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Design of a Wearable Balance Control Indicator

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Table of Contents

AUTHORSHIP PAGE

ABSTRACT-**LT**

EXECUTIVE SUMMARY

1 INTRODUCTION-**LT**

2 LITERATURE REVIEW

- 2.1 UNDERSTANDING BALANCE CONTROL-**LT**
  - 2.1.1 Basics of Balance-**LT**
  - 2.1.2 Human Balance Control and the Inverted Pendulum Model-**LT**
  - 2.1.3 Maintaining Balance during Perturbation-**LT & AM**
  - 2.1.4 Example of Balance Control: Balancing a Pencil-**LT**
- 2.2 RISK FACTORS OF FALLING-**AM**
- 2.3 REDUCING THE NEGATIVE EFFECTS OF FALLS-**AM**
  - 2.3.1 Improving Balance Control-**AM**
  - 2.3.2 Fall Detection Patents-**AM**
  - 2.3.3 Current Fall Detection Technology-**AM**
  - 2.3.4 Limitations of Current Technology-**AM**
- 2.4 METHODS FOR ASSESSING BALANCE CONTROL-**AM**
  - 2.4.1 Vibrotactile Based Patents-**AM**
  - 2.4.2 Current Devices for Assessing Balance Control-**AM & LT**
- 2.5 PROACTIVELY MONITORING BALANCE CONTROL-**AM**
  - 2.5.1 Accelerometry-**KG, AM, LT**
  - 2.5.2 Gyroscopes-**KG & LT**
  - 2.5.3 Strain Gauges and Pressure Transducers-**KG &LT**
  - 2.5.4 Motion Analysis-**AM**
  - 2.5.5 Advantages and Disadvantages of Current Balance Monitoring Methods-**LT**

3 PROJECT STRATEGY-**LT**

- 3.1 CLARIFYING THE ORIGINAL PROBLEM STATEMENT-**LT**
- 3.2 OBJECTIVES & CONSTRAINTS-**LT**

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ABSTRACT

Each year, one in three elderly fall. Studies show that many factors contribute to an elderly person's risk of falling, but if the factors causing imbalance are improved, a person's risk of falling may be reduced. A device that detects and alerts the user of an off-balance situation before the fall occurs could identify a specific need for improved balance control. This paper describes the design, testing, and verification of a prototype wearable device that is worn on the right hip during the sit-to-stand activity (STS) to detect and notify the user of an unbalanced STS. By signaling an off-balance situation during STS, our device notifies the user of poor balance control and identifies the need for balance control improvement.

TABLE OF FIGURES

TABLE OF TABLES

EXECUTIVE SUMMARY

Every year, one in three adults age 65 years and (Tideiksaar, 2002) costing the United States 20 to 30 billion health care dollars annually (Centers for Disease Control & Prevention, 2009). Physicians often help high fall-risk patients improve balance, however many elderly may not have their balance control assessed before a fall occurs. As a result, there is a need for a device that detects and directly notifies the user of their balance control condition before a fall occurs. Current technology is able to assess balance control, however the devices are not catered to the elderly daily life or directly notify the user of their balance control. Therefore, the team designed a wearable device catered to the elderly population that is used only during the sit-to-stand (STS) rehabilitation activity. The device requires the user to perform a rehabilitation activity and detects and directly notifies the user of an unbalanced STS thereby strengthening the user's awareness of their balance condition and forcing rehabilitation to improve balance control.

Through research and client interviews the team determined design objectives, constraints, and necessary functions. The most important objectives were that the device was easy to use, sensitive to balance control, accurate, and adapted to elderly life. According to these objectives, the team determined an activity that the device should monitor, the signal and sensor used as a monitor, the location of the device, and the method of analyzing the signal. The team chose to monitor the STS activity using an accelerometer placed on the right hip, and analyzed the magnitude of the acceleration.

The team performed preliminary experiments to determine the difference between a balanced and unbalanced STS in terms of acceleration. Subjects attached the SparkFun KinetaMap tri-axial accelerometer to their right hip and performed 10 STS's with feet shoulder-width (SW) apart (balanced) and 10 with feet in tandem (unbalanced) with feet placed on the

AMTI force platform. Data from the force platform was used to verify a balanced and unbalanced STS and data from the KinetaMap was used to plot the magnitude of acceleration for each trial. Time between the positive and negative peak amplitudes of the acceleration plots was calculated and the team found that time between peaks of balanced trials was significantly longer ($p< 0.05$) than the time between peaks of the unbalanced trials, which was consistent with research (Pai & Patton, 1997). Preliminary data of subject 1 was used to calculate the time range during which a balanced STS occurred and time range during which an unbalanced STS occurred. The latter end of the unbalanced time range was chosen as the expected time (T_e) separating a balanced from unbalanced STS. This minimized false errors but included the maximum amount of unbalanced situations

The device, *Duino Balance* was built with a tri-axis accelerometer, Arduino Duemilanove Microcontroller Board, and a CEM1203 buzzer, a rechargeable battery pack and slide switch for powering the device, as well as a protoboard for connecting and attaching all the components. *Duino Balance* is enclosed in a plastic project box and attached to a belt to be worn around the user's waist. The device was programmed to detect the minimum and maximum peaks of the STS and measure the time (T_m) between these peaks. If $T_m \leq T_e$, the device was programmed to buzz. If $T_m > T_e$,the device was told to reset. Device verification was conducted using the same tests used during preliminary testing. During SW tests, the device buzzed once when it should not have buzzed (90% accuracy) and during tandem trials, the device reset twice when it should have buzzed (80% accuracy).

Therefore, the design was verified by having greater than 75% accuracy. The team also validated the design by interviewing clients who reported the device was "straightforward, easy to use, and not cumbersome" and could be used in a clinical setting (See Appendix C).

1 INTRODUCTION

Every year, one in three adults age 65 years and older fall and ten thousand elderly die each year as a result of falls (Tideiksaar, 2002). Of those who fall, 20-30% suffer injuries that impair their ability to live healthy, independent lives (Centers for Disease Control & Prevention, 2009). These injuries include moderate injuries such as bruises and arm fractures, and severe injuries such as hip fractures and head trauma. In fact, most fractures among the elderly and traumatic brain injuries are caused by falls (Centers for Disease Control & Prevention, 2009). In addition, falls in the elderly may cause feelings of increasing frailty, fear and stress, ultimately leading to anxiety during activities of daily living (ADLs) (e.g. getting out of bed). Of those who fall, 50% avoid performing ADLs because they fear additional falls (Tideiksaar, 2002).

Furthermore, falls cost the United States 20 to 30 billion health care dollars each year (Services, 2007) and this amount is expected to increase with the increasing elderly population. By 2030, 80 million people will be elderly, an approximate 43% increase since the year 2000, and by 2020, total indirect and direct medical costs of falls may reach 54.9 billion dollars (Centers for Disease Control & Prevention, 2009).The financial burden and effects of falls on the quality of life of the elderly cause the need for a cost-effective solution that minimizes the negative effects of falls, particularly on the elderly population.

The Centers for Disease Control and Prevention (CDC) have administered various fall prevention and education programs aimed at reducing the occurrence of falls in the elderly. These programs address a few of the many risk factors associated with elderly falls. For example, two risk factors that cause falls are decreased balance control and strength in the elderly. As a result, two studies funded by the CDC and conducted over a three-year time span utilized education and exercise programs intended to improve balance and increase strength in the elderly participants, thereby reducing falls. However, neither study produced a significant

reduction in falls (Centers for Disease Control & Prevention, 2009). On the other hand, previous research showed that Tai Chi exercise can improve balance and decrease falls among the elderly, but researchers do not know if the general elderly community can adopt this exercise into daily life (Rose, 2005). According to this research, regular exercise that the elderly can readily integrate into their daily life is one way of reducing the risk falling (Stevens, 2005). The CDC has also identified a need to increase elderly self efficacy and sense of balance control in relation to fall risk in order to prevent elderly falls (Centers for Disease Control & Prevention, 2009). Therefore, the MQP team set out to design a device that would enable elderly to independently assess their balance control and requires daily exercise.

In order to design a device that assesses balance control we had to understand how humans maintain balance and why the elderly experience an increased amount of falls. In addition, we identified current methods used to assess and monitor balance control, and evaluated the advantages and disadvantages of each method.

The final balance control indicator aims to strengthen the user's awareness of their balance condition, while requiring a rehabilitation activity. These features make the device distinct among existing products and the team hopes this will enable the device to reduce elderly falls in the future. This report describes the strategic design of the current balance control indicator. The report will discuss the background of human balance control, advantages and disadvantages of current technology, and the gap in the current market. Following, the report will detail the team's project approach, strategy of design, testing and analysis, and final design and verification.

2 LITERATURE REVIEW

A human's balance control system enables a person to maintain balance while standing, during locomotion (i.e. walking, running), and upon perturbation (e.g. tripping). Three sensory systems in the body are used to maintain balance: vision, the vestibular system, and the somatosensory system. As the functionality of these systems deteriorates, a person's ability to remain balanced decreases. However, studies showed that consistent exercise, especially Tai Chi, can improve balance control and decrease the risk of falling. This chapter discusses the details of balance control, how it is assessed, why elderly are susceptible to falls, and how balance control can be improved. The final sections of this chapter discuss advantages and disadvantages of current technology aimed at reducing the negative effects of falls, and subsequently identify important objectives that will guide the design of the present balance control indicator.

2.1 UNDERSTANDING BALANCE CONTROL

To analyze how each sensory system contributes to balance control, researchers assume the human body behaves like an inverted pendulum (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998); the body is always swaying. Certain factors regarding the body's movement can dictate how well a person is balanced. This section describes how the body maintains balance during locomotion and while standing, how each sensory system contributes to maintaining balance. In addition, this chapter discusses the parameters associated with balance control and ways that these parameters are measured in order to assess a person's balancing ability.

2.1.1 Basics of Balance

Balance is how the body moves relative to the gravitational force vector in order to maintain posture and prevent falling. To understand how balance is maintained, researchers study the relation between the body's center of gravity (COG), the center of mass (COM), the center of

pressure (COP), and the base of support (BOS). The COM is a point on the body equal to the sum of the body's mass as shown in Figure 1. The COG is the vertical projection of the COM shown in Figure 1. The area of the feet in contact with the ground and the area between them when standing is referred to as the BOS. The COP is located at the point of the vertical ground reaction force and represents the weighted average of the pressure on the surface area in contact with the ground (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998) (See Figure 1). Therefore, when standing on one foot the COP is located within the surface area of that foot and when standing on both feet, the COP is located somewhere between the two feet. The COP is totally independent of the COM. The ground reaction force exists between the ground and the surface in contact with the ground (e.g., the person's feet) and is an equal and opposite reaction to the force of the body weight (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998).

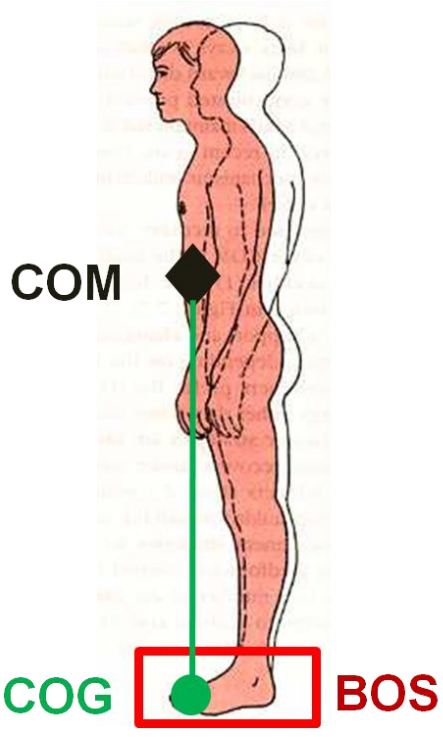


Figure 1 Location of the COM, COG and BOS (Shumway-Cook & Woollacott, 2007).

The weight of the body and the ground reaction force exert a moment about the same point of action. The moment of each force is equal to the product of the force and the perpendicular distance from the force to the point. When the moments due to body weight and ground reaction force are different, the body will sway in the anterior-posterior direction (i.e., forward or backward) and medial-lateral direction (i.e., side to side). While standing, the body continuously attempts to balance these moments to reduce sway (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). The concepts of moments, COG, BOS and COP as they relate to balance will be described as they relate to human balanced control and through an example of balancing a pencil.

2.1.2 Human Balance Control and the Inverted Pendulum Model

Maintaining balance in a human is difficult because the COG is located at a distance 2/3 up from the point of the reaction force, or the feet. Depending on the position of the COG in relation to the point of the reaction force, the body will sway forward or backward. The body is able to react to forward and backward sway in order to recover balance and prevent falling. As a result, the body is continuously swaying forward and backward or in the anterior-posterior (A/P) direction. This phenomenon and factors that determine how the body sways is referred to as the inverted pendulum model shown in Figure 2 (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998).

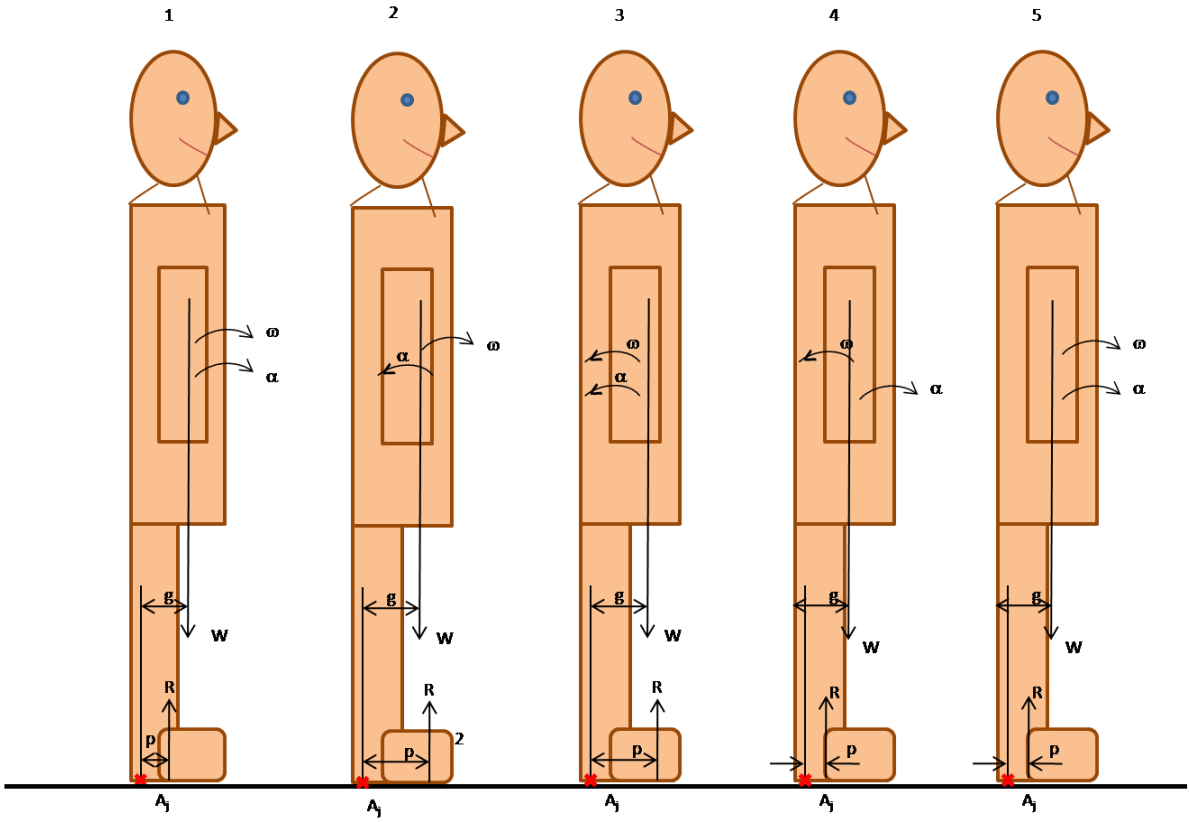


Figure 2 The body modeled as an inverted pendulum adapted from (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998).

In Figure 2, the body's COG, labeled W, acts downward and an equal and opposite reaction force, R, acts upward. Force R represents the COP. These forces remain constant. R acts at a distance, 'p', from the ankle joint, A_j, and W acts at a distance 'g' from the ankle joint. According to the inverted pendulum model, the body sways in the A/P direction while a person is standing still. The body experiences a counterclockwise moment equal to Rp and a clockwise moment equal to Wg, and has a mass moment of inertia equal to the product of the moment of inertia of the whole body about the ankle joint, 'I' and the angular acceleration of the body, 'α'. act to create this forward and backward sway. $Rp - Wg = I\alpha$, where I is the moment of inertia of the whole body about the ankle joint and α = the angular acceleration. At time 1, the COG is ahead of the COP and $Wg > Rp$, resulting in a clockwise angular acceleration or the body

swaying forward. As a result, the body will increase the COP so that it lies anterior or in front of the COG. This causes Rp > Wg and α to reverse. As α reverses, the angular velocity,ω, will start to decrease until at time 3 it reverses. This causes the body to sway backwards. Again, the body needs to adjust to prevent itself. from swaying further backward and so it decreases the COP until it lies behind the COG. This causes α to reverse until at point 5 the angular velocity completely reverses and causes the body to sway forward. This cycle continuously repeats while a person is standing still (Winter D. , 1995).

2.1.3 Maintaining Balance during Perturbation

The COP of the inverted pendulum model has also been found to behave sinusoidally as shown in Figure 3 (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). When the body sways forward, the COP lies outside the COG and has a positive amplitude of acceleration as shown on the graph. When the body sways backward, the COP lies behind the COG and has a negative amplitude of acceleration as shown on the graph. The amplitude of the COP is largest at first and continually decreases. This shows that upon perturbation, a person is accelerating the quickest and sways most. As the body reacts to maintain balance, the body sways less and the acceleration decreases (Shumway-Cook & Woollacott, 2007).

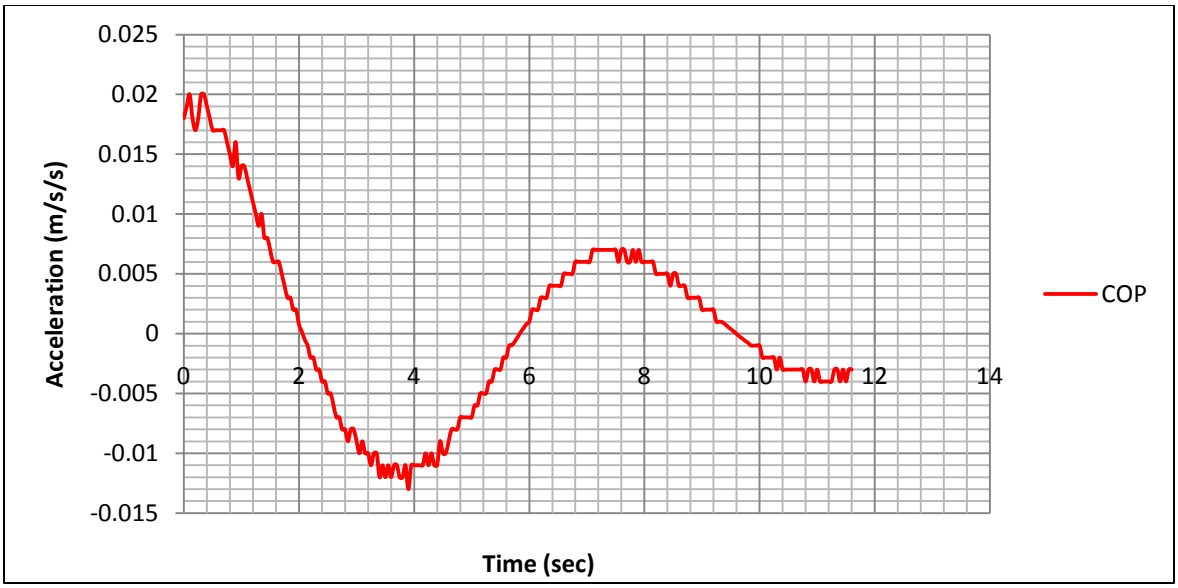


Figure 3: Sinusoidal Behavior of COP- A Working Model Simulation (Gielo-Percak, 2010).

Depending on the acceleration of the perturbation, the body uses different methods to maintain balance. Balance control and anterior-posterior stabilization strategies are utilized to prevent or attempt to prevent a person from falling. There are three major anterior-posterior stabilization strategies: the ankle, hip and stepping strategies, as shown in Figure 4a. The ankle strategy, shown by the number 1 in Figure 4a, only provides a small range of motion and is therefore used when only a small adjustment is needed to maintain balance. The hip strategy, number 2 in Figure 4a, provides a larger range of motion and is used when the person is at their stability limit. Both the ankle and hip strategy can also be combined together in order to maintain balance. The third strategy is the stepping strategy, number 3 in Figure 4a, which is used when the person is about to fall and cannot maintain their balance without taking a step (Pai & Patton, 1997).

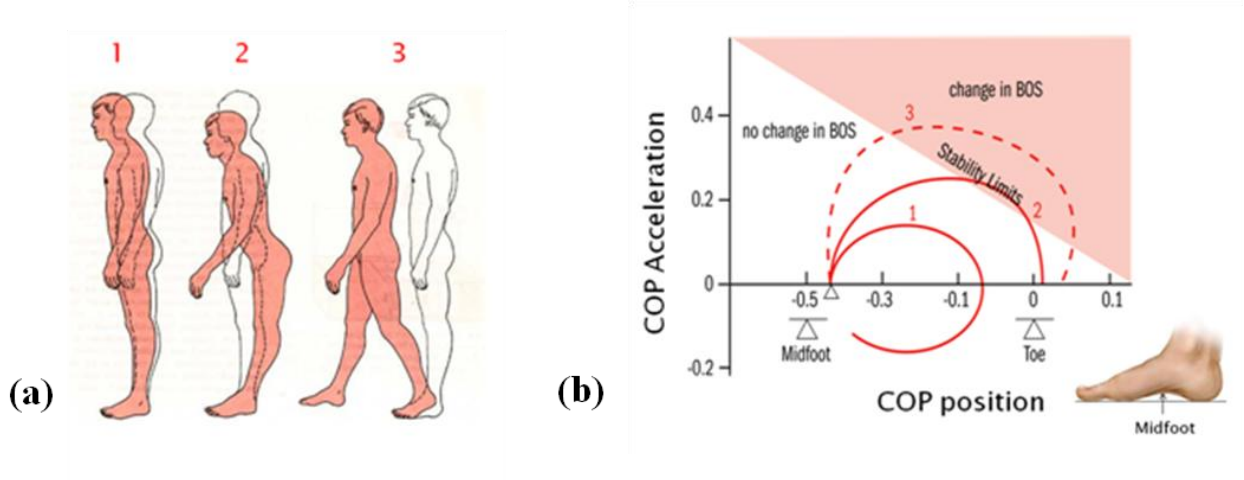


Figure 4 (a) Anterior Posterior Stabilization Strategies and (b) COP acceleration vs. Position (Shumway-Cook & Woollacott, 2007).

These three balance control strategies directly relate to the acceleration and position of the person's COP as shown in Figure 4b. The graph shows the COP acceleration in m/s^2 on the y-axis and the COP position on the x-axis relative to the midfoot or the arch and the toe. Curve 1 corresponds to the ankle strategy, and is a balanced situation because the COP is maintained within the foot. Curve 1 has a small amplitude and acceleration. Curve 2 corresponds to the hip strategy, and when the person is at their stability limit. The COP is just outside the toe region, and has a higher amplitude and acceleration than situation 1. The third curve is the unbalanced situation that requires the stepping strategy to maintain balance. In the off balance situation, the COP is far outside the foot region. This situation also has the highest acceleration and amplitude. From this graph you can see that as the acceleration increases, the loss of balance increases and thus the person has less time to react (Shumway-Cook & Woollacott, 2007).

In addition to amplitude of acceleration, reaction time is an important component of balance. There are two different reaction strategies, the reactive control and proactive control

strategies. The reactive control strategy occurs as a result of a loss of balance or the COP moving outside of the BOS. The second strategy is the proactive control strategy, which occurs in anticipation to an off balance situation. The reaction time involved in a fall is small and thus it is important to utilize both the proactive and reactive strategies. The proactive strategy occurs when a person becomes used to a routine, or occurrence and is able to alter their movements based on anticipation (Shumway-Cook & Woollacott, 2007).

2.1.4 Example of Balance Control: Balancing a Pencil

Balancing a pencil on your finger is an example that illustrates the concepts of balance control. When balancing a pencil at the tip using one finger, the "ground reaction force" is the force of the finger pushing up on the pencil. The COG of the pencil, 'X', is located far above the tip (Figure 5), close to the middle of the pencil. In order to balance a pencil at the tip with one finger, the finger needs to push up at exactly the same point as the COG. If the point of reaction force (R) of the finger does not push up at exactly the same point as the COG, then the weight of the pencil, 'W', creates a moment equal to the product of the weight of the pencil and the perpendicular distance 'd' (distance from point X to point R in Figure 5). The position of COG in relation to 'R' causes the pencil to sway forward or backward. If the COG is located in front of the reaction force, a clockwise moment is created (Figure 5a), causing the pencil to turn forward; if the COG is behind the reaction force a counter-clockwise moment is created (Figure 5b), causing the pencil to turn backward. Balancing a pencil at the tip using one finger is difficult because the COG and point of the reaction force (i.e. the finger) are located far apart.

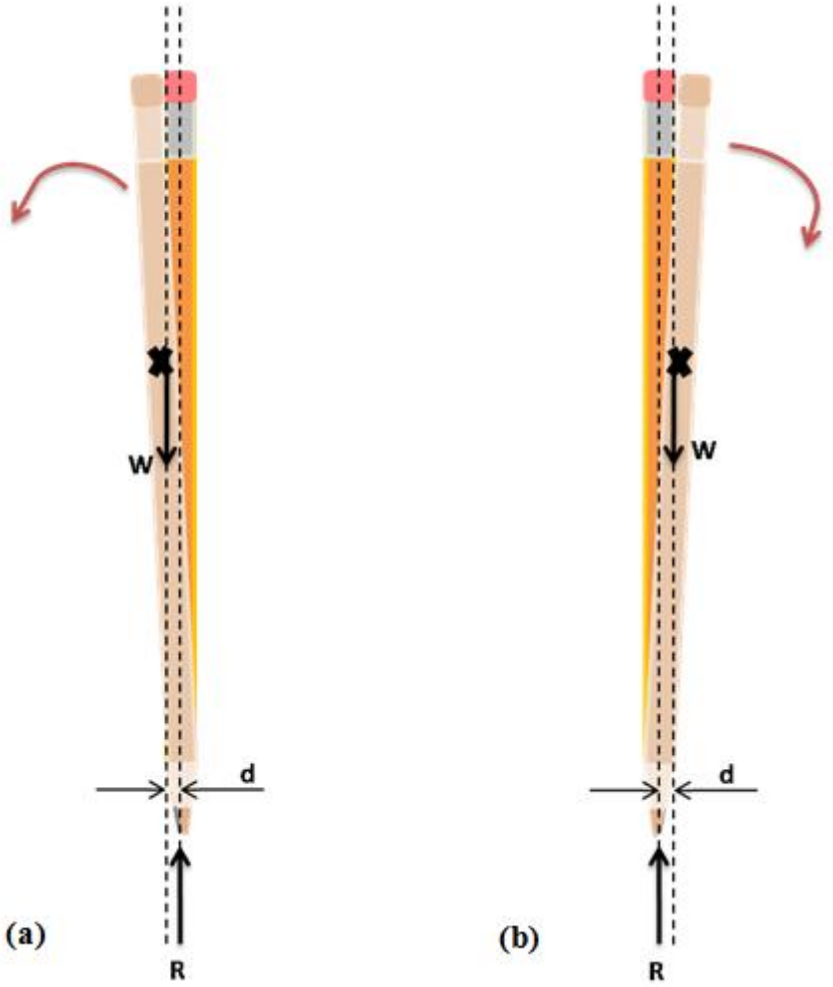


Figure 5: Balancing a pencil on the tip of the finger to illustrate balance control and (a) forward sway and (b) backward sway.

2.2 RISK FACTORS OF FALLING

There are many risk factors that are a result of normal ageing that can cause an elderly person to be more susceptible and likely to fall than a younger adult as shown in Figure 6. Risk factors can be classified into two main categories: extrinsic factors and intrinsic factors. Extrinsic factors are present in the environment or the person's home, and include factors such as stairs, uneven terrain, loose carpet, poor lighting and wet bathroom tiles. Intrinsic factors are present within an individual and generally relate to physiological factors such as balance, vision, proprioception, muscle weakness, reaction time, postural sway, gender, post-fall anxiety syndrome, use of

medication, chronic diseases both neurological and musculoskeletal, and mobility. A fall can be caused by extrinsic factors or intrinsic factors or a combination of the two (Rubenstein, 2006). In the elderly population, physiological factors have a major impact on why the elderly lose their balance and cannot recover when they fall, and why they cannot get up after a fall.

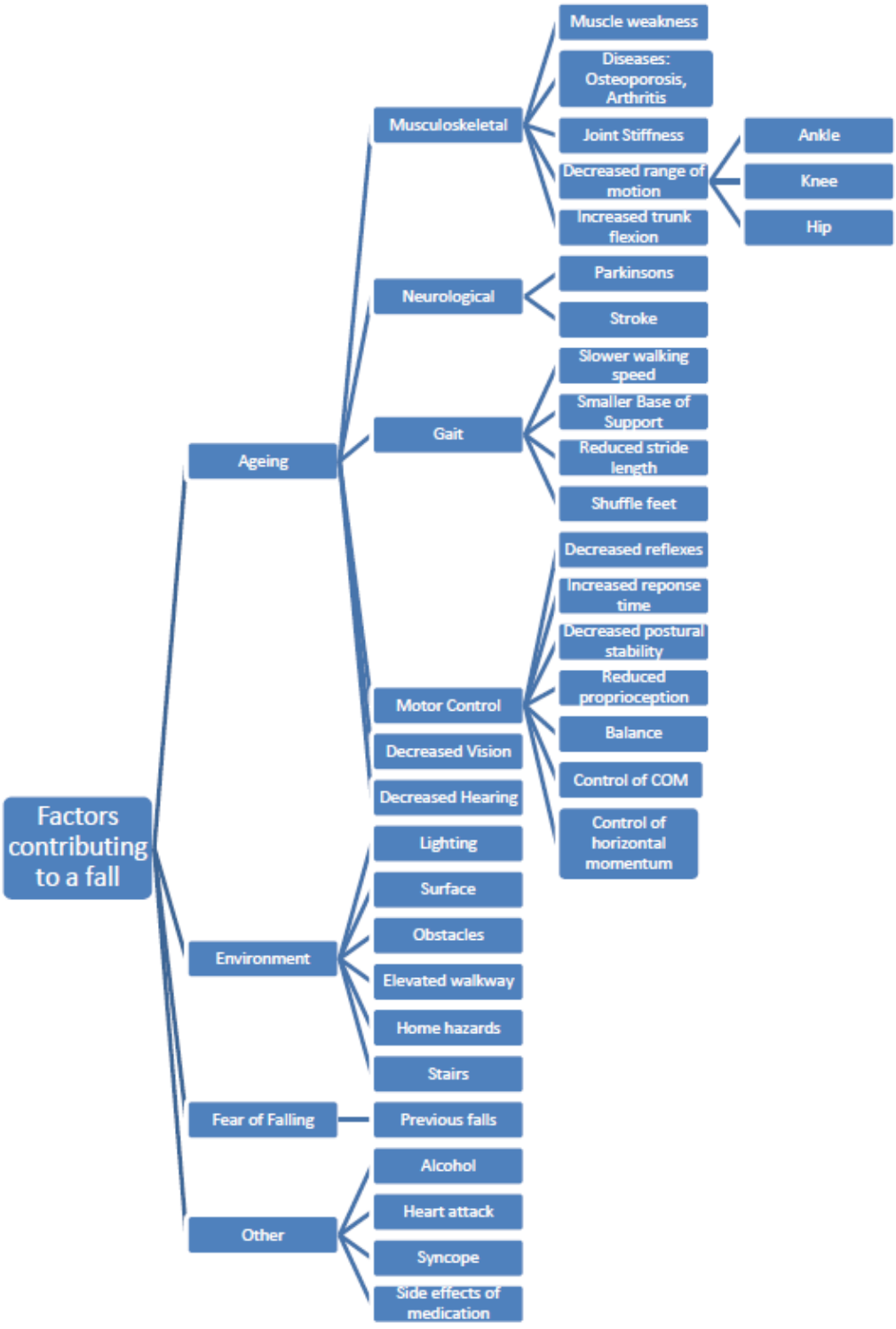


Figure 6: Factors that could contribute to a fall.

Balance control decreases with age. Elderly people have an especially hard time maintaining balance while walking because the COM is outside of the BOS for 80% of the gait

cycle, thus falls often occur while elderly people are walking (Woolley, Czaja, & Drury, 1997). Therefore, it is important to assess and evaluate gait in the elderly because any additional problems will increase their already high risk of falling. For example, elderly people who do not pick up their feet when they walk are more likely to trip over obstacles. Other factors like walking patterns, stride length, response time, and ankle and knee flexion also affect balance control and contribute to why elderly people fall (Voermans, Snijders, Schoon, & Bloem, 2007).

Between the ages of twenty and sixty, there is a 25% increase in response time, which increases the likelihood that an elderly person is going to fall because they are unable to react as quickly to obstacles or changes in their COM (Sturnieks, St George, & Lord, 2008). Response time is an important factor when determining whether a person will be able to recover from a trip or if they will fall. The walking patterns of the elderly also greatly increase their risk of falling. Elderly people have a smaller stride length when they walk and as a result have a slower walking velocity (Winter D. , 1995). Walking velocity is particularly important when an elderly person encounters an obstacle or trips, because the speed of forward rotation of the body related to the person's walking velocity (Bogert, Pavol, & Grabiner, 2002). Elderly people also spend a longer time in the double support phase, which is the phase of the gait cycle where both feet are on the ground (Chong, Chastan, Welter, & Do, 2009). Elderly people have a larger toe out angle, reduced toe pressure, and a higher horizontal heel velocity during heel contact. These gait differences occur in the elderly because they are trying to maintain balance and limit the amount of time during which their COM is outside the BOS. However these gait changes often put them at a greater risk of falling (Woolley, Czaja, & Drury, 1997).

Kinematic and kinetic differences at the trunk, hip, knee, and ankle are also present in the elderly and lead to an increased risk of falling. The elderly have a decreased range of motion in

ankle, knee and hip. Elderly people have reduced ankle range of motion, larger ankle plantar flexion at heel contact, reduced ankle power generation and delayed dorsiflexion. Kemoun et al. found that delay in ankle dorsiflexion during gait is one factor that could prevent falls (Kemoun, Thoumie, Boisson, & Guieu, 2002). When walking down stairs, elderly people also have a substantial decrease in ankle range of motion (Woolley, Czaja, & Drury, 1997). Tripping over an obstacle or falling down stairs can be also be caused by reduced knee flexion (Schillings, Mulder, & Duysens, 2005). Range of motion at the hip is another factor that increases the risk of falling in the elderly. Hip extension in the elderly is reduced during walking. Increased anterior pelvic tilt and hip extension moment during the swing phase also occurs in the elderly (Kemoun, Thoumie, Boisson, & Guieu, 2002). During falls, the elderly also experience increased trunk flexion and trunk velocity. Grabiner et, al found that older adults had trunk flexion angles after a trip that were double the trunk flexion angle of the young subjects (Grabiner, Pavol, & Owings, 2002). Kinetic differences present in the elderly are reduced ankle power generation, increased hip extension moment during swing phase (Grabiner, Pavol, & Owings, 2002), reduced toe pressure and slower generation of joint moments (Lockhart, Smith, & Woldstad, 2005).

The famous slogan "I've fallen and I can't get up!" is both a fear and a reality of many elderly people (Life Alert Emergency Response Inc., 2010). Studies of community dwelling elderly people have shown that approximately 50% of fallers, including those who have not suffered any injuries as a result of the fall, cannot get up on their own after a fall (Tinetti, Lui, & Claus, 1993). The number of people who cannot get up after a fall increases significantly to about 80% over the age of 90 years old (Fleming & Brayne, 2008). The inability to get up after a fall can lead to more serious injuries like dehydration, hypothermia, pneumonia, pressure sores, muscle damage and increased fear of falling (Lord, Sherrington, & Menz, 2000). Lying on the

ground for extended period of time also leads to post fall anxiety syndrome. People with post fall anxiety syndrome alter or avoid daily activities because of the constant fear of another fall. When someone avoids daily activities or changes their gait because of a fear of falling, their muscles become weaker and atrophied and this leads to an abnormal gait and as a result increased risk of falling (Rubenstein, 2006).

2.3 REDUCING THE NEGATIVE EFFECTS OF FALLS

The prevalence and cost of falls in the elderly population has resulted in the development of many rehabilitation methods, risk assessment strategies and devices. All these options are intended to reduce the negative effects of falls (Centers for Disease Control & Prevention, 2009).

2.3.1 Improving Balance Control

Since many factors affect balance, methods of improvement focus on different aspects of balance control. In addition to assessment of fall risk, exercise has been shown to be the most successful form of intervention for reduced fall risk and improved balance control (Stevens, 2005). Several exercise methods for improving balance control are physical therapy, Tai Chi and the Nintendo Wii Fit balance board and gaming console. Physical therapy focuses primarily on increasing muscle strength and flexibility. Tai Chi involves controlling the movement of the COM and reducing body sway (Mao, Hong, & Li, 2006). The Wii Fit balance board and gaming console evaluates COP, BOS, and works on improving reaction time and proprioception (Clark, Bryant, Pua, McCrory, Bennell, & Hunt, 2010). Studies have shown that all three methods are effective for strengthening a person's balance control (Stevens, 2005).

Muscle weakness, especially in the lower extremities, is one cause of falls in the elderly. Studies have shown that muscle weakness negatively affected balance control and postural stability in the elderly because they were not able to generate enough muscle force in response to

a perturbation (Horlings, van Engelen, Allum, & Bloem, 2008). As a result, many studies evaluated the effects of physical therapy and strength training on improved balance and concluded that physical therapy is a successful form of intervention (Rose, 2005).

It is also difficult for the elderly to control movements of their COM and BOS during forward and backward sway, which can cause falls. Tai Chi, shown in Figure 7, is a type of exercise that involves slow forward and backward motions, and has been used in many research studies for the elderly population (Mao, Hong, & Li, 2006). The movements during Tai Chi help the elderly concentrate on slow movements, weight shifting, flexibility, foot positioning and proprioception. Studies have concluded that after participating in Tai Chi the elderly have increased awareness of their body sway and limb movements, reduced fear of falling and considerably improved balance control (Rose, 2005).

Figure 7: Elderly participating in Tai Chi (Rose, 2005).

The Nintendo Wii Balance Board and Wii Fit gaming console has several games and activities that specifically target balance control and COP movements. A study by Clark et al. investigated the validity of the Wii Balance Board for assessing balance control. The study

concluded that the Balance Board is comparable to a laboratory force platform, and can be used to assess standing balance (Clark, Bryant, Pua, McCrory, Bennell, & Hunt, 2010). The Wii Fit video game has balance mini games that are also intended to improve balance through an interactive gaming experience. These games involve moving the user's COP and COM to complete various tasks in the video game and have been used in rehabilitation programs to improve balance control (Clark, Bryant, Pua, McCrory, Bennell, & Hunt, 2010).

2.3.2 Fall Detection Patents

Numerous patents exist for devices that detect falls and the methods by which these devices detect falls vary greatly. These patents can be classified into three categories based on their method of detection: ambient sensors, active protection garments and accelerometer based.

2.3.2.1 Active Protection Garments

Active protection garments (Figure 8) detect certain conditions that could predict a fall and upon these conditions, deploy an airbag intended to cushion the impact of the fall. The design pictured in Figure 8 contains airbags within non-gas porous pockets. The airbags deploy under the following conditions: a rotation rate between sensors on the waistband or torso and at the bottom of the leg exceeding 45 degrees in 0.1 seconds, nearly weightless condition for a period of 0.1 seconds, and lateral and vertical accelerations meeting certain parameters with respect to each other and with respect to normal values (Buckman, 2006). These conditions are not claimed to accurately predict a fall. Therefore, the device could deploy an air bag unexpectedly. Another drawback to this device is that the undergarment could be uncomfortable and hard for an elderly person to put on and take off. However, if the device were optimized to detect only falls, it could be useful in limiting hip injuries.

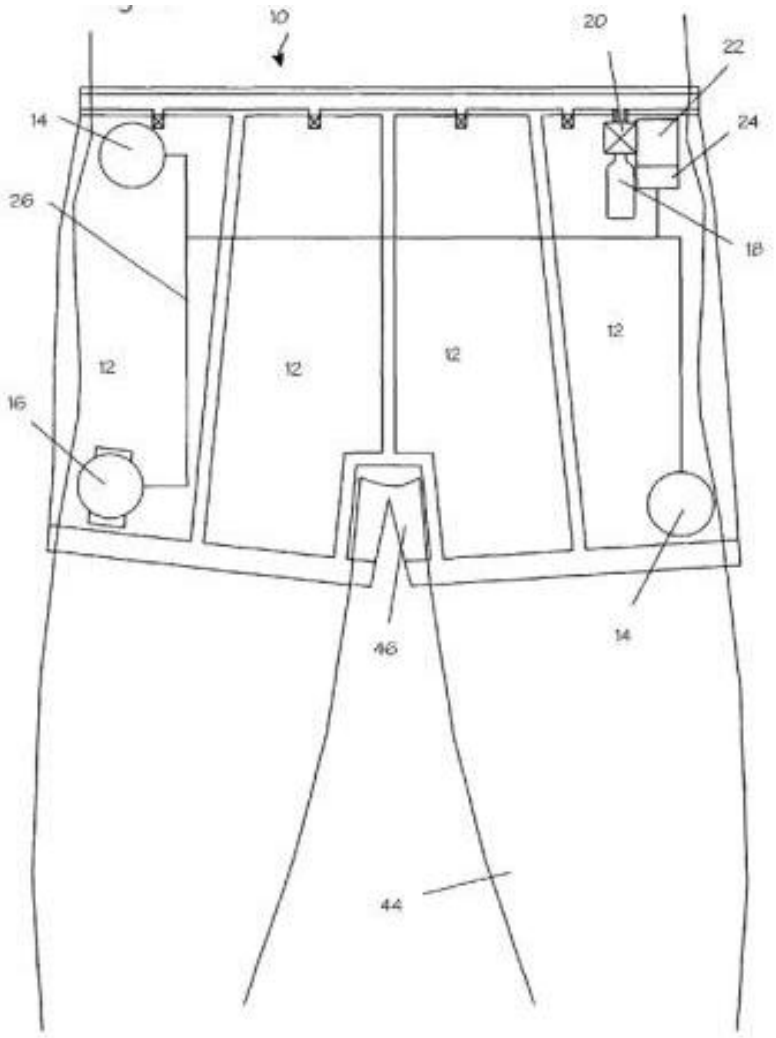
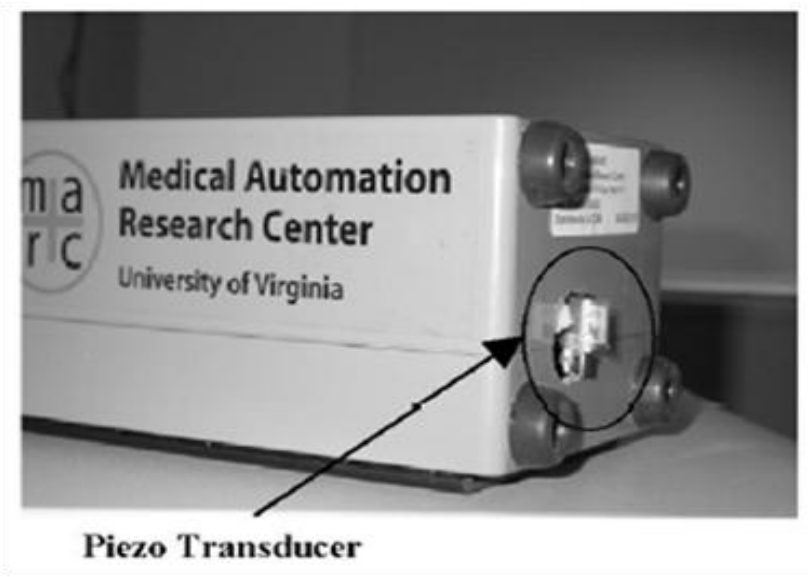


Figure 8: Patent 7,150,048: Active protective garment (Buckman, 2006).

2.3.2.2 Ambient Sensor Patents

Ambient devices differ from wearable fall detection devices in that they detect a fall by measuring indirect factors such as pressure or vibrations in the floor. An ambient floor vibration sensor (Figure 9) was developed and patented by the Medical Automation Research Center at the University of Virginia. This device is positioned on the floor and has a transducer at the bottom of the unit that comes in direct contact with the floor and is used to measure the vibrations patterns in the floor to detect human falls. The device uses parameters such as frequency, amplitude, duration, and succession. The study at the University of Virginia showed that

different activities like walking and running have vastly different vibration patterns and these patterns are also very different from the vibration patterns that occur during and after falling. An anthropomorphic dummy was used to determine the signal and vibration pattern of a 'human' fall, that was then programmed into the device as the only signal and vibration pattern that triggers the device to detect a fall. The fall alert can be sent through a wireless communication as a message to a cell phone or pager. The detection range of this device is around 15 feet, which is large enough to cover most rooms in a home or assisted living facility (Alwan, Felder, Kell, & Dalal, 2004). The obvious limitation of this device is its lack of portability; it cannot be used for active elderly clients that intend to leave one room and go to another room, or go outside, to the supermarket, etc. This device, despite its limitations, was able to perform with 0% false alarms and 100% true positives (Alwan, Felder, Kell, & Dalal, 2004).



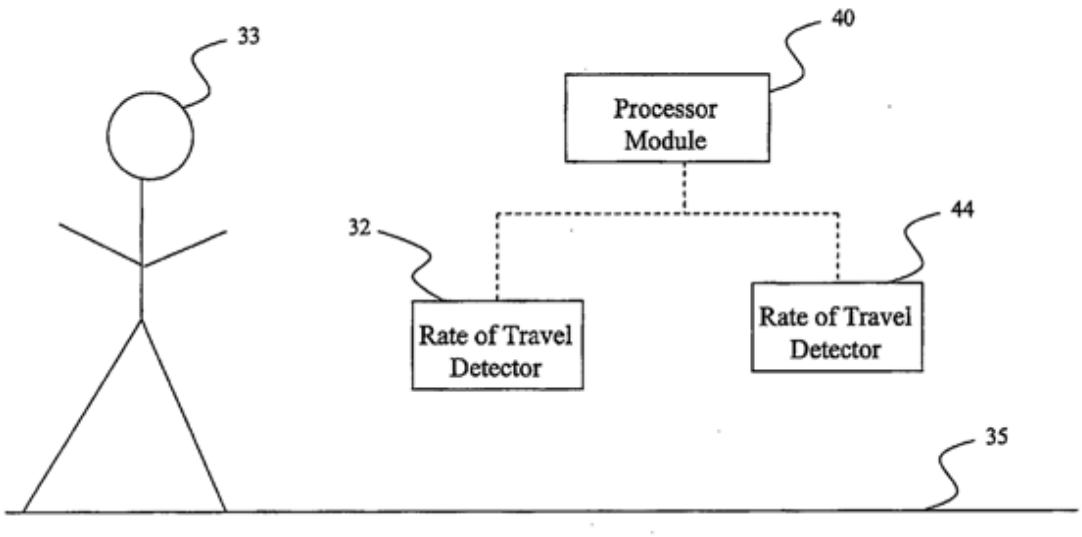


Figure 9: Patent Application Number 2006/0195050 and Floor Vibration Prototpe ((Alwan, Felder, Kell, & Dalal, 2004).

2.3.2.3 Accelerometer-Based Patents

The most common method of detecting a fall is through the use of an accelerometer. Devices that use an accelerometer to detect a fall vary by threshold acceleration values, the algorithm used to confirm the fall, and location of the sensor or sensors on the body.

Patent number 6, 433, 690 (Figure 10) is a wearable accelerometer based device that aims to monitor the user, detect a fall and automatically alert a caregiver or call station that a fall has

occurred. The remote monitoring device (30) is worn on the side of the hip and attaches to the user by a clip. Two biaxial accelerometers with a sensitivity of +/- 2G are enclosed in the monitoring device. These accelerometers are used to measure the angle and acceleration of the body in order to determine if the user is horizontal for longer than two seconds, and if any of the threshold accelerations or angles are exceeded. The patent claims to detect falls with a 95% accuracy rate (Petelenz, Peterson, & Jacobsen, 2002). An advantage of the design is it can automatically detect fall events and rapidly send an alert to a caregiver or call station without any input from the user, which is important if the user were to become unconscious as a result of the fall. By alerting a caregiver or call station of the fall, the faller is able to receive help quickly and reduce long lie injuries and further injuries that could occur as a result of not being able to get up off the floor. The small size of the device allows the user to wear the monitor without interfering with their daily activities. The drawback of the design is that the fall is not detected until after the person is lying horizontal on the ground. Another drawback is that attention needs to be paid to how the device is worn, as false alarms are more likely to occur if the device is not aligned properly with respect to the vertical axis.

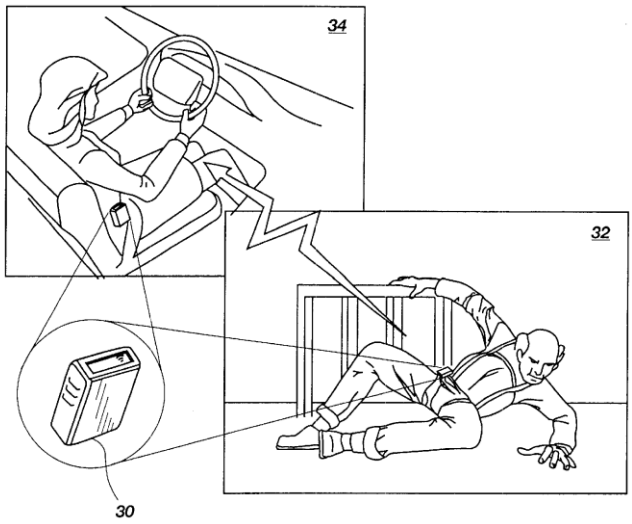


Figure 10: Patent Number 6, 433, 690: Elderly fall monitoring method and device (Petelenz, Peterson, & Jacobsen, 2002).

Accelerometer based designs have also be applied to patient monitoring systems. Patent number 5, 515, 858 (Figure 11) is a monitoring device that resembles a watch and is worn on the wrist. The main objective of this device is to monitor physiological conditions such as temperature and pulse, but the device also incorporates an accelerometer (4) to detect movements of the hand or wrist in order to monitor the actions of the user. The device is able to determine any abnormal acceleration or lack of movement for an extended period of time and automatically transmit an alarm to a surveillance monitor. This design also has a call button which the user can press to send an alert to a caregiver and receive assistance in getting up after a fall (Myllymake, 1996). One benefit of this design is that by monitoring the movements of the user, the device could identify if the user loses consciousness after a fall or remains lying on the ground for a period of time and is not moving. One drawback of the device is that the device does not specifically detect a fall, which is partially addressed with the addition of a call button. This issue is partially addressed with the call button, but if the user becomes unconscious as a result of a

fall and cannot press the button, the faller cannot receive help until a period of time goes by and the device recognizes that the user is not moving.

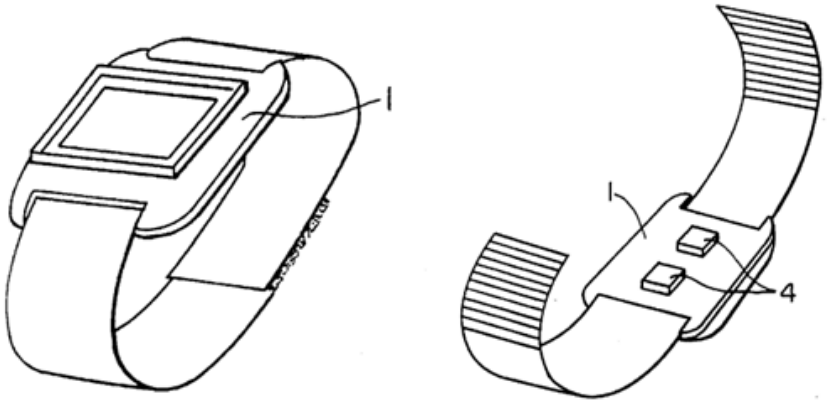


Figure 11: Patent Number 5, 515, 858: Wrist held monitoring device (Myllymake, 1996).

An accelerometer based device can also be used to detect the orientation of the user. Patent number 6, 611, 783 (Figure 12) is a device that uses an accelerometer placed on the back of the user's thigh to detect tilt and variation from a reference angle. The device can be used to monitor patients in a hospital, determine range of motion assessment in physical therapy or prevention and detect falls. The threshold values and reference angles for the accelerometer can be adjusted depending on the application. When threshold value or reference angle is exceeded an alarm is activated to alert and provide feedback to the user. An alert is also sent by a radio frequency transmitter to a caregiver or monitoring station (Kelly & Schoendorfer, 2003). This device is easy to use and does not require any manual activation of the device. One drawback of this device is the risk of false alarms if the user exceeds the reference angle when bending over to pick something up off the ground or sitting in a chair.

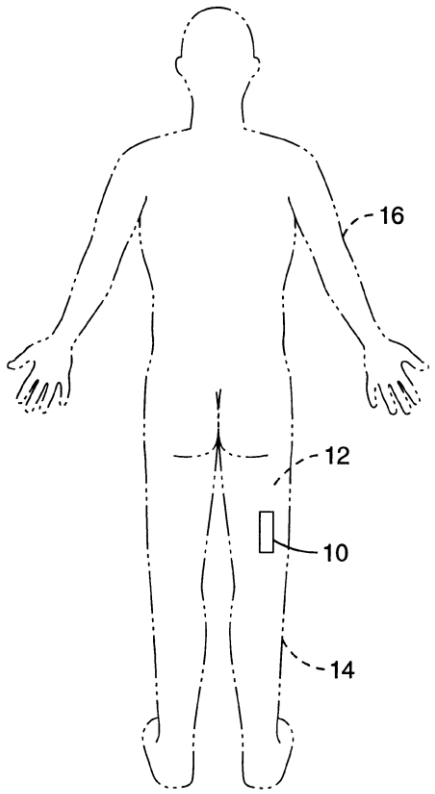


Figure 12: Figure 10: Patent Number 6, 611, 783: Attitude indicator and activity monitoring device (Kelly & Schoendorfer, 2003).

Patent 6, 997, 882 (Figure 13) is a combination of the methods used in the previous patents and monitors the user's activities, measures physiological signals, and detects the velocity, acceleration, orientation and position of the user. This design utilizes sensors attached to a belt with a Velcro strap. The device contains three biaxial accelerometers that can acquire data using six degrees of freedom, and concurrently monitors heart rate using heart rate electrodes. The accelerometers are attached to the belt so that one accelerometer placed over the midline of the back; one accelerometer is placed over the right hip; and the other is placed over the left hip. Bluetooth technology is used to wirelessly transmit the acceleration and physiological data (Parker, Fabeny, Larson, & Monaco, 2006). One major benefit of attaching the sensors at the waist is the sensor's are close to the user's COM. Another advantage of this design is the device is easy to use and does not require any manual activation of the device. A

drawback of this device is potential false alarms as a result of activities of daily living such as bending forward and picking something up off the floor or reaching for something on a shelf.

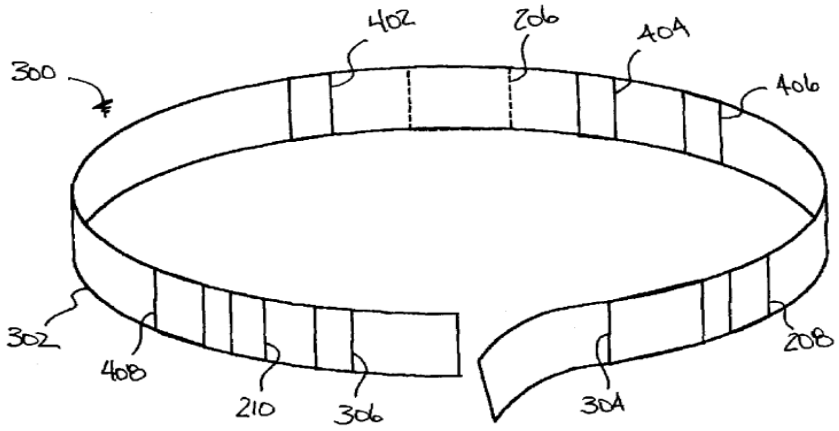


Figure 13: Patent Number 6, 997, 882: Subject monitoring device and method (Parker, Fabeny, Larson, & Monaco, 2006).

2.3.3 Current Fall Detection Technology

Current methods of fall detection are summarized in Table 1. Of these methods, there several commercialized products: MyHalo Monitoring System, BrickHouse Alarm System, and Phillips LifeLine. MyHalo consists of a strap that wraps around the user's chest, a chest strap, and sensing component (Figure 14) (Halo Monitoring, 2009). The sensing component contains a triple-axis accelerometer that automatically detects a fall after the fall has occurred. Upon detection of a fall, the MyHalo system connects to the MyHalo Operating Center who then calls a caretaker. The MyHalo System also sends messages to caretakers via e-mail, text message, or a personal web page. The sensing component is placed no more than 2 inches below the sternum as shown in Figure 14. The device can also be clipped on the pants of the user; however in this position other vital signs (e.g. blood pressure, heart rate) cannot be obtained (Halo Monitoring, 2009). Disadvantages of the MyHalo Monitoring System are that it does not detect a fall before the fall occurs, it is only for use inside the home or immediate area of the home (e.g. yard,

garage), and that wrapping the strap around the chest to put the device on and take it off may be difficult.

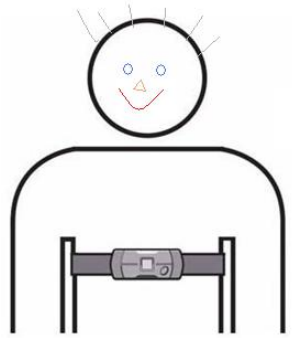
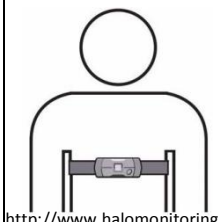


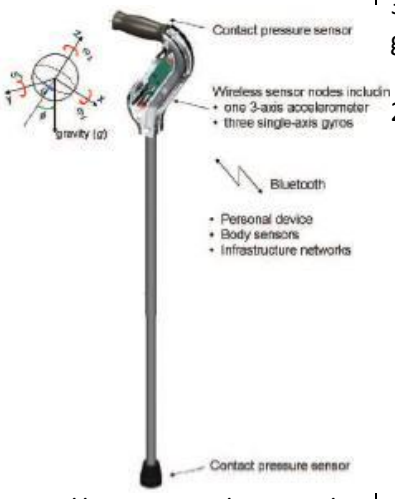
Figure 14: Correct positioning of MyHalo chest strap and device adapted from (Halo Monitoring, 2009).

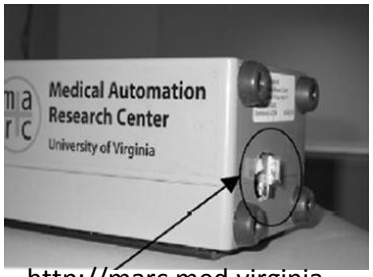
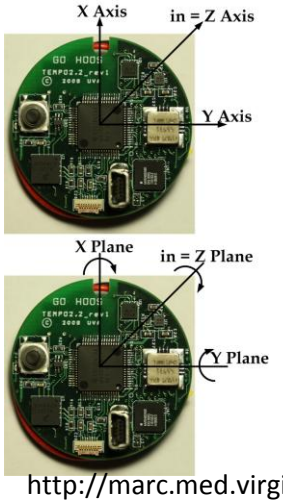
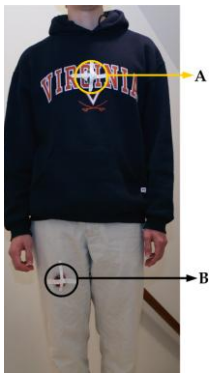
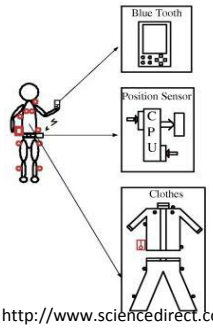
Philips Lifeline Personal Emergency Response System (PERS) is another method of fall protection. This device is contained within an apparatus such as a watch or necklace. The device contains a button and when the buttons is pressed, the system contacts an operator who calls a caretaker or sends emergency response (Figure 15). The button is pressed by the user when in need of assistance and cannot reach help (Koninklijke Philips Electronics, 2009). The main disadvantage of the PERS is that it is user-activated so if the user cannot reach the device, is unconscious, or is in any other way unable to press the button, the device is not beneficial.

Figure 15: Philips LifeLine Personal Emergency Response System (Koninklijke Philips Electronics, 2010).

Table 1: Existing Fall Detection Technology

Device	Sensor for Fall Detection	Location	Pros & Cons
myHalo Fall Monitor  http://www.halomonitoring.com 	3-axis accelerometer	Strap is worn on upper torso near sternum  http://www.halomonitoring.com Clip can be worn on belt	**Pros:** -Automatically detects the fall -Alerts call center that user has fallen who can then alert emergency response -Detects vital signs **Cons:** -Only for home use -Multiple components and set up required -Chest strap could be difficult for the elderly to put on -Does not detect fall before it occurs
Brick House Alert Fall Monitoring and Panic Button System http://www.brickhousealert.com  Fall Detector	3-axis accelerometer	Clips onto side of hip  http://www.brickhousealert.com	**Pros:** -Automatically detects the fall -Alerts central monitoring station who can alert emergency response -Waterproof: can wear in shower **Cons:** -Only for home use -Does not detect fall before it Occurs -Set up required
Philips Lifeline & other Call Buttons  http://www.lifelinesys.com 	No sensor for fall detection	Worn around neck  http://www.lifelinesys.com/ Other buttons are incorporated into a watch 	**Pros:** -May prevent long lie injuries -Contacts a caregiver, monitoring station or emergency response -Device is discrete and doesn't draw attention to user **Cons:** -Does not detect a fall -Not useful is user is unconscious -Requires user activation

Device	Sensor for Fall Detection	Location	Pros & Cons
SmartFall Cane  http://cs.ucla.edu/~alireza/BodyNets08.pdf	3-axis accelerometer 3 single axis gyroscopes 2 Pressure sensors	On cane, not worn  http://cs.ucla.edu/~alireza/BodyNets08.pdf	**Pros:** -Alerts caregiver or monitoring service when user has fallen -Uses subsequence matching Algorithm -No set up required -Does not restrict user's mobility **Cons:** -Not wearable -False alarms -Cane can fall and user not -Bumps into something -Rotation of cane from vertical to horizontal -Does not detect fall before it happens
FallSaver  http://fallsaver.net 	3-axis accelerometer Tilt switch	Back of thigh  http://fallsaver.net	**Pros:** -Minimal false alarms -Easy to use -Detects when user starts to rise from a sitting position -Wireless **Cons:** -Does not detect a fall -Beeps every time user starts to stand up -Going to keep beeping until user sits back down

Device	Sensor for Fall Detection	Location	Pros & Cons
Floor Vibration Sensor  http://marc.med.virginia.edu/pdfs/library/ICTTA_fa	Piezo transducer	On floor, not worn  http://marc.med.virginia.edu/pdfs/library/ICTTA_fall.pdf	**Pros:** -Automatically detects a fall once person is on the ground -No false alarms -Alerts a caregiver of a fall **Cons:** -Does not detect a fall before it occurs -Lack of portability -Only useful for one room
UVirginia TEMPO  http://marc.med.virginia.edu/pdfs/library/ICTTA_fa	3-axis accelerometer 2-axis gyroscope Z-axis gyroscope	Center of trunk and front of thigh  http://marc.med.virginia.edu/pdfs/library/ICTTA_fall.pdf	**Pros:** -Able to recognize different dynamic and static positions -Low cost **Cons:** -Detects fall after it occurs -Difficulty determining whether someone is getting into bed or falling against a wall into a seated position -Attached to user with tape
Smart Coat  http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6V424NMC8781&_user=74021&_rdoc=1&_fmt=&_orig=search&_sort=d&_docanchor=&view=c&_searchStrId=1140042312&_rerunOrigin=google&_acct=C000005878&_version=1&_urlVersion=0	Micro-mercury switches Optical sensors	Embedded in a coat  http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6V424NMC8781&_user=74021&_rdoc=1&_fmt=&_orig=search&_sort=d&_docanchor=&view=c&_searchStrId=1140042312&_rerunOrigin=google&_acct=C000005878&_version=1&_	**Pros:** -Automatically detects a fall -Alerts a monitoring station -May prevent long lie injuries **Cons:** -Does not detect a fall before it occurs -Only detects when user is Horizontal -Can only detect falls in the sagittal plane

Device	Sensor for Fall Detection	Location	Pros & Cons
CSEM wrist fall detector  http://www.csem.ch/docs/Show.aspx?id=9383	2 MEMS 3-axis accelerometers	Wrist  http://www.csem.ch/docs/Show.aspx?id=9383	**Pros:** -Automatically detects a fall -Alerts a monitoring station -Discrete and doesn't draw attention to user -Call button -Able to wear to bed **Cons:** -Does not detect a fall before it occurs -False alarms -Difficult to distinguish fall from wrist acceleration data due to normal movement of the forearm during ADLs
Z Star fall detector  http://www.csem.ch	3-axis accelerometer	Trunk	**Pros:** -Automatically detects a fall -Small size -Alarm when fall occurs - Alerts a monitoring station -Detects orientation of user **Cons:** -Does not detect a fall before it occurs -Limited transmission range -Study not conducted on elderly but by students intentionally falling

The overlying disadvantage of the aforementioned devices is that they do not detect a fall before the fall occurs. Therefore, these device are no proactive and do not help assess a user's risk of falling.

2.3.4 Limitations of Current Technology

The main limitation common among all commercial fall devices is that they do not detect or signal a fall prior to it happening. Therefore, the user does not have a decreased risk of falling

by wearing the device. The devices containing a call button are not useful if the user is unconscious. However, the main downfall of devices on the market is that they do not decrease the risk of falling.

2.4 METHODS FOR ASSESSING BALANCE CONTROL

There are patents and devices on the market that are used for assessing balance control. The patents are vibrotactile based and are used to alert a user of abnormal posture or sensory function. There are two devices currently on the market, the iShoe (Trafton, 2008) and the Wii Fit Balance Board and game (Nintendo, 2009).

2.4.1 Vibrotactile Based Patents

Vibrotactile based patents can be used for monitoring user movements and to correct abnormal posture. The main difference between all of the vibration based feedback devices is the location of the vibration tactors.There are no vibrotactile based patents designed specifically for fall detection or prevention, but some are used to detect balance issues.

Patent number 5, 919, 149 (Figure 16) is a wearable device used to detect balance problems, mainly abnormal postural sway of the upper torso, and to provide feedback to the user to help with rehabilitation. The device collects angular velocity and body tilt angles using two gyroscopes. One of the gyroscopes (12A) is mounted in the middle of the chest for the purpose of measuring side to side movements or roll of the trunk. The other gyroscope is attached on the side of the chest (12B) and is used to measure forward and backward movements or pitch. In this design, feedback is provided to the use through visual and vibration feedback. Visual feedback is provided by projecting an image, of the body sway angle and angular velocity of the trunk, onto a pair of eye glasses (24). Vibrotactile feedback is provided to the user by two vibration tactors that activate when the user has exceeded a particular tilt angle or velocity

(Allum, 1999). One benefit of this design is it can be worn under the user's clothing, so it does not attract any additional attention to the user that could cause embarrassment. This design also attempts to provide the user with feedback about the postural sway which could potentially prevent the user from falling or allow them to anticipate the fall and catch themselves, assuming that the stimuli was activated soon enough and that the user was able to respond to the stimuli quickly. One disadvantage of this design is the purpose of the device is not to detect a fall. A major drawback of this design is the location of the device on the upper chest would make it very difficult for an elderly user to put on and take off.

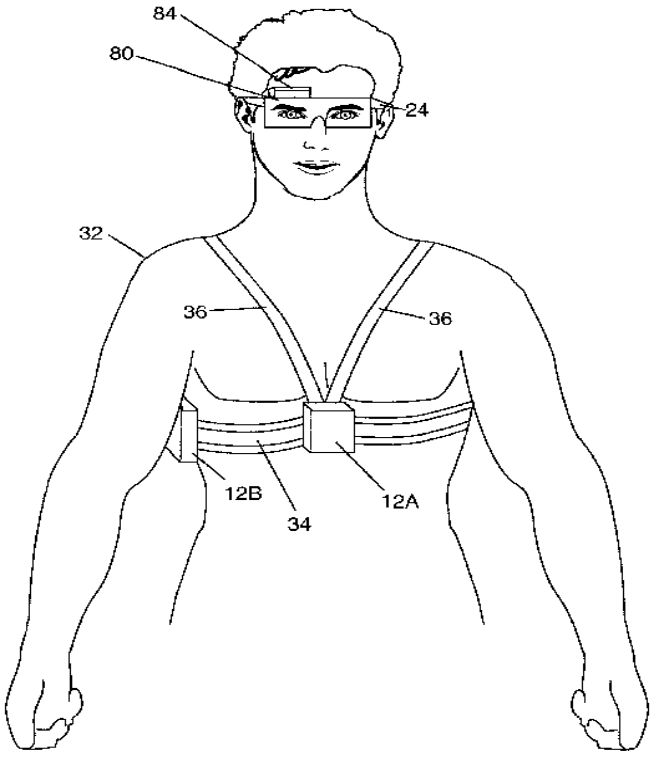


Figure 16: Patent Number 5, 919, 149: Apparatus and method for determination of body sway (Allum, 1999).

Patent number 6, 984, 208 (Figure 17) is a device used to measure the movement and posture of various body parts by transmitting ultrasound signals into the user's muscles. The ultrasound signals are sent to the muscles by ultrasound transducers and recorded by the

receivers (12 & 14). The ultrasound signals are either scattered or reflected by the muscles and this can be used to determine the position, angle, and stiffness of the muscles as well as detect changes in the muscles during particular movements (Zheng, 2006). This design identified an additional method, ultrasound signals, that can be used to measure body movement and detect abnormal body movement. One benefit of this device is the ultrasound sensors and system can be combined with EMG sensors, accelerometers and gyroscopes to apply the design to other applications. A drawback of the design is the design would be beneficial in a clinical setting such as a gait analysis lab or hospital where a trained medical professional could apply the electrodes in the proper location, but would not be practical for home use especially for the elderly because there are too many transducers to apply and proper location is important.

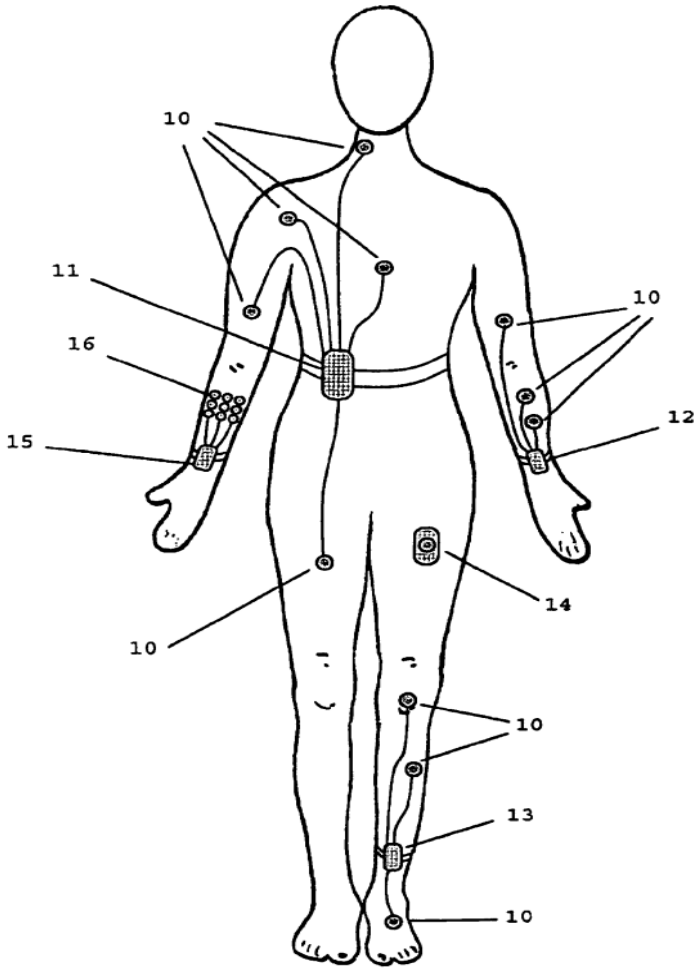


Figure 17: Patent Number 6, 984, 208: Method and apparatus for sensing body movement (Zheng, 2006).

In addition to monitoring body sway and movement, vibrotactile devices can be used to correct posture. Patent number 4, 750, 480 (Figure 18) is a belt that is worn around the waist to detect when the user is slouching and activate a vibration signal to alert the user to correct their posture. The pad (12) is worn on the front of the abdomen and is used to determine whether the abdominal muscles are tightened or relaxed. A switch on the inside of the pad is pressed when the user relaxes their abdominal muscles and this triggers the vibration component to activate and continue vibrating until the user tightens their muscles and releases the switch (Jenness, 1988). A benefit of the design is the device shows that the user is able to respond to the vibration feedback and adjust their posture accordingly. One drawback of the device is the pad is very

large and could be uncomfortable to the wearer. Another drawback of the device is it needs to be worn on the outside of the user's clothing. This would attract unwanted attention to the user, making them self conscious and embarrassed and ultimately not want to use the device.

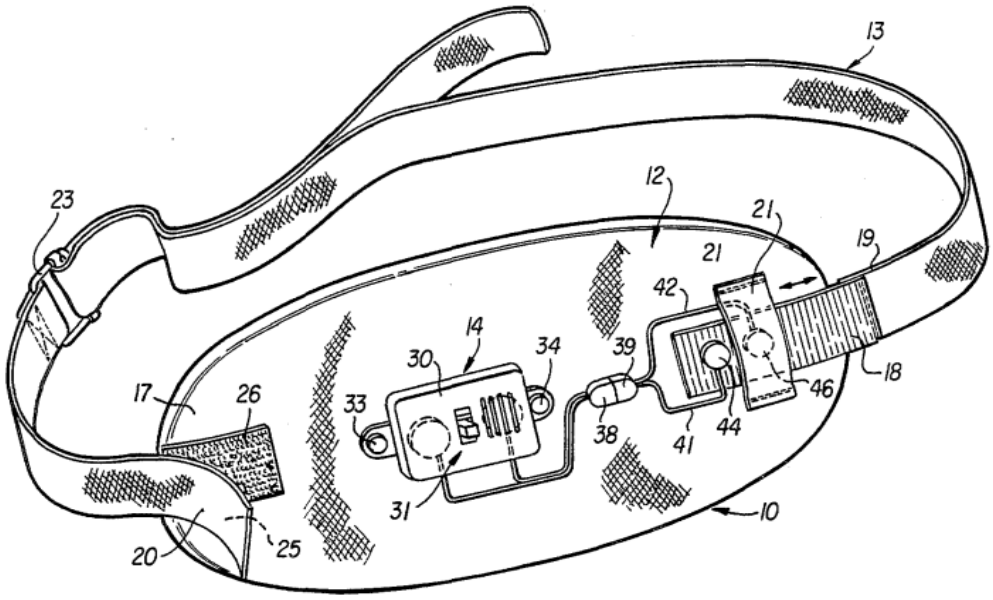


Figure 18: Patent Number 4, 750, 480: Posture-correcting devices (Jenness, 1988).

There is also a design that utilizes vibrotactile feedback to enhance sensory function in the foot, thereby improving human balance. Vibration actuators or electrodes are contained within a wearable system such as a sock or shoe insole (Figure 19).These actuators or electrodes provide electrical stimulation to the mechanoreceptors in the foot and ankle to increase their sensitivity and ability to transmit sensory information to the central nervous system (Harry, Collins, Prplata, & Kleshinkski, 2004). By increasing the sensory performance of the mechanoreceptors in the foot, the device may improve balance in the user. However, this device is located in the user's shoes, which is a difficult location for the elderly to use. Also, the vibration from this device could startle the elderly user and cause them to fall. This device is

being pursued as a therapeutic tool by a small business, Afferent Corporation, however, it is not currently on the market.

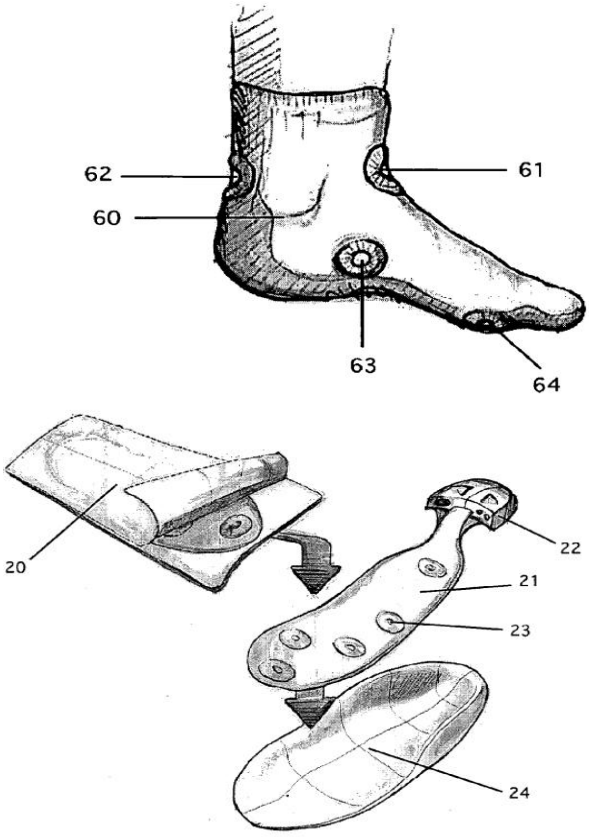


Figure 19: Patent 10/793,729: Method and apparatus for improving human balance and gait (Harry, Collins, Prplata, & Kleshinkski, 2004).

2.4.2 Current Devices for Assessing Balance Control

There are two devices, the iShoe and the Wii Fit Balance Board and gaming system that are currently on the market for assessing and monitoring balance control. A summary of these devices can be found in Table 2.

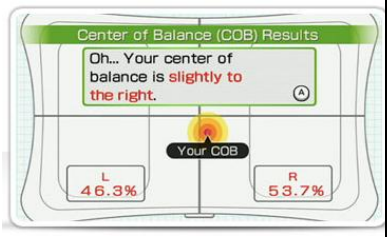
The iShoe is a shoe insert that contains pressure sensors to determine abnormal pressure patterns in the user's feet. This device can be connected to a computer and a medical professional is able to analyze the data to determine and monitor a user's balance control

(Trafton, 2008). One major disadvantage of the iShoe is that the data needs to be brought to a doctor's office and interpreted by a medical professional. Another disadvantage is the device is not proactive and does not provide instant feedback to the user. This device is only beneficial when the user is wearing shoes, and putting on the device could be difficult for an elderly user.

The Wii Fit Balance Board and gaming system provides an interactive way for user's to assess their balance control. The balance board contains pressure sensors that display the user's COP on a television screen through a video game interface. The Wii Fit game has several games for user's to play in order to improve their balance control (Nintendo, 2009). While this device provides visual feedback to the user, and an opportunity to work on their balance control, this device does not directly assess the user's risk of falling. This device is also targeted for the younger population, and the video games are not catered to an elderly user.

While both of these devices are able to assess and monitor balance control, neither device is proactive or directly assesses the user's risk of falling. The devices are also not designed to be easy for an elderly person to use them.

Table 2: Devices that Assess/Monitor Balance Control

Device	Sensor for Fall Detection	Location	Pros & Cons
iShoe  http://web.mit.edu/newsoffice/2008/i-shoe-0716.html	Pressure sensors	Insert for a shoe  http://web.mit.edu/newsoffice/2008/i-shoe-0716.html	**Pros:** -Able to detect abnormal pressure patterns -Bluetooth enabled to transfer data to a doctor -Monitors balance control **Cons:** -Does not detect a fall before it occurs -User must be wearing a shoe to use the device -Data must be interpreted by a medical professional
Wii Fit Balance Board  http://www.nintendo.com/wii/what/accessories/balanceboard  http://wiifit.com/body-test/#body-control	Pressure sensors	Board that users stand on  http://wiifit.com/body-test/#body-control	**Pros:** -Videogame to improve balance control -User awareness of balance control -Measure COP -Cheaper than a force platform **Cons:** -Games are not catered to the elderly -Many steps for operating the device -Games are not related to repetitive daily activities

Although there are currently devices on the market to alert emergency response, detect a fall after it occurs and assess balance control, there is a need for a device that proactively monitors balance control, provides instant feedback, and notifies the user of their risk of falling.

2.5 PROACTIVELY MONITORING BALANCE CONTROL

An important step in preventing falls is identifying of a person's risk of falling as soon as possible (Zijlstra, Bisseling, Schlumbohm, & Baldus, 2010)**.** This can be achieved by assessing and proactively monitoring balance control. One way to proactively monitor balance control is through the use of sensors placed on the body. Sensors such as accelerometers, gyroscopes, pressure transducers or strain gauges are portable, can be used in a person's home and are not very expensive (Janssen, Kulcu, Horemans, Stam, & Bussmann, 2008)**.**

2.5.1 Accelerometry

An accelerometer measures acceleration relative to freefall and is used in many of the fall detection applications. Accelerometers can measure acceleration on one axis, two axis, or three. Accelerometers are micro electro-mechanical systems (MEMs), and behave as a series of small dashpots, damped with a gas (Figure 20). As the spring-mass system moves due to an external acceleration being applied, the electrical impedance of the system changes. These changes are outputted digitally or through an analog signal and must be processed to be interpreted in a circuit. Some considerations taken in choosing an accelerometer for device design include sensitivity (for analog accelerometers, this will mean amplitude of the change of the output voltage relative to g's of acceleration), maximum measurable acceleration, and number of axis on which to measure acceleration. Accelerometers can be combined with gyroscopes, which in the case of device design are sensors that can measure orientation based on the principles of angular momentum (Omega Engineering Inc., 2003).

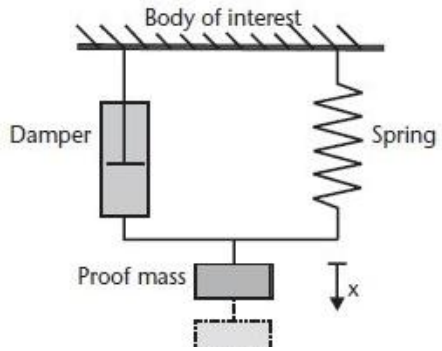


Figure 20: MEMS accelerometer and muscle behavior model (Fung, 1993).

There are a number of functions that an accelerometer can have, one of which is tilt. This is due to the accelerometer's inherent ability to detect gravitational acceleration on each sensing axis relative to its rotational position. A MEMs accelerometer can be modeled after a cantilever beam with a proof mass in a gas-damped chamber between two capacitive plates. As gravity acts on the beam, it displaces from its neutral position toward the lower capacitive plate. This would be read the same as if a force was displacing the sensor upwards accelerating at the equivalent local gravity (Figure 20). This is a useful application for devices that are intended to make use of this rotational property of the accelerometer. However, in applications where the accelerometer is intended to measure motion components as well as rotational components of movement, or only motion components, it can become difficult to discern which portions of the signal are resultant of tilt and which are of motion.

The key to understanding the effects of different components of motion on the signal can be understood from observing the output on the timescale. A rotation of the accelerometer will result in a "DC" or step-like component, where the baseline of the signal will change, and the sensing axis will reflect between zero and one g at rest. The offset from the gravitational acceleration vector will be critical in determining the observed output, where the output will be reflected by $\cos(\Theta)*9.81\ m/s^2$ (gravity). Θ will be the angle which the sensor is offset from

directly measuring the reactive force of gravitational acceleration. Because of this, the accelerometer's output will be a nonlinear (sinusoidal) response to changes in orientation on each sensing axis. It is important to note that an accelerometer will only sense changes in acceleration on the axis that it has defined as measurement axis, and will not observe any changes if there is acceleration on an axis that has no sensor associated with it.

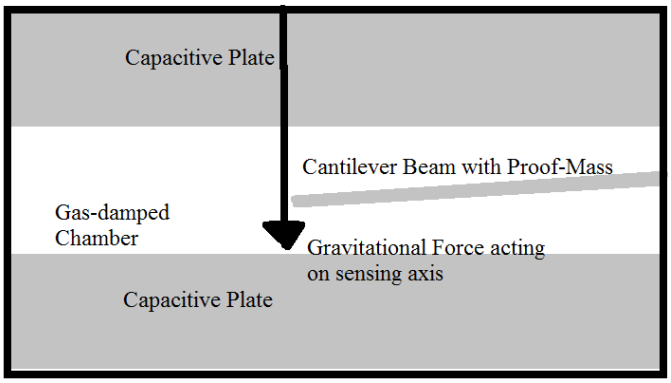


Figure 21: The Effect of Gravitation Acceleration on an MEMs Accelerometer.

Digital Accelerometers output a serial stream of data rather than a voltage on each of the independent measuring axis. They utilize pulse width modulation to determine the specific acceleration magnitude recorded. This means that there is a square wave with a certain frequency and a varying duty cycle. The duty cycle will be proportionate to the severity of acceleration acting on the sensing axis, and obtaining an acceleration measure from this data will be dependent on the sensitivity of the sensing axis (i.e. +/- 2g accelerometer on a 100% duty cycle will be measuring either +2g or -2g, 50% duty cycle will be recording either +1g or -1g). This technique of pulse-width modulation is necessary because of the Boolean nature of digital electronics, as a device can either be fully on or fully off, and an accelerometer requires intermediary measurements.

Accelerometers have been used frequently in motion analysis research studies for both walking and balance control (Janssen W. G., Bussmann, Horemans, & Stam, 2005). According to a study by Janssen et al,

> The results of the study indicate that accelerometer is able to provide a sensitive measure of balance during the sit-to-stand movement. Accelerometery offers the benefit of low cost and portability. An important advantage of accelerometery therefore is that it allows measurement outside of a movement laboratory.
> (Janssen, Kulcu, Horemans, Stam, & Bussmann, 2008)

Accelerometer signals have also been shown to "contain information on kinematic events that will enable us to define time markers to describe the phasing and duration of the sit-to-stand movement without the use of a gait laboratory" (Janssen W. G., Bussmann, Horemans, & Stam, 2005)**.**

Research has shown that the acceleration of a person's COM can determine how close they are to their stability limit. For example, a quick acceleration would cause a person to become more unstable. As a result, studies have used accelerometry to assess balance control and have shown that it is an accurate, affordable alternative to using more expensive motion analysis system. Therefore, the team investigated accelerometry as a design alternative.

2.5.2 Gyroscopes

Gyroscopes are sensors primarily used to measure position, tilt, and orientation. As a mechanical system, a gyroscope is a spinning wheel with a high angular momentum. The system is mounted within two rotors that make the system highly susceptible to external torque forces. This design can be found in gyroscope toys. MEMs gyroscopes, such as those present in gyroscope ICs used in fall detection devices make use of a spinning disc built into a vibrating structure. Because of the spinning disc configuration, larger acceleration forces will act on the outside of the disc versus the inside of the disc. Therefore changed in rotation will act with

greater force on the outside of the disc, causing it to stray from its axis. By using two discs, as they stray from their axis the capacitance between them will change. In order to accommodate the extremely small size of this complex system, complex micromachining is necessary to create a gyroscope sensor. Because of this, these sensors tend to be costly (Torrence, 2008).

Gyroscopes have been used to monitor the tilt of the body, in particular the angle of the trunk at the hip and the angle between the body and the ground when standing. Certain angles of the body have been shown to correspond to unbalanced situations. In addition, gyroscopes have been used in combination with accelerometers in fall detection devices. Therefore, the team investigated gyroscopes for use in the design of a balance control indicator.

2.5.3 Strain Gauges and Pressure Transducers

Strain gauges are electrical resistors configured to measure strain by means of changing resistance/conductance values when the device is deformed. A Typical strain gauge will change in its resistance value when compressive or tensile force is applied. This changing value of resistance can easily be used in a simple circuit to produce a varying voltage or current change as necessary. There are many different types of strain gauges, ranging from simple foil strain gauges through piezoelectric sensors designed from semiconductive material, with capacitive and fiber optic strain gauges in between. Different types of strain gauges tend to have markedly different gauge factors (or sensitivity to strain) from one another. For instance, piezoelectric sensors are much more sensitive to deformation than foil gauges. These devices are also much more sensitive to temperature deviations, like most resistive circuits (Omega Engineering Inc., 2003).

Pressure transducers convert pressure signals into an analog electrical signal. They sometimes implement and are very similar to strain gauges. When a strain gauge, either capacitive, resistive, inductive, or piezoelectric is attached to a wheatstone bridge in order to produce a small amplitude electrical voltage signal, it is considered a pressure transducer. Pressure transducers are available in many different forms with many different applications ranging from small IC component-mounted pressure transducers to large pressure transducers useful in industrial and automotive settings. The analog voltage output of pressure transducers makes them easily useable via basic signal processing (Aston, 1990).

Research has shown that variations in the position of a person's COP along their feet can represent how well the person is balanced. For example, if a person is unbalanced their COP may be positioned closer to their toes while if a person is balanced their COP may be positioned close to the middle of their foot (Pai & Patton, 1997). Current technology, such as the iShoe, has utilized the concept of measuring pressure along the feet as a means of assessing balance control (Trafton, 2008). Since pressure transducers and strain gauges are two methods of measuring pressure, the team investigated these devices for use in the current balance control indicator.

2.5.4 Motion Analysis

Motion analysis laboratories are used to assess balance control and gait. These laboratories have expensive infrared motion capture cameras, and force platforms. The force platform measurements are used to assess COP and BOS during both standing and gait. Spherical reflective markers are placed on anatomical landmarks on the body, and the infrared cameras track the movement of the markers. The video footage is analyzed to assess the movement and location of the markers during the activity, and can be used to analyze body sway and the movement of the COM (Culhane, O'Connor, & Lyons, 2005). While motion analysis can be an

effective and beneficial assessment of balance and gait, the process is expensive, time consuming and can only be performed in a laboratory setting.

2.5.5 Advantages and Disadvantages of Current Balance Monitoring Methods

The accelerometer, pressure transducers and strain gauges, and gyroscope are four devices that can be used to measure acceleration, pressure, and tilt respectively. A person's acceleration, pressure under the foot, and tilt of a person's body has also been shown to relate to balance control. Therefore, each device was investigated for use in the current balance control indicator.

The advantages of the accelerometer are that it is most accurate when placed on the trunk which is an accessible location and could make the device easily positioned. In addition, the accelerometer has shown to be accurate in monitoring balance control during activities such as the sit-to-stand and has been used to monitor balance control in place of a motion analysis system.

Pressure transducers and strain gauges could be used to measure pressure under the foot and monitor a person's balance control by tracking their COP. Although COP has been shown to be one factor that relates to a person's balance control, acceleration of a person's COM has shown to be more reliable in indicating whether a person is balanced. In addition, the sensors need to be placed under the foot, an inaccessible location and requiring the use of footwear.

Gyroscopes have been used in combination with accelerometers, but only measure tilt. Although tilt can indicate whether a person is balanced or unbalanced, it is not a characteristic shown research to be characteristic of balance control. Therefore, using a gyroscope in the balance control indicator may not be the most accurate means of interpreting a person's balance condition.

Through research, pressure and acceleration were identified as the signals most sensitive to balance control. Therefore, preliminary designs utilized the measurement of these signals or a combination of these signals to create the balance control indicator.

3 PROJECT STRATEGY

Falls are the leading cause of unintentional death in the elderly population (Centers for Disease Control & Prevention, 2009). Although there are many physical risk factors associated with aging that are inevitable and can cause falls, balance control can be improved to reduce a person's risk of falling (Stevens, 2005). Current fall-detection technology, such as MyHalo Monitoring is not proactive and detects a fall after it has occurred. Proactive balance control technology is not geared towards the elderly population (e.g. Wii Fit) or requires professional intervention (e.g. iShoe). The Wii Fit is not catered to the elderly population because the games require running, jumping and fast movements which could put an elderly user at risk for injury (Clark, Bryant, Pua, McCrory, Bennell, & Hunt, 2010). The Wii Fit is also a video game interface which could be too complicated for an elderly person to set up and operate. Therefore, we identified the need for a device that indicates a user's balance control and risk of falling and is catered to the elderly population. This chapter details the strategic design process used to determine client needs and wants, objectives, and constraints of the device. The final sections discuss the methods that the team created to reach objectives of the design.

3.1 CLARIFYING THE ORIGINAL PROBLEM STATEMENT

The team was originally given the following problem statement by Professor Gielo-Perczak and Professor Mendelson:

"Design an early balance control device which can be used particularly by the elderly. The first part of a project will involve data collection and analysis of signals during daily movement, in particular situations when individuals can potentially lose their balance. A sensor attached to the subject will be used to collect the data. Based on the data acquired, an early balance control indicator will be designed and tested."

From the original client statement, the team identified the main goal of the project—to design an early and wearable balance control indicator for the elderly. Three steps that were identified to achieve this goal as established by the original problem statement are as follows:

1. Use a wearable sensor to collect data during daily movement and situations when individuals can lose balance
2. Analyze the data
3. Design an early balance control indicator based on the data analysis

A key piece of information that was missing from the original problem statement was the purpose for creating a balance control indicator. To establish the purpose of the design, the team further defined the problem by using a strategy of design thinking called 'decomposition' where the larger problem was broken down into smaller, subproblems (Dym & Little, 2004).

To narrow the problem, the team researched the root cause of falls in the elderly population. Numerous risk factors can contribute to the likelihood of falling. Since an elderly person can obtain any combination of these risk factors, an infinite amount of situations could lead to a fall. Therefore, the team identified the most costly and traumatic effects of falls as shown in Figure 22. Nonfatal fall injuries were found to account for the majority of healthcare expenditures due to falls (Centers for Disease Control & Prevention, 2009) and of these nonfatal injuries, fractures account for 61% of the costs. Among fractures, hip fractures are the most costly and traumatic fall injury as 1 in 5 who suffer a hip fracture dies within a year. Women are 2/3 more likely to sustain a fracture due to an unintentional fall than men and 72% of elderly admitted to the hospital due to a hip fracture were women (Centers for Disease Control & Prevention, 2009).

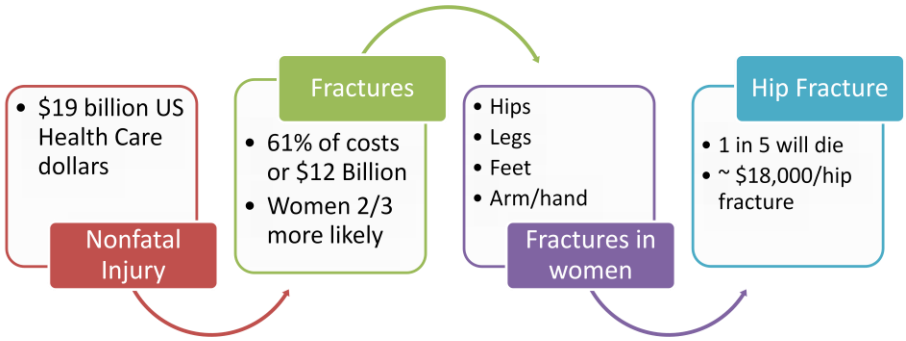


Figure 22: Hip fracture most costly and traumatic effect of falls.

As a result, the team identified the causes of hip fractures and specified a cause that could be ameliorated via the present balance control indicator. The leading causes of hip fracture were tripping and slipping, but numerous variables (e.g. lighting, surface, obstacles) can contribute to tripping and slipping (Kerr, White, Barr, & Mollan, 1997). Fractures were found to be prevalent among people with lower-body weakness, problems with gait and balance, and chronic diseases, e.g. Parkinson's Diseases, arthritis (The National Council on the Aging, 2005). Studies showed that falls can be prevented by increasing strength and balance control by exercises such as Tai Chi and strength training (The National Council on the Aging, 2005).

In addition, clinical assessment where a person is tested for gait, balance, and neurological function, and reviewing medication allows a physician to individually manage a patient's needs in order to prevent falling. For example, the physician may refer the patient to a specialist or change a medication. However, these assessments are recommended for high fall risk patients or those who have already suffered a fall and have gait and balance problems (The National Council on the Aging, 2005). As result, many elderly are not assessed for their fall risk and could be at risk of falling. These people may not be assessed for their fall risk before it is too late.

3.2 OBJECTIVES & CONSTRAINTS

Therefore, the team identified two goals of the project: to strengthen the user's awareness of their balance control, and to proactively force user rehabilitation in order to operate the device. Objectives were the criteria that the project and device addressed in order to meet the defined goals. The objectives of this project are outlined in Figure 23.

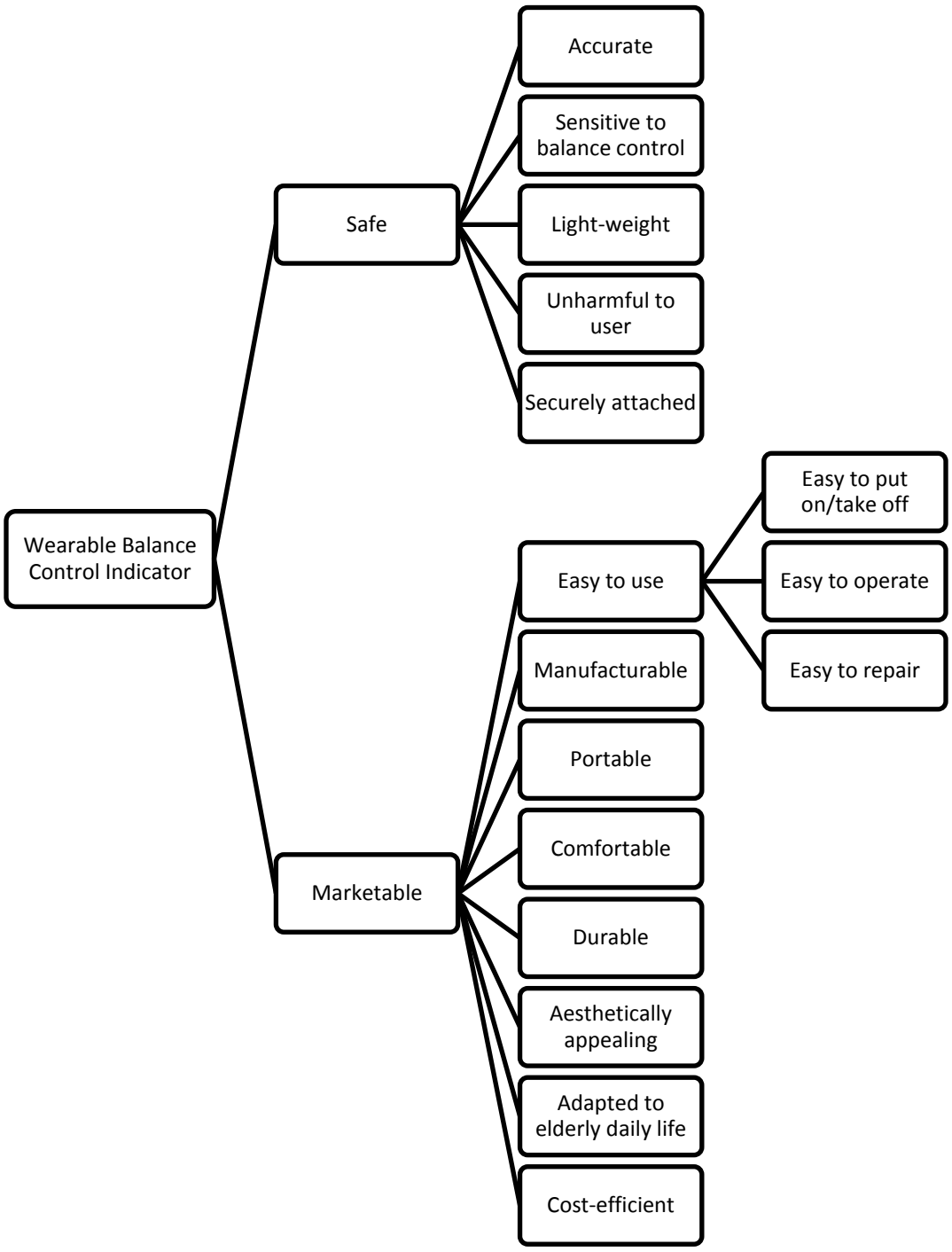


Figure 23: Objectives of the design.

A constraint was a condition that imposes a restriction or limitation on the design (Dym & Little, 2003). Constraints of the design included elderly disabilities, available testing equipment, and affordability. Because of the limited gait analysis capacity we had in the

laboratory, we needed to confirm balance condition by making use of only a single force platform and a tri-axial accelerometer. We also needed to factor in affordability and user acceptance. In order for a device to reach the widest population of elderly clients, it needed to be affordable and uncomplicated in its operation. We needed to account for varying elderly disabilities, making sure that the feedback provided by the device, as well as the controls to operate the device were minimalist and simple. We needed a means of alerting the user that would be able to be understood and sensed by a wide margin of elderly subjects, and directions that were simple for clients to understand.

3.3 REVISED PROBLEM STATEMENT

According to the goals, objectives, and constraints of the project, the design team revised the original client statement as follows:

Design balance control indicator which can be used particularly by the elderly to improve awareness of their balance control. The first part of a project will involve data collection and analysis of acceleration of COP signals during both a balanced and unbalanced STS activity. An accelerometer data logger attached to the subject and an AMTI force platform will be used to collect the data. Based on the data acquired, a lightweight and compact-form-factor balance control indicator will be designed and tested that will be worn on a convenient location for the client and utilize a daily activity to strengthen the user's awareness of their balance condition by notifying the user of an off balance STS situation before a fall has occurred and proactively force user rehabilitation by requiring the STS activity in order for the device to function. The device will need to be easy to use, affordable, reliable, sensitive to changes in balance control, and require no professional intervention to interpret its results. The device should be placed in a

location that is easy to take off and put on, and should utilize a repetitive daily activity so the device can easily be incorporated into the daily life of the user.

3.4 PROJECT APPROACH

To establish an approach, the team identified specific subtasks under each step derived from the problem statement. The subtasks were as follows:

1. Use a wearable sensor to collect data during daily movement and situations when individuals can lose balance
    a. Chose an activity during which a person can lose their balance
    b. Chose a sensor that can be used to collect data during the activity
    c. Chose a location for the sensor
2. Analyze the data
    a. Identify characteristic parameters of the collected data
    b. Choose specific parameter to analyze
3. Design a balance control indicator based on the data analysis
    a. Identify wants and needs of stakeholders (i.e., client, designers, users)
    b. Establish and prioritize design objectives
    c. Identify constraints of the design

Therefore the team completed four tasks:

1. Identify a daily activity for the device to monitor
2. Identify a signal that can be used to monitor balancing control and an accompanying sensor to detect that signal
3. Identify a specific location where the device can be easily worn and the signal can be accurately monitored

4. Identify a specific parameter of the signal to analyze and use to distinguish between a balanced and unbalanced condition

The following chapter details how the design team completed each of the aforementioned tasks.

4 ALTERNATIVE DESIGNS

The goal of this MQP was to design a wearable device for the elderly that detects an unbalanced situation before a fall occurs. The design team identified constraints and objectives by researching and understanding balance control, identified advantages and disadvantages of current fall detection and balance control technology, and interviewed stakeholders (Appendix C). Based on the constraints and objectives, the team developed three design alternatives. A needs analysis was performed and results of the analysis were used to identify necessary functions of the device and develop a conceptual design. Preliminary experiments were conducted to determine feasibility of the design. This section describes the process used and strategic decisions that we made in developing functions and specifications of a feasible conceptual design.

4.1 Preliminary designs

Through extensive literature review and patent search, the team identified advantages, disadvantages, and methods of signal detection of current fall and balance control technology. Advantages of current devices include: sense unbalanced situation before a fall occurs, device is comfortable and catered to the elderly population. Disadvantages of current technology include: detects a fall after it has occurred, not catered to the elderly population, requires professional intervention. In addition, the team found that current devices monitor pressure, acceleration, or tilt to detect imbalance or a fall. Based on these findings, the team developed the four design alternatives described in this section.

4.1.1 Preliminary Design 1: Shoe Insole

The first preliminary design (Figure 24) utilized an array of micro-strain gauges embedded in a shoe or shoe insole. The strain gages would detect a threshold pressure indicative of an

unbalanced situation. Upon detecting this threshold pressure, the insole would vibrate to notify the user that an off-balance situation had occurred.

Advantages of this design were that pressure is a common and accurate means of assessing balance control. In addition, a force platform would be needed to identify the threshold pressure and WPI has possession of this equipment. Therefore, no extra costs would be necessary for testing. However, the main disadvantage of this design was that the user would have to be wearing some type of footwear in order to use the device. This was unfavorable because our client interviews (Appendix C) revealed that elderly have trouble putting on and taking off footwear. Therefore, this design would make the device difficult to manage.

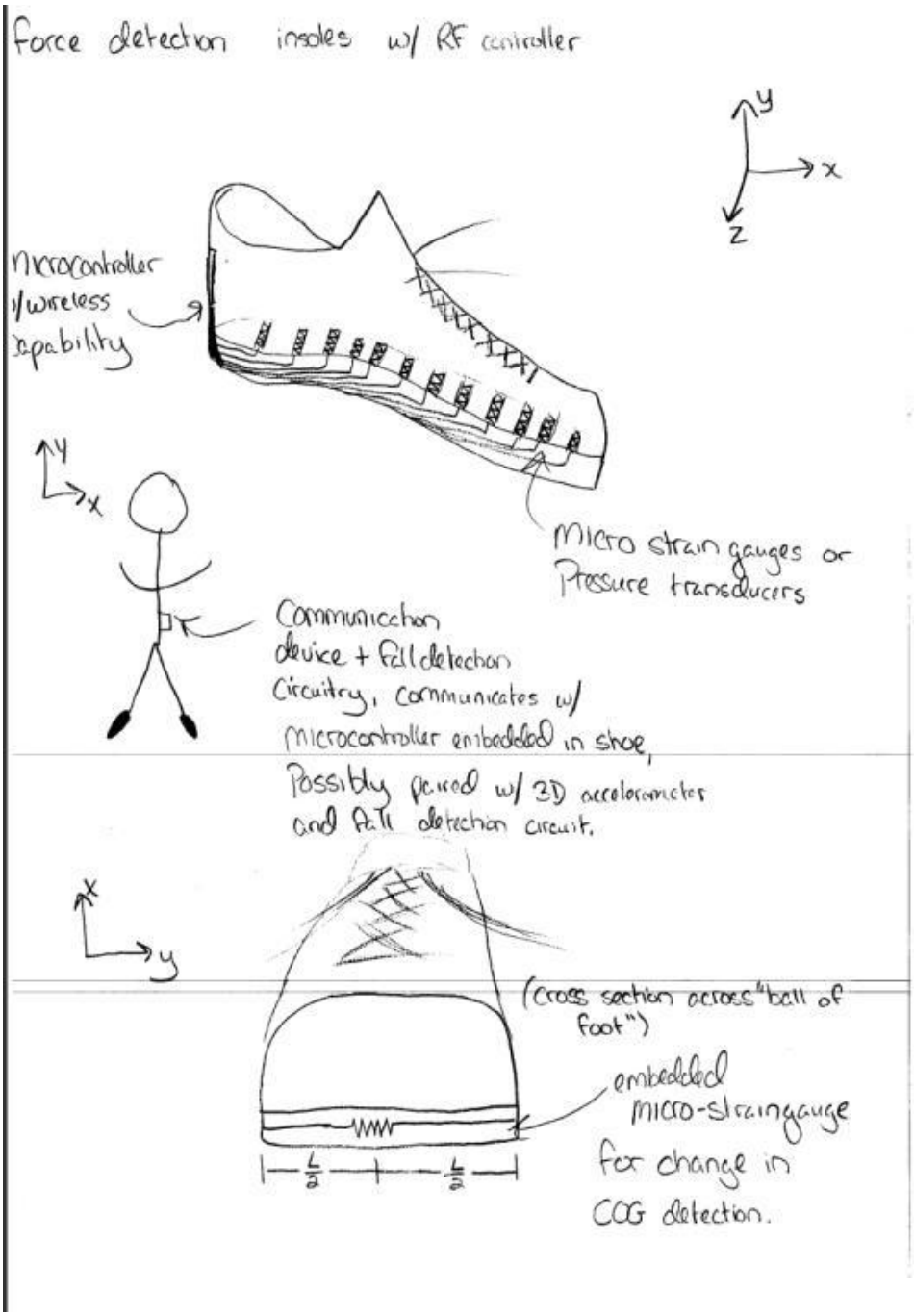


Figure 24: Preliminary design 1, Shoe Insole.

4.1.2 Preliminary Design 2: Waist-mounted Accelerometer

The second preliminary design (Figure 25) utilized a waist-mounted triple-axis accelerometer to monitor the acceleration of the user. The accelerometer would detect a threshold acceleration indicative of an unbalanced situation. Upon detecting this threshold acceleration, the device

would buzz to notify the user that an off-balance situation had occurred. The electrical components of the device would be housed in a plastic box. A clip on the back of the box would be used to attach the device to a belt or pants.

The advantages of this design were that the accelerometer has been shown to accurately monitor balance control, especially when positioned on a person's waist (Ward, Evenson, Vaughn, Rodgers, & Troiano, 2005). Therefore, the device could be placed in an accessible location, easy for the user to locate and attach the device. The location would also enable the user to wear it at any time of day, unlike the Preliminary Design 1 (shoe insole).

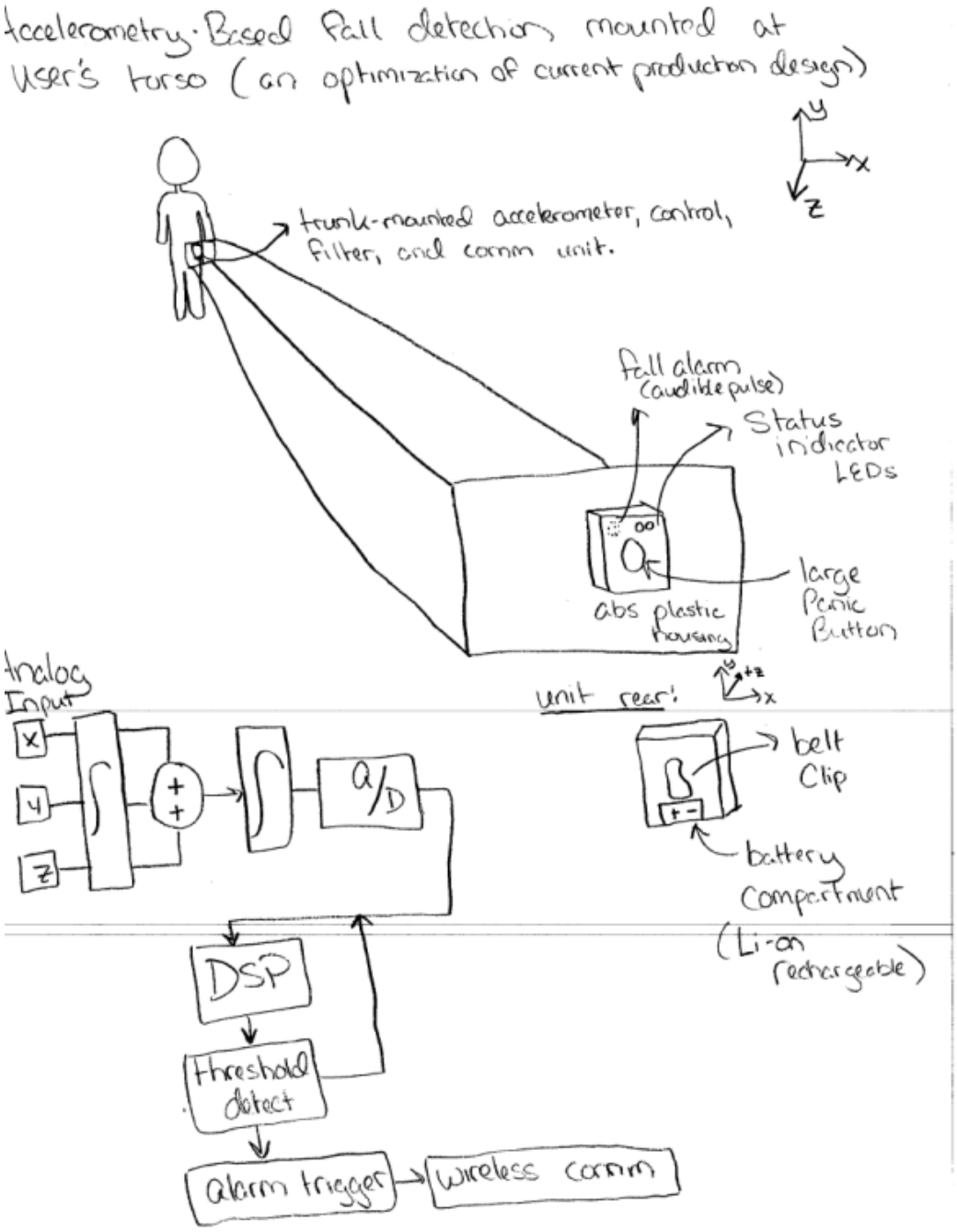


Figure 25: Preliminary Design 2, Waist-mounted accelerometer.

4.1.3 Preliminary Design 3: Ankle Brace Accelerometer/Gyroscope

The preliminary design shown in Figure 26 is based off the concept of vibration therapy described in the vibrotactile shoe insert patent mentioned previously (Harry, Collins, Prplata, & Kleshinkski, 2004) and studies that show changes in the angle and acceleration at the ankle joint help maintain

balance. The design was an ankle brace that contains an accelerometer or gyroscope positioned at the ankle joint of the brace and removable vibrating components located near tendons of the ankle such as the Achilles tendon within pockets of the brace. The accelerometer or gyroscope acts to monitor the acceleration or change of position of the ankle joint, respectively. A threshold acceleration or change in position defined through testing would occur before the individual was going to fall. Upon reaching this threshold, the sensor would activate the vibrating components in the brace. The vibrations would signal to the user that he/she was at risk of falling.

The ankle brace preliminary design also consisted of a removable strap that attaches via Velcro around the top of the ankle brace. The strap houses the battery of the device and wiring components. The wires would plug or snap into the accelerometer or gyroscope and vibrating components. This preliminary design was most closely related to the device described in the Patent Number US 2004/0173220 A1 consists of a wearable system (e.g. a shoe or sock) that contains actuators that create vibration feedback in the ankle or foot (Harry, Collins, Prplata, & Kleshinkski, 2004). The vibration feedback acts to increase the sensitivity of the mechanoreceptors in the foot or ankle and enhance the sensory function of those with decreased sensory performance (e.g. the elderly). Although the patented device could enhance balance control in an elderly individual, the device does not detect and notify the user when a fall is about to occur. Therefore, the preliminary design differs in that it would detect an unbalanced situation before the fall occurs.

The vibrations would also enhance the sensitivity of the mechanoreceptors in the ankle which would help convey information regarding the position of the ankle joint more quickly to the central nervous system. In turn, the vibrating components would increase the ability of the user to recover his or her balance.

An advantage of the ankle brace design shown in Figure 26 was that it could be worn consistently throughout the day. In addition, the location would have a limited interference with the user's everyday activities and could also remain out of the public view. However, it may difficult for

the user to attach the wires from the strap to the vibrating components and gyroscope or accelerometer. The elderly may have limited grip strength and also decreased vision which would make this task more difficult and hence make the device more difficult to put on and take off. In addition, assistive devices that aid the elderly in putting on socks already exist in the market indicating that elderly people have trouble putting on and taking off socks. Therefore, it may also be difficult for an elderly individual to put on the ankle brace. One advantage of this design was that the device directly notifies the user of an off-balance situation before a fall occurs. However, small changes of acceleration or position at the ankle joint may be difficult to detect, making the device inaccurate.

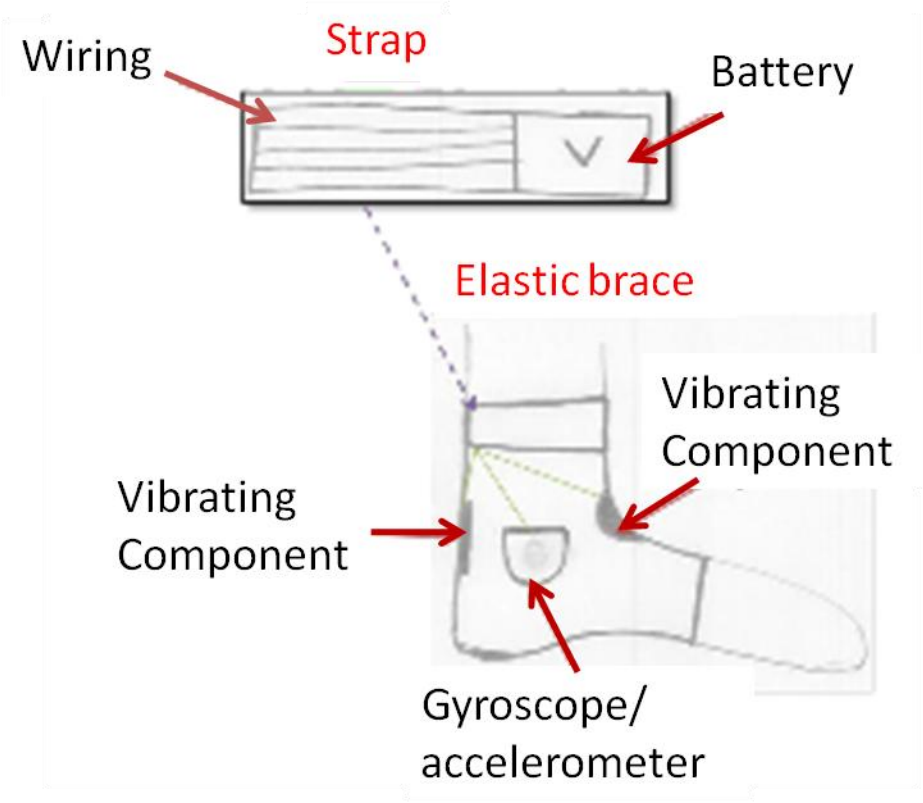


Figure 26: Preliminary Design 3, Ankle Brace.

4.1.4 Preliminary Design 4: V-Tact Belt

The V-Tact Belt design (Figure 27) combined the concept of a vibrotactile navigation device with the SHIMMER platform (accelerometer-based sensor) (Lorincz, Chen, Patel, & Welsh, 2008). The design incorporated small vibration tactors (similar to the ones used in a cell phone or pager) into the inside surface a belt. SHIMMER sensors with the gyroscope board connection will be placed inside MP3 player holders and attached to the outside surface of the belt. The device is worn at the waist because it is close to the user's COM and is attached using Velcro so the user can put it on and take it off easily. The SHIMMER sensors will be programmed and used to detect when the user is off balance and at risk of falling. When the SHIMMER sensors determine the user is off balance, a signal will be transmitted to the vibration tactors on the side which the user is unbalanced. The vibration tactors will then vibrate and alert the user that they are off balance in a particular direction, and the vibration should help them readjust their posture back to a balanced stance or gait. This device can be considered an early fall detection device because it detects and alerts the user when they are off balance and at risk of falling, and allows them to potentially correct their posture and balance before they actually fall. The device would also be able to detect that a fall has occurred.

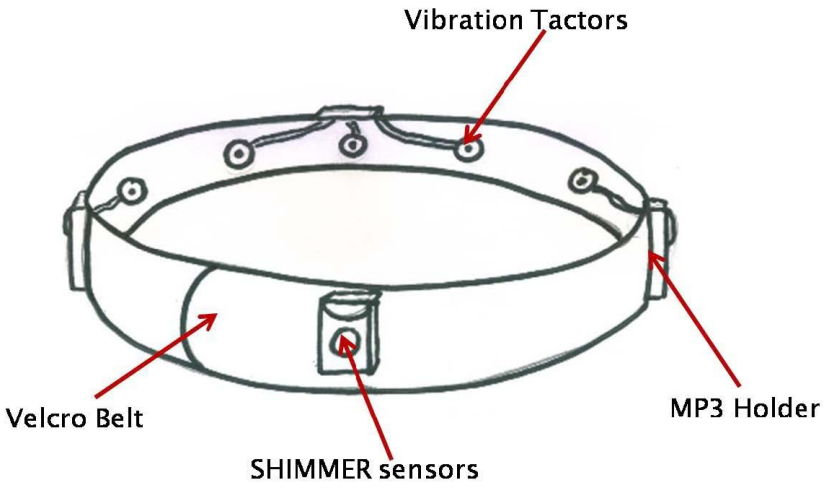


Figure 27: Preliminary Design 4, V-Tact Belt.

4.2 NEEDS ANALYSIS

In order to evaluate the design alternatives, the team first determined the requirements of the design. Designers discussed with all stakeholders the needs of the device, or the attributes it must have, and wants of the device, attributes that a stakeholder would like to have, but may not be possible given other constraints (Gielo-Perczak, 2009). Based on the needs and wants of the stakeholders, objectives were rank-ordered, necessary functions were determined, and specifications were outlined. All design alternatives were then assessed according to how well they met the client's wants and needs, and fulfilled the necessary functions of the device. This sections describes the specific tools used to rank-order the objectives and establish necessary functions and specifications of the device.

4.2.1 Rank-ordering Design Objectives

The team prioritized design objectives by rank-ordering them using pairwise comparison charts (PCC). A PCC is a tool that compares each objective against every one of the other objectives in order to rank order them according to their importance to the final design. In the following PCCs, a '1' means that the objective in the row was more important than the objective in the column. A '0' indicates that objective in the column was more important than the objective in the row. An 'x' was given when comparing an objective against itself. Total points that each objective received were summed in the last column of the table. The objective with the highest total was the most important. The first table (Table 3) evaluated objectives that contributed to the marketability of the design, while the second table (Table 4) evaluated objectives that contributed to the safety of the design.

Table 3: Pairwise Comparison Chart of Marketability Objectives

Marketability PCC	Durable	Aesthetically appealing	Comfortable	Easy to put on/take off	Easy to operate	Cater to elderly daily life	Cost-efficient	Manufacturable	Total
Durable	x	1	0	0	0	0	1	0	**2**
Aesthetically appealing	0	x	0	0	0	0	0	0	**0**
Comfortable	1	1	x	0	0	1	1	0	**4**
Easy to put on/take off	1	1	1	x	1	1	1	0	**6**
Easy to operate	1	1	1	0	x	1	1	0	**5**
Cater to elderly daily life	1	1	0	0	0	X	1	0	**3**
Cost-efficient	0	1	0	0	0	0	x	0	**1**
Manufacturable	1	1	1	1	1	1	1	X	**7**

According to the results of the marketability PCC, the order of importance of these objectives from most to least important was as follows:

1. Manufacturable
2. Easy to put on/take off
3. Easy to operate
4. Comfortable
5. Adaptable to elderly daily life
6. Durable
7. Cost-efficient
8. Aesthetically appealing

Table 4: Pairwise Comparison Chart of Safety Objectives

Safety PCC	Accurate	Sensitive to Balance Control	Minimal interaction with user's skin	Light-weight	Does not interfere with daily activity	Securely attached	No pressure points	Total
Accurate	X	0	1	1	1	1	0	**4**
Sensitive to balance control	0	X	1	1	1	1	0	**4**
Minimal interaction with user's skin	0	0	X	1	1	1	0	**3**
Light-weight	0	0	0	X	0	1	0	**1**
Securely attached	0	0	0	1	1	X	0	**2**
No pressure points	1	1	1	1	1	1	X	**6**

According to the results of the safety PCC, the order of importance of these objectives from most to least important was as follows:

1. No pressure points
2. Accurate
2. Sensitive to balance control
3. Minimal interaction with user's skin
4. Securely attached
5. Light-weight

The team then assigned relative weights to each objective according to their importance. For example, a highly ranked objective was assigned a higher weight indicating that it was more

important to the design. The objectives of each of the two PCCs were weighted as shown in Tables 5 and 6.

Table 5: Weighted Objectives-Marketability

Objective	Score	Adjusted Score	Weight
Manufacturable	7	7 +1 =8	8/36 = 0.22
Easy to put on/take off	6	6 +1 = 7	7/36= 0.19
Easy to operate	5	5 + 1 = 6	6/36= 0.17
Comfortable	4	4 +1 = 5	5/36= 0.14
Adapted to elderly daily life	3	3 +1=4	4/36= 0.11
Aesthetically appealing	0	0 + 1= 1	1/36= 0.03
Durable	2	2+1=3	3/36= 0.08
Cost-efficient	1	1 +1= 2	2/36= 0.06
TOTAL	28	36	1

Table 6: Weighted Objectives- Safety

Objective	Score	Weight
No pressure points	6	6/20 = 0.30
Minimal interaction with user's skin	3	3/20 = 0.15
Accurate	4	4/20 = 0.20
Sensitive to balance control	4	4/20 = 0.20
Lightweight	1	1/20 = 0.05
Securely attached	2	2/20 = 0.10
TOTAL	21	1

A numerical evaluation matrix was then used to assess how well each design alternative met the objectives. The numerical evaluation matrix contained objectives and constraints of the design in the first column. The second column contained the weighted percentage assigned to each objective. The next three columns contained the preliminary designs. Each design was then ranked on a scale from 0-1 in an increment of 0.1 on how well it met the objective (0- it does not met the objective and 1-it completely met the objective). Each team member ranked the design on how well it mets each objective. The rankings for each objective were averaged. A total percentage of how well the objective was met by the design was calculated by multiplying the ranking by the corresponding weighted percentage. Two numerical evaluation matrices were completed: one determined how well

the preliminary designs met the objectives of marketability (Table 7) and the other determined how well the alternative designs met the objectives of safety (Table 8). The Preliminary Design that met the highest percentage of the objectives is highlighted in red.

Table 7: Numerical Evaluation Matrix-Marketability

DESIGN **Constraints & Objectives**	**Weight (%)**	**Shoe Insole**	**Waist-mounted accelerometer**	**Ankle Brace**	**V-Tact**
Manufacturable	22%	0.3 6.6%	0.8 17.6%	0.2 4.4%	0.2 4.4%
Easy to put on/take off	19%	0.4 7.6%	0.8 15.2%	0.2 3.8%	0.2 3.8%
Easy to operate	17%	0.1 1.7%	0.8 13.6%	0.2 3.4%	0.2 3.4%
Comfortable	14%	0.4 5.6%	0.8 11.2%	0.3 4.2%	0.5 7.0%
Cater to elderly daily life	11%	0.6 6.6%	0.7 7.7%	0.4 4.4%	0.5 5.5%
Aesthetically appealing	3%	0.7 2.1%	0.5 1.5%	0.5 1.5%	0.5 1.5%
Durable	8%	0.3 2.4%	0.6 4.8%	0.4 3.2%	0.5 4.0%
Cost-efficient	6%	0.3 1.8%	0.8 4.8%	0.3 1.8%	0.4 2.4%
TOTALS	**100%**	**34.4%**	**76.4%**	**23.3%**	**32%**

Table 8: Numerical Evaluation Matrix-Safety

DESIGN Constraints & Objectives	Weight (%)	Shoe Insole	Waist-mounted accelerometer	Ankle Brace	V-Tact
No pressure points	29%	0.8 23.2%	0.5 14.5%	0.2 5.8%	0.5 14.5%
Minimal interaction with user's skin	14%	0.8 11.2%	0.8 11.2%	0.1 1.4%	0.8 11.2%
Accurate	19%	0.5 9.5%	0.8 15.2%	0.5 9.5%	0.5 9.5%
Sensitive to balance control	19%	0.7 13.3%	0.8 15.2%	0.1 1.9%	0.8 15.2%
Lightweight	5%	0.6 3.0%	0.5 2.5%	0.5 2.5%	0.3 1.5%
Does not interfere with daily activity	5%	0.5 2.5%	0.5 2.45%	0.6 3.0%	0.4 2.0%
Securely attached	9%	0.6 5.4%	0.5 4.5%	0.6 5.4%	0.6 5.4%
TOTALS	**100%**	**65.6%**	**65.6%**	**26.5%**	**48.1%**

Through interviews with Lauren Roberts (Appendix C), a physical therapist of Fairlawn Rehabilitation Hospital, the design team established that marketability and safety were equally important to the design. Therefore, an additional Numerical Evaluation Matrix (Table 9) was created to determine mathematically which alternative design best met the objectives of the design as a whole.

Table 9: Numerical Evaluation Matrix- Overall Objectives

DESIGN Overall Objectives	Weight	Shoe Insole	Waist-mounted accelerometer	Ankle Brace	V-Tact
Safety	0.50	65.6 % 32.2%	65.6% 32.8%	26.5% 13.3%	48.1% 24.1%
Marketability	0.50	34.4% 17.2%	76.4% 38.2%	23.3% 11.7%	32% 16%
TOTALS	**1**	**49.4%**	**71%**	**24.9%**	**50.1%**

According to the Numerical Evaluation Matrices, the waist-mounted belt best satisfied the objectives, as it met 65.55% of the safety objectives, and 76.4% of the marketability objectives, and thus fulfilled 71% of the overall objectives of the design. Therefore, the team chose the waist-mounted conceptual design and determined the functions, specifications, and feasibility of the design.

4.3 FUNCTIONS AND SPECIFICATIONS

The two main goals of the design were to strengthen the user's awareness of their balance control condition and to proactively force the user to perform a rehabilitation exercise. In order to verify that the waist-mounted accelerometer preliminary design would be capable of fulfilling the goals of the project, the team established three necessary functions and used research and client interviews to create specifications that would enable the device to achieve each function. The team also identified constraints that could prevent the design from achieving the goals.

To strengthen the user's awareness of their balance conditions, our device should notify the user if they experience an unbalanced situation. Specifically, our device should detect and notify the user of an unbalanced situation before a fall occurs. In order to proactively force user rehabilitation, operation of the device should require the user to perform a rehabilitation activity. Specifically, the rehabilitation activity should be a daily activity so that operation of the device is integrated into the daily life of the user. Constraints of the design were the varying disabilities of the elderly such as osteoporosis, decreased muscle strength, and decreased range of motion. The device needed to remain easy to use and to operate regardless of disabilities due to normal aging. Two other constraints were the available testing equipment at WPI and affordability. The team only had access to a force platform and the cost of the device should remain equivalent or less than the fall detection technology on the market so that the device can be afforded without the

aid of health insurance. Figure 28 outlines the goals (red), functions (purple), specifications (green), and constraints (grey).

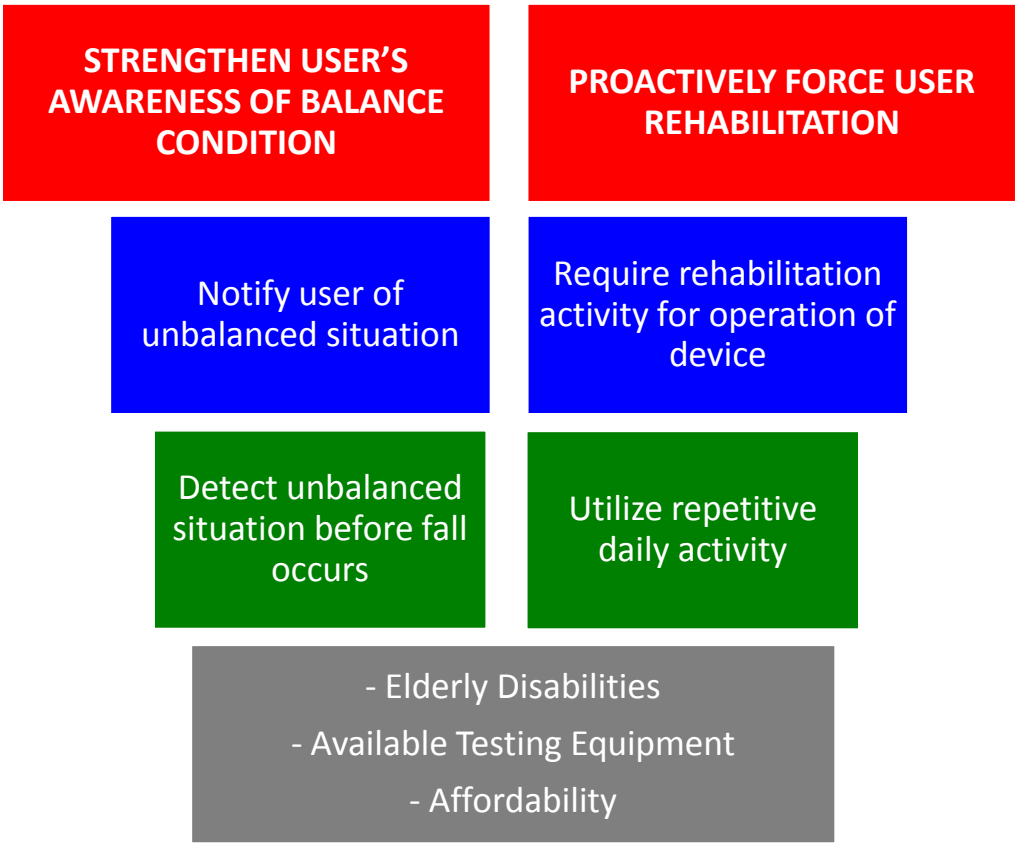


Figure 28: Outline of goals (red), functions (blue), specifications (green), and constraints (grey) of the design.

To determine if the waist-mounted accelerometer met the necessary functions and specifications (Figure 28), the team posed four questions:

1. What activity should be monitored?
2. Where should the device be located?
3. What signal and sensor should be used?
4. How should the signal be analyzed?

The following sections describe the team's process of answering these questions to further identify specifications of the design. These sections also verify that the waist-mounted

accelerometer would be able to fulfill the specifications making it the best choice for the final design.

4.3.1 Choosing the Activity

The two goals of the device were to proactively force user rehabilitation and strengthen the user's awareness of their balance condition. In order to do this, operation of the device should require the user to perform a rehabilitation activity. Specifically, the rehabilitation activity should be a daily activity so that operation of the device is integrated into the daily life of the user. To strengthen awareness, the device should detect and notify the user of an unbalanced situation before a fall occurs. Therefore, the activity should be a daily activity that is repetitive in the life of the elderly, sensitive to balance control, and a feasible rehabilitation technique as shown in Figure 29 (Gross, Stevenson, Charette, Pyka, & Marcus, 1998).

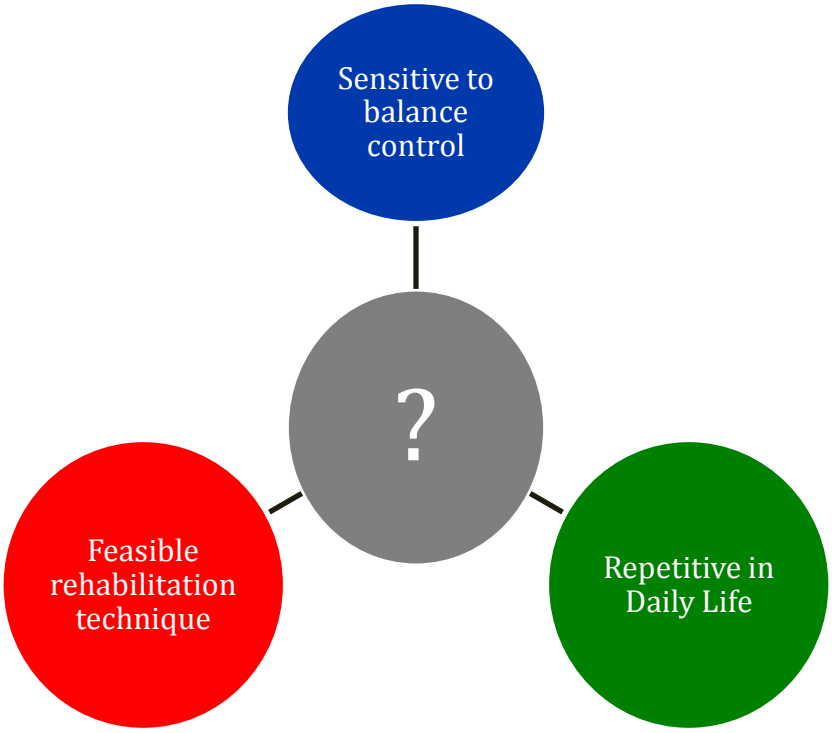
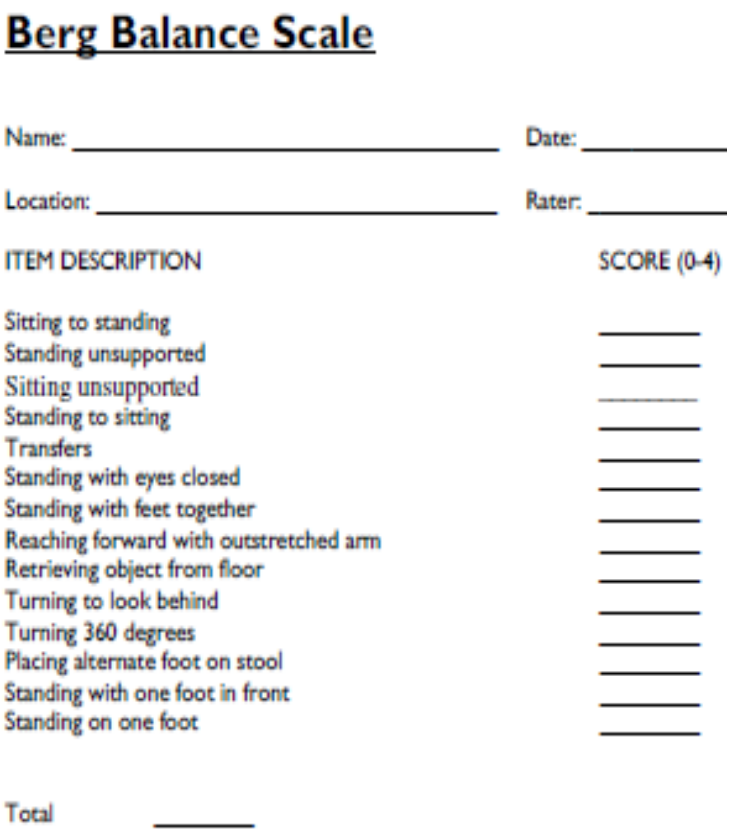


Figure 29: Factors needed in choosing an activity.

To ensure that the chosen activity would be sensitive to balance control, the team looked at the fourteen activities on the Berg Balance Test (BBT) shown in Figure 30. The team identified through research and an interview with a physical therapist at Fairlawn Rehabilitation Hospital that the BBT was the most common exam used to assess balance control and that three activities on the BBT were the most sensitive to balance control: standing on one foot, standing in tandem (one foot in front of the other), and the sit-to-stand (STS) (Lauren Roberts, Fairlawn Rehabilitation Hospital).



Berg Balance Scale

Name: _____ Date: _____

Location: _____ Rater: _____

ITEM DESCRIPTION	SCORE (0-4)
Sitting to standing	_____
Standing unsupported	_____
Sitting unsupported	_____
Standing to sitting	_____
Transfers	_____
Standing with eyes closed	_____
Standing with feet together	_____
Reaching forward with outstretched arm	_____
Retrieving object from floor	_____
Turning to look behind	_____
Turning 360 degrees	_____
Placing alternate foot on stool	_____
Standing with one foot in front	_____
Standing on one foot	_____

Total _____

Figure 30: Berg Balance Test (American Academy of Health and Fitness, 2010).

Standing on one foot and standing in tandem were eliminated because they can be dangerous for the elderly user and are also not repetitive daily activities. STS was chosen as the activity because it was sensitive to balance control (Gross, Stevenson, Charette, Pyka, & Marcus, 1998), and repetitive in daily life (Figure 31).

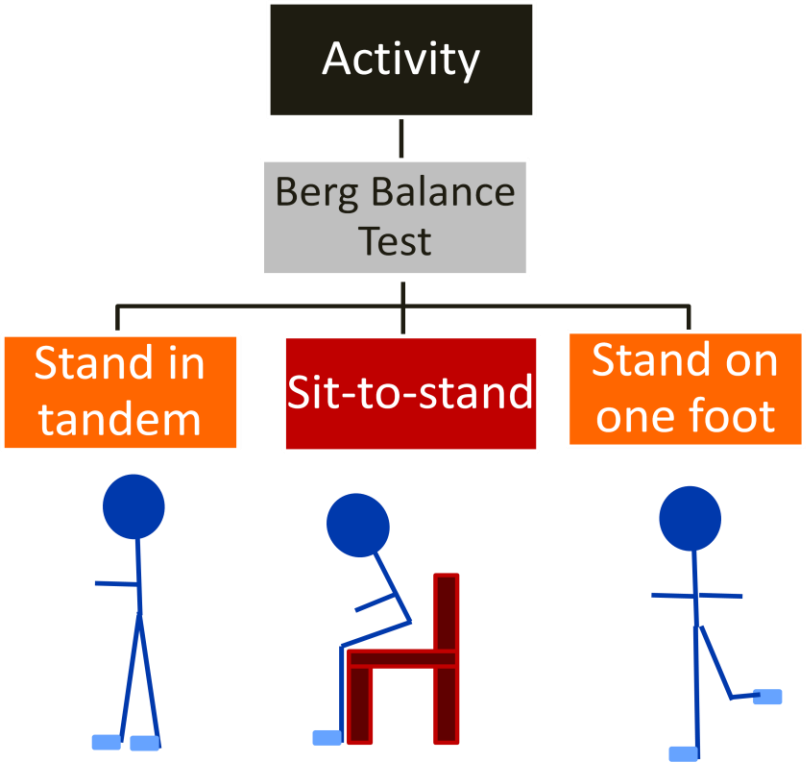


Figure 31: Choosing the Device Activity.

4.3.2 Choosing the Signal and Sensor

The two goals of the device were to proactively force user rehabilitation and strengthen the user's awareness of their balance condition. In order to force rehabilitation, the device required the user to perform the STS activity, which is both a daily activity and a rehabilitation technique. To strengthen awareness, the device detected and notified the user of an unbalanced situation before a fall. To do this, the team had to identify an appropriate signal that was sensitive to balance control and maintains high accuracy. The sensor that monitors this signal must be positioned in an accessible location to maintain ease of use of the device (Figure 32).

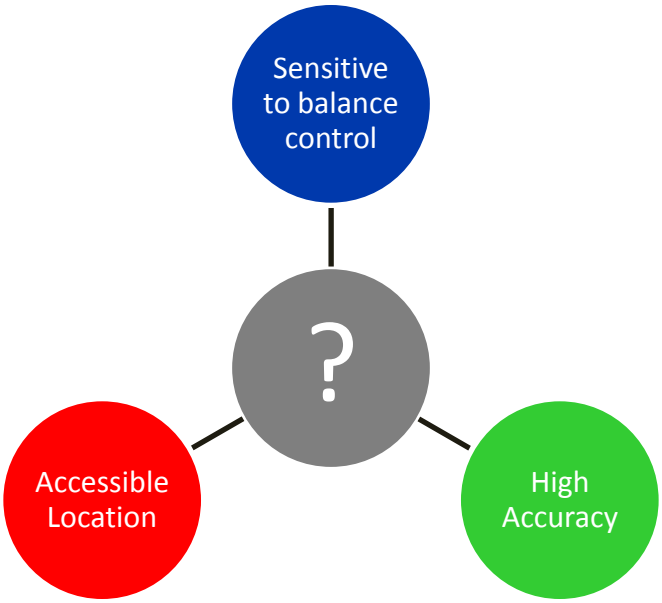


Figure 32: Factors signal and sensor need to fulfill.

The team identified three signals that were currently used to monitor balance control: pressure under the foot, acceleration of the body, and tilt of the upper body at the hip as shown in Figure 33 (Ward, Evenson, Vaughn, Rodgers, & Troiano, 2005). Tilt was eliminated because it was not sensitive to balance control (Mathie, Coster, & Lovel, 2004). Through research, the team identified that acceleration of COM was a key factor in determining a person's ability to stay balanced (Pai & Patton, 1997). In addition, the team found that pressure under the foot can indicate where a person's COM is located. However, research showed that location of COM along the BOS was not the factor that determines a person's balance condition, but the acceleration of a person's mass can dictate if a person is balanced (Pai & Patton, 1997). In addition, acceleration was shown to be the most accurate measure of balance control during the STS activity and is often used in research to assess balance control (Winter D. , 1995), (Gross, Stevenson, Charette, Pyka, & Marcus, 1998). Acceleration is most accurately monitored on the trunk, which would be an accessible location (Ward, Evenson, Vaughn, Rodgers, & Troiano,

2005). On the other hand, pressure under the foot would require the device to be located at the feet. It has been shown that elderly have trouble putting on shoes and socks, so a device similarly worn on the foot could be difficult for the elderly to use (Dunne, Bergman, Rogers, & Rivara, 1993).

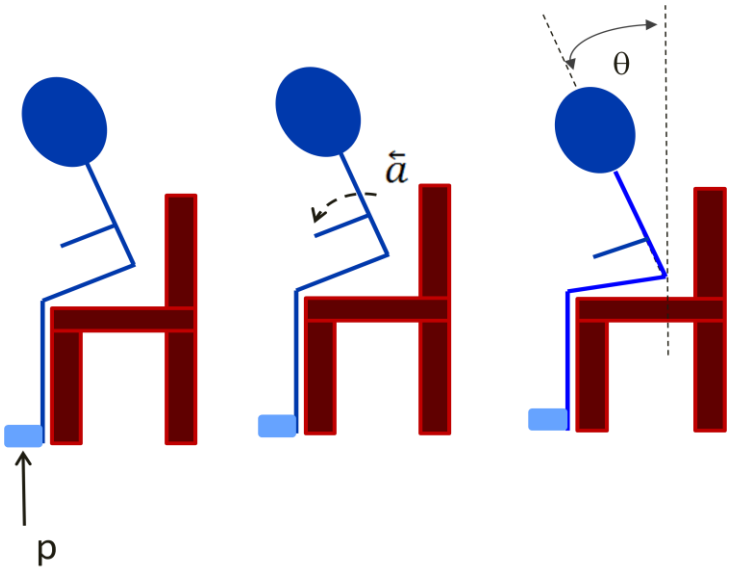


Figure 33: Signals to monitor balance control (pressure, acceleration, tilt).

Therefore, as shown in Figure 34 acceleration was chosen as the signal to monitor during the STS and a triple-axis accelerometer was used as the sensor because it has high accuracy in monitoring balance control in accessible locations (Gross, Stevenson, Charette, Pyka, & Marcus, 1998).

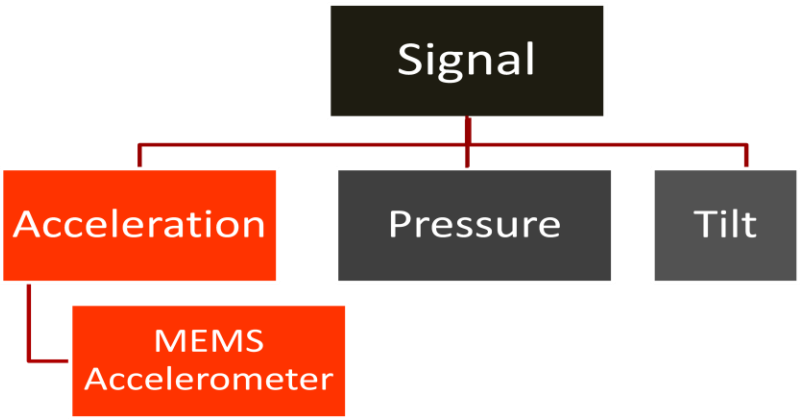


Figure 34: Choosing the Device Signal and Sensor.

4.3.3 Choosing the Location

The two goals of the project were to proactively force user rehabilitation and strengthen the user's awareness of their balance condition. In order to force rehabilitation, the device required the user to perform the STS activity. To strengthen awareness, the device monitored acceleration and detected and notified the user of an unbalanced situation (abnormal acceleration) before a fall. Therefore, the team identified the location to monitor acceleration during the STS that is the most accurate and sensitive to balance control, and maintains comfort and ease of use of the device (Figure 35).

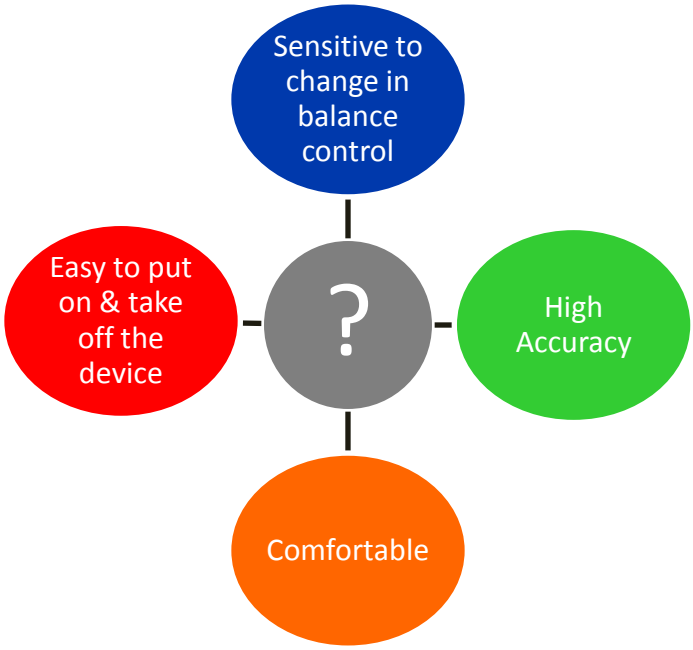


Figure 35: Factors location of device needs to fulfill.

The team identified that the trunk was the most sensitive location for monitoring acceleration of the body (Gross, Stevenson, Charette, Pyka, & Marcus, 1998). Specifically, the sternum, hip, and lower back, shown in Figure 36, were the three locations most sensitive to monitoring balance control because they are closest to the body's COM. Since the device has to be comfortable and easy for the user to put on and take off, the team eliminated the sternum location because a device placed here would require the user to lift their arms. In addition, the team found through research that the most sensitive location on the trunk for monitoring balance control was the hip (Ward, Evenson, Vaughn, Rodgers, & Troiano, 2005). The lower back was a difficult place to reach and locate. While sitting, the device would also be more prone to being bumped if placed on the lower back.

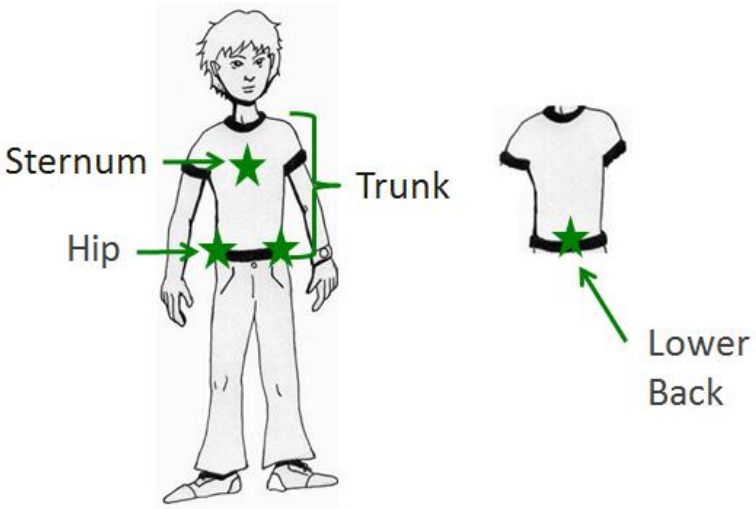


Figure 36: Possible locations of device (http://www.ehow.com/how_4629008_draw-person-standing.html).

Therefore, the team eliminated lower back and chose to monitor acceleration during the STS at the hip as shown in Figure 37. The hip bone is a universal anatomical marker making the device user-friendly and since the hip is a reliable location for measuring acceleration, it ensures high accuracy.

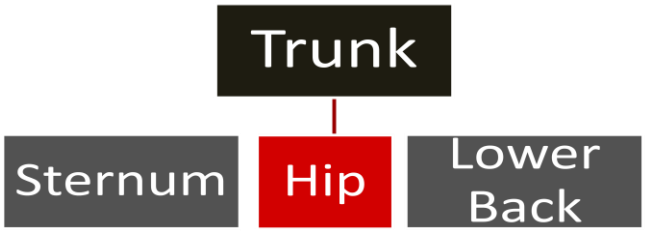


Figure 37: Choosing the Device Location.

4.3.4 Choosing the Method of Analyzing Acceleration Data

The two goals of the project were to proactively force user rehabilitation and strengthen the user's awareness of their balance condition. In order to force rehabilitation, the device required the user to perform the STS activity. To strengthen awareness, the device monitored acceleration and detected and notified the user of an unbalanced situation (abnormal acceleration) before a

fall occurred. Therefore, the team identified the most accurate, repeatable method of analyzing acceleration of the COM that clearly illustrated the difference between a balanced and unbalanced STS (Figure 38).

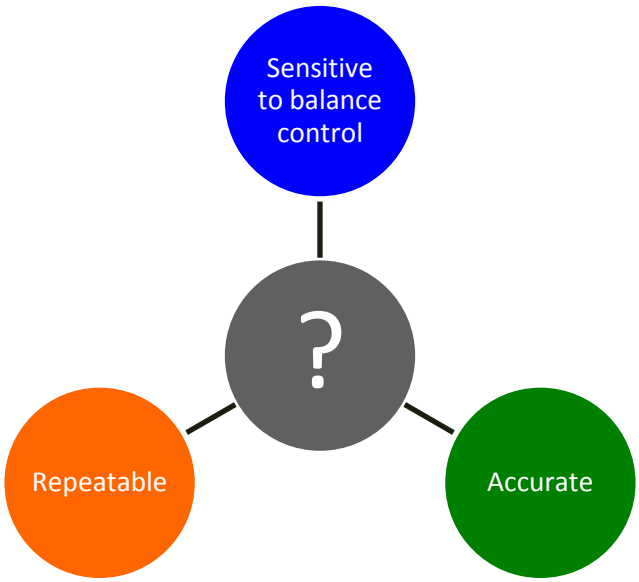


Figure 38: Factors method of analysis needs to fulfill.

Through research the team found that the signal of the COM was directly related to its acceleration in the A/P and M/L directions (Winter D. , 1995). In particular the magnitude and frequency of this signal and reaction time of the subject were the most important in classifying the balance control (Winter D. , 1995). The team identified 5 potential methods for analyzing frequency, magnitude, and time duration of acceleration: Fast Fourier Transform (FFT), acceleration on the X-axis, acceleration on the Y-axis, acceleration on the Z-axis, and route sum of squares of the X-Y-Z axis. FFT is used to enhance or remove periodic noise in a signal and yields the power of the signal as a function of the frequency. Although FFT is an accurate means of assessing balance control, research showed that analysis of a short record can result in erroneously high means and median frequencies (Winter D. , 1995). Since the STS activity is short, only lasting approximately 3 seconds, we eliminated the FFT method of analysis.

Research also shows that the signal of our COM is directly related to its acceleration in both the A/P and M/L directions. Therefore, we eliminated the methods of analyzing acceleration on the X, Y, or Z-axis independently and decided to use the root sum of squares method in order to analyze acceleration of the COM in all planes of movement.

Using the root sum of squares method of analysis, the team could look at one of two parameters of the graph: amplitudes or time variations. Research showed that as acceleration of the COM increased, a person became closer to their "stability limit", meaning they became more likely to lose their balance and fall (Pai & Patton, 1997). Studies also showed that the duration of the STS differs depending on the speed of the movement. According to this research, if a person accelerated quickly during the STS they became more unbalanced and the duration of the STS got shorter. On the other hand, if the person accelerated slowly during the STS they can maintain balance more easily and the duration of the STS got longer. Therefore, the team decided to analyze the time duration of the STS activity (Figure 39).

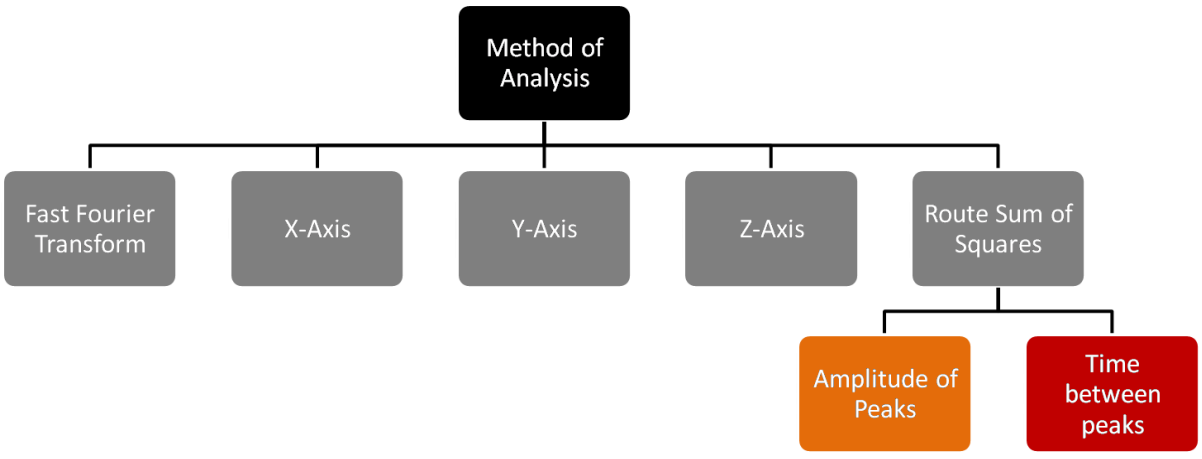


Figure 39: Choosing the Method of Analysis.

4.3.5 Final Solution

The two goals of the device were to proactively force user rehabilitation and strengthen the user's awareness of their balance condition. In order to do this, operation of the device requires the user to perform the STS rehabilitation activity which is a repetitive daily activity and most sensitive to balance control. To strengthen awareness, the device is worn on the right hip and monitors acceleration in all planes of movement during the STS activity. Finally, the device detects an abnormal acceleration indicative of an unbalanced situation and instantly alerts the user of the imbalance. Therefore, the team answered the four questions as follows:

1. What activity should be monitored? The STS rehabilitation and daily activity
2. Where should the device be located? At the right hip bone
3. What signal and sensor should be used? Acceleration will be monitored with a tri-axial accelerometer.
4. How should the signal be analyzed? Analyze time duration of STS looking at the magnitude of the x, y, and z axis of acceleration

Since the most accurate and sensitive means to monitor balance control during STS was by acceleration at the hip, the hip-mounted accelerometer (Preliminary Design 2) was shown to be a feasible conceptual design. The team hypothesized that the time duration of the STS would be longer when a person was balanced and shorter during an unbalanced STS. To verify the accuracy and repeatability of this method, the team performed preliminary testing which is discussed in the following section.

4.4 PRELIMINARY EXPERIMENTS

The purpose of the device was to detect an unbalanced STS in terms of acceleration and notify the user when an unbalanced STS occurs. In order for the device to do this, the team needs to

develop an algorithm or a set of well-defined instructions to completing a task. In this project, the task was to notify the user of unbalanced situation. Therefore, the team needed to conduct preliminary testing to identify a significant difference between the acceleration patterns of a balanced and unbalance STS. In particular, the team needed to verify that the time duration of the STS would be longer when a person was balanced and shorter during an unbalanced STS. This sections details the methods used to verify the hypothesis, results, and conclusions drawn from preliminary testing.

4.4.1 Materials and method

Preliminary tests were conducted to identify significant differences between the acceleration patterns of a balanced STS situation and unbalanced STS situation. Nine healthy subjects participated in preliminary testing. One subject (male, age 11) was eliminated from testing because he did not perform the STS properly. Subject information can be found in Table 10. The SparkFun KinetaMap (SparkFun Electronics, 2009) data logger (Figure 40) containing an ADXL345 tri-axial accelerometer (Sparkfun Electronics, 2009) was used for collecting acceleration data. The AMTI AccuSway force platform and AMTI NetForce and BioAnalysis software were used to record and analyze balance control data, respectively. Acceleration data were collected from each subject. Force platform data were not collected from subjects 5, 6, 7, 8 because these tests were performed in a home setting and the force platform was not available in this location. Force platform data were collected from all other subjects.

Figure 40: KinetaMap Triple-Axis Accelerometer and Data Logger (SparkFun Electronics, 2009).

Table 10: Subject information

	1	2	3	4	5	6	7	8
Gender	F	F	F	M	F	M	M	F
Age	21	20	21	21	53	23	63	50
Height	5’ 8”	5’ 5”	5’ 7”	5’ 1”	5’2”	5’ 11”	5’ 10”	5’ 1”
Mass (kg)	67	70	63	63	67	82	75	59

The force platform and a regular chair were set up as shown in Figure 41. The force platform was placed directly on the ground. A wooden platform, the same height as the force platform was placed adjacent to the force platform. The chair was placed on top of the wooden platform and in front of the force platform so that when the subject sat in the chair her feet rested comfortably on the force platform. The KinetaMap was attached with Velcro to an adjustable belt. The belt was positioned so that the KinetaMap was mounted on the right side of the subject, externally adjacent to the iliac crest (Figure 42).

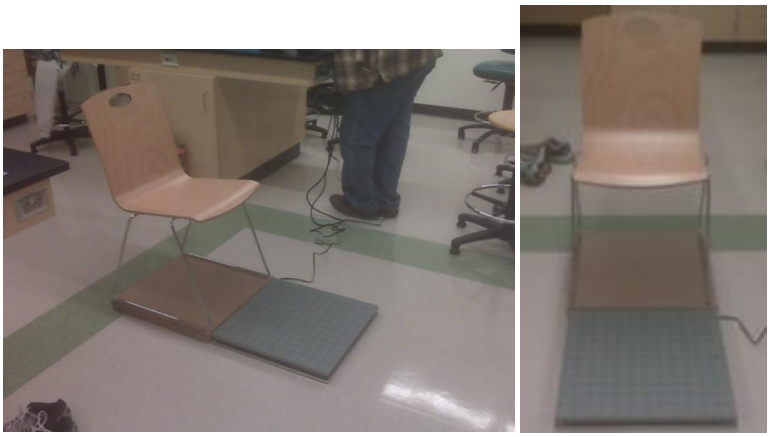


Figure 41: Chair and force platform experimental set-up.

Figure 42: Attachment of the KinetaMap device.

The subject first stood still with feet shoulder-width apart on the force platform. Their weight was collected and saved in NetForce software. The subject then sat in the chair and NetForce data collection was started. The subject then turned on the KinetaMap, and waited for the KinetaMap's LED to start blinking blue, signifying that the accelerometer had started collecting data. Ten seconds after the LED started blinking blue, the subject stood up from the seated position. Ten seconds after the subject reached a balanced standing position, the subject turned the KinetaMap off to stop data collection. This was repeated 10 times with the subject's feet positioned shoulder-width apart while performing the STS (Figure 43) and 10 times with the subject's feet positioned in tandem (Figure 44). Subjects who conducted preliminary testing in the home setting performed a minimum of 5 tandem and shoulder-width STS trials due to time

constraints of the subjects' schedules. After each STS trial, the subject rated their comfort on a scale from 1-5 (1 being completely unbalanced, 5 being completely balanced).

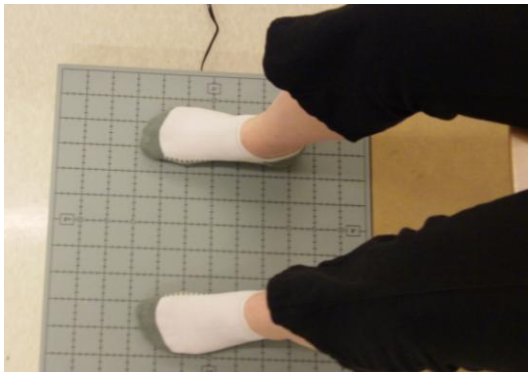


Figure 43: Shoulder-width foot position representing a balanced condition.

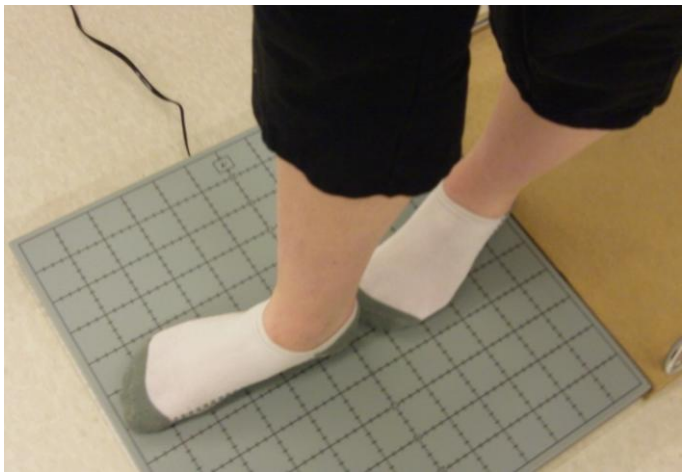


Figure 44: Tandem foot position representing an unbalanced condition.

The KinetaMap collected data at 20Hz and logged each trial in a Microsoft Excel document in terms of time and the X, Y, and Z components of acceleration. The X, Y, and Z components of acceleration were converted into m/s^2 using the ADXL345 Tri-Axial accelerometer data sheet (Appendix A). The raw data were multiplied by 18mg/digit and divided by 9.8 m/s^2 (See Appendix J). These data were used to plot the X, Y, and Z components of acceleration and magnitude of the acceleration in Microsoft Excel. An offset of about 9.8 m/s^2 was observed in each plot due to gravity. Therefore, the average of the first 5 seconds of

acceleration data were subtracted from the entire data set in order to zero each plot. The team then quantitatively compared the X, Y, Z, and root-sum-squares plots of the unbalanced and balanced trials to identify differences. The NetForce data files were imported into BioAnalysis software, and plots of COP were obtained. The COP data for the tandem trials were quantitatively compared to the COP of the shoulder-width trials to determine if the subjects were more off balance and quantitatively represent their balance control. The COP data were also compared to the subject's comfort level.

4.4.2 Results of Preliminary Testing

Results of preliminary testing showed that the STS activity produces an acceleration curve as shown in Figure 45 and contained a positive and negative amplitude of acceleration. The positive amplitude of acceleration corresponded to when the subject flexed the hips to sway forward during the STS activity. The negative amplitude of acceleration corresponded to when the subject extended the lower limbs to sway backward during the STS activity. The peaks of these two amplitudes (red and blue squares) represented the forward sway and backward sway of the STS, respectively.

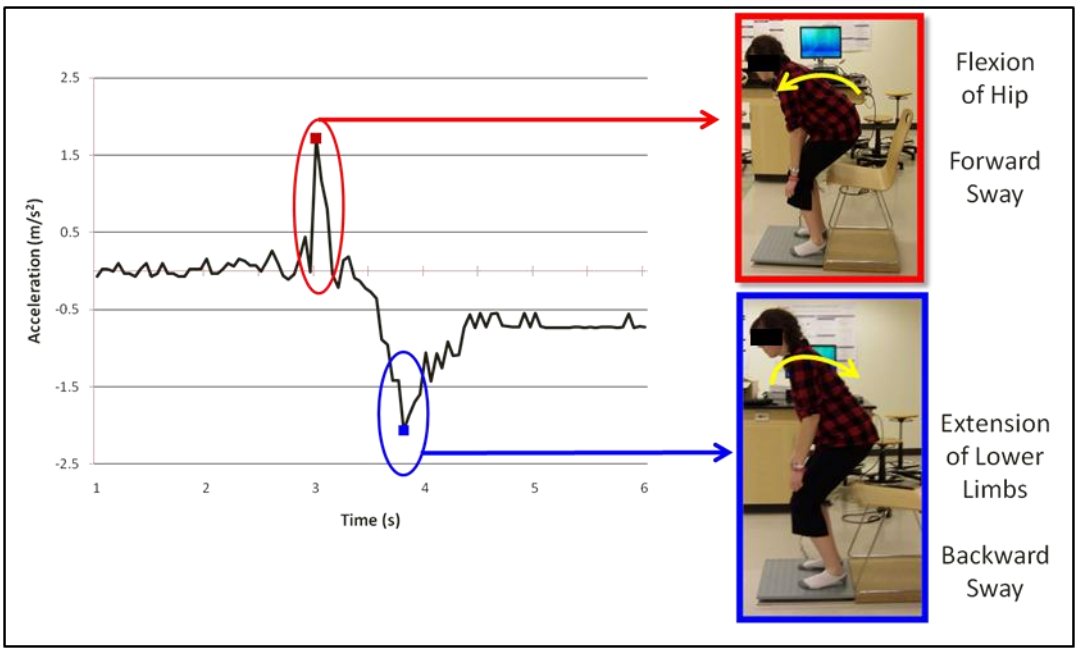


Figure 45: Root Sum of Squares of Acceleration of the STS in relation to hip flexion and extension of lower limbs.

Through quantitative assessment, the team observed that the time between the two peaks was longer in the shoulder-width trials than in the tandem trials as shown by Figure 46. This was consistent with research that showed if a person accelerated quickly during the STS they became more unbalanced and the duration of the STS got shorter. On the other hand, if the person accelerated slowly during the STS they can maintain balance more easily and the duration of the STS got longer. In addition, the team observed that each plot had baseline noise and that in some occasions the final baseline did not equal zero. This is due to the change in tilt of the device from the start to the end of the activity.

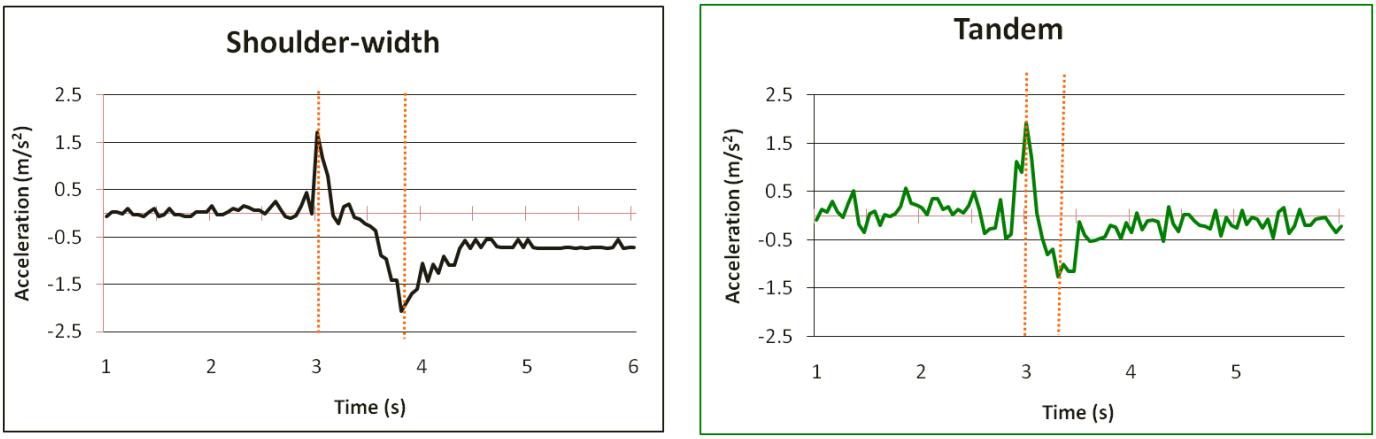


Figure 46: Example of shoulder-width (black) and tandem (green) acceleration curves showing the time duration measured (between organge lines).

The COP plots of the tandem and shoulder-width trials verified that the shoulder-width trials were balanced and tandem trials were unbalanced (Figure 47). The unbalanced tandem trials showed a large variation of the COP (green), which showed that the subject's COM moved across a large area. On the other hand, the balanced shoulder-width trails showed less variation as seen by the more compact black circle. This showed that the subject's COM did not move or sway across a larger area and the subject maintained their balance.

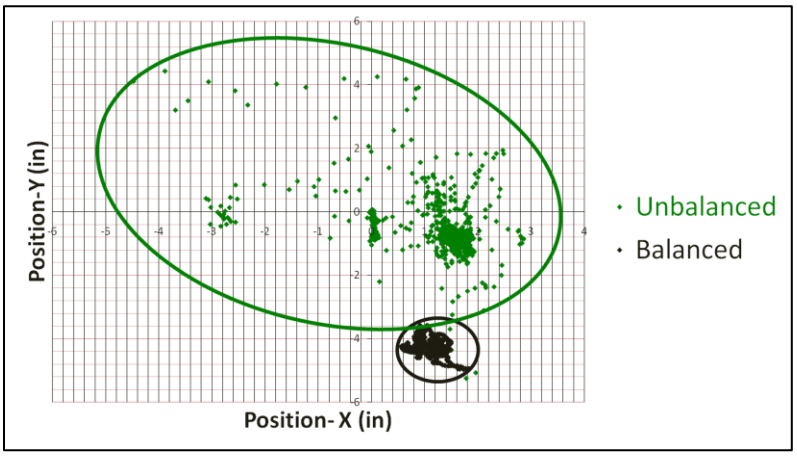


Figure 47: COP plots of tandem (green) and shoulder-width (black) trials.

The average times between the positive and negative peaks are shown for each subject in the bar graph Figure 48 and data are listed in Tables 11 and 12. The average time of shoulder-

width trials was greater than in tandem trials for all subjects tested and was significantly different in subjects who performed 10 shoulder-width and 10 tandem trials. *P*-values are listed in Table 13. The average comfort ratings for each subject (Tables 11 and 12) were greater for shoulder-width trials than tandem trials.

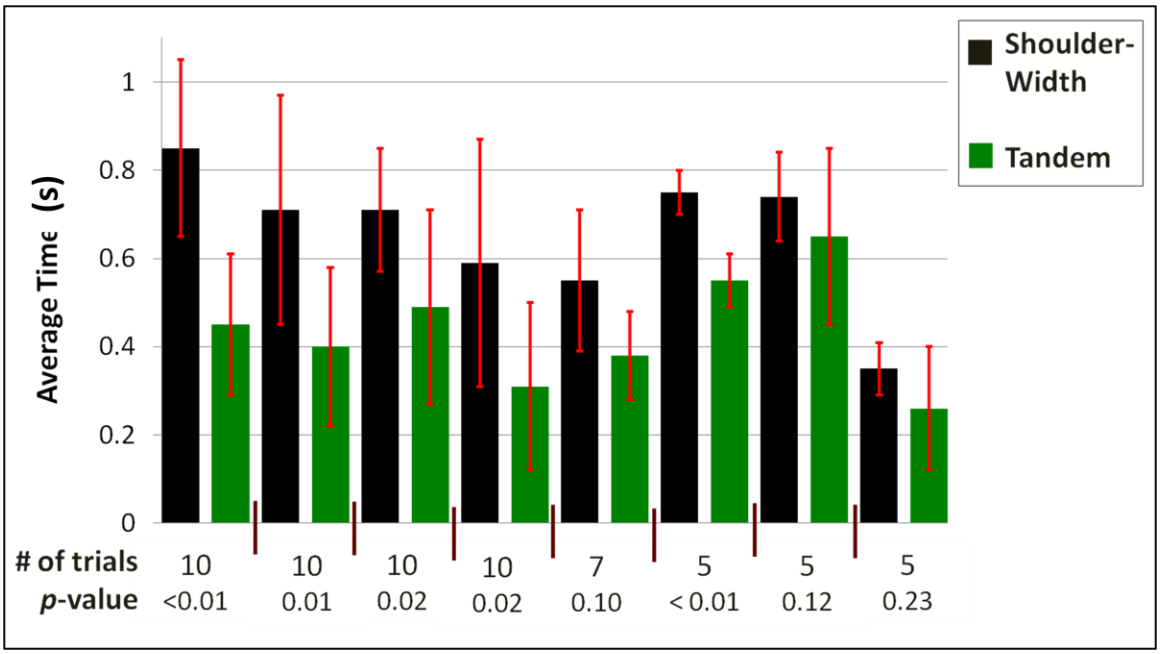


Figure 48: Average time between peaks (R→L: 1, 2, 3, 4, 5, 6, 7, 8,).

Table 11: Results of shoulder-width trials

Subject	# of Trials	Average Comfort Level (1-5)	Average Time (s)	Standard Deviation	Coefficient of Variance	*p*-value
1	35	5	0.85	± 0.20	0.23	< 0.01
2	10	5	0.71	± 0.26	0.37	0.01
3	10	5	0.71	±0.14	0.19	0.02
4	10	5	0.59	±0.28	0.48	0.02
5	7	5	0.55	±0.16	0.29	0.10
6	5	5	0.75	±0.05	0.07	< 0.01
7	5	4.8	0.74	±0.10	0.23	0.12
8	5	5	0.35	±0.06	0.17	0.23

Table 12: Results of tandem trials

Subject	# of Trials	Average Comfort Level (1-5)	Average Time (s)	Standard Deviation	Coefficient of Variance	*p*-value
1	11	3.4	0.45	±0.16	0.35	< 0.01
2	10	4	0.40	±0.18	0.45	0.01
3	10	3.8	0.49	±0.22	0.44	0.02
4	10	3.8	0.31	±0.19	0.60	0.02
5	7	3.5	0.41	±0.18	0.44	0.10
6	5	3	0.55	±0.10	0.18	< 0.01
7	5	3.6	0.65	±0.06	0.09	0.12
8	5	3.8	0.26	±0.14	0.55	0.23

Table 13: *p*-values comparing average shoulder-width to average tandem time between peaks

Subject	# of Trials	*p*-value
1	11	< 0.01
2	10	0.01
3	10	0.02
4	10	0.02
5	7	0.10
6	5	< 0.01
7	5	0.12
8	5	0.23

4.4.3 Conclusions of Preliminary Results

Based on preliminary results the team verified that the time between the positive and negative peaks of the STS acceleration curve was longer in shoulder-width trials than in tandem trials. Through client feedback and COP data, the team also showed that tandem trials represented an unbalanced situation and shoulder-width trials represented an unbalanced situation. Since the time between peaks was significantly different ($p<0.05$) for all subjects who

performed 10 tandem and 10 shoulder-width STS's, the team concluded that a specific time range corresponds to a balanced STS and a specific time range corresponds to an unbalanced STS. The preliminary data showed similar results between subjects of the same gender and age, however not enough data were collected to show that these similarities were significant. Therefore, the time ranges were specific to the individual and the team designed specifications according to the data of one particular subject. The following section discusses how the team conceptualized specifications and made decisions regarding the final design.

4.5 Conceptual design

The purpose of the device is to detect an unbalanced STS in terms of acceleration and notify the user when an unbalanced STS occurs. Based on preliminary results, the time between the positive and negative peaks of these amplitudes was significantly longer in shoulder-width trials than in tandem trials. Therefore, the team concluded that a specific time range corresponded to a balanced STS and a specific time range corresponded to an unbalanced STS. These time ranges were specific to the individual. Based on this finding and necessary functions of the device, the team brainstormed the steps that would enable the device to detect an unbalanced situation and directly notify the user. These steps included:

1. Detect the maximum peak of the positive amplitude
2. Detect the minimum peak of the negative amplitude
3. Calculate the time difference between the maximum and minimum peaks
4. Notify the user if a specified time is calculated

Given these specific steps, the team researched and brainstormed electronic components that would enable the device to achieve these functions. A Morphological chart was developed to organize necessary functions and potential means.

Table 14: Morphological Chart-Electronic functions and means

Functions	**Means**	**1**	**2**	**3**
Detect peaks; Calculate time between peaks		Arduino Microcontroller	VEX	
Notify user		Buzzer	Vibration	LED
Power device		Rechargeable batteries	Throw-away batteries	Lithium Battery pack
Turn on/off		Button	Toggle switch	Slide-switch
Monitor acceleration		ADXL345 accelerometer		

To detect the peaks and calculate the time between the peaks, the device needed to contain a microcontroller. Two microcontrollers were identified: the Arduino Duemilanove and the Vex. The Arduino Duemilanove was chosen because it has open source software, so there were many resource materials, example codes, tutorials and other reference materials online. In order to notify the user of imbalance, the device could buzz, vibrate, or light up. During an interview an elderly client stated that a sound would be the best way to notify the user. Therefore, a buzzer was chosen as the best means of notifying the user. Since the team chose to use the Arduino, the device required a Lithium battery pack as its power source. When determining the best means of turning the device on and off, the team considered which type of switch would be least likely to get bumped and turn the device on or off unintentionally. Therefore, a slide switch was chosen. The ADXL345 accelerometer was chosen to monitor acceleration because it is the accelerometer in the KinetaMap which was used in testing.

In order to understand the gravitational effect on the accelerometer and the offset values we obtained, as well as the change in offset before and after the STS motion, tabletop testing was performed to understand how acceleration offset values changed as the device is tilted.

The first test performed was the z axis rotation test and was used to determine if changes in the orientation relative to gravitational acceleration will yield a different static offset in the device. On each axis, device was positioned with Z axis parallel to gravitation acceleration, with positive end of the axis point upward on a table top. Device was tipped forward, so that the Z axis was offset by approximately 45 degrees relative to gravitational acceleration vector, device was held in this position for some duration of time. Then device was further-tipped forward, such that the switch-face of the device was against the tabletop and Z axis was perpendicular to gravitational acceleration. Plots were created for X (Figure 49), Y (Figure 50), and Z (Figure 51) axis as well as a Magnitude plot (Figure 52) to demonstrate the observed change in baseline value at different orientations.

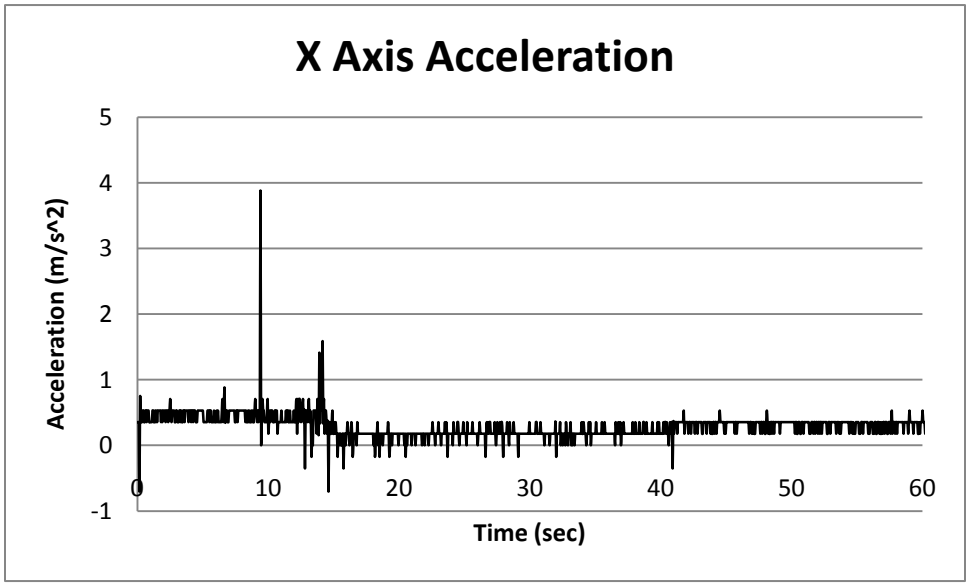


Figure 49: X axis acceleration during Z axis tilt

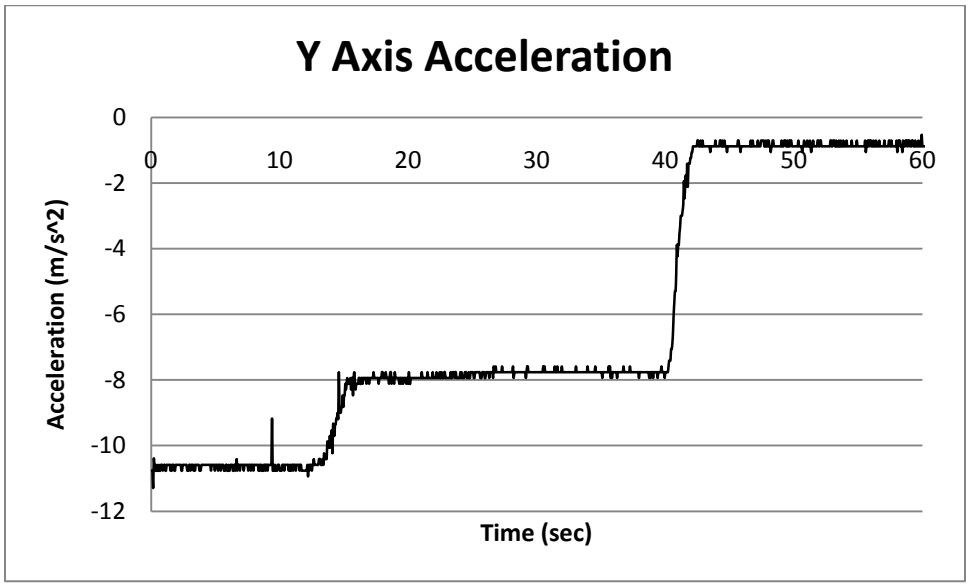


Figure 50: Y axis acceleration during Z axis tilt

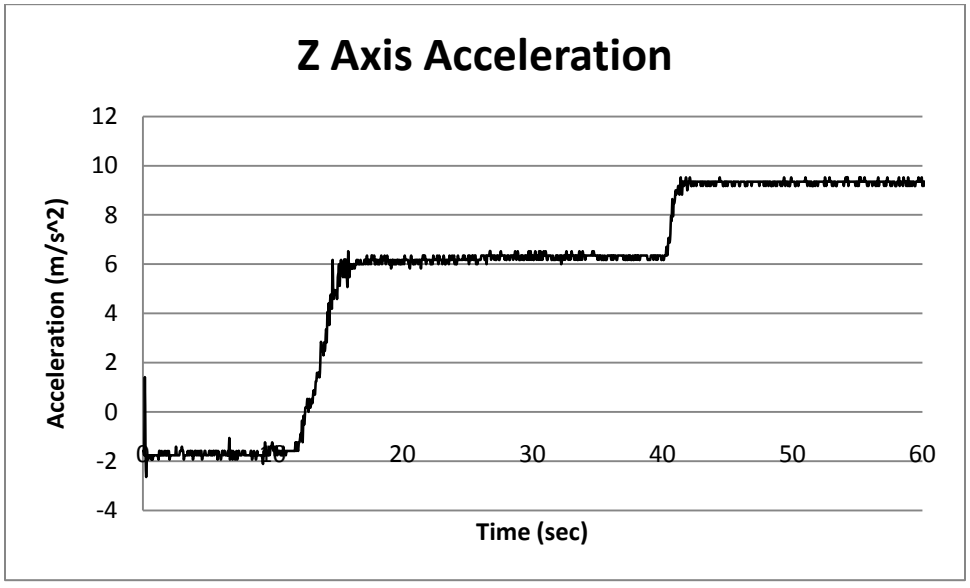


Figure 51: Z axis acceleration during Z axis tilt

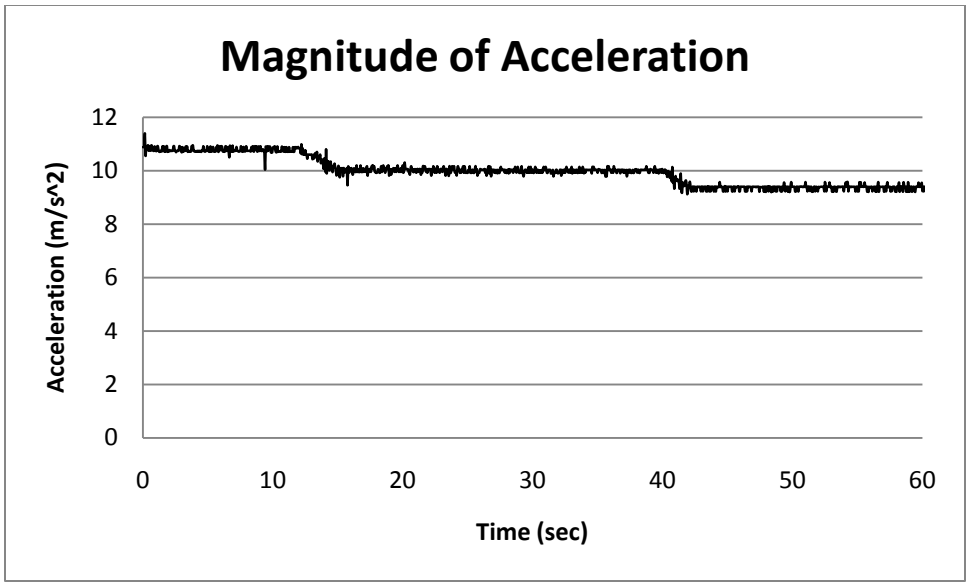


Figure 52: Magnitude of Acceleration during Z axis Tilt

Next x axis rotation tests were performed and the device was positioned with X axis initially positioned 90 degrees askew relative to gravitational acceleration (with the same starting position as described for the Z axis test). The device was then tipped approximately 45 degrees, with the +X axis coming more closer to the gravitational acceleration vector (pointing downward). This position was then held, followed by the device being tipped to a position where the +X axis pointed approximately parallel to the gravitational acceleration vector (downward). X (Figure 53), Y(Figure 54), and Z (Figure 55) axis were again plotted, in addition to a magnitude plot (Figure 56). Pulsatile noise artifacts were due to the imperfections of the movement of the device (human hand).

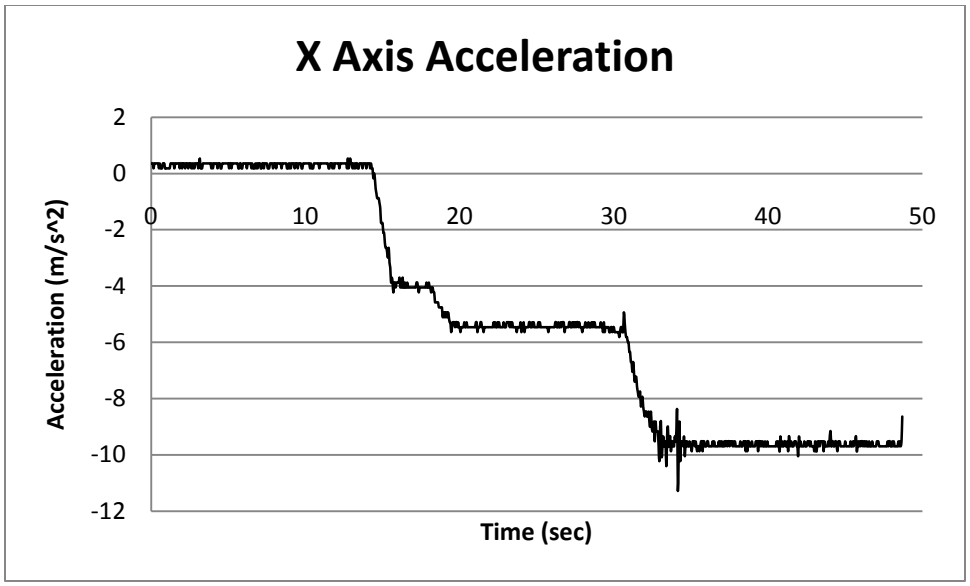


Figure 53: X axis acceleration during X axis tilt

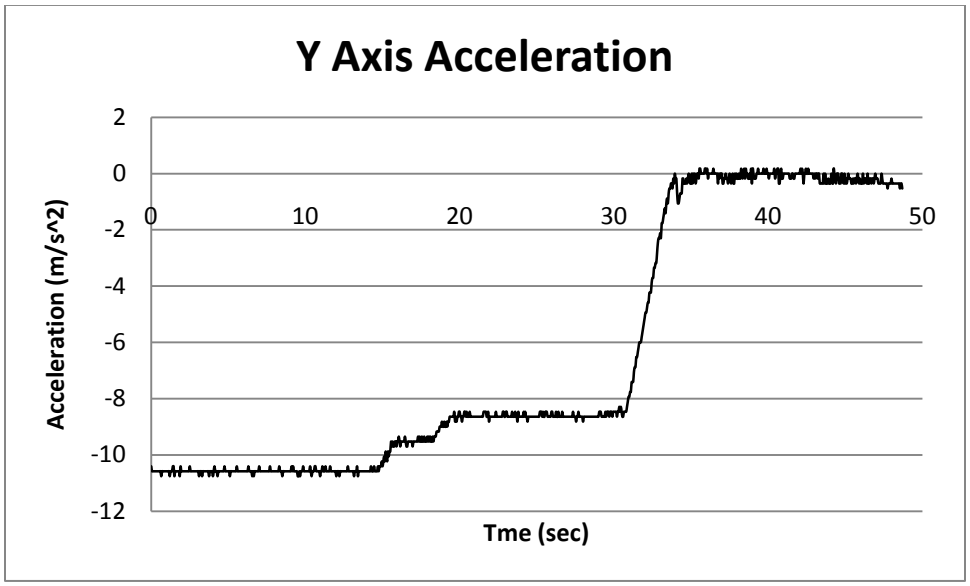


Figure 54 Y axis acceleration during X axis tilt

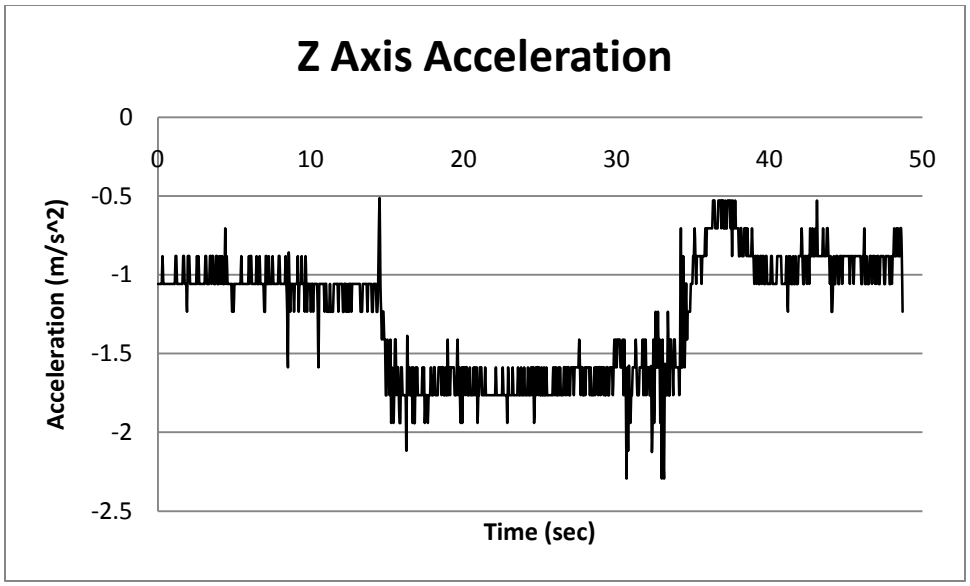


Figure 55 Z axis acceleration during X axis tilt

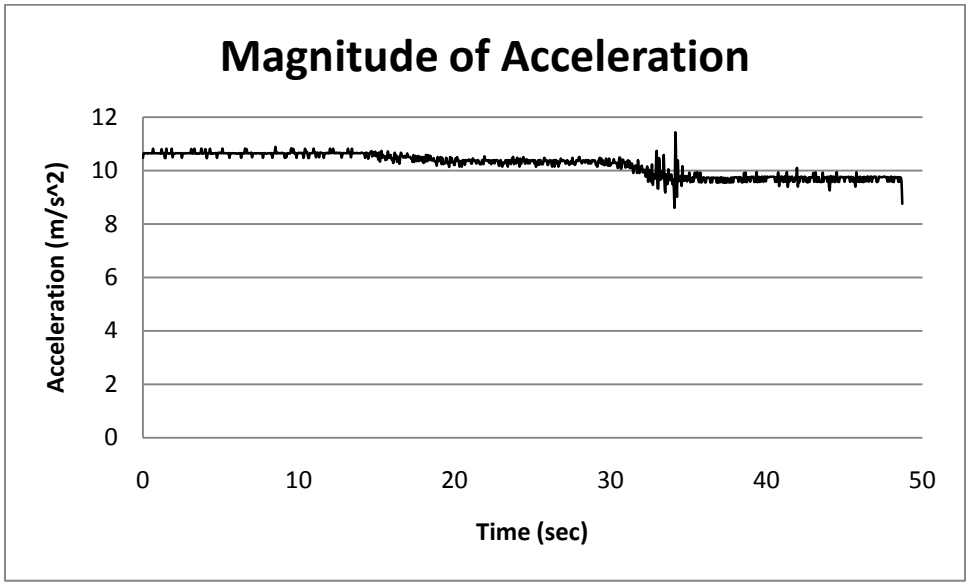


Figure 56 Magnitude of Acceleration during X Axis Tilt

Lastly y-axis rotation tests were performed and the device was positioned with the positive Y axis pointing directly upward, in parallel with the gravitational acceleration vector. The device was then rotated toward the USB port on the 'top' of the device, thusly also creating a change in the orientation of the Z vector relative to gravity, first at approximately 45 degrees, followed by a perpendicular orientation. Observed below are the X (Figure 57), Y (Figure 58), Z (Figure 59), and magnitude (Figure 60) changes for this procedure. Note that during any change in orientation of the device, more than one axis will be effected.

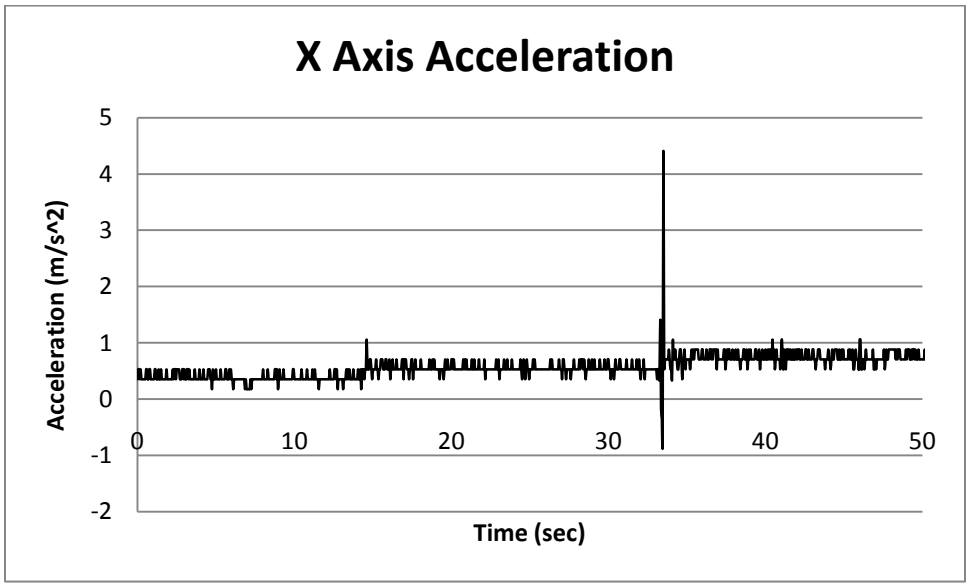


Figure 57: X axis acceleration during Y axis tilt

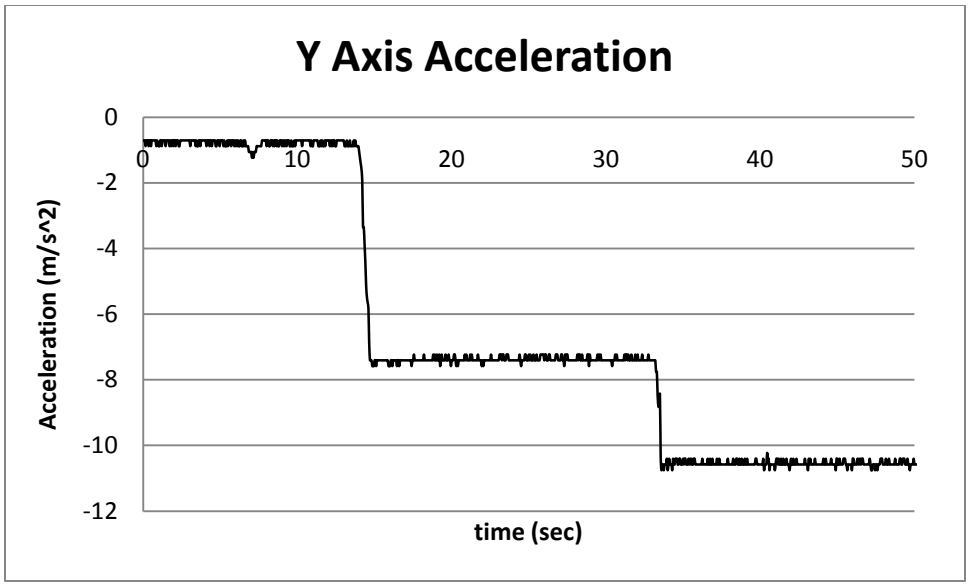


Figure 58: Y axis acceleration during Y axis tilt

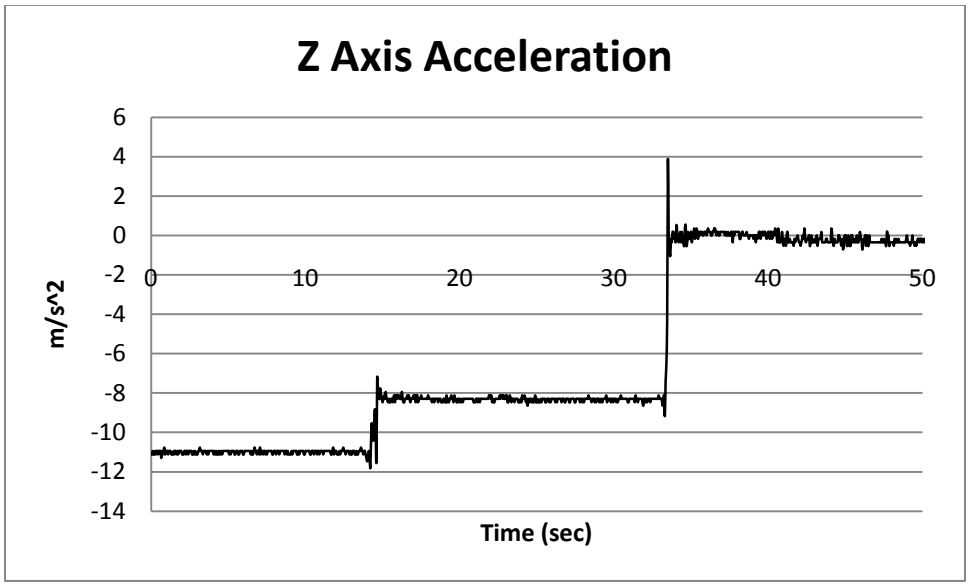


Figure 59: Z axis acceleration during Y axis tilt

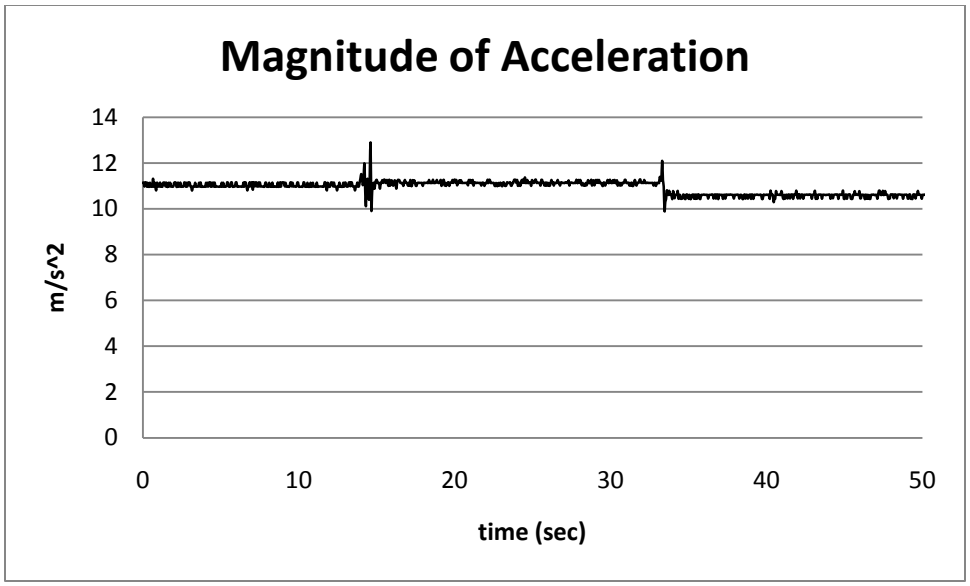


Figure 60: Magnitude of Acceleration during Y axis Tilt

Table 15: Change in magnitude of acceleration due to tilt on X, Y, and Z axis

Axis of rotation	Maximum change in magnitude value
X	0.9 (m/s^2)
Y	0.7 (m/s^2)
Z	1.4 (m/s^2)

Note that this greater observed change in Z axis data due to gravitational acceleration is consistent with the higher 0g bias sensitivity as noted on the ADXL 345 datasheet, as shown below in Figure 61.

Sensitivity Change Due to Temperature			±0.01		%/°C
0 *g* BIAS LEVEL	Each axis				
0 *g* Output for X_{OUT}, Y_{OUT}		−150	±40	+150	mg
0 *g* Output for Z_{OUT}		−250	±80	+250	mg
0 *g* Offset vs. Temperature for x-, y-Axes			±0.8		mg/°C
0 *g* Offset vs. Temperature for z-Axis			±4.5		mg/°C

Figure 61: ADXL345 Data Sheet 0g Bias Level

After the team identified the components of the device, the design team brainstormed and researched how the components would interface to notify the user of an unbalanced situation. First, sample code was used to verify that the buzzer could make a sound and that the accelerometer could collect data. In order for the device to detect the maximum and minimum peaks, the team chose to design the device such that it did not look at the baseline noise. This eliminated the possibility of the device identifying a peak in the baseline as a maximum or minimum peak. Therefore, after a certain baseline was reached the device would start looking for the maximum and minimum peaks. The following (Figure 62) is the basic block diagram of the designed algorithm.

Figure 62: Block diagram of algorithm.

The buzzer would sound if an unbalanced situation occurred and reset if a balanced situation occurred. To specify the difference between subject 1's unbalanced and balanced STS, the time range for a balanced STS and unbalanced STS was calculated by adding and subtracting the standard deviation to and from the corresponding average time. Subject 1's time ranges are shown in Table 16.

Table 16: Calculation of Subject 1's time range

Subject 1	Average Time (s)	Standard Deviation	Time Range (s)
Tandem	0.45	±0.16	0.29 – 0.61
Shoulder-width	0.85	± 0.20	0.65- 1.1

The team decided that above an expected time (T_e) would classify as balanced and below the same expected time, T_e, would classify as unbalanced. In choosing T_e the team chose a threshold acceleration that encompassed as many unbalanced situations as possible without having false errors. Therefore, the team chose 0.60 seconds, the latter end of the tandem time range, as the expected time. If the device calculated a time equal to or less than the expected time, then the device would buzz. If the device calculated a time greater than the expected time, then the device would reset as shown in Figure 63.

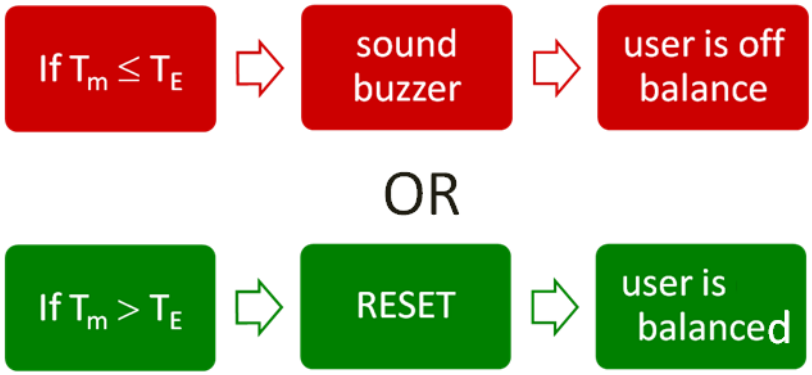


Figure 63: Block diagram of device function.

The device was created for subject 1 based on this conceptual design. The final design is described in following section.

5 FINAL DESIGN

The purpose of the device is to detect an unbalanced STS in terms of acceleration and notify the user when an unbalanced STS occurs. The team found that the STS activity produced an acceleration curve that contained a positive and negative amplitude of acceleration that corresponded to the forward and backward sway of the STS activity, respectively. Based on preliminary results, the time between the positive and negative peaks of these amplitudes was significantly longer in shoulder-width trials than in tandem trials. Comfort ratings of all subjects and COP plots confirmed that shoulder-width trials represented a balanced situation and tandem trials represented imbalance. Therefore, the device was designed to calculate the time interval between the positive and negative peaks of the STS acceleration curve and determine if the time corresponded to a balanced or an unbalanced situation. In preliminary testing the time between peaks was not shown to be significantly similar between subjects, so specifications were designed based on the data obtained from Subject 1. This section details the functions, specifications, and components used in the final design.

5.1 OVERVIEW OF FINAL DESIGN

Our final design and wearable balance control indicator is called *Duino Balance* and is shown in Figure 64. The device was enclosed in a plastic project box and is attached to a belt that can be worn around the user's waist. The overall size of the device is 4 inches long by 2.5 inches wide by 2 inches tall and weighs about 2 pounds.

Figure 64: Duino Balance: A Wearable Balance Control Indicator.

The belt is worn around the user's waist with the device located on the right hip as shown in Figure 65. The device is only be worn when performing the STS activity and provides instant feedback to the user about their balance control. If the user is balanced during the STS activity then the device does not buzz. However, if the user unbalanced during the STS activity, then the device buzzes for 3 seconds.

Figure 65: Placement of device on user.

5.2 DESIGN COMPONENTS

Duino Balance consists of three main components, an ADXL345 tri-axis accelerometer, Arduino Duemilanove Microcontroller Board, and a CEM1203 buzzer. The device also includes a rechargeable battery pack and slide switch for powering the device, as well as a protoboard for connecting and attaching all the components.

The primary component of our device is the Arduino Duemilanove microcontroller board (Figure 66). In our device, the Arduino microcontroller board is used to collect the data from the accelerometer, process the data, and sound a buzzer if necessary. One of the key components on the Arduino board is the USB connector (A to B plug) which can be used to connect the board to a computer to program a code onto the board, or to charge or power the board. The board requires between 7 and 12 Volts of power to run properly. Another important component is the

power connector which can be used to connect the device to a 9V wall adapter plug, allowing the device to be charged through a wall outlet. The digital, analog, and power inputs can be used to connect additional components to the board. Lastly, the main component of the Arduino Duemilanove board is the ATmega168 Microcontroller which was used for the digital signal processing. (Arduino, 2009)

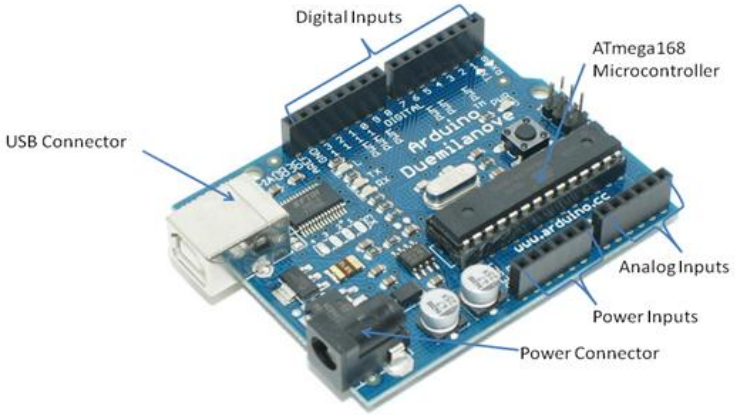
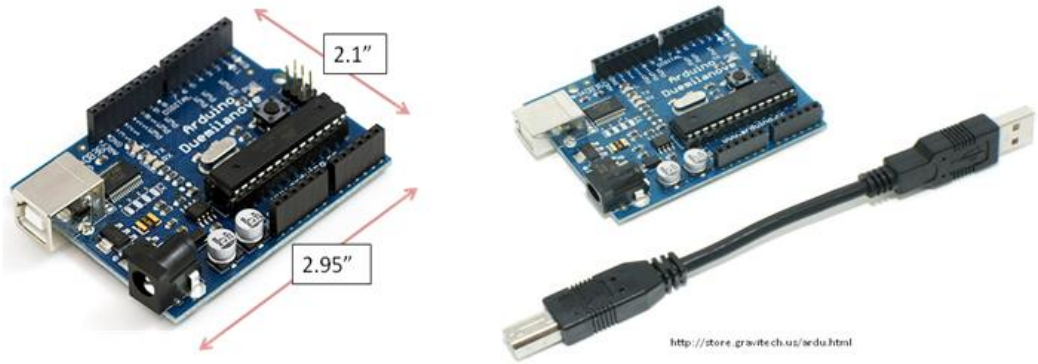


Figure 66: Arduino Duemilanove Microcontroller Board (Arduino, 2009).

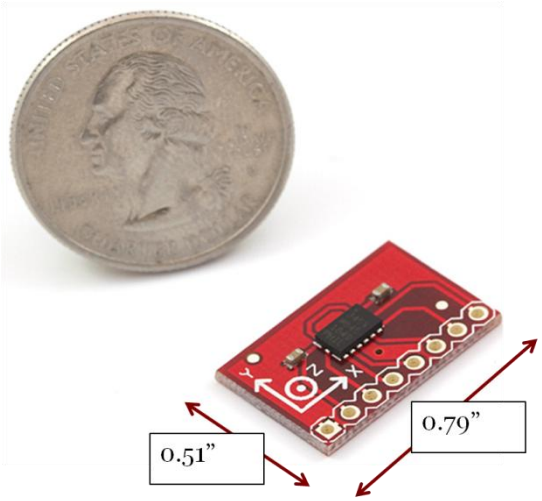
The board can be programmed using Arduino software, which is open source software based on the C/C++ programming language. Code is stored on the Arduino using internal 16KB of Flash memory. Since this is open source software, there are a lot of resource materials, example codes, tutorials and other reference materials online. The board is 2.1" wide by 2.95" long by 0.6" tall and is shown in Figure 67 relative to a 6 inch A to B USB plug. (Arduino, 2009)



http://www.liquidware.com/shop/show/ARD/Arduino+Duemilanove

Figure 67: Size of Arduino and its size relative to a 6 inch A to B USB plug (Arduino, 2009).

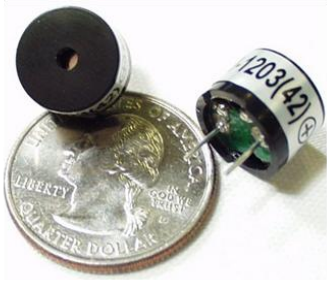
The most important component of our device is the ADXL345 accelerometer (Figure 68) that is used to sense accelerations patterns during the STS activity. One of the reasons we chose this specific accelerometer because it is the same accelerometer in the SparkFun KinetaMap that we used in preliminary testing. Also according to the ADXL345 Data Sheet, “the ADXL345 is well suited for mobile device applications. It measures the static acceleration of gravity in tilt sensing applications, as well as the dynamic acceleration resulting from motion” (Analog Devices, 2009). The ADXL345 is a low power, tri-axial accelerometer that can be set to different sensitivities, but for our application is set to +/-2g. This accelerometer is attached to a breakout board allowing for a simple connection to the Arduino microcontroller board (Sparkfun Electronics, 2009). The accelerometer is connected to the analog pins of the Arduino using an I2C configuration. This allows the Arduino to process the data collected by the accelerometer.



http://www.sparkfun.com/commerce/product_info.php?products_id=9156

Figure 68: ADXL345 Tri-axis Accelerometer (Sparkfun Electronics, 2009).

A buzzer is also included in our device to alert the user of an off balance situation. The buzzer produces a loud sound at 2.04 kHz and is shown in Figure 69, where its size is compared relative to a U.S. quarter (Sparkfun Electronics, 2009). The Arduino processes the data from the sit to stand activity and determines whether or not to sound the buzzer.



http://www.sparkfun.com/commerce/product_info.php?products_id=7950

Figure 69: CEM 1203 Buzzer (Sparkfun Electronics, 2009).

Since the Arduino microcontroller board requires 7-12V of power, another component of our device is a lithium battery pack because otherwise the device would require connection to a computer constant connection to a computer. This component allows the device to be both wireless and portable. The high capacity lithium battery attaches to the Arduino Duemilanove with screws through the holes in the Arduino board, and fits below the Arduino board (Figure

70). The PCB board that the battery is attached to is the same size as the Arduino board, and the height of the battery is approximately 0.4 inches. The battery can provide power to the device for up to 29 hours, this would allow the user to perform the STS activity many times before recharging the device. There are several ways the battery pack can be charged including using the USB plug on the Arduino to connect a computer, a mini-USB to connect the battery pack to a computer or a wall adapter plug (Huynh, 2009).

http://www.liquidware.com/shop/show/ATG/Arduino+to+Go

Figure 70: Rechargeable battery pack (left) and battery attached to Arduino (right) (Allum, 1999) (Arduino, 2009).

The user charges our balance control indicator using a 9V wall adapter plug. This 2.5mm wall adapter plug with a positive center (Figure 71) connects to the power connector on the Arduino board. The Alternative Current (AC) from the wall is converted to 9V Direct Current (DC) that is used to charge the battery pack that powers our device. (Maker Media, 2009)

Figure 71: 9V Wall Adapter Plug (Maker Media, 2009).

A slide switch like the one shown in Figure 72 is used to turn the device on and off. (Digikey, 2009) This type of switch was selected because we want an elderly user to be able to easily turn the device on and off.

Figure 72: On-Off Slide Switch (Digikey, 2009).

Lastly, a ProtoShield board (Figure 73) was used as a PCB with attached header pins that connect directly to the header pins on the Arduino (Sparkfun Electronics, 2009). The board was used to attach the accelerometer, buzzer, two 2k resistors and a 100Ω resistor both of which are needed to reduce the power from the Arduino to the accelerometer and buzzer.

Figure 73: ProtoShield Board (Sparkfun Electronics, 2009).

5.3 BUILDING THE DEVICE

A schematic diagram of the device is shown in Figure 74, where the ADXL345 accelerometer is shown in red, Arduino Duemilanove Microcontroller Board is shown in blue,

and the CEM1203 buzzer is shown in green.

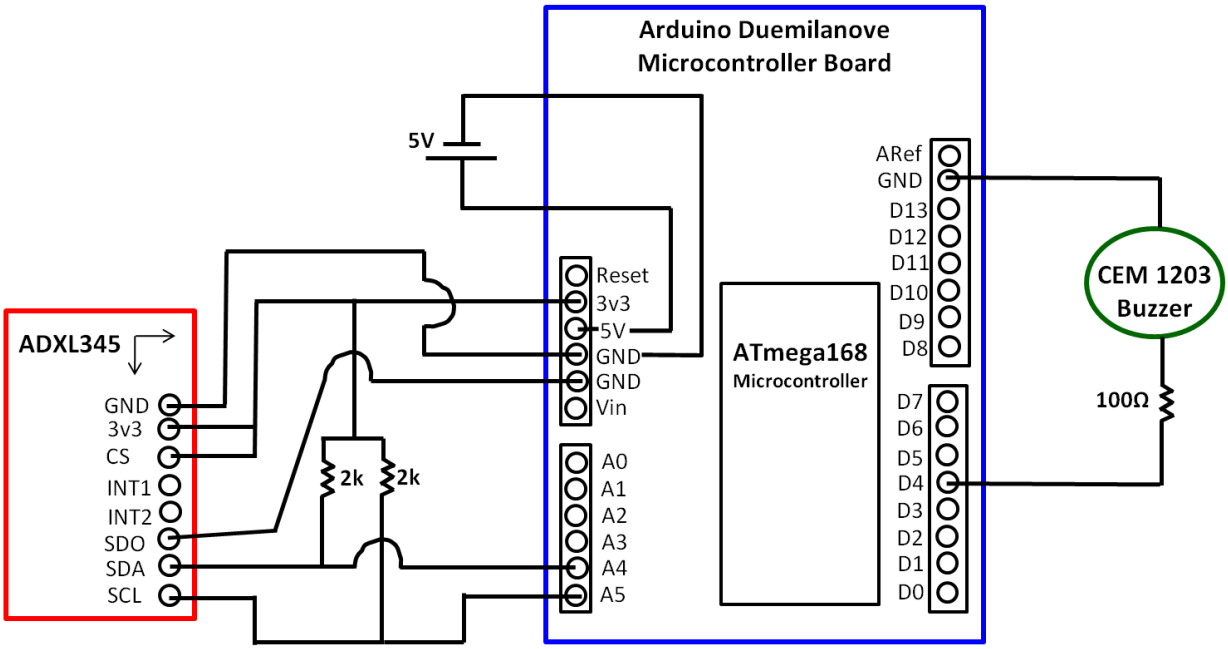


Figure 74: Schematic of Device.

The Arduino Duemilanove Microcontroller supports both SPI and I^2C communication. However, the Arduino programming language does not have functions for SPI communication; therefore we connected the ADXL345 accelerometer to the Arduino using I^2C (Inter-Integrated Circuit). I^2C wiring connections between the Arduino and ADXL345 accelerometer are shown in Figure 74. The I^2C pins on the Arduino are Analog pins 4 and 5, where analog pin 4 is wired to the Serial Data (SDA) pin on the accelerometer and analog pin 5 is connected to Serial Clock (SCL) pin on the accelerometer (Arduino, 2009). Pull up resistors, the 2k resistors shown in Figure 74 were suggested in the data sheet for the accelerometer, are needed to reduce the voltage from the Arduino because the ADXL345 cannot handle more than 3.6V (Analog Devices, 2009). The SDO pin on the ADXL345 was connected to ground on the Arduino. The 3V3 and CS pins on the accelerometer were connected to the 3V3 pin on the Arduino.

One lead of the CEM 1203 buzzer was connected to digital pin 4 on the Arduino Microcontroller Board and the other was connected to ground. A 100Ω resistor was connected between pin 4 and the buzzer because the current from the Arduino was too high for the buzzer to handle. The proper resistor value was determined using the equations and calculations below:

$$Maximum\ current\ for\ Arduino = 40mA$$

$$Coil\ resistance = 42\ \Omega\ \pm 6.3$$

$$Arduino\ Voltage = 5V$$

$$Ohm's\ Law = V = IR$$

$$I = \frac{V}{R}$$

$$I = \frac{5V}{42\Omega} = 0.119A = 119mA$$

$$I = Actual - Max = 119mA - 40mA = 79mA$$

$$79mA > 40mA$$

Since the current is greater than the Arduino can handle we added a resistor. If we add a 100Ω resistor to the circuit:

$$R = 42\Omega + 100\Omega = 142\Omega$$

$$I = \frac{V}{R} = \frac{5V}{142\Omega} = 0.0352A = 35.2mA$$

With the 100 Ω resistor the current was 35.2 mA which is less than the maximum current for the Arduino of 40mA. The 100 Ω resistor provided enough resistance to activate the buzzer.

Lastly, the rechargeable battery pack was connected to the 5V power pin and ground pin on the Arduino Microcontroller Board. The battery pack provides enough power to operate the device without being connected to a computer.

The final assembly of the device is shown in Figure 75. The battery pack was also connected to a slide switch for turning the device on an off. Some foam was also added inside the box on both sides of the device. The foam was used to prevent the device from moving within the box because the box was too long.

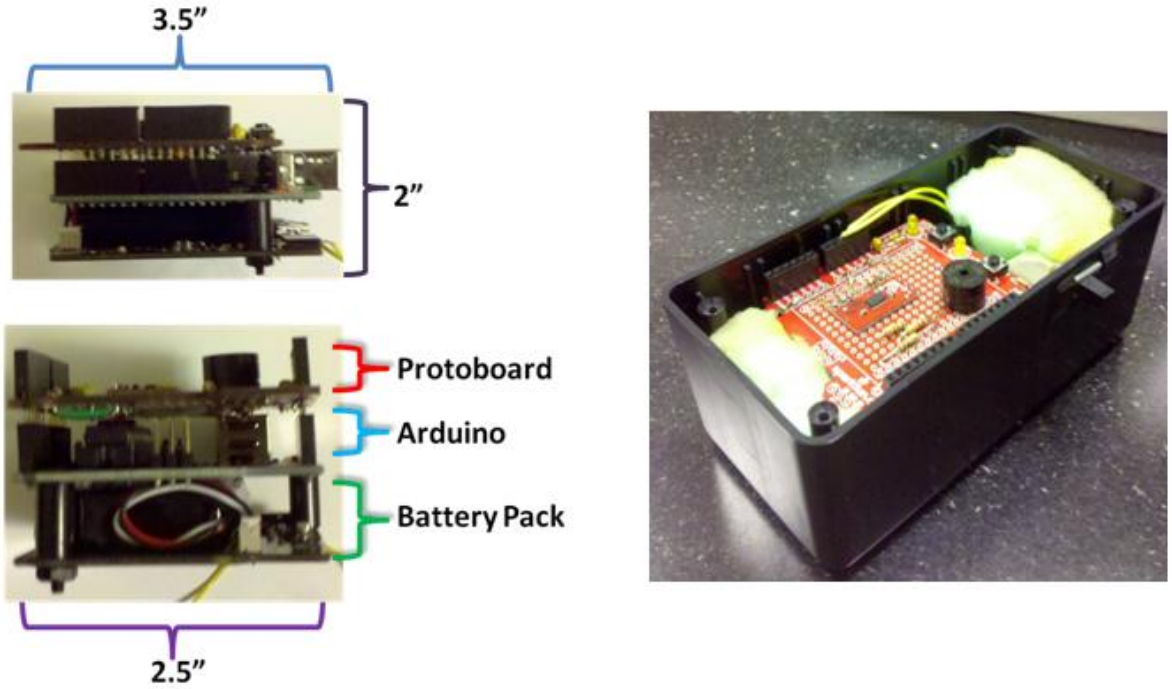


Figure 75: Final Product of Device.

The list and size of all the components in the *Duino Balance* device is included in Table 17. The cost of each component and the total cost of the device is also listed. The total cost to produce our device was $143.69. The myHalo Monitoring and BrickHouse fall detection devices cost about $200 for the device plus additional monthly fees for the monitoring service. (Halo Monitoring, 2009) The Wii gaming console and Wii Fit Balance Board costs approximate

$300. Therefore, the hardware of our device would not only be cheaper than the current monitoring systems but also significantly cheaper in the long term because it would not require monthly fees for monitoring.

Table 17: Cost of Device

Component	Cost	Size
Arduino Duemilanove Microcontroller Board	$29.95	Length: 2.95” Width: 2.1” Height: 0.6” Weight: 0.07 lb
High Capacity Lithium Battery Backpack	$47.35	Length: 2.73” Width: 2.1” Height: 0.498” Weight: 0.22 lb
Arduino ProtoShield Board	$16.95	Length: 2.95” Width: 2.1” Height: 0.6” Weight: 0.1 lb
ADXL345-3 Axis Accelerometer Board	$27.95	Length: 0.79” Width: 0.51”
CEM 1203 Buzzer	$1.95	Diameter: 0.47” Height: 0.55”
Slide Switch	$1.55	Width: 0.2”
9V Wall Adapter Plug	$6.50	
Plastic Project Box	$3.69	Length: 4” Width: 2.5” Height: 4” Weight: 0.6 lb
3 Resistors	$0.30	2k (Qty:2) 100Ω
Laptop Shoulder Strap	$5.00	Weight: 1 lb
Velcro	$2.50	
Total	**$143.69**	Length: 4” Width: 2.5” Height: 2” Weight: 2 lb

5.4 PROGRAMMING THE DEVICE

As mentioned previously, the Arduino Microcontroller is programmed using the Arduino software which is software that utilizes a programming language similar to C/C++. The code was written in an Arduino sketch, compiled and uploaded onto the Arduino Microcontroller Board through a USB.

Since through testing we determined that the time between the maximum positive acceleration peak and minimum negative acceleration peak (Figure 76) was significantly different for balanced and unbalanced trials, the device was programmed to detect these two peaks and measure the time between them.

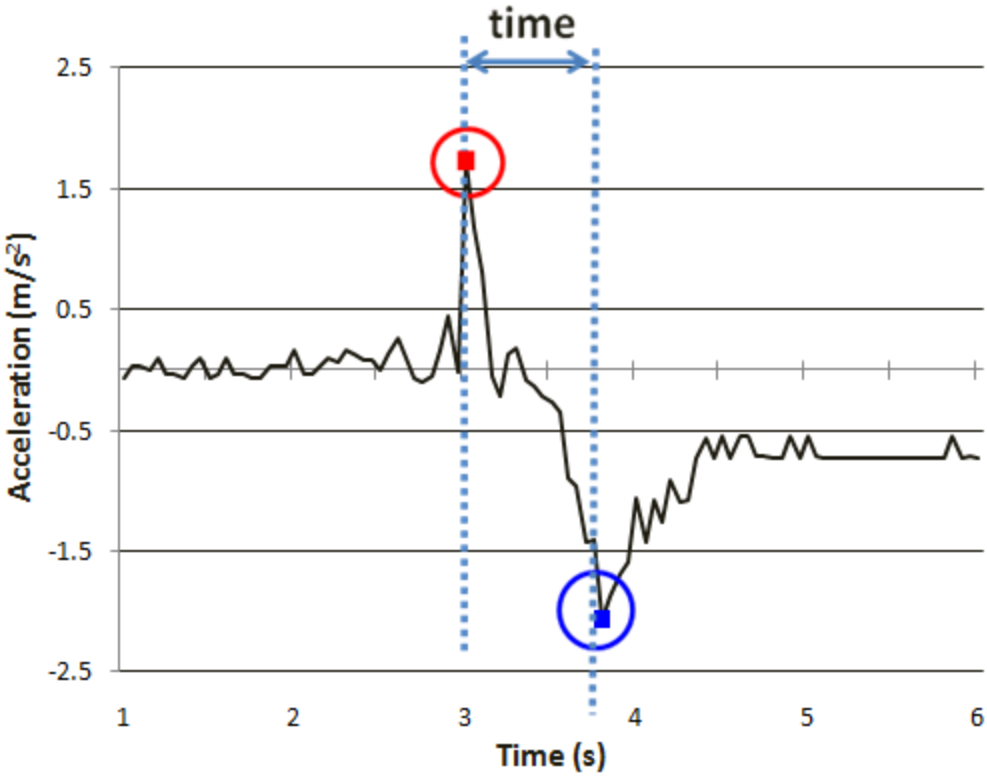


Figure 76: Acceleration curve for the STS activity showing the peaks and time between them.

If the time between the peaks is greater than the expected time, then the user is considered balanced and the device resets. However, if the time between the peaks is less than the expected time the user is unbalanced and the device buzzes. Through testing, we determined that the expected time between the peaks for subject 1 was 0.6 seconds, and the device was programmed specifically for this expected time. The block diagram of the final design is shown in Figure 77.

Figure 77: Block Diagram of Final Design.

The program consisted of four main sections:

1. Defining and initializing the I^2C communication between the Arduino and ADXL345
2. Reading and writing the data from the accelerometer
3. Processing the accelerometer data
4. Sounding the buzzer

Several of the parameters are defined as constants in the beginning of the program using the #define function. The *#define* function was used to assign a constant value to a variable so that whenever this variable is used throughout the code, the compiler replaced the variable with the defined constant (Arduino Reference, 2009). The *#define* function was used specifically to define the expected time between peaks, the threshold acceleration, expected length of time for

the entire STS activity and the duration time for the buzzer. This feature made the device easily programmable for different users.

The first section of the program was used to define the address of the accelerometer and initialize the I2C communication between the ADXL345 and the Arduino and was adapted from an example code on the Arduino Forum (Arduino Forum, 2009). The first line of the program is *#include <wire.h>* function, which was necessary to use the Wire library functions within the Arduino software. The next several lines of code are used to define the address of the ADXL345 accelerometer and numbers of bytes, in this case 6, that the device reads using the *#define* function. The device address found in the ADXL345 datasheet is 0x53 (Analog Devices, 2009). Next the I^2C communication is initialized using the *Wire.begin* function and the serial output is set to 9600 bits/second (Arduino Forum, 2009).

The next section of the program was used to read and write the data from the ADXL345 accelerometer. The ADXL345 is turned on using the *WriteTo* function and the power control register address 0x2D which is used to take the accelerometer out of sleep mode. Finally, the *regAddress* function is used to read the registers for each of the three axes on the ADXL345 (Arduino Forum, 2009). The ADXL345 data sheet states that the address for the x axis is 0x32 and 0x33, the y axis is 0x34 and 0x35, and the z axis is 0x36 and 0x37 (Analog Devices, 2009). The data from each axis is squared using the *sq(x)* function, then each square of each axis is added together using the + operator, and the square root of this sum is taken using the *sqrt(x)* function (Arduino, 2010) to represent the overall magnitude of the acceleration.

The third section of the code was used to process the accelerometer data, and is where the parameters defined at the beginning of the code are used. One of the defined parameters is

StartMinA which is the threshold acceleration, for subject 1 this acceleration was determined through testing to be 0.5 m/s^2. Another important parameter is t_E the expected time between the maximum and minimum peaks, which for subject 1 is 0.6 seconds corresponding to the value 0.6 * 1000 milliseconds. Initially, t_{AA} (time corresponding to the max acceleration), t_{AD} (time corresponding the minimum acceleration), A_A (maximum acceleration) and A_D (minimum acceleration) are set equal to zero. *TestStarted*, *TestFinished*, and *RingBuzzer* were set to false. The test is started and *TestStarted* is true when *A*, the current acceleration, is greater than *StartMinA*. The time the test was started was defined as *TimeStart* and current time was defined as *T*. The expected test length was defined at *TestLength*. In this section of code, *If* statements were used. A moving time window was used to search for the maximum acceleration peak A_A and minimum acceleration peak A_D. Therefore, throughout the duration of the test if *A* was greater than the value stored as the maximum acceleration A_A, and then A_A was now equal to *A*. Similarly, if *A* was less than the value stored as the minimum acceleration A_D, then A_D was now equal to *A*. The values t_{AA} for and t_{AD} correspond the time at which A_A and A_D respectively occurred, and these values change as A_A and A_D change. *TestFinished* was set to true when *T* minus *TimeStart* was greater than the *TestLength*. When *TestFinished* is true, if t_{AD} minus t_{AA} is less than t_E then *RingBuzzer* is true. When TestFinished is true, and t_{AD} minus t_{AA} is greater than or equal to t_E then *RingBuzzer* is false.

The final section of the program was the code for *RingBuzzer* or to sound the buzzer. The RingBuzzer code was modified from an example code by Rob Faludi (Faludi, 2007). The function *pinMode(4, OUTPUT)*was used to define digital pin 4 on the Arduino as the output pin for the buzzer. The buzzer was set to buzz at 2048 Hz for 3 seconds or 3000 milliseconds.

All four sections were integrated together to form the *Duino Balance* code which was used to analyze a STS activity to determine whether the user is balanced or unbalanced, and sound a buzzer for 3 seconds if the user is unbalanced. Figure 78 illustrates how the current device works for subject 1.

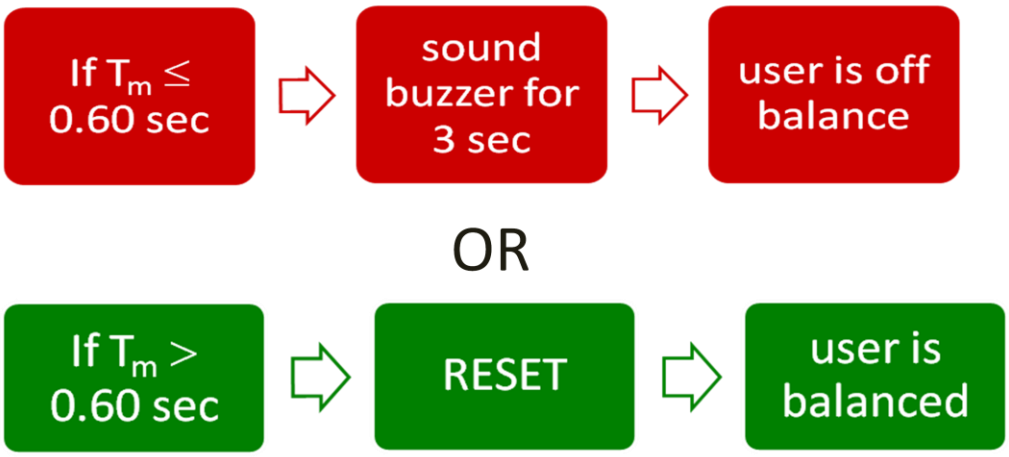


Figure 78: Block diagram of device functions and specifications for subject 1.

6 DESIGN VERIFICATION

In order to verify that our device operated properly, we performed STS tests using the same method that was used during preliminary testing except that our device was placed on the right hip and the KinetaMap on the left hip. The subject sat in a chair with their feet on a force platform, turned on both devices, and stood up from the chair with their feet shoulder width apart. The same test was then performed with their feet in tandem. Ten shoulder width STS tests were performed, followed by ten tandem tests. We used the KinetaMap and force platform data to confirm whether the STS activity was balanced or unbalanced. The device was programmed specifically for subject 1 with an expected time between the maximum and minimum acceleration peaks of 0.6 seconds. During the STS activity, if the time between the peaks was greater than 0.6 seconds then the trial was considered balanced and the device should reset without a buzz. But if the time between the peaks was less than or equal to 0.6 seconds, the device should buzz to indicate that the trial was unbalanced.

An example of the progression of one of the shoulder width STS trials is shown in Figure 79. During this trial, the subject appeared to remain balanced throughout the activity and the device did not buzz.

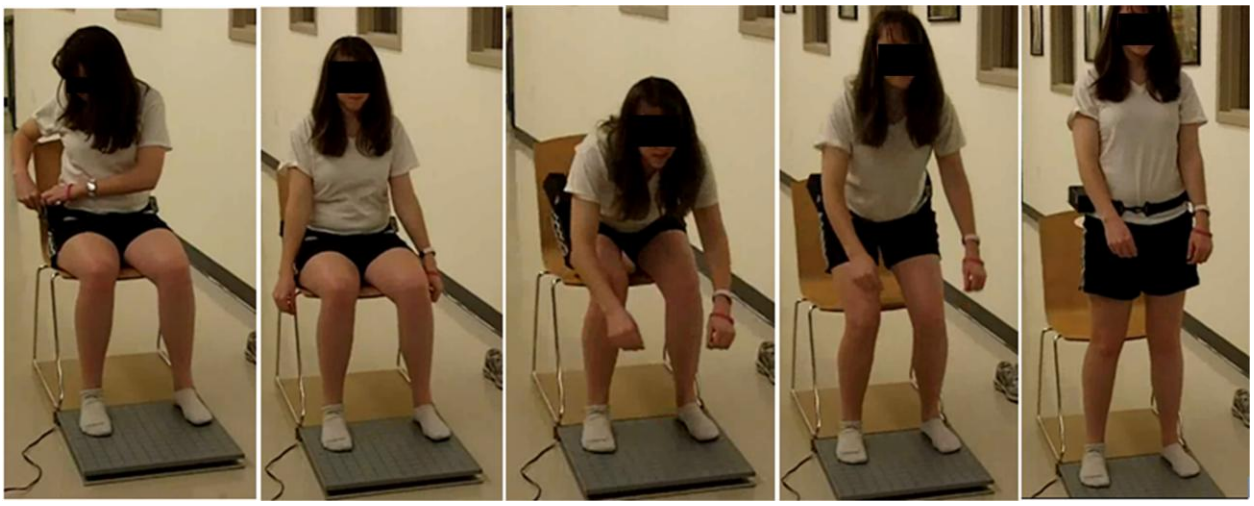


Figure 79: Shoulder Width STS with Device.

In order to confirm whether the trial was balanced, we analyzed the KinetaMap data and generated the acceleration vs. time plot shown in Figure 80. We identified the maximum and minimum acceleration peaks shown by the blue circles in the figure and measured the time between the two peaks. The time between the peaks was 0.95 seconds. This time was greater than the expected time of 0.6 seconds, which was representative of a balanced trial.

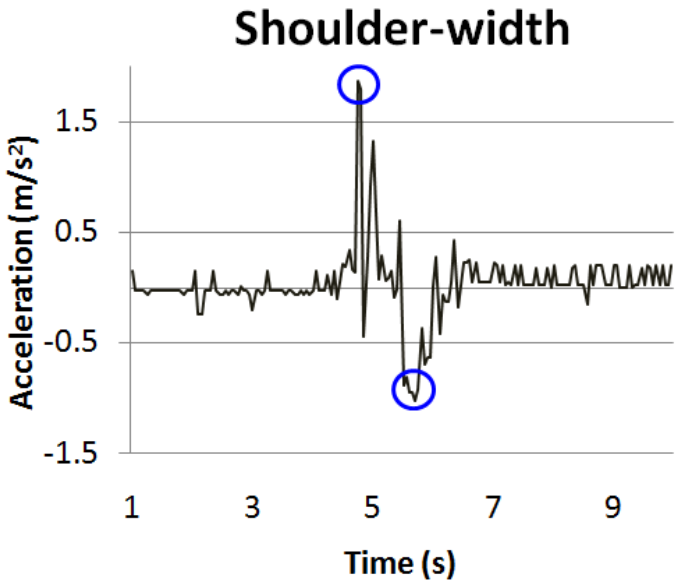


Figure 80: KinetaMap acceleration vs. time data during Shoulder Width STS with device.

In addition, we used the COP data as another method to confirm whether the trial was balanced. The COP data from the shoulder width STS trial is shown in black in Figure 81. The COP was compact and had little variation, which was also representative of a balanced trial.

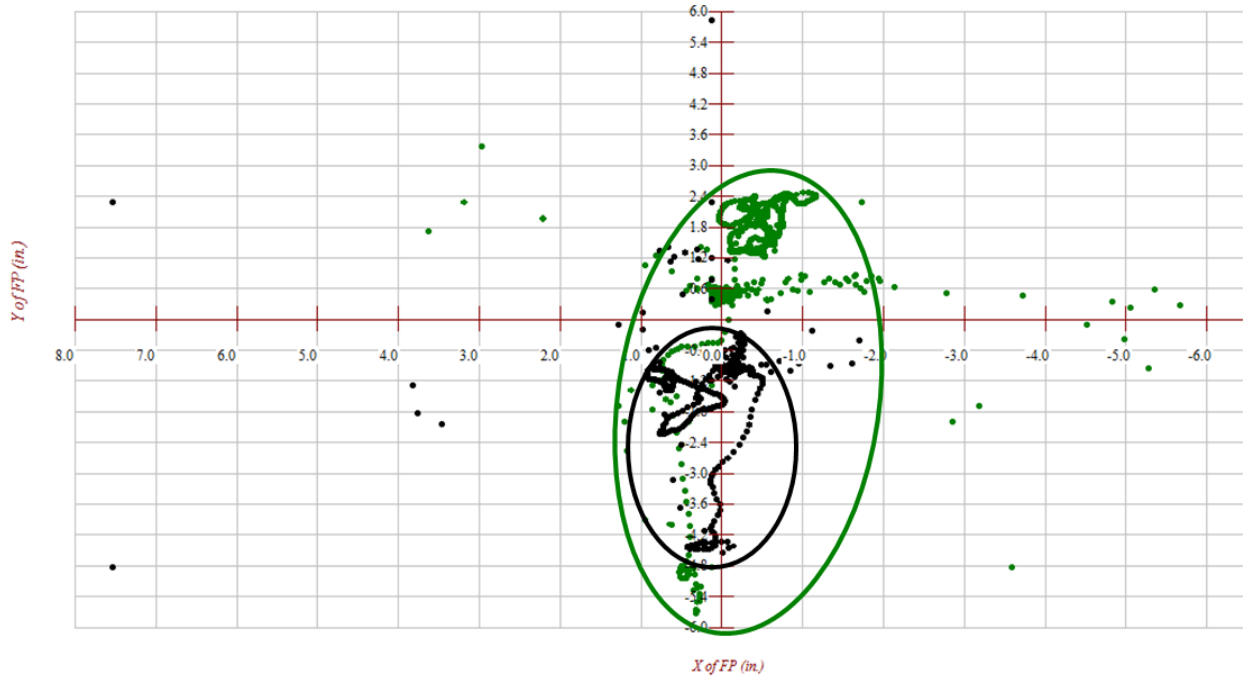


Figure 81 COP graph for shoulder width (black) and tandem (green) STS trial.

An example of a tandem STS test with the device is shown in Figure 82. The device buzzed after this trial and visual analysis also suggested that the subject was unbalanced.

Figure 82: Tandem STS with Device.

In order to confirm whether the device successfully identified an unbalanced trial, we first analyzed the KinetaMap data. The acceleration vs. time graph that we generated for this tandem trial is shown in Figure 83. The maximum and minimum acceleration peaks were identified and circled in blue. We determined that the time between these peaks was 0.45 seconds. This time was less than the expected time of 0.6 seconds, which was representative of an unbalanced trial.

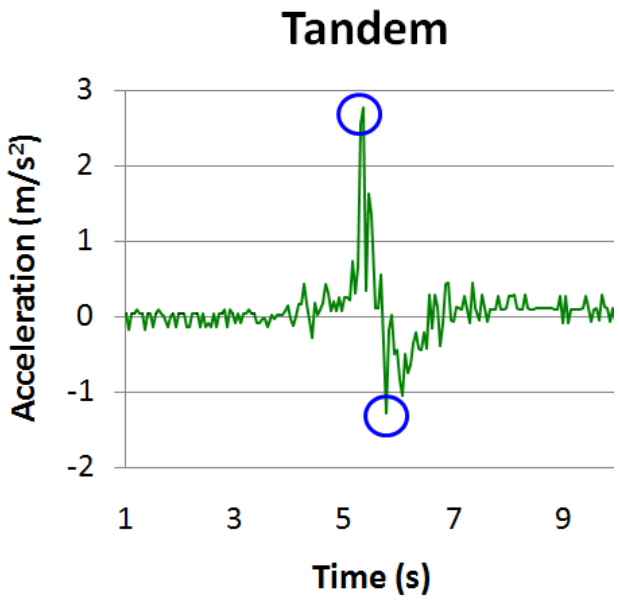


Figure 83: KinetaMap acceleration vs. time data during Tandem STS with device.

Similar to the shoulder width trials, we verified whether the trial was unbalanced by analyzing the COP data from the force platform. The COP data for the tandem trial is shown in green in Figure 81. As you can see, there was a greater variation in the green COP, which was consistent with an unbalanced trial.

After we completed and analyzed the ten shoulder width and ten tandem trials, we compiled the results and determined the accuracy of our device as shown in Figure 85. The COP and KinetaMap data confirmed that all ten shoulder width trials were balanced, so the device was not supposed to buzz at all during these trials. However, the device incorrectly buzzed and

produced a false alarm during one of the shoulder width trials. As a result, the device successfully identified a balanced STS activity for 9 out of 10 shoulder width trials. Whereas for the tandem trials, the COP and KinetaMap data identified all the trials as unbalanced, thus the device was supposed to buzz for all ten trials. But the device correctly buzzed and indicated an unbalanced situation during 8 out of the 10 tandem trials.

Table 18 Results of Device Testing

Trial Type	# of Trials	# of Balanced	# of Buzzes	Accuracy
Shoulder-Width	10	10	1	90%
Tandem	10	0	8	80%

Therefore, the device was 90% accurate for balanced shoulder width STS trials and 80% accurate for unbalanced tandem STS trials.

7 DISCUSSION

Testing of our final device revealed that our device correctly identified 9 out of 10 balanced trials. KinetaMap and COP data showed that all 10 trials were balanced; therefore the device produced a false alarm for one trial. We noticed that when the KinetaMap shifted during preliminary testing this caused a shorter time between the maximum and minimum acceleration peaks. Therefore, the false alarm during the device verification testing could have been caused by shifting of the device during the STS activity.

Testing of our final device revealed that our device correctly identified 8 out of 10 unbalanced trials. Kinetamap and COP data showed that all 10 trials were unbalanced; therefore the device produced a false alarm for two trials. We noticed that when the KinetaMap shifted during tandem preliminary testing, it caused a peak due to noise to have a more negative acceleration than the actual peak that occurred during the backward sway. Therefore, the two false alarms during the device verification testing could have been caused by shifting of the device during the STS activity. Due to these results, one improvement for the device would be creating an attachment that keeps the device secure and eliminates shifting during the STS activity.

The functionality of the *Duino Balance,* balance control indicator device is fairly unique in that no commercial device has been released to perform the same task, either by the same or different means. One current device which bears the most similarity to the balance control indicator is the Wii Fit videogame system, which is roughly based on a force platform with an interactive videogame interface that tests the user's balance and rewards them for strengthening their balance control. This device differs markedly from the balance control indicator in that it does not integrate into the daily lives of the elderly, utilizing a complex graphical user interface

that will be unfamiliar and difficult for many elderly subjects to comprehend. Another device with bears some similarity to the balance control indicator is the iShoe pressure mapping insole, which has not been released to market but remains in a patent-pending status. This device is worn underfoot and capable of measuring COP data and gathering balance control information from the user. However, this device differs from the balance control indicator in that the results are required to be interpreted by a professional and are far more complex than the simple 'balanced or not balanced' logic implemented by the balance control indicator. Because of the complexity of this device and the need for professional interpretation, it would not be useful in a home situation to integrate into a user's daily routine. Because of the simplicity of interpreting the balance control indicator's results as well as the ease of integration into the daily activities of the elderly, it has clear advantages over the similar balance control indicating devices that currently exist on the market or in product development.

The device differs from wearable fall indicators and PERS systems in its core functionality. The fall indicators and emergency response systems are a capable means of ensuring that elderly will receive proper attention after a fall has occurred, but neither addresses the need for preventative measures that can minimize or eliminate the risk of fall. Whereas the balance control indicator device will allow the elderly to strengthen their balance control, potentially eliminating the risk of fall and resultant injury, the fall detection device will simply be able to detect the fall once it has happened and reduce the risk of long-lie injuries and increase the survival rate. PERS systems may be less effective as they require that the user be conscious and capable of activating a call signal in order to receive help. The primary advantage present in the balance control indicator device is the ability to reduce the risk of fall through preventative exercises with instant feedback.

The objectives for the device, specifically strengthening of the user's awareness of their balance condition and proactive forcing of user rehabilitation, were both met in the design of the balance control indicator device. The Berg balance test indicates that the STS activity is a valid indication of the user's overall balance control, and therefore by creating a device that provides understandable feedback on the balance condition with which a user performs this activity the device has succeeded in assisting the user in becoming aware of their balance control condition. Whereas the user may not have been aware that he or she was in need of assistance in order to become balanced before the application of the device, a sounding buzzer will notify the user that he or she is not so balanced as he may have perceived. In the daily performance of the STS activity, and by attempting to reduce imbalance so as to not trigger the device to sound, the user will be forced to actively participate in rehabilitation. Without taking any extraordinary measure (as using the device can easily be incorporated into the daily routine of the elderly), the elderly user has taken a step towards the recognition and improvement of his or her balance condition.

The limitations to the data used to design the device would pose a quite passable challenge in verifying the functionality of the device for direct use in the elderly population. Foremost, the subjects used to gather information used to set device thresholds and values for programming were healthy young subjects, mostly under the age of 30. The values that would be obtained in elderly testing may vary, which would necessitate the reprogramming of the device. As with any real world device, there is the possibility for error in the results given by the device as well. Through preliminary testing, thresholds were collected that represented the data obtained from most of the trials collected. There are, of course, trials which may have been balanced but resulted in a peak-to-peak time duration of less than 0.6 seconds, and likewise for unbalanced

trials. Though there is this limitation, we do feel that in most instances our device will be able to accurately distinguish balanced from unbalanced situations with minimal opportunity for error.

The clear and present economic benefits of the widespread implementation of a device of this type are abundantly apparent. Minimization of healthcare dollars spent treating victims post-fall could reduce overall national spending. With an aging population and a need for reduction in healthcare dollars spent, the utility of preventative maintenance shines. A low cost in-home device that can reach a widespread population and assist them in regaining their balance and mobility could reduce overconsumption of natural resources used to produce other devices and facilities that are needed after falls have occurred, specifically increased rehabilitation facilities and products as well as mobility devices and intensive care resources.

The device has the capacity to reach a global market, in that it is low cost and that it is needed globally in any culture where there are elderly people. In areas where there is a reduced availability of healthcare a low-cost device that could prevent the need for expensive treatment would be a welcomed alternative to having lower income brackets not receive the healthcare that they need. Likewise, in developed areas with modernized healthcare a reduction of expenditures related to post-fall healthcare costs would also be welcomed as well as the obvious benefits to the quality of life of individuals that can avoid the devastating effects of a fall.

Despite advancements in treatment available to fall victims, the simplest means to avoid lasting fall-related disabilities and quality of life limitations would be to avoid falling and injuring oneself in the first place. Because of this, the device would succeed in addressing improvements to the quality of life of subjects who used it properly to address balance control issues before the issues resulted in a fall. By not only helping to prevent the lasting and

sometimes devastating effects of a fall injury, but also mitigating the financial ruin that can be resultant of a costly hospitalization and rehabilitation following a fall, the device serves to address the need for the elderly to live a good and satisfying life. This device would have a clear influence on the health and personal safety of its users, as it would attempt to remediate a potentially dangerous situation of an elderly person lacking proper balance.

Because of a simple and condensed list of components, and the low cost nature of the device, it is very manufacturable. The prototype was built entirely from off-shelf components, and could easily be optimized to minimize cost as well as reduce size and weight. The lithium ion rechargeable battery pack contained within the device reduces the need for repeated use of costly and environmentally-damaging alkaline batteries present in some electronics, though the need for proper disposal of the device and rechargeable battery pack would need to be addressed. In the prototype, all components in the design were RoHS compliant, or compliant in regard to restriction of hazardous substances such as lead, cadmium, and mercury. In this sense, the device is no more damaging to the ecological system than any other small consumer electronic device such as a phone or mp3 player.

As can be seen above, the balance control indicator device addresses a need for a low-cost in-home balance indicator for use by the elderly to analyze their balance and alert them of a lack of balanced state. The device was required by team-designed objectives to increase user awareness of his or her balance control and proactively force user rehabilitation, both of which the designed device was capable in its ability to do. The need for a low cost preventative device that can reduce or eliminate the occurrence of costly hospitalizations and needless diminishment of the quality of life of the elderly population due to falls is strongly presented in the staggering healthcare costs associated with elderly post-fall hospitalization, and the fact that these costs will

be rising exponentially due to a growing number of elderly within only the next ten years. The issue of an increased elderly population and thusly the need to maintain their quality of life is a global one, and the creation of a low cost device that could be accessible to lower income societies globally is an important step in the direction of improvement of the lives of elderly from many different situations. Manufacturability and environmental consciousness were factors that were considered during the production of the device, and because of this a device has been designed that is very manufacturable and additionally is RoHS compliant.

8 DESIGN VALIDATION

Through preliminary testing, the team designed a device to be used during the STS activity that detects and directly notifies the user of an unbalanced STS. The final device was tested for subject 1 (college-age female) and buzzed during an unbalanced STS situation and did not buzz during a balanced STS situation. Operation of the device required the user to perform the STS, both a rehabilitation and daily activity. Thus, the device demonstrated its ability to strengthen the user's awareness of their balance control and force user rehabilitation.

In addition, the design team verified that the design met client objectives by interviewing physical therapist and an elderly user. Both reported that the design will be "straightforward, easy to use, and not cumbersome". In addition, both clients reported that it could be used in a clinical setting such as a nursing home, physical therapy session, and also as additional tool to assess a patient's fall risk (Appendix C). However, the device was not tested and verified on an elderly user. Therefore, this section describes how the study can be continued to prove the device's viability for elderly users.

8.1 PRELIMINARY TESTING WITH ELDERLY SUBJECTS

The designers need to prove that the current device is able to accurately detect and notify an elderly user of an unbalanced STS and that it does not buzz during a balanced STS. This section outlines the preliminary testing that needs to be conducted in order to customize the device for an elderly user.

8.1.1 Materials and Methods

The team needs to gather all materials needed for the study. This includes at least 10 healthy elderly subjects age 70 years or older. The health of the subjects should be evaluated by a physical therapist. Subjects should score a 41 or above on the Berg Balance Test, meaning they

can walk independently without an assistive device or supervision (Internet Stroke Center, 2010). In addition, the SparkFun KinetaMap data logger will be used for collecting acceleration data. Set the parameters of the KinetaMap according to the parameters outlined in Appendix B. The AMTI AccuSway force platform and AMTI NetForce and BioAnalysis software will be used to record and analyze balance control data, respectively. A physical therapist should be present during all tests to rate the subject's STS performance according to how well they maintained their balance.

The force platform and a regular chair will be set up as shown in Figure 84. The force platform will be placed directly on the ground and a wooden platform, the same height as the force platform, will be placed adjacent to the force platform. The chair will be placed on top of the wooden platform and in front of the force platform so that when the subject sits in the chair their feet rest comfortably on the force platform. The KinetaMap will be attached with Velcro to an adjustable belt. The belt will be positioned so that the KinetaMap is mounted on the right side of the subject, externally adjacent to the iliac crest (Figure 85).

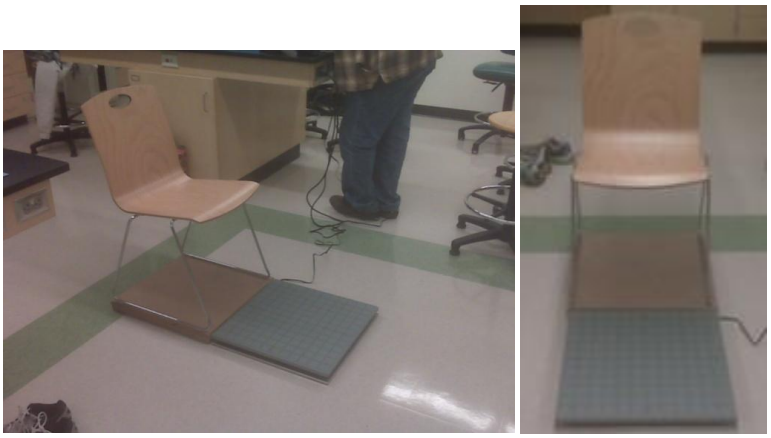


Figure 84: Chair and force platform experimental set-up

Figure 85: Attachment of the KinetaMap device

Using the NetForce Software, create a subject account for each subject. Press subject and then add record. Enter the subject information. Select the first subject. Press weigh and then tare and ask subject 1 to stand still with feet shoulder-width apart on the force platform. Click weigh. Their weight will be collected and saved in NetForce software.

With the KinetaMap attached properly to the subject's right hip, ask the subject to sit in the chair. Start NetForce data collection and press tare to start data collection. Ask the subject to turn on the KinetaMap. Wait for the KinetaMap's LED to start blinking blue, which signifies that the accelerometer has started collecting data. Ten seconds after the LED starts blinking blue, tell the subject to stand from the seated position. Ten seconds after the subject reaches a balanced standing position, tell them to turn the KinetaMap off in order to stop data collection.

Repeat this procedure 10 times with the subject's feet positioned shoulder-width apart (Figure 86), giving the subject at least 3 minutes rest between trials and at least 2 hours rest every 5 trials. If the subject is capable of standing with their feet in tandem, have the subject repeat the same test except rising from the chair with their feet in tandem (Figure 87). The test should be repeated 10 times with at least 5 minutes rest between trials and at least 2 hours rest every 5 trials. If the subject is unable to stand or perform the STS with feet in tandem, the STS should be repeated 20 times with feet shoulder-width apart, with 3 minutes rest between trials and at least 2 hours rest every 5 trials. After each STS trial, ask the subject to rate their comfort

on a scale from 1-5 (1 being completely unbalanced and 5 being completely balanced). The physical therapist should also rate the patient's balance control on the same scale after each trial.

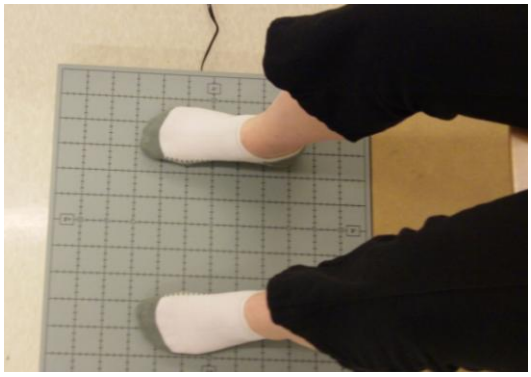


Figure 86: Shoulder-width foot position representing a balanced condition.

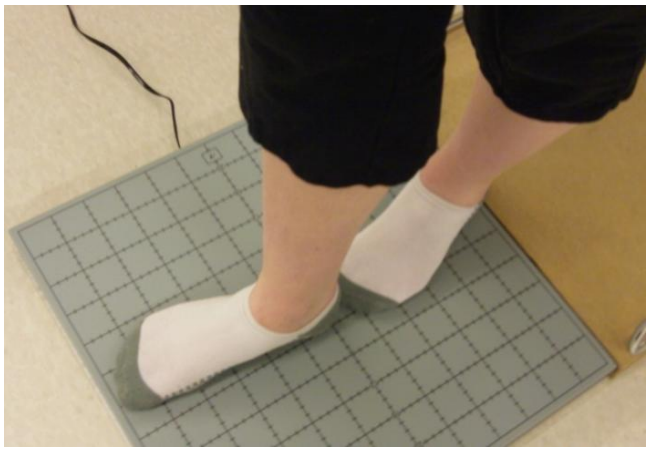


Figure 87: Tandem foot position representing an unbalanced condition.

The KinetaMap collects data at 20Hz and logs each trial in a Microsoft Excel document in terms of time and the X, Y, and Z components of acceleration. The X, Y, and Z components of acceleration need to be converted into m/s^2 using the procedure outlined in (see Appendix J). Subtract the average of the first 5 seconds of acceleration data from the entire data set in order to zero each plot. Calculate the magnitude of the acceleration as outlined in Appendix J for each trial and plot the acceleration as a function of time. Import the NetForce data files into BioAnalysis software to obtain the plots of COP.

8.1.2 Determine the Difference Between Balanced and Unbalanced STS

Next the team then needs to establish the difference between the subject's balanced and unbalanced STS. This will be done by determining the expected time (T_e) above which is considered balanced and equal to or below which is considered unbalanced.

In order to calculate this value, first separate all trials into two groups: one containing all data belonging to trials where the physical therapist rated the STS ≤ 3 (unbalanced) and the other containing all data belonging to trials where the physical therapist rated the STS > 3 (balanced). There should be at least 10 trials in each category to prove that results are significant. If there are less than 10 trials in either category, more data needs to be collected.

Verify that the trials rated ≤ 3 are unbalanced and the trials rated > 3 are balanced by qualitatively comparing the COP plots of each group. Unbalanced trials are characterized by a larger variation of the COP than balanced trials (Figure 88).

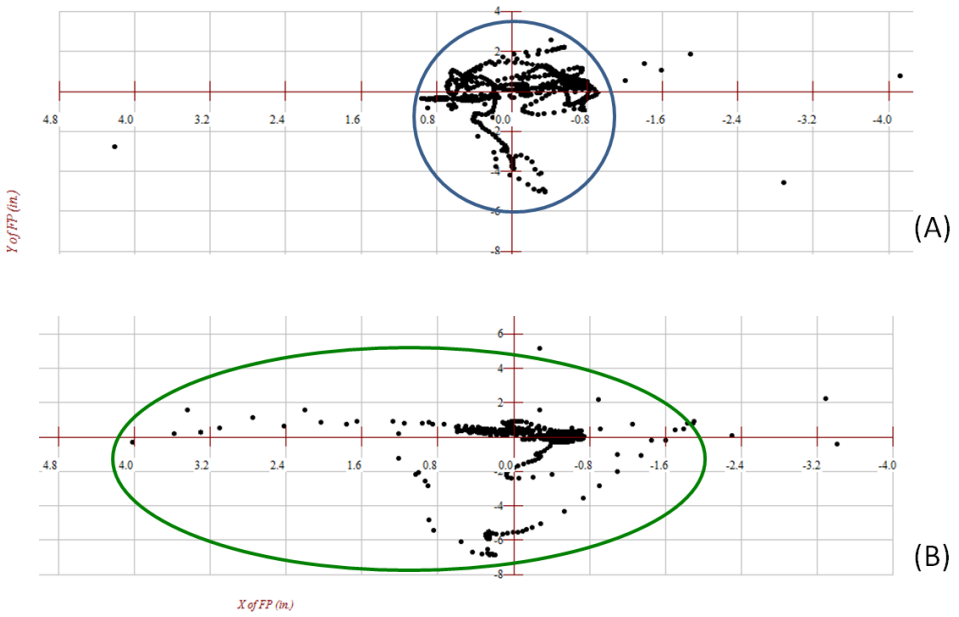


Figure 88: a) COP plot for a balanced STS b) COP plot for an unbalanced STS.

After verifying that all trials are properly categorized as balanced or unbalanced, the team needs to evaluate the acceleration plots. Each acceleration plot should have a positive and negative amplitude. For each plot, calculate the time between the maximum peak and minimum peak. Calculate the average time between peaks for balanced trials (STS > 3) and unbalanced trials (rated ≤ 3) and the corresponding standard deviation. Calculate the time range for a balanced STS and unbalanced STS by adding and subtracting the standard deviation from the corresponding mean. For example, if the mean is 5 and the standard deviation is ±1, the range will be 4-6. Choose T_e by identifying the greatest time within the unbalanced time range that is not included in the balanced time range. For example, if the balanced time range is 5.5-7.3 and the unbalanced time range is 3.5-6, T_e would be 5. Insert the new T_e into the code as explained in Section 5.4.

8.1.3 Verify the device

To verify the design, attach both the KinetaMap and the new device to the belt such that when the belt is attached to subject 1 the new device is positioned at the right hip and the KinetaMap is positioned at the left hip. With the belt attached, conduct the same testing procedure as was performed in preliminary testing. During all trails, observers need to note whether the device buzzes. And after each trial, the physical therapist and subject should rate the comfort level as done in preliminary testing.

Using data from the KinetaMap, plot the magnitude of the acceleration for each trial and calculate the time between the positive and negative peaks. Label the trial unbalanced if the calculated time was less than or equal to T_e and it should have buzzed. Label the trial balanced if the calculated time was greater than T_e and it should not have buzzed. For each trial, label what actually happened during the test (buzz or no buzz). For balanced trials calculate the percentage

of times the device did not buzz and for unbalanced trials calculate the percentage of times it did buzz.

Results should be at least 90% accurate for balanced trials, meaning that the device did not incorrectly buzz for more than 10% of the trials. Results should be at least 80% accurate for unbalanced trials, meaning that the device did not buzz for more than 20% of the trials when it should have buzzed. If the device achieves the accuracy rates, then the device was successful at identifying a balanced and unbalanced STS activity and thus notifying the user of their balance control.

9 CONCLUSIONS AND FUTURE RECOMMENDATIONS

By creating a device that can proactively monitor balance control in the elderly population and utilize a repetitive daily activity, we have impacted the ability of the elderly to become aware of their own balance control and helped them to prevent falls and fall injuries. By developing a device that is low cost and can be used with no professional intervention, we have enabled the elderly to frequently observe their own balance control from their home without the requirement of transportation to a clinical setting the without incurring the large expense of clinical visits and post-fall hospitalization. By developing a device that is simple to use, with no outward complexity beyond a simple toggle switch and status indicator, we have developed a device that will be unintimidating to the elderly and promote frequent use. Therefore, a device of this nature bridges the void between existing balance control indicators and in-home fall monitoring devices for the elderly. Combining simplicity of operation present in the fall indicators and the proactive nature of the balance control indicators, the device creates a simplistic means for proactive balance control monitoring. This addresses issues with devices not catered to the elderly population, e.g. Wii Fit, and those requiring professional intervention, e.g. the iShoe, and those which do not proactively monitor balance control or force user rehabilitation, e.g. the Phillips Lifeline pendant and myHalo fall monitoring system. This system, using principles of monitoring acceleration of COM and how it relates to the stability limits of the user, implements a simple means for balance analysis that has been proven through testing to be as effective as a conventional gait monitoring system for the process of monitoring the sit-to-stand motion. This eliminated the need for costly motion detection systems, markers, and force plates for the user to conduct a simple at-home balance test.

For future development of the device, a larger population study would improve the viability of the time duration between peaks idea. With a larger sample population including the elderly, more support could be made for the observed pattern of a longer time duration between peaks in acceleration. This would also confirm that this pattern was observable for elderly subjects. With a larger population, it could be either confirmed or refuted that there is significant variability between elderly subjects necessitating a reprogramming of the device between subjects.

Additionally, improvements could be made to the device to make it more effective in a home or clinical setting. For instance, the added capability of the device to log number of STS performed and the number of balanced/unbalanced trials and the accessibility of this information by a clinician would help a clinician or physical therapist to be sure that a subject was performing the activity and reporting correct results. Finally, a user interface for the computer to the device could allow not only for better clinician accessibility of this information, but also for an easier means to program and reprogram the device based on the changing needs of the user. Such device/personal computer communication would allow for greater flexibility of the device in that it could be reprogrammed within a clinician's office and used during physical examinations for multiple elderly clients.

GLOSSARY

Base of Support (BOS)- The area of the body that is in contact with the support surface (Gielo-Perczak, 2009).

Bluetooth- An open wireless protocol for exchanging data over short distances (using short length radio waves) from fixed and mobile devices.

Center of Mass (COM)- In a uniform gravity field to represent the unique point in an object or system which can be used to describe the system's response to external forces and torques (Nave, 2010).

Center of Pressure (COP)- The point on a body where the total sum of the aerodynamic pressure field acts, causing a force and no moment about that point (Hurt, 1965).

Dorsiflexion- Flexion of the foot in an upward direction (Encyclopedia, 2010).

Inverted Pendulum Model- A pendulum with a mass above its pivot point.

Personal Emergency Response Systems (PERS)- An electronic device designed to let the user summon help in an emergency (Federal Trade Commission, 2002).

Post-fall anxiety Syndrome- Post-fall syndrome is commonly observed in geriatric medicine, affecting near one out of five fallers. Left untreated, this condition can lead to a regressive syndrome, with physical, psychological and social consequences. To avoid such an evolution, specific physical therapy must be proposed as soon as possible (Morisod & Coutaz, 2007).

Pressure Ulcers- An area of skin that breaks down when the patient remains in one position for too long without shifting weight.

Proprioceptive System- The sense of the orientation of one's limbs in space (Anissimov, 2010).

Mechanoreceptors- A sensory receptor that responds to mechanical pressure or distortion.

Mediolateral Movement- movement along the frontal plane of a body (i.e. lateral sway) (Gielo-Perczak, 2009).

Microelectromechanical Systems (MEMS)- the integration of mechanical elements, sensors, actuators, and electronics on a common silicon substrate through microfabrication technology. micromechanical components are fabricated using compatible "micromachining" processes that selectively etch away parts of the silicon wafer or add new structural layers to form the mechanical and electromechanical devices (Reithel, 2010).

Somatosensory System- A diverse sensory system comprising the receptors and processing centres to produce the sensory modalities such as touch, temperature, proprioception (body position), and nociception (pain) (Boulpaep & Boron, 2003).

Tai Chi- A meditative form of gentle stretching and postural changes in a slow and flowing manner (Rose, 2005).

Vestibular System- A sensory system in mammals that determines body position with respect to gravity and orientation with respect to self-generated movements. It allows for the transmission of information that allows for compensatory movement and adjustment in body positioning (Gray, 2000).

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Appendix A: ADXL 345 Data Sheet

ANALOG DEVICES

3-Axis, ±2 *g*/±4 *g*/±8 *g*/±16 *g* Digital Accelerometer

ADXL345

FEATURES

- Ultralow power: as low as 40 µA in measurement mode and 0.1 µA in standby mode at V_S = 2.5 V (typical)
- Power consumption scales automatically with bandwidth
- User-selectable resolution
  - Fixed 10-bit resolution
  - Full resolution, where resolution increases with *g* range, up to 13-bit resolution at ±16 *g* (maintaining 4 mg/LSB scale factor in all *g* ranges)
- Embedded, patent pending FIFO technology minimizes host processor load
- Tap/double tap detection
- Activity/inactivity monitoring
- Free-fall detection
- Supply voltage range: 2.0 V to 3.6 V
- I/O voltage range: 1.7 V to V_S
- SPI (3- and 4-wire) and I^2C digital interfaces
- Flexible interrupt modes mappable to either interrupt pin
- Measurement ranges selectable via serial command
- Bandwidth selectable via serial command
- Wide temperature range (−40°C to +85°C)
- 10,000 *g* shock survival
- Pb free/RoHS compliant
- Small and thin: 3 mm × 5 mm × 1 mm LGA package

APPLICATIONS

- Handsets
- Medical instrumentation
- Gaming and pointing devices
- Industrial instrumentation
- Personal navigation devices
- Hard disk drive (HDD) protection
- Fitness equipment

GENERAL DESCRIPTION

The ADXL345 is a small, thin, low power, 3-axis accelerometer with high resolution (13-bit) measurement at up to ±16 *g*. Digital output data is formatted as 16-bit twos complement and is accessible through either a SPI (3- or 4-wire) or I^2C digital interface.

The ADXL345 is well suited for mobile device applications. It measures the static acceleration of gravity in tilt-sensing applications, as well as dynamic acceleration resulting from motion or shock. Its high resolution (4 mg/LSB) enables measurement of inclination changes less than 1.0°.

Several special sensing functions are provided. Activity and inactivity sensing detect the presence or lack of motion and if the acceleration on any axis exceeds a user-set level. Tap sensing detects single and double taps. Free-fall sensing detects if the device is falling. These functions can be mapped to one of two interrupt output pins. An integrated, patent pending 32-level first in, first out (FIFO) buffer can be used to store data to minimize host processor intervention.

Low power modes enable intelligent motion-based power management with threshold sensing and active acceleration measurement at extremely low power dissipation.

The ADXL345 is supplied in a small, thin, 3 mm × 5 mm × 1 mm, 14-lead, plastic package.

FUNCTIONAL BLOCK DIAGRAM

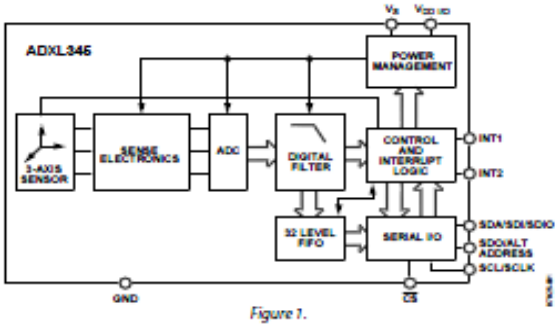

Figure 1.

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Fax: 781.461.3113

SPECIFICATIONS

T_A = 25°C, V_S = 2.5 V, $V_{DD\ I/O}$ = 1.8 V, acceleration = 0 *g*, C_S = 1 µF tantalum, C_{ID} = 0.1 µF, unless otherwise noted.

Table 1. Specifications[1]

Parameter	Test Conditions	Min	Typ	Max	Unit
SENSOR INPUT	Each axis				
Measurement Range	User selectable		±2, ±4, ±8, ±16		*g*
Nonlinearity	Percentage of full scale		±0.5		%
Inter-Axis Alignment Error			±0.1		Degrees
Cross-Axis Sensitivity[2]			±1		%
OUTPUT RESOLUTION	Each axis				
All *g* Ranges	10-bit resolution		10		Bits
±2 *g* Range	Full resolution		10		Bits
±4 *g* Range	Full resolution		11		Bits
±8 *g* Range	Full resolution		12		Bits
±16 *g* Range	Full resolution		13		Bits
SENSITIVITY	Each axis				
Sensitivity at X_{OUT}, Y_{OUT}, Z_{OUT}	±2 *g*, 10-bit or full resolution	232	256	286	LSB/*g*
Scale Factor at X_{OUT}, Y_{OUT}, Z_{OUT}	±2 *g*, 10-bit or full resolution	3.5	3.9	4.3	mg/LSB
Sensitivity at X_{OUT}, Y_{OUT}, Z_{OUT}	±4 *g*, 10-bit resolution	116	128	143	LSB/*g*
Scale Factor at X_{OUT}, Y_{OUT}, Z_{OUT}	±4 *g*, 10-bit resolution	7.0	7.8	8.6	mg/LSB
Sensitivity at X_{OUT}, Y_{OUT}, Z_{OUT}	±8 *g*, 10-bit resolution	58	64	71	LSB/*g*
Scale Factor at X_{OUT}, Y_{OUT}, Z_{OUT}	±8 *g*, 10-bit resolution	14.0	15.6	17.2	mg/LSB
Sensitivity at X_{OUT}, Y_{OUT}, Z_{OUT}	±16 *g*, 10-bit resolution	29	32	36	LSB/*g*
Scale Factor at X_{OUT}, Y_{OUT}, Z_{OUT}	±16 *g*, 10-bit resolution	28.1	31.2	34.3	mg/LSB
Sensitivity Change Due to Temperature			±0.01		%/°C
0 *g* BIAS LEVEL	Each axis				
0 *g* Output for X_{OUT}, Y_{OUT}		−150	±40	+150	mg
0 *g* Output for Z_{OUT}		−250	±80	+250	mg
0 *g* Offset vs. Temperature for x-, y-Axes			±0.8		mg/°C
0 *g* Offset vs. Temperature for z-Axis			±4.5		mg/°C
NOISE PERFORMANCE					
Noise (x-, y-Axes)	Data rate = 100 Hz for ±2 *g*, 10-bit or full resolution		<1.0		LSB rms
Noise (z-Axis)	Data rate = 100 Hz for ±2 *g*, 10-bit or full resolution		<1.5		LSB rms
OUTPUT DATA RATE AND BANDWIDTH	User selectable				
Measurement Rate[3]		6.25		3200	Hz
SELF-TEST[4]	Data rate ≥ 100 Hz, 2.0 V ≤ V_S ≤ 3.6 V				
Output Change in x-Axis		0.20		2.10	*g*
Output Change in y-Axis		−2.10		−0.20	*g*
Output Change in z-Axis		0.30		3.40	*g*
POWER SUPPLY					
Operating Voltage Range (V_S)		2.0	2.5	3.6	V
Interface Voltage Range ($V_{DD\ I/O}$)	V_S ≤ 2.5 V	1.7	1.8	V_S	V
	V_S ≥ 2.5 V	2.0	2.5	V_S	V
Supply Current	Data rate > 100 Hz		145		µA
	Data rate < 10 Hz		40		µA
Standby Mode Leakage Current			0.1	2	µA
Turn-On Time[5]	Data rate = 3200 Hz		1.4		ms
TEMPERATURE					
Operating Temperature Range		−40		+85	°C
WEIGHT					
Device Weight			20		mg

[1] All minimum and maximum specifications are guaranteed. Typical specifications are not guaranteed.
[2] Cross-axis sensitivity is defined as coupling between any two axes.
[3] Bandwidth is half the output data rate.
[4] Self-test change is defined as the output (*g*) when the SELF_TEST bit = 1 (in the DATA_FORMAT register) minus the output (*g*) when the SELF_TEST bit = 0 (in the DATA_FORMAT register). Due to device filtering, the output reaches its final value after 4 × τ when enabling or disabling self-test, where τ = 1/(data rate).
[5] Turn-on and wake-up times are determined by the user-defined bandwidth. At a 100 Hz data rate, the turn-on and wake-up times are each approximately 11.1 ms. For other data rates, the turn-on and wake-up times are each approximately τ + 1.1 in milliseconds, where τ = 1/(data rate).

ABSOLUTE MAXIMUM RATINGS

Table 2.

Parameter	Rating
Acceleration	
Any Axis, Unpowered	10,000 *g*
Any Axis, Powered	10,000 *g*
V_S	−0.3 V to +3.6 V
$V_{DD I/O}$	−0.3 V to +3.6 V
Digital Pins	−0.3 V to $V_{DD I/O}$ + 0.3 V or 3.6 V, whichever is less
All Other Pins	−0.3 V to +3.6 V
Output Short-Circuit Duration (Any Pin to Ground)	Indefinite
Temperature Range	
Powered	−40°C to +105°C
Storage	−40°C to +105°C

Stresses above those listed under Absolute Maximum Ratings may cause permanent damage to the device. This is a stress rating only; functional operation of the device at these or any other conditions above those indicated in the operational section of this specification is not implied. Exposure to absolute maximum rating conditions for extended periods may affect device reliability.

THERMAL RESISTANCE

Table 3. Package Characteristics

Package Type	θ_{JA}	θ_{JC}	Device Weight
14-Terminal LGA	150°C/W	85°C/W	20 mg

ESD CAUTION

ESD (electrostatic discharge) sensitive device. Charged devices and circuit boards can discharge without detection. Although this product features patented or proprietary protection circuitry, damage may occur on devices subjected to high energy ESD. Therefore, proper ESD precautions should be taken to avoid performance degradation or loss of functionality.

PIN CONFIGURATION AND FUNCTION DESCRIPTIONS

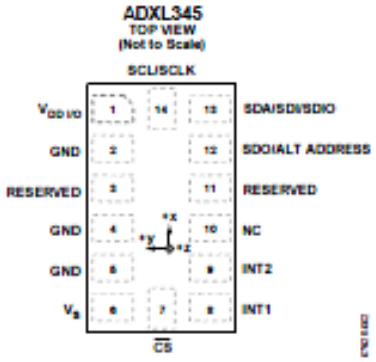


Figure 2. Pin Configuration

Table 4. Pin Function Descriptions

Pin No.	Mnemonic	Description
1	$V_{DD I/O}$	Digital Interface Supply Voltage.
2	GND	Must be connected to ground.
3	Reserved	Reserved. This pin must be connected to V_S or left open.
4	GND	Must be connected to ground.
5	GND	Must be connected to ground.
6	V_S	Supply Voltage.
7	$\overline{CS}$	Chip Select.
8	INT1	Interrupt 1 Output.
9	INT2	Interrupt 2 Output.
10	NC	Not Internally Connected.
11	Reserved	Reserved. This pin must be connected to ground or left open.
12	SDO/ALT ADDRESS	Serial Data Output/Alternate I²C Address Select.
13	SDA/SDI/SDIO	Serial Data (I²C)/Serial Data Input (SPI 4-Wire)/Serial Data Input and Output (SPI 3-Wire).
14	SCL/SCLK	Serial Communications Clock.

THEORY OF OPERATION

The ADXL345 is a complete 3-axis acceleration measurement system with a selectable measurement range of ±2 *g*, ±4 *g*, ±8 *g*, or ±16 *g*. It measures both dynamic acceleration resulting from motion or shock and static acceleration, such as gravity, which allows the device to be used as a tilt sensor.

The sensor is a polysilicon surface-micromachined structure built on top of a silicon wafer. Polysilicon springs suspend the structure over the surface of the wafer and provide a resistance against acceleration forces.

Deflection of the structure is measured using differential capacitors that consist of independent fixed plates and plates attached to the moving mass. Acceleration deflects the beam and unbalances the differential capacitor, resulting in a sensor output whose amplitude is proportional to acceleration. Phase-sensitive demodulation is used to determine the magnitude and polarity of the acceleration.

POWER SEQUENCING

Power can be applied to V_S or $V_{DD\ I/O}$ in any sequence without damaging the ADXL345. All possible power-on modes are summarized in Table 5. The interface voltage level is set with the interface supply voltage, $V_{DD\ I/O}$, which must be present to ensure that the ADXL345 does not create a conflict on the communication bus. For single-supply operation, $V_{DD\ I/O}$ can be the same as the main supply, V_S. In a dual-supply application, however, $V_{DD\ I/O}$ can differ from V_S to accommodate the desired interface voltage, as long as V_S is greater than $V_{DD\ I/O}$.

After V_S is applied, the device enters standby mode, where power consumption is minimized and the device waits for $V_{DD\ I/O}$ to be applied and for the command to enter measurement mode to be received. (This command can be initiated by setting the measure bit in the POWER_CTL register (Address 0x2D).) In addition, any register can be written to or read from to configure the part while the device is in standby mode. It is recommended to configure the device in standby mode and then to enable measurement mode. Clearing the measure bit returns the device to the standby mode.

Table 5. Power Sequencing

Condition	V_S	$V_{DD\ I/O}$	Description
Power Off	Off	Off	The device is completely off, but there is a potential for a communication bus conflict.
Bus Disabled	On	Off	The device is on in standby mode, but communication is unavailable and will create a conflict on the communication bus. The duration of this state should be minimized during power-up to prevent a conflict.
Bus Enabled	Off	On	No functions are available, but the device will not create a conflict on the communication bus.
Standby or Measurement	On	On	At power-up, the device is in standby mode, awaiting a command to enter measurement mode, and all sensor functions are off. After the device is instructed to enter measurement mode, all sensor functions are available.

POWER SAVINGS

Power Modes

The ADXL345 automatically modulates its power consumption in proportion to its output data rate, as outlined in Table 6. If additional power savings is desired, a lower power mode is available. In this mode, the internal sampling rate is reduced, allowing for power savings in the 12.5 Hz to 400 Hz data rate range but at the expense of slightly greater noise. To enter lower power mode, set the LOW_POWER bit (Bit 4) in the BW_RATE register (Address 0x2C). The current consumption in low power mode is shown in Table 7 for cases where there is an advantage for using low power mode. The current consumption values shown in Table 6 and Table 7 are for a V_S of 2.5 V. Current scales linearly with V_S.

Table 6. Current Consumption vs. Data Rate
(T_A = 25°C, V_S = 2.5 V, $V_{DD\ I/O}$ = 1.8 V)

Output Data Rate (Hz)	Bandwidth (Hz)	Rate Code	I_{DD} (μA)
3200	1600	1111	145
1600	800	1110	100
800	400	1101	145
400	200	1100	145
200	100	1011	145
100	50	1010	145
50	25	1001	100
25	12.5	1000	65
12.5	6.25	0111	55
6.25	3.125	0110	40

Table 7. Current Consumption vs. Data Rate, Low Power Mode
(T_A = 25°C, V_S = 2.5 V, $V_{DD\ I/O}$ = 1.8 V)

Output Data Rate (Hz)	Bandwidth (Hz)	Rate Code	I_{DD} (μA)
400	200	1100	100
200	100	1011	65
100	50	1010	55
50	25	1001	50
25	12.5	1000	40
12.5	6.25	0111	40

I²C

With $\overline{CS}$ tied high to $V_{DD\ I/O}$, the ADXL345 is in I²C mode, requiring a simple 2-wire connection as shown in Figure 8. The ADXL345 conforms to the *UM10204 I²C-Bus Specification and User Manual*, Rev. 03—19 June 2007, available from NXP Semiconductor. It supports standard (100 kHz) and fast (400 kHz) data transfer modes if the timing parameters given in Table 11 and Figure 10 are met. Single- or multiple-byte reads/writes are supported, as shown in Figure 9. With the SDO/ALT ADDRESS pin high, the 7-bit I²C address for the device is 0x1D, followed by the R/$\overline{W}$ bit. This translates to 0x3A for a write and 0x3B for a read. An alternate I²C address of 0x53 (followed by the R/$\overline{W}$ bit) can be chosen by grounding the SDO/ALT ADDRESS pin (Pin 12). This translates to 0xA6 for a write and 0xA7 for a read.

If other devices are connected to the same I²C bus, the nominal operating voltage level of these other devices cannot exceed $V_{DD\ I/O}$ by more than 0.3 V. External pull-up resistors, R_P, are necessary for proper I²C operation. Refer to the *UM10204 I²C-Bus Specification and User Manual*, Rev. 03—19 June 2007, when selecting pull-up resistor values to ensure proper operation.

Table 10. I²C Digital Input/Output Voltage

Parameter	Limit[1]	Unit
Digital Input Voltage		
Low Level Input Voltage (V_{IL})	0.25 × $V_{DD\ I/O}$	V max
High Level Input Voltage (V_{IH})	0.75 × $V_{DD\ I/O}$	V min
Digital Output Voltage		
Low Level Output Voltage (V_{OL})[2]	0.2 × $V_{DD\ I/O}$	V max

[1] Limits based on characterization results; not production tested.
[2] The limit given is only for $V_{DD\ I/O}$ < 2 V. When $V_{DD\ I/O}$ > 2 V, the limit is 0.4 V max.

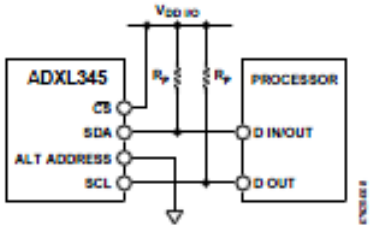
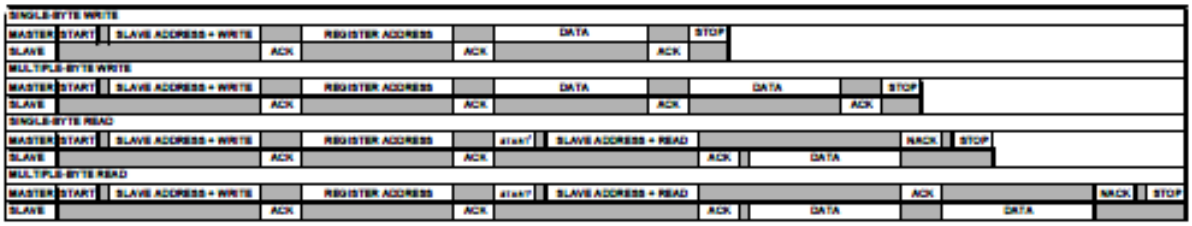


Figure 8. I²C Connection Diagram (Address 0x53)



[1]THIS START IS EITHER A RESTART OR A STOP FOLLOWED BY A START.

NOTES
1. THE SHADED AREAS REPRESENT WHEN THE DEVICE IS LISTENING.

Figure 9. I²C Device Addressing

I²C

With $\overline{CS}$ tied high to $V_{DD\ I/O}$, the ADXL345 is in I²C mode, requiring a simple 2-wire connection as shown in Figure 8. The ADXL345 conforms to the *UM10204 I²C-Bus Specification and User Manual*, Rev. 03—19 June 2007, available from NXP Semiconductor. It supports standard (100 kHz) and fast (400 kHz) data transfer modes if the timing parameters given in Table 11 and Figure 10 are met. Single- or multiple-byte reads/writes are supported, as shown in Figure 9. With the SDO/ALT ADDRESS pin high, the 7-bit I²C address for the device is 0x1D, followed by the R/$\overline{W}$ bit. This translates to 0x3A for a write and 0x3B for a read. An alternate I²C address of 0x53 (followed by the R/$\overline{W}$ bit) can be chosen by grounding the SDO/ALT ADDRESS pin (Pin 12). This translates to 0xA6 for a write and 0xA7 for a read.

If other devices are connected to the same I²C bus, the nominal operating voltage level of these other devices cannot exceed $V_{DD\ I/O}$ by more than 0.3 V. External pull-up resistors, R_P, are necessary for proper I²C operation. Refer to the *UM10204 I²C-Bus Specification and User Manual*, Rev. 03—19 June 2007, when selecting pull-up resistor values to ensure proper operation.

Table 10. I²C Digital Input/Output Voltage

Parameter	Limit[1]	Unit
Digital Input Voltage		
Low Level Input Voltage (V_{IL})	0.25 × $V_{DD\ I/O}$	V max
High Level Input Voltage (V_{IH})	0.75 × $V_{DD\ I/O}$	V min
Digital Output Voltage		
Low Level Output Voltage (V_{OL})[2]	0.2 × $V_{DD\ I/O}$	V max

[1] Limits based on characterization results; not production tested.
[2] The limit given is only for $V_{DD\ I/O}$ < 2 V. When $V_{DD\ I/O}$ > 2 V, the limit is 0.4 V max.

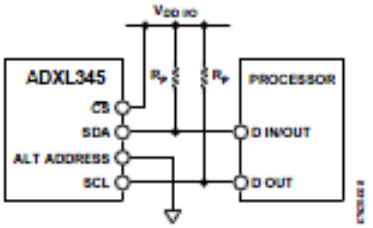


Figure 8. I²C Connection Diagram (Address 0x53)

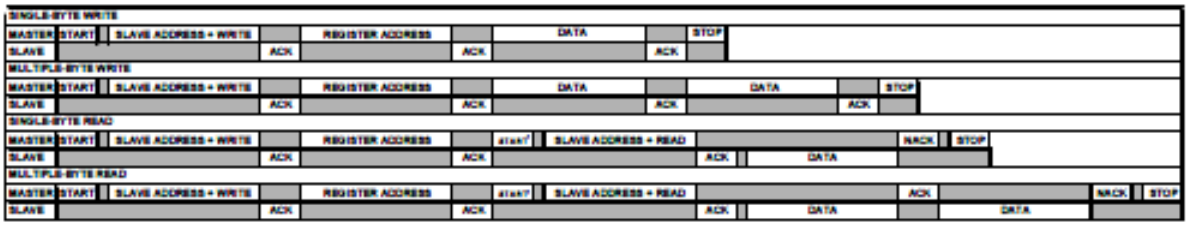


Figure 9. I²C Device Addressing

Table 11. I²C Timing (T_A = 25°C, V_S = 2.5 V, $V_{DD\ I/O}$ = 1.8 V)

Parameter	Limit[1, 2] Min	Max	Unit	Description
f_{SCL}		400	kHz	SCL clock frequency
t_1	2.5		µs	SCL cycle time
t_2	0.6		µs	t_{HIGH}, SCL high time
t_3	1.3		µs	t_{LOW}, SCL low time
t_4	0.6		µs	$t_{HD;STA}$, start/repeated start condition hold time
t_5	350		ns	$t_{SU;DAT}$, data setup time
t_6[3, 4, 5, 6]	0	0.65	µs	$t_{HD;DAT}$, data hold time
t_7	0.6		µs	$t_{SU;STA}$, setup time for repeated start
t_8	0.6		µs	$t_{SU;STO}$, stop condition setup time
t_9	1.3		µs	t_{BUF}, bus-free time between a stop condition and a start condition
t_{10}		300	ns	t_R, rise time of both SCL and SDA when receiving
	0		ns	t_R, rise time of both SCL and SDA when receiving or transmitting
t_{11}		250	ns	t_F, fall time of SDA when receiving
		300	ns	t_F, fall time of both SCL and SDA when transmitting
	20 + 0.1 C_b[7]		ns	t_F, fall time of both SCL and SDA when transmitting or receiving
C_b		400	pF	Capacitive load for each bus line

[1] Limits based on characterization results, with f_{SCL} = 400 kHz and a 3 mA sink current; not production tested.
[2] All values referred to the V_{IH} and the V_{IL} levels given in Table 10.
[3] t_6 is the data hold time that is measured from the falling edge of SCL. It applies to data in transmission and acknowledge times.
[4] A transmitting device must internally provide an output hold time of at least 300 ns for the SDA signal (with respect to $V_{IH(min)}$ of the SCL signal) to bridge the undefined region of the falling edge of SCL.
[5] The maximum t_6 value must be met only if the device does not stretch the low period (t_2) of the SCL signal.
[6] The maximum value for t_6 is a function of the clock low time (t_2), the clock rise time (t_{10}), and the minimum data setup time ($t_{5(min)}$). This value is calculated as $t_{6(max)} = t_2 - t_{10} - t_{5(min)}$.
[7] C_b is the total capacitance of one bus line in picofarads.

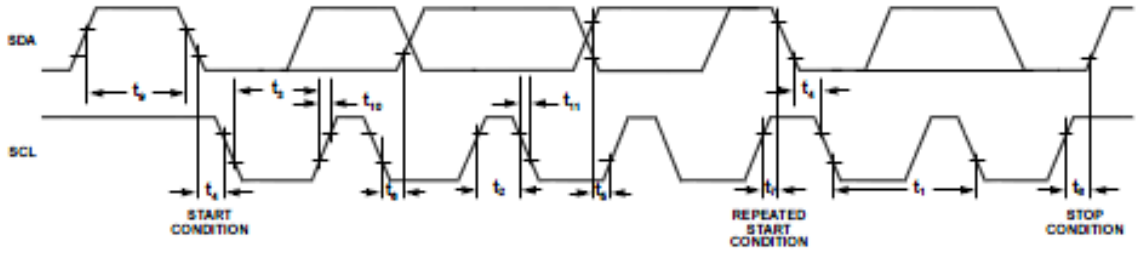


Figure 10. I²C Timing Diagram

Trigger Mode

In trigger mode, FIFO accumulates samples, holding the latest 32 samples from measurements of the x-, y-, and z-axes. After a trigger event occurs and an interrupt is sent to the INT1 or INT2 pin (determined by the trigger bit in the FIFO_CTL register), FIFO keeps the last n samples (where n is the value specified by the samples bits in the FIFO_CTL register) and then operates in FIFO mode, collecting new samples only when FIFO is not full. A delay of at least 5 µs should be present between the trigger event occurring and the start of reading data from the FIFO to allow the FIFO to discard and retain the necessary samples. Additional trigger events cannot be recognized until the trigger mode is reset. To reset the trigger mode, set the device to bypass mode and then set the device back to trigger mode. Note that the FIFO data should be read first because placing the device into bypass mode clears FIFO.

Retrieving Data from FIFO

The FIFO data is read through the DATAX, DATAY, and DATAZ registers (Address 0x32 to Address 0x37). When the FIFO is in FIFO, stream, or trigger mode, reads to the DATAX, DATAY, and DATAZ registers read data stored in the FIFO. Each time data is read from the FIFO, the oldest x-, y-, and z-axes data are placed into the DATAX, DATAY and DATAZ registers.

If a single-byte read operation is performed, the remaining bytes of data for the current FIFO sample are lost. Therefore, all axes of interest should be read in a burst (or multiple-byte) read operation. To ensure that the FIFO has completely popped (that is, that new data has completely moved into the DATAX, DATAY, and DATAZ registers), there must be at least 5 µs between the end of reading the data registers and the start of a new read of the FIFO or a read of the FIFO_STATUS register (Address 0x39). The end of reading a data register is signified by the transition from Register 0x37 to Register 0x38 or by the $\overline{CS}$ pin going high.

For SPI operation at 1.6 MHz or less, the register addressing portion of the transmission is a sufficient delay to ensure that the FIFO has completely popped. For SPI operation greater than 1.6 MHz, it is necessary to deassert the $\overline{CS}$ pin to ensure a total delay of 5 µs; otherwise, the delay will not be sufficient. The total delay necessary for 5 MHz operation is at most 3.4 µs. This is not a concern when using I^2C mode because the communication rate is low enough to ensure a sufficient delay between FIFO reads.

SELF-TEST

The ADXL345 incorporates a self-test feature that effectively tests its mechanical and electronic systems simultaneously. When the self-test function is enabled (via the SELF_TEST bit in the DATA_FORMAT register, Address 0x31), an electrostatic force is exerted on the mechanical sensor. This electrostatic force moves the mechanical sensing element in the same manner as acceleration, and it is additive to the acceleration experienced by the device. This added electrostatic force results in an output change in the x-, y-, and z-axes. Because the electrostatic force is proportional to Vs^2, the output change varies with Vs. The self-test feature of the ADXL345 also exhibits a bimodal behavior that depends on which phase of the clock self-test is enabled. However, the limits shown in Table 1 and Table 12 to Table 15 are valid for all potential self-test values across the entire allowable voltage range. Use of the self-test feature at data rates less than 100 Hz may yield values outside these limits. Therefore, the part should be placed into a data rate of 100 Hz or greater when using self-test.

Table 12. Self-Test Output in LSB for ±2 *g*, Full Resolution

Axis	Min	Max	Unit
X	50	540	LSB
Y	−540	−50	LSB
Z	75	875	LSB

Table 13. Self-Test Output in LSB for ±4 *g*, 10-Bit Resolution

Axis	Min	Max	Unit
X	25	270	LSB
Y	−270	−25	LSB
Z	38	438	LSB

Table 14. Self-Test Output in LSB for ±8 *g*, 10-Bit Resolution

Axis	Min	Max	Unit
X	12	135	LSB
Y	−135	−12	LSB
Z	19	219	LSB

Table 15. Self-Test Output in LSB for ±16 *g*, 10-Bit Resolution

Axis	Min	Max	Unit
X	6	67	LSB
Y	−67	−6	LSB
Z	10	110	LSB

REGISTER MAP

Table 16. Register Map

Address					
Hex	Dec	Name	Type	Reset Value	Description
0x00	0	DEVID	R	11100101	Device ID.
0x01 to 0x1C	1 to 28	Reserved			Reserved. Do not access.
0x1D	29	THRESH_TAP	R/$\overline{W}$	00000000	Tap threshold.
0x1E	30	OFSX	R/$\overline{W}$	00000000	X-axis offset.
0x1F	31	OFSY	R/$\overline{W}$	00000000	Y-axis offset.
0x20	32	OFSZ	R/$\overline{W}$	00000000	Z-axis offset.
0x21	33	DUR	R/$\overline{W}$	00000000	Tap duration.
0x22	34	Latent	R/$\overline{W}$	00000000	Tap latency.
0x23	35	Window	R/$\overline{W}$	00000000	Tap window.
0x24	36	THRESH_ACT	R/$\overline{W}$	00000000	Activity threshold.
0x25	37	THRESH_INACT	R/$\overline{W}$	00000000	Inactivity threshold.
0x26	38	TIME_INACT	R/$\overline{W}$	00000000	Inactivity time.
0x27	39	ACT_INACT_CTL	R/$\overline{W}$	00000000	Axis enable control for activity and inactivity detection.
0x28	40	THRESH_FF	R/$\overline{W}$	00000000	Free-fall threshold.
0x29	41	TIME_FF	R/$\overline{W}$	00000000	Free-fall time.
0x2A	42	TAP_AXES	R/$\overline{W}$	00000000	Axis control for tap/double tap.
0x2B	43	ACT_TAP_STATUS	R	00000000	Source of tap/double tap.
0x2C	44	BW_RATE	R/$\overline{W}$	00001010	Data rate and power mode control.
0x2D	45	POWER_CTL	R/$\overline{W}$	00000000	Power-saving features control.
0x2E	46	INT_ENABLE	R/$\overline{W}$	00000000	Interrupt enable control.
0x2F	47	INT_MAP	R/$\overline{W}$	00000000	Interrupt mapping control.
0x30	48	INT_SOURCE	R	00000010	Source of interrupts.
0x31	49	DATA_FORMAT	R/$\overline{W}$	00000000	Data format control.
0x32	50	DATAX0	R	00000000	X-Axis Data 0.
0x33	51	DATAX1	R	00000000	X-Axis Data 1.
0x34	52	DATAY0	R	00000000	Y-Axis Data 0.
0x35	53	DATAY1	R	00000000	Y-Axis Data 1.
0x36	54	DATAZ0	R	00000000	Z-Axis Data 0.
0x37	55	DATAZ1	R	00000000	Z-Axis Data 1.
0x38	56	FIFO_CTL	R/$\overline{W}$	00000000	FIFO control.
0x39	57	FIFO_STATUS	R	00000000	FIFO status.

Register 0x32 to Register 0x37—DATAX0, DATAX1, DATAY0, DATAY1, DATAZ0, DATAZ1 (Read Only)

These six bytes (Register 0x32 to Register 0x37) are eight bits each and hold the output data for each axis. Register 0x32 and Register 0x33 hold the output data for the x-axis, Register 0x34 and Register 0x35 hold the output data for the y-axis, and Register 0x36 and Register 0x37 hold the output data for the z-axis. The output data is twos complement, with DATAx0 as the least significant byte and DATAx1 as the most significant byte, where x represent X, Y, or Z. The DATA_FORMAT register (Address 0x31) controls the format of the data. It is recommended that a multiple-byte read of all registers be performed to prevent a change in data between reads of sequential registers.

Register 0x38—FIFO_CTL (Read/Write)

D7	D6	D5	D4	D3	D2	D1	D0
FIFO_MODE		Trigger	Samples				

FIFO_MODE Bits

These bits set the FIFO mode, as described in Table 19.

Table 19. FIFO Modes

Setting			
D7	D6	Mode	Function
0	0	Bypass	FIFO is bypassed.
0	1	FIFO	FIFO collects up to 32 values and then stops collecting data, collecting new data only when FIFO is not full.
1	0	Stream	FIFO holds the last 32 data values. When FIFO is full, the oldest data is overwritten with newer data.
1	1	Trigger	When triggered by the trigger bit, FIFO holds the last data samples before the trigger event and then continues to collect data until full. New data is collected only when FIFO is not full.

Trigger Bit

A value of 0 in the trigger bit links the trigger event of trigger mode to INT1, and a value of 1 links the trigger event to INT2.

Samples Bits

The function of these bits depends on the FIFO mode selected (see Table 20). Entering a value of 0 in the samples bits immediately sets the watermark status bit in the INT_SOURCE register, regardless of which FIFO mode is selected. Undesirable operation may occur if a value of 0 is used for the samples bits when trigger mode is used.

Table 20. Samples Bits Functions

FIFO Mode	Samples Bits Function
Bypass	None.
FIFO	Specifies how many FIFO entries are needed to trigger a watermark interrupt.
Stream	Specifies how many FIFO entries are needed to trigger a watermark interrupt.
Trigger	Specifies how many FIFO samples are retained in the FIFO buffer before a trigger event.

0x39—FIFO_STATUS (Read Only)

D7	D6	D5	D4	D3	D2	D1	D0
FIFO_TRIG	0	Entries					

FIFO_TRIG Bit

A 1 in the FIFO_TRIG bit corresponds to a trigger event occurring, and a 0 means that a FIFO trigger event has not occurred.

Entries Bits

These bits report how many data values are stored in FIFO. Access to collect the data from FIFO is provided through the DATAX, DATAY, and DATAZ registers. FIFO reads must be done in burst or multiple-byte mode because each FIFO level is cleared after any read (single- or multiple-byte) of FIFO. FIFO stores a maximum of 32 entries, which equates to a maximum of 33 entries available at any given time because an additional entry is available at the output filter of the device.

APPENDIX B: KinetaMap User Manual

sparkfun ELECTRONICS

[303]

KinetaMap User Guide

2009.04.01

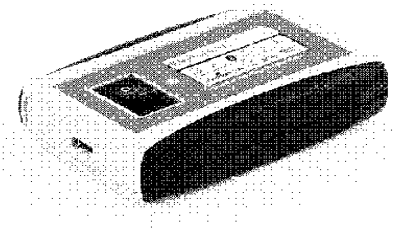
Overview

KinetaMap is a combination of technologies that allow data logging and transmission of GPS location and accelerometer data. Kinesiology is a growing field of research where human movement and motion can be analyzed for various applications. KinetaMap has the ability to capture things like pedestrian gait, vehicle braking, or package handling. The logs give you GPS location and the raw acceleration readings - it's up to you to decipher what they mean!

KinetaMap comes with basic firmware that currently supports acceleration and GPS logging. Flip the power switch and GPS will be logged once per second, and accelerometer readings (X,Y and Z axis) will be logged at up to 100 Hz. We designed KinetaMap to be as flexible and hackable as possible. It has the LPC2148 USB bootloader built-in which allows updates to the firmware easily and quickly over USB.

KinetaMap contains 1GB of flash that will last for weeks of continuous logging. It attaches to a computer over USB and shows up on any operating system as a flash drive. Quickly move files onto or off the KinetaMap without having to remove the flash card. All log files are comma separated text files for easy parsing with Excel, Octave, MatLab, or your favorite data analysis package.

KinetaMap includes a Bluetooth module that supports SPP (serial port profile). This allows a computer to wirelessly connect to and download the log files. With a range of 100m, a computer is capable of querying a KinetaMap located within a vehicle outside. A computer may also wirelessly modify the configuration and settings internally saved within KinetaMap. KinetaMap has also been tested with Bluetooth and Google Maps enabled phones. KinetaMap allows precise location and map directions on your phone!



Specifications

- Dimensions: 3.7x2.5x1.1"(28x63x94mm)
- Weight: 3.6oz(103g)
- Battery: 3.7V 1100mAH LiPo Battery
- Current Draw: 140mA-Bluetooth Active
  110mA-Bluetooth Inactive
  4mA-Sleep Mode

the side of the device is the miniUSB connector. Plugging a USB cable into a powered KinetaMap from a computer will open a Mass Storage Device. The Mass Storage Device allows access to the log files and configuration file, as well as provides the ability to load custom firmware.

Using the KinetaMap Default Firmware

While SparkFun encourages it's customers to create and modify custom code to load onto the KinetaMap, we've also created a new and robust default firmware version that should be able to meet the needs of many applications. This section of the User Guide will describe the new firmware (version 1.2) and give an overview of the flow of the firmware.

Quick Start

If you don't really care how the firmware works, and you just want to get going, well, this section is for you! To get the device working out of the box all you need to do is flip the switch to "ON!" By default all of the bluetooth functions are disabled. The device will begin logging to two files: a data file which contains the accelerometer readings, battery voltage readings and GGA GPS data; and a NMEA data which logs the unedited GGA GPS messages. The accelerometer and battery readings will be logged at 10Hz, and the GPS messages will be logged no faster than 1 Hz. Invalid GPS messages will not be logged. The accelerometer will be set to 2G mode. WAAS will be enabled on the GPS module.

Data will be logged to each file until either the device is turned off, the device runs out of battery power, or a USB cable is plugged in from a computer. To retrieve the data files from the device plug a USB cable into the device from a computer(with the power to the KinetaMap turned on). This will open a Mass Storage Device. The log files are located on this drive and can be downloaded to your hard drive.

You may also open the config.txt file to edit the configuration parameters. Just save your changes on the SD card, and don't alter the file name. The edited parameters will be loaded the next time you turn the KinetaMap on. Don't worry about corrupting the configuration file, a corrupted config file will be detected by the firmware and handled.

Power-up Sequence

Turn the KinetaMap on by flipping the rocker switch to the '-' position. Once the device is powered it will go through the startup sequence. First the KinetaMap will load the configuration settings from the config file (config.txt which is located on the SD card). The firmware checks to see if the config file exists. If the file is not on the card then a config file with default parameters is loaded. If the firmware finds the config file, then the file is read and the parameters are loaded. However, if the firmware reads the config file and detects that invalid settings have been saved, or if the file is not in the correct firmware, than the existing config file is deleted and a new one with default parameters is created. While the KinetaMap is loading the parameters, the tri-color LED on the front of the KinetaMap will be solid GREEN. More information on the config file and it's parameters will be discussed in the Configuration File section of this user guide.

After the parameters are loaded on the KinetaMap, the initialization process takes place. The initialization process largely depends on the loaded parameters. If gps logging or communication is enabled than the GPS module is initialized with the proper parameters; if neither of these options are selected then the GPS module is left off. The Bluetooth module is only turned on and initialized if the "config_menu" setting or the "send_gps_to_bt" parameters are enabled. The accelerometer is initialized to the selected range, and then the firmware configures the internal timer for readings based on the accel_frequency parameter. The Real Time Clock, used to timestamp each log, is initialized to 00:00:00. Once a valid GPS reading has been logged, the RTC is updated to the current time indicated in the GPS message; the time is only updated to GPS time if a GPS function is enabled though. While the KinetaMap is initializing the peripherals, the tri-color LED on the front of the KinetaMap will be solid Blue. (Note: More information on the parameters and how they affect the initialization can be found in the Configuration File section of this user guide.)

Parameter Name	Default Value
config_menu	0(Off)
log_gps	1(On)
log_accel	1(On)
send_gps_to_bt	0(Off)
log_batt	1(On)
log_nmea	1(On)
enable_waas	1(On)
full_scale_accel	0(Off)
enable_rmc	0(Off)
enable_gga	1(On)
accel_frequency	10(Hz)

After the peripherals are initialized the device is almost ready to start it's main routine. Before it can begin, though, it needs to create files to log too if logging is enabled. There are two files that are created in the default settings: the data file and the NMEA file. If any type of logging is enabled than a data file is created. The data file consists of the time, the acceleration values, the battery voltage, and some basic GPS information. Also, if NMEA logging is enabled, an NMEA file is created. The NMEA file saves the unaltered GPS messages from the EM408 module. All of the data logged by the KinetaMap will be logged to these two files. The KinetaMap will create a new set of files each time the device is turned on.

The names of the created files follow a specific format. The data files are all named "KinetaMapXXX.csv," where XXX is replaced by an integer between 0 and 255. When the KinetaMap creates a file, it will start with the name "KinetaMap000.csv." If this file already exists, then the KinetaMap will increment the number following "KinetaMap." It will do this until if finds a filename that doesn't exist yet.

The NMEA files follow the same type of system. All of the NMEA files are named "KinetaMapNMEAXXX.csv," where XXX is replaced by an integer between 0 and 255. There are two things you should be aware of when accessing the files. The first is that the newest files won't necessarily be the ones with the highest number. Let's assume that you've run the KinetaMap 4 times, there would be 8 files on the SD card; one pair of files for each time the KinetaMap was run (assuming your logging both NMEA and data files). The files would be numbered '000' to '003.' Now if you delete the '000' files then the next time the KinetaMap is run, it will create file '000' again.

The other thing to be aware of is that the file numbers for the data file and the NMEA file might not match each other. If you've turned the NMEA logging off and created several data files, then you turn the NMEA logging back on, the file numbers for the NMEA files will be several numbers behind the file numbers for the data files.

Logging Scheme

After finishing the startup procedure the KinetaMap will go into the main logging routine, unless the default parameters have been changed to enable the configuration menu. If the configuration menu has been enabled, the device must receive a command via bluetooth to start logging. This will be covered in depth in the Configuration Menu section of this user guide.

The main logging routine will continue running until either the KinetaMap is turned off, the battery dies or a USB cable is plugged into the device from a computer (or any powered port). The logging routine is drastically affected by the system parameters, however it should be relatively easy to understand how the system parameters affect the logging scheme.

The routine will be continually checking for incoming messages from the GPS module. The GPS module will be sending either or both GGA and RMC messages depending on the enable_gga and enable_rmc parameters. GPS messages are received at 1 Hz intervals. By default, only the GGA messages are enabled. The GPS messages are loaded into two places; the coordinates, along with the altitude if it's a GGA message, are saved to the data file. The entire message is saved directly to the NMEA file.

The routine will also read the accelerometer and battery voltage at the rate defined in the accel_frequency parameter. If the frequency is greater than 1 Hz, there will be more accelerometer readings than GPS messages. This will be reflected in the log file by blank spots in the log. Each log in the data file consists of a timestamp, the accelerometer readings, the battery voltage and the GPS data. GPS data will only be received when valid GPS messages are received from the GPS module. If there is not valid fix, or if the message was corrupted, the GPS data will not be logged. If one of these fields has been disabled in the configuration file, or if there is no new data for the field, the field will be blank for the given log. Every time data is saved to the data file, the tri-color LED will blink; the LED will blink Blue if data has been saved but there wasn't any valid GPS data saved. If valid GPS data is saved, the LED will blink Green. The LED should blink at roughly 1 Hz.

All of the data in the logging sequence is saved to the data file and the NMEA file as long as the parameters are enabled properly. Only one set of files is used for the entire session. A session is only ended if power is no longer provided to the KinetaMap, or if a USB cable is plugged into the device from a powered port.

LED Behavior

Below is a table which explains the behavior of the tri-color LED on the KinetaMap while using the default firmware.

Function	LED Status
Charging	RED
Loading System Parameters	GREEN
Initialization/Waiting for Bluetooth Connection	BLUE
Logging Data (No Valid GPS Fix)	Blinking BLUE (1Hz)
Logging Data (Valid GPS Fix)	Blinking GREEN (1Hz)

Retrieving Data from the Log Files on the SD Card

There are two ways to access the logged data which resides on the SD card. The easiest way to retrieve the data is to plug a USB cable into the KinetaMap (with the power on) from a computer. Doing this will create a Mass Storage Device drive on the computer. The files on the SD card will appear in this drive. These files can just be copied onto your hard drive, or viewed directly from the SD card. You can also modify the configuration file (config.txt) from here.

You can also view the contents of the SD card from the configuration menu. This method will be covered in detail in the Configuration Menu section of this user guide.

If you've let the KinetaMap run a few times with the default firmware and default parameters then two types of file should be on the SD card: a data file (KinetaMapXXX.csv) and a NMEA file (KinetaMapNMEAXXX.csv). There will be one pair of files for each KinetaMap session. A CSV file is a 'Comma Separated Variable' file. These files can be opened in any text editor but are specially created for easy use with spreadsheet programs like Excel and Open Office Calc.

The data file created by KinetaMap will always have the following format, regardless of what parameters are enabled or disabled:

UTC	X	Y	Z	Batt	Fix	Lat.	Lat. Dir.	Long.	Long. Dir.	Altitude	Type
0	2	0	-16	5014							
0	1	0	-17	5014							
1	2	0	-16	5014							
1	2	0	-16	5020							
170937.44					1	4003.9	N	10512.6	W	1485	GGA
170937	2	0	-16	5020							
170937	2	0	-16	5014							

Each log consists of one row of data; if there are no values to be logged for a specific field, that field is left blank in the log. The UTC field is the timestamp. The timestamp is initially set to 00:00:00 (HH:MM:SS); however if GPS functions are enabled this value is updated to the current time once a valid GPS message is received. If GPS functions are not enabled, the UTC will measure the duration of the KinetaMap session.

The X, Y and Z acceleration fields are the acceleration values retrieved from the LIS302 triple axis accelerometer. The acceleration values can fall anywhere in the range of -127 to 127 depending on the current acceleration. The LIS302 accelerometer has two modes: 2G mode and 8G mode. To find the G force of the accelerometer the values in these fields must be scaled with respect to the selected mode. Instructions on how to scale these values can be found in the data sheet of the LIS302.

The Batt. Volt field represents the current battery voltage. This reading is represented in mV. The battery voltage is read every time the accelerometer is queried.

The next 6 fields are for GPS data and the names should explain their function. GPS data is only logged when a new message is received; if there is no new data than the fields are left blank. Also, the altitude is only logged if GGA messages are enabled. Altitude will not be logged when an RMC message is received. The last field in the data file is the Type field; the Type field indicates if the GPS message received was a GGA message or an RMC message.

The NMEA file is much simpler. The only data saved to the NMEA file are the GPS messages. While the data file contains all of the relevant data, it is not easy to use the position data in it's format. If you want to find a path, you must use a scripting program to parse out the GPS data in order to make it meaningful. To make it easier to use the GPS data, the KinetaMap has the option of logging data to a NMEA file. The file contains only the GPS data received from the GPS module. This file can be directly imported to websites like GPSVisualizer (http://www.gpsvisualizer.com) to view position data.

The Configuration File

The configuration file is used during the powerup sequence to load the system parameters for the KinetaMap. To

access the configuration file plug a USB cable from a computer into the KinetaMap and turn the KinetaMap on. When the Mass Storage Device drive opens on the computer you will see the configuration file in the drive. The configuration file is named "kineta_config.txt," and if you open the file in a text editor you will see eleven parameters. You can also view the file from the config menu.

config_menu=0
log_gps=1
log_accel=1
send_gps_to_bt=0
log_batt=1
log_nmea=1
enable_waas=1
full_scale_accel=0
enable_rmc=0
enable_gga=1
accel_frequency(hz)=10

The parameters in the configuration file must be kept in this order with a carriage return separating each parameter. All of the parameters except the "accel_frequency" parameter only have two options, 1 or 0. If the parameter is set to 1 than the function is enabled, if the parameter is set to 0 than the function is disabled. Don't forget, you can also change the parameters from the Config Menu. However if you want to enable the config menu you must first do this by directly editing the config file on the SD card.

As a side note, if the configuration file has somehow been corrupted don't worry about it. The firmware will detect a corrupted configuration file, delete it, and create a new one with default parameters. Also, if you want to revert to the default parameters just delete the configuration file and turn the KinetaMap on without a configuration file on the SD card; a new one will automatically be created.

config_menu parameter (Default: 0)

The config_menu parameter enables/disables the configuration menu. If the configuration menu is enabled the KinetaMap will not begin logging until a command is received from the bluetooth connection; so be careful enabling this parameter. The configuration menu will allow you to manipulate the SD card, edit the system parameters, and perform function checks on the GPS module and LIS302 accelerometer over a bluetooth connection.

If the config_menu and send_gps_to_bt parameters are both disabled, the bluetooth module will be turned off. This will significantly improve the battery life of the KinetaMap.

log_gps parameter (Default: 1)

The log_gps parameter tells the KinetaMap if it should log GPS data or not. By default the log_gps parameter is enabled. If the parameter is changed to 0, the KinetaMap will not save the GPS coordinates to the SD card.

log_accel parameter (Default: 1)

The log_accel parameter tells the KinetaMap if it should log the acceleration values or not. By default the log_accel parameter is enabled. If the parameter is changed to 0, the KinetaMap will not save the acceleration values to the SD card.

send_gps_to_bt parameter (Default: 0)

The send_gps_to_bt parameter tells the KinetaMap that it should send the NMEA messages received from the EM-408 module to the bluetooth module. This means that if an SPP bluetooth connection has been established

between another device and the KinetaMap, the GPS messages will be sent to the connected device (i.e. blackberry). If the config menu is also enabled, then the GPS messages will only be transmitted once the config menu has been exited.

If the config_menu and send_gps_to_bt parameters are both disabled, the bluetooth module will be turned off. This will significantly improve the battery life of the KinetaMap.

log_batt parameter (Default: 1)

The log_batt parameter tells the KinetaMap if it should log the battery voltage value or not. By default the log_batt parameter is enabled. If the parameter is changed to 0, the KinetaMap will not save the battery voltage values to the SD card.

log_nmea parameter(Default: 1)

The log_nmea parameter tells the KinetaMap if a NMEA file should be created in the current session. If this parameter is enabled a NMEA file will be created, and the NMEA GPS messages from the EM-408 module will be saved directly to this file. If the parameter is disabled, a NMEA file will not be created for the session.

enable_waas parameter (Default: 1)

This parameter will enable or disable the WAAS function on the EM-408 GPS module.

full_scale_accel parameter (Default: 0)

The full_scale_accel parameter sets the range of the LIS302 accelerometer. If the parameter is set to 0, the range will be set to 2G mode, if the parameter is set to 1, the range will be set to 8G mode.

enable_rmc parameter (Default: 0)

This parameter enables or disables RMC messages from the GPS module.

enable_gga parameter (Default: 1)

This parameter enables or disables GGA messages from the GPS module.

accel_frequency(Hz) parameter (Default: 10)

The accel_frequency parameter sets the frequency that acceleration and battery values are read. This parameter must be an integer between 1 and 100. 100 Hz is the maximum bandwidth of the LIS302 accelerometer.

NOTE: Testing was performed using a 20 Hz read frequency. Setting the frequency to higher values should work, but may cause adverse affects to the data file logging.

The Configuration Menu

The configuration menu allows the user to manage the SD card, perform function checks on the accelerometer and gps module, and edit the system parameters over a bluetooth connection using the serial port protocol. Instructions on how to connect to the device over bluetooth are provided in the Connecting over Bluetooth section of this user guide. By default, the config_menu parameter in the configuration file is disabled. In order to enable configuration menu you must edit the config.txt file by plugging in a USB cable to the KinetaMap with the power turned on. Then edit the config.txt file to enable the config_menu parameter. More information on editing the configuration file can be found in the Configuration File section of this user guide.

APPENDIX C: Client Interviews

1. *The following is an interview conducted by team member, Liz Tuite. The interviewee is an 82 year-old female. Interview questions are bolded and interviewee comments are not bolded.*

Yes I have, I hate to tell you, but I have. Twice two bad falls.

Can you explain?

Yes I can once I was very careless, Do you want a long story or short story?

Just as long as it needs to be, we can cut it out.

Well, I was up at Hampton Beach, had a nice seafood festival, I was going to meet Joe up there, my son. I was calling him on my cellphone and I couldn't hear him, so I took a walk to get away from the music. I'm talking on the phone and walking and I walked right off the stage. Fell down about two steps and I tore my rotator cuff, very painful. They put the ice on my knees because that's where the blood was. I said, 'no I'll get up myself' you know

And how did it effect you after?

I've suffered with it for a few years. Because I went for PT and got pretty good strength and pretty good range of motion. I went back a second time for PT, and I neglected doing exercises. I know I'm going to do them all my life, but now I'm seriously considering having it repaired, but now I'm 82 years old and I wonder will I need rehab and how am I going to manage to get dressed and take care of myself, so that's a problem.

What's the biggest change you've had in your daily life after the fall?

One of the things I loved was once a month I'd go to mustard seed and volunteer to cook for like 200 people with my church, and I liked being there and I liked being with the crowd and cooking and doing the serving. I can't rotate that arm, I cannot like big bowls, I can't stir the things. They asked me to just do the dishes, they wanted me there, and I said I can't even do that. I can't rotate to dry or wash them. And that was one thing, now it's cooking and carrying bundles there's a lot of things. I can't talk with my hands like I like to, I didn't know how much I did that. I have to sometimes pick up a mug of beer with two hands, and at the end of the day <> can you bring me a mug of beer? And my other fall was with the dog, I was standing talking with my son in law, whatever, all a sudden he turns around to go out in the backyard and the dog goes to chase him, I didn't realize the leash was behind me laying on the ground, picked me right up off the ground, landed on my back, and I had two fractures which weren't found for quite a while by MRI finally of my sacrum. Took so long to heal because I wasn't aware of the fracture, that's been painful.

How has that effected your daily life?

Well, I can't carry heavy things, weighty. When I'm bringing in my bundles, grocery shopping, which I always used to do, I could do it, but it causes pain in my lower back. Even today, I just had 5 lbs of flour, 8 lbs of groceries, and I could feel it. So I tire more easily, and I was always very active. I can't do as much as I used to, I like to be active.

What type of therapy or medicine or PT have you done or tried to do after your falls to help you?

I went to therapy, they had me do exercises, leg exercises, laying down, picking up your butt off the floor, with straight knees, leg raises, laying flat with bent knees. For my arms it was mainly range of motion exercises, *stretches* you know, like rattle? I tried to do more, they'd be like 'no, you're burnt, you can't do any more', and it was like, once you're done, you knew you could not. The hardest one was laying flat on my left side, raising that arm up, gravity working against me. And now, waking up in the morning, I can't use that arm to get my blanket off me. It's been really been debilitating, and it makes me angry. You know, my doctor's tell me how healthy I am, my cholesterol is fine, my everything is fine, no liver problems, I'm just so healthy, I say 'yeah, now if I could just learn to stand on my own two feet I'd be fine.'

Did you ever feel before your falls that you were off balance or you had balance issues?

No, I don't think I was off balance because I've always been a walker. You know, I've walked a lot so I've always been strong. I would have been fine if I hadn't been on the phone, and it was noisy too, so I had to go away from the music.

So, do you think in your case the device I was describing would have helped at all?

In my case? Not really, I think as you use better sense. How many accidents do you think occur from people using poor judgement? I always think about it now, I think I'll always be careful where I walk, where I step.

Before either of your falls, did you know that falls caused so many problems in the elderly?

I did, but frankly, I don't always know I'm as old as I am. But I work nursing homes so I've seen…

I am aware of slippery floors, I've always been, because I've known people who've gotten hurt that way, I've always been aware of scatter rugs, I never have scatter rugs because you can get hurt that way. I'm kind of aware of hazards, even if I'm in someone's grocery store and I see a spill I'll say "young man you'd better get that picked up right away because someone could fall and get hurt"

Do you know of anyone who you think could benefit from the device?

Sure I do, personally… my sister. She complains of being lightheaded and unsteady in the morning when she gets up. I've told her you know, watch your posture and things. I think that's something she could benefit from before she gets out of bed in the morning, you know see how she could use it cautiously, getting out of bed because she's so unsteady.

So Why do you think it would help her, do you think she would use it?

Oh I think she would, you know… it's not troublesome, it's very easy to use, I think it would be very good in nursing home use, you know the people who could use it to make themselves aware. You know, people want to stay well, so it would be nice for them to know if they're unsteady or have a chance of falling so maybe they sit until their head clears, you know, maybe get some fluids in them, change their medicines, or tell the doctor's what times this happens so they can adjust their medicine so they don't have drops in blood pressure or something.

Do you worked at St. Pat's, right? How long did you work there for?

Oh I worked there for… I don't want to say too long. I really enjoyed the old people .No, I was there for at least 5 years, and they've improved a lot of things since then. They have all kinds of sensors and alarms to watch people get out of bed.

Do you think this device would be more useful than anything currently used?

I think it would be, as long as they're well aware, for the old people. I think it would be good for people who are recuperating, you know young people like me who have never had to deal with an issue like this, and then something happens and they need a device like this just to remind them that 'oh maybe I need to sit for a while, maybe I need to slow down until I get my balance,' Sometimes people are medicated for pain to, when they're recuperating, and they're not as steady as they think they are, and maybe it would be useful in that respect, postoperatively or when they're home recuperating.

Do you think if they're forgetful about other things they'd be forgetful about this?

I think they'd forget to put it on, or if they're really confused they'd forget what the buzz meant. But I think if they had a person watching them who could remember to put it on overnight, so that it could buzz in the morning, if it was comfortable enough to be worn all the time.

Do you think they'd remember to turn it on if they could wear it all day, and they sat down would they remember to turn it on before they sat up?

It depends, on how alert they are, their minds. You know, myself, I wouldn't forget, and if I were sick in bed I'd be thankful for it… you know, my sister… people don't want to fall, because they don't want to get fractures, and they know it might be the last straw for independence. If you need this, then you probably need people around you anyway don't you think? I mean, like if people are confused and they're on medicine, and they're home alone, and they might just forget an move quick, and maybe it'd be nice to have it on at night., can you have it on overnight?

Or maybe it could just be on when you're about to get up?

And would it be better if it was on overnight?

I think bedtime would be a good time to put it on, and then have it start working as soon as you get up and get on the move.

Do you think we picked the best way to notify the person?

I think a noise would be best, some quick sound that would alert you to sit for a while.

I'm trying to think of any other times, I think post-op patients, when they're on medication and they're asleep, you get up and you try to do things automatically, you know, you just try to get up, and I think a tone.

Do you think it would be helpful for otherwise healthy people?

Like, if they're mentally well, and physically well, but maybe not so physically well, I think of my sister, you know, she's had a few falls, and I think she can be careless, so I think of Rose.

She can be careless, and get up without thinking. And with her recuperating I think it would be helpful.

What's she recuperating from?

She had a fall, she was up in the middle of the night on a bare floor, she has a fractured tibia. She was feeding the cat in the middle of the night, she was in her bare feet because she has some neuropathy of her feet, she was feeding the cat, she took a terrible fall on some water she spilled. She had had her left knee replaced last year, and she was careful to protect that so she broke her tibia right where it connects to the knee. She had a lot of issues, and MRIs. So she has a special needs boy, so she needs to hurry around in the morning and get him off to school and she's always hurrying. When we went on vacation with her she used to get up and blast around and make the beds, that's Rose.

Do you think before this fall she would have used this device?

No, I think she wouldn't have… because she'd say "I don't do that," you know, because we all think different. I mean my sister would think that way, but not Rose, because we all think different.

Do you think that maybe after her surgery on her knee she was unstable?

She was slower, it was painful walking but she was steady.

Do you think if a Dr. told her to use this she would have?

No, because she was steady.

But if she walks slower than she's probably not steady, she probably had balance problems before.

No, she had pain. I actually shouldn't say that, you know she probably did have some balance problems

Yes, through our research we found that walking slowly is a sign of balance problems.

Rose does have some balance problems, but she's cautious and she's not unsteady, so I don't think the alarm would have helped her. But that had nothing to do with the fall, she slipped on water, you know. I think the alarm would help if she were going to recuperate, you know, just to slow her down.

The purpose, in Rose's case, was that she probably could have had balance problems, even before the surgery to do strengthening and things like that.

Did she have PT after her surgery, she must have?

Yes, a while, and then she had people come to the house and do her exercises with her too. Rose's main issue was that she's always moving too fast, and someone needs to slow her down. You know, I guess that could be useful if you could just use it to help slow people down.

Do you think this would be helpful if we could incorporate it into a physical and have a person just stand up 5 times and if it buzzed they could use that to do physical therapy to improve their balance?

Yes, you know I think that could work. Because I don't think Doctors put enough time into physical therapy in older people, you know, they really don't. The older you get, "it's part of the aging process," that's what everyone says, you know, everything is part of the aging process. I think Physical therapy should almost be something that the doctors do at a certain age or when

they see certain symptoms like instability or weakness. Physical therapists can pick up on a lot of things, you know. And my friend, you know, she has M.S., she's going to start some PT because she has a lot of weaknesses and she's having trouble walking, PT is wonderful for the elderly, and for young people with problems. It's hard to keep it up when you get home though, doing the same thing over and over.

It's one thing we were thinking, you go to physical therapy and then you go home and it's all up to you, and if you had this device you could see how you improved and then keep using it after and see if your balance started to get worse, do you think that's useful?

That's a good point. I know, that's why my shoulder has gotten worse from not doing exercises, I could always walk, I could do that forever, but those exercises just made me crazy. I know I need the strength back for the summer, because raking and mowing the lawn and all those things I like to do… they're going to hurt, you know, they're going to hurt.

Do you think doing those things is going to make it worse?

I do, because I was outside raking one day before vacation and I was trying to get done raking before the sanders and a couple day later it hurt like a son of a gun and I remembered back and I said 'oh that must have hurt.' But now I'm careful, when I rake, I rake with my left hand and the other one just goes with the rake.

Okay, I'm done with the interview, thanks that was great.

2. The following are written responses to an e-mail interview. The interviewee is a physical therapist from Fairlawn Rehabilitation Hospital. Questions are bolded and responses are not bolded.

Would the device be useful in a clinical setting? Home setting? If so, how? If not, why?

yes, probably in a clinical rehab setting... we could use it in addition to balance scales that we use during a PT eval (ie Berg or Tinetti scale) which all can help identify those who are a high fall risk.
it may be too costly for home?! insurances may not buy into it?!

What type of patient do you think this would be used for?

patients who are in a fall risk catagory: Parkinson's, MS, post stroke, general medical elderly pts over the age of 80

Would you change anything about this device?

it seems simple in terms of either off balance with 'sit-to-stand' or not. it is good that it is clear cut.

How could this device benefit an elderly individual?

again, it could identify high fall risk patients; need for a person to get P.T. for balance training, need for an assistive device such as cane or walker

APPENDIX D: Data from Preliminary Testing

Table 19: Time between maximum and minimum peaks of shoulder-width trials (SW: shoulder-width, AVG: average, STDEV: standard deviation, CV: coefficient of variance)

Trial	Subject 1 Time (s)	Subject 2 Time (s)	Subject 3 Time (s)	Subject 4 Time (s)	Subject 5 Time (s)	Subject 6 Time (s)	Subject 7 Time (s)	Subject 8 Time (s)
SW	0.75	0.85	0.6	0.6	0.55	0.6	0.7	0.35
SW	0.8	0.7	0.1	0.1	0.5	0.65	0.65	0.15
SW	0.8	0.8	0.25	0.35	0.1	0.6	0.55	0.25
SW	0.9	0.9	0.6	0.3	0.4	0.4	0.65	0.1
SW	0.95	0.2	0.45	0.2	0.5	0.5	0.7	0.45
SW	0.9	0.3	0.48	0.55	0.2			
SW	0.85	0.85	0.9	0.15	0.55			
SW	1.15	0.95	0.5	0.25	0.5			
SW	0.9	0.9	0.6	0.1	0.1			
SW	0.9	0.65	0.4	0.5	0.4			
SW	0.95				0.5			
SW	0.95							
SW	0.55							
SW	0.8							
SW	0.5							
SW	0.1							
SW	1							
SW	0.85							
SW	0.65							
SW	1							
SW	0.8							
SW	0.95							
SW	0.8							
SW	0.75							
SW	0.95							
SW	0.85							
SW	0.9							
SW	0.95							
SW	1.3							
SW	0.75							
SW	0.85							
SW	0.9							
SW	0.9							
SW	0.85							
AVG	**0.85**	**0.71**	**0.71**	**0.59**	**0.55**	**0.75**	**0.74**	**0.35**
STDEV	**0.20**	**0.26**	**0.14**	**0.28**	**0.16**	**0.05**	**0.10**	**0.06**
CV	**0.23**	**0.37**	**0.19**	**0.48**	**0.29**	**0.07**	**0.13**	**0.17**

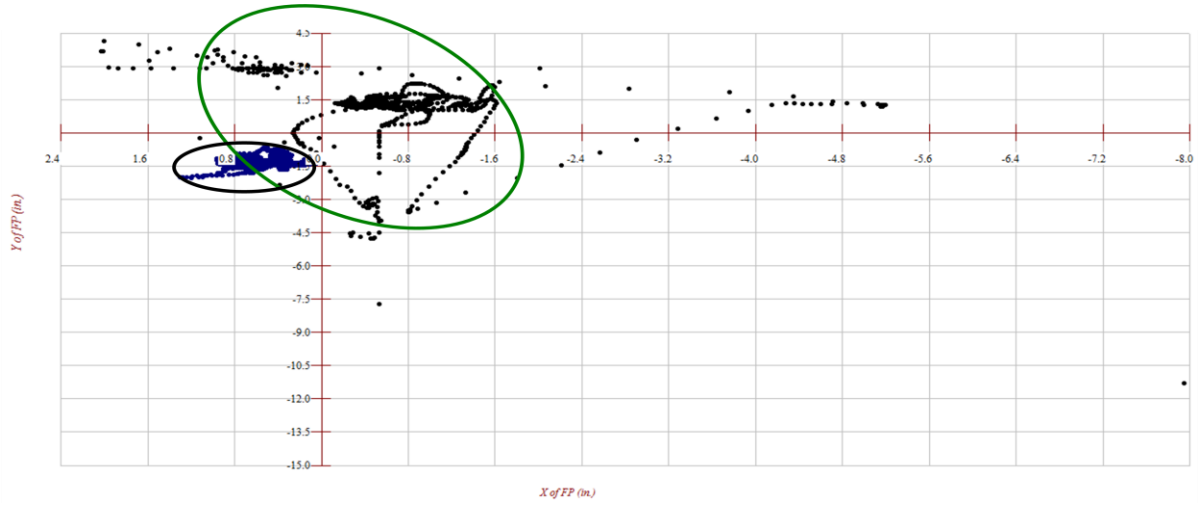
Table 20: Time between maximum and minimum peaks of tandem trials (T: tandem, AVG: average, STDEV: standard deviation, CV: coefficient of variance)

Trial	Subject 1 Time (s)	Subject 2 Time (s)	Subject 3 Time (s)	Subject 4 Time (s)	Subject 5 Time (s)	Subject 6 Time (s)	Subject 7 Time (s)	Subject 8 Time (s)
T	0.3	0.3	0.6	0.6	0.55	0.6	0.7	0.35
T	0.7	0.55	0.1	0.1	0.5	0.65	0.65	0.15
T	0.6	0.65	0.25	0.35	0.1	0.6	0.55	0.25
T	0.55	0.45	0.6	0.3	0.4	0.4	0.65	0.1
T	0.4	0.2	0.45	0.2	0.5	0.5	0.7	0.45
T	0.7	0.35	0.48	0.55	0.2			
T	0.4	0.35	0.9	0.15				
T	0.45	0.7	0.5	0.25				
T	0.3	0.2	0.6	0.1				
T	0.3	0.25	0.4	0.5				
T	0.3							
AVG	**0.45**	**0.40**	**0.49**	**0.31**	**0.41**	**0.55**	**0.65**	**0.26**
STDEV	**0.16**	**0.18**	**0.22**	**0.19**	**0.18**	**0.10**	**0.06**	**0.14**
CV	**0.35**	**0.45**	**0.44**	**0.597718**	**0.44**	**0.18**	**0.09**	**0.55**

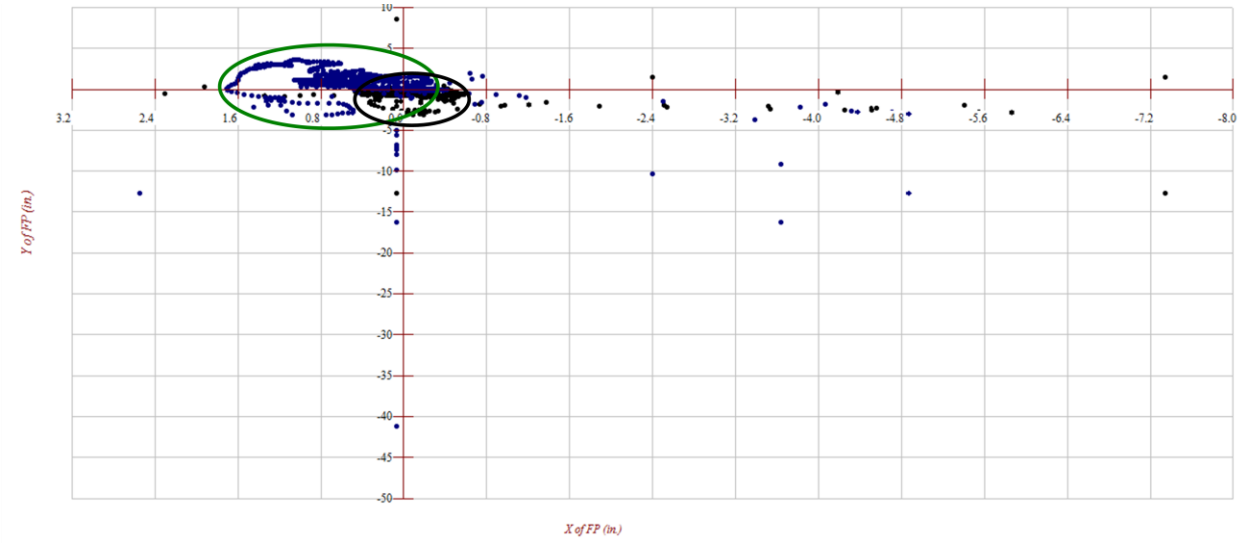
APPENDIX E: Center of Pressure Data from Preliminary Testing

The following shows a representative balanced and unbalanced sit-to-stand trial for subjects 1-4 labeled by a black and green circle, respectively.

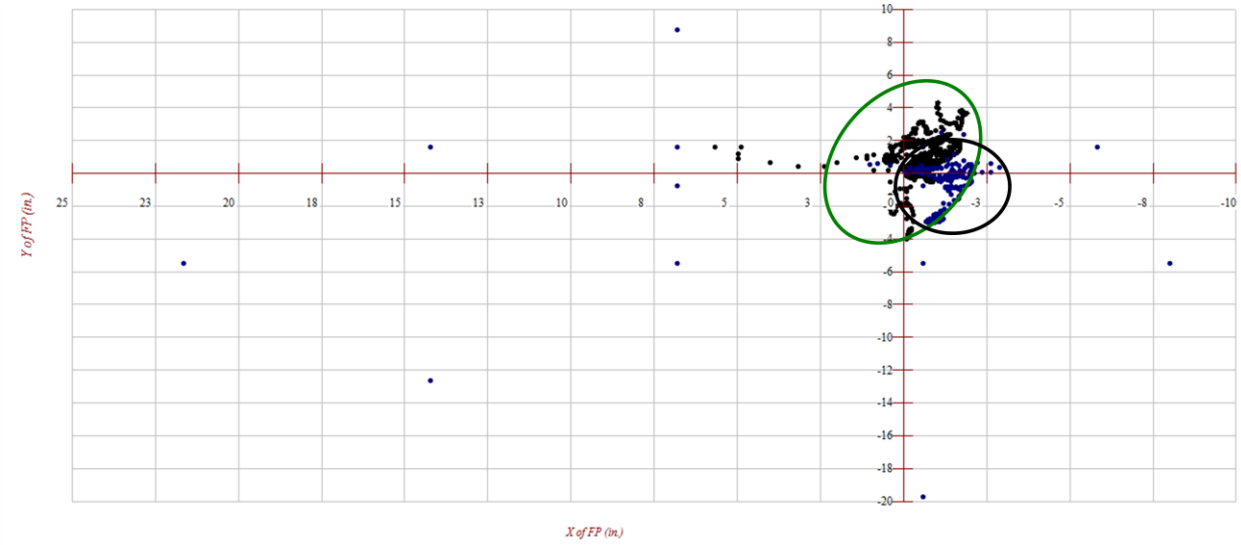
Subject 1



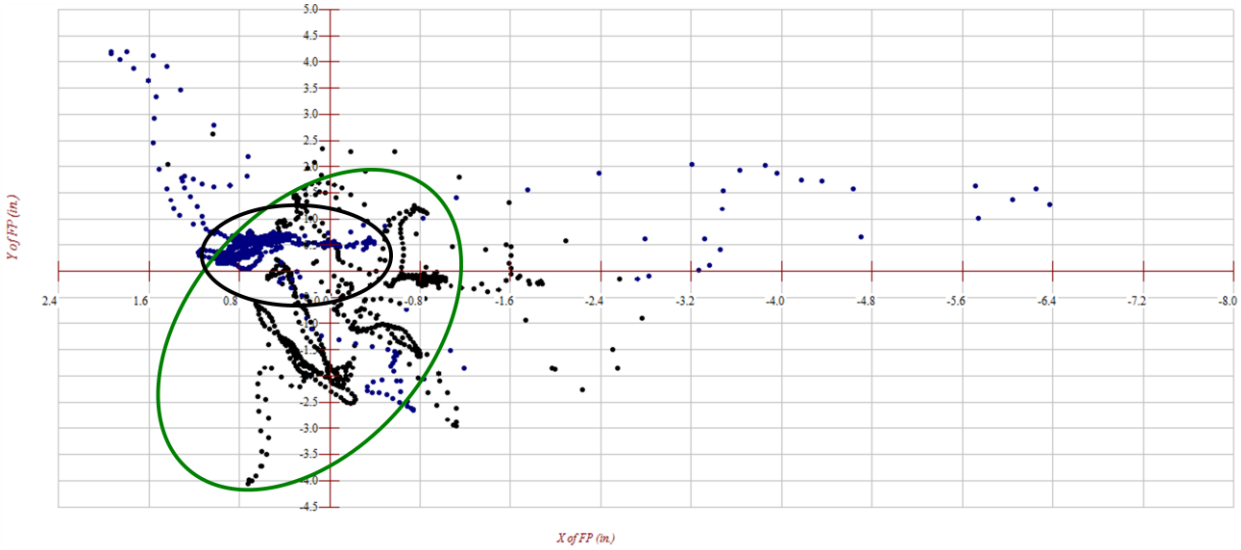
Subject 2



Subject 3



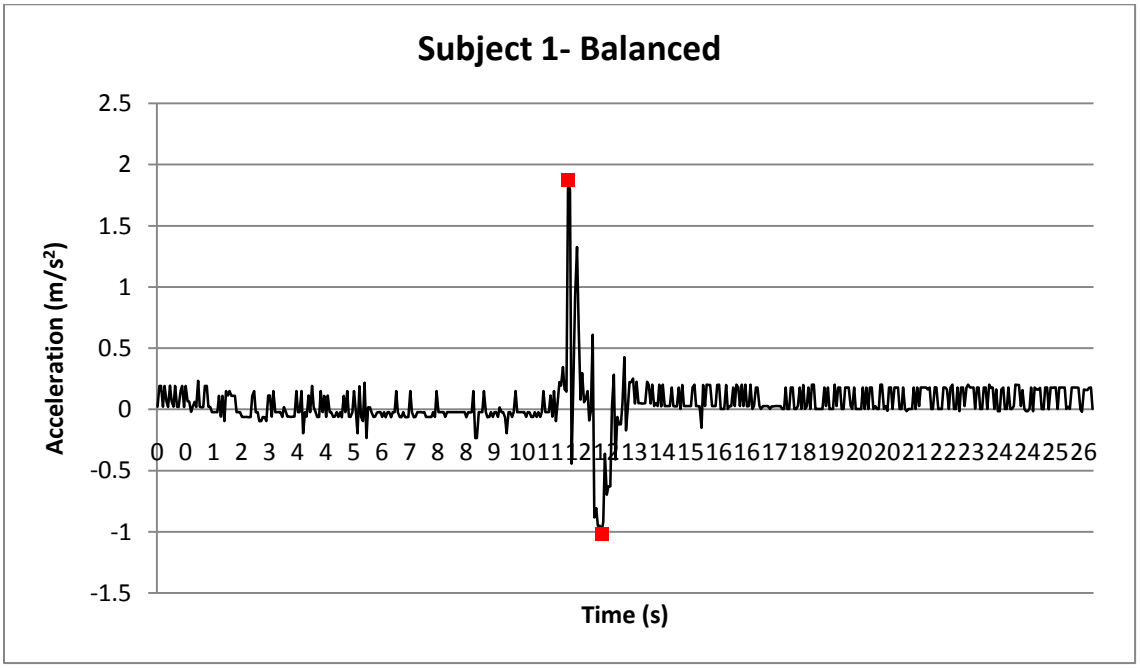
Subject 4

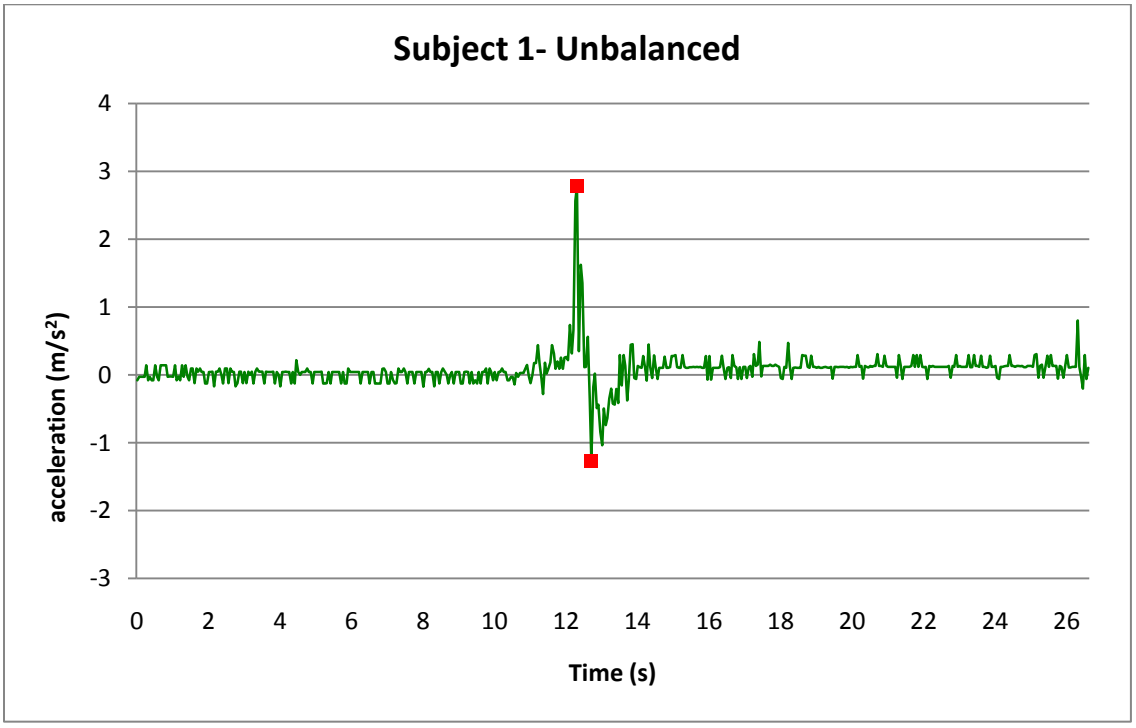


APPENDIX F: Acceleration Plots from Preliminary Testing

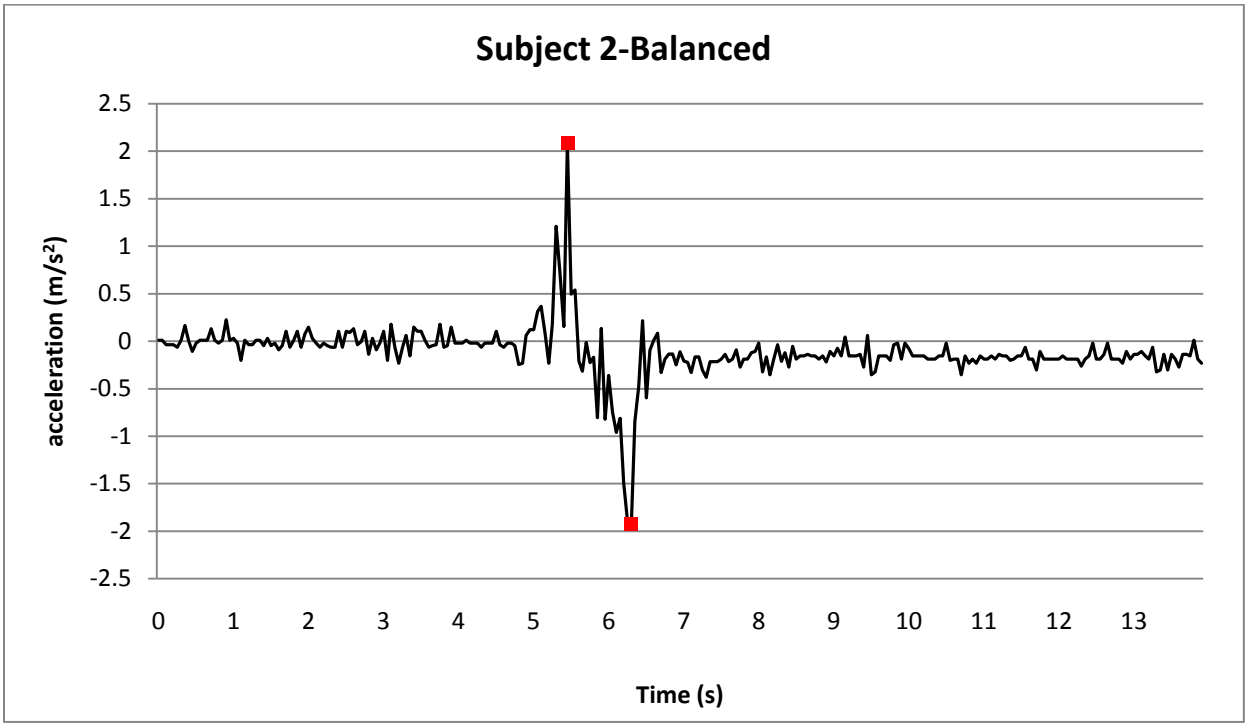
The following show a representative balanced (black) and unbalanced (green) sit-to-stand trial for subjects 1-8.

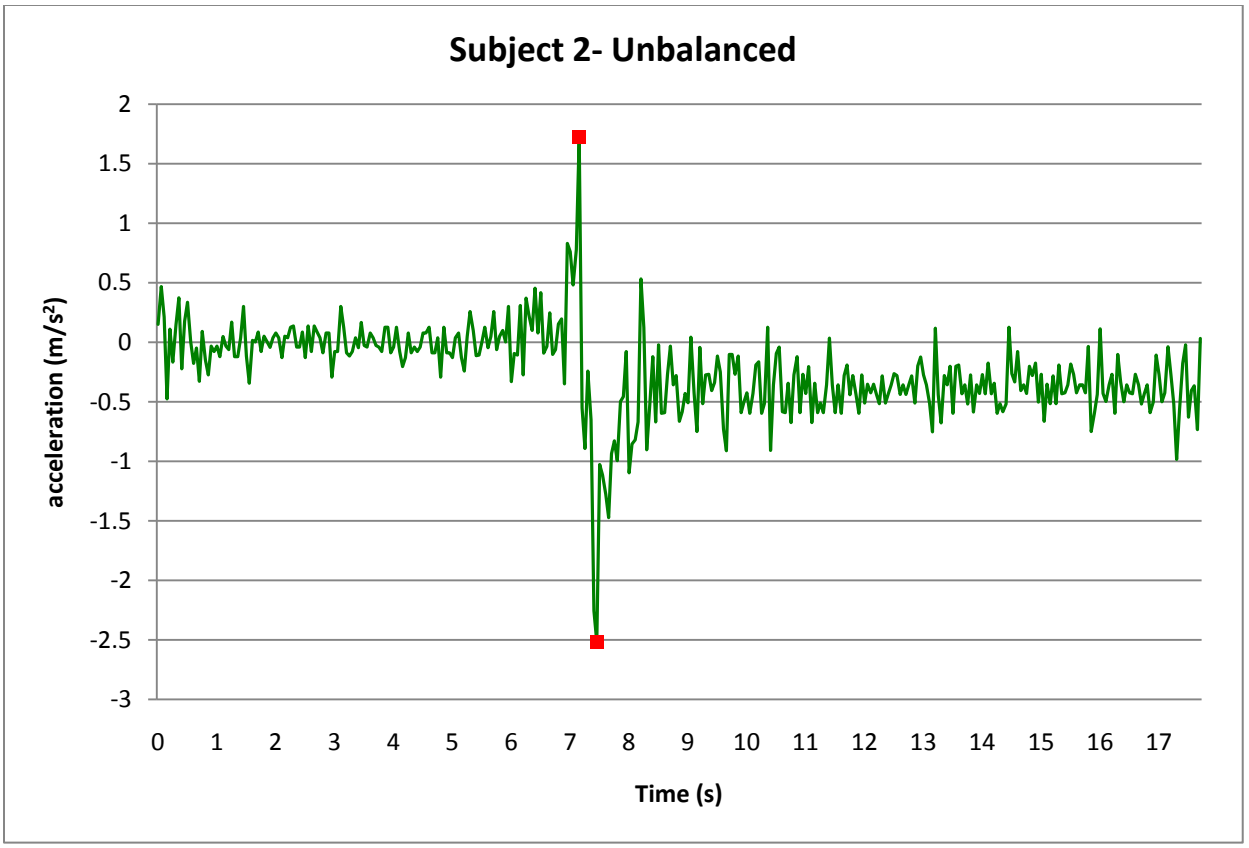
Subject 1



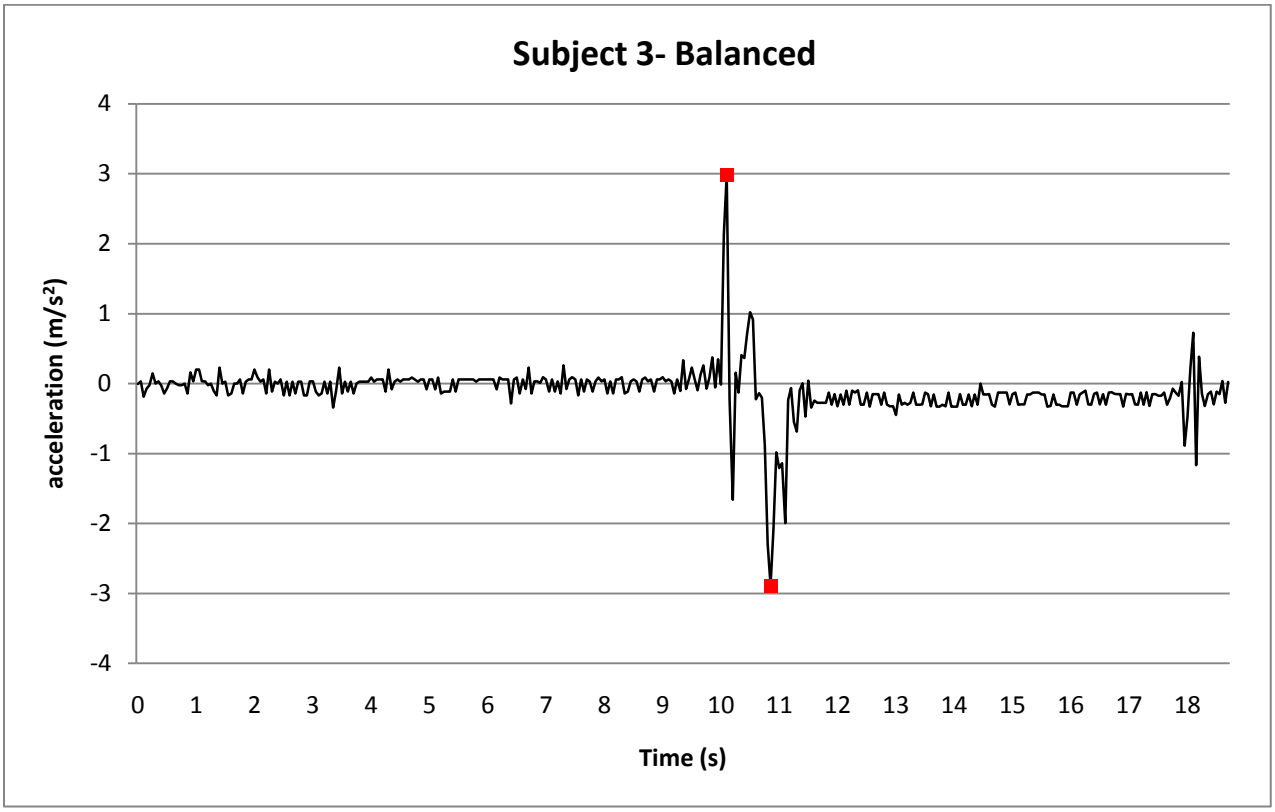


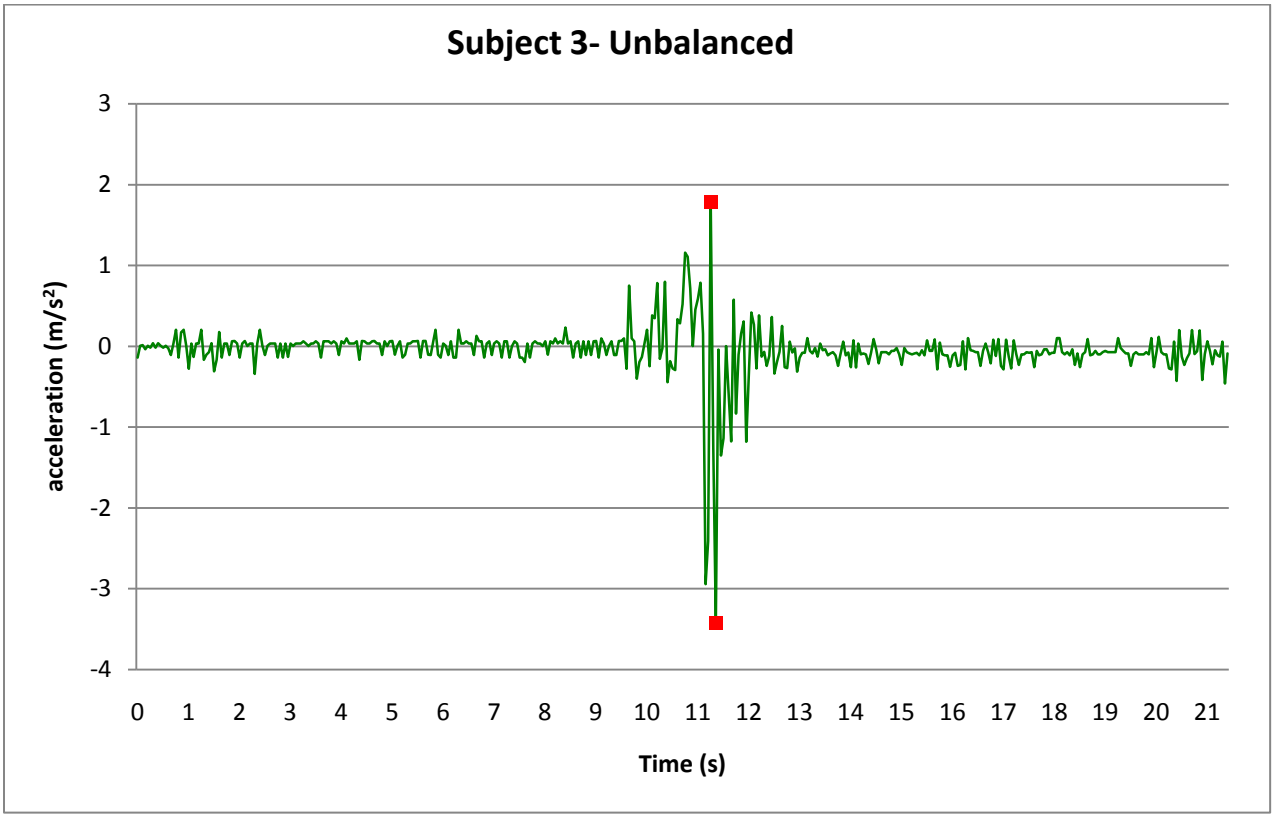
Subject 2


Subject 2-Balanced
acceleration (m/s²)
2.5
2
1.5
1
0.5
0
-0.5
-1
-1.5
-2
-2.5
0 1 2 3 4 5 6 7 8 9 10 11 12 13
Time (s)

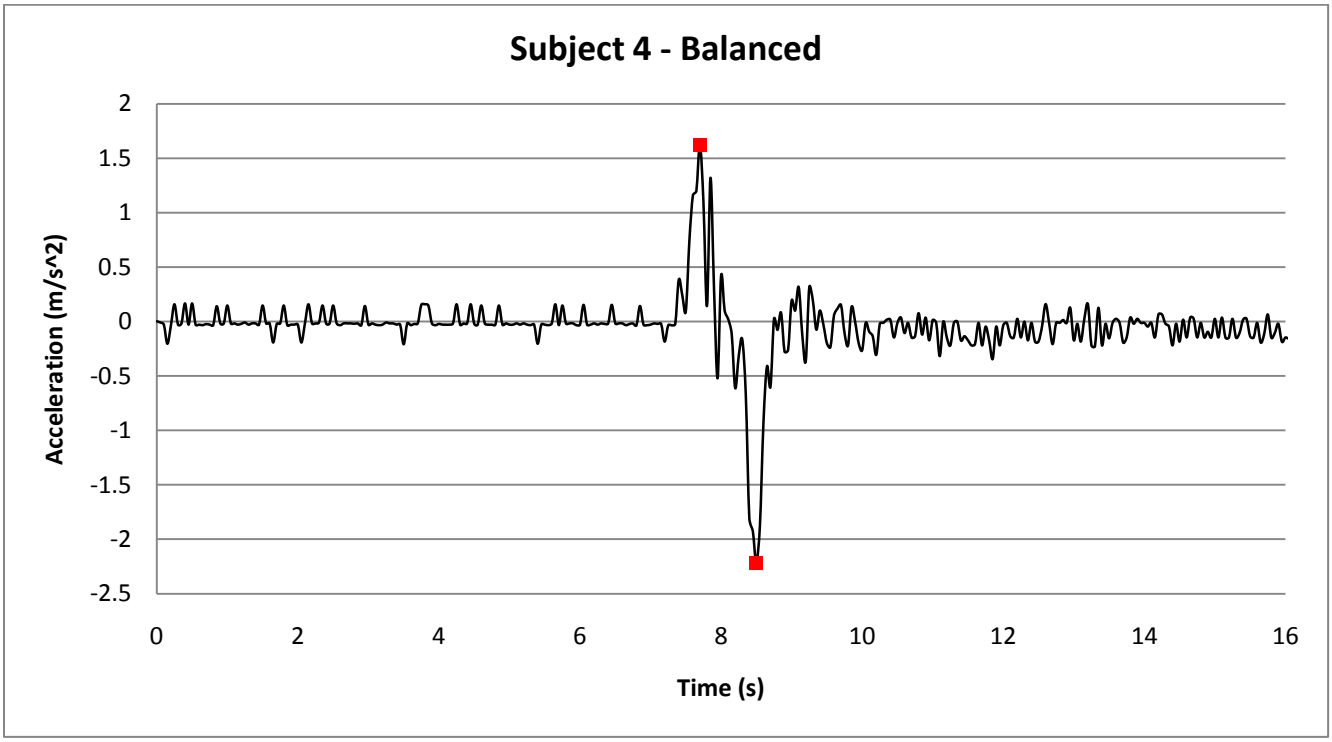

Subject 2- Unbalanced
acceleration (m/s²)
2
1.5
1
0.5
0
-0.5
-1
-1.5
-2
-2.5
-3
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17
Time (s)

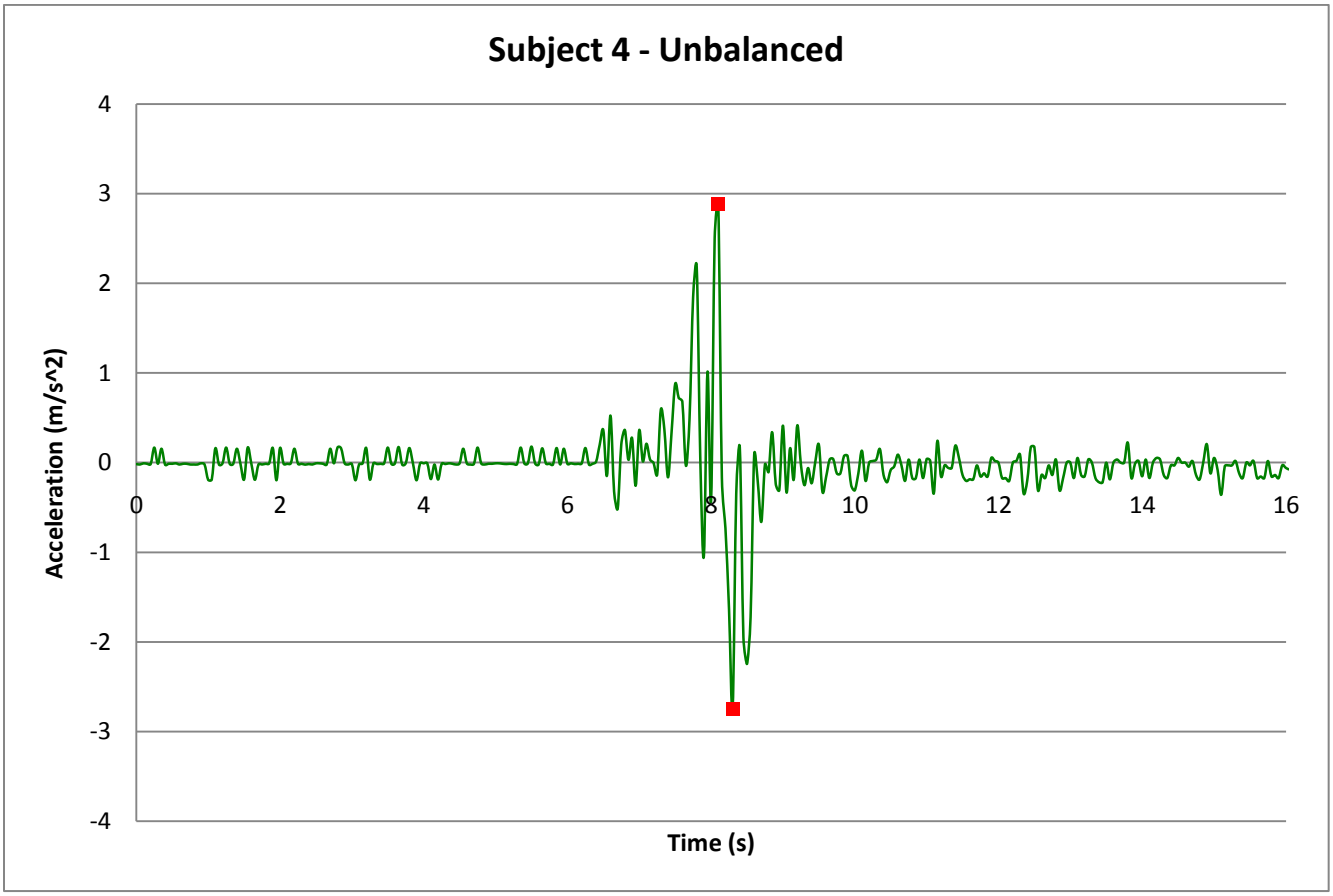
Subject 3


Subject 3- Balanced
acceleration (m/s²)
4
3
2
1
0
-1
-2
-3
-4
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18
Time (s)

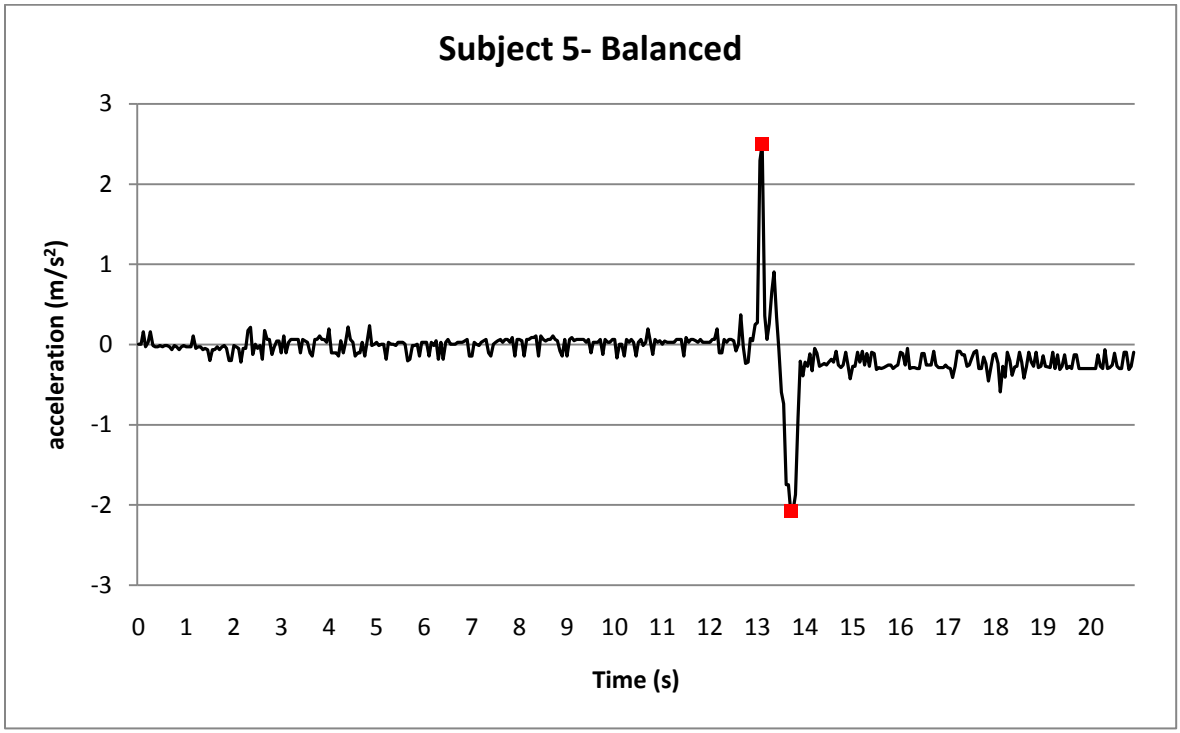

Subject 3- Unbalanced
acceleration (m/s²)
3
2
1
0
-1
-2
-3
-4
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20 21
Time (s)

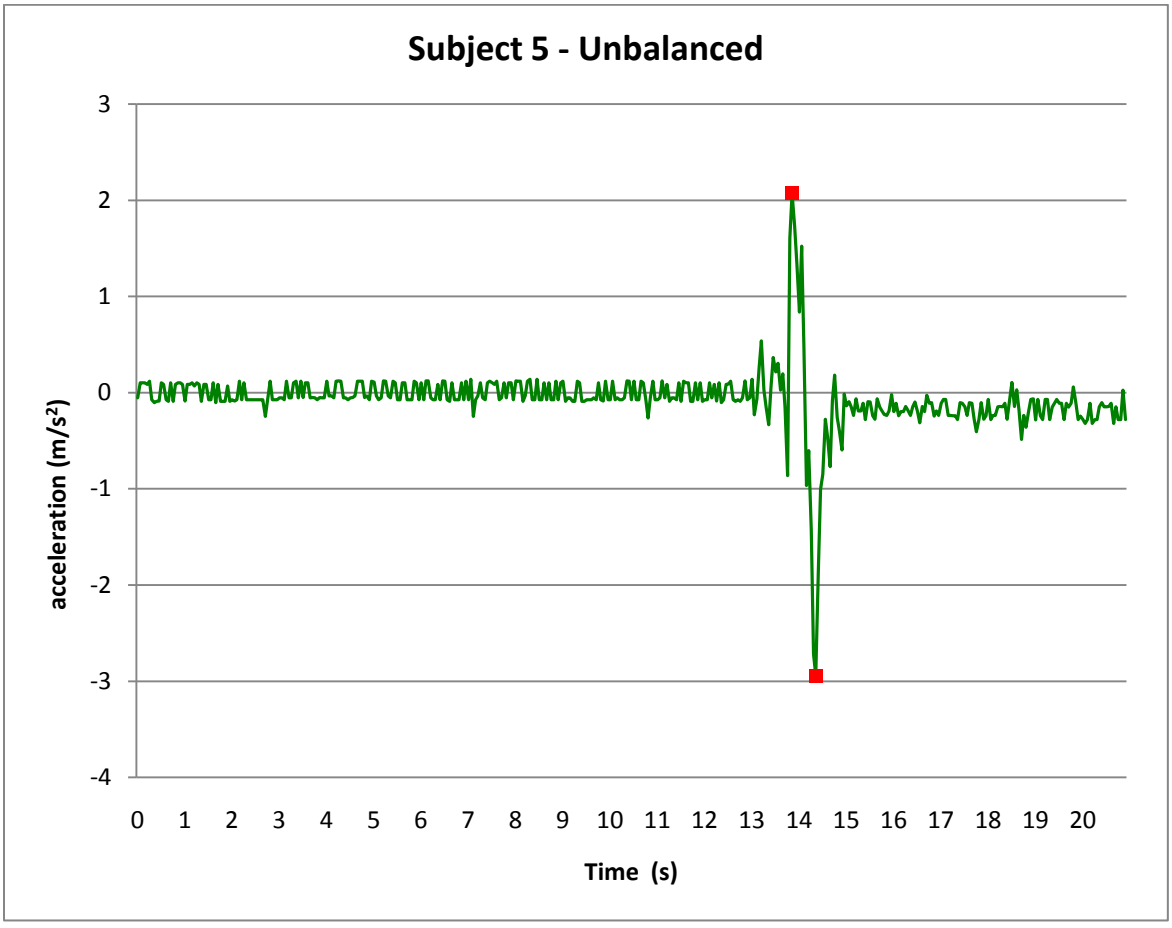
Subject 4


Subject 4 - Balanced
Acceleration (m/s^2)
2
1.5
1
0.5
0
-0.5
-1
-1.5
-2
-2.5
0
2
4
6
8
10
12
14
16
Time (s)

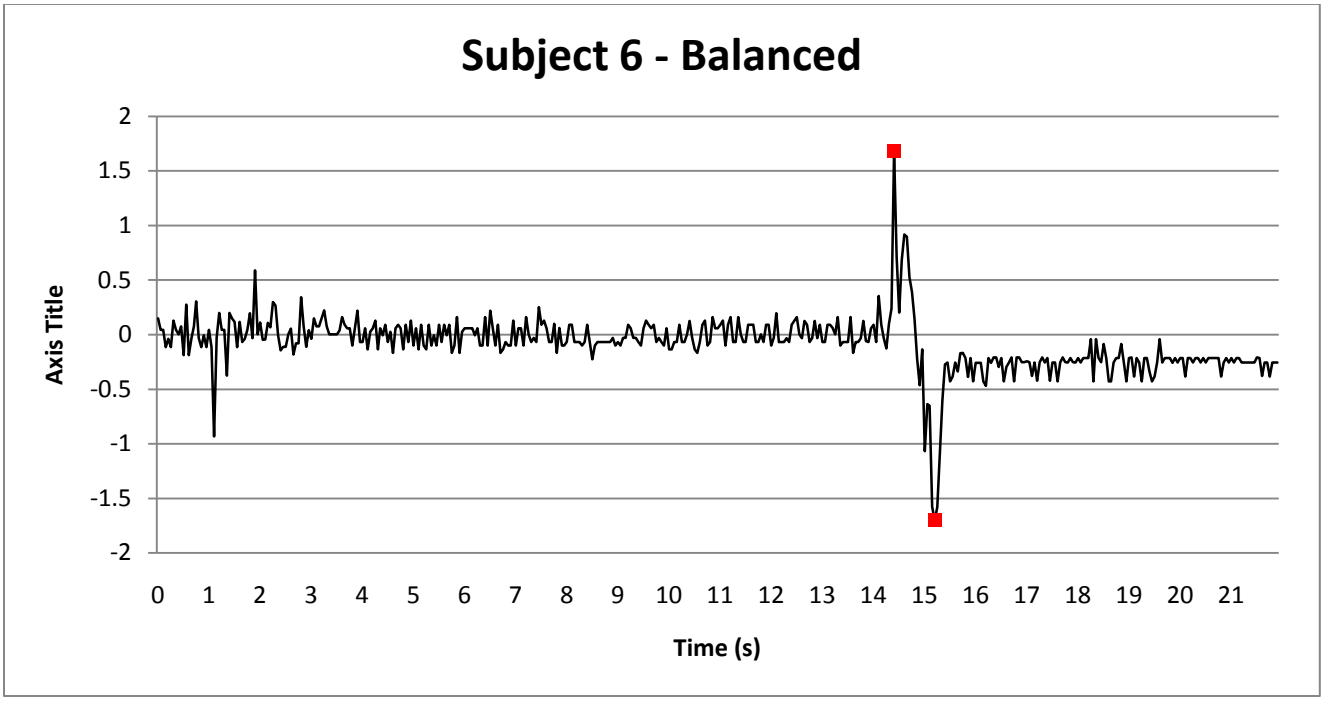

Subject 4 - Unbalanced
Acceleration (m/s^2)
4
3
2
1
0
-1
-2
-3
-4
0
2
4
6
8
10
12
14
16
Time (s)

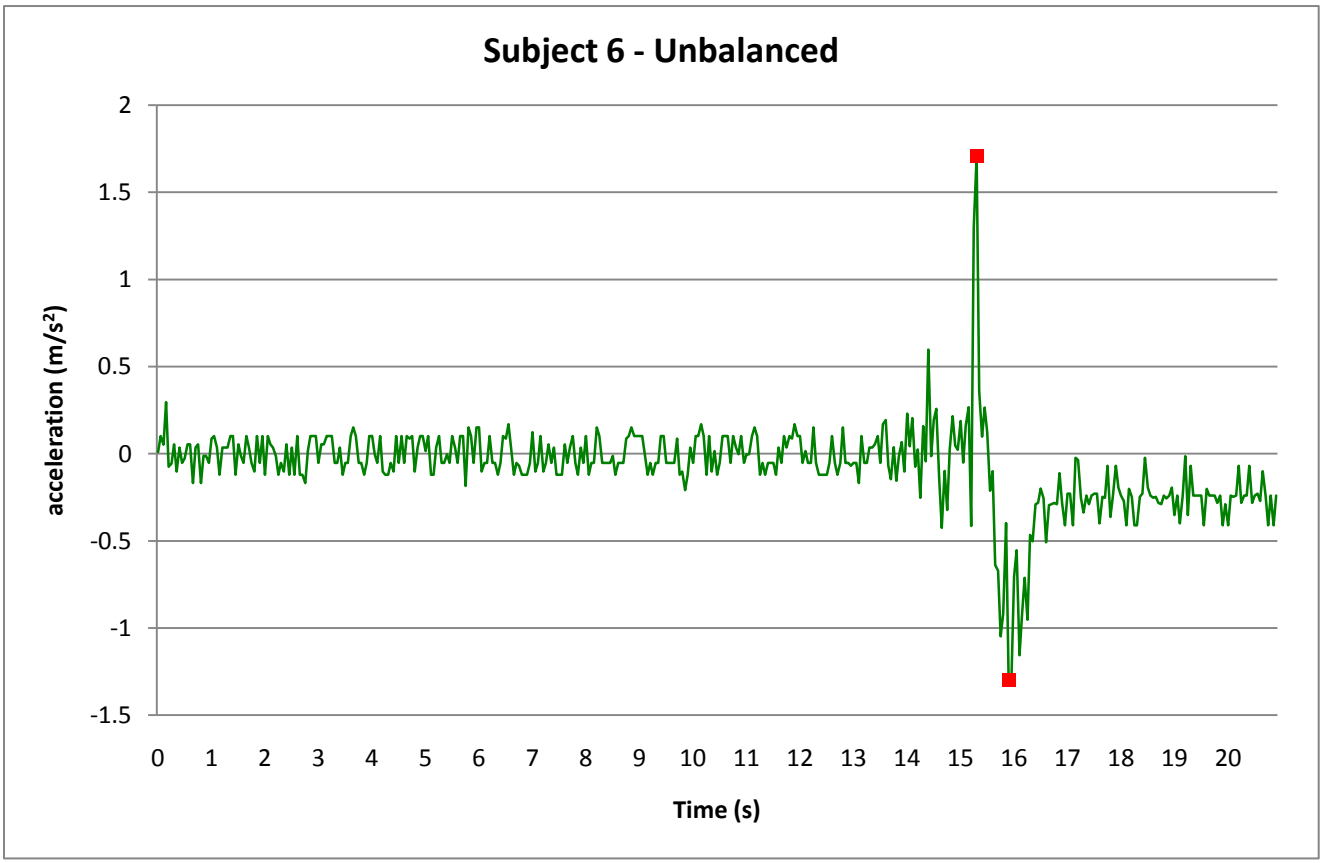
Subject 5


Subject 5- Balanced
acceleration (m/s²)
3
2
1
0
-1
-2
-3
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20
Time (s)

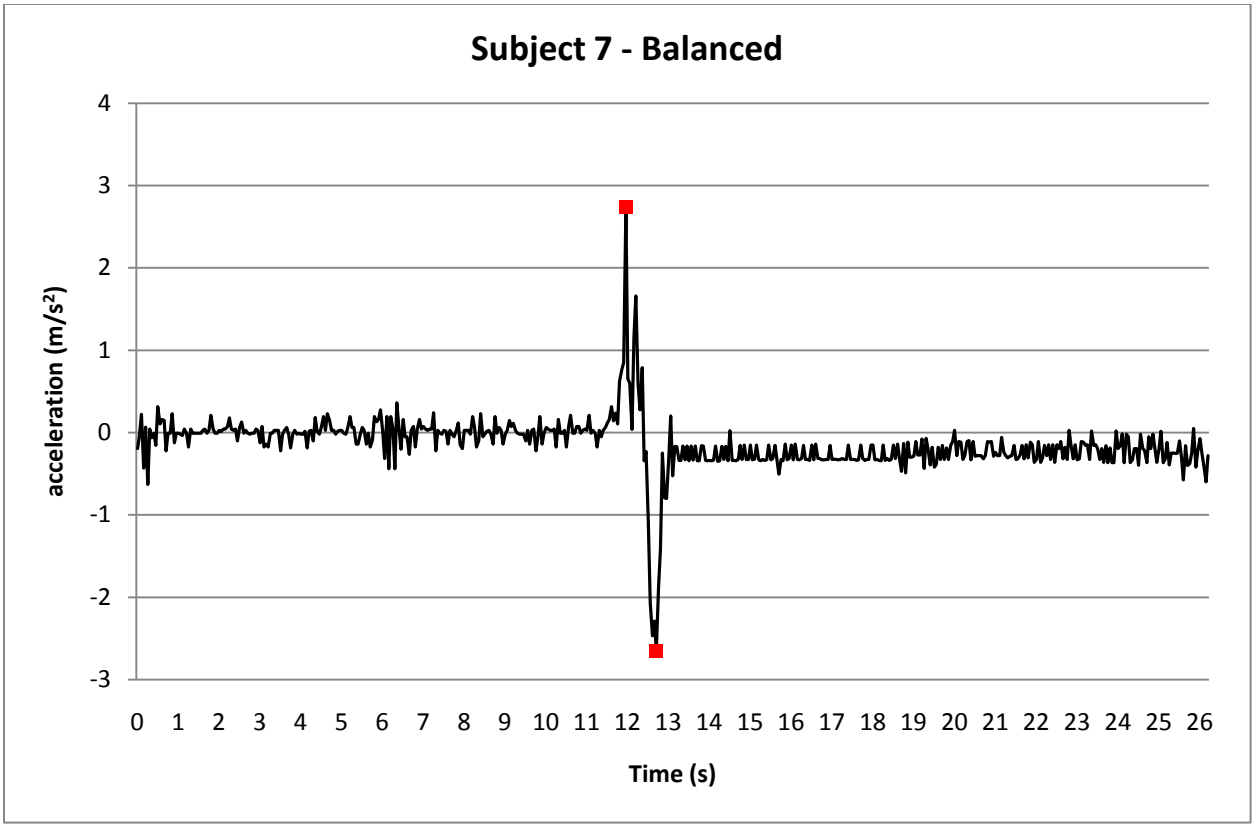

Subject 5 - Unbalanced
acceleration (m/s²)
3
2
1
0
-1
-2
-3
-4
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20
Time (s)

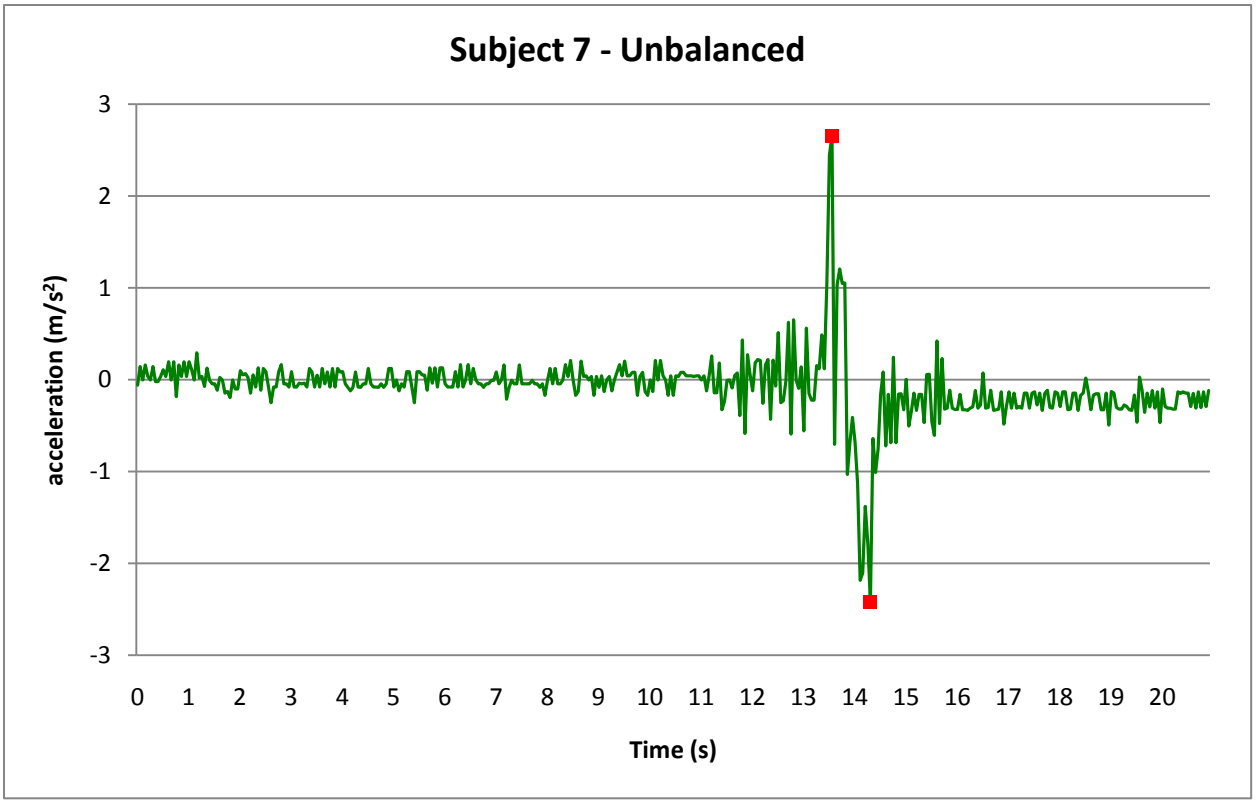
Subject 6


Subject 6 - Balanced
Axis Title
2
1.5
1
0.5
0
-0.5
-1
-1.5
-2
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20 21
Time (s)

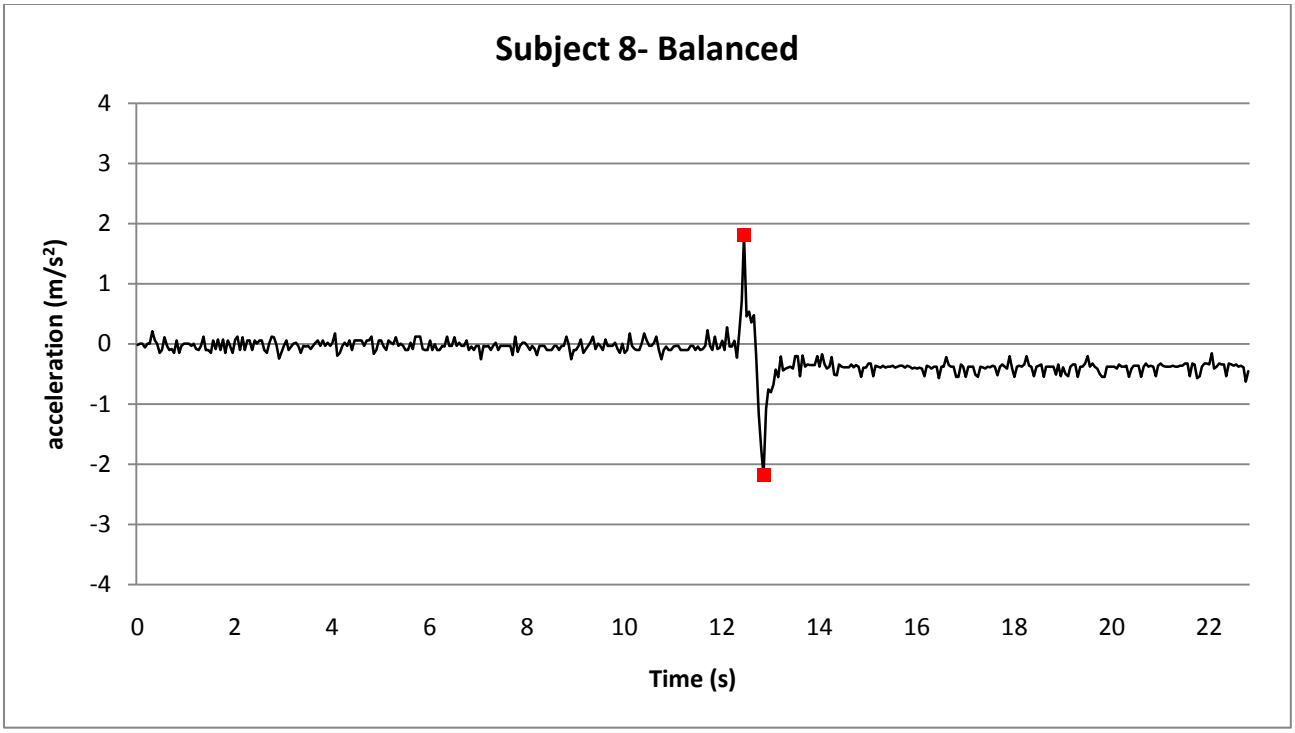

Subject 6 - Unbalanced
acceleration (m/s²)
2
1.5
1
0.5
0
-0.5
-1
-1.5
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20
Time (s)

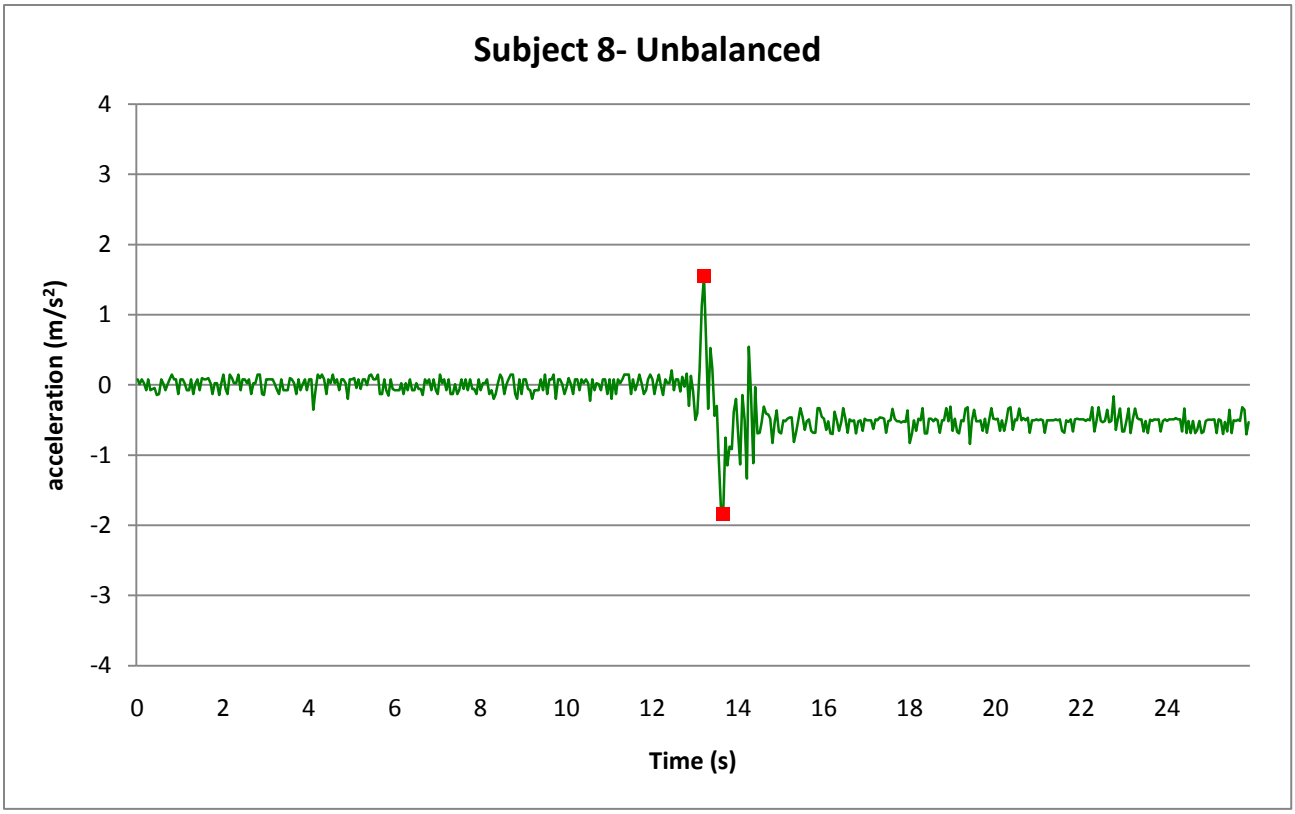
Subject 7


Subject 7 - Balanced
acceleration (m/s²)
4
3
2
1
0
-1
-2
-3
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20 21 22 23 24 25 26
Time (s)


Subject 7 - Unbalanced
acceleration (m/s²)
3
2
1
0
-1
-2
-3
0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20
Time (s)

Subject 8


Subject 8- Balanced
acceleration (m/s²)
4
3
2
1
0
-1
-2
-3
-4
0
2
4
6
8
10
12
14
16
18
20
22
Time (s)


Subject 8- Unbalanced
acceleration (m/s²)
4
3
2
1
0
-1
-2
-3
-4
0
2
4
6
8
10
12
14
16
18
20
22
24
Time (s)

APPENDIX G: Data from Device Verification

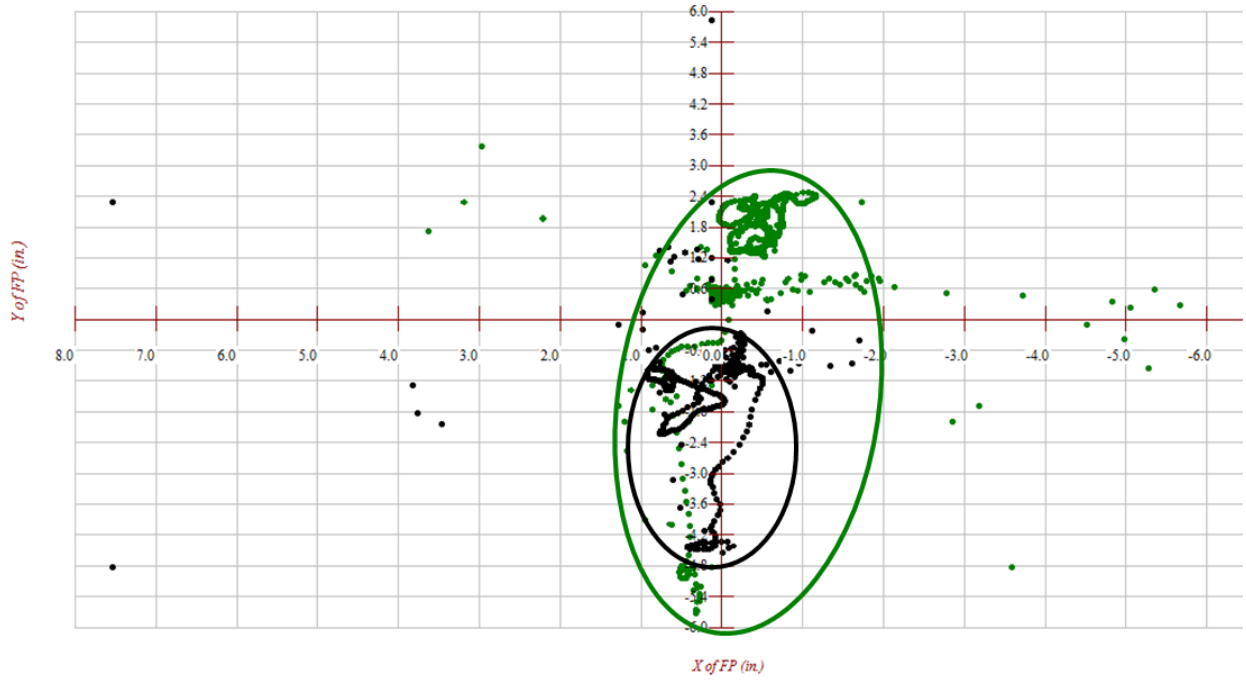
Table 1: Time between positive and negative peaks of shoulder-width design verification trials

Trial	Subject 1 Time (s)
SW	0.80
SW	0.90
SW	1.10
SW	0.80
SW	0.70
SW	0.70
SW	0.80
SW	1.10
SW	0.55
SW	0.95

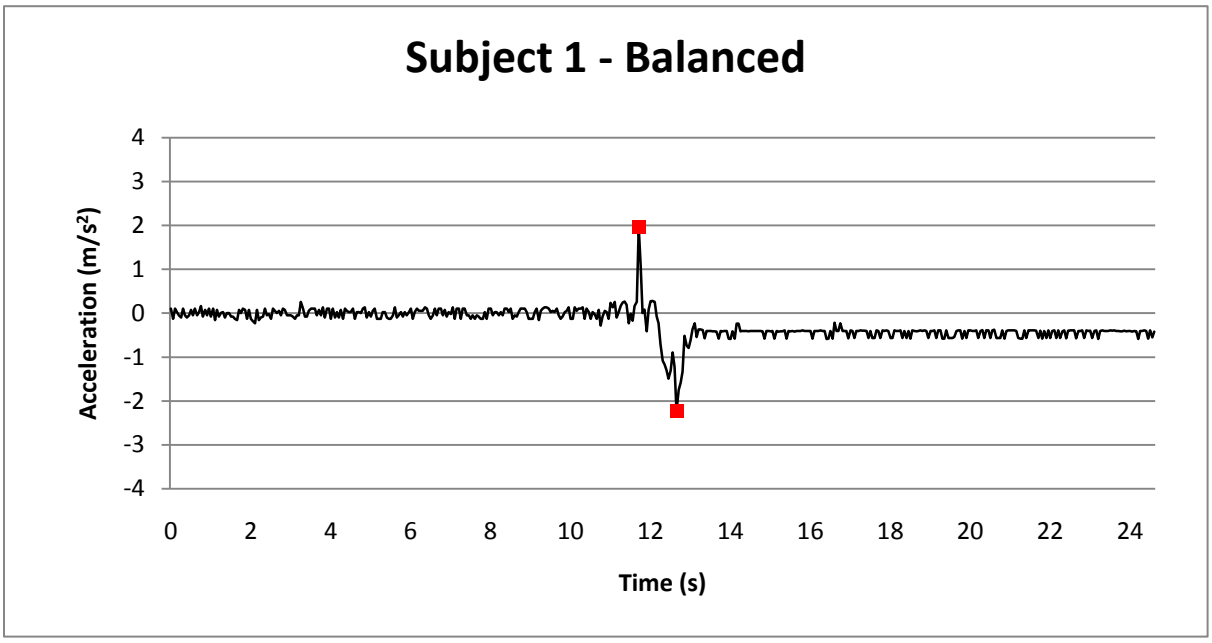
Table 2: Time between positive negative peaks of tandem device verification trials

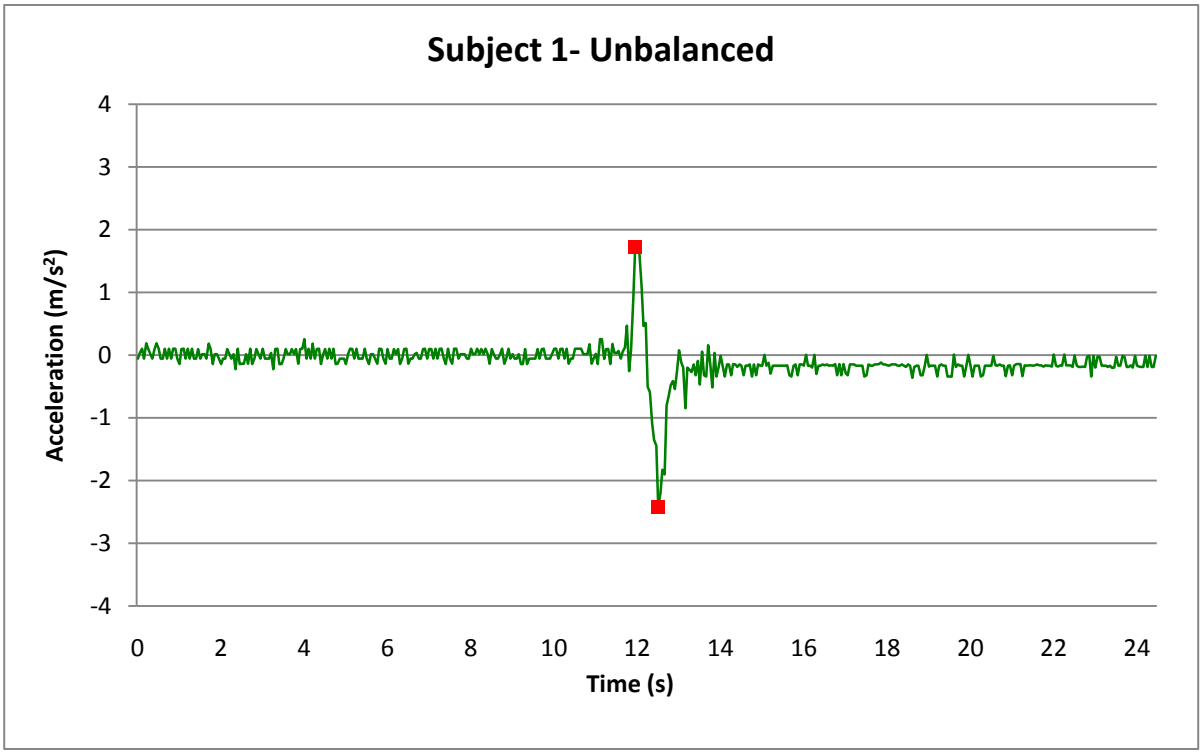
Trial	Subject 1 Time (s)
T	0.45
T	0.40
T	0.50
T	0.60
T	0.35
T	0.45
T	0.70
T	0.55
T	0.30
T	0.75

The following shows a representative center of pressure plots for balanced (black circle) and unbalanced (green circle) sit-to-stand trials completed in design verification testing of subject 1.



The following show a representative balanced (black) and unbalanced (green) sit-to-stand trial for subjects 1 design verification testing.





APPENDIX H: Code for Device

Parts of code were adapted from (Arduino Forum, 2009) and (Faludi, 2007)

```
#include <Wire.h>

#define DEVICE (0x53)    //ADXL345 device address

#define TO_READ (6)        //num of bytes we are going to read each time (two bytes for each axis)

//------Values specific to user

#define StartMinA (0.5)  //threshold acceleration

#define TE (0.6*1000)   //expected time between peaks in milliseconds

#define RingDuration (3.0*1000)  //time in milliseconds to ring buzzer

#define TestLength (2.5*1000)  //total time in milliseconds for test

TestStarted = false;

TestFinished = false;

RingBuzzer = false;

BuzzerOff = 0;

TAA = 0;  //time of max peak

TAD = 0; //time of min peak

AA = 0;  //max acceleration

AD = 0;  //min acceleration

//-----------------------------------------

byte buff[TO_READ] ;    //6 bytes buffer for saving data read from the device

char str[512];                    //string buffer to transform data before sending it to the serial port
```

```
void setup()
{
  Wire.begin();        // join i2c bus
  Serial.begin(9600);  // start serial for output

  //Turning on the ADXL345
  writeTo(DEVICE, 0x2D, 0);
  writeTo(DEVICE, 0x2D, 16);
  writeTo(DEVICE, 0x2D, 8);
}

void loop()
{
  int regAddress = 0x32;  //x-axis registers on the ADXL345
  int regAddress = 0x33;  //x-axis register
  int regAddress = 0x34;  // y-axis register
  int regAddress = 0x35;  //y-axis register
  int regAddress = 0x36;  //z-axis register
  int regAddress = 0x37;  //z-axis register

  int x, y, z;

  readFrom(DEVICE, regAddress, TO_READ, buff); //read the acceleration data from the
ADXL345

   //each axis reading comes in 10 bit resolution, ie 2 bytes.
   //thus we are converting both bytes in to one int
```

```
x = (((int)buff[1]) << 8) | buff[0];
y = (((int)buff[3])<< 8) | buff[2];
z = (((int)buff[5]) << 8) | buff[4];

//we send the x y z values as a string to the serial port
sprintf(str, "%d %d %d", x, y, z);
Serial.print(str);
Serial.print(10, BYTE);

//Delay is needed in order not to clog the port
delay(15);

 int a = sqrt(sq(x)+sq(y)+sq(z));  //magnitude of acceleration

 if (a > StartMinA && ! TestStarted){
    TestStarted = true;
    TimeStart = T;
 }

 if (T-TimeStart > TestLength && ! TestFinished) {
    TestFinished = true;
   if (TAD-TAA < TE) {
      RingBuzzer = true;
      BuzzerOff = T + RingDuration;
   }
 }
```

```
  if (RingBuzzer) {
    if (T>BuzzerOff) RingBuzzer = false;
      :RingBuzzer
      ;
  }

  if (TestStarted && ! TestFinished) {
    if (a>AA) {
      AA = a;
      TAA = T;
    }
  else if (a < AD) {
      AD = a;
      TAD = T;
  }

}

void setup() {
  pinMode(4, OUTPUT); // set a pin for buzzer output
}

void loop() {
  RingBuzzer(4, 2048, RingDuration);  // ring buzzer on pin 4 at 2048Hz for RingDuration
}
```

```
//---------------- Functions
//Writes val to address register on device
void writeTo(int device, byte address, byte val) {
   Wire.beginTransmission(device); //start transmission to device
   Wire.send(address);        // send register address
   Wire.send(val);        // send value to write
   Wire.endTransmission();     //end transmission
}

//reads num bytes starting from address register on device in to buff array
void readFrom(int device, byte address, int num, byte buff[]) {
  Wire.beginTransmission(device);    //start transmission to device
  Wire.send(address);        //sends address to read from
  Wire.endTransmission();      //end transmission

  Wire.beginTransmission(device); //start transmission to device
  Wire.requestFrom(device, num);    // request 6 bytes from device

  int i = 0;
  while(Wire.available())    //device may send less than requested (abnormal)
  {
    buff[i] = Wire.receive(); // receive a byte
    i++;
  }
  Wire.endTransmission(); //end transmission
```

APPENDIX I: Abstract Accepted to Northeast Bioengineering Conference

A Wearable Balance Control Indicator

Amanda Martori, Kevin Goggins, Elizabeth Tuite
Advisors: Profs. Krystyna Gielo-Perczak and Yitzhak Mendelson,
Department of Biomedical Engineering
Worcester Polytechnic Institute, Worcester, MA

Abstract— **Each year, one in three elderly fall. Studies show that many factors contribute to an elderly person's risk of falling, but if the factor causing imbalance is improved, a person's risk of falling may be reduced. A device that detects and alerts the user of an off-balance situation before the fall occurs could identify a specific need for improved balance control. This paper describes the design of a prototype wearable device that can give the user instant feedback regarding their balance control.**

I. Introduction

Every year, one in three elderly fall and 10,000 die as a result of falls [1]. Injuries due to falls cost the United States 20 to 30 billion health care dollars each year and with an increasing elderly population, this amount could rise to about 54.9 billion by 2020 [2].

Studies show that multiple factors increase the risk of falling in the elderly including muscle weakness, and abnormal balance patterns [3]. These factors can be improved to decrease a person's risk of falling. Medical professionals often use a "sit-to-stand" (STS) test, which requires the patient to rise from a chair, to assess balance control [4].

The proposed device can detect imbalance in an elderly person and indicate a need for physical therapy intervention to improve balance control. One device on the market, the iShoe, detects abnormal pressure patterns via a shoe insole [5]. Data collected by the iShoe can be interpreted by a medical professional to evaluate the user's balance control. However, this device does not directly notify the user of their balancing ability.

This paper describes the design of a prototype balance control device for the elderly that gives the user instant feedback regarding their balance control. The device will enable the user to seek medical attention in order to improve their balance control and decrease their risk of falling.

II. Materials and Methods

The first part of the project involved data collection using the SparkFun KinetaMap data logger that contains an ADXL345 tri-axial accelerometer (Analog Devices). The KinetaMap was attached with Velcro to an adjustable belt. The belt was positioned so that the KinetaMap was mounted on the right side of the subject, externally adjacent to the iliac crest (Figure 1). Balance control data were also collected using an AMTI AccuSway force platform. AMTI NetForce and BioAnalysis software were used to record and analyze the balance control data, respectively.

Figure 1: Belt Attachment of KinetaMap and Experimental Set-Up

The force platform was placed directly on the floor and a wooden platform, at the same height, was placed adjacent to the force platform. A chair was placed on top of the wooden platform (Figure 1) so that when the subjects sat down their feet were positioned comfortably on the force platform.

Preliminary studies were conducted on 5 subjects, 1 male and 4 females, all ages 20-21 years. To determine their weight, subjects stood on the force platform with feet shoulder-width (SW) apart for 10 seconds. Next, the subjects sat in the chair with their feet positioned on the force platform; NetForce data collection was started and the KinetaMap was turned on.

After 10 seconds, subjects were asked to rise from the chair at a comfortable speed with their feet SW apart (control). The KinetaMap was turned off 10 seconds after the activity was completed. This process was repeated for 10 separate trials. The process was also repeated for 10 trials where subjects stood up with their feet in tandem (T) (variable). During the STS trials, subjects were asked to rate their comfort level on a scale from 1-5, where 1 corresponds to completely off balance and 5 indicates balance.

The NetForce data files were imported into BioAnalysis software, and plots of center of pressure (COP) were obtained. The COP data were used to determine if the subjects were off balance and quantitatively represent balance control. Raw acceleration data from the x,y, and z components of the KinetaMap were saved in Microsoft Excel for further analysis. Acceleration was converted into m/s^2 and the magnitude of the acceleration profile for each trial was plotted.

The maximum acceleration (a+), minimum acceleration (a), time between these points (t), and threshold acceleration (a_t) below which only baseline noise occurred, were identified for each subject (Figure 3). The mean time(t) was calculated for tandem and SW trials of each subject and compared using a two-tailed, Student's T-Test.

Twenty additional SW and T trials were conducted for subject 1. A time (t_a) was established for subject 1 such that $t \leq t_a$ corresponded to an unbalanced STS. Preliminary results were used to develop an algorithm to detect an unbalanced STS. The algorithm was designed to detect a+, a-, and to calculate time (t). If $t \leq t_a$ the algorithm activates a buzzer. The device includes an ADXL345 tri-axial accelerometer, that is used to sense the acceleration patterns when the user sits down and rises from a chair. The algorithm was programmed on an Arduino microcontroller development kit, and interfaced with the accelerometer and buzzer to alert the user of an off-balance situation.

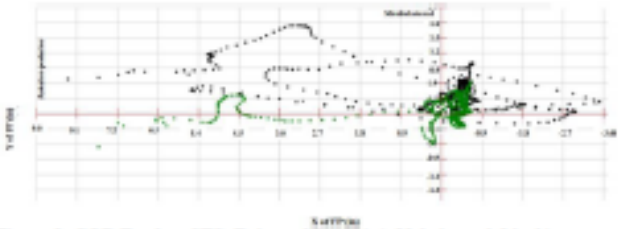


Figure 2: COP-Tandem STS: Balanced (green) & Unbalanced (black)

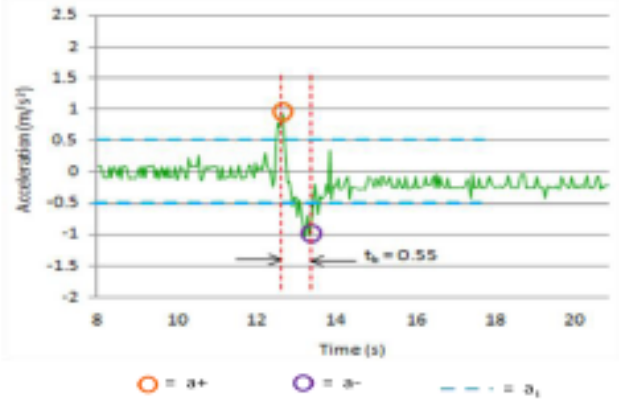


Figure 3: Acceleration signal of balanced (SW) STS

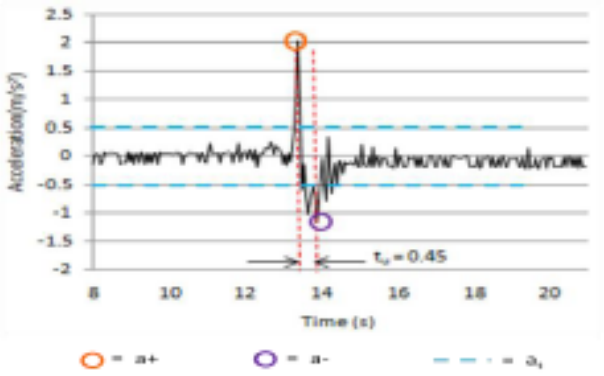


Figure 4: Acceleration signal of unbalanced (T) STS

III. RESULTS

Preliminary testing of 5 subjects resulted in 50 trials rated as a '5', 34 trials rated as a '4', and 14 trials rated as a '3'. All subjects reported feeling more unbalanced during tandem STS and all trials rated $\leq$4 were in tandem.

COP plots showed that the unbalanced STS trials had a greater variation in the anterior-posterior direction, and a less compact COP than the balanced trial (Figure 2). The subject's comfort level in the balanced trial was rated as a 4, and the unbalanced trial was rated as a 3. Acceleration plots (Figures 3 and 4) had a positive acceleration and negative acceleration component, corresponding to the forward lean that initiates the STS and backward lean that completes the motion, respectively. The average time (t) between a+ and a- was significantly less ($p \leq 0.05$) for tandem trials than SW trials in all subjects (Table 1). For all SW trials of subject 1, t>0.70sec and in all tandem trials t<0.65sec. Therefore, t_u of subject 1 was equal to 0.60 sec such that $t \leq 0.60$sec corresponded to an unbalanced STS.

Table 1: Results of SW and T trials

	Average t (sec)		Standard deviation (sec)		Coefficient of variance		p
Subject	SW	T	SW	T	SW	T	
1	0.85	0.45	±0.20	±0.16	0.23	0.35	**0**
2	0.71	0.4	±0.26	±0.18	0.37	0.45	**0.01**
3	0.71	0.49	±0.14	±0.22	0.19	0.44	**0.01**
4	0.59	0.31	±0.28	±0.19	0.48	0.60	**0.02**
5	0.55	0.36	±0.16	±0.17	0.29	0.48	**0.05**

IV. DISCUSSION

The results confirmed that the prototype device can be used to detect an unbalanced situation during STS. Unlike current devices on the market such as the iShoe, the device can assist elderly patients to assess their balance control independently. The device also utilizes a universal anatomical marker, the hip bone, for its positioning. This makes the device user-friendly and since the hip is a reliable location for measuring acceleration, it ensures high accuracy. Once turned on, the device operates on its own requiring no user intervention. The device, used during the STS test, can be integrated into daily life. Balance control parameters vary among users. Therefore, the balance control parameters for each individual need to be collected and replaced in the algorithm.

In the future, the device can be improved to assess balance control during various tests. Due to its ability to instantly detect imbalance for each individual, the device can be used by physicians, physical therapists, or an elderly individual to find out if there is a need for balance control improvement.

V. CONCLUSION

We have designed a prototype wearable device that can detect an abnormal acceleration pattern during a STS activity and alerts the user of an off-balance situation. Our device will allow an elderly person to monitor their balance control while rising from a chair. By signaling an off-balance situation during STS, our device will be able to notify the user of poor balance control.

ACKNOWLEDGEMENTS

We would like to thank our Lab Manager Lisa Wall, Adriana Hera, and all test subjects for their assistance with this project.

REFERENCES

[1] R. Tideiksaar (2002). *"Falls in Older People: Prevention & Management."* Baltimore: Health Professionals Press, Inc.

[2] Centers for Disease Control & Prevention. (2009, January 19). *Cost of Falls Among Older Adults.* Retrieved September 25, 2009, from CDC: http://www.cdc.gov/HomeandRecreationalSafety/Falls/fallcost.html

[3] Winter, D. A., Patla, A. E., & Frank, J. S. (1990). Assessment of balance control in humans. *Medical Progress through Technology*(16), 31-51.

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[5] iShoe, Inc. (2009). *News.* Retrieved February 1, 2010, from iShoe: http://www.i-shoe.net/news.html
rising from a chair in healthy elderly and young women. *Gait and Posture , 8*, 175-185.

APPENDIX J: Conversions of Acceleration Data

Raw data from the accelerometer was output into a comma-separated variable file type, with columns for acceleration in the X, Y, and Z axis. This file type was opened in Microsoft Excel in order to process and analyze the data. First, magnitude was calculated via a root-sum-of-squares method. This can be seen below.

$$Mag_{acceleration} = \sqrt{(x^2 + y^2 + z^2)}$$

Next, Because the data from the accelerometer was outputted on a scale of -128 to 128, rather than $\frac{m}{s^2}$, it was necessary to convert the measured value to a more traditional acceleration measurement. In order to do this, the team implemented the function

$$Acceleration_{\frac{m}{s^2}} = ((Mag_{acceleration} * 18)/1000) * 9.8$$

In order to calculate the acceleration in $\frac{m}{s^2}$. Then, in order to eliminate the gravitational offset value which was generally within a $\pm 1\frac{m}{s^2}$ range of $9.81\frac{m}{s^2}$, the team calculated the average of all the values preceding the STS motion (e.g. the pre-motion baseline), and then subtracted this value throughout the range of the values.

Finally, because the accelerometer was collecting data at 20Hz, we calculated that each datapoint was collected at a 0.05 second interval, and associated each data point with its respective time in seconds. The data was then plotted in magnitude of acceleration in $\frac{m}{s^2}$ versus time in seconds. The curve was then analyzed for minimum value, maximum value, and time duration between minimum and maximum values using built-in excel functionality.

For each subject, the three parameters collected for each plot were statistically analyzed using an unpaired t-test, and the significance in time duration between balanced and unbalanced trials was assessed.

In order to assess the sensitivity of device attachment angle, a similar sit-to-stand test was conducted, and a similar means for data processing was implemented. The minimum, maximum and time duration values were assessed for similarity between the offset trials and the control (0° offset) for significant difference.

APPENDIX K: IRB APPROVAL

Informed Consent for Testing

Informed Consent Agreement for Participation in a Research Study

Investigators: Project Advisors: Krystyna Gielo-Perczak & Yitzhak Mendelson
WPI Students: Kevin Goggins, Amanda Martori and Liz Tuite

Contact Information: BME Department
WPI
100 Institute Road
Worcester, MA 01609
Tel. 508-831-5100, Email: falldetection@wpi.edu

Title of Research Study: Acquisition of Acceleration and Center of Pressure Data while Rising from a Chair

Introduction

You are being asked to participate in a research study. Before you agree, however, you must be fully informed about the purpose of the study, the procedures to be followed, and any benefits, risks or discomfort that you may experience as a result of your participation. This form presents information about the study so that you may make a fully informed decision regarding your participation.

Purpose of the study:

In this experiment, we will investigate how the hip moves while sitting down and rising from a chair. This data will be collected in order to identify a movement pattern that is indicative of an unbalanced situation. This information will be used to develop a device that will detect the point when a user is off balance.

Procedures to be followed:

The subject will wear a belt with an attached accelerometer device. You will stand on a force platform in front of a chair and sit down. Then you will be seated in a chair with your hands on your hips and with your feet on the force platform. Then you will be asked to rest your feet on the force platform with one foot directly behind the other, so that the heel of one foot is touching the toes of the other. Next you will be asked to rise from the chair at a comfortable speed and you will be able to rest for 2 minutes between trials. You will also be asked to rise from the chair as fast as possible, with 2 minutes of rest between trials. You will be provided a break after half an hour. Your participation will last for approximately 1 hour.

Risks to study participants:
There is a minimal risk of falling. The student investigators will be holding the chair and standing next to the platform in order to catch the subject if they do lose their balance and fall.

Benefits to research participants and others:
The benefits are you will become more aware about balance control difficulty, and gain awareness about your body and mobility. The benefit to other is this will help with our project and lead to the design of an early fall detection device for the elderly community.

Record keeping and confidentiality:
Records of your participation in this study will be held confidential so far as permitted by law. However, the study investigators, the sponsor or it's designee and, under certain circumstances, the Worcester Polytechnic Institute Institutional Review Board (WPI IRB) will be able to inspect and have access to confidential data that identify you by name. Any publication or presentation of the data will not identify you

Compensation or treatment in the event of injury:
In the unlikely event of physical injury resulting from participation in the research, you understand that medical treatment may be available from WPI, including first aid emergency care, and that your insurance carrier may be billed for the cost of such treatment. No compensation for medical care can be provided by WPI. You further understand that making such medical care available, or providing it, does not imply that such injury is the fault of the investigators. You do not give up any of your legal rights by signing this statement.

For more information about this research or about the rights of research participants, or in case of research-related injury, contact:
Krystyna Gielo-Perczak, BME Department, WPI, 100 Institute Road, Worcester, MA (Tel. 508-831-5100) You may also contact the Chair of the WPI Institutional Review Board (Professor Kent Rissmiller, Tel. 508-831-5019, Email: kjr@wpi.edu) or WPI's University Compliance Officer (Michael J. Curley, Tel. 508-831-6919, Email: mjcurley@wpi.edu).

Your participation in this research is voluntary. Your refusal to participate will not result in any penalty to you or any loss of benefits to which you may otherwise be entitled. You may decide to stop participating in the research at any time without penalty or loss of other benefits. The project investigators retain the right to cancel or postpone the experimental procedures at any time they see fit. Data obtained in this experiment will become the property of the investigators and WPI. If you withdraw from the study, data already collected from you will remain in the study.

By signing below, you acknowledge that you have been informed about and consent to be a participant in the study described above. Make sure that your questions are answered to your satisfaction before signing. You are entitled to retain a copy of this consent agreement.

_____ Date: _____

Study Participant Signature

Approved by WPI IRB
From: 4/8/2010
To: 4/7/2011

_____

Study Participant Name (Please print)

_____ Date: _____

Signature of Person who explained this study

100 Institute Road
Worcester, MA 01609-2280, USA
508-831-5000, Fax: 508-831-6090
www.wpi.edu

Worcester Polytechnic Institute IRB #1
IRB 00007374

8 April 2010
File: 10-067

Worcester Polytechnic Institute
100 Institute Road
Worcester, MA 01609

Re: IRB Expedited Review Approval: #10-067, "Fall Detection Device"

Dear Dr. Gielo-Perczak,

The WPI Institutional Review Committee (IRB) approves the above-referenced research activity, having conducted an expedited review according to the Code of Federal Regulations 46.

Consistent with CFR 46.116 regarding the general requirements for informed consent, we remind you to only use the **attached stamped approved consent form** and to give a copy of the signed consent form to your subjects. You are also required to store the signed consent forms in a secure location and retain them for a period of at least three years following the conclusion of your study. You may also convert the completed consent forms into electronic documents (.pdf format) and forward them to the IRB Secretary for electronic storage.

The period covered by this approval is 8 April 2010 until 7 April 2011, unless terminated sooner (in writing) by yourself or the WPI IRB. Amendments or changes to the research that might alter this specific approval must be submitted to the WPI IRB for review and may require a full IRB application in order for the research to continue.

Please contact the undersigned if you have any questions about the terms of this approval.

Sincerely,

Kent Rissmiller
WPI IRB Chair

Participant Email:

Dear _____,

We are recruiting subjects to participate in a research study for our Major Qualifying Project (MQP). In our study, we are investigating how the hip moves while sitting down and rising from a chair. This data will be collected in order to identify a movement pattern that is indicative of an unbalanced situation. This information will be used to develop a device that will detect when a user is off balance.

In our experiment, you will be asked to wear a belt with an attached accelerometer device, and sit and rise from a chair with your feet on a force platform (a device similar to a bathroom scale). You will be asked to rise from a chair several times. There is a minimal risk of falling during the experiment, but we will be holding the chair and standing next to you in order to catch you if you do lose balance. The benefit for participating in our study is you will become more aware of balance control difficulty, and gain awareness about your body and mobility. This will also help with our project and lead to the design of a balance control indicator device for the elderly community.

If you are interesting in participating in our study and helping with our MQP, please reply to this email and we can schedule a time for testing.

Thank you,

Amanda Martori, Liz Tuite and Kevin Goggins

Fall Detection MQP Team