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A Comparison of Commonly Used Accelerometer Based Activity Monitors in Controlled and Free-Living Environment

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To the Graduate Council:

I am submitting herewith a dissertation written by Yuri Feito entitled "A Comparison of Commonly Used Accelerometer Based Activity Monitors in Controlled and Free-Living Environment." I have examined the final electronic copy of this dissertation for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Doctor of Philosophy, with a major in Exercise and Sport Sciences.

David R. Bassett, Major Professor

We have read this dissertation and recommend its acceptance:

Dixie L. Thompson, Eugene C. Fitzhugh, Edward T. Howley, Betty Greer

Accepted for the Council:
Carolyn R. Hodges

Vice Provost and Dean of the Graduate School

(Original signatures are on file with official student records.)

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A COMPARISON OF COMMONLY USED ACCELEROMETER-BASED ACTIVITY MONITORS IN CONTROLLED AND FREE-LIVING ENVIRONMENT

A dissertation

Presented for the

Doctor of Philosophy Degree

The University of Tennessee, Knoxville

Yuri Feito

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DEDICATION

Este trabajo está dedicado a mis padres, quienes lo dejaron todo para darme la oportunidad de poder realizar todos mis sueños en libertad. Quiero que conste lo agradecido que les estoy por todos los sacrificios que han hecho durante todos estos años. A los dos les doy las gracias por haber tomado las decisiones que nos permitieron "conocer" el mundo, aunque haya sido obligado, y por librarnos de un gobierno opresivo y aniquilante. Papi, gracias por la mano dura y por tu inquebrantable labor en hacerme un hombre de trabajo y de bien. Nunca me enseñaste a dejar las cosas a medias, exigiendo siempre lo mejor de mí. Mami, darte las gracias por tu comprensión sería poco. Siempre has sido una madre ejemplar.

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A los más pequeños, espero haberles dado un buen ejemplo y quiero dejarles saber que sí se puede lograr todo en la vida cuando se tiene empeño. No dejen que nada ni nadie les destruya sus sueños. El trabajo, la determinación y el sacrificio son la fórmula que se requiere para llegar lejos en la vida. Aunque este doctorado sea el primero de la familia espero que no sea el último.

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ABSTRACT

This dissertation was designed to determine the effects of body mass index (BMI) and walking speed on activity monitor outputs. A secondary purpose was to compare the activity monitors' performance in a free-living environment.

In the first experiment, 71 participants wore three waist-mounted activity monitors (Actical, ActiGraph, and NL-2000) and an ankle-mounted device (StepWatch 3) while walking on a treadmill (40, 67 and 94 m/min). The tilt angle of each device was measured. The Actical recorded 26% higher activity counts ($P < 0.01$) in obese persons with a tilt < 10 degrees, compared to normal weight persons. The ActiGraph was unaffected by BMI or tilt angle.

In the second experiment, the steps recorded by the devices were compared to actual steps. Speed had the greatest influence on the accuracy these devices. At 40 m/min, the ActiGraph was the least accurate device for normal weight (38%), overweight (46%) and obese (48%) individuals. The Actical, NL-2000 and StepWatch averaged 65%, 73% and 99% of steps taken, respectively.

Lastly, several generations of the ActiGraph (7164, GT1M, and GT3X), and other research grade activity monitors (Actical; ActivPAL; and Digi-Walker) were compared to a criterion measure of steps. Fifty-six participants performed treadmill walking (40, 54, 67, 80 and 94 m/min) and wore the devices for 24-hours under free-living conditions. BMI did not affect step count accuracy during treadmill walking. The StepWatch, PAL, and the AG7164 were the most accurate across all speeds; the other devices were only accurate at the faster speeds. In the free-living environment, all devices recorded about 75% of StepWatch-determined steps, except the AG7164 (99%).

Based on these findings, we conclude that BMI does not affect the output of these activity monitors. However, waist-borne activity monitors are highly susceptible to under-counting steps at walking speeds below 67 m/min, or stepping rates below 100 steps/min. An activity monitor worn on the ankle is less susceptible to these speed effects and provides the greatest accuracy for step counting.

TABLE OF CONTENTS

List of Tables

List of Figures

CHAPTER 1

INTRODUCTION

The benefits of physical activity (PA) are well documented. An active lifestyle has been linked with reductions in overall and disease specific mortality and morbidity leading to increases in longevity and improved quality of life [1-9]. In 1996, the publication of the Surgeon General's Report on Physical Activity and Health, was a landmark event in documenting the health benefits of physical activity [10]. In 2008, the Department of Health and Human Services (HHS) published the Physical Activity Guidelines for all Americans, emphasizing the reduction of sedentary behaviors and recommended a minimum of 150 minutes per week of moderate-intensity physical activity or 75 minutes of vigorous physical activity, which can be accumulated in bouts of at least 10 minutes [11]. Other recommendations have promoted the use of mechanical devices, such as step counters, and encourage people to accumulate 10,000 steps per day as a way to achieve adequate PA levels [12]. Even though the benefits of PA are well know, half of the U.S. population does not engage in enough physical activity to meet the HHS recommendations, and about 25% report no physical activity at all [13].

Accurately measuring PA has been important to investigators over the years. Although direct observation would be considered the best way to measure a person's activity level, it is neither practical nor realistic. Thus, investigators have primarily relied on self-report methods (e.g. surveys, diaries) to assess PA patterns [14]; however, these methods have limitations that may provide imprecise estimates of overall physical activity as purposeful, more intense activities are more easily recalled than everyday activities [14-17]. Therefore, investigators have suggested the use of objective monitoring devices, such as pedometers and accelerometers, to more accurately measure PA [14].

Pedometers and accelerometers are small, portable devices that provide a means of measuring PA while being minimally intrusive to the participants. Their validity and reliability have been widely studied, with good results [18-23], and their use in PA interventions has been shown to be beneficial [24-25]. However, some limitations have been reported that may limit the applicability of these devices for all populations (i.e. elderly, obese). Primarily, there seems to be a speed threshold below which the devices are less accurate [21-22, 26-27]. Several studies have suggested that at the slower speeds, the magnitude of the vertical acceleration is below the devices' sensitivity thresholds, causing these devices to underestimate activity levels [21-22, 27]. Additionally, abdominal adiposity, and tilt-angle seem to have a negative effect on spring-levered devices, causing them to significantly underestimate PA in overweight and obese. On the other hand, piezoelectric monitors do not seem to be affected by abdominal adiposity or tilt-angle, thereby making them more appropriate devices to measure PA among overweight and obese populations [28-30]. Moreover, some investigators have suggested the use of ankle-borne devices as a way to provide accurate estimates of physical activity, as these devices are not affected by slow walking speed or adiposity [31-32].

Statement of the Problem

Considering their small size and portability, as well as their accuracy for measuring ambulatory activity, accelerometer-based activity monitors are presently being used to objectively measure physical activity levels in various countries. In recent years, surveillance systems in the U.S. [33-34], Canada [35-36] and Europe [37] have incorporated accelerometer-based activity monitors to estimate secular trends in physical activity among adults and children, and have developed activity cut-offs in order to categorize population-based activity levels and estimate health outcomes [38-40]. Therefore, there is a need for accurate and reliable devices that can measure PA in normal weight, overweight, and obese individuals.

Statement of Purpose

The purpose of this dissertation is to determine if measures of adiposity (i.e. BMI and waist circumference) have a significant effect on the accuracy of commonly used accelerometer-based activity monitors in controlled free-living conditions. The first study (Part III) examines the role of body mass index (BMI) and speed on the Actical and ActiGraph activity monitors. The second study (Part IV) assesses the effect of BMI on the step-function of accelerometer based activity monitors. The third study (Part V) has two purposes: (1) to compared the most widely used activity monitors and assess their validity for step counting in a controlled and free-living environment; and (2) to determine the validity of three generations of the same device during treadmill walking and in the free-living environment.

Significance of this Study

As physical activity monitors continue to gain momentum in physical activity research and surveillance systems, it is important to determine if these devices are accurate and reliable. These series of studies will be the first to examine how markers of adiposity affect the output of commonly used accelerometer-based activity monitors. In addition, it will compare three different generations of the same device (i.e. ActiGraph) to a criterion method.

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CHAPTER 2

LITERATURE REVIEW

ACTIVITY MONITORS

Walking is the preferred mode of physical activity for over 30% of Americans [1-2], and thus it is important to find ways to measure this mode of physical activity accurately. Self-report methods (e.g. surveys, diaries) permit researchers to gather data from large number of individuals at relatively low cost and allow patterns of behavior to be examined [3]. However, these methods have a number of limitations which may hinder the accurate assessment of PA [4]. First, social desirability might lead to over-reporting by individuals who knowingly do not engage in recommended amounts of PA [5]. Secondly, the recalling of PA has been considered a highly complex cognitive task [6], thus limiting the information that some individual may be able to provide. Lastly these instruments are limited by response rates and the extent to which participants can follow instructions [3].

In 2000, Sallis and Saelens [3] assessed the reliability and validity of physical activity self-report instruments developed or used in the 1990's and concluded that psychometric tools that used an interview process had stronger psychometric characteristics than self-administered methods. However, not all types of self-report provide accurate estimates of PA; therefore, objective monitoring should be used if this is the primary outcome of interest.

Considering the limitations of these instruments and recent technological advances, researchers have been able to develop accurate and reliable devices to objectively monitor physical activity (e.g. motion sensors, heart rate monitors, global positioning systems (GPS), etc). Motion sensors, such as pedometers and accelerometers, have received the most attention, providing good evidence of their validity and reliability [7-8].

ACCELEROMETERS

Accelerometers are deigned to measure human movement through changes in acceleration, which can then be used to estimate PA over time. Technological advances have allowed the development of accelerometers that accurately assess movement patterns using small, portable, and minimally intrusive devices [7, 9]; however, the high cost, technical expertise required for accessing and interpreting the data, and need for additional hardware and software limits the usefulness of these devices for large-scale studies [7].

Different types of accelerometers exist, (e.g. piezoelectric crystals, piezoresistive and electronic piezoelectric sensors). Most of these devices use a variation of a spring mass system containing a seismic mass and a piezoelectric sensor in a cantilever beam, or integrated chip sensor design. In either design, when acceleration is applied, the seismic mass responds by applying force to the piezoelectric sensor, causing it to bend or compress [9-10]. . Accelerometers designed to measure ambulatory activity use one or more piezoelectric sensors that respond to changes in acceleration in either a single or multiple orthogonal planes (anteroposterior, mediolateral, and vertical) [9]. The piezoelectric sensor is most sensitive in a vertical direction, therefore it is often referred to as uniaxial, as it primarily records acceleration in the vertical plane [9]. Devices that contain two or more accelerometers that measure accelerations in the anteroposterior and/or lateral planes are said to be biaxial or triaxial. Omnidirectional devices theoretically assess accelerations in multiple planes, but are most sensitive to movement in a single plane [9].

In physical activity research, the raw accelerations are converted to “activity counts” by the summation of the absolute values of the sampled change in acceleration, through a specified period of time (i.e. counts·min^{-1}) [9, 11]. The accelerations recorded are proportional to muscular forces; hence these counts can hypothetically be translated into energy expenditure (EE) [12]. The general consensus is that accelerometers provide an accurate assessment of physical activity, but less accurate prediction of EE; especially in the free-living environment [13].

ActiGraph

In 1993, Computer Science and Applications (CSA), Inc. (Shalimar, FL) designed the first model of this accelerometer-based activity monitor (Model 7164). Later, several generations (GT1M and GT3X) were introduced. The 7164 is a waterproof device measuring 5.1 x 3.8 x 1.5 cm, weighing 42.6 g and able to measure acceleration in the vertical direction between 0.05 and 2.0 G's [14]. The internal mechanism is design as a cantilevered arm, which generates a charge when movement occurs and is then filtered and digitized at 10 samples per second (10 Hz) by an eight-bit Analog/Digital (A/D) converter [14].

Initialization and downloading of recorded data is achieved through software provided by the manufacturer and a reader interface unit (RIU) connected to a personal computer through a serial port. Downloading consists of the transferring of data from the device to the computer so that it can be analyzed in a commonly used spreadsheet format [14]. A 3-Volt lithium coin cell battery (#2430) provides 4-6 months of power. Continuous data recording is limited to 22 days, using a 1-minute epoch using 64K of nonvolatile random access memory (RAM). Smaller sample intervals (i.e. 1, 10, 15 or 30 seconds) result in subsequent decrease in recording time [14].

Advancements in microchip technology have allowed for the development of devices with greater memory capacity. The ActiGraph GT1M and the GT3X are the new generation of activity monitors, which similar to the 7164 are relatively small (3.8 x 3.7 x 1.8 cm; 27 g) and record accelerations in the range of 0.05 – 2.0 G's. Unlike the 7164 however, the newer versions digitize PA data through a 12-bit A/D converter at 30 Hz, thus providing three times the sampling capabilities of the 7164 model [11]. In addition, both devices use a direct USB 2.0 connection, which makes it easier to use than the serial connection of the 7164. The GT1M and GT3X are capable of measuring activity counts, steps taken, and energy expenditure and activity levels. The main difference between these two models is the greater memory capabilities (4 MB vs. 1 MB), and higher battery life (20 days vs. 14 days) of the GT3X compared to the GT1M [11].

Considering the ActiGraph has been available since the mid-1990 it makes sense that it has been the most studied. Although various versions have been introduced over the years, numerous investigators have reported on the validity of the ActiGraph to assess physical activity in the laboratory and in free-living conditions [15-18], to estimate energy expenditure [19-25] and in comparison to other activity monitors [26-35].

Janz [15] in 1994, and Melanson and Freedson [16] in 1995, validated the first generation of the ActiGraph (model 5032), which later became known as the Computer Science Application (CSA) accelerometer. Janz studied the validity of the device in children and found good correlations between the ActiGraph counts and heart rate over three days [15]. Melanson and Freedson [16] examined its validity in adults who walked and ran on a treadmill at different speeds (80, 107, and 134 $m \cdot min^{-1}$) and elevations (0%, 3% and 6% grade) while wearing devices on the hip, ankle and wrist. In general, investigators found the ActiGraph was able to detect changes in speed, but not changes in grade. In addition, significant correlations were reported between measured energy expenditure and the ActiGraph counts for hip, ankle, and wrist (0.80, 0.66, and 0.81; $P < 0.01$) [16].

In 2000, Bassett et al. [8], Hendelman et al. [18] and Welk et al. [26] concluded that although accelerometers were highly correlated with energy expenditure during ambulatory activity ($r = 0.77 – 0.86$), they were poorly correlated with lifestyle activities ($r = 0.55 – 0.59$). This underestimation of energy expenditure is due to the devices' inability to measure upper body movement, changes in terrain and/or loading activities accurately [8, 18, 26]. Furthermore, they suggested that equations developed in the laboratory to estimate energy expenditure from motion sensors are not appropriate for "lifestyle" activities in the free-living environment [8, 18].

Freedson et al. [19] developed one of the first regression equations to estimate EE from activity counts for the ActiGraph 7164. In addition, they were the first to identify specified activity counts cut-point corresponding to different activity levels (i.e. light, moderate, hard and very hard). In their study, 50 adults performed three exercising conditions for six-minutes each; walking at 80 and at 107 $m \cdot min^{-1}$, and running at 167 $m \cdot min^{-1}$`. Investigators measured oxygen consumption through open circuit spirometry, while participants wore the ActiGraph 7164 on the right hip, secured to a belt. Based on the information gathered from indirect calorimetry and the ActiGraph counts, and considering METs are metabolic equivalents (1 MET = 3.5 $ml \cdot kg^{-1} \cdot min^{-1}$) investigators developed an equation to estimate MET levels that resulted in good agreement with measured values (METs = 1.439008 + (0.000795 * $counts \cdot min^{-1}$; $r^2 = 0.82$; SEE = ± 1.12 METs). Based on this equation, subsequent accelerometer-based cut-points were design to identify levels of physical activity. Activities recording less than 1952 $counts \cdot min^{-1}$ were considered light activity (< 3.0 METs); activities recording between 1952 – 5724 $counts \cdot min^{-1}$ were classified as moderate (3.0 – 5.99 METs); hard activities (6.0 – 8.99 METs) recorded 5725 – 9498 $counts \cdot min^{-1}$; and very hard activities (> 8.99 METs) recorded over 9498 $counts \cdot min^{-1}$ [19].

In addition, based on a subsample of thirty-five participants, Freedson et al. [19] developed a prediction equation to estimate energy expenditure (kcals x min^{-1}). The equation ($kcals \cdot min^{-1} = (0.00094 * counts \cdot min^{-1}) + (0.1346 * mass\ in\ Kg) – 7.37418$); $r^2 = 0.82$; SEE= ± 1.40 $kcals \cdot min^{-1}$) was cross-validated with the other fifteen participants. The kcal equation resulted in an excellent correlation compared to the measured value (r = 0.93; SEE = ± 0.93 $kcals \cdot min^{-1}$). Furthermore, the mean difference between the actual and predicted EE were small and non-significant: -0.19, -0.46, and 0.12 kcals x min^{-1} for 80, 107 and 167 $m \cdot min^{-1}$, respectively ($P > 0.05$) [19].

Considering the limitations of a single accelerometer in trying to measure physical activity EE, Swartz et al. [20] established a prediction equation using two devices, one on the wrist and the other on the hip. Participants included relatively healthy men (N=31; ages 41 ± 17 years) and women (N=39; ages 42 ± 14 years) who completed one to six different activities within the following categories: yard work, occupation, housework, family care, conditioning and recreation. A total of 28 activities were completed, with 12 participants performing each activity. Participants wore a portable indirect calorimetry device (Cosmed $K4B^2$, Cosmed , Rome, Italy) while completing each activity to measure energy expenditure and two ActiGraphs, one on the hip at the right anterior axillary line and the other on the dominant wrist. The regression equations developed by the authors from the accelerometer counts for the wrist, hip and wrist plus hip accounted for 3.3%, 31.7% and 34.3%, respectively, of the variance in METs [20]. Thus, even though there was an improvement of 2.6% in the prediction of EE when using two accelerometer-based activity monitors, the authors concluded that this improvement is not warranted given the additional time and cost associated with the wrist-mounted device [20].

In 2004, King et al. [21] compared the validity of five different accelerometer-based activity monitors while walking and running on a treadmill. For the ActiGraph, they used the equation of Freedson et al. [19] to convert the activity counts into kcals·min^{-1}. When compared with indirect calorimetry, most monitors overestimated EE at most treadmill speeds. However, the ActiGraph only underestimated EE at the lowest and highest speeds ($P < 0.001$). The ActiGraph was the most accurate at estimating total EE during walking and jogging [21]. This would make sense, considering that the Freedson equation was developed on treadmill walking and jogging.

In 2006, Crouter et al. [23], developed a new method to estimate EE from accelerometer activity counts. Unlike previous investigators, and based on the variability of activity counts observed among activities, Crouter et al. [22], developed their regression equation based on the coefficient of variation (CV = standard deviation / mean) and hypothesized that by calculating the CV for six 10-s epochs within a 1-min period, they could distinguish walking and running from all other activities. Forty-eight participants completed at least one of three routines specified that included low, moderate and vigorous intensity activities while wearing a Cosmed K4b^2 portable indirect calorimetry device and an ActiGraph on the right anterior axillary line.

Investigators achieved their purpose and established that for activities with a CV $\leq$ 10 (e.g. walking and running) an exponential curve was better suited to estimate EE. Meanwhile, for activities with CV > 10 (e.g. lying, washing dishes, raking leaves, etc.) a cubic curve was found to be a better fit. Overall, the regression equations developed by Crouter et al. had a significant correlation with the measured METs ($r = 0.96$, SEE = 0.73; $p<0.001$) and was within 0.75 METs of the measured values for all 17 activities, which was not significantly different from actual METs for any activity, or for all activities combined [23]. In 2010, this method was modified to provide even closer estimates of energy expenditure throughout all activities performed [25].

Crouter et al. [22] and Rothney et al. [36] both compared ActiGraph regression equations that were developed over more than a decade of research (1997 – 2006). In general, they concluded that one equation is unable to estimate EE for all activities accurately, and for the most part, equations developed to measure EE during walking are not accurate for most other activities [22, 36].

Considering the number of devices available and the different versions of each (e.g. 7164, GT1M), it is important for researchers to determine if the different monitors continue to be reliable and valid. Investigators are beginning to test the different generations and different models of these devices to further examine if differences exist among them [29, 31-32, 34]. Fudge et al. [31] were the first to elucidate that differences existed between the GT1M and the 7164. In their study, investigators measured the activity counts of the GT1M, 7164 and a triaxial accelerometer (3dNX model, BioTel Ltd., Bristol, UK) in endurance-trained individuals who completed two exercise tests on the treadmill. Although linear relationships were observed for all activity monitors during walking, the uniaxial accelerometers (GT1M and 7164) plateau during running, while the triaxial accelerometer increased linearly with increases in speed up to 20 km·hr^{-1} [31]. The authors concluded these differences were probably due to the biomechanics of running. At higher speeds, a leveling-off of vertical accelerations occurs, limiting the accelerations measured by the uniaxial mechanisms within these devices. However, differences in the GT1M electronics (increases in sampling rate and a wider band-pass filter) allowed the device to plateau at a higher speed than the 7164 (14 – 16 vs. 10 – 12 km·hour^{-1}) [31].

Rothney et al. [32] compared the performance of three generations of ActiGraph on a mechanical system (models 7164, 71256, and GT1M). This study used mechanical oscillations to determine the dynamic response and reliability of the three monitors and found significant differences among the three. Analysis of intermonitor CV revealed all three generations demonstrated high CV values at the lower frequencies (>20%), while at frequencies above 40 rpm, the GT1M had lower CVs compared to the 7164 and 71256. Intermonitor CV was low (0.55%) for all frequencies above 40 rpm [32]. Based on these findings, investigators concluded the GT1M has undergone some changes in either device sensitivity, or the filtering approach. However, the lower intermodel CV values represent an improvement over previous monitors [32].

Similarly, Fudge et al. [31], and John et al. [34] demonstrated the limitations of uniaxial activity monitors, when measuring running intensities, with activity counts peaking at 14 $km \cdot hour^{-1}$. More significant, however, was John and colleagues' findings that the activity counts obtained from four generations of the ActiGraph activity monitor, 7164, GT1M-V1, GT1M-V2, and GT1M-V3, were not statistically different while walking or running. This suggests that researchers interested in measuring physical activity could use any of four available versions of the ActiGraph, and could adequately compare the data among studies [34].

According to the ActiGraph manufacturer, the GT1M and GT3X models are comparable in their technology, except the GT3X has a triaxial, instead of a uniaxial accelerometer [11]. As of May 2010, there were no published studies comparing the two monitors or suggesting a benefit of one over the other.

Actical

The Actical (Phillips Respironics, Bend, OR) is considered an “omni-directional” device, capable of recording in multiple directions, although is most sensitive in the vertical plane [37]. This device is the smallest device available (2.8 x 2.7 x 1.0 cm, and it weights 17 g) and has the capability to record up to 44 days of data time in 1-minute epochs. Even though the device can be worn on multiple sites (i.e. wrist, hip, or ankle) with the use of hook and loop straps, the hip is the preferred wearing site. Unlike the ActiGraph capability to return 1-second epochs, this devices smallest sample interval is 15 seconds. The Actical detects low frequency accelerations in the range of 0.5 – 3.2 Hz, and g-forces (0.05 – 2.0 Hz) common to human movement [37]. The accelerations recorded by the internal mechanism generate an analog voltage that is filtered, amplified, and digitized through an A/D converter at 32 Hz. The device is initialized and downloads data through a serial port reader (Actireader), which allows the device to be completely waterproof [38]. Data from the device is downloaded to a personal computer as a ‘.csv file’, which can later be manipulated through spreadsheet software (e.g. Excel).

Similar to the ActiGraph, the Actical has been subject to various reliability and validity studies [36, 39-41]. Currently, Canada is using the device as an objective measure of physical activity through the incorporation of the device in the Canadian Health Measures Survey, which was developed to collect health information of a representative sample of the Canadian population [42-43].

In 2006, Esliger and Tremblay [40] conducted the first mechanical study to assess the validity of multiple accelerometers simultaneously. They compared three commonly used accelerometers, the Actical, ActiGraph (7164) and the RT3 (Stayhealthy, Inc., Monriva, CA), using a hydraulic shaker table. In this manner, investigators were able to manipulate the magnitude of the acceleration as well as the frequency of oscillation, which are the two key variables that contribute to the accelerometer's output [40]. Investigators mounted 15 monitors (five of each model) to the surface of a shaker table, positioned vertically along their sensitive axis to maximize and standardize the output of the piezosensor [40]. When comparing the accelerometer output in counts, the Actical had better intra-instrument reliability than the ActiGraph and RT3 (CV = 0.4%, 4.1% and 46.4%, respectively). However, the ActiGraph had better inter-instrument reliability than the other two accelerometers, with a CV of 4.9%, compared to 15.5% for the Actical, and 42.9% for the RT3 [40].

In a secondary comparison of a larger number of devices, somewhat similar findings were observed. When 39 Actical devices were tested, investigators found relatively stable intra-instrument reliability compared to the first experiment (CV = 0.50% vs. 0.40%). Inter-instrument reliability however was markedly improved (CV = 4.02% vs. 15.5%). The authors suggested the improvement in the Actical inter-instrument reliability of the second experiment compared to the first, was due to a improvement in inter-device calibration by the manufacturer, considering that the devices used in the first experiment were not from the same lot than those used in the second experiment [40]. Although these findings seem to suggest that the Actical is a more reliable accelerometer-based activity monitor, Esliger and Tremblay's findings are not in agreement with a previous study by Welk et al. [44], which showed the Actical to have the poorest reliability (CV = 20%) when individuals walked on a treadmill at 80 $m \cdot min^{-1}$. In light of these discrepancies and considering the differences found in their study among the devices sampled, Eslinger and Tremblay suggested these inconsistencies might be due to calibration changes done by the manufacturer after becoming aware of Welk et al.'s finding two years earlier.

Paul et al. [41] sought to compare the outputs of the Actical and ActiGraph in a group of fifty-six men and women while wearing both devices for 15-days. In this study of relatively healthy adults, the Actical recorded significantly lower activity counts compared to the ActiGraph ($P < 0.0001$), even though a strong correlation existed between the two devices ($r=0.90$; $P < 0.0001$). The authors suggested the reason for these discrepancies was due to the different A/D conversion filters used by each manufacturer (i.e. 10 Hz ActiGraph vs. 32 Hz Actical) [41]. In order to compare the devices, investigators performed a log-transform of the raw data and found a decreased in the coefficient of variation (15.5% vs. 3.1%), suggesting a way by which the outputs of each device could be compared ($r= 0.90$; $P < 0.0001$). Based on the result of this transformation, investigators developed two regression equations to convert raw outputs from one device to another and concluded that although raw outputs from these two accelerometer-based activity monitors are not comparable, the use of their equation could help compare the activity outputs for each device [41]. However, this conversion should be used cautiously as the participants in this study wore the devices for approximately 17 hours per day, thus the applicability of these equations is limited to studies that have at least 17 hours per day of wear time [41].

Considering that the raw outputs of accelerometers do not provide an easy-to-understand measure of physical activity (i.e. counts per minute), investigators have begun to use the simple measure of a “step counts” as a way to measure physical activity since it provides a stable measure of ambulation [45]. Therefore, using a dual-mode accelerometer would provide investigators the opportunity to obtain both measures, counts per minute and steps per minute. The Actical has been recently modified to provide both measures; therefore, Esliger et al. [39] sought to assess its validity by conducting a technical assessment using a mechanical shaker and a practical assessment using direct observation of steps taken during treadmill walking at three different speeds. They further wanted to assess how the Actical compared to a previously validated dual-mode accelerometer (i.e. ActiGraph 7164) [28].

Technical assessment on the shaker table demonstrated a perfect correlation (r= 1.00) and a very low intra- and interdevice coefficient of variation (CV < 0.1%) between the steps per minute detected by the Actical and the shaker plate oscillations per minute. Therefore, the authors concluded that the addition of the step function to the Actical did not negatively impact its accuracy [39]. The practical assessment included thirty-eight adult volunteers (16 males and 22 females), who walked on a treadmill at three preselected speeds (50, 83 and 133 $m \cdot min^{-1}$) at 0% grade for six minutes while wearing eight Actical and eight ActiGraph 7164 accelerometers in the mid-axillary line on the right and left hip respectively. The comparison of mean step counts per minute recorder by the Actical was only significantly different from the actual steps at 50 $m \cdot min^{-1}$ (88 vs. 95 steps per minute; $P < 0.001$). At the two faster speeds, significant differences were not observed between methods. Similar findings were reported for the ActiGraph 7164 (90 vs. 95 steps per minute at 50 $m \cdot min^{-1}$; $P < 0.001$). More significantly however was their finding of 7.4% difference at the slowest speed, which although small can result in significant underestimations over 24-hours [39]. Based on these findings, Esliger et al. concluded that the step function of the Actical was accurate at speeds used by most healthy individuals for ambulation [39].

The Actical has also been used to measure energy expenditure. Since the accelerometer outputs are proportional to the amount of energy expended during activity investigators can estimate EE through the use of regression equations. Thus, activities with higher activity counts would result in higher EE, while activities with lower activity counts will have lower EE.

In 2006, Heil [37] developed a number of equations to estimate activity EE from the Actical acceleration outputs while performing every day activities (e.g. typing, hand writing, floor sweeping, carpet vacuuming, slow and moderate treadmill walking and jogging). Participants included children and adults performing a number of sedentary, light, moderate and vigorous intensity activities while wearing a portable metabolic system and Acticals on the wrist, hip and ankle. When developing the equations, Heil assumed differences existed among the activities performed and used certain activity count cut-offs to illustrate these changes in intensity among the activities performed. Therefore, one of two regression equations could be used depending on the activity counts recorded by the accelerometer.

In general, for children the two-regression models were most accurate when estimating energy expenditure, although it tended to over predict activity EE for the moderate intensities by and under predict activity EE vigorous intensities by 10 and 9 kcals, respectively [37]. For adults, the single and double regression equations developed for the hip monitor resulted in the best estimate of EE (r^2= 0.75, and 0.85, respectively) for all activities regardless of intensity. However, the single regression model had a tendency to over predict most variables, while the two-regression model was more accurate and had a lower tendency to over predict [37]. Although good agreement was found between activity EE and the equations developed, Heil concluded that further research to validate these algorithms under free-living conditions should be considered, as all the activities performed in this study were done in a laboratory environment [37].

More recently, Crouter and Bassett [46] developed a new regression algorithm to predict energy expenditure using the Actical activity outputs. Unlike Heil who used count per minute cut-points to distinguish between the types of activities performed, Crouter et al. used the coefficient of variation (CV) of ambulatory and lifestyle activities. They observed that ambulatory activities (walk/run) have less variation compare to other lifestyle activities, which may result in more intermittent activity. For walking/jogging activities, which resulted in a CV $\leq$ 13% an exponential regression line demonstrated a better fit compared to a linear regression (r^2= 0.895; SEE = 1.051 vs. r^2= 0.912; SEE = 0.149) [46]. For activities resulting in a CV > 13%, such as lifestyle activities the relationship between Actical counts and the intensity of exercise was best described by a cubic curve (r^2= 0.884; SEE = 0.804).[46]. In addition, the Crouter et al. Actical regression model also demonstrated accuracy when estimating MET levels and time spent in light, moderate and vigorous activity [46]. In light of these findings, Crouter et al. concluded this new regression model improved upon previously published equations [37, 47-48].

In 2008, Rothney et al. [36] compared the predictive performance of Actical prediction equations (Heil's single regression and 2-regression models) and compared them to room calorimetry. Eighty-five adults were asked to stay over-night in a room calorimeter while the device recorded minute-by-minute activity data. Participants performed a number of ambulatory (e.g. walking and jogging) and sedentary activities (e.g. desk work) for 10 minutes, followed by 10 minutes of rest. While individuals were not performing any of the activities prescribed, they were asked to engage in their normal activity patterns within the room calorimeter, which was equiped like a one-bedroom apartment [36]. When comparing the physical activity levels measured by room calorimetry and the predicted values of the Actical, the Actical single regression tended to significantly overestimate sedentary activity (1 – 1.5 METs) by 15%, and vigorous intensity (> 6 METs) activities by 20%; while underestimating light intensity activities (1.5 – 3 METs) by 77% [36]. No significant differences were seen between the room calorimetry and the single regression equation at moderate intensity activities (3 – 6 METs). The 2-regression model, significantly overestimated sedentary, moderate and vigorous activities (14%, 66%, and 22%, respectively) while significantly underestimating light intensity activity (80%). Investigators concluded that each regression equation had its strength and weaknesses; therefore, neither equation is superior in all instances.

ActivPAL

The ActivPAL (PAL Technologies Limited, Glasgow, UK) is another physical activity monitor that has been used in PA research since 2002. This device contains a uniaxial accelerometer and it is much smaller than the ActiGraph, but slightly bigger than the Actical (3.5 x 5.3 x 0.7 cm; 20 g). Unlike either the ActiGraph and the Actical, the ActivPAL is placed on the mid-line of the thigh, about a third of the way down between the hip to the knee [49]. The manufacturer recommends the use of PALstickies, which are "self-adhering, removable and hair friendly" to secure the device to the thigh; however other hypo-allergenic medical tape/dressing can be used (e.g. 3M Tegaderm dressings) [49]. The ActivPAL uses a uniaxial accelerometer and responds to changes in gravitational acceleration as well as acceleration resulting from segmental movement to sense limb position and activity. This allows for the reliable discrimination of periods of upright activity from seated or lying activities through proprietary algorithms [49]. The accelerometer has a range of 0 – 2 g, records at a frequency of 10 Hz, and is able to record steps, cadence and time spent sitting, standing and walking through proprietary algorithms [49].

The ActivPAL has a 4 MB memory capacity that allows recording in excess of 7 days of activity (maximum recording period is dependent on the activities performed) [49]. The device uses a rechargeable lithium polymer battery that is charged through an USB port when connected to the docking station and will recharge to “full” in less than 2 hours after 7 days of use, or in less than 10 minutes after 24 hours of use [49].

Several investigators have established the validity of the ActivPAL through the use of direct observation [50-52], or the use of discrete accelerometers [53]. Overall, the ActivPAL has been shown to have good reliability with intraclass coefficients (ICC) > 0.90, which is considered excellent [54].

In 2006, Ryan et al. [50] were the first to report on the validity and reliability of the ActivPAL among a group of healthy adults. In this study, participants were asked to walk on a treadmill at five preselected speeds (54, 67, 80, 94, and 197 $m \cdot min^{-1}$) and on an outdoor course at three self-selected speeds (slow, normal and fast) while wearing four ActivPAL’s. All trials were recorded with a video camera, which served as the criterion method to measure steps. Interdevice reliability for number of steps taken for all treadmill-walking speeds and outdoor walking speeds were nearly perfect (ICC = ≥ 0.99) [50]. In addition, percent of agreement for steps taken between the ActivPAL and direct observation was < 1% for all treadmill speeds and less than 0.02% for the outdoor walking. Considering the accuracy of the ActivPAL to measure steps at various speeds, the authors concluded that the ActivPAL could serve as a good tool to monitor the activity patterns of people with normal walking patterns as it is unaffected by walking speed [50].

Considering the claims by the manufacturer that this device is able to accurately monitor the time one spends sitting, standing and walking, Grant et al. [51] sought to evaluate the validity and reliability of this device in a controlled environment and free living condition (performing activities of daily living) compared to a criterion method (video observation). When comparing the device to the criterion method, the ICC was > 0.97 for all postures (sitting, standing and walking) in both the controlled and the free-living condition. When comparing the interdevice reliability, all of the ActivPAL performed similarly (ICC > 0.99) for all conditions, except for walking in the free-living environment (ICC >0.79). In addition, investigators also found that the transitions between sit-stand and stand-sit were identical when compared to the video observation [51].

Moreover, when analyzing the percent of agreement between the device and the criterion method, the device was accurate estimating postural changes 96% of the time when the controlled and free-living condition were combined (98.5% and 93.6%, respectively). The difference was due to a decreased sensitivity of the devices when estimating standing and walking during the free-living condition [51]. Investigators suggested that these discrepancies might be due to how the activities were performed during the free-living condition, where participants performed walking with short time intervals of standing between them. The observer correctly identified these periods as walk/stand periods; however the ActivPAL interpreted these periods as one long walking interval and thereby leading to an overestimation of walking periods and an underestimating of standing intervals compared to the observer [51]. Researchers did not consider this a limitation of the device; instead it was considered a limitation of the study design [51]. Thus, concluding that the ActivPAL is a valid and reliable device to monitor postural changes throughout the day.

In 2007, Godfrey et al. [53] compared the ActivPAL to a discrete accelerometer-based system to determine its accuracy while performing activities of daily living (ADL's). When comparing the percent differences in time measured between the two devices, the ActivPAL showed the biggest difference for stepping (1.64%), followed by standing (0.50%) and sitting (0.06%). When comparing the direct time recording comparisons, which give a good representation of the ActivPAL's accuracy, the ActivPAL averaged 98% for all three conditions. Therefore, investigators concluded that in a population of healthy adults, the ActivPAL accurately estimated 98% of the activities perform during a six-hour period [53] .

More recently, Grant et al. [52] evaluated the accuracy of the ActivPAL to measure steps in a group of community-dwelling older adults. Twenty-one participants, aged 65 to 87 years who were participating in community-based exercise classes and did not use a walking aid took part in the study. Each person was asked to walk on a treadmill at five preselected speeds (40, 54, 67, 80, and 94 $m \cdot min^{-1}$) and around a 500-meter course at three self-selected speeds (slow, normal and fast). Each trial was recorded on video and this served as the criterion method. Overall, the mean difference between the ActivPAL and the observer was less than three steps (0.6%). Furthermore, the absolute percent error was less than 1% between the two methods, thereby allowing authors to conclude the ActivPAL is an accurate and reliable device to measure ambulation among community dwelling older adults.

PEDOMETERS

The use of pedometers can be dated back to Leonardo DaVinci [55]; and it is believed that Thomas Jefferson introduced it to the U.S., after traveling to France [56]. Previous research on pedometers in the 1970's and 1980's did not find them suitable for research due to large errors [57-58]; however, most pedometers today use an electronic circuit that responds to changes in vertical accelerations when a person walks. Pedometers are designed to measure ambulatory activity and provide a user-friendly output measure (i.e. a step count). Traditional pedometers are worn at the waist and house a spring-suspended lever arm, which moves up and down as a result of vertical accelerations produced at the waist during walking [17, 59]. A step is recorded when an acceleration above a manufacturer-design threshold (i.e. 0.35 g for the

DigiWalker) causes the lever arm to move up and down opening and closing an electrical circuit [17, 59]. Pedometers are designed to provide immediate feedback to the end-user in the form of accumulated steps during walking. However, they are unable to record below or above a certain "threshold", thus limiting their accuracy at slower (e.g. shuffling) or faster (e.g. running) speeds [12]. Pedometers cannot provide accurate estimates of PA energy expenditure [17]; nor can they detect non-ambulatory activity (e.g. swimming, weight training, or cycling) [17]. Most pedometers are unable to store data to measure habitual physical activity [12] (although new advances in microchip technology have allowed the implementation of on-board memory functions to recall previous day activity [59]). Despite these limitations, pedometers provide a small, easy to use, and inexpensive way to promote physical activity that has been shown to be effective [60-61].

DigiWalker

The DigiWalker (Yamax Corp., Tokyo, Japan) is a small (5.2 x 3.9 x 1.9 cm) and inexpensive pedometer that has been used extensively in research [62-66], although some limitations exist [67-69]. Bassett et al. [62] investigated the accuracy of five different electronic pedometers for measuring steps and distance on a sidewalk, a rubberized track and on the treadmill. After completing a 4.88 km (3.03-mile) walk on a sidewalk course, investigators found that the Digiwalker (DW-500) recorded distance walked with the most precision. In addition, good agreement was shown between pedometers placed in opposite locations (i.e. right and left hip). Although two other pedometers (Pacer and Accusplit) showed close estimates of distance walked, placement of the device on one hip differed from the other. When comparing the effects of the walking surface, no significant differences were seen. Finally, when they assessed the effect of speed, the DW-500 was the most accurate at the slow-to-moderate speeds

(54 – 80 m·min^{-1}) recording about 75% of actual steps. At the faster speed all devices seemed to be within 10% of actual steps [62].

In 2003, Crouter et al. [63] followed up on Bassett et al.'s [62] 1996 findings considering the advancement made in pedometer technology. In this study, Crouter and colleagues investigated the validity and reliability of ten electronic pedometers for steps taken, distance travelled, and energy expenditure at various walking speeds on a treadmill. Overall, they found that all electronic pedometers tended to underestimate actual steps at 54 and 67 m·min^{-1}, and several pedometers were within 1% of actual steps at speeds of 80 m·min^{-1} or above (Yamasa Skeltone EM 180, Omron HJ-105, Digiwalker SW-701, New Lifestyles NL-2000, Kenz Lifecoder, and Walk4Life LS 2525). Only the DigiWalker SW-701 was found to not differ from actual steps at any speed ($P > 0.05$) [63]. Of the ten devices included in the study, six-displayed distance traveled and all tended to overestimate distance at the slowest speeds, and underestimate at the faster speeds. In addition, the analysis of estimated EE from each pedometer showed significant overestimation of actual values of gross and net EE during treadmill walking.

Based on these findings, Crouter et al. [63] concluded that at slower speeds, the vertical accelerations acting on the waist are below the threshold (i.e. 0.35 g, for the DigiWalker) needed to record a step. Additionally they showed pedometers are most accurate in predicting distance traveled at speeds above 80 m·min^{-1}. Similarly to Bassett et al.'s [62] findings, Crouter and colleagues concluded that the DigiWalker SW-701 was the most accurate electronic pedometer for predicting steps, distance, and gross kilocalories for walking [63].

Schneider et al. [65] sought to determine the accuracy and reliability of ten electronic pedometers for measuring steps. Each participant completed two trials around a 400-meter track with two pedometers of the same model each time worn on opposite sides of the hip. No significant differences were found between the pedometers on opposite sides ($P > 0.05$). Two models, the Oregon Scientific PE316CA and the Sportline 330 significantly underestimated and overestimated actual steps taken, respectively. Although no other device demonstrated any other significant difference from actual steps, investigators demonstrated that the New-Lifestyle NL-2000, the Yamax DigiWalker SW-701 and the Kenz Lifecorder were most accurate (within ± 3%) [65]. Moreover, of these three devices the Yamax had the least error and highest intramodel reliability (>0.99). The authors further pointed out that the level of error for these three devices all met the Japanese Industrial Standard set by the Ministry of Industry and Trading regulations (± 3%) [70].

In a similar study, Schneider et al. [64] evaluated the accuracy of thirteen-models of electronic pedometers, over a 24-hour period using the DigiWalker SW-200 as the criterion method. The SW-200 used in this study is similar to the SW-701 used by Schneider et al. [65], the only differences are the functions displayed. Similar to Schneider et al.'s [64] finding during a 400 m walk, the Kenz Lifecoder, DigiWalker SW-200 and the New Lifestyles NL-2000 demonstrated the smallest percent difference in the free-living environment over a 24-hour period. The authors concluded that the discrepancies among pedometer models might be due to sensitivity thresholds for the magnitude of vertical acceleration that would trigger the recording of a step among different devices [64]. In addition, the filtering mechanism of each device may lead to underestimation of the actual number of steps taken. Some of the devices used in this study (e.g. Sportsline 345, Accusplit Alliance 1510 and Freestyle Pacer Pro); in addition to more recent ones (i.e. Omron HJ-720 ITC), do not record a step immediately when someone begins to walk [64, 71]. Instead, they have a filtering mechanism that will not count a step until a person accumulates four steps or four seconds of activity have elapsed, depending on the model. This represents a significant limitation, considering that recent research has shown the most common walking bout includes 4 ± 1 steps in a row [72], which might include most household (e.g. laundry, dishes, etc) or occupational (e.g. office work) activities. In addition, they explained that abdominal adiposity might play a role in the accuracy of these electronic devices, as significant levels of abdominal adiposity among obese individuals could affect how the device is positioned on the waist (e.g. tilt forward), which would reduce the magnitude of accelerations recorded at the waist.

Several investigators [66-68] have tried to determine if this final point made by Schneider et al. [64] was correct. Swartz et al. [66] used the same model pedometer used by Schneider et al. [64], the DigiWalker SW 200, to determine the accuracy of the device among a group of apparently healthy men and women with different degrees of adiposity. Their findings did not indicate a BMI effect at any of the five-speeds tested (54 $m \cdot min^{-1}$, P = 0.991; 67 $m \cdot min^{-1}$, P = 0.0556; 80 $m \cdot min^{-1}$, P = 0.591, 94 $m \cdot min^{-1}$, P = 0.426; and 107 $m \cdot min^{-1}$, P = 0.869) [66]. When investigators looked at the effect of placement of the pedometer on the waist (i.e. front, side, or back), step count accuracy was reduced at the slowest speeds (i.e. 54 and 67 $m \cdot min^{-1}$) up to 20% for the front pedometer, up to 33% for the side pedometer and up to 26% for the back pedometer. At the fastest speeds (80, 94 and 107 $m \cdot min^{-1}$) differences in steps counts or placement were not seen ($p>0.05$). These findings are in agreement with those of Bassett et al. [62], Crouter et al. [63] and Schneider et al. [65] who demonstrated significant underestimation of actual steps taken for electronic pedometer at speeds less than 80 $m \cdot min^{-1}$.

Melanson et al. [67] tested the accuracy of the DigiWalker SW-200 in a large sample of adults (N=259) of varied ages and body weights. Overall, their findings were in line with those previously reported [62-63, 65-66], where the accuracy of the DigiWalker was directly proportional to the walking speed. However, when participants self-selected their walking speeds to "normal walking" and "brisk walking" significant decreases in walking speed were observed with age ($P < 0.001$) at the "normal walking speed" (2.9 ± 0.4 for 18 – 30 years vs. 2.3 ± 0.4 for > 70 years) and at the "brisk walking speed" (3.8 ± 0.4 for 18 – 30 years vs. 2.8 ± 0.5 for > 70 years). When comparing BMI groups at the self-selected "normal walking speeds" regardless of BMI classification the DigiWalker significantly underestimated steps taken by 6% - 12% ($p<0.01$). At the "brisk speed" the DigiWalker only underestimated for the obese groups ($P < 0.01$) [67].

In a second part to their study, Melanson and colleagues [67] compared the accuracy of two electronic spring-levered pedometers (Walk4Life LS-2500, and Step Keeper HSB-SKM) with a piezoelectric device (Omron HF-100) at three preselected walking speeds (27, 48 and 54 $m \cdot min^{-1}$). The findings were consistent with their first study, with all pedometers significantly underestimating at the slowest speed and improving in accuracy at the faster speeds. The accuracy of the HF-100 was greater than the Walk4Life LS-2500 and Step Keeper HSB-SKM at 26.8 $m \cdot min^{-1}$ (56.4 ± 33.8% vs. 7.5 ± 16.3% and 20.5 ± 28.4%, respectively) and 48 $m \cdot min^{-1}$ (97.8 ± 9.6% vs. 52.1 ± 38.7% and 73.4 ± 36.7%, respectively). At 54 $m \cdot min^{-1}$ no significant differences were observed among each device; and greater than 90% accuracy was achieved at self-selected speeds (56.3 ± 13.4 $m \cdot min^{-1}$). Based on these findings, investigators concluded that piezoelectric pedometers may be better equipped to measure the lower accelerations experienced at the slowest speeds, and may be more suitable for special population that ambulate at these slower speeds (i.e. elderly, obese) [67].

Unfortunately, Melanson et al. [67] did not include the DigiWalker on the final comparison between the piezoelectric pedometers, which would have been relevant considering the DigiWalker is one of the most widely used pedometers. Therefore, Crouter et al. [68] examined the effects of several anthropometric measures (i.e. BMI, waist circumference) and tilt angle on the accuracy of the DigiWalker and the New Lifestyles NL-2000 (NL 2000) during treadmill walking and in a 24 hour period among overweight and obese individuals. Their findings clearly indicate the differences between devices when measuring actual steps. Overall, the NL-2000 outperformed the DigiWalker at all walking speeds (54, 67, 80, 94 and 107 $m \cdot min^{-1}$) recording greater percentage of actual steps ($P < 0.05$). The DigiWalker's accuracy was inversely related to BMI and waist circumference. Most significantly however was the effect of tilt angle, which resulted in underestimations of up to 60% at the slowest speed and about 40% at 94 $m \cdot min^{-1}$ in those with tilt angles greater than 15 degrees [68]. Therefore, authors concluded that the primary factor affecting the DigiWalker's accuracy was pedometer tilt, as a result of increased abdominal adiposity. Moreover, these findings support Melanson et al.'s [67] finding and confirm the enhanced capabilities of piezoelectric pedometers for measuring ambulatory activity among overweight and obese individuals.

Tudor-Locke et al. [27] and Le Masurier and Tudor-Locke [28] both compared the accuracy of the DigiWalker to a dual-mode accelerometer (ActiGraph, model 7164) during controlled and free-living conditions. For the free-living condition, the DigiWalker steps were significantly correlated with the ActiGraphs counts per minute per day ($r = 0.74$; $P < 0.0001$), total counts per day ($r = 0.80$; $p < 0.0001$), and ActiGraph steps per day ($r = 0.86$; $P < 0.0001$). Nonetheless, a significant difference in steps were seen between devices during free-living activities, with the ActiGraph recording 1845 ± 2116 more steps than the DigiWalker ($P < 0.0001$) [27]. Unlike Tudor-Locke et al.'s [27] findings in the free-living environment, Le Masurier and Tudor-Locke only found significant differences between devices while walking on a treadmill for five minutes at a slow speed (54 m·min^{-1}) [28]. The DigiWalker detected over 96% of steps taken at speeds over 67 m·min^{-1}, whereas only 75% of steps taken were recorded at 54 m·min^{-1}. The ActiGraph recorded over 98% of steps taken at each speed [28].

Step Activity Monitor

The Step Activity Monitor, or Step Watch-3 (SW), is a relatively new kind of activity monitor that uses an accelerometer and an electronic filter inside a polycarbonate case to measure ambulatory activity. The device is a small (7.0 x 5.0 x 2.0 cm, 38 g), waterproof, self-contained device that attaches to the ankle above the lateral malleolus on the right leg, or the medial malleolus on the left leg, through the use of an elastic strap . The SW continuously records steps during a user specified period of time at specific intervals (epochs). The minimum sampling interval is 6 seconds allowing for a total of 1.12 days of ambulation data. The maximum sampling interval is 25.5 min, providing for 285.6 days of continuous monitoring [73]. At one-minute epochs, the SW can store step data for up to 2 months before requiring data to be downloaded [74]. The SW uses a docking station that plugs into a computer through a standard USB port for programming and downloading of data. The monitor and dock communicate through an infrared link, which allows the SW to be completely sealed [74]. Sensitivity to movement, frequency with which steps are detected, and acceleration required to record a step are all adjusted through the software provided by the manufacturer.

The SW has been shown to be a highly reliable device in the measurement of ambulatory activity among healthy individuals [75-76], the obese [77], the elderly [69, 78], amputees and diseased populations [73, 79]. Through a series of case studies, Coleman et al. [73] demonstrated the accuracy of the SW among a group of diabetics or individuals with lower-limb amputations after walking over level ground at a self-selected speed, uphill and downhill at 9%

grade, and up and down stairs. Overall, the SW accurately counted 99.7% of steps taken, 98.8% for normal walking, and 96.2% for walking on stairs [73]. Investigators concluded that the SW was a highly accurate, reliable, instrument that can be used to perform long-term step monitoring on a wide range of subjects and activities [73].

In 1999, Shepherd et al. [77] provided the first evidence on the impact of obesity on the accuracy of a waist-worn activity monitor. In this study, investigators compared an ankle-borne device versus a spring-levered pedometer worn at the waist. Overall, the SW had an absolute error of less than one percent (0.54 ± 0.7%), where as the pedometer had close to 3% error (2.82 ± 3.8%). When looking at each individual activity, the pedometer performed best during the 400-m walk with a mean error of 2.30 ± 3.8%. However, when going upstairs the percent error was nearly 20% (19.9 ± 21.3%). On the other hand, the SW showed less than 1% error for the 400-meter walk, and less than 5% error during the stair ascent (0.31 ± 0.7%, and 3.58 ± 5.2%, respectively). Furthermore, the highest absolute error for the SW was seen during 10-meter walk (5.25 ± 5.7%) and stair descend (7.25 ± 11.6%). However, these values were well below the pedometer estimates (15.5 ± 16.2% and 10.8 ± 10.8%, respectively) [77]. When the participants were divided into those with a BMI below 30 $kg{\cdot}m^{-2}$ (N=21 for each device) and over 30 $kg{\cdot}m^{-2}$ (N=8 for each device) the errors were more significant. Among those with BMI less than 30 $kg{\cdot}m^{-2}$ both the SW and the pedometer were very accurate (0.6 ± 0.7% vs. 1.6 ± 1.4%, respectively). However, the pedometer had more than a 5% higher error than the SW for those with a BMI over 30 $kg{\cdot}m^{-2}$ (6.1 ± 6.0% vs. 0.5 ± 0.5%). Using univariate regression analysis, investigators demonstrated that the magnitude of pedometer error was significantly related to BMI ($r = 0.792$; $P < 0.001$) and weight ($r = 0.753$; $P < 0.001$). Considering these findings, the researchers suggested that using a ankle-born device may be more appropriate for individuals with high levels of adiposity, as waist-borne devices may have limited accuracy [77].

In 2005, Foster et al. [75] and Karabulut et al. [76] both reported on the validity of the SW among healthy individuals. Foster et al. [75] recruited twenty lean (BMI < 25 $kg \cdot m^{-2}$) and obese (BMI > 30 $kg \cdot m^{-2}$) participants to walk on a treadmill at three different speeds (27, 54, and 80 $m \cdot min^{-1}$) while wearing four activity monitors. A piezoelectric pedometer (Omron HF-100, Tokyo, Japan) on the left hip above the knee, a spring-levered pedometer (Accusplit Digi-Walker 2, San Jose, CA) on the right hip, and two SWs, one attached to the inside of the left ankle over the medial malleolus, and the other to the outside of the right ankle above lateral malleolus. The accuracy of all devices was compared to a gold standard (manual counting). Collectively, the accuracy of the SW across the three velocities was 99.7 ± 0.67%, with an intra-class correlation coefficient for SW and manual counts of 0.9995. The Omron HF-100 performed better than the Accusplit recording about 60% and 98% of steps taken compared to 20% and 80% of steps recorded by the Accusplit at 27 and 54 $m \cdot min^{-1}$, respectively. At 80 $m \cdot min^{-1}$, no significant differences were observed among any of the devices. In addition, comparing the data between lean and obese subjects did not show any significant differences among the devices or between the two groups. However, the SW was the most accurate and more precise, having less variability at all speeds [75].

The second aim of Foster et al.'s [75] study was to examine if energy expenditure could be estimated using the SW pedometer, considering that previous investigators had reported the inability of pedometers to accurately estimate walking EE [7, 63]. The researchers proposed two regression equations with good agreement (r^2= 0.89 and r^2= 0.81). When the calculated EE's were compared with the measured values, the calculated EE were within 11% of the measured values for all walking speeds. Therefore, investigators concluded that the SW was a precise and accurate device for measuring EE of walking among a range of different velocities for individuals of different body compositions. They additionally concluded that considering the device's output, which provides not only steps, but also a time stamp of the activity performed the SW could effectively measure EE and PA in the free-living environment, becoming a reliable tool to assess the efficacy of physical activity interventions.

Karabulut et al. [76] confirmed Foster et al.'s [75] findings and suggested that the SW could serve as a useful criterion tool when monitoring physical activity in the free-environment, where people perform activities at slower speeds. In their study, participants wore two waist mounted activity monitors (New Lifestyle (NL-2000) and DigiWalker SW 701 (SW-701)) and two ankle-borne devices (StepWatch-3 (SW) and AMP 331 (AMP)) while walking on a treadmill and during a 24-hour period. In addition, investigators wanted to find out how potential sources of error (e.g. leg swinging, heel tapping, stationary cycling and driving a car) would influence the accuracy of the devices.

For the walking condition, the SW was the most accurate of the devices used giving mean counts within 1% of actual steps at all speeds. The other pedometers (NL-2000, SW-701, AMP) tended to underestimate at the slowest speeds, with accuracy improving with greater speed [76]. During the 24-hour free-living condition, investigators found significant differences among the NL-2000, SW-701 and AMP recording up to 18% lower steps compare to the SW. Investigators suggested this underestimation by the other devices might be due to the SW ability to record higher percentage of actual steps during free-living conditions, such as slow walking or lifestyle activities [76]. When considering the effects of additional sources of error, such as foot tapping, cycling, etc., Karabulut et al. found that the SW is more sensitive to recording heel tapping and leg swinging compared to the other devices. Conversely, cycling or car driving did not affect the SW unlike the NL-2000, SW-701 and the AMP. Investigators suggested that these discrepancies should not affect the overall application of the SW considering that heel tapping and leg swinging might be a small percentage of an individual's daily activity [76].

More recently, Bergman et al. [69], Storti et al. [78], and Mudge et al. [79] have contributed to the SW literature by showing that among all devices tested the SW had the greatest accuracy among community dwelling older adults [78], older adults living in assisted-living facilities [69], and in patients after a stroke [79]. These findings are significant, and support the notion of the SW as a criterion method for measuring ambulatory activity. Waist-borne devices have shown limitations in terms of accurately measuring steps taken at slower walking speeds, and accurately measuring steps in overweight and obese individuals.

Nonetheless, the SW is not without limitations. In their study of adults after a stroke, Mudge et al. [79] demonstrated good agreement between the SW and more advance methods of measuring gait (i.e. 3-dimensional gait analysis (3-DGA) and footswitches. However, correlations were lower for the paretic limb ($r = 0.896$) compared to the nonparetic limb ($r = 0.959$) when comparing the SW to the 3-DGA, respectively. These findings further suggest the applicability of the SW as a reference criterion in walking research, even though some limitations may exist among individuals with limited mobility, due to a stroke or other neurological conditions.

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CHAPTER 3

EFFECTS OF BMI AND TILT ANGLE ON OUTPUT OF TWO WEARABLE ACTIVITY MONITORS

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ABSTRACT

Background: Accelerometer-based activity monitors have been used to provide objective measures of physical activity and energy expenditure (EE) in free-living individuals. However, output from these devices has not been compared among normal, overweight and obese individuals. **Purpose:** To examine the effects of body mass index (BMI) and device tilt angle on activity counts recorded by wearable monitors, in a controlled laboratory setting. A secondary aim was to examine the effects of these variables on estimated EE. **Methods:** Seventy-one healthy adults wore an Actical and an ActiGraph GT1Mon the right and left hip, respectively, while walking at 40, 67, and 94 $m \cdot min^{-1}$. EE was measured by indirect calorimetry and compared with estimated values using published equations. Three-way repeated measures ANOVAs were used to examine differences in outcome variables (activity counts and energy expenditure) between speeds, BMI and tilt angle for each device. **Results**: No significant differences in activity counts were observed among BMI categories for either the Actical or ActiGraph ($P > 0.05$). For the Actical, however, among those with an absolute tilt angle < 10°, the obese group recorded higher activity counts than the normal weight group ($P = 0.01$). Using the Heil 2-regression model, the Actical overestimated EE by up to 35% at the intermediate speed and up to 12% at the fastest speed ($P < 0.001$). The Freedson METs regression equation yielded closer estimates of EE than the Freedson Kcal regression equation. **Conclusion:** Our findings indicate that the Actical has limitations when comparing individuals with varying BMI

and tilt angles, in a controlled laboratory environment. The ActiGraph seems to be a more suitable device for making these comparisons.

Key Words: Walking, Actical, ActiGraph, Energy Expenditure

INTRODUCTION

Accelerometer-based activity monitors and pedometers are often used to provide objective measures of physical activity [1-3]. However, the effect of adiposity on these devices is unclear. Previous studies have reported inverse associations between objectively monitored physical activity and adiposity [1-6]. While it may seem logical that overweight and obese individuals are less likely to engage in activities throughout the day [7], this cannot be concluded with certainty. Accelerometer-based activity monitors and pedometers may be subject to errors that could hinder the ability to accurately measure physical activity in these individuals.

Previous studies have shown that spring-levered pedometers are less accurate in obese individuals than in normal weight persons [1, 4-7], but pedometers with an accelerometer mechanism are not subject to this limitation [8-10]. For example, Crouter et al. [8] examined the effects of adiposity on pedometers. They found that a spring-levered pedometer significantly underestimated ambulatory activity in obese individuals by up to 40%, and they suggested that adipose tissue and/or the device's tilt angle contribute to errors with this type of device. By comparison, the accuracy of an accelerometer-based pedometer was not affected by body mass index (BMI) or tilt angle.

It is important for researchers to understand the limitations of accelerometer-based activity monitors, as they are currently being used in large epidemiological studies in the U.S. [11], Canada [12], and Europe [13]. Therefore, it is important to know if these devices accurately measure physical activity patterns in overweight and obese individuals. Without this information, researchers cannot determine if measured differences among groups are artifact or truly representative of group differences in physical activity. To date, no study has closely examined the effects of BMI on the ActiGraph and Actical activity monitors. Thus, the purpose of this study was to determine if these devices are affected by BMI and tilt angle, when measuring physical activity in a controlled laboratory setting. A secondary aim is to assess the accuracy of various prediction equations among different BMI groups.

METHODS

Seventy-one adult participants (32 men, 39 women) volunteered to take part in this study. Participants were recruited across BMI categories to have an even distribution of participants among all three groups. All participants signed university-approved informed consent forms and completed a Physical Activity Readiness Questionnaire (PAR-Q) prior to testing. Participants were free of cardiovascular, pulmonary and metabolic disease as reported on the PAR-Q. They were instructed to arrive at the Applied Physiology Laboratory following a four-hour fast, and having refrained from moderate or vigorous exercise for the same period of time. The university's Institutional Review Board approved the study protocol.

Anthropometry

Participants, wearing light clothing and no shoes, had their height measured to the nearest 0.1 cm using a stadiometer (SECA, Corp., Columbia, MD). Body mass was measured to the nearest 0.01 kg on an electronic scale (Life Measurement, Inc., Concord, CA). The scale was calibrated according to the manufacturers' specifications prior to testing. BMI was calculated using the standard formula, body mass (kg) divided by height (m) squared. Abdominal (waist) circumference (WC) was measured on the skin at the level of the navel using a Gullick tension-gauged tape measure (Creative Health Products, Inc., Plymouth, MI) following established guidelines [14]. BMI and WC are both indicators of adiposity, and are strongly correlated with the percentage of body fat in the overall population [15].

Treadmill Walking

Prior to treadmill walking, two activity monitors were attached at the anterior axillary line, on opposite sides of an elastic belt, fastened to the participant's waist. We used an omnidirectional (Actical, Mini Mitter Co., Inc., Bend, OR) and a uniaxial (ActiGraph Model GT1M, ActiGraph, LLC, Fort Walton Beach, FL) activity monitor. Technical information regarding the two devices can be found elsewhere [16-19]. Briefly, the Actical is a small (28 x 77 x 10 mm) omnidirectional activity monitor weighing 17g and able to measure accelerations in the range of 0.5 to 3.0 Hz. The ActiGraph is a slightly bigger (38 x 37 x 18 mm) uniaxial activity monitor weighting 43g and capable of measuring accelerations in the range of 0.25 to 2.5 Hz.

Each device was initialized and synchronized using the software provided by the manufacturer (Actireader V. 2.10, Actilife Lifestyle Monitoring System, V. 3.3; for the Actical and ActiGraph, respectively). Both activity monitors have the ability to modify the physical activity collection times (epochs). In order to obtain the greatest time resolution, both devices were set to their smallest collection interval; 15-sec for Actical and 1-sec for ActiGraph and summed to obtain counts per minute ($\text{counts}\cdot\text{min}^{-1}$).

Participants were asked to walk on a motorized treadmill (Q65 Series 90, Quinton Instrument Co., Bothell, WA) for five minutes at three different speeds (40, 67 and 94 $\text{m}\cdot\text{min}^{-1}$). Prior to each stage, investigators used a protractor (Sears Craftsman) to measure the tilt angles of each activity monitor while mounted at the waist. A negative tilt angle indicated that the top of the device was closer to the body, whereas a positive tilt resulted from the top being further away from the body. Each subject was asked to straddle the belt prior to beginning each stage, and hold on to the handrails before starting to walk. Participants were instructed to release the handrails once they began walking. At the end of each stage, participants were asked to straddle the belt, in order to minimize activity at the end of the bout.

Indirect Calorimetry

To determine energy expenditure (EE) during exercise, we measured oxygen uptake via ParvoMedics TrueOne® 2400 Metabolic Measurement System (ParvoMedics, Sandy, Utah). The gas analyzer and flow meter were calibrated prior to each test. Throughout the test, oxygen uptake (VO_2), and EE ($kcal \cdot min^{-1}$) were recorded in 20-second intervals. All values were adjusted to reflect minute data and only the last two minutes of each stage were used for statistical analysis, in order to obtain values reflective of steady state exercise.

Statistical treatment

Prior to statistical treatment, all subjects were categorized by BMI into normal weight, overweight and obese groups following established cut points [20] . Tilt angles were used to indicate a deviation of the face of the wearable monitor away from the vertical plane. Tilt angles were measured with the participant standing up before each stage and the values were then averaged. If the absolute value of the tilt angle was less than 10 degrees, then we considered that the face of the wearable monitor was approximately vertical. If the absolute value of the tilt angle was greater than or equal to 10 degrees, then we considered that there was either a forward or backward tilt, either of which could affect the wearable monitor's output. Based on previous research from our laboratory [8], we decided to use a dichotomous variable (< 10 or ≥ 10 degrees) for the absolute value of the tilt angle. This dichotomous variable was used in all subsequent statistical analyses.

To determine differences in anthropometric measures among our sample, we conducted BMI comparisons using a one-way analysis of variance with Bonferroni adjustments.

Considering that a reference method to compare the output of accelerometer-based activity monitors is not currently available, we analyzed each device (Actical and ActiGraph) independently. The effects of BMI and tilt angle on activity counts at the three established speeds were analyzed using a 3-way repeated measures analysis of variance (ANOVA) with speed as the within-subject variable and BMI and tilt angle as between-subject variables. Post hoc analyses with Bonferroni adjustments were performed to further investigate the interaction between tilt angle and BMI.

Activity EE (AEE) by the Actical was calculated using the 2-regression model equation of Heil [21]. In order to compare the Actical prediction with measured values (ParvoMedics), we needed to obtain total EE (TEE). Therefore, resting EE (REE) was estimated using the Mifflin-St Jeor Equations (r^2= 0.71) [22]. The estimated REE was added to the AEE to determine TEE during the exercise bout (TEE = REE + AEE). For the ActiGraph, we used both equations developed by Freedson et al. [23]:

$$\text{TEE (kcal}\cdot\text{min}^{-1}) = (0.00094 \times \text{counts}\cdot\text{min}^{-1}) + (0.1346 \times \text{kg}) - 7.37418 \qquad \text{(Equation 1)}$$

and

$$\text{TEE (METs)} = 1.439008 + (0.000795 \times \text{counts}\cdot\text{min}^{-1}) \qquad \text{(Equation 2)}$$

Freedson et al. [23] showed that both equations are highly correlated with TEE during walking (r^2= 0.85) and running (r^2= 0.82), but most of their subjects were young, normal weight university students. Thus, we felt it was important to assess their accuracy among individuals of varying BMI. Given that 1 MET = 1 $\text{kcal}\cdot\text{kg}^{-1}\cdot\text{hr}^{-1}$ [24], we calculated all values and expressed TEE relative to body weight ($\text{kcal}\cdot\text{kg}^{-1}\cdot\text{min}^{-1}$).

The EE analysis was performed using repeated measures ANOVA. Post hoc analysis with Bonferroni adjustments were performed on the percent of actual values ((predicted / actual) x 100) to determine if each of the equations used over- or underestimated TEE and if differences existed among BMI groups. We did not intend to compare each of the prediction equations to each other. Thus, our analyses were done independently and compared each of the prediction methods to the criterion method.

Data were entered into Excel 2003 (Microsoft Co., Redmond, WA) and all statistical analyses were carried out using SPSS statistical software, version 17.0 for Windows (SPSS Inc., Chicago, IL). A significance level of $P < 0.05$ was chosen to denote statistical significance. All values are reported as mean ± standard deviation (SD).

RESULTS

Anthropometrics

Participant characteristics by BMI category are shown in Table 3-1.

Table 3-1: Physical characteristics of participants by BMI category (mean ± SD)

	Normal Weight (N= 28)	**Overweight (N=24)**	**Obese (N=19)**
Age (yr)	27.8 ± 8.0	34.6 ± 14.2	31.5 ± 11.1
Height (m)	1.71 ± 0.09	1.72 ± 0.12	1.71 ± 0.08
Weight (kg)	65.5 ± 9.8	80.3 ± 11.7‡	97.4 ± 9.3$^{‡ *}$
BMI (kg/m^2)	22.2 ± 1.9	26.9 ± 1.2‡	33.5 ± 3.5$^{‡ *}$
Waist Circumference (cm)	75.9 ± 12.7	85.3 ± 19.3	99.0 ± 10.5$^{‡ *}$

‡ Significantly different from normal group ($P < 0.001$)
* Significantly different from overweight group ($P < 0.001$)

Activity Counts

The Actical, as expected, showed significantly higher activity counts with progressively higher speeds ($P < 0.001$). In multivariate analysis, the interactions between speed and BMI, and speed and tilt angle were not significantly different (Table 3-2). However, an interaction was observed between speed, BMI, and tilt angle ($P = 0.034$). This was further analyzed by dividing the sample into those with absolute tilt angles less than 10 degrees, or greater than or equal 10 degrees (Table 3-3).

Table 3-2: Activity Counts for each device by BMI category (mean ± SD)

Device/Speed (m·min^{-1})		Normal Weight	Overweight	Obese
Actical	40	380 ± 200	412 ± 141	522 ± 216
	67	1419 ± 444	1500 ± 234	1594 ± 434
	94	2904 ± 583	3091 ± 372	3239 ± 604
ActiGraph	40	513 ± 288	581 ± 217	772 ± 363
	67	2154 ± 551	2165 ± 390	2206 ± 527
	94	3904 ± 797	3987 ± 741	4053 ± 695

Table 3-3: Absolute value of tilt angles for each BMI category (mean ± SD)

Device		Normal Weight	Overweight	Obese
Actical	**< 10 Degrees**	4.6° ± 3.6° (N=13)	6. 3° ± 2.7° (N=15)	4.0° ± 1.9° (N=9)
	≥ 10 Degrees	15.5° ± 4.8° (N=15)	14.6° ± 4.2° (N=9)	18.5° ± 8.4° (N=10)
ActiGraph	**< 10 Degrees**	3.8° ± 2.8° (N=13)	2.5° ± 2.5° (N=15)	4.1° ± 2.9° (N=11)
	≥ 10 Degrees	15.2° ± 4.6° (N=15)	15.5° ± 6.2° (N=9)	20.3° ± 9.3° (N=8)

Post hoc analysis in those with tilt angles of < 10 degrees revealed significantly higher activity counts at the fastest speed (94 m·min^{-1}) for the obese (26%) group, compared to the normal weight (3,046 ± 568 counts·min^{-1} vs. 2,693 ± 484 counts·min^{-1}, respectively). Although the overweight group recorded 17% higher Actical activity counts than the normal weight group, this difference was not statistically significant (P > 0.05) (Figure 3-1). For those with tilt angles ≥10 degrees, no significant differences were observed among BMI categories (Figure 3-2).

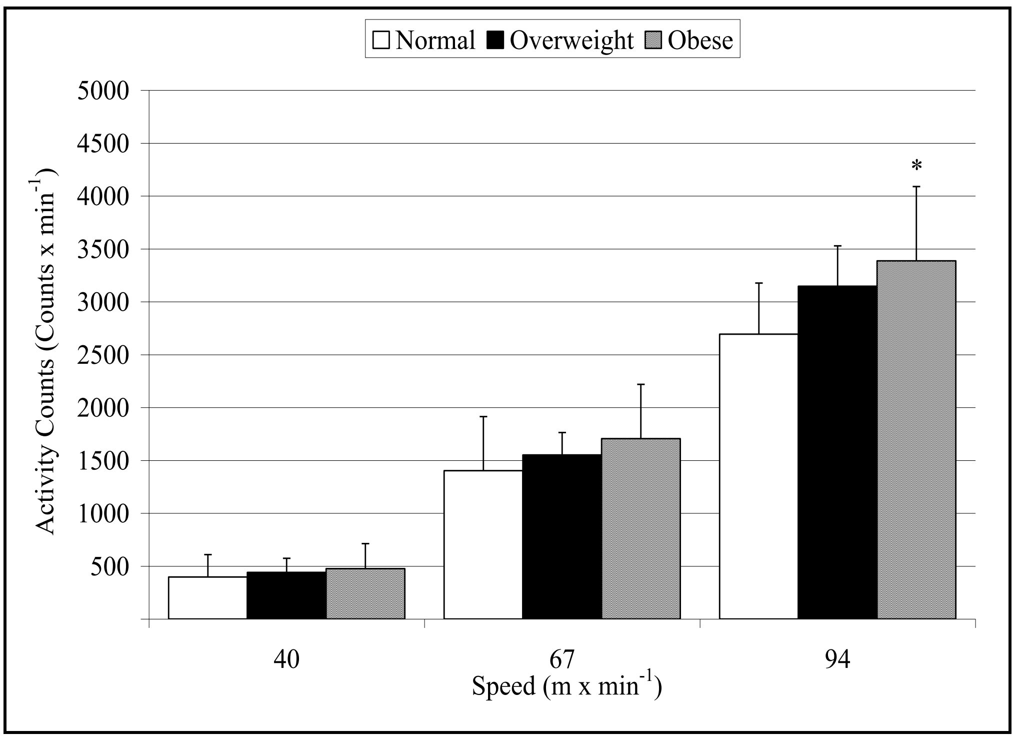

Figure 3-1: Effect of BMI on activity counts for the Actical (tilt angles < 10 degrees). *Error bars* are standard deviations. * Significantly different from normal weight group (P < 0.05).

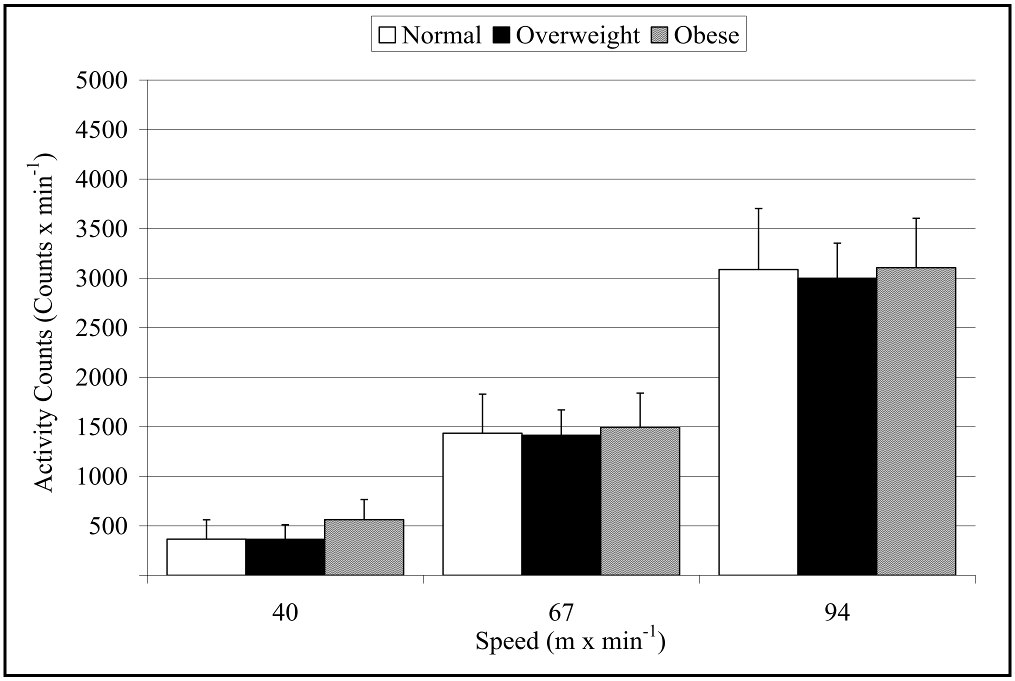

Figure 3-2: Effect of BMI on activity counts for the Actical (tilt angles >= 10 degrees). *Error bars* are standard deviations.

The ActiGraph results are based on sixty-six subjects, as data from five participants were lost due to battery problems (N= 2), or download failure (N= 3). Similar to the Actical, the ActiGraph showed significant increases in activity counts with increasing speeds ($P < 0.001$) (Table 3-2). However, neither BMI nor tilt angle had significant effects on activity counts ($P > 0.05$) (Figure 3-3 and 3-4).

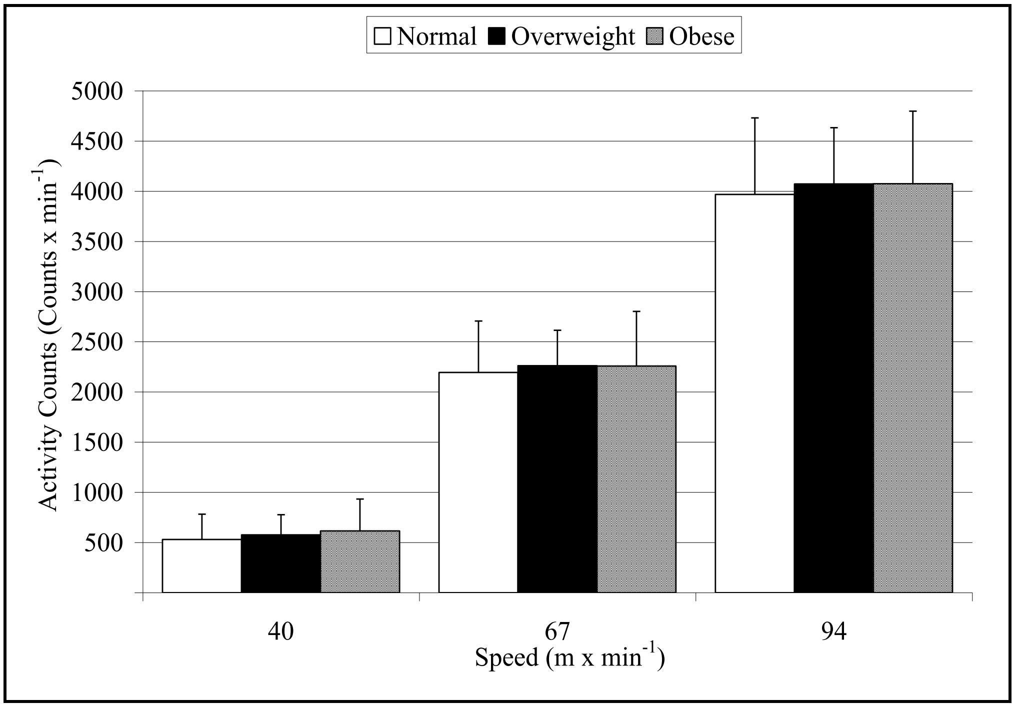

Figure 3-3: Effect of BMI on activity counts for the ActiGraph (tilt angles < 10 degrees). *Error bars* are standard deviations.

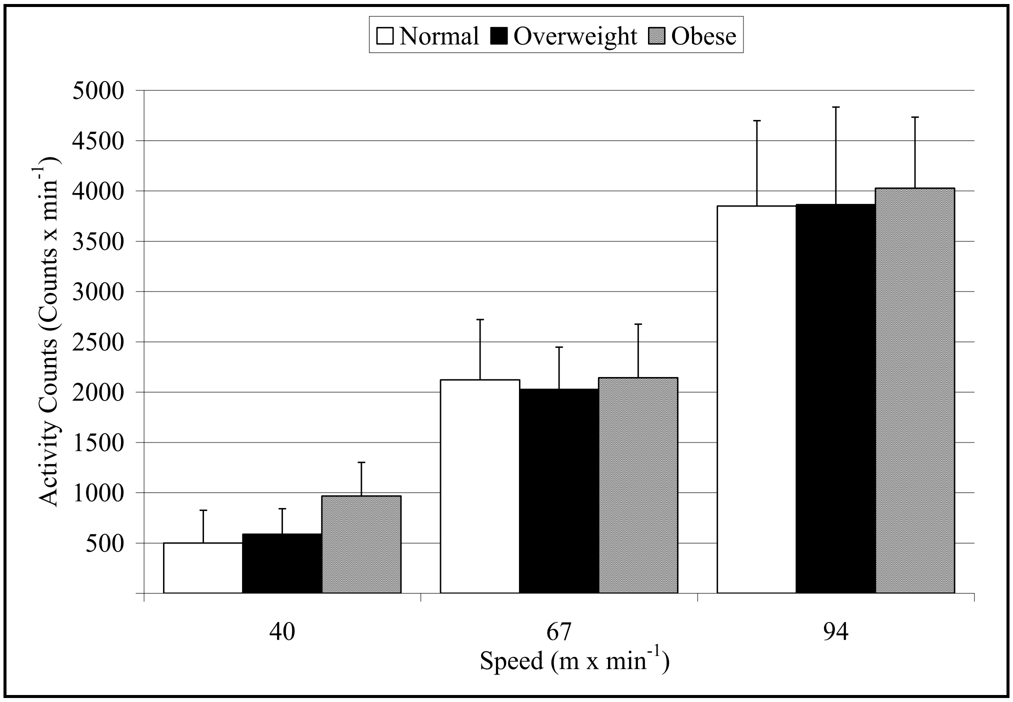

Figure 3-4: Effect of BMI on activity counts for the ActiGraph (tilt angles >= 10 degrees). *Error bars* are standard deviations.

Energy expenditure

As expected, TEE significantly increased with increasing speed ($P < 0.001$). Overall, BMI had no significant effect on Actical estimates of TEE ($P > 0.05$). However, within BMI groups, significant differences were found between measured and estimated values ($P < 0.025$). The Heil 2-regression model for the Actical overestimated EE by up to 35% at the faster speeds for the normal weight and overweight groups. For the obese group, significant differences were only seen at 67 $m \cdot min^{-1}$, with the Heil equation overestimating by 24%.

The ActiGraph estimates of TEE increased with increases in speeds ($P < 0.001$), but there were no significant effects of BMI. However, within BMI groups, differences were observed between measured and predicted values. In normal weight participants (Figure 3-5), the Freedson Kcal regression equation underestimated TEE by 39% at the slowest speed, yet it provided reasonable estimates at the intermediate and fast speeds. In overweight and obese participants (Figure 3-6 and 3-7), the Freedson Kcal regression equation greatly overestimated EE at virtually all speeds (P < 0001). In contrast, the Freedson MET regression equation closely estimated EE for the moderate speed, though it underestimated EE at the slowest speed in all BMI groups (P < 0.001) and overestimated EE at the fastest speed for the normal and overweight groups (P < 0.05).

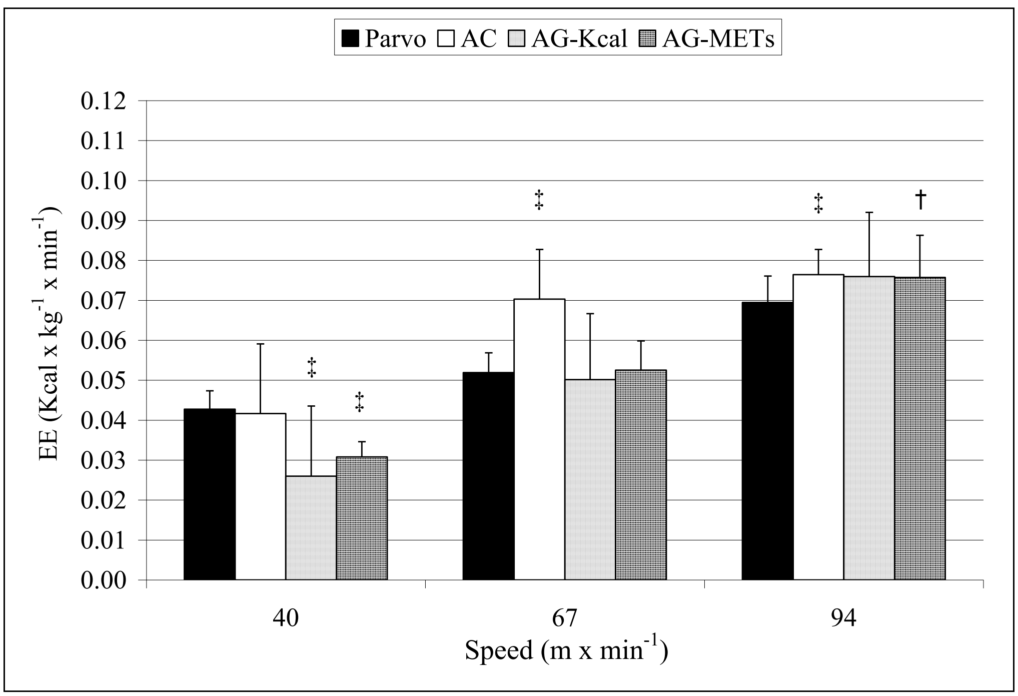

Figure 3-5: Energy expenditure comparisons between measured and predicted values for normal weight individuals.

Error bars are standard deviations. ‡ Significantly different from measured (P < 0.001); † Significantly different from measured (P < 0.05).

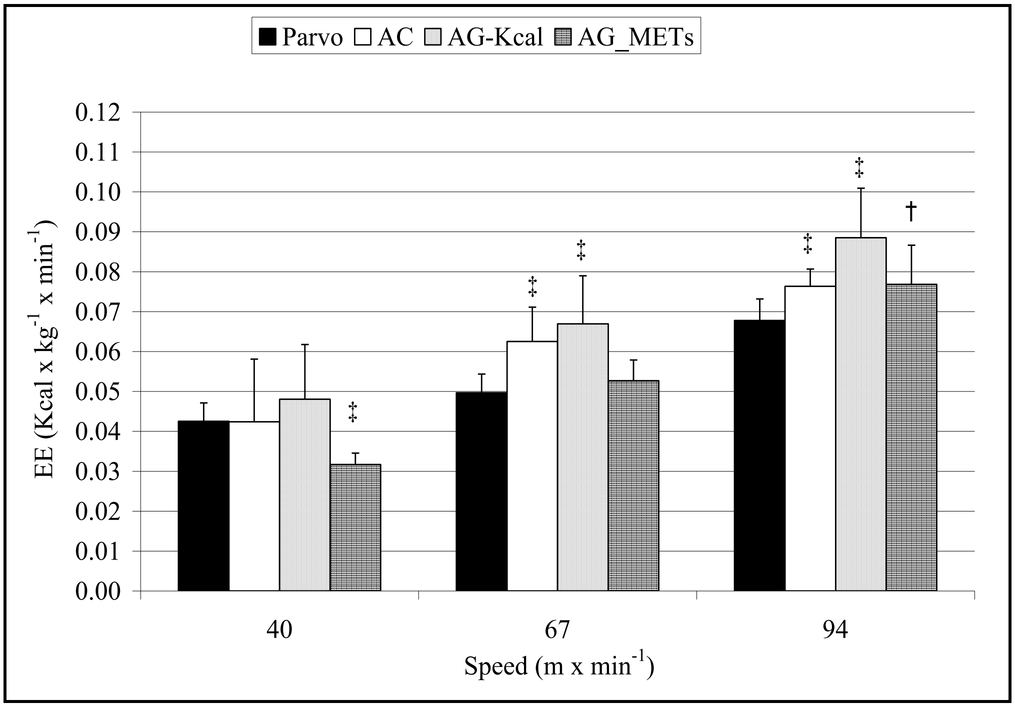

Figure 3-6: Energy expenditure comparisons between measured and predicted values for overweight individuals.

Error bars are standard deviations. ‡ Significantly different from measured ($P < 0.001$); † Significantly different from measured ($P < 0.05$).

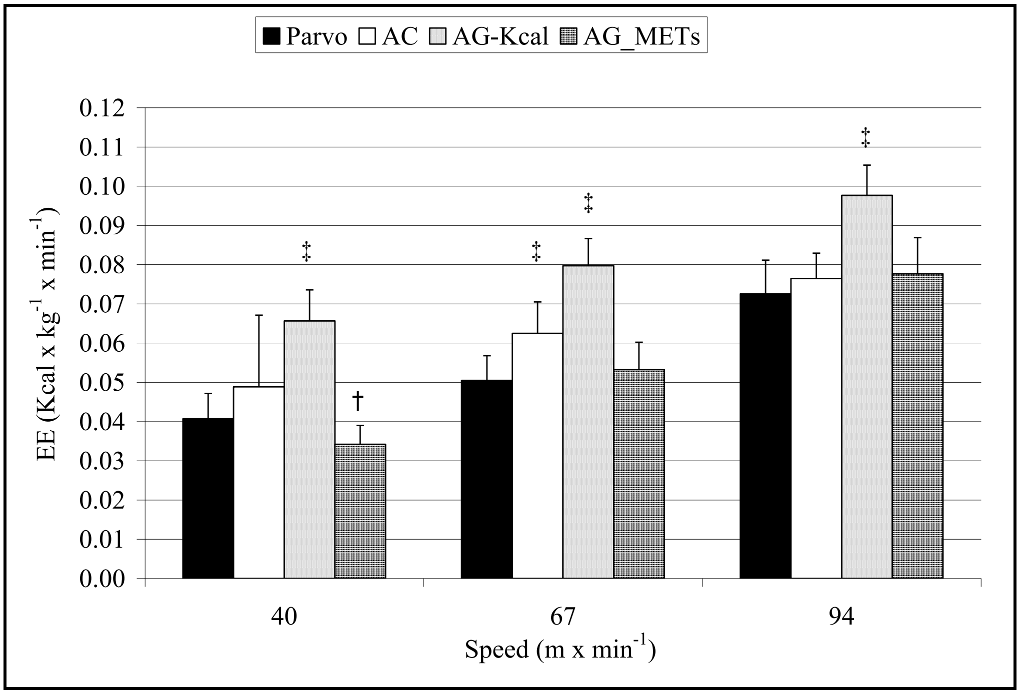

Figure 3-7: Energy expenditure comparisons between measured and predicted for obese individuals.

Error bars are standard deviations. ‡ Significantly different from measured ($P < 0.001$); † Significantly different from measured ($P < 0.05$).

DISCUSSION

Previous investigators have reported that pedometers are less accurate in individuals with BMI values over 30 $kg \cdot m^{-2}$ [1, 4-7], but studies examining the effects of BMI and tilt angle on the ActiGraph and Actical are not available. To our knowledge, this is the first study that has examined the effects of BMI and tilt angle on these two accelerometer-based activity monitors.

BMI and tilt angle significantly affected the Actical count values. In cases where the accelerometer tilt angle was less than 10 degrees, the obese group demonstrated higher activity counts than the normal and overweight groups. This may be related to the Actical's 'omnidirectional' mechanism, which detects acceleration changes in multiple planes, although it is most sensitive in the vertical plane [21]. Hence, it can detect subtle movements of abdominal fat in individuals with higher BMI's. No effects of BMI and tilt angle were seen on the count values recorded by the ActiGraph.

We also sought to determine if predicted EE would be affected by BMI and tilt angle, since EE estimates are dependent on activity counts. We hypothesized that differences in activity counts among the various groups would influence the EE calculations; however, this was not the case. We believe that the over- or underestimation of EE is a function of the prediction equations and not an effect of BMI or tilt angle. In the case of the Actical, Heil's equation overestimated EE at 67 $m \cdot min^{-1}$ for all three BMI groups, but it especially overestimated EE in the normal weight group. At first, this seems impossible because the normal weight group had lower Actical counts than the overweight and obese groups, but it is a function of the speeds used in this investigation and how the equation was derived.

The Heil 2-regression model was based on multiple activities from which two cut-points were derived. This model uses one of two regression lines, depending on the intensity of the activity. For activity counts between 350 – 1200 $counts \cdot min^{-1}$ a regression with a particularly steep slope is used. For activity counts above 1,200 $counts \cdot min^{-1}$ a different regression with a much gentler slope is used. With both of Heil's regression equations, as activity counts increase, the estimated EE increases accordingly. However, as one crosses over 1200 $counts \cdot min^{-1}$, one switches between the two regression lines and EE drops dramatically (e.g. 1,199 $counts \cdot min^{-1}$ = 4.5 METs and 1,200 $counts \cdot min^{-1}$ = 2.4 METs). Thus, the effect of differences in accelerometer counts may be counter-intuitive, since an individual with lower count values can have a higher estimated EE.

Using the regression equations of Freedson et al. [23], we found that the Kcal regression equation was reasonably accurate for the normal weight group, but it greatly overestimated TEE for the overweight and obese groups. Freedson's subjects had a mean age of 24 years and a mean body mass index of 22.8 $kg{\cdot}m^{-2}$. Thus, it is not surprising that the Freedson Kcal equation is invalid in the overweight and obese groups, given that it was not developed on them. Further, it is important to note that both Freedson equations were developed using a previous version of the ActiGraph (ActiGraph 7164), which has a different filter and sampling frequency than the GT1M model we used [25]. Therefore, we conclude that researchers should use caution when interpreting the outcomes of the Freedson Kcal prediction equation in overweight and obese individuals.

The Freedson MET equation slightly overestimated TEE for the normal and overweight group at the faster speed ($P < 0.05$). There was a large underestimation of TEE for all BMI groups at the slowest speed. However, the slowest speed in the present study (1.5 mph) was well below the range of speeds used by Freedson et al. [23] in developing their regression equation (i.e. 3-6 mph). In summary, the Freedson MET equation is more appropriate to estimate TEE than the Kcal equation, although it has some limitations.

Our study is not without limitations. We realized that BMI is not the best indicator of adiposity, and that other variables such as percent body fat are preferred. However, we also collected waist circumference on the participants and those in the highest BMI category would have been classified as "obese" by that measure as well. An additional limitation is the measurement of tilt angle, which was assessed while the participant was standing. Measuring actual tilt angles during the walking bout would be a better way to assess the device's tilt angle while in use.

We believe our findings are novel and expand the current body of literature on objective measures of physical activity. To our knowledge, this is the first study to compare the effects of BMI and tilt angle on the activity counts of the Actical and ActiGraph. Future investigations should seek to determine whether these variables affect the output of these devices under free-living conditions.

In summary, our findings indicate that the Actical has limitations when comparing individuals of varying BMI, as the activity counts may be impacted by BMI and tilt angle. Based on our findings, the ActiGraph seems to be a more suitable device when trying to estimate activity patterns of individuals with a wide range of BMI values. In addition, researchers should be aware of the limitations of predicting EE using published regression equations derived from activity monitor outputs.

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CHAPTER 4

EFFECTS OF BODY MASS INDEX AND SPEED ON STEP COUNT OUTPUT OF WIDELY USED ACTIVITY MONITORS

ABSTRACT

Background: Activity monitors have been widely used in research and are currently being used to study physical activity (PA) trends in the U.S. and Canada. The purpose of this study was to determine if body mass index (BMI) and walking speed affect the step output of commonly used accelerometer-based activity monitors during treadmill walking. **Methods:** Participants were classified into BMI categories and walked on a treadmill at three different speeds (40, 67, and 94 $m \cdot min^{-1}$) while wearing four accelerometer-based activity monitors (ActiGraph™ GT1M, ActiCal™, NL-2000, and StepWatch™). **Results:** At the slowest speed, all waist-mounted devices significantly underestimated actual steps regardless of BMI (P <0.001), with the NL-2000 recording the greatest percentage (72%). At the intermediate speed, the ActiGraph was the least accurate, recording only 80% of actual steps. At the fastest speed, all of the activity monitors demonstrated a high level of accuracy, although the StepWatch slightly underestimated steps in the obese group (P < 0.05). **Conclusion:** Our data suggest that BMI does not greatly affect the step-counting accuracy of accelerometer-based activity monitors. However, walking speed affects the accuracy of the ActiGraph, Actical and NL-2000. The ankle-mounted StepWatch was the most accurate device across a wide range of walking speeds.

Key Words: Adiposity, Physical Activity, Walking, Pedometers

INTRODUCTION

Over the last two decades, investigators have examined the validity and reliability of accelerometer-based activity monitors [1-3]. These devices have been used in the U.S. [14-16, 21] and Canada [4] to objectively monitor PA, estimate temporal trends in PA and examine the associations with health variables. Because some researchers are reporting the step counts given by these devices [5], knowledge of these devices' step count accuracy among individuals of all body types is important.

A number of investigators have examined the validity and reliability of pedometers to measure ambulatory activity [2-3, 6-7]. Some of these devices are accurate for recording the number of steps taken by normal weight individuals [8]. However, Shepherd et al. [9] suggested that pedometer accuracy was influenced by body mass index (BMI). In their study, individuals with a BMI less than 30 $kg \cdot m^{-2}$ had about 1% error when comparing waist and ankle-borne activity monitors. However, among individuals with a BMI greater than 30 $kg \cdot m^{-2}$ the mean absolute percent error was greater for the waist-borne pedometer (6%) compared to the ankle-borne monitor (0.5%) [9]. More recently, other investigators have sought to determine if these differences among BMI categories do indeed exist, with conflicting results [1, 9-12]. Swartz et al. [10] and Elsenbaumer et al. [11] tested the accuracy of the Yamax SW 200, a spring-levered pedometer, and found no significant differences among BMI categories. However, contrary to these findings, several other researchers have reported significant differences among overweight and obese individuals when using a spring-levered device compared to a piezoelectric device [12-14].

The ActiGraph and Actical are two accelerometer-based activity monitors currently being used in large epidemiological studies in the U.S. and Canada [13-16] to assess PA trends and establish associations with health outcomes. Considering the findings from these previous studies, we believe it is important to determine if BMI negatively impacts the accuracy of accelerometer-based activity monitor. Thus, the purpose of this study was to determine whether the step outputs of accelerometer-based activity monitors are affected by BMI.

METHODS

Seventy-one adult volunteers (39 females and 32 males) from the University of Tennessee, Knoxville and surrounding community agreed to participate in this study, and completed an informed consent and Physical Activity Readiness Questionnaire (PAR-Q) prior to data collection. Based on PAR-Q criteria, all participants were free of cardiovascular, pulmonary or metabolic disease and were not taking any medication for blood pressure or heart conditions. The university's Institutional Review Board (IRB) approved the study protocol.

Anthropometry

Height (m) and body mass (kg) were measured with light clothing and without shoes using a stadiometer (SECA, Corp., Columbia, MD) and calibrated scale (Life Measurement, Inc., Concord, CA), respectively. Body Mass Index (BMI) was calculated by dividing the body mass (kg) by height squared (m^2).

We used several commonly used accelerometer-based monitors, the Actical (AC, Mini Mitter Co., Inc., Bend, OR), the ActiGraph (AG) model GT1M (ActiGraph, LLC, Fort Walton Beach, FL), the NL-2000 (NL) (New Lifestyles, Inc., Lee's Summit, MO) and a step activity monitor (StepWatch 3 (SW), OrthoCare Innovations, Seattle, WA). After the anthropometric measurements were completed, the PA monitors were initialized and synchronized through a USB port (AG) or a docking station (AC and SW) using the software provided by their respective manufacturers (Actireader V. 2.10, Actilife Lifestyle Monitoring System, V. 3.3; and

StepWatch for Mac v. 3.1b, respectively). Both the AC and AG were set to their shortest possible epochs (15-sec and 1-sec, respectively). The SW was set to “normal” walking speed and leg motion during initialization. The NL was initialized manually by the investigators.

All the waist-mounted devices were secured to an elastic belt and this was attached to the participant’s waist with the AC and AG on the right and left sides, respectively, over the anterior axillary line. The NL was placed on the right side of the belt, to the left of the Actical, mid-way between the umbilicus and the hip. Finally, the SW was attached to the left ankle using a Velcro strap over the lateral malleolus. During each test, a trained investigator used a hand-tally counter to record the actual steps, and a step was counted every time the right heel made contact with the ground.

Treadmill Walking

Participants walked on a motorized treadmill (Q65 Series 90, Quinton instrument Co., Bothell, WA) for five minutes at three different speeds (40, 67 and 94 $m \cdot min^{-1}$). They were instructed to straddle the belt prior to starting the treadmill. Once the proper speed was reached, the participant was asked to step on the belt and begin walking. After five minutes of walking, participants were asked to straddle the belt again while holding onto the handrails and were asked to remain motionless for two additional minutes. This caused the devices to record zero steps for at least a full minute, and allowed for easier data interpretation.

Statistical treatment

All subjects were placed into one of three BMI categories following established guidelines [15]; normal weight (≤ 24.9 $kg \cdot m^{-2}$), overweight (25.0 – 29.9 $kg \cdot m^{-2}$) and obese (≥ 30 $kg \cdot m^{-2}$). To determine the accuracy of each device, we calculated the percentage of actual steps recorded by each device [(measured steps / actual steps) x 100]. All statistics were run on percent of actual values.

We assumed that a greater number of steps would be recorded at the faster speeds [16] and thus we decided to use a separate statistical analysis for each walking speed. We used a two-way repeated measures ANOVA (device x BMI) to determine whether differences existed among the four devices and determine the effects of BMI, at each of the three prescribed speeds. Pairwise comparisons with Bonferroni adjustments were used to determine if differences existed among devices or BMI categories. In addition, to test the hypothesis that at each speed the percent of actual steps for each device was equal to 100%, we performed single sample t-tests. This allowed us to establish whether significant differences existed between each device and the criterion method.

Bland-Altman plots [17] were constructed to show the variability of the devices' error scores. This technique allows the graphical representation of the mean error score and the 95% prediction intervals. With this method, data points above zero represent underestimations, whereas data points below zero represent overestimations. Greater accuracy of a device results in individual error scores with a tighter prediction interval around zero.

Data were entered into Excel 2003 (Microsoft Co., Redmond, WA) and all statistical analyses were carried out using SPSS statistical software, version 17.0 for Windows (SPSS Inc., Chicago, IL). A significance level of alpha = 0.05 was chosen to denote statistical significance. All values are reported as mean ± standard deviation (SD).

RESULTS

The final analysis was performed using sixty-five participants, as data for six individuals were lost due to battery problems, (AG; N= 2), or download failure (AG; N= 3 and SW; N=1).

The characteristics of the participants are shown in Table 4-1. For each speed, we divided our sample into BMI categories and compared each device to the criterion at each of the three different speeds (40, 67 and 94 m·min^{-1}).

Table 4-1: Physical Characteristics by BMI and Gender (mean ± SD)

	All Participants (N= 65)	**Normal (N=25)**	**Overweight (N=22)**	**Obese (N=18)**
Age (yrs)	30.8 ± 11.2	27.2 ± 6.6	34.3±14.1	31.7 ± 11.4
Height (m)	1.72 ± 0.10	1.71 ± 0.09	1.73 ± **0.12**	1.71 ± 0.08
Weight (kg)	79.3 ± 16.8	65.1 ± 10.0	80.0 ± 12.1 †	97.4 ± 9.6 †
BMI (kg·m $^{-2}$)	26.9 ± 5.2	22.1 ± 1.9	26.9 ± 1.2	33.6 ± 3.6 †

† Significant differences between BMI categories ($P < 0.001$)

Accuracy at the slowest speed

We observed the biggest discrepancy among the devices at 40 m·min^{-1}, with all three waist-mounted devices significantly under-counting steps regardless of BMI ($P < 0.001$). Among the waist-mounted activity monitors, the AG recorded the lowest percentage of steps for the normal weight (38%), the overweight (46%) and the obese (48%) (Figure 4-1). However, this difference within BMI categories was not statistically significant ($P > 0.05$). The AC recorded an average of 65% of all steps taken and did not seem to be greatly influenced by adiposity. Of the three waist-mounted devices, the NL was found to be the most accurate, averaging 73% of actual steps. The SW was the most accurate device at the slowest walking speed, recording 100%, 102% and 96% of all actual steps for the normal weight, overweight and obese groups, respectively.

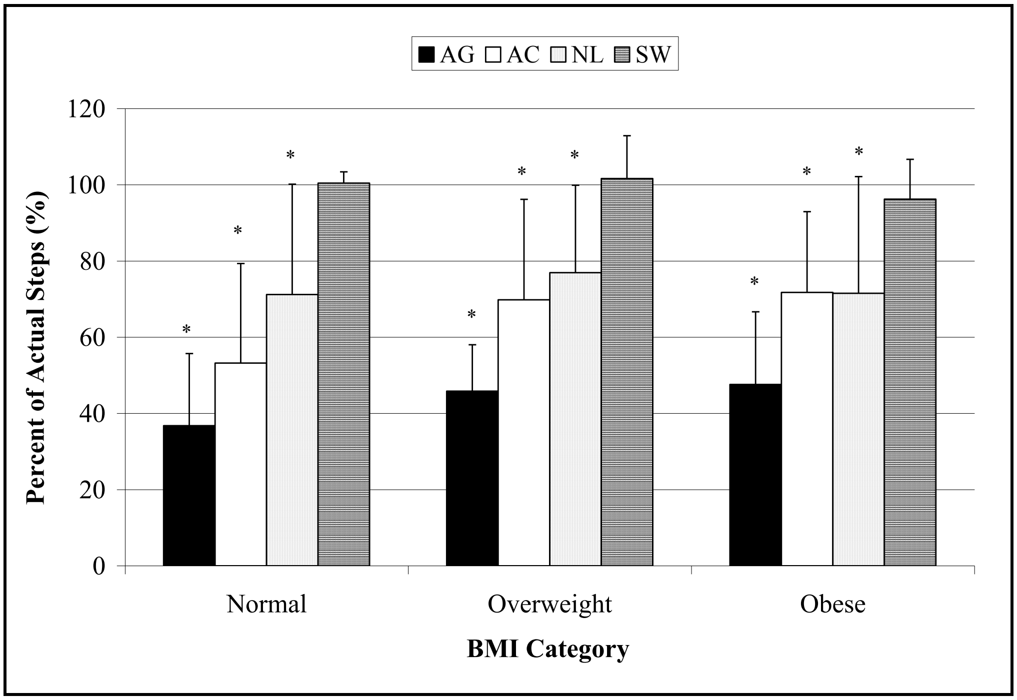

Figure 4-1: Percent of actual steps by Actical (AC), ActiGraph (AG), NL-2000 (NL) and StepWatch (SW) by BMI category at 40 $m \cdot min^{-1}$.

Errors bars represent standard deviations. * $P < 0.001$; † $P < 0.05$

Accuracy at the moderate speed

At a moderate speed, no significant differences were observed among devices for the normal and overweight groups ($P > 0.05$). However, when compared to the criterion, the AG recorded 80% of steps taken for those in the obese group ($P = 0.005$). All other devices recorded 95% to 97% of total steps ($P > 0.05$) (Figure 4-2).

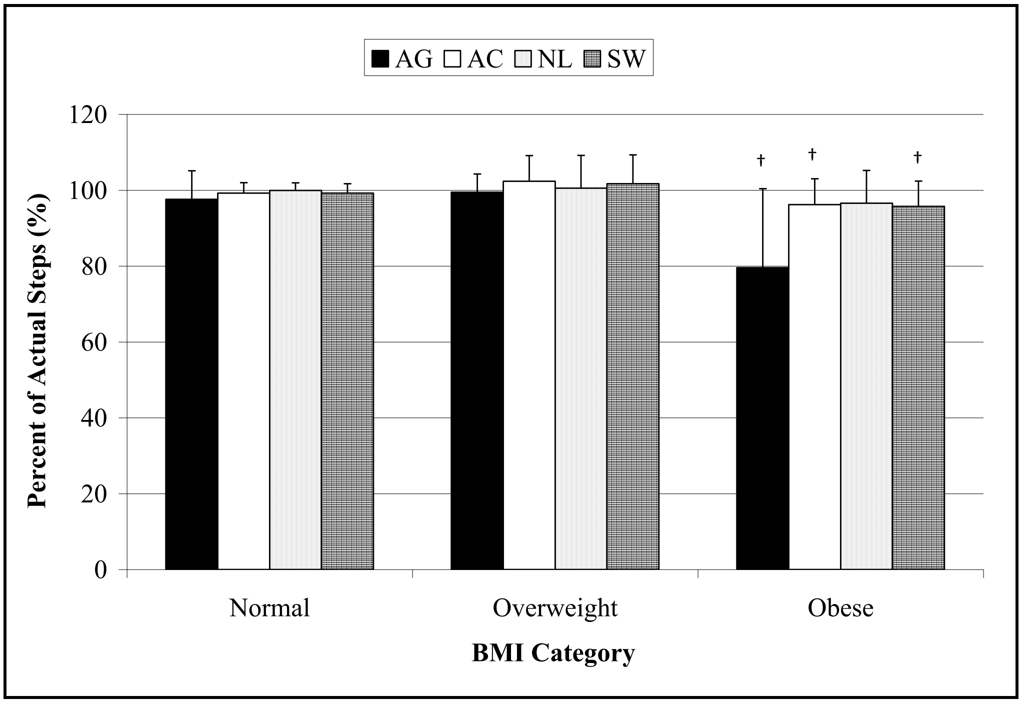

Figure 4-2: Percent of actual steps taken by Actical (AC), ActiGraph (AG), NL-2000 (NL) and StepWatch (SW) by BMI category at 67 m·min^{-1}

Errors bars represent standard deviations. † P < 0.05.

Accuracy at the faster speed

At a fast walking speed (94 m·min^{-1}) no significant differences were observed in the percentage of actual steps taken within BMI categories. For the normal weight category, all devices recorded over 99%. For the overweight, a slight overestimation (2.25%) was seen; and among the obese all devices recorded over 97% of all steps taken (*P* > 0.05) (Figure 4-3). Among the obese group, the SW recorded statistically significantly lower steps; however, the practical significance of this small difference (< 5%) is limited.

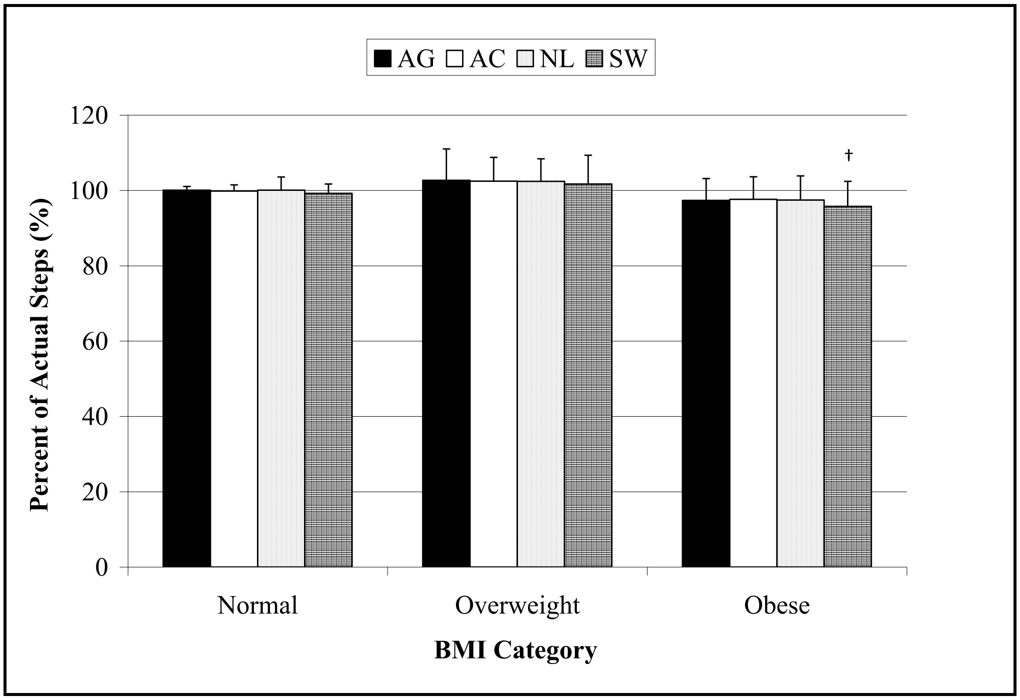

Figure 4-3: Percent of actual steps by Actical (AC), ActiGraph (AG), NL-2000 (NL) and StepWatch (SW) by BMI category at 94 m·min^{-1}.

Errors bars represent standard deviations. † P < 0.05.

Figures 4-4 to 4-7 shows the Bland-Altman plots for all four devices. All three waist-mounted devices demonstrated wide variability in error at step frequencies below 100 steps·min^{-1}; the errors became smaller as step frequency increased above this threshold. The SW showed the smallest variability in error (SD ± 8 steps·min^{-1}), while the AG showed the highest variability in error (SD ± 52 steps·min^{-1}).

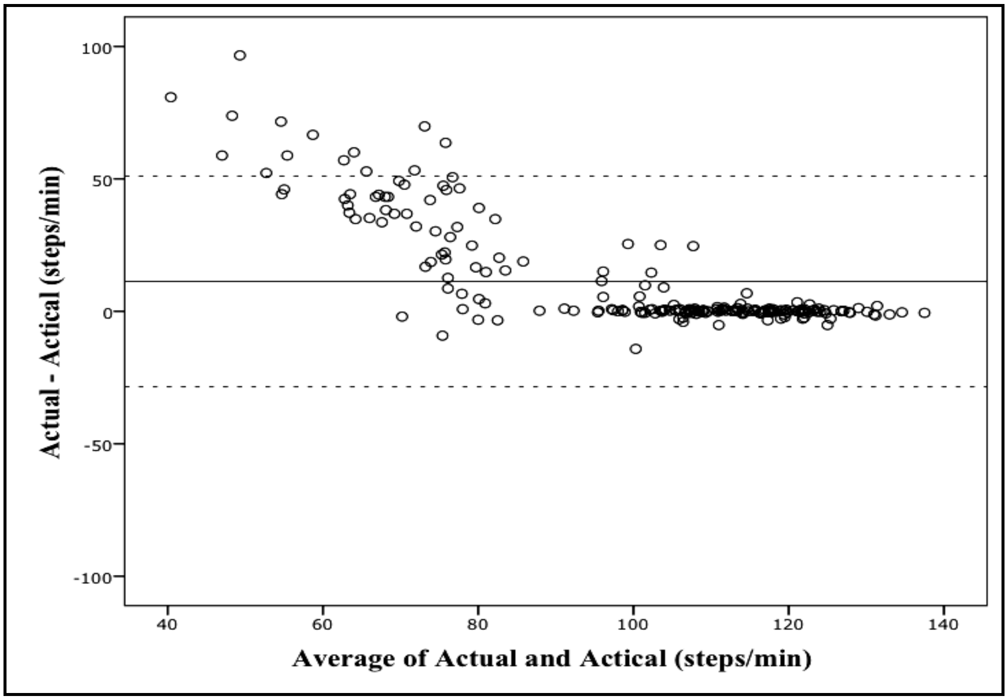

Figure 4-4: Bland-Altman plot depicting error scores (actual steps per minute - device steps per minute) for the Actical.

Solid line represents mean differences; dashed lines represent 95% prediction intervals.

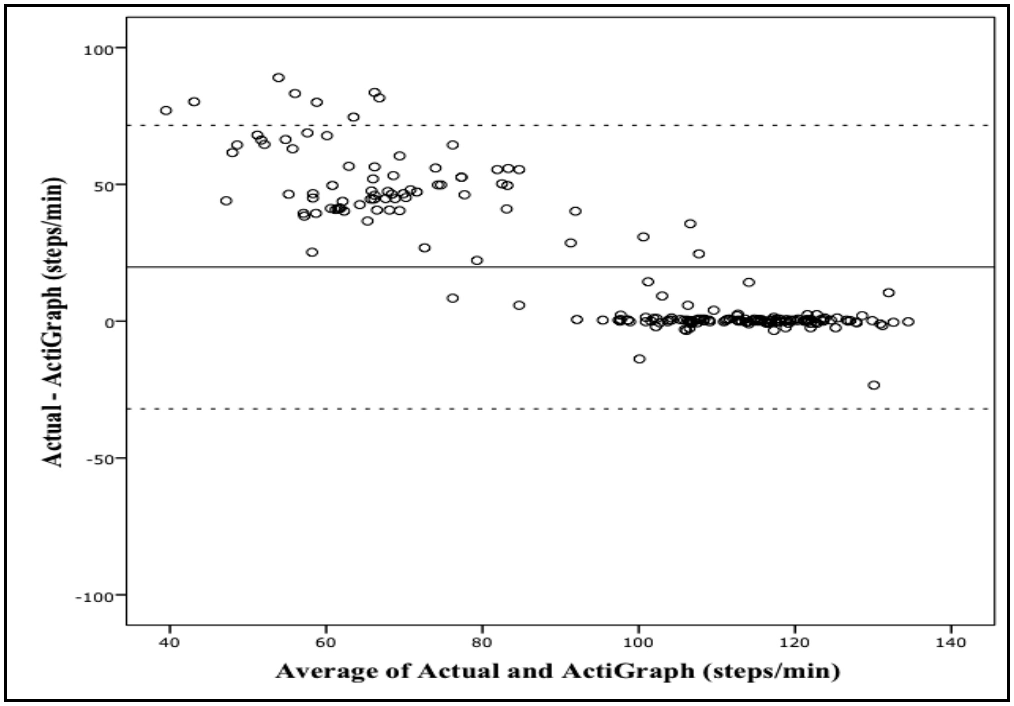

Figure 4-5: Bland-Altman plot depicting error scores (actual steps per minute - device steps per minute) for the ActiGraph.

Solid line represents mean differences; dashed lines represent 95% prediction intervals.

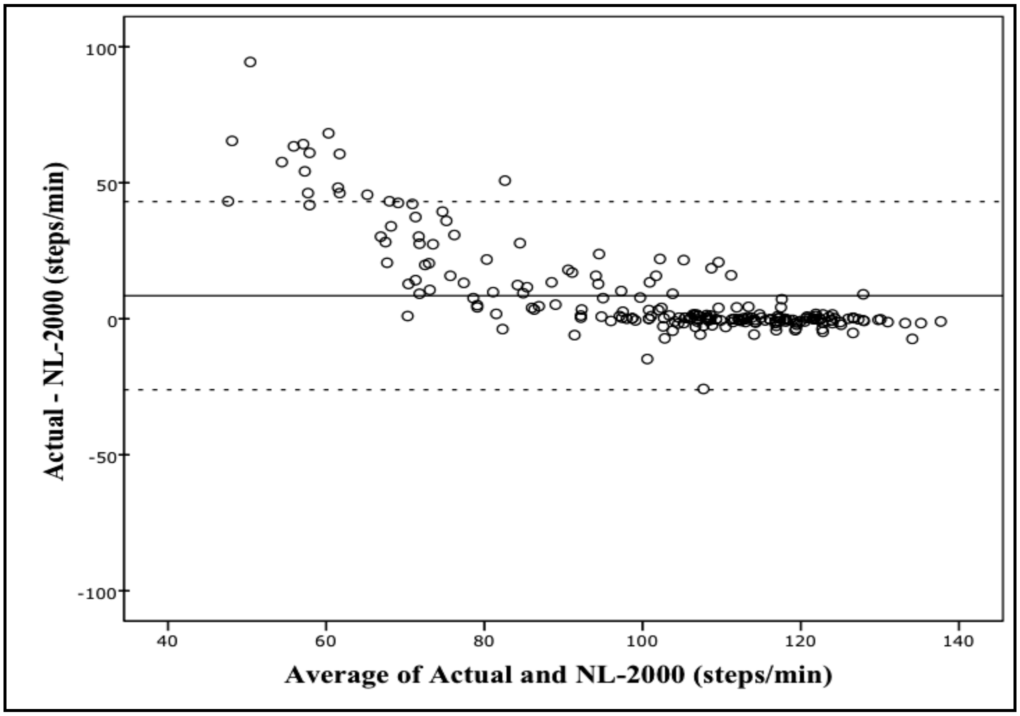

Figure 4-6: Bland-Altman plot depicting error scores (actual steps per minute - device steps per minute) for the NL-2000.

Solid line represents mean differences; dashed lines represent 95% prediction intervals.

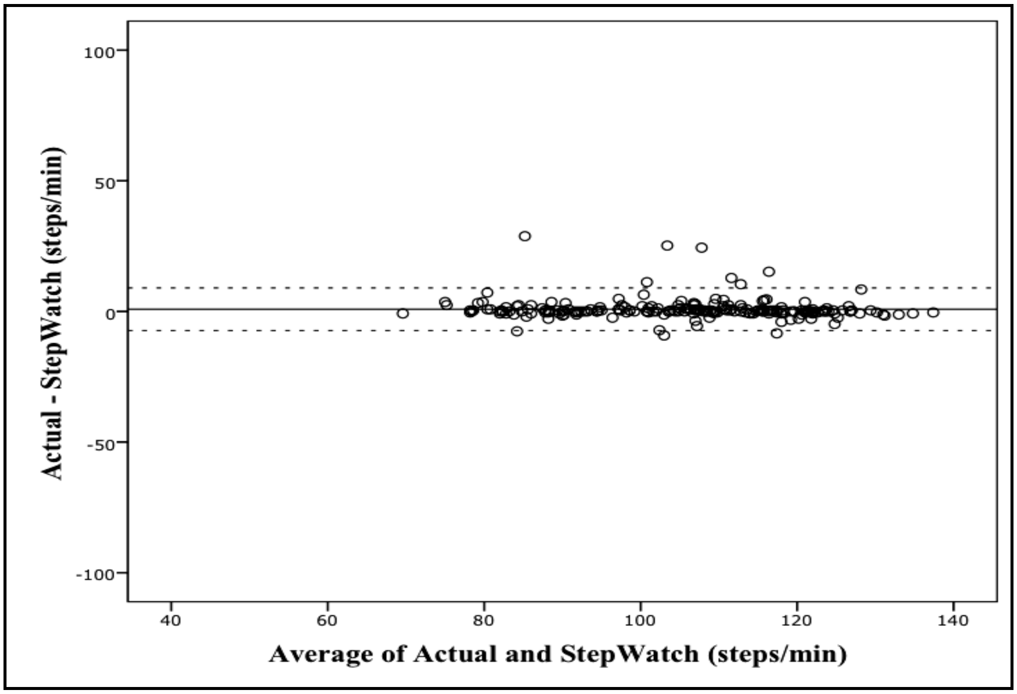

Figure 4-7: Bland-Altman plots depicting error scores (actual steps per minute - device steps per minute) for the StepWatch.

Solid line represents mean differences; dashed lines represent 95% prediction intervals.

DISCUSSION

We sought to determine if BMI, commonly referred to as a surrogate measure of adiposity, affects the step counts of waist-borne activity monitors while waking on a treadmill at three preselected walking speeds. Our findings indicate that waist-mounted activity monitors underestimated steps by approximately 20% to 60% at the slowest speed, regardless of adiposity. At 40 $m \cdot min^{-1}$, the AG recorded the lowest percentage of steps (44%), with higher accuracy in the obese group. By comparison, the AC and NL averaged 66% and 73% of actual steps taken, respectively. The SW was the most accurate device at 40 $m \cdot min^{-1}$, recording 99% of steps. At the moderate and fast speeds, all devices seemed to estimate percentage of steps taken with good accuracy, except for the AG, which under-estimated the number of steps taken by obese individuals by 20% at a 67 $m \cdot min^{-1}$.

Our study agrees with the findings of previous research [6, 9, 12]. Similar to Shepherd and colleagues [9], we found that the SW provides the most accurate measure of ambulatory activity across a wide range of speeds and BMI values. At the slowest speed, waist mounted devices recorded 44% to 73% of actual steps. However, step counting accuracy improved as speed increased, and most devices recorded over 99% of actual steps at speeds over 67 $m \cdot min^{-1}$.

In addition, our findings are consistent with the general observation that waist-mounted devices have decreased accuracy at slower walking speeds [6]. Even though we did not use spring-lever arm pedometers as in the study of Bassett et al. [6] we did compare steps to the same criterion method (i.e. hand-tally counter) over a similar range of speeds. Thus, we conclude that our three waist-mounted devices, similar to spring-levered pedometers, also seem to be affected by the reduced vertical accelerations experienced at these slow speeds.

Furthermore, this conclusion is consistent with a study by Crouter et al. [12], which found that piezo-electric devices are most accurate at speeds above 67 m·min^{-1}.

Recently, Tudor-Locke and colleagues [18] reviewed the typical values for steps per day in individuals living with chronic disease. In this review, they report on studies using both waist-mounted and ankle-borne activity monitors for different clinical populations. They found that in those studies that used the ankle-borne activity monitor (SW) the mean steps per day were higher than in those using waist-mounted pedometers [18]. In studies of individuals with heart and vascular disease, including chronic heart failure, myocardial infarction, peripheral arterial disease and post-stroke, in which waist-mounted devices were used, the median value for habitual activity was 4,684 steps·day^{-1}. However, in studies that used the ankle-mounted device, the habitual daily activity was 6,515 steps·day^{-1} (39% more than in studies using the waist-mounted pedometer). Moreover, in individuals with arthritis, joint arthroplasty and fibromyalgia, the average value was about 4,500 steps·day^{-1} when using a waist-mounted device, and over 10,400 steps·day^{-1} when using the SW [18]. Taken together, these findings suggest that waist-mounted pedometers undercount steps per day in many patient populations.

As surveillance systems begin to use these accelerometer-based activity monitors to objectively measure the PA patterns of adults in the U.S. [19] and Canada [4], it is important for researchers to be aware of differences between the devices in order to make adequate health related recommendations. For instance, a recent study by Sisson et al. [20] reported on the cross-sectional relationships between walking volume and metabolic syndrome, using steps data (ActiGraph 7164) from the National Health and Nutrition Education Survey (NHANES). The study found that adults who took more steps per day had lower waist circumferences, higher HDL cholesterol levels, and lower triglyceride levels than those who were less active. Taking our results into consideration, we can conclude that the inverse relationship of steps per day and health measures shown by Sisson et al. [20] is real, as opposed to artifact caused by a device that under-records steps in obese individuals. However, it is possible that individuals who walked slowly were under-credited for the steps they accumulated.

In summary, we have shown that BMI contributes very little to the error of these waist-mounted, accelerometer-based activity monitors. However, walking speed could be an important source of device error. Thus, for individuals who walk at slower speeds, either due to age or disability, we agree with Shepherd et al.'s [9] recommendation of using an ankle-mounted device (i.e. StepWatch) as a more accurate way to objectively measure physical activity than a waist-mounted activity monitor. Considering the trade off between device cost and accuracy, we challenge manufacturers to develop ankle-borne devices that are as accurate as the SW, but are more cost effective so that researchers, and perhaps individuals, could have an accurate measure of ambulatory activity.

For individuals with normal walking patterns, and without any physical limitations, the waist-mounted or ankle-borne devices seem to accurately measure walking volume. We believe that public health practitioners, epidemiologists, and researchers should be aware of these differences in order to accurately assess the ambulatory levels of various populations.

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CHAPTER 5

ACCURACY OF ACCELEROMETER-BASED STEP COUNTERS IN CONTROLLED AND FREE-LIVING ENVIRONMENTS

ABSTRACT

Introduction: Over the last decade, numerous studies have established the usefulness of pedometers and accelerometers as objective activity monitors. Under controlled laboratory conditions, some of these devices have been shown to provide accurate and reliable measures of ambulatory activity. However, limited data exist on the accuracy of these devices under free-living conditions. **Purpose**: The purpose of this study was two-fold: 1) to examine the effects of walking speed and BMI on step count accuracy of accelerometer-based activity monitors (ActiGraph 7164, GT1M, GT3X; Actical, ActivPAL, StepWatch) and a waist-mounted pedometer (Digi-Walker) in a controlled environment; and 2) to assess the step count accuracy for these same devices among individuals in a free-living environment. **Methods**: Fifty-six individuals wore six accelerometer-based activity monitors while performing treadmill walking (40, 54, 67, 80 and 94 m·min-1) and during one day of free-living activity. The criterion for steps during treadmill walking was the hand-tally counter, while the criterion for steps during the free-living condition was the StepWatch. **Results**: BMI had no effect on step count accuracy during treadmill walking. The StepWatch, PAL, and the AG7164 were the most accurate across all speeds; the remaining devices were only accurate at speeds over 67 $m \cdot min^{-1}$. In the free-living environment, the AG7164 recorded 99.5 ± 27% (mean + SD) of StepWatch-determined steps. **Conclusion**: We demonstrated that BMI does not affect the step output of commonly used activity monitors during walking. In addition, 67 $m \cdot min^{-1}$ appears to be the minimum speed required for accurate step counting, at least for most waist-mounted activity monitors. Finally, the StepWatch, AG7164 and PAL were the most accurate of all devices tested on the TM, but only the AG7164 yielded comparable results to the StepWatch in the free-living environment.

Key Words: BMI, Accelerometers, Step counts

INTRODUCTION

Physical activity has been found to have a number of health-related benefits [1]. More specifically, walking is a health promoting activity that can be performed by most people, without great effort or risk [2]. Pedometers and accelerometers are small devices used to objectively measure walking. Over the last decade, numerous studies have established the usefulness of these devices under controlled laboratory conditions, providing accurate and reliable measures of ambulatory activity [3-6]. However, limited data exist on the accuracy of these devices in free-living conditions. Today, the U.S. [7], Canada [8], and Europe [9] use accelerometer–based monitors for physical activity surveillance ; and health promotion campaigns to combat obesity and increase physical activity patterns among adults and children [10-11]. Thus, determining the accuracy of these devices in the free-living environment is of great importance.

Even though waist-borne pedometers have been shown to provide accurate measures of physical activity in normal weight individuals [5, 12-13], their accuracy in overweight/obese individuals has been questioned [14-15]. Hence, Sheppard et al. [15] suggested that the ankle might be a more suitable location when trying to monitor ambulatory activity among overweight and obese individuals. Moreover, the accuracy of the ActiGraph, which is a device worn at the waist, has been questioned in special populations. For instance, Chou et al. [16] reported that among individuals with lower limb amputation, the ActiGraph recorded 90% of steps when placed on the ankle of the prosthetic limb, compared to 64% of steps taken when on the waist. Recent findings from our laboratory demonstrate that walking speed affects the step count accuracy of the ActiGraph GT1M, Actical and NL-2000 in a laboratory setting [17]. Tyo et al. [18] showed that the NL-2000 and Digi-Walker (SW 200) are both highly influenced by walking speed in the free-living environment. Furthermore, Kozey et al. [19] recently recently found differences in the step counts of two generations of the ActiGraph accelerometer (models 7164 and GT1M) while walking on a level track.

Therefore, the goals of this study were: (1) to compare the effects of speed and BMI on the accuracy of five different accelerometer-based activity monitors and a pedometer (Digi-Walker) during treadmill walking between 40 and 94 m/min, and (2) to assess the step count accuracy of these same devices in a free-living environment.

METHODS

Participants

Fifty-six individuals from the University of Tennessee, Knoxville and surrounding community were recruited for this study. We recruited participants across a wide range of BMI values, and attempted to have similar numbers of individuals in each BMI group. Participants were limited to those with negative responses to a Physical Activity Readiness Questionnaire (PAR-Q) and without orthopedic or physical limitations. All participants completed an informed consent form prior to enrolling in the study, and the study protocol was approved by the university's Institutional Review Board (IRB).

Activity Monitors

To compare step-count accuracy of different activity monitors while performing treadmill walking, we used a commonly used pedometer [Digiwalker SW-200 (DW) Yamax Corp., Tokyo, Japan], and six research grade accelerometer-based activity monitors [Actical™ (AC), Phillips Respironics, Bend, OR; ActiGraph™, models 7164 (AG7164), GT1M (GT1M) and GT3X (GT3X), ActiGraph, Pensacola, FL; ActivPAL™ (PAL), PAL Technologies Limited, Glasgow, UK; and the StepWatch 3 (SW), OrthoCare Innovations, Seattle, WA]. Technical descriptions of each device can be obtained elsewhere [20-25].

Device placement was standardized based on the manufacturers' recommendations and previous validation studies [26-27]. The DW, AC and three generations of the ActiGraph were affixed to two different elastic belts and worn at two different times during the study. The first belt included the AG7164 and GT1M, and the DW. The AG7164 and GT1M were worn over the right and left hip, respectively, at the anterior axillary line. The DW was worn in the mid-line of the right thigh, medial to the AG7164. The AC and GT3X were attached to a second belt and were placed over the left and right anterior axillary line, respectively.

Two additional activity monitors (PAL and SW) were placed on the right leg. The PAL was secured to the anterior aspect of the right thigh with Tegaderm™ adhesive dressing. The SW was attached to the right ankle, above the lateral malleolus. During treadmill walking, the first author counted the steps using a hand-tally counter and this was used as the criterion measure.

The SW served as the criterion measure of steps taken during the free-living condition, due to its accuracy over a wide range of walking speeds [28]. To our knowledge, the StepWatch is by far the most accurate device for walking speeds ranging from 27 to 106 $m \cdot min^{-1}$ (1 to 4 mph) [28-32]. In addition, the StepWatch gives an accurate measure of step counts even in obese individuals, unlike many waist-mounted pedometers [14]. In part, this is due to its wear location on the ankle, as opposed to the waist. However, the StepWatch is expensive ($425 + $1500 for docking station and software- get most recent prices), making it cost-prohibitive for many applications. To minimize any "order" effect, a counter-balanced design was used with half the sample wearing belt one first followed by belt two, and the other half wearing them in the opposite order.

Testing Protocol

The testing protocol consisted of the initial visit to the lab and two consecutive days in the free living-environment. During the initial visit, participants had their height (cm) and weight (kg) measured wearing light clothing (e.g. shorts and t-shirt) without shoes using a standard stadiometer (SECA, Corp., Columbia, MD) and calibrated scale (Tanita Body Composition analyzer, Model BC-418), respectively. Circumferences of the waist, abdomen and hip were measured in duplicate (cm), using a tension-gauge measuring tape over clothing.

Because we used multiple devices, all activities were performed in two trials; trial one (T1) included the AG7164, GT1M, DW and SW; for trial two (T2) we used the GT3X, AC, PAL and SW. Participants were asked to remain motionless for one minute before and after each condition, to facilitate the data analysis.

Prior to beginning each trial, all the devices were initialized and synchronized using their respective software (AC, Actireader V. 2.10; AG7164, ActiSoft Analysis Software V. 3.2.1.1; GT1M and GT3X, Actilife Lifestyle Monitoring System, V. 4.4.1; SW, StepWatch V. 3.0; PAL, ActivPAL Professional- Research edition, V. 5.8.5.0). We set all devices to record steps every 15-seconds (epoch) which is the maximal recording epoch allowed by the AG7164. Fifteen-second epochs gave us enough recording time for participants to take the devices home for at least three days.

Treadmill Walking

Participants were instructed to walk 100 steps on a treadmill (Medtrack ST55, Quinton, Bothell, WA) at five different speeds (40, 54, 67, 80 and 94 $m \cdot min^{-1}$). Prior to each walking bout, participants were asked to straddle the treadmill belt and remain motionless for one minute. After each bout, the DW data were recorded by hand and the DW was reset to zero.

Free Living Condition

Upon completion of the treadmill activities, participants were instructed to take all the devices home and wear them for the following two days, during all waking hours. Each participant was sent home with specific instructions on what device to wear each day depending on their group assignment. In addition, they were instructed to record the time at which they put the devices on in the morning, and the time at which they took them off at the end of the day. If they needed to remove the devices for a period of time during the day, participants were asked to write down the on and off times (as well as the amount of steps on the DW for trial 1) to eliminate those values from the total wear time and total DW steps. Participants were instructed to wear the devices during all waking hours, except when showering, swimming, or exercising. Participants were asked to refrain from wearing the devices while exercising (other than walking), as this may influence the accuracy of the devices. Participants were instructed to log the DW steps in a space provided in the instruction sheet after the completion of trial 1.

Statistical treatment

We calculated BMI ($kg{\cdot}m^{-2}$) and categorized each participant into normal weight (< 24.9 $kg{\cdot}m^{-2}$), overweight ($25.0 – 29.9$ $kg{\cdot}m^{-2}$) and obese (> 30 $kg{\cdot}m^{-2}$) categories. Differences between BMI categories for anthropometric measures and total wear time were determined using a one-way ANOVA.

We were only interested in the step data provided by each device. Therefore, we extracted the step data recorded during the treadmill and free-living conditions, and we recorded the total "wear time," reported by each participant for each device. For each activity monitor with a computer interface, the data were downloaded and the steps within the designated time interval were added as the steps for that activity. In order to account for discrepancies in the participant's recorded time and the monitor's internal clock for the free-living condition, we included all steps within three minutes of the time the participant recorded manually. The percentage of actual steps taken was calculated for each device [(measured steps / actual steps) x 100] and this variable was used in all statistical analysis. During the treadmill condition, participants were asked to walk 100 steps while a trained investigator counted steps by hand-tally. During the free-living environment condition, we used the SW as the reference measure due to its high accuracy [29].

We did not intend to compare the treadmill walking to the free-living condition, thus we analyzed each activity separately. The treadmill walking experiment was analyzed using repeated measures ANOVA to examine the interactions among device, speed and BMI categories. A one-way ANOVA with Bonferoni adjustments allowed us to determine if differences existed between devices at each of the prescribed speeds (device x speed), while an independent sample t-test (test value = 100%) was used to locate differences between each device and the criterion method.

The free-living condition was also analyzed using a repeated measures design (ANOVA) to determine differences within device and among BMI categories. One-way analysis of variance with Bonferoni adjustments was used to determine differences between each device. Additionally, an independent sample t-test was used to examine differences between devices and the criterion method (test value = 100%). Activity monitor wear times were compared using a paired sample t-test to examine if wear time was different between testing days.

Data were entered into Excel 2003 (Microsoft Co., Redmond, WA) and all statistical analyses were conducted with SPSS, version 17 for Windows (SPSS Inc., Chicago, IL). The significance level was set at P < 0.05 for all statistics. However, because a value may be statistically significant, but not necessarily have practical relevance, we felt that reporting only devices that had a p-value < 0.05 and a percent difference ≥ 5% of actual steps would be most applicable.

RESULTS

Table 5-1 show participant demographics by BMI categories. Total wear times (hours) and total steps recorded by the SW during the free-living activity are included in table 5-2. Overall, males (N=28) were taller, heavier and had larger waist circumferences than their female counterparts. No significant differences (*P* > 0.05) were observed among participants for wear time, or total steps taken between each of the testing days.

Table 5-1: Participants demographic by BMI category (mean ± SD)

	Normal Weight (N=21)	**Overweight (N=19)**	**Obese (N= 16)**
Age (y)	28.3 ± 10.5	31.2 ± 9.9	29.0 ± 7.9
Height (m)	1.73 ± 0.09	1.71 ± 0.10	1.69 ± 0.07
Weight (kg)	67.5 ± 8.4	80.6 ± 9.8**	97.5 ± 9.6**,†
BMI (kg·m^2)	22.5 ± 2.0	27.4 ± 1.1**	34.1 ± 3.2**,†
Circumferences			
Waist (cm)	74.8 ± 5.8	83.5 ± 8.4*	95.4 ± 7.4**,†
Abdominal (cm)	78.4 ± 7.2	89.0 ± 7.7*	102.1 ± 9.6**,†
Hip (cm)	94.4 ± 6.5	103.6 ± 3.9**	115.3 ± 8.7**,†

* Significantly different from normal weight (*P* < 0.001); * (*P* < 0.05);
† Significantly different from overweight (*P* < 0.001)

Table 5-2: Total wear time and StepWatch recorded steps for free-living condition

	Normal Weight (N=21)	Overweight (N=19)	Obese (N= 16)
Total Wear Time (Free-living)			
Day 1 (h)	12.8 ± 1.9	13.7 ± 1.9	14.0 ± 1.8
Day 2 (h)	13.0 ± 1.9	13.2 ± 1.6	13.8 ± 1.5
Steps recorded by StepWatch (Free-living)			
Day 1 (steps)	7481 ± 6014	7464 ± 4020	8922 ± 5387
Day 2 (steps)	6251 ± 2899	7638 ± 4624	7821 ± 4033

There was no significant effect of BMI on percentage of steps taken for the treadmill walking condition. However, a significant interaction was observed between speed and device ($P < 0.001$). The largest differences among devices occurred at the slowest speeds, where the SW and PAL had the greatest accuracy (100 ± 1% and 98 ± 3% of steps, respectively) (Figures 5-1 - 5-3). At faster walking speeds (80 and 94 $m\cdot min^{-1}$); all devices recorded over 97% of steps taken. The GT1M, AC, PAL and GT3X recorded statistical significant lower steps than the SW ($P < 0.001$); however these differences were less than the 5% "cut-off" we had considered practically significant (Figure 5-4 and 5-5).

Moreover, when comparing each device to directly observed step counts, we found that all devices, with the exception of the SW, significantly under-estimated the percentage of steps taken at the slowest two speeds (Figures 5-1 and 5-2). At the intermediate speed of 67 $m\cdot min^{-1}$, the GT1M, DW, and GT3X significantly underestimated the percentage of steps taken ($P < 0.01$) (Figure 5-3). At 80 $m\cdot min^{-1}$, the GT1M, GT3X, AC and PAL significantly underestimated steps, but the underestimations were less than 5% (Figures 5-4). At the fastest speed, only the GT1M, GT3X and PAL showed any significant undercounting of steps, but once again they underestimated by less than 5% (Figure 5-5).

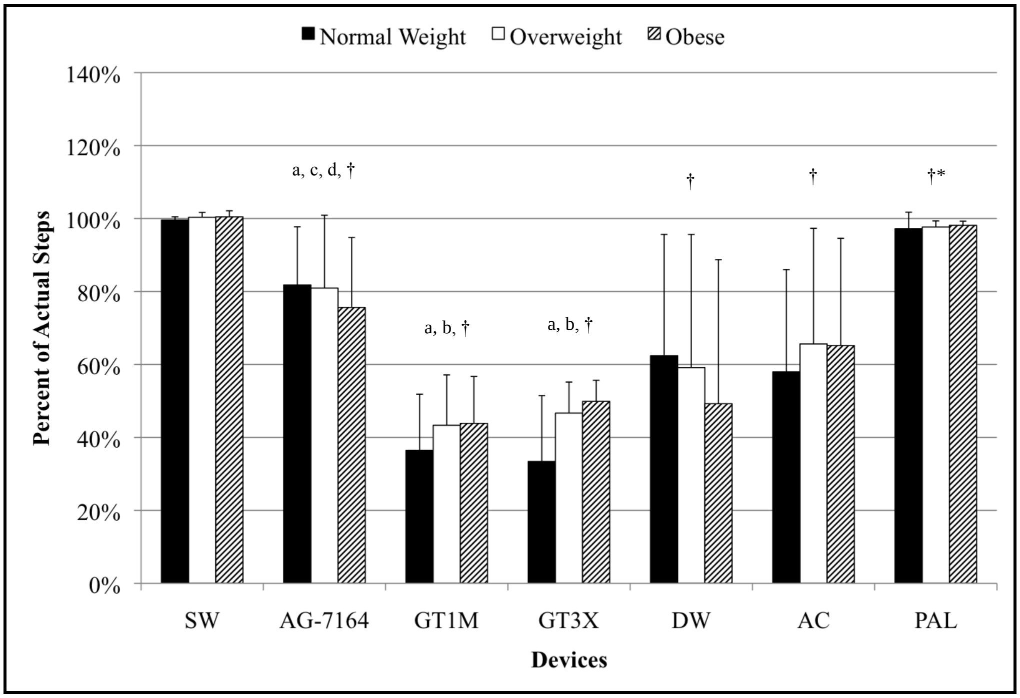

Figure 5-1: Percent of actual steps taken during treadmill walking at 40 m·min^{-1} for all devices compared with hand tally criterion method.

Bars represent mean ± SD

^{a} Significantly different from SW (*P* < 0.001)

^{b} Significantly different from AG7164 (*P* < 0.001); $^{b\#}$ (*P* < 0.01); ^{b*} (*P* < 0.05)

^{c} Significantly different from GT1M (*P* < 0.001)

^{d} Significantly different from GT3X (*P* < 0.001)

† Devices significantly different from hand tally (P < 0.001); †* Differences < 5%

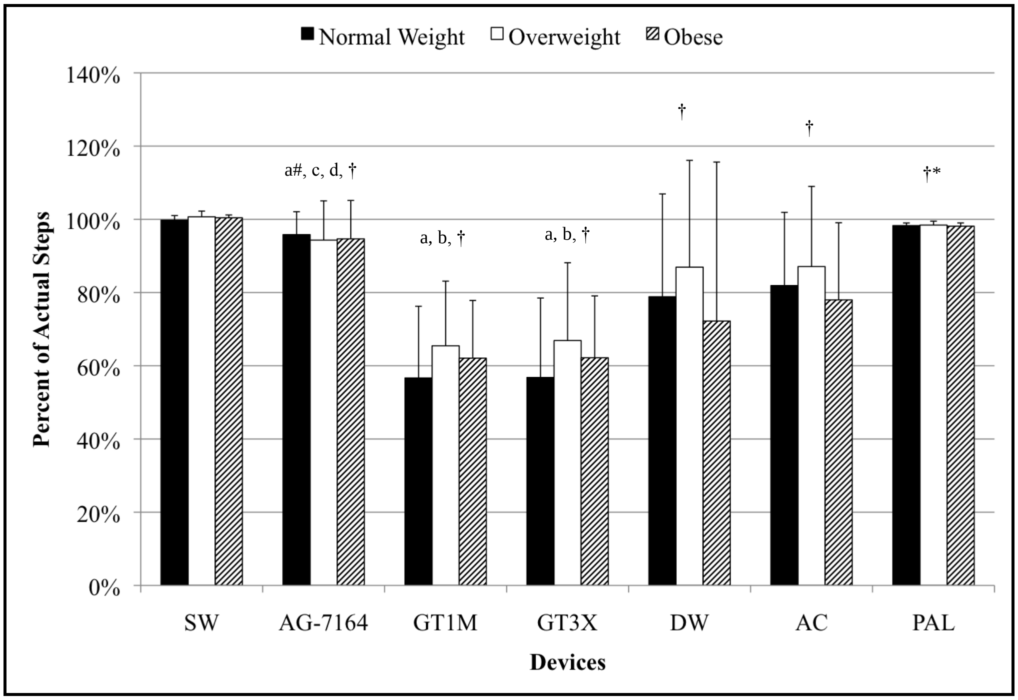

Figure 5-2: Percent of actual steps takend during treadmill walking at 54 m·min^{-1} for all devices compared with hand-tally criterion method.

Bars represent mean ± SD

[a] Significantly different from SW (*P* < 0.001), [a#] (*P* < 0.01)

[b] Significantly different from AG7164 (*P* < 0.001)

[c] Significantly different from GT1M (*P* < 0.001)

[d] Significantly different from GT3X (*P* < 0.001)

† Devices significantly different from hand tally (P < 0.001); †* Differences < 5%

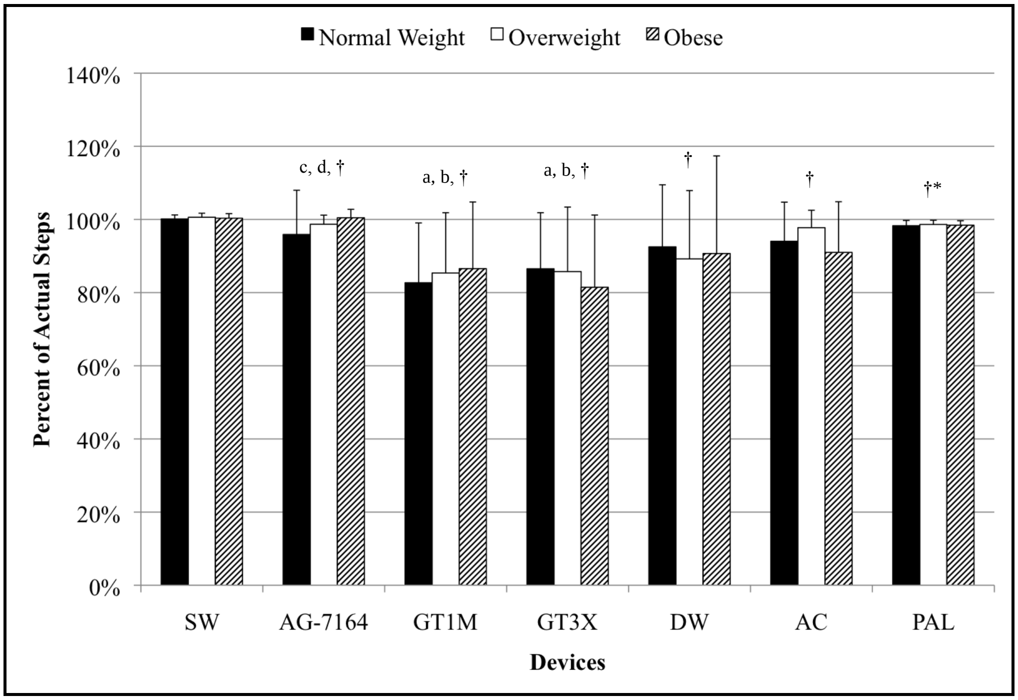

Figure 5-3: Percent of actual steps taken during treadmill walking at 67 m·min^{-1} for all devices compared with hand-tally criterion method.

Bars represent mean ± SD

[a] Significantly different from SW (P < 0.001)

[b] Significantly different from AG7164 (P < 0.001)

[c] Significantly different from GT1M (P < 0.001)

[d] Significantly different from GT3X (P < 0.001)

[†] Devices significantly different from hand tally (P < 0.001); [†*] Differences < 5%

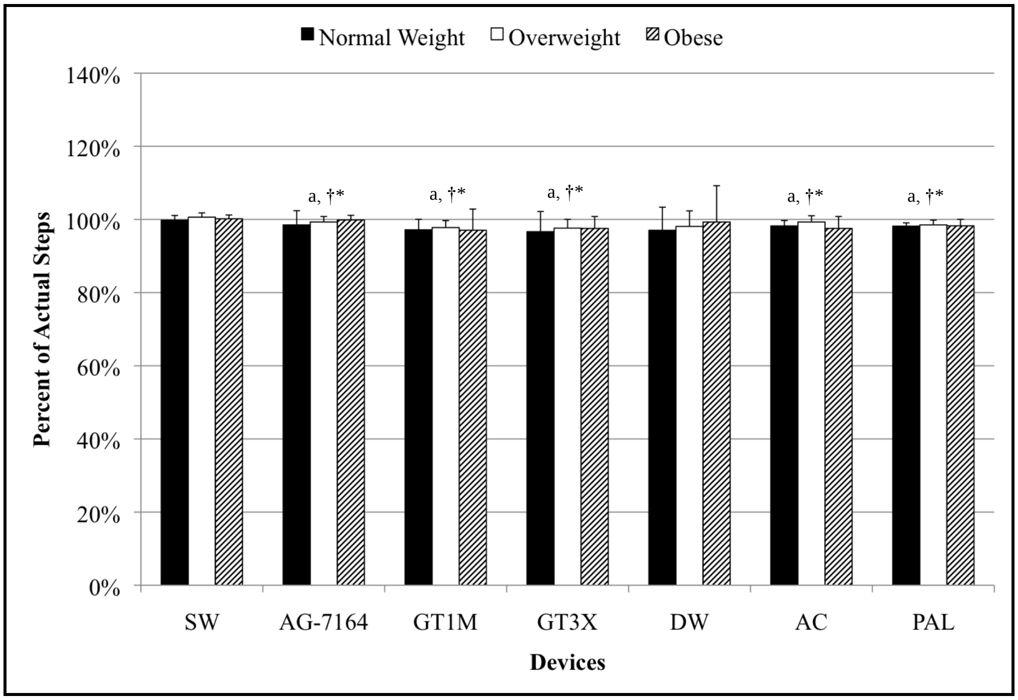

Figure 5-4: Percent of actual steps taken during treadmill walking at 80 $m \cdot min^{-1}$ for all devices compared with hand-tally criterion method.

Bars represent mean ± SD

[a] Significantly different from SW (*P* < 0.001);

† Devices significantly different from hand tally (P < 0.05); †* Differences < 5%

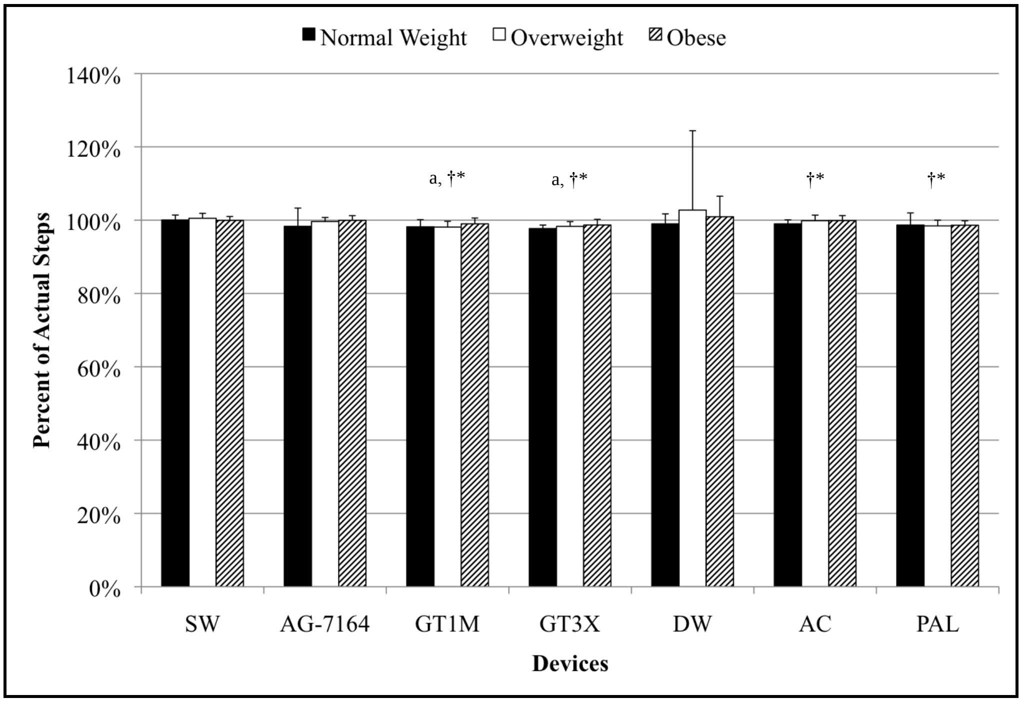

Figure 5-5: Percent of actual steps taken during treadmill walking at 94 m·min^{-1} for all devices compared with hand-tally criterion method.

Bars represent mean ± SD

[a] Significantly different from SW (*P* < 0.001);

[†] Significantly different from hand tally (*P* < 0.001)

[*] Differences less than 5%

During the free-living condition, BMI did not significantly influence the number of steps taken ($P > 0.05$). However, we did find significant differences in step-counting ability among the devices ($P < 0.001$). The AG7164 had the highest level of accuracy when compared with our criterion method (the SW), recording over 99% of steps (99.3% ± 29.6%). The AC recorded close to 80% of steps (77.6% ± 23.1%) while the GT1M, GT3X and DW recorded about 75% of steps (75.6 ± 15.3%, 74.4 ± 13.3%, and 75.3 ± 4%, respectively). Meanwhile, the PAL recorded a just over 70% of actual steps (71.6% ± 19.7%) during the free-living condition. When compared to the criterion method, all devices except the AG7164 were significantly different from the criterion method ($P < 0.05$) (Figure 5-6).

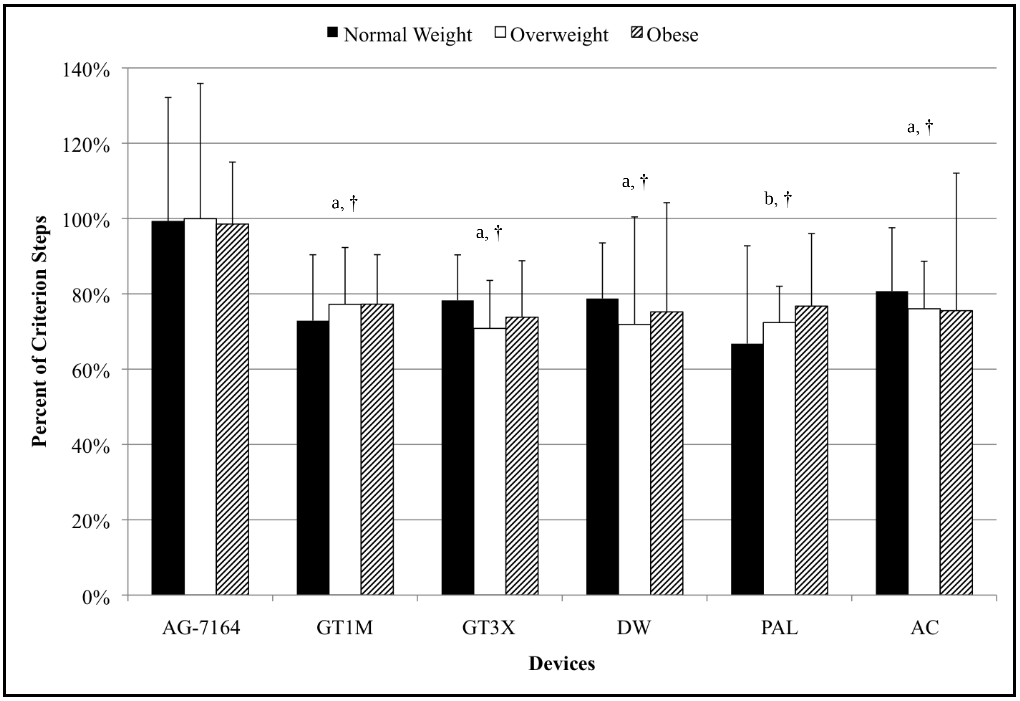

Figure 5-6: Percent of criterion steps for free-living condition

Bars represent mean ± SD. In this phase of the study, the StepWatch ankle-mounted device served as the criterion.

[a] Significantly different from AG7164 ($P \leq 0.001$);

[b] Significantly different from AG7164 ($P < 0.01$);

[†] Significantly different from criterion method ($P \leq 0.001$)

DISCUSSION

The purpose of this study was to compare the validity of several commonly used physical activity monitors in controlled and free-living environments. To our knowledge, this is the first study that has compared these devices to a criterion method in these settings.

During treadmill walking, the PAL was the most accurate device at all speeds, recording 99% of steps taken. These results concur with previous findings from Busse et al. who reported significant correlation (r= 0.93, $P < 0.001$) and reliability (intraclass correlation coefficient (ICC) = 0.95) of the PAL when compared to the SW [33]. The AG7164 also showed relatively high accuracy when compared to the SW, recording over 95% of steps at most speeds. These findings are in agreement with those of Le Masurier et al. [34] who compared the number of steps taken by the AG7164 to hand-tally counts during a walking bout using similar walking speeds as in this study (54, 67, 80, 94, and 107 $m \cdot min^{-1}$). Their study however, did not include any speed lower than 54 $m \cdot min^{-1}$. Nonetheless, compared to the accuracy of the other devices used, the AG7164 may be a suitable device to measure physical activity levels among various populations as it can accurately record steps in individuals who walk at various speeds. In addition, the AG7164 gives very similar step data to the SW during free-living activity, while other waist-mounted activity monitors measure only 75% as many steps as the SW. This explains why Tudor-Locke et al. found that the average U.S. adult takes an average of 9676 ± 107 uncensored steps per day [35] while other pedometer studies [36] have reported much lower numbers. Thus, based on the findings of these study, we can be confident that the uncensored daily step counts reported by Actigraph 7164 in NHANES 2003-2006 are an accurate reflection of ambulatory activity in the U.S. population. Therefore, investigators should feel confident when reporting associations between NHANES step data and health variables.

An interesting finding was the inconsistency in accuracy of the PAL device. Whereas it was highly accurate during treadmill walking (99%), it only recorded around 70% of SW-determined steps during the free-living condition. These findings are in contrast to those of Busse et al. who found that the PAL systematically recorded higher steps than the SW during outdoor activities [33]. However, in Busse et al.'s study the cadence was standardized to a "comfortable walking speed" using a metronome while walking on an outdoor circuit. In our study, we did not control any component of the participant's free-living environment other than to remove the devices if they were performing any activity that did not include walking. The manner in which our participants probably performed their activities during the day (i.e. a large amount of intermittent activity accumulated in short walking bouts [37]) may have influenced the PAL accuracy. We surmise that the PAL may have something similar to the 4-second filter in the Omron pedometer, which does not record steps accumulated in bouts of less than four consecutive seconds; because this is the only way it could be highly accurate for continuous walking but still underestimate 24-hr free-living step counts.

Overall, we did not find a significant effect of BMI on the step count accuracy of these devices while walking on the treadmill, or over the course of a 24-hour period of free-living activity. This conclusion is in agreement with recent findings from Feito et al. [17] who concluded walking speed had a greater influence on step count accuracy than BMI, for accelerometer-based activity monitors. In a similar study, Swartz et al. [38] did not find any significant effect of BMI during treadmill walking when comparing an electronic pedometer (Yamax SW 200) to a hand tally method. Our findings show that at speeds less than 67 $m \cdot min^{-1}$, the DW exhibited considerable inter-individual variability with standard deviations as high as 40% of the mean. This may explain the discrepant results observed by different authors [14, 39].

What it is certain, however, is that regardless of BMI category, the DW records significantly fewer steps during treadmill walking at speeds below 67 m·min^{-1} and in the free-living environment, when compared to the SW. This, however, was the case for most of the devices used in this study.

Another purpose of our study was to compare the accuracy of three generations of the ActiGraph to a criterion method. Our findings suggest that the AG7164 yielded the most comparable data to the SW device, in controlled and free-living conditions. The later generations of the device, GT1M and GT3X, are in close agreement with each other. However, when compared to the SW both devices significantly underestimated steps in the free-living environment, as well as slower treadmill walking speeds (< 67 m·min^{-1}). These findings may be related to upgrades in the internal mechanisms of the newer devices, which decrease the sensitivity to low-frequency accelerations (e.g. noise), thereby improving the accuracy of the device during ambulatory activity [40]. This reduction in sensitivity may be responsible for the GT1M's and GT3X's inability to accurately measure low intensity, or intermittent activities, such as those performed throughout the day [37, 40].

Rothney et al. [40] first described differences between the ActiGraph 7164 and GT1M generations in a mechanical setting and suggested a low-frequency extension may be needed to accurately measure low intensity movements. These findings, lead of the ActiGraph company to introduce a low-frequency extension option in version four (v. 4) of the ActiLife software. This option needs to by chosen by the researcher when initiating the device, thus adjusting the bandwidth at which the GT1M and GT3X record activity levels. Hypothetically, this low-frequency extension makes the devices more sensitive, thereby improving accuracy when measuring low intensity activities. Considering that Rothney et al.'s [40] comparisons were performed using a mechanical setup, and direct comparisons between ActiGraph generations and a reference device have not been performed until now, we opted not to select the low-frequency extension option, when initializing the ActiGraph devices.

Our findings suggest that when compared to a criterion method, the accelerometer-based activity monitors used in this study do not accurately measure walking speeds below 67 $m \cdot min^{-1}$. These findings are significant considering that population studies are reporting daily step counts and making comparisons with health variables , such as blood pressure, cholesterol and waist circumference [41-42]. Despite some inaccuracies in step counting with most pedometers, recent investigators have reported significant improvements in body mass [43] and systolic blood pressure [44] in individuals participating in pedometer based intervention programs that increased their physical activity by approximately 2500 steps per day [43-44]. Therefore, regardless of their limitations, practitioners find that these devices are useful tools to promote physical activity.

This study is not without limitations. First, we used the SW as our criterion method, which even though it has been shown to accurately measure steps during various activities [28, 32], it is not considered a reference method. In addition, because we wanted to compare the ActiGraph devices to their original settings, we did not use the low-frequency extension recently introduced in version 4.4.1 of the Actilife Lifestyle Monitoring System for the GT1M and GT3X. This may limit the GT1M and GT3X's accuracy during the free-living condition as reported by Rothney et al. [40] and in this study. Further studies should look at the effect of the low-frequency extension in step outputs when compared to a criterion method.

In conclusion, our findings suggest BMI does not affect the accuracy of accelerometer-based activity monitors during treadmill walking or daily activity. Furthermore, most devices are not accurate when walking at speeds less than 67 m·min^{-1}. In addition, the AG7164 seems to be more accurate than the GT1M and GT3X for step counting, when the low-frequency extension is not in place.

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APPENDIX

APPENDIX A

INFORMED CONSENT PART III AND IV

Project Title: Accuracy of accelerometers: The role of central adiposity.

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Purpose

You are invited to participate in a research study at the University of Tennessee's Obesity Research Center that examines the accuracy of a number of objective measures (pedometers, accelerometers, SenseWear Pro 2 Armband) to assess physical activity and caloric expenditure.

Procedures

The testing session is composed of two sections. First, we will obtain some basic body composition measures. Second, you will be asked to walk on a treadmill at various speeds while wearing a number of small, pager size devices, along your waistline, ankle and upper arm. Additionally, you will be connected to a computer system through a mouthpiece and hoses, which will allow us to measure the air you exhale. From this, we can calculate the number of calories burned during the exercise. You will be asked to wear comfortable clothing in which you can perform light to moderate physical activity and to refrain from eating 4 hours prior to testing.

Your height, weight, percent body fat, waist circumference will be measured during the first stage of the test, prior to beginning the exercise portion of the session. Weight and percent body fat will be measured using a common device known as a Bod Pod, which will require wearing minimal amounts of clothing to accurately estimate body fat. Usually, a bathing suit is appropriate. This test takes less than 5 minutes during which time you will sit in the Bod Pod chamber. The Bod Pod estimates your body fat by comparing your body weight to your body size.

The exercise will require walking at different speeds: 1.5 mph (slow walk), 2.5 mph (normal walk) and 3.5 mph (fast walk) for five minutes at each speed. The total walking time will be 15 minutes. While walking, the investigator will attach four-pager (4) type devices to your waistline or belt, a step activity monitor to your right ankle and an armband to your left arm. All devices should remain in place while walking on the treadmill. Additionally, you will be asked to breathe through a mouthpiece supported by headgear that will allow for the collection of expired air while you walk on the treadmill.

Your total commitment for the study will be no more than 60 minutes and will be completed in one day.

Risks and Benefits

There are few health risks associated with moderate exercise. These risks include abnormal blood pressure responses and heart rhythm disturbances, as well as musculoskeletal discomfort.

Our screening suggests you have no major diseases/conditions that limit your ability to exercise safely. The risks of participating in this study are equivalent to the risks of activities requiring moderate exertion (yard work, light sport activities, etc.) that you engage in during everyday activities. The benefits to participation in this study include exposure to a number of devices that may provide accurate information about steps taken and calories expended.

Confidentiality
The information obtained from these tests will be treated as privileged and confidential. This information will not be released to any person without your prior written consent. However, the information will be used in research reports or presentations, but your name or any other forms of identity will not be disclosed.

Contact Information
If you have any questions at any time about the study or the procedures, or you experience adverse effects as a result of participating in this study, please contact the investigator Yuri Feito (see page 1 for contact information). If you have questions about your rights as a participant in this study, contact the University of Tennessee's Research Compliance Services of the Office of Research at 865-974-3466.

Right to Ask Questions and to Withdraw
You are free to decide if you want to participate in this study and withdraw from it at any time without penalty. If you decide to withdraw, your data will be destroyed.

Before you sign this form, please ask questions about any aspects of the study that are unclear to you.

Consent
By signing this consent form, I am indicating that I understand and agree to take part in this research study.

____
Your name

___________________________________ ____________
Your signature Date

____
Investigator's Name

___________________________________ ____________
Investigator's Signature Date

APPENDIX B

INFORMED CONSENT FOR PART V

Project Title: Comparison of physical activity monitors in a controlled and free-living environment.

Investigators: Yuri Feito, MS, MPH
David R. Bassett, Jr. PhD

Address: The University of Tennessee; Department of Exercise, Sport and Leisure Studies
1914 Andy Holt Ave.; Knoxville, TN- 37996

Telephone: 865-974-5091 (Yuri Feito)

Purpose

You are invited to participate in a research study at the University of Tennessee that examines seven devices to measure ambulatory activity.

Procedures

After reading this informed consent form, you will be asked if you have any questions regarding the study. Once I have answered your questions, I will ask you to sign the informed consent form, if you wish to participate. The study is divided into three days. Day one will include an initial visit to the Applied Physiology Laboratory at UT where all testing will be performed wearing light clothing (e.g. shorts and t-shirt). Your height, weight, and abdominal circumference will be measured. Then, an elastic belt will be attached to your waist with three small, pager-like devices that measure your steps. An additional device, similar to those on the waist, will be attached to an elastic band around the right ankle and another device will be attached to your mid-thigh with an adhesive dressing. You will then be asked to walk 100 steps on a treadmill at five preselected speeds (1.5, 2.0, 2.5, 3.0 and 3.5 mph). Because of the number of devices used require similar placement on the waist, I will need you to repeat the treadmill walking twice while wearing different devices.

Once you have completed the treadmill portion, you will be sent home with all the devices and you will wear them one more time, during all waking hours. On days two and three, the devices should be worn the entire day, from the time you wake up until the time you go to bed, except when swimming or showering. During these two days, you should go about your normal daily routine.

Risks and Benefits

There are few health risks associated with moderate exercise. These risks include abnormal blood pressures and heart rhythm disturbances, as well as musculoskeletal discomfort. There is a very small likelihood of suffering a heart attack. However, our screening suggests you have no major diseases/conditions that limit your ability to exercise safely. The risks of participating in this study are equivalent to the risks of activities requiring moderate exertion (yard work, light sport activities, etc.). Participation in this study will have no direct benefit to you as a participant. However, it will allow us to determine the accuracy of these devices in a controlled and free-living environment.

Confidentiality

The information obtained from these tests will be treated as privileged and confidential. This information will not be released to any person without your prior written consent. However, the information will be used in research reports or presentations, but your name or any other forms of identity will not be disclosed.

Contact Information

If you have any questions at any time about the study or the procedures, or you experience adverse effects as a result of participating in this study, please contact the investigator Yuri Feito (see page 1 for contact information). If you have questions about your rights as a research participant, contact the University of Tennessee's Research Compliance Services of the Office of Research at 865-974-3466.

Right to Ask Questions and to Withdraw

You are free to decide if you want to participate in this study, and you can withdraw from it at any time without penalty. If you decide to withdraw, your data will be destroyed.

Consent

Before you sign this form, please ask questions about any aspects of the study that are unclear to you. By signing this consent form, I am indicating that I understand and agree to take part in this research study.

____________________	____________________	__________
Name (please print)	Signature	Date
____________________	____________________	__________
Investigator's Name (please print)	Signature	Date

APPENDIX C

PHYSICAL ACTIVITY READINESS QUESTIONNAIRE (PAR-Q) FOR PARTS III – V

PHYSICAL ACTIVITY READINESS QUESTIONNAIRE (PAR-Q)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age and you are not used to being very active, check with your doctor.

No	Yes	
☐	☐	• Has your doctor ever said that you have a heart condition <u>and</u> that you should only do physical activity recommended by a doctor?
☐	☐	• Do you feel pain in your chest when you do physical activity?
☐	☐	• In the past month, have you had chest pain when you were not doing physical activity?
☐	☐	• Do you lose your balance because of dizziness or do you ever lose consciousness?
☐	☐	• Do you have a bone or joint problem that could be made worse by a change in your physical activity?
☐	☐	• Is your doctor currently prescribing drugs (for example water pills) for your blood pressure of heart condition?
☐	☐	• Do you know of <u>any other reason</u> why you should not do physical activity?

Please note: If your health changes so that you then answer YES to any of these questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

If you answered YES to one or more questions

Talk to your doctor by phone or in person BEFORE you start becoming much more physically active of BEFORE you have a fitness appraisal. Tell you doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want as long as you start slowly and build up gradually. Or you may need to restrict your activities to those which are safe for you. Talk to your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

If you answered NO to all questions

If you have answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- Start becoming much more physical active – begin slowly and build up gradually. This is the safest and easiest way to go.
- Take part if a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively.

Delay becoming much more active if:

- You are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better, or
- If you are or may be pregnant – talk to your doctor before you start becoming more active.

I understand that my signature signifies that I have read and understand all the information on the questionnaire, that I have truthfully answered all the questions, and that any question/concerns I may have had have been addressed to my complete satisfaction.

____
Name (please print)

____________________________ ____________________
Signature Date

VITA

Yuri Feito was born in La Havana, Cuba on March 16, 1978. After two short years in the Caribbean Island, his parents fled the communist regime and settled in Madrid, Spain where he lived for 10 years, before coming to the United States. In 1992, Yuri was selected to attend high school at the Maritime and Science Technology (MAST) Academy, a blue-ribbon school of excellence, where he graduated in 1996. That same year, pursuing his passion for sports and exercise, he enrolled in Barry University's exercise science program where he completed a dual degree (BS to MS) in five years. In 2001, he was hired as an exercise physiologist in the cardiac and pulmonary rehabilitation programs at the Wellness Center at Broward General Medical Center in Fort Lauderdale, FL. In 2005, after a short tenure as Barry University's director of fitness and wellness, he returned to Broward General Medical Center as an administrator where he managed the wellness programs; including the fitness center, cardiac & pulmonary rehabilitation departments and employee wellness services throughout the hospital. In 2007, prior to attending the University of Tennessee, he completed a master's in public health from Nova Southeastern University, where he earned the Public Health Research Award from the School of Osteopathic Medicine for his work in cancer research. He graduated the University of Tennessee in 2010 with a Doctor of Philosophy in Exercise and Sport Sciences with a concentration in Exercise Science and a specialization in Exercise Physiology. He is an active member of the American College of Sports Medicine and has recently accepted a faculty position in the department of Sport and Exercise Science at Barry University.