}$$

G Coefficient:

$$\mathbf{G} = \frac{\boldsymbol{\sigma^2 s}}{\boldsymbol{\sigma^2 s} + \frac{\boldsymbol{\sigma^2 sp}}{\boldsymbol{n_p}} + \frac{\boldsymbol{\sigma^2 su}}{\boldsymbol{n_u}} + \frac{\boldsymbol{\sigma^2 spu, e}}{\boldsymbol{n_p n_u}}}$$

APPENDIX E

Data Collection Forms

Demographic Questionnaire

ID CODE: _____

Date: _____

Participant Information:

Birthday: _____ Age: _____

Gender: Male Female

Do you (the participant) have Down syndrome?

_____ Yes _____ No

1. _____ / 2. _____

ID CODE: _____

Date: _____

Height: _____ cm Weight: _____ kg

Waist Circumference	Belt Circumference	Hip Circumference
1. _____ cm	1. _____ cm	1. _____ cm
2. _____ cm	2. _____ cm	2. _____ cm
3. _____ cm	3. _____ cm	3. _____ cm

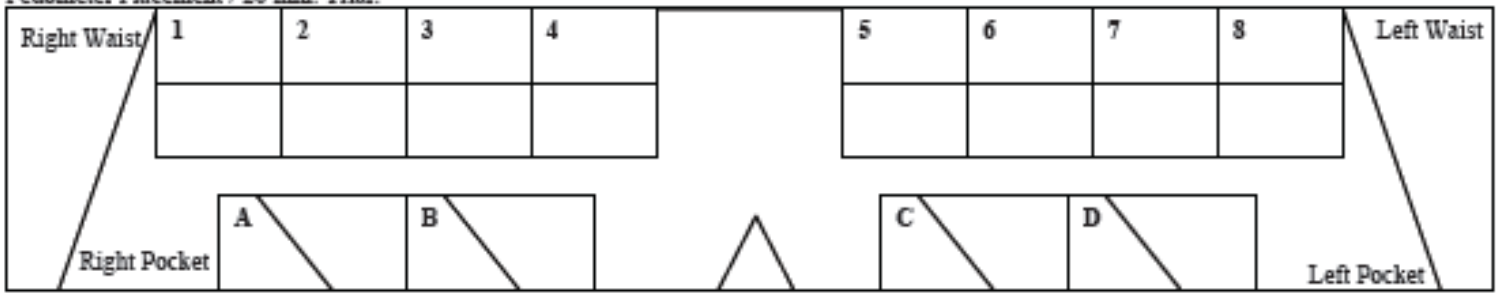
Waist Circumference	Belt Circumference	Hip Circumference
cm	cm	cm

Hip Width _____ cm

Waist Width _____ cm

Participant ID # _____ Date _____ 1. _____ / 2. _____

Pedometer Placement / 20 min. Trial:



Trial 1: Speed m/s mph

Hand Count

1.	2.	3.	4.		5.	6.	7.	8.
	A.	B.				C.	D.	

Zone 1: _____ s Zone 2: _____ s

Zone 3: _____ s Zone 4: _____ s

Zone 5: _____ s Zone 6: _____ s

Zone 7: _____ s TOTAL: _____ s

Average: _____ s M/S: _____ s

MPH: _____ s

Retest 1: Speed m/s mph

Hand Count

1.	2.	3.	4.		5.	6.	7.	8.
	A.	B.				C.	D.	

Zone 1: _____ s Zone 2: _____ s

Zone 3: _____ s Zone 4: _____ s

Zone 5: _____ s Zone 6: _____ s

Zone 7: _____ s TOTAL: _____ s

Average: _____ s M/S: _____ s

MPH: _____ s

Trial 2: Speed _____ m/s _____ mph FAST SLOW

Hand Count

1.	2.	3.	4.		5.	6.	7.	8.
	A.	B.				C.	D.	

Zone 1: _____ s Zone 2: _____ s

Zone 3: _____ s Zone 4: _____ s

Zone 5: _____ s Zone 6: _____ s

Zone 7: _____ s TOTAL: _____ s

Retest 2: Speed _____ m/s _____ mph FAST SLOW

Hand Count

1.	2.	3.	4.		5.	6.	7.	8.
	A.	B.				C.	D.	

Zone 1: _____ s Zone 2: _____ s

Zone 3: _____ s Zone 4: _____ s

Zone 5: _____ s Zone 6: _____ s

Zone 7: _____ s TOTAL: _____ s

Trail 3: Speed _____ m/s _____ mph FAST SLOW

Hand Count

1.	2.	3.	4.		5.	6.	7.	8.
	A.	B.				C.	D.	

Zone 1: _____ s Zone 2: _____ s

Zone 3: _____ s Zone 4: _____ s

Zone 5: _____ s Zone 6: _____ s

Zone 7: _____ s TOTAL: _____ s

Retest 2: Speed _____ m/s _____ mph FAST SLOW

Hand Count

1.	2.	3.	4.		5.	6.	7.	8.
	A.	B.				C.	D.	

Zone 1: _____ s Zone 2: _____ s

Zone 3: _____ s Zone 4: _____ s

Zone 5: _____ s Zone 6: _____ s

Zone 7: _____ s TOTAL: _____ s