Validation and Reliability of the Hexoskin and FitBit Flex Wearable BIO Collection Devices

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VALIDATION AND RELIABILITY OF THE HEXOSKIN AND FITBIT FLEX WEARABLE BIO-COLLECTION DEVICES

By

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Bachelor of Science in Kinesiology
University of Nevada, Las Vegas
2012

A thesis submitted in partial fulfillment
of the requirements for the

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May 2015
ABSTRACT

Validation and Reliability of the Hexoskin and Fitbit Flex wearable bio-
collection devices

by

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The purpose of this study was to investigate whether the Hexoskin shirt and the
wrist worn Fitbit Flex activity monitor were both valid and reliable for the
physiological functions they were designed to monitor. Heart rate (beats·min⁻¹),
respiratory rate (breaths·min⁻¹), step count, and energy expenditure results were
collected for the Hexoskin. Step count and energy expenditure were collected for
the Fitbit Flex. 49 adolescent participants performed a walking treadmill protocol at
1.5 mph, 2.5 mph, and 3.5 mph for 3 minutes at each speed. 46 subjects returned to
perform the same protocol a second time. 31 of the participants were used to
determine reliability. Each trial required the participants to walk while wearing a
Hexoskin shirt, a Fitbit Flex on their right wrist, a Polar T-31 heart rate monitor,
and to be monitored by an Applied Electrochemistry Moxus Metabolic System.

Hexoskin heart rate correlation was inconsistent between the two protocols with
some minutes/stages being highly related in one protocol and not in the other. A
number of stages showed significant differences in the mean values. Interclass
correlation was acceptable for half of the measurements compared
Hexoskin respiration rate values were highly correlated for the every minute of the first two stages (1.5 mph and 2.5 mph) but showed variations between protocols in the final (3.5 mph). All but one minute’s heart rate value was significantly underestimated. All stages exhibited high interclass correlation scores.

Hexoskin energy expenditure had no stages that were correlated. However, all stages showed no significant differences though the Hexoskin did slightly overestimated caloric count values. The interclass correlation was acceptable for all stages.

Fitbit Flex energy expenditure was not acceptably correlated for any stage, the values were significantly higher than the MOUXS calculated values, and no stage could be considered acceptable for interclass correlation purposes.

Hexoskin step count was highly related only at the 3.5 mph stage. The 1.5 mph and 2.5 mph stages were not correlated and also significantly underestimated the steps taken. Only the 3.5 mph walk could be accepted as reliable.

Fitbit flex step count was not correlated for any stage, the values were significantly lower than the observed count, and no stage could be considered acceptable for interclass correlation.

Overall, the Hexoskin and fitbit Flex do not appear to be acceptable tools for research purposes.
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CHAPTER 1

INTRODUCTION

Over the previous decades, mankind has looked to the computer and the technology that operates it as a tool to improve the quality of life we experience. Early computers were large, bulky, and impractical due to their size. The financial costs to operate them were beyond the means of all but the most well-to-do organizations or persons. In comparison, today’s hand-held smart phones have more computation power than could ever be imagined when the first computers were being invented. It is this strive to continually improve current technology that permits us to advance forward and make it even faster, smaller, and more convenient for the general population to operate.

Most consider February 1946 as the start of the modern age of computers. This when the Electrical Numerical Integrator and Calculator (ENIAC) was officially presented to the public at the Moore School of Electrical Engineering at the University of Pennsylvania [126]. The ENIAC was the first general-purpose computing machine in which mathematical computations were done entirely in an electronic manner [16]. Upon ENIAC’s completion, the machine measured a total of 10 ft. high, 100 ft. long, and 3 ft. deep [16, 49] and had a total power consumption of 174 kilowatts. It could only perform 5,000 addition, 357 multiplication, or 38 division calculations in one second. [16, 40, 45, 49, 121]. By comparison, a modern smartphone, the iPhone 4, which fits in your hand can perform 2 billion operations per second [106]. It is this progression in technology and computational power that allows electronic devices to become smaller and more powerful over time. Moore’s law is a theory that states processor speeds, or the overall processing power for computers will double every two years based on technological advances in
computer design [43]. This continual increase in computer power combined with the continual decrease in actual device size has led to an assortment of measurement instruments that have become highly beneficial and user friendly for all persons. Some of the more promising advances in the application of these new measurement tools have come in the fields of exercise physiology, sports performance, and medical applications. Devices such as heart rate monitors (HRM) and pedometers (Pd) are continually being made smaller, more portable, and suitable for the everyday user to operate. Simultaneously, these devices are acquiring the ability to measure complex physiological functions, measure several of these functions at the same time, and calculate estimated values such as calorie usage, or energy expenditure (EE) in a rapid manner. It is unknown whether advances in technology equate to device designs being valid and/or reliable in a research setting.

Prior to the emergence of portable technology, manual pulse palpation of an artery for heart rate (HR), watching someone’s chest for the respiratory rate (RR), or counting steps (SC) during a test session were the only genuine way to monitor a person’s physiological capacity in a field setting. These methods were also used in laboratory settings as technology first emerged. Because early technology required large and/or sophisticated laboratory equipment and a high level of tester expertise to competently operate, the simple manual measurements were more practical to employ [19]. For these reasons, physiological monitoring devices have continually become smaller, easy to operate, financially feasible, reliable, and valid for not only the common person but also for research professionals in many fields.
Manual data collection, though the norm in early research and still sometimes utilized today, is not free from error and can be costly in terms of time and personnel. Observer based methods lead to numerous measurement errors. Improper assessing techniques, measuring on the wrong spot on a subject's body, inconsistent application, and/or not paying attention to the subject can all lead to missing data which can affect the overall result and any inferred conclusions the study may be trying to determine [33, 113]. According to Pate (1993), observational techniques require more personnel as the number of measurements that are taken increases. Also, the observer must either be in the immediate presence of the subject, usually for a considerable period of time, or must view previously recorded video tapes. Depending on the measurements taken, considerable effort might have to be expended in training observers or the subjects themselves (if they are to record their own data) in the proper and accepted manner to collect said data [91]. Self-recorded data can lead to specific issues as participants may try to “skew” the data in a favorable direction or they may simply record the measurements inaccurately. In a study by Clapp III and Little (1994), 89 women who exercised regularly were asked to evaluate their HR over two aerobic training sessions. During the first session, they used a Polar Eelectro Oy, (Vantage XL, CIC Uniq) HRM to record their HR in beats per minute (bpm). During a second, similar, aerobic session, they gauged their perceived exercise intensity with the Borg Rating of Perceived Exertion Test (RPE) and also stopped at intervals to manually take their pulse by measuring palpations for 6 seconds and multiplying by 10. The corresponding RPE results gave a HR value of 10-15 bpm lower than was actually experienced. Also, manual palpations recorded a HR rate of 10-30 bpm lower than the readings from the HRM in the previous session [23].
The difference in HR in this study shows that manual and device measurements can greatly differ.

However, instruments are not immune to measurement errors. They are limited by their level of precision and sometimes by an individual user’s interpretation of the results. The precision of an instrument refers to the smallest difference between two quantities that the instrument can recognize. As a result, it is not possible to determine with certainty the exact capacity of a result [33]. For instance, reagent strips that measure urinary concentrations through pH levels rely on a fresh unused strip that is blue colored at the start. After exposure to urine, it changes color through different shades of green and finally to yellow when the local pH is reduced. A technician compares the strip’s color change to a standardized chart that represents a color and its corresponding urinary concentration. Individual interpretation of shades of color can give variations in results of the same sample when evaluated by different persons [21]. These random errors can never be completely eliminated because instruments can never make measurements with absolute certainty. However, random errors can be reduced by making measurements with instruments that have better precision and instruments that make the process less qualitative [33].

Almost all aerobic and some anaerobic forms of activity use HR to determine appropriate training intensities for optimal training gains [89]. The technology to measure this element along with other quantifiable aspects such as step count (SC), respiratory rate (RR), and energy expenditure (EE) have become the subject of interest for the research investigator, especially for those in the field of exercise physiology.
Using HR to measure the intensity of exercise has been employed for over 100+ years. In a study by Knox (1940), HR increases were measured during simple exercises for assessment of cardiovascular fitness by administering the precursor of what we know today as the 3 minute step test. Knox commented that HR evaluation testing for health purposes had long been used in medical practice. Knox agreed (as pointed out by Bowen (1903)) that acceleration of the HR begins immediately as exercise starts. Because HR was considered a medical standard for cardiovascular health, especially in cardiac patients, detecting and recording the HR became an area of interest to physicians. This comparison of HR to exercise intensity along with the medical implications of measuring HR has prevailed to this day [9].

The traditional and most common way that RR is evaluated is to watch a person’s chest wall rise and fall. The chest wall is considered by the medical field to consist of all structures on the outside of the body that surrounds the lungs and move with them as they fill with air and expand or release air and contract. The ribcage and abdominal walls are included in this anatomical area [63]. This measurement can be performed easily if the patient is motionless (standing, sitting, or lying down) or the chest can be clearly viewed when breathing. However, observing the rise and fall of the chest wall in persons that are moving during daily activity, exercising, or have clothes on can complicate measuring the RR. Also, measuring the RR in infants, the elderly and persons of abnormal weight may not be possible due to their size and/or physical characteristics that prevent a person from seeing any chest movement. A less used but not unknown way to measure RR is by placing the back of the hand close to the nose to monitor exhaled air [109].
The RR can fluctuate in reaction to certain metabolic demands, physical activity intensities, or in reaction to the presence of a disease such as an infection. An abnormally low or high RR can often indicate the presence of a more serious condition [34, 124]. Some studies have shown that the RR may be a better measurement when compared to other vital signs such as HR and blood pressure (BP) in discriminating between stable patients and patients at risk that would need to be watched closely [109]. RR measurement is a non-invasive way to assess a subject’s health for exercise and medicinal purposes. Long term monitoring can reveal apnea conditions such as tachypnea (abnormally fast RR), bradypnea (abnormally slow RR), or apnea (no RR) [98]. Lastly, RR monitoring can reveal abnormal chest movements in physiological areas such as the ribcage or abdomen. These abnormal movements allow us to evaluate and help predict certain medical conditions by how the chest wall is moving and the related air flow that results from it [98].

A daily SC is a useful tool for health and wellness. The American College of Sports Medicine (ACSM) has recommend persons do at least 30 accumulated minutes of moderate-intensity physical activity on at least 5 days of the week [71]. To accomplish this goal, one of the more common, though not definitively proven activities is a daily count of 10,000 steps. Tudor-Locke1 and Bassett (2012) measured the SC of various groups and found that 6,000-7,000 step were taken in an average day. By also doing a 30 minute walk at a HR gauged moderate intensity, another 3,000-4,000 steps can be easily added. The combination of these two walking scenarios can assist persons to achieve a daily exercise quantity that results in a healthier lifestyle [115]. One longitudinal study by Moreau et al., (2001) gave evidence that by increasing daily SC to ≈ 10,000 a day,
hypertensive women were able to reduce their systolic BP ~ 11mm Hg and body mass ~ 1.3kg after 24 weeks of increased walking. A Pd that accurately measures the daily SC would be a valuable way for persons to increase their daily walking levels by giving them a "hard number" reminder of how close or far they in their daily travels to their fitness goals. Also, it would provide a psychological boost because the device would remind the wearer to increase their SC through alternative activities such as using stairs instead of elevators or walking to nearby stores rather than driving.

EE, or calorie counting has been around since the 1860s [75]. This is the time period that scientists began to investigate analytically the chemical composition of foods. They were particularly interested in measuring the energy (heat) value of foods. Their first investigations were based on livestock food consumption but realized it could also be applied to humans. The calculated caloric value of food can be used as a prediction of EE and subsequently, exercise activity [65]. These caloric values can be used to fight and prevent obesity. The main cause of obesity is an imbalance between calories consumed in food and calories expended in daily activities. A lack of regular exercise contributes to caloric excess. The primary treatment for obesity is dieting, augmented by physical exercise, both of which require an accurate caloric measurement [36]. It is accepted that 1 pound of fat equals 3,500 calories. Losing 1-2 pounds a week is the healthy way to lose weight. However, care must be taken to not lose weight too fast. According to the Mayo Clinic in Rochester, Minnesota, when weight is reduced quickly, it is not only fat that is lost but actually a combination of fat, lean tissues and water weight [17]. Because of this, fast weight loss can easily reduce lean muscle mass below that which is desired. The Mayo Clinic cautions that the type of diet followed can affect what type of body
composition will be lost and that the weight loss can be independent from the number of calories cut from a diet [17]. An accurate measurement of EE can help fight rising obesity rates in the U.S. in the same manner as the measuring of SC.

The aim of the current study is to investigate the validity of two recently released physiological measurement devices. The first device is the Hexoskin wearable body metrics shirt by Hexoskin (Carré Technologies Inc. San Francisco, CA). This device measures HR, RR, EE, and SC. HR and RR are detected through sensors embedded in the shirt. EE and SC are estimated by the use of a 3d accelerometer to measure activity [12]. The second is the Fitbit Flex (Fitbit Inc. San Francisco, CA). It can be worn like a wrist bracelet but the manufacturer also states it can be put in a clothing pocket or placed on the hip with no effect on accuracy [111]. The Fitbit Flex measured SC and estimates EE by use of a 3d accelerometer also [111].

Any device that is to be used for field or lab measurements should follow four important criteria in order to be considered an accurate and dependable data collection instrument. First, the device must measure the factor it claims to measure within an acceptable range of results. Second, it must give consistent results when used in similar settings at different times or places. Third, the financial cost must be within reason and it must be easy for both the participant and the user to use. Fourth, its use should minimally alter the behavior of the subject that it is being applied to so as not to influence a measurement [65].

This study will only address the first criteria; the validity of the Hexoskin and Fitbit Flex to record the physiological functions they claim to measure. Based on the way the Fitbit is worn on the wrist, it is hypothesized that there will be significantly lower
measurements for both EE and SC. It is also hypothesized that there will be significantly lower EE, SC, and RR measurements for the Hexoskin. HR however, it is hypothesized that there will be no significant difference due to the proximity of the imbedded ECG leads in the shirt to the heart.
Purpose of this Study

The purpose of this study is to examine the validity of both the Fitbit Flex and Hexoskin shirt. EE and SC validity will be tested for the Fitbit Flex. EE, SC, RR and HR validity will be tested for the Hexoskin.

Research Hypothesis

There will be significantly lower measurements in both EE and SC for the Fitbit Flex. There will be significantly lower measurements in EE, SC, and RR for the Hexoskin. HR will not have a significant measurement difference.

Significance of the Study

There is great interest in the professional sport and recreational exercise community for small, unobtrusive but yet accurate measurement devices. Instruments that can accurately measure physiological functions while exercising and not be intrusive or influence the mechanics of the subject performing the activity can be of immense value by giving realistic results under actual conditions. Devices that are small and reliable will not only benefit those who exercise, but will be valuable to the medical community as well. Wearable technology along with wireless advances in information transfer can open up a large, new realm of medical observation and evaluation possibilities between patients and physicians. These new devices will help reduce extensive hospital stays and/or bulky, intrusive measuring equipment. Therefore, it is important to evaluate promising new technology that measures physiological functions for accuracy. Coaches, athletes, physicians and patients will likely depend on the information collected to set training intensities, evaluate health conditions, and make medical or rehabilitative
decisions. In this sense, inaccurate results could lead to serious injuries due to training at inappropriate intensities, serious medical problems due to the administration of incorrect medical protocol, underperformance due to not reaching optimal training levels, and possible improper medical treatment and injury to the patient.

Limitations

The study had potential limitations due to the interaction of the monitoring devices used to measure physiological functions. The Hexoskin shirt has ECG sensors embedded in the shirt. These sensors are designed to touch the subject’s skin laterally on each side of the chest, just below the pectoral muscles. This is the same position on the body that the polar transmitter belt is designed to be placed for HR monitoring. The belt was lowered slightly to accommodate the shirt sensors. SC was measured manually and had the potential for observer errors due to simple miscounting or miscounting due to attention from the subject being diverted resulting in wrong final measurements. EE had the potential for calculation errors due to the way the MOXUS monitor recorded HR and pulmonary respiration. Calorie consumption was calculated from the MOXUS by using the measured relative VO2, converting it to an absolute VO2, and multiplying that by the nonprotein respiratory exchange ratio (RER) caloric equivalent. The MOXUS was programmed to record HR and pulmonary ventilation at 15 second intervals. Depending on when the computer recorded those vital signs, the prior or subsequent 15 second interval could be larger or smaller due to this variation in which interval the values were used to average out the vital sign recorded.
Delimitations

The study involved the validity of the Hexoskin shirt and the Fitbit Flex activity monitor during walking and jogging conditions. Both are marketed to be able to measure vital signs not only during these physical activities but while one is sitting, lying down, or sleeping. This study chose to focus on the measurements of walking and jogging in order to concentrate on one aspect that both monitors claim to accurately measure. Measurements during other daily modes were left for further research so they could be analyzed fully and properly without complication. The age population chosen (18-44) was to comply with the ACSM’s guidelines’ for exercise intensities and risk classification. Being of age 18-44 is one factor to be considered “low” risk.” Literature reviewed for this thesis was limited to a historical evolution of previously validated devices that measured vital signs, the application of these devices with the general public, and the benefits to using them. There are no known studies that directly address the Hexoskin or the Fitbit Flex for validity.
CHAPTER 2
REVIEW OF RELATED LITERATURE

Heart Rate

In a study by Knox (1940), an electrocardiogram (ECG) was used during a step test protocol to record the electrical activity of the subjects’ heart via a marking device that used a metal rod and a smoked drum that it marked measurements on. Two saucer-shaped copper discs 2.5 cm. in diameter, using soap as an electrolyte, were attached to the chest with elastic straps. One electrode was applied over the apex beat and the other over the second right costal cartilage. Knox was able to read a satisfactory HR voltage and a uniform base line in this manner. Knox acknowledged, however, the device was only usable for clinical use [13]. The time and effort needed to analyze the tracings on the smoked drum after the test did not allow for practical use of the device in a real time setting or in an expeditious manner.

It was not until 1950 that Himmelstein and Scheiner were able to use newly available technology to invent the cardiotachoscope [51]. Up to this time, tracings on a recording medium had to be developed or analyzed while surgery was in progress. This delay in developing and analyzing recorded heart activity was not beneficial for surgeons. Important information was not able to be used as needed due to the delay. Recognizing the need for a continuous data mechanism for surgical purposes, a direct writing machine connected straight to the ECG was created. These ECG readings were drawn out and were also directed to large cathode ray tubes in the operating room that could be viewed at a distance by all. By having a written record and an immediate viewing of the hearts
function, surgeons were able to better serve patients [51]. Though ECG readings were displayed, there were issues with the presentation of HR information. Many early units did not provide HR data and a technician had to manually calculate it through ECG wave analysis [15]. In 1954 the Cambridge Operating Room Cardioscope from the Cambridge Instrument Company was introduced. The display included a small screen and analog indicators behind a round glass portal of a torpedo-shaped explosion proof housing. The entire device was mounted on a stand and was considered portable for the time. This was the beginning of HRM downsizing [15]. It was not until 1968 and the Hewlett Packard Model 7803A “Monitorscope” that HR information was displayed on a monitor along with ECG readings. The Monitorscope used a horizontal line that would progress across the bottom of the screen with the HR shown as a bar graph. The screen itself was stamped with graduations and numbers for reading of the HR [15]. The first practical portable HRM that was able to take readings while doing various activities was the Holter Monitoring device (HMD). It was previously and is still to this day used as an instrument that continuously records the heart’s electrical activity over a period of time. A series of electrodes are attached to the chest. They are then connected to a small device worn on the patient's belt or neck. The HMD keeps a record of the heart's electrical activity throughout the recording period [52]. Hinkle, Meyer, Stevens, and Carver conducted validity tests in 1967 with two HMD’s. The equipment used was the “Electrocardiocorder” Model 350A and "Electrocardiocorder" Model 350CG by the Avionics Research Products Corporation Medicine Corporation. After recording 151 men performing numerous physical activities over a period of 6-10 hours each, it was determined that the 350A model had a mean error percentage of + 1.32% and the Model
CG had a mean error percentage of +2.13% for ECG and HR reading. [52] HMDs along with the ECG are the primary HRM’s that are used as the standard of measurement for HR that past and current HRM have been validated with [15].

It was these steps in technology that gave rise to the portable HRM’s used today. The ECG and HMD are accurate, but they are not 100% appropriate for use in field settings due to cost, size, and complexity of their use [67]. Polar Electro Oy (Polar Electro, Finland) introduced the first, small retail HRM, the Tunturi Pulser, in 1978. All one had to do was press a finger to a sensor on the face of the monitor and the HR was displayed. It was able to be worn on the waist on a belt but was unwieldy and awkward [67]. In 1983 the Sport Tester PE 2000 was unveiled. It was the first wireless HRM to use electronic field data transfer. Comparable HRs were taken and compared between the Sport Tester 2000 and an ECG. The results showed that the mean bpm’s measured by the ECG and the PE 2000 varied from each other by up to 5 bpm. At one time, the difference was ± 10 bpm at each workload. Though the difference in results between the two was attributed to the method that each used to calculated HR, the Sport Tester 2000 was deemed valuable for measuring HR during exercise due to its size and ease of use [67]. Building on this, in 1984 Polar Electro Oy then introduced the Sport Tester PE 3000. This was the first HRM equipped with a computer interface and a transmission done by a magnetic field. Vogelaere et al. (1986) compared HR readings of the Sport Tester PE 3000 to those obtained from an ECG. 20 subjects performed 50 minutes of variable walking exercises. The Sports Tester PE 3000 was shown to be a valid alternative for measuring HR. There was no significant difference in HR readings between it and the ECG (P < 0.05) and there was a high correlation coefficient (r) of 0.982 [120].
Leger and Thivierge, (1988), tested the validity of 13 commercially available HRM’s by comparing their measured values with simultaneous ECG readings. The HRMs, all wireless, varied in the types of leads used for ECG readings. Conventional chest leads, earlobe connections and alternate lead placements were evaluated. Validity was measured by using several ergometric devices. All 13 HR monitors were classified as follows; if \( r \) was \( \geq 0.93 \) and the standard error (SE) of the estimate was less than 6.8\%, it was considered “excellent”; if \( (r) > 0.65 \) and the SE was \( 6.8 \pm 15\% \), is was “good; or lastly, if \( (r) < 0.65 \) and the SE > 15\%, it was “inadequate.” There were excellent correlations between readings obtained by ECG and HRM using conventional chest electrodes. 3 of 4 HRM using conventional chest electrodes had excellent readings. Only one of the 4 had a good rating. All other HRMs that used alternate types of leads or the earlobe as a placement were deemed inadequate [69].

In 1995, improving on advances in technology, Polar introduced the Vantage NV HRM. This was the first HRM to use a coded transmission from a chest transmitter to a computer receiver. It measured electrical activity of the heart but also had a Polar R-R recording (beat-to-beat) system and an analysis system [67]. Kaikkonen, Karppinen and Laukkanen (1997) used the Vantage NV HRM to study recovery and overtraining vital signs in male orienteers. Orienteering being a sport where participants use navigational skills, a map, and compass to quickly traverse unfamiliar terrain going from one point to another. The participants all stated that the vital sign measurements that they recorded while orienteering were easy to understand and analyze both at home and during training [58]. Technology had advanced to a point that complex readings were being made available to the general public in ways they could understand. Also, no longer were the
readings limited to the HRM when measured but could be saved into a computer. They then assessed the timing accuracy of the Polar Vantage NV for measurement of the R-R intervals. In 99.9% of the R-R intervals measured, there was only a ±5 ms. difference between the Polar Vantage NV field device and the included Polar R-R recorder software used for home analysis. Technology was becoming more complex but at the same time it was becoming easier to use and understand for the common person.

By the mid-2000’s, HRM costs had dropped considerably. At the same time there was an increase in computation power in HRM technology. Vanderlei, Silva, Pastre, Azevedo, and Godoy (2008) wanted to test the Polar S810i HRM for validity. Unlike an ECG and its accompanying software that was expensive and hard to access, the Polar S810i HRM was an everyday device that was far cheaper than the ECG but gave the same measurements. The R-R intervals were recorded by electrodes attached to an elastic band placed around the thorax. Electronic signals were then continuously conveyed and kept in a receiver for storage and calculation purposes. 15 subjects were attached to both devices and told to lay in the supine position for 20 minutes. Next they pedaled on a stationary bike using the pre-established intensity for an additional 20 minutes. A comparison of the two instruments over the two 20 minute periods was made. No significant differences were found between the two in terms of the number of R-R intervals recorded during rest (ECG; 401.70 ± 16.14, Polar; 401.10 ± 16.10) or during exercise. (ECG; 634.40 ± 6.62, Polar 635.00 ± 6.17). The (r) between the two for the resting period was 1.000 and the (r) for the exercise period was 0.981. It was reasoned that the Polar S810i HRM was arguably as reliable in its measurements as those obtained by an ECG [119]. This was an important moment because now instead of having to
depend on large, expensive, and normally inaccessible instruments to have vital statistics recorded in a usable manner, they could now be collected inexpensively and in a far more easy-to-use manner. Most of HRMs up this period utilized a chest strap transmitter positioned around an individual’s thoracic region with a wrist-watch style receiver worn on the wrist that usually displayed a continuous HR reading. These chest leads were the most valid overall measuring devices for the time. [68]

By 2007, however, technology allowed for the development of wrist-watch styled HRMs that did not required a chest lead. Not needing a chest strap made this style of HRM more practical and easier to put on and wear. Also, because there was no constricting chest lead to wear throughout the day, it had a better comfort level. This was considered a big leap for HRMs [68].

The Smarthealth wrist-watch style HRM uses the back battery cover as the one electrode and has two front casing electrodes, located above and below the wristwatch display. A HR is displayed when the index and middle finger of the opposing hand are placed on the front casing electrodes. This HRM measures the electronic signals emitted from heart beats that pass through the body as the heart contracts. Lee and Gorelick (2011) tested the Smarthealth watch on twenty-five individuals. They participated in 3-min periods of standing, walking at 2.0 mph, walking at 3.5 mph, jogging at 4.5 mph, and running at 6.0 mph. HR was concurrently measured and recorded at 60-sec intervals using three methods: the Smarthealth wristwatch, a Polar Vantage XL monitor with an accompanying chest strap and an ECG which served as the standard measurement. The accuracy and validity of The Polar Vantage XL had been previously determined by Godsen, Carroll, and Stone (1991). 2633 heart beats were measured during the treadmill
activities. The Vantage XL yielded HR values ± 6 bpm 95% of the time when compared to the ECG. HR from the Smarthealth watch were highly correlated with those from the ECG, \((r) \geq .95\) and the SE was below 5 bpm for all measurements. The \((r)\) between the Smarthealth watch and Polar HRM was \(\geq .97\) while the SE between the two was < 3.7 bpm. The Smarthealth watch did exhibit a reduced ability to detect HR during the 4.5 and 6.0 mph run. The Smarthealth watch seemed to be a valid HRM when standing and during treadmill exercise involving walking and jogging.

A second type of wrist worn HRM are those that use Photoplethymography (PPG). PPG is a noninvasive optical monitoring technique where peripheral blood volume changes in living tissue due to the pulse that radiates from a heartbeat. In brief, there is a light source on one side of a tissue bed and a light detector on the other. The difference in light absorption during pressure waves in the blood due to a heartbeat can be analyzed for HR [88]. PPG has been used clinically since 1936. The bulky equipment (lighting and sensors) that was needed for it to be utilized barred its widespread use at the time. The practical uses of PPG did not become available until 1962 with the invention of light-emitting diodes (LED) [95]. By 2007, advances in technology made PPG portable and thus practical. Selvaraj, Santhosh, and Anand (2007) helped established the accuracy of PPG by concurrently measuring a finger-tip PPG signal to an ECG reading. Both were measured on stationary subjects during normal and deep breathing conditions. \((r)\) Was calculated at 0.9698 during normal breathing and 0.7389 with normal but deep respiration. Zhang, Pi, and Liu (2014) expanded on the PPG idea by applying new analyzing techniques to the procedure. They compared the PPG wrist-watch to an ECG during intense exercise. 12 subjects ran at a top speed of 15 km/hour. The results showed
that the SE between the two was 2.34 bpm and that the (r) was 0.992 [125]. This can potentially lay the groundwork for wearable devices such as smart-watches which use PPG signals to monitor HR for fitness. The biggest advantage for PPG is that it is simpler to build, less costly, but is still as reliable [99].

**Respiratory Rate**

Just as a person’s HR is an important vital sign that can be a measure of cardiovascular health or help establish exercise intensity, so too can the monitoring of the RR be applied. RR is defined as the rate of ventilation, or the number of breaths taken in a set amount of time. The respiratory system delivers oxygen to tissues and simultaneously removes carbon dioxide. These actions help to regulate the partial pressure of both gases in arterial blood. This gas regulation is partially accomplished by setting the RR which in turn is regulated by chemoreceptors that sense oxygen, carbon dioxide and pH levels, mechanoreceptors of the lungs, and the respiratory centers of the medulla and pons [124].

Unlike the measuring of a HR which specifically requires the detection of electrical impulses generated from the heart, RR can be measured in a number of ways. Because we can either view the chest wall moving, measure chest displacement as it rises and fall, or measure exhaled breath in a variety of ways, the recording of RR has developed in two directions; contact and non-contact methods.

Contact RR monitoring instruments are usually based on measuring one of the following parameters: respiratory sounds, respiratory airflow, respiratory related chest or abdominal movements, respiratory CO₂ emission and oximetry probe SpO₂. RR can also
be derived from an ECG [3]. Non-contact methods include Doppler radar, optical, and thermal based sensing [3]. Though there are numerous ways to record the RR, there is no one way that is the considered the “gold standard” with which to compare RR data against like an ECG is used to compare HRM readings for validity [39, 94]. This lack of a baseline standard to compare devices for RR recording makes the evaluation of devices a relative comparison against each other rather than an absolute that is measured against an accepted standard magnitude taken under similar conditions [37]. Sensitivity, specificity, and how a monitor is applied to the patient for measurement are the factors that need to be considered in RR recording [53].

In the previously mentioned study by Knox (1940) that involved early ECG measurements, RR was also measured using the same copper discs, soap as an electrolyte, and elastic chest straps that measured HR. Knox wanted to see if the device could be used to detect sinus arrhythmia. Sinus arrhythmia is a cyclic variation of the sinus rhythm of more than 10% variance in HR or 120ms due to respiratory breathing. HR changes as you breathe. When you breathe in, your heart rate speeds up slightly. When you breathe out, your heart rate slows back down [74]. Knox concurred with Treadgold (1930) that sinus arrhythmia was usually of no significance in regards to cardiac efficiency unless it was extremely marked. In most cases, sinus arrhythmia disappears during the exercise but becomes prominent after it, perhaps due to deeper breathing. In the case of the step test administered by Knox (1940), it could be easily detect the sinus arrhythmia in the RR after the test was complete [61]. But just as for HR, the RR data had to be extracted manually from the recording device and analyzed after
the subject had completed the test. Depending on the skill and competency of the analyzing technician, some data may be missed or wrongly recorded.

Early studies such as those conducted by Peabody, Gregory, and Willis (1979) and Arnson, Rau, and Dixon (1981) showed that even though technology was advancing, the monitoring of RR was not something that could be done through a machine with an acceptable amount of validity or reliability. Peabody et al. (1979) evaluated whether a commonly used for-that-time thoracic impedance device (Hewlett-Packard Cardiorespirograph, model No. 78200 series) could detect changes in RR and conditions of apnea in infants. The infants RR was also visually monitored by the attending nurses. RR was recorded on a polygraph (Rikendenski model KA series, Grass model 7) for analysis and comparison purposes after the observations were complete to check the accuracy of the readings. Twenty-one infants were watched over forty-three separate occasions for 2-7 hours each time. 145 hours total were recorded. In the 145 hours, 544 apneas > 15 seconds were manually recorded by the polygraph device. 487 were concurrently detected by the thoracic impedance device while the nurse’s observation only caught 179 of those 487. Thoracic impedance monitors were determined to be unreliable because they could detect only a fixed duration of respiratory pause. The pause time being manually set when the instrument was employed. Other apneic conditions of < 15 seconds were detected by the polygraph but not the impedance device. Thoracic impedance was also sensitive to outside influences that were unavoidable in a clinical setting such as apneic episodes induced by established nursing procedures and airway obstruction. Finally, ineffective breathing patterns such as disorganized breathing,
obstructive apnea, and paradoxical breathing can undetectable by both thoracic impedance monitoring and observation [93].

Arnson et al. (1985) compared the results from two simultaneously employed RR monitor systems and from a visually observed RR count on ten subjects over an 8.5 hour period. The first device used ECG impedance as a way to measure RR. Two ECG leads were placed on a subject’s chest and the RR was measured by chest lead displacement due to chest expansion. The second used a thermal sensor attached to a nasal cannula whose electrical resistance varies inversely with the temperature of exhaled gasses [6]. The observed RR was 21.61 ± 4.91, impedance was 23.87 ± 13.54, and thermal was 16.49 ± 9.91. All the measurements had a significant difference (p < 0.10). They discerned a pattern of increasing divergence from the counted RR. The impedance results gave progressively higher readings and the thermal gave progressively lower [6].

In the 1980’s, computers were just beginning to emerging as useful measurement aids. However, their use for RR measurement was still considered technically complicated, expensive, and prone to measurement errors due to a lack of variation discretion caused by low computation power. Computers were unable to distinguish the RR’s of various respiratory situations such as movement, coughing, hiccups, or respiratory maladies [31]. Some early non-invasive RR research was conducted by Erikson, Berggren, and Hallgren (1986). Erikson et al. (1986) used two 15” strain gauges of a mercury in silastic (Medimatic, Hellerup, Denmark). These strain gauges are designed to alter voltage inside a metal band when tension is applied. They were calibrated with the least-squares method. This method measures the circumferential changes of the rib cage and abdominal compartments during respiration in a supine
position. It was known that alterations to RR could be influenced by the prevalent invasive monitoring techniques used at that time such as nose-clips or mouthpieces. These items had an effect of producing sensory stimulus or psychological feedbacks that interfered with RR [42]. The strain gauge measurements were compared to capnographic measuring. Fourteen subjects were studied for thirty minutes. For the first ten minute period, only the strain gauge and capnographic were used. During the second ten minutes, nose clips and a mouthpiece were added. And finally, in the last ten minutes, the original measurement condition using strain gauges and capnographics were employed again. The following RR results were recorded on average for all subjects; 1-10 minutes, (13.3 ± 3.84 breaths per minute), 11-20 minutes (11.9 ± 3.01 breaths per minute), 21-30 minutes (13.1 ± 3.41 breaths per minute). The overall absolute difference was 4.7 ± 3.66% which compared well with the values obtained in previous studies with similar strain gauges and calibrations [1, 20]. These strain gauges were sensitive to body movements and positional changes though. They would require either a cooperative or immobile subject to obtain an accurate RR [31].

Larsonn and Staun (1999) performed research using fiber optics to measure RR. RR for eighteen subjects was recorded simultaneously by three methods; fiber optics, capnography and observation by three nurses. Data collection was done in four 3-minute periods with the subjects lying still. A total of 516 minutes were analyzed. Paired t-tests showed a mean of 0.50 and 0.30 more recorded breaths by the fiber optics compared to observation and capnography. The 95% confidence interval of the differences between the three was -0.5 to +1.5 and the mean RR was fourteen breaths per minute [66]. These
results showed that the fiber optic RR recordings were satisfactory when compared to capnography and observation counting.

Folke, Granstedt, Hok, and Scheer (2002) conducted a similar experiment that compared three non-invasive methods of respiratory monitoring. Ten healthy subject would be simultaneously monitored by 1) A qualified observer (experienced registered nurse) that marked inspired breaths by pressing a push-button. 2) A publically available fiber optic device that detected water droplet condensation of undiluted expired air (Optovent RR9700, Optovent AB, Sweden). 3) A prototype device that detected variable CO₂ concentrations between a subjects inspired and expired gases. The participant’s would perform a series of actions that simulated different conditions that the RR monitors would have to accommodate. First, the subjects breathed normally for ten minutes while remaining still. During this time they held their breath twice for as long as they wanted. After this “normal” period, two provocation tests were made. The initial provocation test had all the subjects move all their limbs and touch their face in order to test the tolerance of the measuring methods to movement. The other provocation test used augmented nasal oxygen (L/min) for three minutes. These tests and the order were repeated twice for all subjects. The prototype sensor was mounted in two different ways for each test to determine which was more apt to detect RR. The first had the sensor placed just over the patient’s mouth via a head harness [B] while in the second, it was positioned in the same manner except that a chin strap was added for additional anchoring in position [A]. The results corresponded to the conclusions made by Hok, Wiklund, & Henneberg (1993) that RR monitor sensitivity, specificity, and how it is applied to a patient created discrepancies in data collection. In 75.1% of the 4219 respirations total, all 3 methods
detected the subject’s breath. But when compared to each other and as stand-alone data, it was found that the prototype in either position A or B detected RR with the highest accuracy, missing an average of 7.3% of the total RR. The fiber optic sensor missed 7.6% while the nurse’s visual observation missed 21.4%.

Though we cannot compare RR to a set “gold” standard, the two studies by Larsonn and Staun (1999) and Folke et al. (2002) show that human error can play a significant part of an inaccurate RR measure and that RR monitors had evolved to a more reliable way of monitoring than observation. The argument could be made that they were now a better method for RR evaluation purposes [37].

There have been numerous other advances in RR technology methods that have been determined to be valid and reliable sources of measurement. Most advances have been in the area of non-contact and non-invasive methods. These methods are considered best for monitoring as it takes out the compliance factor or immobile requirement that was required from a patient with earlier monitors. There was a concern with earlier technology and measurement interference from either patient movement or monitor displacement issues. Technology had become advanced enough that computer processing was capable of making high numbers of calculations and could better differentiate between breathing and other artifact disturbances such as coughing, sneezing, or forms of apnea [3].

Murthy, Pavlidist, and Tsiamyrtzi (2004) tested a thermal based RR monitor that used infra-red imaging to detect heat changes in the air surrounding the mouth and nose during respiration. A Phoenix Camera (Indigo, Goleta, CA) was used for monitoring but utilized new computation methods to determine exhaled breath temperature. An advanced
statistical algorithm based on the methods of moments and Jeffrey’s divergence measure was used to correct previously encountered errors during data collection. Jeffrey’s divergence measure was determined to be the most efficient for image noise removal based on the statistical models of wavelet coefficients [110]. By using the tip of the nose as a reference point, the RR for ten subjects was measured from a distance of 6-8 feet. Thermal data was compared to data collected by a piezo strap transducer wrapped around the subject’s diaphragm. Overall, the new computation measurements recorded a 92% accuracy when compared to the piezo strap [80]. Not only was the RR monitor accurate but could be done at a distance, making it useful for uncooperative subjects.

Tan, Saatchi, Elphick, and Burke (2010) researched the use of a visual based (non-contact) respiration monitoring system that would be cost effective, accurate and easy to use. The application recorded a subject during respiration and used computer analysis to determine the RR. This visual method was based on the comparison of a currently recorded image and sequential ones collected prior. Each was marked with a unique time stamp. An algorithm was designed to perform image subtraction. The computer compared the current pixel size of the chest area of a recorded subject with 0.5 seconds worth of pixel sizes from previous images. This combining of images occurred for every image recorded. Combined images became a binary image that contained the magnitude of the scenery changes during the 0.5 combination period. The increasing pixel values indicated inspiration. A second algorithm was developed to detect repetitive respiration sequences to assist in the measuring of RR [112]. The video image was pilot tested on only one subject. Along with this visual method; thermal based, air flow based, and stain gage based systems were simultaneously employed for comparison purposes.
After ten minutes the visual method measured RR at 19.4 breaths per minute. A manual count of RR, and all other monitors measured RR at 20 breaths per minute. The advantage of this RR collection was the ability to record the breathing of an individual with a standard video device or smart phone and store or transport those images to a medical facility for off-line analysis [112].

**Calorie Expenditure**

Calorimetry is the scientific field that focuses on the measurement of heat production. By measuring the quantity of heat produced from foods, we can evaluate the energy we can extract from them for everyday use [65]. This is based on the fact that heat evolved from biochemical reactions in the body are exactly the same as those measured when food is converted into the same end-products by simple combustion in a calorimeter [75]. Therefore, we can measure the EE of various physical activities by the measurements of the heat we generate to an outside medium [90] or by the comparing the differences in oxygen consumption vs carbon dioxide production during rest and steady-state exercise [27]. Measuring calories, or heat expenditure for the purpose of weight loss or the measure of EE during exercise has been around since the mid 1700’s. It was at this time that scientists discovered that heat given off from living beings was the result of the oxidation of carbon and the formation of carbon gas when oxygen was consumed [75]. For every liter of oxygen used, a known amount of heat is released contingent on the type of nutrient oxidized [57]. There are three techniques that are commonly used as the base measurement for evaluating EE in a newly designed devices. Direct calorimetry, indirect calorimetry and doubly labeled water [97]. All three are highly accurate and precise for measuring EE but all are not very practical for everyday research due to either the high
cost of materials needed, large intrusive equipment that impairs or modifies normal behavior during measurements, and/or the amount of time needed by the tester and testee to properly utilize the methods for the best results [97].

The earliest used method for EE measurement was direct calorimetry. This was the direct measurement of generated heat from a living being. It is defined by the generalized equation that is referred to as the Principle of Energy Conservation [57];

\[
M = (R + C + K + E) + S
\]

Where M is heat production from metabolism, R is the radiant heat exchange, C is the convective heat transfer, K is the conductive heat transfer, E is the evaporative heat transfer, and S is the rate of storage of body heat. Direct calorimetry requires individuals to be sequestered in a special chamber. This chamber has a secondary medium (water or ice for example) surrounding it. Heat is transferred from the inside of the chamber to the surrounding medium where it absorbs the generated body heat. This change in the medium temperature can be measured and the EE can be determined for the person inside and for the activity they were doing [65]. These direct calorimetry chambers have an excellent accuracy rate (~1%) and a precision rate ~2-3%) [97]. However, there are significant factors that make them impractical for all but the most precise and important research. The construction of these special chambers is extremely expensive. Also, they cannot be built excessively large. This means that the types of the activities performed inside are limited to certain specific tasks. This makes direct calorimetry impractical for the study of EE during normal daily physical activities or for use on large populations of beings [65].
The second method is the use of indirect calorimetry. Indirect calorimetry measures the consumption of oxygen that closely correlates with heat production [65]. Subjects utilizing indirect calorimetry use a portable respiratory gas-exchange monitor. This method removes the requirement for a special chamber and a lab setting like direct calorimetry uses [97]. Instead, respiratory monitors in the form of a face mask or mouth piece are worn by the subject to collect respiration gases as well as collect a gas samplings [65]. These devices have a general accuracy rate of ~2-3% [65]. However, just like the direct calorimetry method, the equipment can be expensive. Because the monitors are worn by a subject, they allow for a greater measure of freedom for the type of activities that can be recorded by removing the confining and limited chamber space of direct calorimetry [75]. This does not mean the freedom granted by the device is without limitations. The monitors and masks still restrict to a certain degree the type of activity that can be performed and recorded. Activities are limited to those that the participant is comfortable doing with the appropriate worn equipment and can be safely performed [65]. Because indirect calorimetry is carried out on an individual basis, it can be a fairly time-consuming process that is more suited for smaller studies with continuous gas exchange measures limited to 1-5 hours [97].

To put it simply, the difference between direct and indirect calorimetry is that direct calorimetry measures total heat loss from the body while indirect calorimetry measures total energy production by the body.

The last method for EE measuring is the use of doubly-labeled water. It is considered an indirect calorimetry method [97]. Doubly-labeled water is a noninvasive procedure involving the ingestion of a quantity of water labeled with the known
concentration of naturally occurring hydrogen and oxygen stable isotopes. These isotopes are known respectively as deuterium and $^{18}\text{O}$. As energy is expended in the body, carbon dioxide and water are produced as by-products. Deuterium is eliminated from the body in the water by-products while $^{18}\text{O}$ is eliminated in both the water and carbon dioxide by-products [72].

Formula for $^{18}\text{O}$ elimination [72];

$$\text{CO}^{18}\text{O} + \text{H}_2\text{O} = \text{HCOO}^{18}\text{O}^- + \text{H}^+ = \text{CO}_2 + \text{H}_2^{18}\text{O}$$

Formula for deuterium elimination; [210]

$$^{2}\text{H} + \text{H}_2\text{O} = ^{2}\text{HHO}$$

The differences between the two isotope elimination rates in the end-products can be compared to one other and then used to calculate the EE. This method can be performed by a wide range of individuals over long periods of time. It is normally conducted for a period of 4 to 21 days [97]. This makes doubly-labeled water advantageous for analyzing habitual EE patterns in a long-term analysis of one’s activity habits. Because urine specimens only need to be collected periodically, it lessens the interference by the researcher on a subject’s normal activity during the day [72]. This makes the doubly-labeled water method the “gold standard” for EE measurements in free-living conditions as there is no equipment or confining space interference issues [72]. However, the cost and scarcity of the two isotopes and the expertise of the trained technicians required to analyze their concentrations via mass spectrometry prohibits the use of doubly-labeled water in large epidemiological studies [122]. Doubly-labeled water has an accuracy rate of $\sim$1\% and a precision rate of $\sim$4-7\% [97].
When all three methods are compared to one another, indirect calorimetry becomes the most commonly preferred method for EE measurements by researchers. The advances in technology have made it possible to adapt this method to several data collection methods, all the while improving on the sensitivity and time scale required for long-term studies. The reducing size of EE computers and their concurrent increase in computational power have given rise to large advancements in the measurement of EE in the uninterrupted everyday life of persons performing various physical activities [32].

Because HR has a linear association with exercise intensity, it is correct to associate HR with the measurement of respiratory gases that are created during exercise. The concurrent increase in oxygen consumption can be used to evaluate EE values [84]. However, directly measuring VO$_2$ can be difficult. Attempts have been made to develop ways to estimate EE based on HR [2, 92]. Oja, Ilmarinen, and Louhevaara (1982) evaluated this premise by monitoring the vital signs of 9 postman in Finland. All were evaluated with numerous lab VO$_2$ tests during different treadmill walks. This was done for various speeds and loads and also for several cycle ergometer tests. The VO$_2$ results were then compared to equivalent tasks performed in the field during mail delivery duties. Oja et al. acknowledged that for a given VO$_2$, HR would be higher in static muscle contractions over dynamic [60] and higher in small muscle over large muscle mass during similar work [108]. These variations indicate errors for the estimation of VO$_2$ from HR due to the various workloads measured. Because most of the comparisons between the two were no more accurate than a “lottery drawing” for results, Oja et al. determined that HR and VO$_2$ could only be correlated when the VO$_2$ test activity resembled the actual work performed. Researchers also had to account for subject
emotions, the frequently of stress [65], and environmental temperature [54]. All of these could cause an increase in HR without a proportional rise in oxygen consumption.

One way to account for these variations in work intensities and to regulate for the subject’s weight, age, and outside influences was to use an observer based EE calculation system. The manual based method is grounded on an observer watching someone perform an activity and recording the time they spend doing various portions of that activity. Time spent walking, sitting, standing, and the activity itself with seeming intensities are all recorded. Calculations for EE are then made based on previous research for that activity and the established EE values that had been already been pre-determined to perform them. Charts like those created by Spitzer and Hettinger (1965) or from other literature sources [29] have the energy cost of numerous activities pre-recorded for researchers to use in their investigations. Subjects would be observed for a period of time with the nature and duration of all activities recorded on a diary card. The time intervals on the diary card would be divided into as small as needed time periods to best divide out the activities performed. The observer can then properly figure the EE for that period. At the end of the period, the mean value of all measurements would be calculated. The time in the activity would then be multiplied by a pre-determined chart element. This adjustment would correct the EE tally for things such as body weight, age, gender, and all other influences for the day’s activity costs in calories. Though the method does not interfere with the work procedure, it is time consuming, expensive, and requires a disciplined and well-trained observer to record the pertinent information [2].

In 1982, there were only two commercially available instruments that were not only portable but also able to measure oxygen consumption directly. The first was the
“Oxylog” (P.K. Morgan Ltd. Chatham, Kent, U.K.). The Oxylog gas monitor weighed 2.6 kg, and was worn on the waist. It collected VO₂ respiration gases via an oronasal mask held to the face by an elastic harness. The mask was connected to the monitor by flexible hosing. This greatly reduced any restrictions on the physical activity being performed by the wearer [75]. When compared to Douglas-Bag respiration collection and analyzation under resting conditions, the Oxylog was shown to be accurate to within 3% and have a precision of 6% [97]. The second instrument was the “Oxycon P” (Mijnhardt, Netherlands). This apparatus was worn like a backpack and weighed 3.5 kg. The fraction of Oxygen in the exhaled air is measured by polarographic methods. [75] Polarography is a specific type of measurement where an electrodes potential is altered in a linear fashion from the initial potential to the final potential by some interaction. In this case, the change in expired O₂ causes a change in electrical potential [73]. A flowmeter is used to measure pulmonary ventilation. The Oxycon P was determined to have an accuracy of 1% [54]. Despite both instruments being smaller and portable for the time, they were both expensive and still had some limitations. Limitations being mostly in situations where the subject was uncomfortable while wearing the equipment and performing the activity being monitored [97].

Harrison, Brown, and Belyavin (1982) tested the Oxylog monitor using 4 male subjects who performed four quick cycle ergonmeter experiments. All four experiments involved similar criteria. Pedal for three to ten minutes, 50 rev·min⁻¹ at 30-50 W. The subjects O₂ exchange was measured by the Oxylog and a Parkinson Cowan dry gas meter for accuracy at 1 minute intervals. The only difference in the 4 tests was the type of gas collection device that was used. Experiment one used the Oxylog provided mask,
experiment two used an R.A.F. mask, experiment three used a mouthpiece, and experiment four a valve box. 433 comparative measurements were made in total. Averaging the mean differences between the two devices for VO$_2$ showed a non-significant 1.5% underestimation by the Oxylog. Results were also able to show a significant relationship between the measured VO$_2$ and work rate ($p < 0.05$). The conclusion derived was that the Oxylog was sufficient for reliable determination of VO$_2$ and thus EE in the field.

Horwat and Meyer (1988) tested the Oxycon P. This EE device was evaluated along with simultaneous measurements of HR recordings, self-evaluation using the RPE scale by the subject, and the EE calculated by the observer based charts from Spitzer and Hettinger (1965) with all observed values entered into a small computer. Fourteen subjects worked thirty-five periods of between four to sixteen minutes in eleven different steel and metallurgy factory work situations. All wore both the Oxycon P backpack and a telemetric HELLTGE system for HR recording. Observation entries were made on a Hewlett-Packard 71 pocket computer for calculation of EE by activity and RPE was recorded at the end of each activity period.

The Oxycon P underestimated the VO$_2$ by 3.5% with an (r) of 0.96 when compared to actual VO$_2$ reading. It overestimated HR values with the VO$_2$ by 24.6%, $r = 0.81$. Howert and Meyer concurred with Oja et al. (1982) that the VO$_2$ - HR relationship has to be established by comparing the two during the work that is similarly performed and under the same environmental conditions. The observation based computer calculations overestimated EE by 5-8% and had an (r) of 0.94. RPE was the least favorable with an overestimation of 26.6% and an (r) of 0.45. The study showed that
even though HR is still used, it is as accurate or as valid for EE as could be achieved by other devices such as the Oxycon P. But the overwhelming advantage of the HR method for EE is that it is an inexpensive, simple, and non-invasive compared to the other methods. One advantage for the pocket computer observation method is that it can be slightly amended to provide measurements in respect to postures, movement types, and type of work done. Along with HR, it too can be a beneficial tool for EE calculation by being able to describe what is done and how much EE it costs for the subject [54].

By The mid 1990’s, technology had evolved in ways that allowed for EE measurements to be taken without VO₂ measurements. No longer was there a requirement for invasive equipment or devices that were uncomfortable to wear or that affected normal activities performed throughout the day. Computers microprocessors were then small and powerful enough to be worn on a wrist, ankle, or on the waist without the need for bulky respiration gas analysis devices. EE could now be analyzed by subject acceleration measurements that could be converted to a digital form and then to numeric values for calculation purposes by proprietary algorithms [25].

Melanson and Freedson (1995) tested the validity of one of the earliest models to use this new technology, the CSA (Computer Science and Applications Inc. activity monitor). What made the CSA unique was the fact it was one of the first EE monitors that could be worn as previously stated; on the wrist, ankle, or waist [78]. Fifteen subjects were tested on three occasions with monitored treadmill tests. Subjects slow walked (4.8 km·h⁻¹), fast walked (6.4 km·h⁻¹), or jogged (8.1 km·h⁻¹) on each of the three visits at 0%, 3%, and 6% grade for 8 minutes each. Measurements were compared to a HRM (Quantum XL HeartWatch, Polar Electro, Inc., Finland), an on-line computer based O₂
acquisition system (Hans Rudolph, Kansa City, MO), and a dry gas meter (Rayfield Equipment, Waitsfield, VT) for validity purposes. One non-factor discovered was that the CSA did not discriminate for EE based on the treadmill grade and activity counts, only for treadmill speed ($r = 0.00-0.03, p > 0.05$). At grade, the CSA activity counts were significantly and similarity correlated with measured EE ($r = 0.66$ ankle, 0.80 waist, 0.81 wrist). The ankle position was the most accurate for predicted mean EE by being within 1% but had a large SEE at 0.85 (11.4%). For actual mean EE, there was no significant difference between the CSA and actual readings with a $-2.86 \pm 3.08$ kcal·min$^{-1}$.

Koehler, Marees, Braun, and Schaanzer (2013) evaluated two portable sensors for EE assessment during high-intensity running. Most EE devices are geared for the common population and the average fitness enthusiast. But as shown by Oja et al. (1982) and Horwat and Meyer (1988), EE measurements during certain exertion periods need to be made with relation to the specific activity being performed. In this case, EE was examined during high intensity running conditions. The first monitor was an Actiheart chest monitor (CamNtech Ltd., Cambridge, United Kingdom) which combines readings for HR and uniaxial acceleration to calculate EE. The monitor had already been validated for sedentary activities and low intensity exercise but had yet to be validated in high intensity activities [14]. The second was the SenseWear Pro3 armband (BodyMedia, Pittsburgh, USA) that measured biaxial acceleration, body heat lost, and galvanic skin response. Just like the Actiheart monitor, it was previously validated but in free-living conditions only [11]. Twenty-nine male endurance and strength trained athletes volunteered for a treadmill running exercise using a 1% grade. The speed started at 2.8 m·s$^{-1}$ and increased by 0.4 m·s$^{-1}$ every 5 minutes until a speed of 4.8 m·s$^{-1}$ was reached or
exhaustion occurred. EE was validated by measuring oxygen uptake and carbon dioxide production with a Zan 600 spirograph (Zan, Oberthulba, Germany) and HR was kept by a Polar S610 recorder (Polar Electro, Finland) [30].

The Actiheart monitor underestimated EE for all speeds by 1.1 to 8.3 kcal·min⁻¹ and had correlation with the Zann 600 of (r) = 0.61. HR was evaluated correctly with the Actiheart when compared to a Polar 610 recorder with activity counts increasing only between 2.8 and 3.6 m·s⁻¹. The SenseWear armband significantly underestimated EE for all speeds by 1.0 to 9.5 kcal·min⁻¹ and the correlation with the Zann 600 was (r) = 0.66. Both devices underestimated EE during high-intensity running [30].

One of the more popular brands of activity monitors for purchase by the public is made by Fitbit Inc.in San Francisco. The Fitbit line includes at least six different styles of monitors that vary in their individual measurements and functions. All models monitor both EE and SC by use of a tri-axial accelerometer and proprietor algorithms. Collected information is uploaded to a computer via wireless technology for the user to see. Fitbit claims that the device can be worn in multiple ways beyond the normal positions of on the wrist or hip. It can also be placed in a pocket or attached to a sports bra without compromising accuracy [111]. Noah, Spierer, Gu, and Bronner (2013) compared both a Fitbit Tracker and a Fitbit Ultra against indirect calorimetry for accuracy. One of each Fitbit type mentioned was attached to the belt of each participant on each body side. Indirect calorimetry used a telemetric gas analysis system (K4b2 Cosmed Inc., Rome, Italy) that has been validated in previous research [111]. It is comprised of a small metabolic analyzer, battery pack and face-mask to collect and measure respiration gases [82]. Twenty-three subjects performed four stages of testing.
Stage one began with 6 minutes of sitting in a chair. The next three stage were walking and running on a treadmill for 6 minutes each; 3.5 miles mph at 0% incline, 3.5 mph at 5% incline, 5.5 mph at 0% incline. Though the ICC for Fitbit and Fitbit Ultra in respect to average kcal expended was high (r =0.94–0.97), neither was valid when compared to the K4b2; r =0.18. The Fitbit Tracker was between 88 – 90% accurate and the Fitbit Ultra was between 91–113% for EE compared to the K4b2. It was determined that the Fitbit devices under-estimated the EE that was measured in the walk and jog activities [82].

As can be discerned, EE is a complex item to calculate due to many factors being involved. What type of activity and what level of exertion arguably are the major contributors to the errors in measurements that are recorded. Just as a person’s physiological and biological data must be entered into the EE monitor for accurate calculations of EE, so too must the predetermined EE of the activity and intensity level at which the individual will perform at be accounted for also.

**Step Count**

The use of a Pd for the measurement of physical activity is not a new idea. The ancient Romans used a crude form of a Pd called a hodometer to measure the distance a soldier would travel on foot per day. The hodometer operated by an elaborate gear assembly attached to a wagon wheel. The gears were calibrated to measure a standard Romans mile. When the one mile distance was reached via the appropriate number of rotations from the wagon wheel, a pebble would be dropped into a bowl. Though not the exact same thing as a Pd, the idea of measuring distance walked by the user is the same [55]. This method to measure distance is still used in the form of a survey wheel used by surveyors and construction personal. Approximately 500 years ago, drawings from
Leonardo da Vinci show that he also envisioned a mechanical Pd based on the Roman hodometer to be used for military applications [102].

History has attributed the creation of what is considered to be the predecessor to the modern Pd to both French inventor Jean Fernel and Thomas Jefferson, the 3rd president of the United States. Fernel is widely considered to have actually invented the first Pd in 1525. It was shaped like a watch with 4 dials (units, tens, hundreds, thousands) all linked by small gears. The user would fasten the device to a belt on the left of his body. A line would then be run to a matching lever on the right knee. As one walked, the cord would pull the lever, and a pointer on a bottom dial would advance by one unit for each step. The Pd would be introduced to America by Thomas Jefferson in the 1780’s. Jefferson purchased a Pd while on a trip to France. However, it is not known if he kept the Pd as is or modified it to a different configuration. It is well known, however, that Jefferson did promote the use of the Pd for counting steps from correspondence letters that have been archived and preserved to this day [86].

The current idea of counting steps was made popular in Japan in 1965 by Dr. Y. Hatano. Hatano invented and marketed a Pd called the manpo-kei, or as translated, "10,000 steps meter." Hatano determined the average person in Japan took 3,500 to 5,000 daily steps. He calculated that by increasing the daily SC to 10,000, people could burn about twenty percent of their daily caloric intake and be thinner and healthier [50]. By 1985, the idea of 10,000 daily steps had reached the United States. It had been determined that U.S. adults take approximately 6,500 steps/day [22]. Many public health information booklets from various governments then started to recommend 10,000 steps as the daily goal for adults and 12,000 for youth [Appendix A]. For adults, 10,000 steps
equates to roughly 5 miles of walking and it burns between 300 to 400 calories [22].

These goals can be achieved with an active lifestyle that includes a 30-minute walk each day [116]. Tudor-Locke and Bassett (2004) introduced the concept of a graduated step index for healthy adults that helped correlate the steps taken each day with the health of a person. The lower the daily steps, the lower the perceived health of that person. It was later expanded by Tudor-Locke, Johnson, and Katzmarzyk (2009). See table #1; Step Classification below.

Table 1: Step Classification

1) < 2,500 steps/day (‘basal activity’)
2) 2,500-4,999 steps/day (‘limited activity’)
3) 5,000-7,499 steps/day (‘low active’)
4) 7,500-9,999 steps/day (‘somewhat active’)
5) 10,000-12,499 steps/day (‘active’)
6) ≥ 12,500 steps/day (‘highly active’)

Most Pds used for research utilize either a spring-levered or piezo-electric accelerometer mechanism. Spring-levered Pds have a spring suspended horizontal lever arm. The lever arm moves vertically (up and down) in response to vertical accelerations of the hip. The motion of the body causes the lever to open and close an electrical circuit. When the lever arm moves with the appropriate force above the sensitivity threshold, electrical contact is made and a step is registered [24]. The vertical force most often used is a hip acceleration of ≥ 0.35g [70]. However, a spring-loaded Pd must be used in the vertical plane to work properly. Tilting the Pd in the wrong plane can create inaccurate measurements or no measurements at all [47]. Previous research has shown that this tilt
of a Pd was the most important factor for accuracy, affecting results more than a person’s waist circumference or BMI when EE was calculated [59].

Piezo-electric Pds utilize a horizontal cantilevered beam with a weight on one end. When accelerations above the sensitivity threshold occur, the beam compresses a piezo-electric crystal. This compression generates voltage in proportion to the beam acceleration. These voltage oscillations are used to record steps [24]. Piezoelectric Pds were shown to be less affected by tilt in comparison to spring-lever Pds. However, they cost more than their spring-lever equivalents [47].

There is now a third measurement mechanism that Pds use for SC. It is a magnetic reed proximity switch. It utilizes a spring-suspended horizontal lever arm. But instead of a lever arm being used to close an electrical circuit, a magnet is attached to the lever arm. The magnetic field causes two overlapping pieces of metal encased in a glass cylinder to touch, resulting in a step being counted. [96].

The use of a Pd to measure habitual physical activity has the following advantages for the monitoring of EE and daily activity: 1) There is no modification of everyday activity. Pds are small, simple and worn in ways (on waist, belt or wrist) that do not have an impact on normal daily movements [24]. The standardized steps-per-day unit of measurement allows for universal interpretation [117]. 2) Because they are inexpensive ($10-$160 USD) and simple to use, Pds can be applied to large numbers of subjects quickly and efficiently [117]. 3) Pds can be used for long period of time from 24 hours to 7 days for a valid assessment of EE [30]. 4) A Pd can avoid measurement biases caused by inactivity or intermittent activity. This gives it a more accurate measured quantity of a subject’s actual physical activity level [26]. 5) Pds can estimate physical
activity in terms of a period of time such as a day or days as opposed to being limited to a single activity and its corresponding interval [26].

But just as there are benefits to Pds, there are drawbacks. 1) The accuracy of the Pd can be compromised by outside incidental happenings that interfere with the measurement such as the walking surface being soft like sand or weather conditions that affect normal walking patterns like high winds or rain [70]. 2) There can be uncertainty as to what bodily movements are being recoded. Just as walking can affect the pedometer level arms, so too can other activities such as skipping, hopping or jumping [104]. Both going up steps at low velocity [8] or riding a bike [59] will give a lower SC than is actually performed. Any activity that results in the lever arm moving needs to be evaluated and standardized for that specific activity’s equivalent SC [77]. 3) Deducing the final SC from the pedometer scores can challenging. The number of variables such as frequency, duration and intensity of the physical activity need to be considered before a final SC can be finalized [59]. 4) Based on the prior disadvantage, Craig et al (2010) and McNamara et al. (2010) have both commented on the lack of standardization in terms of how data was reported from earlier Pds. A strong need exists for a Pd protocol to assure accurate measurements among participants of various weight classifications (normal, overweight, and obese), movements (hop, skip, jump) and/or intensity levels (low, intermediate, or high) [47].

A study done by Shepherd, Toloza, McClung, and Schmalzried (1999) compared the accuracy of numerous SC devices, one of which was a digital pedometer worn on the belt line compared with an observed SC measurement. Twenty-nine subjects were assessed while they quickly walked 400 meters, slowly walked 10 meters, and then
ascended and descended a flight of 11 standard stairs. The results were given in terms of the final SC for all subjects as well as a comparison of results between those subjects with a BMI < 30 and those with a BMI of ≥ 30. The digital pedometer had an overall error rate of 2.82%. However, the slow walking and stair protocol were extremely high in the individual overall error calculations with an error rate of 15.5% and 19.9% respectively. When the subjects SC was compared based on BMI scores, the absolute error difference was more pronounced in obese participants (BMI > 30) with an error rate of 6.12% and 1.56% in those of BMI < 30. The highest error rate for obese persons was the stair ascent with 30.5% as oppose to 14.1% in all others. Absolute error of the digital pedometer was positively correlated with body mass index ($r$) = 0.792, $p < 0.00$ and weight ($r$) = 0.753, $p < 0.00$. When the results from the pedometer were grouped by gender, the mean absolute error was 1.81% for men and 3.06% but it was not statistically significant ($p = 0.45$) [100]. This research shows the importance of accounting for factors such as the activity and BMI of the user in order to properly measure their activity level.

Leicht and Crowther (2007) performed Pd testing that compared the validity of the measured SC when subjects walked over different surfaces such as concrete, grass, dry beach sand, wet beach sand. The gender of the participants was also recorded for comparison purposes. Fifty-two persons volunteered and performed six 150 meter walks on four different surfaces. All wore a YAMAX SW-700 Digiwalker pedometer. The time of the walks, visually observed SC, and the SC recorded by the Pd were compared. Walking over dry beach sand significantly reduced walking speed (concrete, 5.6 ± 0.5 km·hr⁻¹; grass, 5.6 ± 0.5 km·hr⁻¹; dry beach sand, 5.0 ± 0.5 km·hr⁻¹; wet beach sand, 5.4 ± 0.5 km·hr⁻¹) and increased the observed SC (concrete, 190 ± 13; grass, 186 ± 12; dry
beach sand, 207 ± 12; wet beach sand, 207 ± 12) and the Pd SC (concrete, 195 ± 14; grass, 191 ± 14; dry beach sand, 213 ± 15: wet beach sand, 201 ± 16) Compared with males, females registered a greater number of Pd steps (204 ± 18 vs 197 ± 15) and a greater absolute (9 ± 12 vs 3 ± 7 steps) and relative (4.46 ± 5.72 vs 1.63 ± 3.57%) Pd error walking over dry beach sand. The conclusion was that walking on a soft surfaces significantly slowed walking speeds and at the same time increased Pd error for females as compared to males. The study showed the need to account for the gender of the subject and the walking surface for proper SC tallies during normal, daily movement. Documenting the influences of these factors would be beneficial for future Pd measurements [70].

Smith and Schroeder (2008) tested Pd accuracy while subjects were walking, skipping, galloping, sliding, and hopping. One hundred-two college students wore a Walk4Life LS-7010 pedometer at mid-thigh on the right and left of the hip. All subjects executed the previously mentioned movements over a 26 meter hardwood court. SC was also visually counted by the researchers with a hand counter. All movements were done in a brisk manner. When all participants reached the end of the court they stopped immediately for recording purposes. Significant differences were evident between the hand tally counts and readings from the right and left pedometers during all movements (p < 0.01). Mean error was lowest between the hand tally and the average of the right and left pedometers while walking (-1.35 ± 1.60) and hopping (-2.94 ± 2.33), and increased while sliding (-6.42 ± 4.78), galloping (-8.22 ± 4.63), and skipping (-8.30 ± 4.45). Results indicate the Pd may not consistently register the vertical force produced by the trail foot contact, the lead foot contact, or a combination of the two while skipping, galloping, and
sliding. Just as shown by Leicht and Crowther (2007), the activity being performed can influence the Pd measurement capability just like a soft or unstable walking surface [104].

During the same time that Smith and Schroeder (2008) were performing Pd accuracy tests on various movements, Ayabe, Aoki, Ishii, Takayama, and Tanaka (2008) were examining Pd accuracy while subjects ascended and descended stairs and when performing a bench stepping exercise. As shown by Shepherd et al. (1999), stair ascent and descent could have an effect on a SC if it was not taken into account. Ten healthy men performed two different experiments to measure SC during stair and step climbing. All wore three commercial Pds (DW-800, YM, HJ-700IT; OM, Lifecorder; KZ) for SC and were visually observed as well. The first measurement taken was while they walked up and down a staircase with eleven, 18 cm high steps. Second, they performed a step protocol by using 10, 20 and 30 cm high platforms. There were 5 stages for each test that measurement for ascent and descent were taken. Each stage had a faster rate of climb and descent. The rates were 40, 50, 80, 100 and finally 120 steps•min\(^{-1}\). All the Pds underestimated the SC during stair climbing at slower stepping rates and/or the lower platform heights. During the stair ascending and descending and the bench stepping exercise using 20 to 30 cm high platforms at 80 to 120 steps • min\(^{-1}\), the magnitude of the measurement error was -3.8 ± 10.8% for the KZ, -2.1 ± 9.8% for YM and -11.0 ± 18.9% for OM. These results indicate that the KZ and the YM can accurately assess the number of steps during stair climbing using 20 to 30 cm high platforms at 80 to 120 steps • min\(^{-1}\) [8].
Several studies have been done to test the SC correlation of the various Fitbit models. Because the Fitbit activity monitors measure both SC and EE, making sure both readings are valid is important for healthy living and monitoring. Part of the Fitbit testing involved using the Fitbit Tracker and the Fitbit Ultra for validity. Testing was done by Noah et al. (2013) for EE and also included SC measurements. SC was measured against an industry-standard accelerometer, (Actical, Philips Respironics, Inc., Andover, MA) during the walk and job treadmill stages (5 miles mph at 0% incline, 3.5 mph at 5% incline, 5.5 mph at 0% incline). There were no differences in mean steps taken between the Fitbit Tracker and Fitbit Ultra devices and there was no significant difference between the Fitbits and the Actical monitors (p ≤ 0.05) with r = (0.90 – 0.96) [82].

Takacs, Pollock, Guenther, Bahar, Napier, and Hunt (2014) tested the validity of the Fitbit One activity monitor by attaching one device to the belt above each hip and one in the front pants pocket of the dominant leg. Motion cameras and visual observations were used to compare the Fitbit results for accuracy. Thirty subjects walked on a treadmill for 25 minutes total in five, 5 minute stages that varied in speed (0.90 m·s⁻¹, 1.12 m·s⁻¹, 1.33 m·s⁻¹, 1.54 m·s⁻¹, and 1.78 m·s⁻¹). The relative error for all three Fitbit Ones at every speed was less than 1.3%. The three devices had an inter-device reliability ≥ 0.95

New Technology

With today’s technological advances in smartphones, Wi-Fi capabilities, and wearable bio-feedback clothing, the interest to create new methods to measure vital signs has become a growing field of research. The use of any of these items can be extremely
helpful in many fields of learning. Training for a sport and actual real-time performance would be greatly improved. The incorporation of these devices could give instant bio-feedback information for both the athlete and the coach to view. This would allow for instant modification or analysis of an activity’s intensity in a real-time setting [123].

One example is the relationship between RR and the anaerobic threshold (AT). RR has been shown to be a valid indicator for anaerobic exercise and AT [19]. AT is defined as a break point in an individual’s metabolic production of blood lactate (La) during exercise [76]. La levels will initially increase in a steady, linear manner that corresponds with lower levels of exercise intensity. The body can eliminate any La accumulation during this stage [107]. At a certain point or threshold, La production will shift from this linear La increase to a rapid, disproportioned escalation that the body will not be able to eliminate at the same rate as it is being metabolically produced. After the AT is crossed, metabolic acidosis will increase sharply while exercise time to exhaustion will decrease quickly due to high levels of La in the blood [107]. The exercise intensity point just before the AT has been termed the maximum lactate steady state (MLSS). The MLSS is considered the “gold standard” in AT assessment [7]. MLSS is the point where the body’s accumulation of La is balanced with its ability to eliminate it. It is a La concentration and exercise intensity that can be maintained overtime without a continual blood lactate accumulation [10]. It is considered to be the ideal training point for athletes. RR has been shown to be a valid measure to measure the AT [18, 19]. A portable or wearable device that accurately measures RR, especially with real-time analysis, would be beneficial to a coach or an athlete for the training purposes. Training intensities could
be adjusted directly to the MLSS as the athlete performs for maximum performance gains.

For the population as a whole, there is an increasing need for practical medical applications that will help monitor the health status for a large number of persons while at the same time keeping the rising costs of medical attention down [81]. One way to do this would be to develop precise medical bio-feedback clothing or smartphone applications that would monitor the health status of a person independent of a medical facility. Coupled with the speed and versatility of Wi-Fi internet, a subject could be monitored by computer from a hospital. A physician’s direct attention would only be required when certain vital sign criteria were met. By taking monitoring equipment that up to today could only be accessed in a hospital and placing it right into a home setting, hospital space could be freed up for the most critical of patients requiring immediate attention. Lesser medical conditions could be monitored in the safety and comfort of their own home [123]. This would be extremely helpful for being able to provide medical attention to the ever-increasing elderly population, infants, and those affected by chronic diseases due to changes in lifestyle populations [123].

The advent of home monitoring would not only reduce medical costs, but would enable persons to become more involved in their own care and treatment. No longer would patients be unaware of their vital signs or status. This home-based monitoring would make medical treatment more patient-centric. Wearable bio-feedback clothes and smartphones would give patients greater self-awareness of their basic parameters such as HR, BP, or body temperature as well as how that data relates to their diet, exercise, or physical activities for the day [5]. This increased awareness also leads to patients being
able to gain their functional freedom faster. Studies have shown that when persons have knowledge of their activity level, duration of performance, corresponding HR, RR, and EE values and they know they are being monitored, they are psychologically influenced to become more active. Patient begin to measure their own progress in relation to prior achievements. They try to walk farther in a day, push physical limits in performance, and be more active. This creates a positive increase in exercise, diet, and healthy behavior habits [28, 101]. Remote monitoring of vital signs means that certain medical situations such as those caused by chronic disease, the beginning or ending of a medication prescription, or post hospitalization complications can possibly be caught early enough and treated. This immediate treatment would happen without having to wait until a return visit or a possible hospital admittance to realize there is a medical issue to deal with [28]. One monitoring aspect that has resulted in an increased level of patient-awareness and involvement with personal health has been the linking of physical activity and/or diet to social media groups. When groups of persons can see and compare their own health status and activity levels with others, it has been acknowledged that most persons in the group will have an enhanced interest in their health [5]. They begin to measure their own achievements to those of others. Whether it is the psychological need to compete with others, the personal shame of not doing as much as others such as walking a certain amount of distance in a day, or just the fun of comparing results, social media and the sharing of data has been shown to positively influence a person’s involvement in a healthy lifestyle [5].

However, for any home based or personal system to work, all portions of it need to be mobile, user-friendly and unobtrusive. Bio-feedback clothing must be comfortable
to the wearer and has to consider many factors. How much does it weigh? Even the smallest of weight can become burdensome and feel heavy after a sustained period of time. Is it ergonomically suited for the activity it is measuring? Just because a device can measure a physiological function does not mean that its use or body placement will allow certain activities to be fully performed without interference. If it is to be worn for extended periods of time, is it made of comfortable material? Is it easy to put on, take off, and clean? Any sensors used in the clothing must be must be small, low-weight, low-power and wireless [38, 123]. Smartphone applications must be simple and easy to use. Medical information can be confusing for most people and keeping the shared information as simple and understandable as possible is a necessity. The apps must not only be simple but must also be user friendly. Not everyone is proficient with technology and this can create a sense of anxiety or dread when having to utilize it. This is a concern especially directed to the elderly population. Technology such as smart phones or even a computer can be a challenge for them to use. These are items that they did not grow up with and employ a completely different way to do things that they have consistently been doing all their lives. Smartphone apps should try to employ voice commands in their method of operation. This would be extremely useful for those that are bedridden, unable to move due to injury or paralysis, and for those that have a hard time manually manipulating the phone itself due to arthritis or pain [56, 123].

But the idea of evaluating physiological functions does not just apply to home monitoring. It can also be applied to persons during the performance of their chosen work. This is especially helpful for those that work in high stress or physically demanding fields. To be able to evaluate the vital signs of a person in the performance of
dangerous, high-risk jobs can help keep those persons safe from physical exhaustion or over exertion. Situations such as these can lead to heart attacks, respiratory issues, or any of a number of maladies that can be prevented by having knowledge of how that activity is affecting someone [65]. Biofeedback shirts have been used to monitor fire-fighter vital signs while performing such duties as climbing flights of stairs or searching on hands-and-knees for rescue victims. Comparisons have been made between the exertion levels of these activities both in and out of their full turn-out gear [103]. Construction workers have been evaluated with the same bio-feedback shirts to measure the physiological stresses they endure while working for long period. The researchers can then measure the amount of change they physiologically experience due to extended, high intensity activities and the effect of the weather [41].

The easiest type of monitoring for sport, medical, or health purposes are wearable biotech devices. Once they are put on, the user has no need to interact with them until they have finished their activity. Smartphones usually must be placed in certain positions on the body or be manually manipulated either during or between exertion periods. The current trend for wearable devices seems to be directed in two ways. Wearable clothing with data collection devices embedded in the fabric and wrist worn bracelets. There are various wearable shirts that measure vital signs. Athos Inc. has a wearable shirt and short combination. When worn together, they contain 22 Electromyography (EMG) (14 in the shirt, eight in the shorts) sensors, four HR sensors (two each, shirt and short) and two RR sensors in the shirt. There are small docking stations on the outer thigh of the shorts and the upper arm of the shirt where a removable data storage device is held. The software collects bio signals and sends them via Bluetooth to your phone. The claims by Athos is
that HR, RR, and the EMG of the actively working muscles can be recorded [118]. The Omsignal biotech shirt has ECG sensors built into the fabric and claims to measure HR, RR, breathing depth, target heart rate zones, activity intensity, SC, and EE. It also utilizes Bluetooth technology to transmit data to a smartphone [85]. Lastly, the Hexoskin wearable shirt measures HR, heart rate variability, heart rate recovery, ECG waves, RR, minute ventilation, peak acceleration, SC, SC cadence and EE. In addition to using Bluetooth technology, a detachable USB port is included for direct computer downloading. The Hexoskin is waterproof which allows it more freedom for data collection purposes [12]. There are currently no known studies that confirm any of the products validity or reliability. As can be seen, the variation of wearable biofeedback clothing is growing and becoming more and more popular for persons desiring quick and accurate vital sign stats

With a retail cost of $400.00 and more, wearable biofeedback clothes can be too expensive for most persons. However, most persons these days do own smart phones and the possibility of using them for activity monitoring and EE evaluations are a viable and rapidly growing field of research. This type monitoring is cheap, easy to use and available to all. Smart phones have been experimented with as monitoring devices in many medical applications [35]. One reason smart phones are easily used for monitoring is their built-in digital camera. Tan, Saatchi, Elphick, and Burke (2010) proved this with their study that video footage can determine RR through pixel analysis. Because smart phones are continually increasing their picture resolution and processing power, they make simple and ideal devices for monitoring, storing, and analyzing collected data.
Research has shown that a steady HR can be monitored with video taken on a smartphone. RR and heart rate variability can also be extracted using digital analysis from patient recording [64, 87]. The use of smartphones for this type of monitoring is very appealing because medical applications can be purchased cheaply, downloaded quickly, used easily by the patient, require no advanced training, and utilize no additional equipment [4]. Because of the widespread ownership of smartphones, these health related benefits would be available to any and all that owned one.

Selvaraj, Santhosh, and Anand (2007) helped established the accuracy of PPG to obtain a HR by measuring a finger-tip PPG signal with a wrist-watch style device. The same technological application can also be used by smartphones. An area of the skin is illuminated with the white LED flash and color changes are recorded by the video camera to generate a red-green-blue video [64]. The high resolution and power of their digital cameras allows these video images to be analyzed by looking at color changes in the skin due to cardiac activity [87]. These videos have been compared to those obtained from a pulse oximeter and have been shown to produce similar waveforms with cardiac pulse peaks [64]. These peaks are the used to determine a continuous HR signal.

In 2008, Granot, Ivorra, and Rubinsky realized that the power and small design of smartphones along with their user-friendly interaction could be beneficial for health monitoring. They argued that because conventional medical imaging systems were self-contained units that combined data collection devices, all the necessary software, and an imaging system to display results, that the cost for these devises would remain high and would be unnecessarily reproduced for every lab they were used in. Also, the sheer size and physical connection of the devices made them unwieldy and uncomfortable for the
patient. They reasoned that the three components of a medical monitor could be separated and more efficiently utilized. With the advances in smartphone technology, data collection could be done through a home collection monitor connected to a smartphone or it could involve high resolution photographs taken by the phone. Either way, the information could be sent via Wi-Fi to a central processing center. Once downloaded, the information could be opened with any appropriate software by anyone with access to the processing unit [46]. Unlimited readings and analysis could be made. No longer would a trained technician be needed for every device in every location. The only technicians that would be required would be those at the central processing unit. Also, areas of the world that once did not have access to medical monitoring equipment could now utilize smartphones to send in data and receive the same care and medical consultation as anyone else [46].

By 2014, smartphones and tablets were being viewed as valuable tools for exercise monitoring. Their high computational speed along with virtually unrestricted Wi-Fi access created a situation where either could serve as a physical activity monitor that could be used practically anywhere. Because the iPad touch and the iPhone contained the same pre-installed 3-D accelerometer, Nolan, Mitchell, and Doyle-Baker (2014) tested whether the iPad Touch was valid as a physical activity monitor. It was reasoned due to the same 3-D accelerometers installed that the results of the iPad testing would be applicable to the iPhone as well. Twenty-five participants performed two sessions of activity, treadmill walking (4.0 km · hr⁻¹ to 7.2 km · hr⁻¹) and running (8.1 km · hr⁻¹ to 11.3 km · hr⁻¹) while wearing the device. A HRM (FS2c, Polar Electro, NY) and a metabolic cart equipped with gas analyzers (TrueOne2400, Parvomedics, UT) were used
to compare the final results. The activity type for the iPad was classified with 99% accuracy. The monitored iPad speed had a bias of 0.02 \(8.1 \text{ km} \cdot \text{hr}^{-1}\) with an SEE of 0.57 \(8.1 \text{ km} \cdot \text{hr}^{-1}\) for the walking protocol. There was a -0.03 \(8.1 \text{ km} \cdot \text{hr}^{-1}\) bias with an SEE of 1.02 \(8.1 \text{ km} \cdot \text{hr}^{-1}\) for the running. EE was calculated to have a bias of 0.35 METs and an SEE of 0.75 METs for walking and a -0.43 METs bias with an SEE of 1.24 METs for running. The iPod Touch had an accuracy that was comparable to other accelerometer-based monitors.

As can be deduced, technology, especially in the wearable and smartphone fields, is expanding rapidly. These devices are opening new areas of research and giving users the ability to be better informed, more involved, and a better understanding of the physiological functions of their bodies and the effects various outside influences can have on them.
CHAPTER 3

METHODOLOGY

Subject characteristics

Forty-nine adults were recruited from the community through word of mouth and from various lecture classes at the University of Nevada, Las Vegas. Males were between 18-44 years of age and females were between 18-54 years of age. All participants were determined to be of sufficient health to perform the treadmill exercise protocol. This was assessed by the completion of an ACSM health risk questionnaire with a “low” risk category classification based on the participant’s answers. An informed consent form was signed prior to any testing. The protocol was approved by the University of Nevada, Las Vegas Institutional Review Board (protocol number 1408-4894).
**Instrumentation**

Participant’s body composition data were taken through Bioelectrical Impedance Analysis, using a Tanita TBF-521 Body Fat Monitor/Scale. (Arlington Heights, IL, USA).

A Nautilus T9.14 commercial treadmill (Vancouver, WA, USA) was used for the treadmill protocol.

Physiological measurements were taken by 3 monitoring devices. 1) HR, RR, pulmonary ventilation, and expired air samples were analyzed continually through the use of an Applied Electrochemistry Moxus Metabolic System (Bastrop, TX, USA) and accompanying Applied Electrochemistry Oxygen (S – 3A) and Carbon Dioxide (CD – 3A) analyzers (Naperville, IL, USA). HR was transmitted from a Polar T31 Coded Transmitter and Belt Set with the recorded data sent directly to an accompanying computer. 2) HR, RR, SC, and EE were recorded by a Hexoskin wearable body metrics shirt (Carré Technologies Inc. San Francisco, CA, USA) with embedded ECG sensors and a detachable data collection unit. 3) SC and EE were recorded with A Fitbit Flex (Fitbit Inc. San Francisco, CA, USA) and data uploaded via Wi-Fi internet connection.

SC was visually measured by use of a hand tally counter.
Collection of the Data

Prior to any data collection, the Hexoskin application was downloaded to an iPad for real-time monitoring purposes. The Fitbit Flex application was downloaded to a Samsung Galaxy 4 for the same purpose. Data collection was completed during two scheduled visits to the Exercise Physiology Laboratory in MPE 326 with a minimum of 3 days between tests. The initial visit began with the completion of the ACSM health risk questionnaire and informed consent form. Subjects then stood on a bioelectrical impedance analysis device with bare feet in order to measure body composition. The subject’s age, height, weight, and body composition (BC) were recorded. Body Mass Index (BMI) was determined by the formula [Weight (lbs.) * 703 / Height² (in)]. Once anthropometric and body composition data was obtained, the information was entered into the Hexoskin and Fitbit Flex applications and the MOXUS computer for accurate measurements of EE.

The participant was then prepared for the treadmill protocol. First, they were fitted with the chest HRM and a Hexoskin shirt. When used individually, the built-in Hexoskin shirt HRM sensors and the chest HRM belt would normally be positioned in about the same position along the side of the subject’s pectoral muscles. To avoid any interference between the two, the chest HRM belt was lowered by approximately 1 inch which had no effect on its heart rate monitoring ability. A visual inspection was conducted on every subject prior to walking to ensure that this 1 inch gap separated the two. The Hexoskin shirt is packaged with an elastic band that can be fastened around a subject to hold the shirt sensors tight to the skin and help prevent them from shifting on the skin. This band was placed on every subject to ensure maximum connectivity of the
shirt HR sensors. The male version of the shirt has belt loops placed at sternum level for this reason. The female version does not have sternum level belt loops. An elastic band was still placed on the female subjects for connectivity assistance. The band was placed on the sternum, just below the breasts and wrapped around the upper torso so the band was on top of the shirt sensors along the side of the chest. Once the subject was properly prepped, they were instructed to stand on the treadmill. The Hexoskin data collection device was then connected to the Hexoskin shirt. Both the Hexoskin and chest HRM were visually observed to confirm that there was a solid heart rate signal from both devices that was within 2-3 bpm of each other for at least 15 seconds. When there was difficulty making a connection, the Hexoskin sensors were slightly dampened with water to help facilitate the Bluetooth contact from the monitoring devices. Once this was complete, the subject was connected to the MOXUS system and the Fitbit Flex was placed on their right wrist.

All participants performed an approved treadmill walking protocol. The protocol consisted of three separate stages of walking at three different speeds. Each stage was separated by a rest interval that allowed for data collection and preparation for the next stage. Subjects began by walking at 1.5 mph at 0% grade for three minutes. After the rest interval, the speed was increased by 1.0 mph while then grade remained the same. This continued until the final stage of 3.5 mph was completed. Participants began each stage by stepping onto the treadmill while it was already moving at that stages speed. They were instructed not to hold onto the treadmills handrails and to walk in the natural manner for that speed.
The Hexoskin has a default of 70 bpm if it does not identify the subjects HR. If a HR of 70 was viewed continuously for at least half of the 1.5 mph or 2.5 mph stage, the entire preparation process to include visual inspection of the HRM and shirt sensor placement, adjustment of the elastic band, and reconnection of the Hexoskin to ensure a signal from both devices was repeated prior to walking the next stage. If the Hexoskin gave any consistent reading other than 70 bpm or appeared to have fluctuations above and/or below 70, that was considered a completed stage and no adjustments were made.

Upon completion of the treadmill protocol, the participant was allowed to self-select a walking or jogging speed for a cool-down.

The treadmill protocol during the second scheduled visit was performed in the same manner but without paperwork or anthropometric and body composition measurements.
**Statistical Analysis Methods**

The data was analyzed using IBM SPSS statistics 22.0 software (Armonk, New York). The dependent variables investigated were HR, RR, EE and SC.

Validity for each protocol was conducted at various time and speed combinations for HR, RR, SC, and EE using Pearson’s correlation coefficient (r). An (r) of < 0.70 was considered to be not correlated and the null hypothesis would be rejected under this circumstance.

Significant mean differences were also conducted for each protocol in the same manner as validity in regards to the various time and speed combinations for HR, RR, SC, and EE. Because the same persons were tested for each protocol and it was not known whether the results from either device would be greater or lesser than the actual measurement, 2-tailed, dependent paired t-tests were performed. A (p) value ≤ 0.05 was considered a significant difference in the mean values.

Pearson’s correlation and paired t-tests of the Hexoskin and MOXUS HR and RR values was conducted for the 1, 2, and 3 minute point for each of the three stages (1.5 mph, 2.5 mph, and 3.5 mph). However for RR, an additional statistics were performed for the 3 minute RR sum of the 1.5 mph, 2.5 mph, and 3.5 mph stages was also done. The Hexoskin HR value was compared to the HR recorded by the Polar T-31 HRM. The recorded Hexoskin RR value was compared to that recorded by the MOXUS respiratory system.

Pearson’s correlation and paired t-tests of the Hexoskin and MOXUS and the Fitbit Flex and MOXUS EE were performed separately using the 3 minute total EE of the three stages. Both devices EE were separately compared to the EE calculated from the
measured MOXUS absolute VO$_2$ and the appropriate respiratory exchange ratio (RER) at that stage.

Pearson’s correlation and paired t-tests of the Hexoskin and the observed value and the Fitbit Flex and the observed value for SC were performed separately using the 3 minute total SC of the three stages compared to the actual physical count for the corresponding stage walked.

Reliability was performed for HR, RR, SC, and EE. Cronbach's alpha ($\alpha$) was used as the coefficient of reliability or consistency. An $\alpha \geq 0.7$ was considered an acceptable score for the lower-bound estimate of reliability for scores measured in protocol #1 vs protocol #2. The average means was used for significance (AMS).

HR and RR reliability was determined by using the Intraclass Correlation (ICC) option in SPSS by comparing the measured Hexoskin values from protocol #1 and protocol #2 at the 1, 2, and 3 minute point for all three stages. EE reliability was determined by comparing the 3 minute cumulative value from protocol #1 and protocol #2 at all three stages. SC reliability was determined by comparing the 3 minute cumulative value from protocol #1 and protocol #2 at all three stages.
CHAPTER 4
RESULTS

Forty-nine subjects signed up to participate in the two day walking protocol. Three were unable to return for the second day of testing due to personal reasons. N=49 (26 males, 23 females) was used for the first validity test. Because of the inconsistencies with the Hexoskin and its measuring of physiological readings, a second validity was calculated using the protocol #2 data with N=46 (24 males, 22 females). Because it was not possible to replicate the environment and/or conduct the second test on a similar day/time of the first test, eighteen persons were not utilized for reliability purposes. N=31 (18 males, 13 females) was used to determine Hexoskin HR and RR reliability and Hexoskin and Fitbit Flex SC and EE reliability. The anthropometrical baseline data from the both validity protocols and the reliability analysis are included below (Table 2a; protocol #1 Anthropometrics, Table 2b; protocol #2 Anthropometrics, Table 2c; Reliability Anthropometrics)

<table>
<thead>
<tr>
<th>Table 2a; protocol #1 Anthropometrics</th>
<th>Table 2b; protocol #2 Anthropometric</th>
</tr>
</thead>
<tbody>
<tr>
<td>Protocol #1</td>
<td>Protocol #2</td>
</tr>
<tr>
<td>N=49; 26 males, 23 females</td>
<td>N=46; 24 males, 22 females</td>
</tr>
<tr>
<td>Age (yrs)</td>
<td>Age (yrs)</td>
</tr>
<tr>
<td>23.43 ± 6.57</td>
<td>23.39 ± 6.69</td>
</tr>
<tr>
<td>Body mass (kg)</td>
<td>Body mass (kg)</td>
</tr>
<tr>
<td>76.15 ± 18.46</td>
<td>76.52 ± 18.73</td>
</tr>
<tr>
<td>Height (m)</td>
<td>Height (m)</td>
</tr>
<tr>
<td>1.72 ± 0.11</td>
<td>1.72 ± 0.11</td>
</tr>
<tr>
<td>Body fat (%)</td>
<td>Body fat (%)</td>
</tr>
<tr>
<td>27.15 ± 6.92</td>
<td>27.62 ± 6.83</td>
</tr>
<tr>
<td>Body mass index (BMI)</td>
<td>Body mass index (BMI)</td>
</tr>
<tr>
<td>25.36 ± 3.90</td>
<td>25.46 ± 3.93</td>
</tr>
<tr>
<td>Values are Mean ± SD</td>
<td>Values are Mean ± SD</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Table 2c; Reliability Anthropometrics</th>
</tr>
</thead>
<tbody>
<tr>
<td>Reliability</td>
</tr>
<tr>
<td>N=31; 18 males, 13 females</td>
</tr>
<tr>
<td>Age (yrs)</td>
</tr>
<tr>
<td>24.39 ± 7.59</td>
</tr>
<tr>
<td>Body mass (kg)</td>
</tr>
<tr>
<td>77.95 ± 21.52</td>
</tr>
<tr>
<td>Height (m)</td>
</tr>
<tr>
<td>1.73 ± 0.10</td>
</tr>
<tr>
<td>Body fat (%)</td>
</tr>
<tr>
<td>28.24 ± 7.00</td>
</tr>
<tr>
<td>Body mass index (BMI)</td>
</tr>
<tr>
<td>25.56 ± 4.53</td>
</tr>
<tr>
<td>Values are Mean ± SD</td>
</tr>
</tbody>
</table>
Hexoskin HR Validity

HR validity was inconsistent between the two protocols. While all minutes were significantly correlated at the 1.5 mph speed in protocol #1, \(r = 0.77, 0.72, \text{and } 0.70; \text{ all } p < 0.00\), none of the HR’s for the corresponding minutes and speed were highly related in protocol #2 \(r = 0.64, 0.52, 0.61; \text{ all } p < 0.00\). Only one minute displayed a significant correlation in each protocol during the 2.5 mph and 3.5 mph stages; protocol #1 @ 3.5 mph, minute -3 \(r = 0.72; p < 0.00\) and protocol 2 @ 3.5 mph, minute -3 \(r = 0.72; p < 0.00\). The highest observed correlation for HR obtained by the Hexoskin compared to the Polar T-31 was no higher than \(r = 0.77\).

HR during walking protocol #1 showed significant mean differences during the 1.5 mph speed at minute-2 and -3 \(p = 0.01, p = 0.03\) and at 2.5 mph minute-2 \(p = 0.01\) see table 3 below). 1.5 mph minute-1, 2.5 mph minute-1 and -3, and all three minutes of the 3.5 mph stage had no significant mean differences between the Hexoskin and the Polar T-31 readings.

Walking protocol #2 showed significant mean HR differences only during the 1.5 mph walk at minute-2 and -3 \(p = 0.05; p = 0.05\). See Table 3; Hexoskin Heart Rate Validity. (Full table; Appendix B)
Table 3: Hexoskin Heart Rate Validity

<table>
<thead>
<tr>
<th>Protocol #1 (n=49)</th>
<th>Correlation; p value</th>
<th>HR Ob (bpm)</th>
<th>HR Hx (bpm)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>minute -1</td>
<td>r = 0.77; 0.00</td>
<td>97.39±13.48</td>
<td>95.78±17.29</td>
<td>p = 0.31</td>
</tr>
<tr>
<td>minute -2</td>
<td>r = 0.72; 0.00</td>
<td><strong>97.1±14.67</strong></td>
<td><strong>92.14±18.88</strong></td>
<td>p = 0.01</td>
</tr>
<tr>
<td>minute -3</td>
<td>r = 0.70; 0.00</td>
<td><strong>96.61±14.57</strong></td>
<td><strong>92.27±19.09</strong></td>
<td>p = 0.03</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Protocol #2 (n=46)</th>
<th>Correlation; p value</th>
<th>HR Ob (bpm)</th>
<th>HR Hx (bpm)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>minute -1</td>
<td>r = 0.58; 0.00</td>
<td>101.76±12.24</td>
<td>98.55±22.72</td>
<td>p = 0.23</td>
</tr>
<tr>
<td>minute -2</td>
<td>r = 0.55; 0.00</td>
<td><strong>102.31±12.44</strong></td>
<td><strong>95.59±20.17</strong></td>
<td>p = 0.01</td>
</tr>
<tr>
<td>minute -3</td>
<td>r = 0.66; 0.00</td>
<td>102.27±13.09</td>
<td>98.9±19.37</td>
<td>p = 0.11</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Protocol #1 (n=49)</th>
<th>Correlation; p value</th>
<th>HR Ob (bpm)</th>
<th>HR Hx (bpm)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>minute -1</td>
<td>r = 0.77; 0.00</td>
<td>97.39±13.48</td>
<td>95.78±17.29</td>
<td>p = 0.31</td>
</tr>
<tr>
<td>minute -2</td>
<td>r = 0.72; 0.00</td>
<td><strong>97.1±14.67</strong></td>
<td><strong>92.14±18.88</strong></td>
<td>p = 0.01</td>
</tr>
<tr>
<td>minute -3</td>
<td>r = 0.70; 0.00</td>
<td><strong>96.61±14.57</strong></td>
<td><strong>92.27±19.09</strong></td>
<td>p = 0.03</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Protocol #2 (n=46)</th>
<th>Correlation; p value</th>
<th>HR Ob (bpm)</th>
<th>HR Hx (bpm)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>minute -1</td>
<td>r = 0.58; 0.00</td>
<td>101.76±12.24</td>
<td>98.55±22.72</td>
<td>p = 0.23</td>
</tr>
<tr>
<td>minute -2</td>
<td>r = 0.55; 0.00</td>
<td><strong>102.31±12.44</strong></td>
<td><strong>95.59±20.17</strong></td>
<td>p = 0.01</td>
</tr>
<tr>
<td>minute -3</td>
<td>r = 0.66; 0.00</td>
<td>102.27±13.09</td>
<td>98.9±19.37</td>
<td>p = 0.11</td>
</tr>
</tbody>
</table>

Hexoskin HR Reliability

Hexoskin HR interclass correlation (ICC) was considered acceptable only during minute-1 at 1.5 mph (α = 0.75), minute-2 and -3 at 2.5 mph (α = 0.76, 0.82), and at minute-3 at 3.5 mph (α = 0.76). See Table 4; Hexoskin Heart Rate Reliability.

(Scatterplot; Appendix F)
Hexoskin RR Validity

RR for both protocol #1 and #2 was significantly correlated for all minutes and the total RR sums for both the 1.5 mph and 2.5 mph stages. However, during the 3.5 mph stage of protocol #1, none of the minutes or the cumulative sum were related (r = 0.46, 0.41, 0.46, 0.50 respectively; all p < 0.00) while only minute -1 of the 3.5 mph walk in protocol #2 was not significantly correlated (r = 0.63; p < 0.00).

RR showed significant mean differences for all minutes over both protocols except for the 3.5 mph, minute -1 stage of protocol #1 (p = 0.19; see Table 5; Hexoskin Respiratory Rate Validity). This included the comparison of the sum total of respiratory breathes over the entire 3 minute period for each speed. (Full table; Appendix C)

### Table 4: Hexoskin Heart Rate Reliability

<table>
<thead>
<tr>
<th>Heart Rate</th>
<th>Interclass Correlation</th>
<th>n=31</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>α</td>
<td>p</td>
</tr>
<tr>
<td>1.5 mph</td>
<td>0.75</td>
<td>0.00</td>
</tr>
<tr>
<td>2.5 mph</td>
<td>0.68</td>
<td>0.00</td>
</tr>
<tr>
<td>3.5 mph</td>
<td>0.53</td>
<td>0.02</td>
</tr>
</tbody>
</table>

### Table 5: Hexoskin Respiratory Rate Validity

<table>
<thead>
<tr>
<th>Protocol #1 (n=49)</th>
<th>Correlation; p value</th>
<th>RR Ob (brpm)</th>
<th>RR Hx (brpm)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td>Total</td>
<td>r = 0.93; 0.00</td>
<td>62.67 ± 16.2</td>
<td>57.69 ± 15.74</td>
</tr>
<tr>
<td></td>
<td>minute -1</td>
<td>r = 0.85; 0.00</td>
<td>20.47 ± 5.55</td>
<td>19.08 ± 5.84</td>
</tr>
<tr>
<td></td>
<td>minute -2</td>
<td>r = 0.88; 0.00</td>
<td>20.94 ± 5.74</td>
<td>19.04 ± 5.41</td>
</tr>
<tr>
<td></td>
<td>minute -3</td>
<td>r = 0.92; 0.00</td>
<td>21.27 ± 5.81</td>
<td>19.57 ± 5.95</td>
</tr>
<tr>
<td>2.5 mph</td>
<td>Total</td>
<td>r = 0.89; 0.00</td>
<td>71.08 ± 16.15</td>
<td>63.53 ± 17.56</td>
</tr>
<tr>
<td></td>
<td>minute -1</td>
<td>r = 0.85; 0.00</td>
<td>23.8 ± 6.07</td>
<td>20.84 ± 6.43</td>
</tr>
<tr>
<td></td>
<td>minute -2</td>
<td>r = 0.85; 0.00</td>
<td>23.65 ± 5.46</td>
<td>21.51 ± 6.06</td>
</tr>
<tr>
<td></td>
<td>minute -3</td>
<td>r = 0.82; 0.00</td>
<td>23.63 ± 5.7</td>
<td>21.16 ± 6.28</td>
</tr>
<tr>
<td>3.5 mph</td>
<td>Total</td>
<td>r = 0.46; 0.00</td>
<td>87.78 ± 20.42</td>
<td>77.86 ± 37.06</td>
</tr>
<tr>
<td></td>
<td>minute -1</td>
<td>r = 0.41; 0.00</td>
<td>28.75 ± 7.41</td>
<td>26.43 ± 13.32</td>
</tr>
<tr>
<td></td>
<td>minute -2</td>
<td>r = 0.46; 0.00</td>
<td>29.23 ± 6.9</td>
<td>25.78 ± 12.34</td>
</tr>
<tr>
<td></td>
<td>minute -3</td>
<td>r = 0.50; 0.00</td>
<td>29.57 ± 7.24</td>
<td>25.86 ± 12.69</td>
</tr>
</tbody>
</table>
Table 5 Hexoskin Respiratory Rate Validity cont.

<table>
<thead>
<tr>
<th>Protocol #2 (n=46)</th>
<th>Validity</th>
<th>RR Ob Mn</th>
<th>RR Hx Mn</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total</td>
<td>r = 0.90; 0.00</td>
<td>67.41 ± 17.03</td>
<td>61.04 ± 18.53</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>1.5 mph</td>
<td>r = 0.76; 0.00</td>
<td>22.3 ± 5.76</td>
<td>20.22 ± 6.3</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>minute -1</td>
<td>r = 0.89; 0.00</td>
<td>22 ± 5.83</td>
<td>20.48 ± 6.65</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>minute -2</td>
<td>r = 0.85; 0.00</td>
<td>23.11 ± 6</td>
<td>20.15 ± 6.6</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>minute -3</td>
<td>r = 0.76; 0.00</td>
<td>22.3 ± 5.76</td>
<td>20.22 ± 6.3</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>2.5 mph</td>
<td>r = 0.92; 0.00</td>
<td>76.13 ± 19.44</td>
<td>67.76 ± 20.76</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>minute -1</td>
<td>r = 0.82; 0.00</td>
<td>25.37 ± 6.54</td>
<td>22.93 ± 7.98</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>minute -2</td>
<td>r = 0.85; 0.00</td>
<td>25.41 ± 6.7</td>
<td>22.3 ± 7.74</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>minute -3</td>
<td>r = 0.91; 0.00</td>
<td>25.35 ± 6.98</td>
<td>22.52 ± 7.4</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>3.5 mph</td>
<td>r = 0.63; 0.00</td>
<td>29.87 ± 8.11</td>
<td>27.09 ± 12.19</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>minute -1</td>
<td>r = 0.74; 0.00</td>
<td>20.63 ± 7.8</td>
<td>24.96 ± 10.05</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>minute -2</td>
<td>r = 0.70; 0.00</td>
<td>31.09 ± 8.52</td>
<td>27.44 ± 11.86</td>
<td>p = 0.00</td>
</tr>
</tbody>
</table>

Hexoskin RR Reliability

Hexoskin RR ICC was significantly correlated for all minutes at all walking speeds. The lowest Cronbach alpha being minute-2 at 2.5 mph (α = 0.79). See Table 6;

Hexoskin Respiratory Rate Reliability. (Scatterplot; Appendix G)

Table 6: Hexoskin Respiratory Rate Reliability

<table>
<thead>
<tr>
<th>Respiratory Rate</th>
<th>Interclass Correlation</th>
<th>n=31</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Total</td>
<td>1 minute</td>
</tr>
<tr>
<td></td>
<td>α</td>
<td>p</td>
</tr>
<tr>
<td>1.5 mph</td>
<td>0.90</td>
<td>0.00</td>
</tr>
<tr>
<td>2.5 mph</td>
<td>0.89</td>
<td>0.00</td>
</tr>
<tr>
<td>3.5 mph</td>
<td>0.93</td>
<td>0.00</td>
</tr>
</tbody>
</table>

Hexoskin EE Validity

None of the stages of walking for either protocol were highly related. The highest correlation being r = 0.59; p = 0.00 for protocol #2 @ 1.5 mph
Hexoskin EE had no significant mean differences for any stage of walking for either protocol. Most Hexoskin EE values were slightly overestimated. See Table 7; Hexoskin Energy Expenditure Validity. (Full table; Appendix D)

Table 7; Hexoskin Energy Expenditure Validity

<table>
<thead>
<tr>
<th>Protocol #1 Hx (n=49)</th>
<th>Correlation; p value</th>
<th>EE Ob (cal)</th>
<th>EE Hx (cal)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td></td>
<td>11.93 ± 3.1</td>
<td>13.39 ± 8.21</td>
<td>p = 0.15</td>
</tr>
<tr>
<td>2.5 mph</td>
<td></td>
<td>14.46 ± 3.71</td>
<td>14.96 ± 8.28</td>
<td>p = 0.62</td>
</tr>
<tr>
<td>3.5 mph</td>
<td></td>
<td>19.47 ± 4.92</td>
<td>20.4 ± 10.29</td>
<td>p = 0.46</td>
</tr>
</tbody>
</table>

Table 8; Hexoskin Energy Expenditure Reliability

<table>
<thead>
<tr>
<th>Protocol #2 Hx (n=46)</th>
<th>Correlation; p value</th>
<th>EE Ob (cal)</th>
<th>EE Hx (cal)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td></td>
<td>11.86 ± 3.11</td>
<td>12.74 ± 7.89</td>
<td>p = 0.37</td>
</tr>
<tr>
<td>2.5 mph</td>
<td></td>
<td>14.39 ± 3.67</td>
<td>14.52 ± 8.07</td>
<td>p = 0.90</td>
</tr>
<tr>
<td>3.5 mph</td>
<td></td>
<td>19.38 ± 4.62</td>
<td>19.67 ± 10.84</td>
<td>p = 0.83</td>
</tr>
</tbody>
</table>

Hexoskin EE Reliability

Hexoskin EE ICC was acceptable at all three speeds (1.5 mph, α = 0.85; 2.5 mph α = 0.83; 3.5 mph, α = 0.80). See Table 8; Hexoskin Energy Expenditure Reliability.

(Scatterplot; Appendix H)
Fitbit Flex EE Validity

None of the Fitbit Flex stages for either protocol were related. The closest being the 3.5 mph stage in protocol #1 with an \( r = 0.68; \ p = 0.00 \).

The Fitbit Flex had significant mean differences for every stage walked (\( p = 0.00 \) for all stages on both protocols). See Table 9; Fitbit Flex Energy Expenditure Validity.

(Full table; Appendix D)

Table 9: Fitbit Flex Energy Expenditure Validity

<table>
<thead>
<tr>
<th>Protocol #1 Fb ( n=49 )</th>
<th>Correlation; p value</th>
<th>EE Ob (cal)</th>
<th>EE Fb (cal)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph ( r = 0.54; 0.00 )</td>
<td>11.93 ± 3.1</td>
<td>19.41 ± 6.95</td>
<td>( p = 0.00 )</td>
<td></td>
</tr>
<tr>
<td>2.5 mph ( r = 0.58; 0.00 )</td>
<td>14.46 ± 3.71</td>
<td>25.31 ± 7.22</td>
<td>( p = 0.00 )</td>
<td></td>
</tr>
<tr>
<td>3.5 mph ( r = 0.68; 0.00 )</td>
<td>19.47 ± 4.92</td>
<td>28.71 ± 7.85</td>
<td>( p = 0.00 )</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Protocol #2 Fb ( n=46 )</th>
<th>Correlation; p value</th>
<th>EE Ob (cal)</th>
<th>EE Fb (cal)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph ( r = 0.49; 0.00 )</td>
<td>11.86 ± 3.11</td>
<td>19.46 ± 7.36</td>
<td>( p = 0.00 )</td>
<td></td>
</tr>
<tr>
<td>2.5 mph ( r = 0.48; 0.00 )</td>
<td>14.39 ± 3.67</td>
<td>24.67 ± 7.46</td>
<td>( p = 0.00 )</td>
<td></td>
</tr>
<tr>
<td>3.5 mph ( r = 0.54; 0.00 )</td>
<td>19.37 ± 4.62</td>
<td>25.59 ± 6.94</td>
<td>( p = 0.00 )</td>
<td></td>
</tr>
</tbody>
</table>

Fitbit Flex EE Reliability

The Fitbit Flex had an acceptable ICC only at 2.5 mph (\( \alpha = 0.72 \)). The 1.5 mph and 3.5 mph were not (\( \alpha = 0.56, \alpha = 0.67 \)). See Table 10; Fitbit Flex Energy Expenditure Reliability. (Scatterplot; Appendix I)

Table 10: Fitbit Flex Energy Expenditure Reliability

<table>
<thead>
<tr>
<th>Energy Expenditure</th>
<th>Interclass Correlation</th>
<th>n=31</th>
<th>Fitbit</th>
<th>( \alpha )</th>
<th>( p )</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td></td>
<td>0.56</td>
<td>0.02</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2.5 mph</td>
<td></td>
<td>0.72</td>
<td>0.00</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3.5 mph</td>
<td></td>
<td>0.67</td>
<td>0.00</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
Hexoskin SC Validity

Hexoskin SC was significantly correlated only at the 3.5 mph stage of both protocols (#1, r = 0.74; #2, r = 0.94). Of particular note is the negative (r) value at the 1.5 mph stage for both protocols (#1, r = -0.09; #2, r = -0.01).

SC correlation for the Hexoskin showed no significant mean differences during minute-3 for both protocols (p = 0.10 and 0.31). SC during minutes -1 and -2 for both protocols had significant lower differences (p = 0.00 for all). See Table 11; Hexoskin Step Count Validity. (Full table; Appendix E)

<table>
<thead>
<tr>
<th>Protocol #1 Hx (n=49)</th>
<th>Correlation; p value</th>
<th>SC Ob</th>
<th>SC Hx</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td>r = -0.09; 0.56</td>
<td>274.65 ± 20.67</td>
<td>97.78 ± 67.6</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>2.5 mph</td>
<td>r = 0.01; 0.50</td>
<td>333.22 ± 17.88</td>
<td>212.37 ± 84.71</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>3.5 mph</td>
<td>r = 0.74; 0.00</td>
<td>381.62 ± 19.85</td>
<td>377.76 ± 23.52</td>
<td>p = 0.10</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Protocol #2 Hx (n=46)</th>
<th>Correlation; p value</th>
<th>SC Ob Mn</th>
<th>SC Hx Mn</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td>r = -0.01; 0.98</td>
<td>262.87 ± 28.19</td>
<td>83.11 ± 64.77</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>2.5 mph</td>
<td>r = 0.14; 0.34</td>
<td>329.87 ± 24.35</td>
<td>207.35 ± 98.87</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>3.5 mph</td>
<td>r = 0.94; 0.00</td>
<td>377.91 ± 23.56</td>
<td>379.22 ± 24.65</td>
<td>p = 0.31</td>
</tr>
</tbody>
</table>

Hexoskin SC Reliability

SC for the Hexoskin was reliable for 1.5 mph (α = 0.70) and 2.5 mph (α =0.86). 3.5 mph was not (α = 0.61). See Table 12; Hexoskin Step Count Reliability. (Scatterplot; Appendix J)
Table 12: Hexoskin Step Count Reliability

<table>
<thead>
<tr>
<th>Step Count</th>
<th>Interclass Correlation</th>
<th>Hexoskin</th>
<th>n=31</th>
<th>α</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td></td>
<td></td>
<td></td>
<td>0.70</td>
<td>0.00</td>
</tr>
<tr>
<td>2.5 mph</td>
<td></td>
<td></td>
<td></td>
<td>0.86</td>
<td>0.00</td>
</tr>
<tr>
<td>3.5 mph</td>
<td></td>
<td></td>
<td></td>
<td>0.61</td>
<td>0.01</td>
</tr>
</tbody>
</table>

Fitbit Flex SC Validity

No stages for either protocol were significantly correlated for the Fitbit.

The Fitbit Flex SC had significant mean differences for all stages except for minute-2 of protocol #2 (p = 0.58). See Table 13; Fitbit Flex Step Count Validity. (Full table; Appendix E)

Table 13: Fitbit Flex Step Count Validity

<table>
<thead>
<tr>
<th>Protocol #1 Fb (n=49)</th>
<th>Correlation; p value</th>
<th>SC Ob Mn</th>
<th>SC Fb Mn</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td>r = 0.43; 0.00</td>
<td>274.65 ± 20.67</td>
<td>230.94 ± 65.26</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>2.5 mph</td>
<td>r = 0.41; 0.00</td>
<td>333.22 ± 17.88</td>
<td>319.1 ± 40.92</td>
<td>p = 0.01</td>
</tr>
<tr>
<td>3.5 mph</td>
<td>r = 0.65; 0.00</td>
<td>381.63 ± 19.85</td>
<td>372.55 ± 27.9</td>
<td>p = 0.00</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Protocol #2 Fb (n=46)</th>
<th>Correlation; p value</th>
<th>SC Ob Mn</th>
<th>SC Fb Mn</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mph</td>
<td>r = 0.43; 0.00</td>
<td>262.87 ± 28.2</td>
<td>231.17 ± 51.64</td>
<td>p = 0.00</td>
</tr>
<tr>
<td>2.5 mph</td>
<td>r = 0.37; 0.01</td>
<td>329.87 ± 24.35</td>
<td>326.41 ± 44.74</td>
<td>p = 0.58</td>
</tr>
<tr>
<td>3.5 mph</td>
<td>r = 0.43; 0.00</td>
<td>377.91 ± 23.36</td>
<td>359.07 ± 33.57</td>
<td>p = 0.00</td>
</tr>
</tbody>
</table>

Fitbit Flex SC Reliability

All three stages for the Fitbit Flex had (α) scores below 0.70 (1.5 mph, α = 0.46; 2.5 mph, α = 0.50; and 3.5 mph, α = 0.66). See Table 14; Fitbit Flex Step Count Reliability. (Scatterplot; Appendix K)
Table 14: Fitbit Flex Step Count Reliability

<table>
<thead>
<tr>
<th>Speed (mph)</th>
<th>α</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5</td>
<td>0.46</td>
<td>0.02</td>
</tr>
<tr>
<td>2.5</td>
<td>0.50</td>
<td>0.03</td>
</tr>
<tr>
<td>3.5</td>
<td>0.66</td>
<td>0.00</td>
</tr>
</tbody>
</table>

Step Count
Interclass Correlation
n=31
Fitbit
CHAPTER 5
SUMMARY, CONCLUSIONS, AND RECOMMENDATIONS

Discussion of Results

The Hexoskin shirt and Fitbit Flex activity monitor are two products that are currently being marketed to both the public and professional community for the monitoring of physiological measurements in real time. A first step to ensuring these devices are suitable for use by anyone is to test their validity and reliability under normal circumstances. The four most common metabolic data measurements monitored by persons are HR, RR, EE, and SC. It was hypothesized that the HR data collected by the Hexoskin would not have any significant difference to the T-31 chest HRM. It was hypothesized, however, that RR, EE, and SC measurements by the Hexoskin and EE and SC measurements by the Fitbit Flex would all be significantly lower.

Hexoskin

Correlation values (r) for HR were inconsistent at best when comparing the results from both protocols. For example, all 3 minutes were significantly correlated to the Polar T-31 during the protocol #1 walk at 1.5 mph while none were during protocol #2 at the same speed. Both the 2.5 and 3.5 speeds had one minute that was highly related in one protocol but not acceptably related in the other. In short, five of the total nine minutes that were looked at side-by-side were not highly related from one protocol to another. On a smaller scale, the same was true for the comparison of the average mean differences. All three minutes of the 3.5 mph stage from both protocols had no significant differences. The 1.5 mph stage had two minutes (minutes -1 and -2) and 2.5 mph had one (minute -2)
that alternated in significant difference between protocols. Overall, the Hexoskin tended to underestimate HR for every stage. HR reliability between the two protocols was split between the nine stages compared. Two stages at 3.5 mph (minutes -2 and -3) and 1.5 mph (minutes -2 and -3) and one at 2.5 mph (minute -1) had an (α) below 0.7.

There was a noticeable issue with the Hexoskin and its ability to collect HR data at certain points. This issue was not unknown to the researchers. During use of the Hexoskin in the collection of pilot data for other research, it was found that there were instances where the Hexoskin did not record HR data properly (fluctuating values) or failed to record at all (70 bpm default). This was a point of interest that needed to be addressed for two reasons. First, an accurately displayed HR during exercise allows a participant to stay within desired training intensities for maximum results. Second, the Hexoskin estimates EE by using HR values. Thus, incorrect HR measurements will create incorrect EE values. The greatest contributing factor that affected HR correlation and ICC values for either protocol appears to be the fit of the Hexoskin shirt on the individual. The shirt is designed to be skin tight on the participant so the HR sensors will be flat on the skin along the side of the chest. Using the elastic band seems to improve the connectivity by holding the sensors close during movement but this is not always effective. For this study, sensor placement may have been affected by the subject’s natural walking motion or their physical anatomy (large chest or slim frame for example). Both instances appeared to effect the sensors detection abilities by either creating a gap between the sensor and skin or shifting the sensor to a position where it could no longer detect the HR properly.
Using the basis of forty-nine persons in protocol #1 and forty-six in protocol #2, there were 285 individual stages of walking. Notes were made during both protocols to examine how often and what HR sensor issues occurred. There were 30 stages of walking where the HR measurement was not consistent due to one of the following circumstances. 10 had no HR data collected after walking began even though connectivity was checked prior to starting. 20 stages registered abnormally high or low values for the entire 3 minute stage, or had no connection for the first 1-2 minutes and then an extremely high or low value for the remainder of the stage when compared to the observed HR. Additionally, 38 separate stages of walking had various issues such as the HR not registering immediately (over 1 minute) and then giving a normal value or the HR went up and down repeatedly for the entire stage. The remaining 227 stages solidly connected and gave a consistent reading ~7.82 seconds after beginning the walk. Either the subject’s physical attributes, their natural walking motion, the shifting of the sensors or a combination of all three can be considered to have contributed to the Hexoskin’s inconsistent readings.

Detection of RR by the Hexoskin was not a concern. The Bluetooth connection to the iPad was instantaneous and continuous. The data obtained for the RR correlation and ICC included one extra measurement, however. Just like with HR where RR values were obtained at each minute for all three stages, the cumulative sum of all three minutes worth of RR values for each stage was also evaluated and compared to the MOXUS. All the minutes and total sum RR for both the 1.5 mph stage and the 2.5 mph stage were acceptably correlated. In protocol #1, none of the minutes or the total sum at 3.5 mph were correlated while only minute -1 of the 3.5 mph stage in of protocol #1 had no
significant difference. In stark contrast, every ICC between the two protocols was acceptable with the lowest (α) being a definitive 0.79.

This high number of RR stages that were not correlated to include the cumulative RR sums can best be explained by the same reason that the HR correlation was not consistent, the fit of the shirt on the subject. All mean values produced between the observed value (MOXUS) and the Hexoskin recorded value with one exception were significantly lower. There are two RR sensors that run parallel across the anterior side of the Hexoskin. One is chest level, directly in line with the HR sensors, and the other lies a little above the naval. Because the enclosed elastic band was being used on the chest for every participant, it can be speculated that a second band placed in this lower position may have permitted the naval sensor to more accurately detect RR. There are belt loops on the Hexoskin at this level for both the male and female versions of the shirt for this very reason. The addition of a second band, especially with leaner persons, may have a positive effect on RR correlation values. Because the RR ICC was considered acceptable at all stages, an inferred conclusion can be made that whatever is done to improve the Hexoskin’s correlation, ICC will not be affected.

The Hexoskin did not have any stages of walking between the two protocols that showed significant correlation for EE. The highest (r) value being 0.59 for the 1.5 mph walk in protocol #2. While the Hexoskin did slightly over-estimate EE for every stage, there were no significant mean differences in caloric estimation. This was contrary to the hypothesis that there would be significantly lower values. The smallest mean difference was at 2.5 mph during protocol #2 (0.13 Calories) and the largest was at 1.5 mph during protocol #1 (1.46 Calories). While the differences were not much more than the observed
values, it is preferred that with any EE data collection device, the Hexoskin included, that
the measurements be slightly below the real value if the EE is not being recorded exactly.
Taking these higher values of 0.13 and 1.46 and applying them across a 16 hour day that
someone may be awake, the Hexoskin would actually overestimate EE by 41.6 to 467.2
Calories. This actual Calorie count will depend on the day’s activity but there will be a
higher estimated value regardless. This can be an issue for those trying to lose weight.
The perception will be that they are burning more Calories than they actually are. This in
turn can affect how these persons diet and plan their daily exercise habits.

Hexoskin SC was highly correlated only at the 3.5 mph stage for both protocols.
The mean difference at this speed also showed no significant differences. However, for
both the 1.5 mph and 2.5 mph stages, there was extremely low correlations (0.01, 0.14)
and significantly lower mean differences than the actual physical count. This is in direct
accordance with previous studies that show pedometers and accelerometers have
significantly smaller SC’s at speeds of approximately 2.5 mph and slower [70, 78]. One
reason for the significant differences in SC at the slower speeds for the Hexoskin may be
in how the data collection pack is worn during activities. The Hexoskin has a small
pocket that is positioned along the right mid-axillary line, just above the iliac crest where
the data pack is placed during exercise. During faster walking speeds, there is enough
motion for the 3d accelerometer within the data pack to register as steps. At slower
speeds, a person may have to modify their walking motion to accommodate the unusually
slow motion required to walk at these decreased walking speeds. This adjustment to their
walking mannerism may not induce enough movement for the 3d accelerometer to
register as a step. Hexoskin reliability for SC showed a low (α) for the 3.5 mph stage only (α = 0.613).

**Fitbit Flex**

No stages of walking were acceptably correlated for the Fitbit Flex. The highest (r) being at 3.5 mph, minute -3 in protocol #1 (r = 0.68). EE for the Fitbit Flex was significantly higher for all stages of walking. This differed from the alternate hypothesis that the Hexoskin EE would be significantly lower. The smallest mean difference was at 3.5 mph during protocol #2 (6.21) and the largest was at 2.5 mph during protocol #1 (10.85). Taking these higher values and performing a similar calculation like that which was done for the Hexoskin EE for a 16 hour day, the Fitbit Flex would overestimate EE between 1987.26 to 3289.6 Calories. This extremely high over-estimation would be very detrimental to all using the devise to help monitor their daily fitness levels and Caloric expenditure. The Fitbit was only reliable at the 2.5 mph speed and not by much (α = 0.72).

SC for the Fitbit Flex was only valid at 2.5 mph in protocol #2. All other stage were significantly lower as hypothesized. One reason for the significant differences in Fitbit Flex SC, just like the Hexoskin, is how a person walks at slow speeds (1.5 mph) as compared to faster speeds (3.5 mph). The Fitbit Flex relies on the arm motion while walking to count steps. It uses this count along with previously entered personal data (Ht. Wt., and BC) to calculate EE. When persons were walking at slower speeds on the treadmill, it was observed that there was a noticeable reduction in arm motion which would directly affect the SC value. There were no reliable stages. The highest (α) being 0.66 @ 3.5 mph.
Practical Application

Previous research suggests that wearable physiological devices can help athletes improve performance by monitoring metabolic values in a real time manner [13, 19, 61, 71]. The use of these devices, especially those similar to the Hexoskin shirt which record multiple vital signs simultaneously, have the ability to help improve certain health care related issues that are present in today’s medical community. Medical related financial costs can be lowered by promoting home monitoring of less serious medical conditions. By using a measuring device and a Wi-Fi connection, patients can be monitored remotely, freeing up valuable hospital space. Also, by using Wi-Fi connections, medical records can be centrally located and accessed by any medical personal at any time [98, 109].

But beyond these specific applications, these devices can be used by all persons on an everyday basis. The Hexoskin is designed to be worn under any clothing, from casual t-shirts to business suits. It is unobtrusive and can be used to monitor a whole day’s activity when downloaded at the end of the day. The Fitbit Flex is small and worn on the wrist with no hindrances to daily activities. Persons that are aware of their current activity level, duration of performance, corresponding HR, RR, and EE values can be psychologically influenced to become more active. They will begin to use the data collected to compare current achievements to prior ones. This then creates healthy improvements to diet and daily behavior habits [28, 101]. Currently, ~67% of persons in the US are classified as overweight or obese. These devices can help reduce those numbers and improve the lives of many [65, 71]
Conclusions and Recommendations

For Future Study

The primary aim of this study was to establish an initial pool of data for the evaluation of the Hexoskin shirt’s ability to accurately and consistently measure vital signs. As of the writing of these results, there is no published literature that directly addresses the Hexoskin in any manner. The logical first goal, the one that was addressed in this research, was to establish whether the basic concepts of validity and reliability were supported by the Hexoskin’s measurements. Our research focused on measuring four physiological functions, HR, RR, EE, and SC during a simple activity such as walking. The three speeds used, 1.5 mph, 2.5 mph, and 3.5 mph were chosen because they would elicit notable differences in these four categories. Based on this, there are numerous research topics that can be expanded upon by further study.

First, research can be extended into validity and reliability of metabolic values obtained when activities other than walking are performed. Walking in itself is an extremely mild activity, even at higher speeds. With the HR connection issues that were observed with this simple motion, it would be prudent to examine whether activities with greater motion and higher intensities such as cycling, jogging, sprinting, or resistance training would affect measurements and by what degree. For this product to be marketed to professional athletes, research institutes, and those requiring accurate data, it will need to be tested in the same manner and conditions for which it is to be used.

Second, the Hexoskin needs to be evaluated under conditions that mimic what the common person will do after donning the shirt and up until they exercise. Because this technology is in essence a shirt, it will be viewed as simple to use by the common person.
It will be treated as a wear-and-go type monitoring device that only needs to be put on to be used. Most people will put it on and then stretch and/or prepare for their activity before actually doing it. The Hexoskin should be evaluated for this situation and ensure it is able to connect and measure all the vital signs appropriately after this preparation period. During the initial research, the only interaction the subjects had with the Hexoskin was to put it on and take it off. All sensor placements, elastic band attachments, data manipulation, and connectivity issues were handled by the researchers. This will not be the case in real life. There needs to be an evaluation as to whether this device can be properly used and whether data collection, especially HR will be able to be suitably recorded by just the user.

Third, in regards to the Hexoskin and SC measurement, placement of the data pack can be evaluated to see if a different location would be better for overall SC values. The current location on the right side of the body may not be in the optimal location to detect certain movements such as slow walking or for that matter cycling. This can be seen in the results stated previously. The data pack should be moved to other areas such as the front or back of the shirt pocket to see if this location works better for the 3d accelerometer inside. Also, lowering the data pack to the side of the hip as far as possible may be an option.

The Hexoskin and Fitbit Flex are both excellent ideas for metabolic measuring and the possibilities for their use are numerous. However, like all products in their initial generations of production, there are issues with both devices that need to be addressed for validity and reliability purposes. The Hexoskin’s monitoring of HR is an important function that needs to be rigorously tested. Instantaneous HR readings are not only used
for real-time determination of a person’s training intensity but are also used for the Hexoskin’s EE estimation. As seen in the results, Hexoskin HR measures were valid for the 3.5 mph speed for both protocols and only had one significant difference between the six total minutes over the 2.5 mph stages. There were four out-of-six significantly different minutes for 1.5 mph. This is an issue that will need to be analyzed and corrected if this device is being marketed as a home health care aid for persons that are sedentary, injured, elderly, or sick. These persons will have lower HR values and the Hexoskin must be able to accurately measure HR at these lower intensities for any home monitoring to be acceptable.

It is the opinion of the researchers that the Hexoskin is a tool that may be valuable for research purpose. However, the HR detection issue is a major factor that will need to be thoroughly investigated and corrected.

The Fitbit Flex does not accurately measure what it claims to. The best opinion that can be given for it is that it should be used as a psychological tool to motivate a person to become more active. Even if the values measured are wrong, there will still be a level of self-induced motivation and need to better oneself through comparing current results with previous.
Appendix A; Government/agency/professional organization step-based recommendations from around the world

<table>
<thead>
<tr>
<th>Government/agency/professional organization</th>
<th>Step-based recommendation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Queensland Health (Australia)</td>
<td>Sponsors 10,000 Steps: “aims to increase the day-to-day activity of Australians by encouraging you to use a step-counting pedometer to accumulate ‘incidental’ physical activity as part of your everyday living” (<a href="http://www.10000steps.org.au/">http://www.10000steps.org.au/</a>)</td>
</tr>
<tr>
<td>U.S. President’s Challenge Physical Activity and Fitness Awards Program</td>
<td>Recommends 8,500 steps/day for adults, and 13,000 and 11,000 steps/day for boys and girls respectively (<a href="http://www.presidentschallenge.org/challenge/active/index.shtml">http://www.presidentschallenge.org/challenge/active/index.shtml</a>)</td>
</tr>
<tr>
<td>America on the Move</td>
<td>Promotes walking an extra 2,000 steps in addition to eating 100 less calories each day to stop weight gain. (<a href="http://aom3.americaonthemove.org/">http://aom3.americaonthemove.org/</a>)</td>
</tr>
<tr>
<td>National Obesity Forum (U.K)</td>
<td>Indicates that 3,000 to 6,000 steps/day is sedentary, 7,000 to 10,000 steps is moderately active, and &gt; 11,000 steps/day is very active. (<a href="http://www.nationalobesityforum.org.uk/healthcare-professionals/155/treatmentmainmenu-169/192-useful-tools-and-agencies.html">http://www.nationalobesityforum.org.uk/healthcare-professionals/155/treatmentmainmenu-169/192-useful-tools-and-agencies.html</a>)</td>
</tr>
<tr>
<td>Northern Ireland’s Public Health Agency</td>
<td>Promotes an additional 30 minutes of daily walking or 3000 steps (<a href="http://www.getalifegetactive.com/adults/walking/walking">http://www.getalifegetactive.com/adults/walking/walking</a>)</td>
</tr>
<tr>
<td>Ministry of Health, Labour and Welfare of Japan</td>
<td>Recommends: “for individuals who intend to promote health mainly through physical activity, a daily walk of 8,000 to 10,000 steps is set as the target. The report indicates that 8,000 to 10,000 steps/day is approximately equivalent to 60 minutes of walking per day at an intensity of 3 METs, and that it is also approximately equivalent to 23 MET-hours/week of MVPA which is the recommended physical activity level in this guideline.”</td>
</tr>
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Appendix B; Hexoskin Heart Rate

### Protocol #1 (n=49)

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<th>HR (Ob)-(Hx) @ 1.5 mph</th>
<th>t</th>
<th>Significance</th>
<th>HR Ob Mn</th>
<th>HR Ob SD</th>
<th>HR Hx Mn</th>
<th>HR Hx SD</th>
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<th>95% LC</th>
<th>95% UC</th>
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<tr>
<td></td>
<td>t(48) = 1.03</td>
<td>p &gt; 0.05; 0.31</td>
<td>97.39</td>
<td>13.48</td>
<td>95.78</td>
<td>17.29</td>
<td>1.61</td>
<td>-1.55</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>t(48) = 2.64</td>
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<td>14.67</td>
<td>92.14</td>
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<tr>
<td></td>
<td>t(48) = 2.23</td>
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<td>92.27</td>
<td>19.09</td>
<td>4.35</td>
<td>1.95</td>
<td>8.27</td>
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</table>

- Shaded areas indicate significant difference between observed measurements and the monitor indicated.
- A positive mean represents a lower measured average value for the device compared to observed measures.
- A negative mean (-) represents a higher measured average value for the device compared to observed measures.

### Protocol #2 (n=46)

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<th>HR (Ob)-(Hx) @ 1.5 mph</th>
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<th>HR Ob SD</th>
<th>HR Hx Mn</th>
<th>HR Hx SD</th>
<th>MEAN</th>
<th>95% LC</th>
<th>95% UC</th>
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<tr>
<td></td>
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<td>p &lt; 0.05; 0.05</td>
<td>96.02</td>
<td>11.95</td>
<td>92.44</td>
<td>15.45</td>
<td>3.59</td>
<td>0</td>
<td>7.17</td>
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<tr>
<td></td>
<td>t(45) = 0.72</td>
<td>p &gt; 0.05; 0.47</td>
<td>95.98</td>
<td>12.85</td>
<td>94.11</td>
<td>20.38</td>
<td>1.87</td>
<td>-3.34</td>
<td>7.08</td>
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<td></td>
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<tr>
<td></td>
<td>t(45) = 2.03</td>
<td>p &lt; 0.05; 0.048</td>
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<td>91.24</td>
<td>18.41</td>
<td>4.43</td>
<td>0.03</td>
<td>8.84</td>
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</tbody>
</table>

- Shaded areas indicate significant difference between observed measurements and the monitor indicated.
- A positive mean represents a lower measured average value for the device compared to observed measures.
- A negative mean (-) represents a higher measured average value for the device compared to observed measures.

HR = Heart Rate
Ob = Observed measurement
Hx = Hexoskin
Mn = Mean
SD = Standard Deviation
LC = Lower confidence bound
UC = Upper confidence bound
Appendix C; Hexoskin Respiratory Rate

- Shaded areas indicate significant difference between observed measurements and the monitor indicated
- A positive mean represents a lower measured average value for the device compared to observed measures
- A negative mean (−) represents a higher measured average value for the device compared to observed measures

**RR = Respiratory Rate**
**Ob = Observed measurement**
**Hx = Hexoskin**
**Mn = Mean**
**SD = Standard Deviation**
**LC = Lower confidence bound**
**UC = Upper confidence bound**

<table>
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<tr>
<th>Protocol #1 (n=49)</th>
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<th>Significance</th>
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<th>RR Ob SD</th>
<th>RR Hx Mn</th>
<th>RR Hx SD</th>
<th>MEAN</th>
<th>95% LC</th>
<th>95% UC</th>
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<td>RR (Ob)-(Hx) @ 1.5 mph</td>
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<td></td>
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<td></td>
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<tr>
<td>Total</td>
<td>t(48) = 5.76</td>
<td>p &lt; 0.05; 0.00</td>
<td>62.67</td>
<td>16.2</td>
<td>57.69</td>
<td>15.74</td>
<td>4.98</td>
<td>3.24</td>
<td>6.72</td>
</tr>
<tr>
<td>minute -1</td>
<td>t(48) = 3.13</td>
<td>p &lt; 0.05; 0.00</td>
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<td>5.55</td>
<td>19.08</td>
<td>5.84</td>
<td>1.39</td>
<td>0.5</td>
<td>2.28</td>
</tr>
<tr>
<td>minute -2</td>
<td>t(48) = 4.84</td>
<td>p &lt; 0.05; 0.00</td>
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<td>5.74</td>
<td>19.04</td>
<td>5.41</td>
<td>1.9</td>
<td>1.11</td>
<td>2.69</td>
</tr>
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<td>p &lt; 0.05; 0.00</td>
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<td>1.01</td>
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<td>RR (Ob)-(Hx) @ 2.5 mph</td>
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<td></td>
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<td></td>
<td></td>
<td></td>
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<tr>
<td>Total</td>
<td>t(48) = 6.64</td>
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<td>71.08</td>
<td>16.15</td>
<td>63.53</td>
<td>17.56</td>
<td>7.55</td>
<td>5.27</td>
<td>9.84</td>
</tr>
<tr>
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<td>6.43</td>
<td>2.96</td>
<td>1.97</td>
<td>3.95</td>
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<td>5.46</td>
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<td>6.06</td>
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<td>t(48) = 4.78</td>
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<td>21.16</td>
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<td>2.47</td>
<td>1.43</td>
<td>3.51</td>
</tr>
<tr>
<td>RR (Ob)-(Hx) @ 3.5 mph</td>
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<tr>
<td>Total</td>
<td>t(48) = 2.1</td>
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<td>-1.18</td>
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<tr>
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<th>RR Ob SD</th>
<th>RR Hx Mn</th>
<th>RR Hx SD</th>
<th>MEAN</th>
<th>95% LC</th>
<th>95% UC</th>
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<tbody>
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<tr>
<td>Total</td>
<td>t(45) = 5.34</td>
<td>p &lt; 0.05; 0.00</td>
<td>67.41</td>
<td>17.03</td>
<td>61.04</td>
<td>18.53</td>
<td>6.37</td>
<td>3.97</td>
<td>8.77</td>
</tr>
<tr>
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<td>t(45) = 3.36</td>
<td>p &lt; 0.05; 0.00</td>
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<td>5.76</td>
<td>20.22</td>
<td>6.3</td>
<td>2.09</td>
<td>0.84</td>
<td>3.34</td>
</tr>
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<td>19.47</td>
<td>6.65</td>
<td>1.52</td>
<td>0.62</td>
<td>2.43</td>
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<tr>
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<td>t(45) = 5.69</td>
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<td>21.16</td>
<td>6.28</td>
<td>2.47</td>
<td>1.43</td>
<td>3.51</td>
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<td>RR (Ob)-(Hx) @ 2.5 mph</td>
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<td>23.43</td>
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<td>32.87</td>
<td>11.96</td>
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<td>18.71</td>
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<td>8.11</td>
<td>27.09</td>
<td>12.19</td>
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<td>5.67</td>
<td>3.66</td>
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<td>11.86</td>
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## Appendix D: Hexoskin and Fitbit Energy Expenditure

### Hexoskin EE

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<th>EE Hx Mn</th>
<th>EE Hx SD</th>
<th>MEAN</th>
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</tr>
<tr>
<td>EE (Ob)-(Hx) @ 1.5 mph</td>
<td>t(48) = -1.45</td>
<td>p &gt; 0.05; 0.15</td>
<td>11.93</td>
<td>3.1</td>
<td>13.39</td>
<td>8.21</td>
<td>-1.46</td>
<td>-3.48</td>
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<tr>
<td>EE (Ob)-(Hx) @ 2.5 mph</td>
<td>t(48) = -0.5</td>
<td>p &gt; 0.05; 0.62</td>
<td>14.46</td>
<td>3.71</td>
<td>14.96</td>
<td>8.28</td>
<td>-0.5</td>
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<tr>
<td>EE (Ob)-(Hx) @ 3.5 mph</td>
<td>t(48) = -0.74</td>
<td>p &gt; 0.05; 0.46</td>
<td>19.47</td>
<td>4.92</td>
<td>20.4</td>
<td>10.29</td>
<td>-0.94</td>
<td>-3.48</td>
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### Protocol #2 Hx | n=46 | t | Significance | EE Ob Mn | EE Ob SD | EE Hx Mn | EE Hx SD | MEAN | 95% LC | 95% UC |
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<td></td>
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<tr>
<td>EE (Ob)-(Fb) @ 1.5 mph</td>
<td>t(45) = -8.9</td>
<td>p &lt; 0.05; 0.00</td>
<td>11.93</td>
<td>3.1</td>
<td>19.41</td>
<td>5.59</td>
<td>-7.48</td>
<td>-9.17</td>
<td>-5.79</td>
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<tr>
<td>EE (Ob)-(Fb) @ 2.5 mph</td>
<td>t(45) = -12.85</td>
<td>p &lt; 0.05; 0.00</td>
<td>14.46</td>
<td>3.71</td>
<td>25.98</td>
<td>7.22</td>
<td>-10.85</td>
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<td>EE (Ob)-(Fb) @ 3.5 mph</td>
<td>t(45) = -11.23</td>
<td>p &lt; 0.05; 0.00</td>
<td>19.47</td>
<td>4.92</td>
<td>28.71</td>
<td>7.85</td>
<td>-9.24</td>
<td>-10.9</td>
<td>-7.59</td>
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### FitBit Flex EE

<table>
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<tr>
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<th>EE Ob Mn</th>
<th>EE Ob SD</th>
<th>EE Fb Mn</th>
<th>EE Fb SD</th>
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<th>95% UC</th>
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<tr>
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<td></td>
<td></td>
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<tr>
<td>EE (Ob)-(Fb) @ 1.5 mph</td>
<td>t(45) = -8.01</td>
<td>p &lt; 0.05; 0.00</td>
<td>11.86</td>
<td>3.11</td>
<td>19.46</td>
<td>6.29</td>
<td>-7.59</td>
<td>-9.5</td>
<td>-5.68</td>
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<tr>
<td>EE (Ob)-(Fb) @ 2.5 mph</td>
<td>t(45) = -10.64</td>
<td>p &lt; 0.05; 0.00</td>
<td>14.39</td>
<td>3.67</td>
<td>24.67</td>
<td>7.46</td>
<td>-10.28</td>
<td>-12.23</td>
<td>-8.34</td>
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<tr>
<td>EE (Ob)-(Fb) @ 3.5 mph</td>
<td>t(45) = -7.15</td>
<td>p &lt; 0.05; 0.00</td>
<td>19.37</td>
<td>4.62</td>
<td>25.59</td>
<td>6.94</td>
<td>-6.21</td>
<td>-7.96</td>
<td>-4.46</td>
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</table>

- Shaded areas indicate significant difference between observed measurements and the monitor indicated
- A positive mean represents a lower measured average value for the device compared to observed measures
- A negative mean (-) represents a higher measured average value for the device compared to observed measures

EE = Energy Expenditure
Ob = Observed measurement
Hx = Hexoskin
Fb = Fitbit Flex
Mn = Mean
SD = Standard Deviation
LC = Lower confidence bound
UC = Upper confidence bound

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Appendix E: Hexoskin and Fitbit Step Count

<table>
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<tr>
<th>Protocol #1 Hx (n=49)</th>
<th>t</th>
<th>Significance</th>
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<th>SC Ob SD</th>
<th>SC Hx Mn</th>
<th>SC Hx SD</th>
<th>MEAN</th>
<th>95% LC</th>
<th>95% UC</th>
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</thead>
<tbody>
<tr>
<td>SC (Ob)-(Hx) @ 1.5 mph</td>
<td>t(48) = 17.11</td>
<td>p &lt; 0.05; 0.00</td>
<td>274.65</td>
<td>20.67</td>
<td>97.78</td>
<td>67.6</td>
<td>176.88</td>
<td>156.09</td>
<td>197.66</td>
</tr>
<tr>
<td>SC (Ob)-(Hx) @ 2.5 mph</td>
<td>t(48) = 9.98</td>
<td>p &lt; 0.05; 0.00</td>
<td>333.22</td>
<td>17.88</td>
<td>212.37</td>
<td>84.71</td>
<td>120.86</td>
<td>96.5</td>
<td>145.22</td>
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<tr>
<td>SC (Ob)-(Hx) @ 3.5 mph</td>
<td>t(48) = 1.68</td>
<td>p &gt; 0.05; 0.10</td>
<td>381.62</td>
<td>19.85</td>
<td>377.76</td>
<td>23.52</td>
<td>3.88</td>
<td>-0.76</td>
<td>8.51</td>
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<table>
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<th>Protocol #2 Hx (n=46)</th>
<th>t</th>
<th>Significance</th>
<th>SC Ob Mn</th>
<th>SC Ob SD</th>
<th>SC Hx Mn</th>
<th>SC Hx SD</th>
<th>MEAN</th>
<th>95% LC</th>
<th>95% UC</th>
</tr>
</thead>
<tbody>
<tr>
<td>SC (Ob)-(Hx) @ 1.5 mph</td>
<td>t(45) = 17.23</td>
<td>p &lt; 0.05; 0.00</td>
<td>262.87</td>
<td>28.19</td>
<td>83.11</td>
<td>14.7</td>
<td>179.76</td>
<td>158.75</td>
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<tr>
<td>SC (Ob)-(Hx) @ 2.5 mph</td>
<td>t(45) = 8.8</td>
<td>p &lt; 0.05; 0.00</td>
<td>328.87</td>
<td>24.35</td>
<td>201.67</td>
<td>94.87</td>
<td>122.52</td>
<td>94.46</td>
<td>150.58</td>
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<tr>
<td>SC (Ob)-(Hx) @ 3.5 mph</td>
<td>t(45) = -1.02</td>
<td>p &gt; 0.05; 0.31</td>
<td>377.91</td>
<td>23.36</td>
<td>379.22</td>
<td>24.65</td>
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<thead>
<tr>
<th>Protocol #1 Fb (n=49)</th>
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<th>Significance</th>
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<th>SC Ob SD</th>
<th>SC Fb Mn</th>
<th>SC Fb SD</th>
<th>MEAN</th>
<th>95% LC</th>
<th>95% UC</th>
</tr>
</thead>
<tbody>
<tr>
<td>SC (Ob)-(Fb) @ 1.5 mph</td>
<td>t(48) = 5.147</td>
<td>p &lt; 0.05; 0.00</td>
<td>274.65</td>
<td>20.67</td>
<td>230.87</td>
<td>64.32</td>
<td>43.71</td>
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<tr>
<td>SC (Ob)-(Fb) @ 2.5 mph</td>
<td>t(48) = 2.65</td>
<td>p &lt; 0.05; 0.01</td>
<td>333.22</td>
<td>17.88</td>
<td>319.1</td>
<td>40.92</td>
<td>14.12</td>
<td>3.39</td>
<td>24.86</td>
</tr>
<tr>
<td>SC (Ob)-(Fb) @ 3.5 mph</td>
<td>t(48) = 3.00</td>
<td>p &lt; 0.05; 0.00</td>
<td>381.63</td>
<td>19.85</td>
<td>372.55</td>
<td>27.9</td>
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<table>
<thead>
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<th>Protocol #2 Fb (n=46)</th>
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<th>Significance</th>
<th>SC Ob Mn</th>
<th>SC Ob SD</th>
<th>SC Fb Mn</th>
<th>SC Fb SD</th>
<th>MEAN</th>
<th>95% LC</th>
<th>95% UC</th>
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</thead>
<tbody>
<tr>
<td>SC (Ob)-(Fb) @ 1.5 mph</td>
<td>t(45) = 4.56</td>
<td>p &lt; 0.05; 0.00</td>
<td>262.87</td>
<td>28.2</td>
<td>231.17</td>
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<tr>
<td>SC (Ob)-(Fb) @ 2.5 mph</td>
<td>t(45) = 0.553</td>
<td>p &gt; 0.05; 0.58</td>
<td>329.87</td>
<td>24.35</td>
<td>326.41</td>
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<td>SC (Ob)-(Fb) @ 3.5 mph</td>
<td>t(45) = 4.05</td>
<td>p &lt; 0.05; 0.00</td>
<td>377.91</td>
<td>23.36</td>
<td>359.07</td>
<td>33.57</td>
<td>18.85</td>
<td>9.46</td>
<td>28.23</td>
</tr>
</tbody>
</table>

- Shaded areas indicate significant difference between observed measurements and the monitor indicated.
- A positive mean represents a lower measured average value for the device compared to observed measures.
- A negative mean (-) represents a higher measured average value for the device compared to observed measures.

SC = Step Count
Ob = Observed measurement
Hx = Hexoskin
Fb = Fitbit Flex
Mn = Mean
SD = Standard Deviation
LC = Lower confidence bound
UC = Upper confidence bound
Appendix F: Hexoskin HR ICC Scatterplot

HR Reliability; 1.5 mph, minute 1

HR Reliability; 1.5 mph, minute 2

HR Reliability; 1.5 mph, minute 3
Appendix F: Hexoskin HR ICC Scatterplot cont.

HR Reliability; 2.5 mph, minute 1

α = 0.68

HR Reliability; 2.5 mph, minute 2

α = 0.76

HR Reliability; 2.5 mph, minute 3

α = 0.82
Appendix F: Hexoskin HR ICC Scatterplot cont.
Appendix G: Hexoskin RR ICC Scatterplot.

RR Reliability; 1.5 mph, Total

RR Reliability; 2.5 mph, Total

RR Reliability; 3.5 mph, Total
Appendix G: Hexoskin RR ICC Scatterplot cont.
Appendix G: Hexoskin RR ICC Scatterplot cont.

RR Reliability; 2.5 mph, minute 1

\[ \alpha = 0.80 \]

RR Reliability; 2.5 mph, minute 2

\[ \alpha = 0.86 \]

RR Reliability; 2.5 mph, minute 3

\[ \alpha = 0.87 \]
Appendix G: Hexoskin RR ICC Scatterplot cont.
Appendix H: Hexoskin EE ICC Scatterplot

EE Reliability; Hx, 1.5 mph

$\alpha = 0.85$

EE Reliability; Hx, 2.5 mph

$\alpha = 0.83$

EE Reliability; Hx, 3.5 mph

$\alpha = 0.80$
Appendix I: Fitbit Flex EE ICC Scatterplot

EE Reliability; Fb, 1.5 mph
\( \alpha = 0.56 \)

EE Reliability; Fb, 2.5 mph
\( \alpha = 0.72 \)

EE Reliability; Fb, 3.5 mph
\( \alpha = 0.67 \)
Appendix J: Hexoskin SC ICC Scatterplot

SC Reliability; Hx, 1.5 mph
\[ \alpha = 0.070 \]

SC Reliability; Hx, 2.5 mph
\[ \alpha = 0.86 \]

SC Reliability; Hx, 3.5 mph
\[ \alpha = 0.61 \]
Appendix K: Fitbit Flex SC ICC Scatterplot

SC Reliability; Fb, 1.5 mph
\[ \alpha = 0.45 \]

SC Reliability; Fb, 2.5 mph
\[ \alpha = 0.50 \]

SC Reliability; Fb, 3.5 mph
\[ \alpha = 0.66 \]
Bibliography


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Publishing Group


Curriculum Vitae

Jeffrey Montes

GENERAL INFORMATION

Education
2013-2015 M.S., University of Nevada, Las Vegas, Las Vegas, NV
       Kinesiology: Exercise Physiology
2010-2012 B.S., University of Nevada, Las Vegas, Las Vegas, NV
       Kinesiology
1998-2000 A.S., College of Southern Nevada, North Las Vegas, NV
       General Studies: Science Emphasis

Certifications
       American College of Sports Medicine (ACSM): Certified Personal Trainer, Certified Fitness Specialist
       National Strength and Conditioning Association (NSCA): Certified Personal Trainer, Certified Strength and Conditioning Specialist
       United States Weightlifting Association: Level 1 Coach
       American Heart Association: CPR and AED certified

Memberships
       American College of Sports Medicine
       National Strength and Conditioning Association
       United States Weightlifting Association

RESEARCH EXPERIENCE

Graduate Assistant

       Cosmed BODPOD
       Hydrostatic Weighing Tank
       Orca Metabolic Cart
       Moxus Metabolic Cart
       Blood draw and analysis
       Hexoskin shirt

Professional Presentations and Refereed Published Abstracts

Montes, J., Young, J., Navalta, J. Validation and Reliability of the Hexoskin and Fitbit Flex wearable bio-collection devices, NV, 2015


Mercer, J., Prado, A., Montes, J. Triathlon Research; Running while wearing a Wet Suit. University of Nevada, Las Vegas, NV, 2014


Harvel, A.C., Miller, B.L., Trilleras, G., Montes, J., Girouard, T.J., Navalta, J.W. Association of Total and Regional Lean Body Mass Tissue Percentage and Upper and
Lower Limb Isokinetic Strength. Annual Meeting of the Southwest American College of Sports Medicine, Newport Beach, CA, 2013.


Classes Taught

UNLV
KIN 175 Physical Activity and Health; Fall 2014, Spring 2015
KIN 491 Exercise Physiology Laboratory; Fall 2014, Spring 2015
KIN 245 Anatomical Kinesiology; Spring 2015

Text Books Reviewed


Journal Articles Reviewed


December 2013, International Journal of Exercise Science: "The Effect of Music as a Motivational Tool on Isokinetic Concentric Performance in College Aged Students"