The Effect of Spring Loaded Single-Tip Support Cane Mechanisms on Upper and Affected Lower Limb Ground Reaction Forces, Muscle Activity, and Self-Perceived Ease of Use

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A Thesis
Submitted in partial fulfillment
of the requirements for the degree of
Master of Science in
Kinesiology

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August, 2016

Acknowledgements

After an intensive two year period, I am writing this thank you note as the finishing touch on my Master's thesis. These last two years have not only taught me a lifetime worth of knowledge in the field of rehabilitation (more specifically, canes), but also a lot about independent learning and has completely changed the method by which I undertake my academic endeavours. Through this short note, I would like to reflect on the people who have helped and supported me throughout this period. First and foremost, I would like to express my sincere gratitude to my supervisor, Dr. Carlos Zerpa, for his continuous support, motivation, and immense knowledge. I would also like to thank the rest of my thesis committee, Dr. Derek Kivi and Dr. Paolo Sanzo, for their encouragement, insightful comments, and enthusiasm. I would not be where I am today without the help and support I received from you all thus far. Last but not least, I would like to thank my parents, husband, family, and friends for their emotional support throughout this process.

Thank you all,

Aya

Abstract

The primary purpose of this study was to determine the effect of commercial spring loaded single-tip canes on ground reaction forces, impulse, and EMG activation in the upper limb during ambulation. Ground reaction forces and impulse were also assessed for a simulated injured lower limb. A secondary purpose was to assess both traditional and spring loaded cane designs for subject-perceived ease of use. Healthy participants (n=21) were fitted with three types of canes (traditional, Miracle Cane®, and Stander Cane®) and a T-Scope knee brace to simulate an injury. Each participant walked over two force plates, where EMG, force, impulse, and Ease of Use data were collected. Intra-class correlation (ICC) values were calculated for all dependent variables to examine the consistency across replications of the protocol. The result values ranged from 0.558 to 0.999, indicating strong correlations between trials for all measured variables. A one-way ANOVA was performed to analyze differences in walking speed between cane types and no significant differences were found. Multiple two-way mixed factorial ANOVAs were performed to answer research questions regarding differences in muscle activation, ground reaction forces, and impulses between the three types of canes. Statistically significant differences were found in EMG activation between cane types, ($F(2, 280) = 732.48$, $p < .05$, partial $\eta^2 = 0.11$), in which the Miracle Cane® produced less EMG output than all other canes. There was a statistically significant interaction between the type of cane and type of limb on vertical, ($F(2,78) = 35.16$, $p < .05$, partial $\eta^2 = .47$), medial, ($F(2,78) = 4.07$, $p < .05$, partial $\eta^2 = .09$) lateral ground reaction forces, ($F(2,78) = 5.29$, $p < .05$, partial $\eta^2 = .12$) and vertical impulse, ($F(2,78) = 9.93$, $p < .05$, partial $\eta^2 = .2$). There was also statistically significant difference in anterior force production between cane types, ($F(1.645, 64.164) = 7.74$, $p < .05$, partial $\eta^2 = 0.16$). Means, standard deviations, and participant testimonials were analyzed for the Ease of Use Questionnaire. The results from the qualitative and quantitative data indicate that individuals preferred the spring loaded canes over the traditional cane; however, participants preferred the Stander Cane® over the Miracle Cane®. The findings of this research may have implications for the design of standard single-tip support canes and suggest avenues for future research.

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Chapter One – Literature Review

Research findings reveal that approximately 6.6 million individuals living outside of health care institutions use mobility aids (Kaye, Kang, & LaPlante, 2000). Of these mobility aids, canes are by far the most widely used devices, with approximately 19% of these individuals using canes in United States (U.S.) alone (Ipsos, 2009).

Canes are often prescribed to improve people's mobility and help them maintain balance while performing activities of daily living. By decreasing weight bearing on one leg, canes may also help alleviate pain from injury or clinical pathology (e.g., hip fracture, arthritis), or compensate for weakness or impaired motor control of the leg (e.g., from stroke) (Bradley & Hernandez, 2011; Brand & Crowninshield, 1980). Additional clinical benefits ascribed to cane use include improvement of balance control due to a widened base of support (BOS) and increased somatosensory feedback (Jeka, 1997; Tagawa, Shiba, Matsuo, & Yamashita, 2000). Conversely, clinical observation and empirical evidence indicate a high prevalence of disuse and abandonment of mobility aids (Becker, Glad, Nebelsick, & Yernberg, 2013).

Becker et al. (2013) recently reported that 30-50% of individuals abandon their cane devices after receiving it. Problems associated with cane use reported in the clinical literature include discomfort, pain, and injury. Specifically, individuals most frequently complain of pain and injuries in the upper extremity due to repetitive stresses resulting from chronic cane use (Koh, Williams, & Povlsen, 2002; Parr & Faillace, 1999). There are also several mechanisms by which canes are thought to adversely affect balance control (Bouisset & Zattara, 1981); however, it is important to note that inappropriate device prescription, inadequate user training, or use of non-prescription devices may exacerbate the problems listed above (Gitlin & Burgh, 1995; Mann, Hurren, & Tomita, 1993; Schemm & Gitlin, 1998).

The rehabilitation sciences field is always striving to improve upon cane product designs that may be deemed less effective with emerging technology. For example, several modifications have been made to the standard single-tip support cane since its emergence in order to address the aforementioned concerns. Of these modifications, the addition of a spring loading mechanism to the shaft of the cane is the most novel and least researched. The goal of the spring loading mechanism is to store the energy of the impact from cane strike and use this elastic energy to provide propulsion after the midstance stage of ambulation (Liu, Xie, & Zhang, 2011). This cycle of storing and releasing mechanical energy is thought to reduce the magnitude of the impact during the initial contact phase and propel the body after midstance. Furthermore, the spring mechanism is hypothesized to reduce extra push-off being exerted by the upper extremities after midstance, thereby reducing the incidence of upper extremity injuries (Liu et al., 2011).

There are several canes that are currently on the market with spring loaded shafts; however, these canes are in the preliminary stages of research. That is, companies producing these types of canes depend on testimonials to support the efficacy of their products. To the best of our knowledge, there has been no research conducted establishing a causal relationship between spring loading mechanisms in canes and decreased forces on the upper extremities. Studies performed on such mechanisms in auxiliary crutches report a decrease in vertical ground reaction force by up to 26% and increased subjective comfort and ease of use reported (Segura & Piazza, 2007). These decreases are thought to significantly limit the jarring movements seen with standard crutch use and thereby decrease the likelihood of overuse injuries. Furthermore, the literature also suggests that handgrip force and stride length are decreased when spring loaded crutches are used (Parziale & Daniels, 1989). In summary, the use of spring loaded crutches has been shown to possibly alter the mechanics of crutch gait in ways that are likely to reduce injury in crutch users.

The findings related to the addition of spring loading mechanisms to auxiliary crutches suggest that such mechanisms can also be useful in improving current cane designs. This mechanism is an important concept to examine as it addresses a large portion of the concerns associated with current cane designs and benefits a large population. Based on the gaps in existing cane literature, the primary purpose of the current study was to examine the effect of spring loaded cane mechanisms in minimizing the magnitude of ground reaction forces (GRFs), impulse, and levels of muscle activity, as a possible avenue to diminish upper and lower extremity injuries. The secondary purpose of the study was to assess both traditional and spring loaded cane designs for subject-perceived ease of use. The findings of this study may have implications on the rate of abandonment of current cane designs by allowing practitioners to make appropriate recommendations with regards to the best cane design when attempting to minimize the negative effects of repetitive stresses on patients' upper and lower limbs.

The Gait Cycle

The goal of walking is to move the body toward a desired location while using the least amount of energy. The efficiency of walking is moderated by joint mobility and appropriate muscle forces (Cavagna & Kaneko, 1977). As the body moves forward, one limb typically acts as the support limb while the other limb is being advanced. The gait cycle, in its simplest form, is comprised of the stance and swing phases. The stance phase is subdivided into three components, including the initial double stance, single limb stance, and terminal double limb stance (Perry & Davids, 1992).

Sixty percent of the total time of the gait cycle is spent in the stance phase, where at least one foot is in contact with the ground (Paterno & Hewett, 2008). Each double stance period accounts for 10% of the 60% of total time spent in this phase. The remaining 40% of the gait cycle is represented by the swing phase for this same limb. Slight variations in the percentage of stance and swing can be attributed to gait velocity (Jordan, Challis, & Newell, 2007). That is, the duration of each aspect of the stance phase decreases as walking velocity increases. The transition from walking to running is marked by the elimination of double support period(s).

Analysis of the human gait cycle has revealed that a consistent sequence of motions can be observed at each of the joints of the lower extremity during locomotion. Each gait cycle contains a total of eight relevant phases. The stance phase is comprised of five gait phases, which include initial contact, loading response, midstance, terminal stance, and pre-swing (Dekoster, 2014). The remaining three stages take place during the swing phase.

The first two stages of gait occur during initial double support (one of the three components of the stance phase). These stages include initial contact and loading response. Initial contact is often referred to as heel strike. While the term heel strike is appropriate in normal gait, some individuals achieve heel contact later on in the gait cycle, if at all. The main purpose of this stage is to transfer the weight onto the new stance limb while minimizing the magnitude of GRFs, maintaining gait velocity, and maintaining stability (Dekoster, 2014). The loading response phase includes initial contact and continues until the contralateral foot is raised to begin swing. The purpose of this phase is to absorb GRFs as weight is rapidly transferred on the outstretched limb (Astephen & Deluzio, 2005).

The third stage of the stance phase, midstance, occurs during single-limb stance and acts to progress the body's center of mass (COM), which is located approximately in the pelvic area, over the support foot. This progression continues through terminal stance. The terminal stance phase includes heel rise of the support foot and is concluded with contralateral foot contact (Leung, Evans, & Mak, 1998). During this stage of walking, the forces that are translated through the foot are quite significant, often 2-3 times the individual's body weight (Bogey, 2015). Given these high forces and the fact that the average person takes three to five thousand steps per day (with active individuals taking an average of ten thousand steps per day), it is not surprising that the foot can easily be injured or develop chronic stress related issues (Bumgardner, 2015). The final stage of the stance phase, pre-swing, begins initial contact of the contralateral limb and ends with ipsilateral toe-off (Magee, 2008). Rapid unloading of the limb occurs as weight is transferred to the contralateral limb. A major objective of this phase is to position the limb for swing (Magee, 2008). Refer to Figure 1 for a visual illustration of the sub-stages of the stance phase.

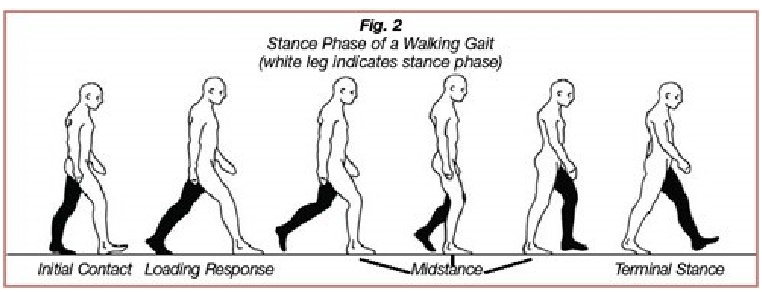

Figure 1. Stages of the stance phase of gait. Reprinted from Biomechanics of Walking, in FootEducation, 2015. Retrieved April 4, 2015, from http://www.footeducation.com/foot-and-ankle-basics/biomechanics-of-foot-and-ankle/biomechanics-of-walking-gait/.

The swing phase is characterized by three unique stages, including initial swing, mid-swing, and terminal swing. Initial swing begins when the foot is lifted from the floor and ends when the swinging foot is opposite the stance foot (Magee, 2008). Two important objectives of this stage are to advance the swing limb forward and achieve foot clearance. The mid-swing phase begins when the swinging foot is opposite to the stance foot and ends when the swinging limb is forward and tibia is vertical (Magee, 2008). Terminal swing, the final third of the swing phase, begins with a vertical tibia and ends when the foot strikes the floor. Limb advancement is completed as the tibia moves ahead of the thigh and the knee maximally extends (Magee, 2008). The two main objectives of this stage are the completion of limb advancement and preparation for stance.

Kinematics of normal gait. Kinematics of the lower extremities, in terms of joint positions, can also be described by the actions that occur at each of the stages of the gait cycle (Dicharry, 2010). At the instance of initial contact (Figure 2A), the hip is positioned at 30 degrees of flexion, the knee at 5 degrees of flexion, and the ankle in a neutral position (Dicharry, 2010). During the loading response phase (Figure 2B), the hip remains at about 30 degrees of flexion. The knee continues to flex, nearly reaching a peak flexion angle of 20 degrees. The ankle begins this phase in neutral (as the loading response includes initial contact), plantarflexes rapidly to achieve a flat foot position, then reverses this motion to return to neutral (Dicharry, 2010).

During the midstance phase (Figure 2C), the hip steadily extends, achieving a position of approximately 5 degrees of flexion. Flexion of the knee ceases very early in the midstance phase and the knee begins extending, reaching a final position of 8 degrees of flexion at the end of this phase. The ankle gradually dorsiflexes to approximately 10 degrees through this phase (Dicharry, 2010). Continuing into the terminal stance phase (Figure 2D), the hip continues to extend through neutral, reaching a final position of 10 degrees of hyperextension. It is important to note that several degrees of this apparent hyperextension can be attributed to pelvic tilting and extension of the lumbar spine; however, this is difficult to distinguish through observation

(Dicharry, 2010). Initially, the knee continues to extend during the terminal stance phase, reaching approximately 5 degrees of flexion. This motion is then reversed (becomes knee flexion) primarily due to heel rise. The knee reaches a final position of 12 degrees of flexion at the end of this phase. As the heel begins to rise, the ankle continues to dorsiflex, reaching a peak angle of 12 degrees. As gastrocnemius and soleus muscle activity increases, this motion ceases and the ankle begins to plantarflex, reaching approximately 10 degrees of dorsiflexion (Dicharry, 2010).

During pre-swing (Figure 2E), the hip reverses directions and flexes to an approximately neutral position. The knee rapidly flexes to approximately 40 degrees of flexion during this phase. The ankle experiences rapid dorsiflexion from 10 degrees of dorsiflexion to approximately 20 degrees of plantarflexion as weight is shifted onto the other limb (Dicharry, 2010).

Moving into the swing stage of the gait cycle, the objective of the hip, knee, and ankle joints are to work together to advance the limb forward and ensure foot clearance. During initial swing (Figure 2F), the hip rapidly flexes to approximately 25 degrees of flexion, the knee flexes to a peak of 60 degrees, and the ankle dorsiflexes to approximately 10 degrees of plantarflexion in order to clear the toes during swing (Dicharry, 2010). As the limb advances into mid swing (Figure 2G), the hip continues to flex to approximately 35 degrees of flexion, the knee rapidly extends to approximately 20 degrees of flexion, and the ankle continues to dorsiflex until a neutral position is achieved. During terminal swing (Figure 2H), the hip extends slightly to a position of 30 degrees of flexion, the knee extends to a neutral position and begins to flex to approximately 5 degrees of flexion, and the ankle remains in a neutral position (Dicharry, 2010).

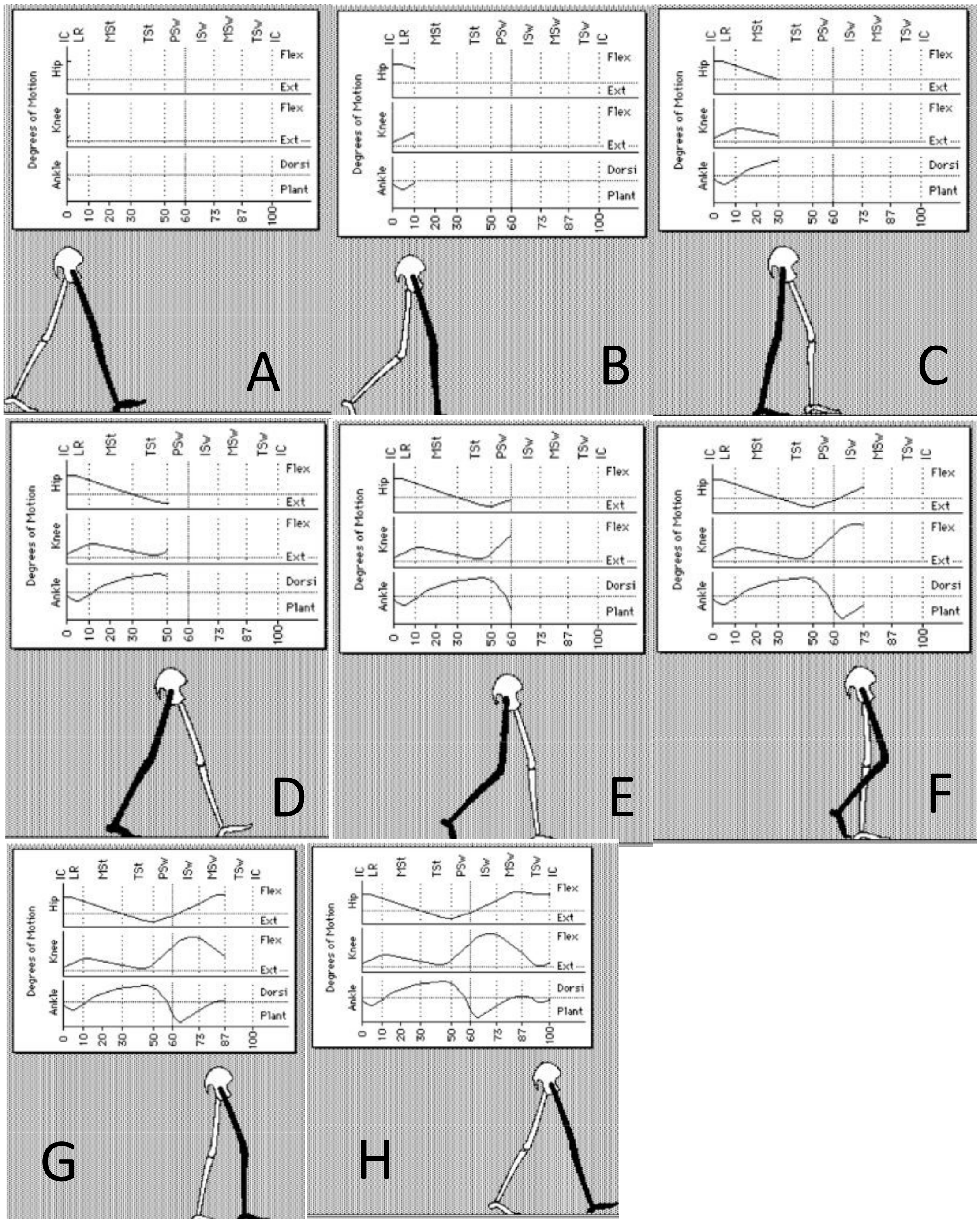

Figure 2. Joint positions during normal gait cycle. Reprinted from *Observational Gait Analysis*, by Los Amigos Research and Education institute, Rancho Los Amigos National Rehabilitation Center, 2001.

Kinetics of normal gait. In addition to observational or kinematic analysis, gait can also be analyzed through assessment of GRFs and impulses, a branch referred to as kinetics (Winter, 1984).

***Ground reaction forces*.** Ground reaction forces are typically measured with a force transducer, which provides an electrical signal that is proportional to the applied force (Simon et al., 1981). Force transducers are often embedded within force platforms. Ground reaction forces acting on a foot during standing, walking, or running, are traditionally measured by force platforms (Simon et al., 1981). Force plate output data provides three ground reaction force vector components: vertical, anterior-posterior (AP), and medial-lateral. Normal gait can be represented by typical force-time graphs for each of these vector components (Simon et al., 1981).

The vertical component of ground reaction force, shown in Figure 3, is the largest and accounts for the vertical acceleration of the body's COM during walking. This force curve is often referred to as the "M curve" due to its resemblance to the corresponding letter in the English alphabet (Marasovic, Cecic, & Zanchi, 2009). At the instant of initial contact, zero vertical force is produced. As the limb advances into loading response, the force begins to quickly rise. This increase in force is attributed to the increase in body weight being supported by the limb. During midstance, force decreases below body weight as the COM experiences a downward acceleration, which creates an upward inertial force that must be subtracted from the body weight. The change in COM mass position is caused by the sinusoidal motion of the pelvis during walking, which rises and falls approximately 10 cm in space (DeLisa, 1998). In the final phases of stance, a second peak is created (due to downwards deceleration of the COM) and force rapidly reaches zero as the foot transitions into the swing stage (Marasovic et al., 2009).

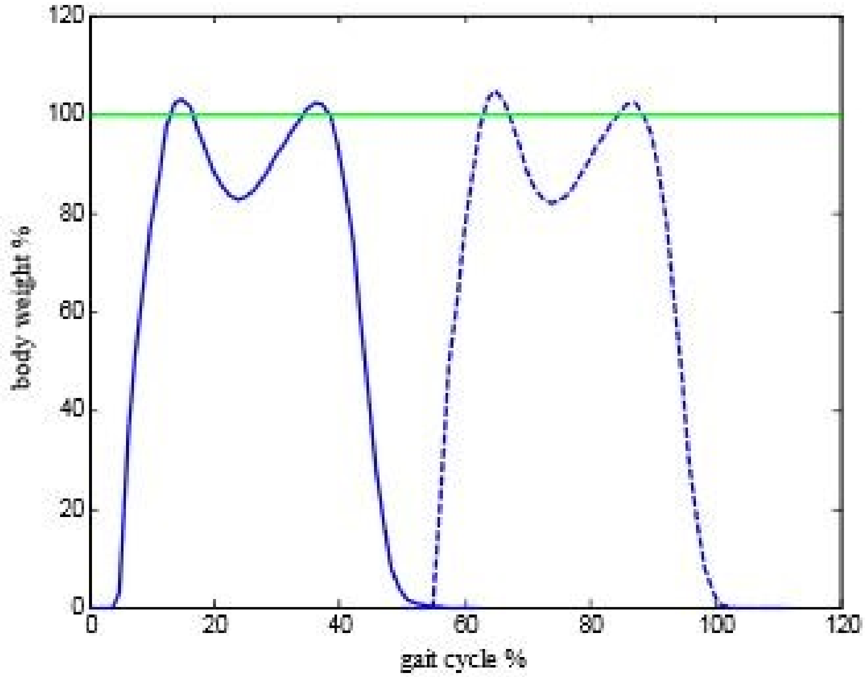

Figure 3. Vertical GRF during normal gait. Reprinted from “Analysis and Interpretation of Ground Reaction Forces in Normal Gait,” by Marasovic, Cecic, and Zanchi, 2009, *WSEAS TRANSACTIONS On SYSTEMS*, *8*(9), 1105-1114.

The AP GRF, seen in Figure 4, represents the horizontal force exerted during contact. This GRF acts in the direction of the human walking forwards and backwards (Marasovic et al., 2009). Initially, the force-time curve shows a breaking force (negative direction) until midstance in order to decelerate the body’s COM. This breaking force is followed by a propulsive force (positive direction) following midstance.

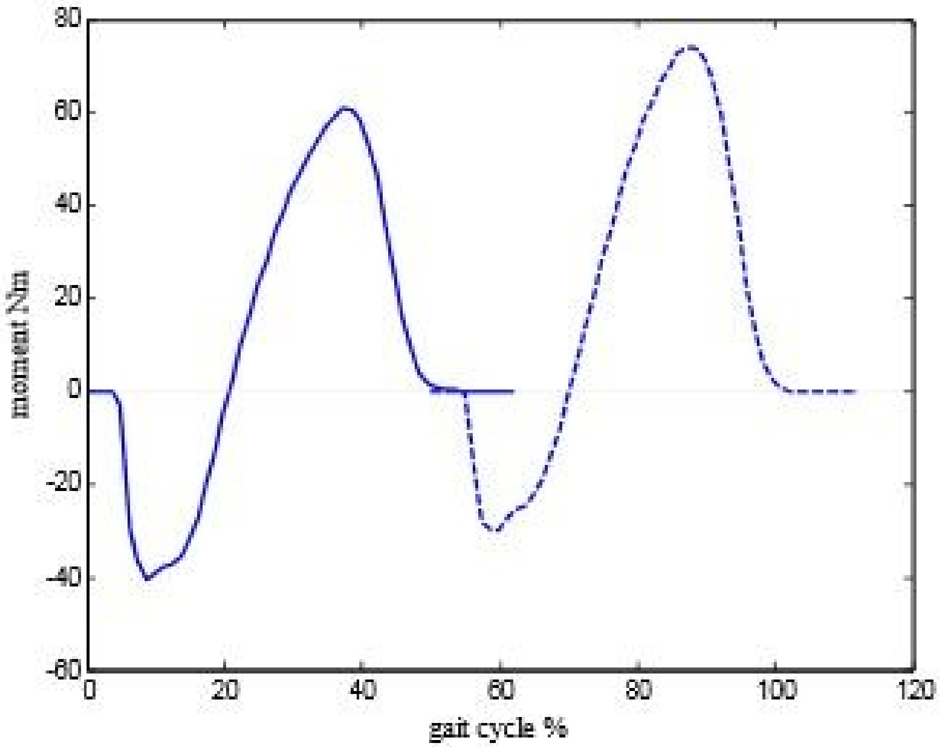

Figure 4. Anterior-posterior GRF during normal gait. Reprinted from “Analysis and Interpretation of Ground Reaction Forces in Normal Gait,” by Marasovic, Cecic, and Zanchi, 2009, *WSEAS TRANSACTIONS On SYSTEMS*, *8*(9), 1105-1114.

The final component of GRFs, medial-lateral GRF, represents the magnitude of the medial-lateral shear force. The magnitude of this force is dependent on the position of the COM relative to the foot; therefore, as step width increases, shear force increases due to the increased angle between the lower extremity and the point of contact. In a typical walking pattern, the COM tends to move laterally at heel strike and during the loading response and moves medially through the rest of the stance phase. This pattern is illustrated in Figure 5.

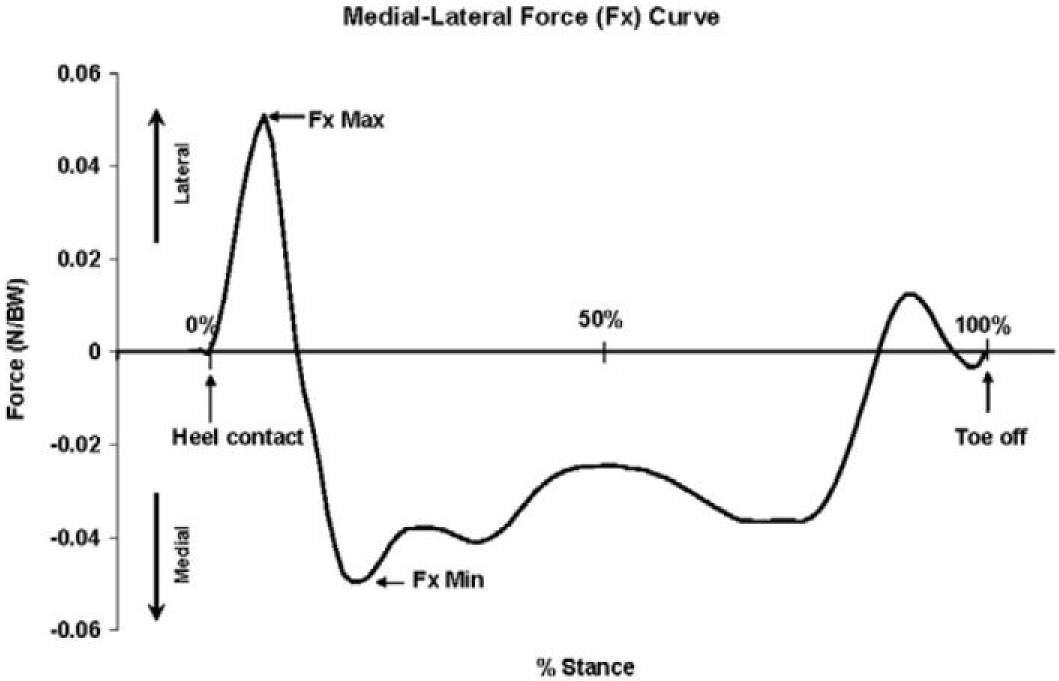

Figure 5. Medial-lateral GRF in normal gait. Reprinted from "Peripheral arterial disease affects ground reaction forces during walking," Scott-Pandrof et al., 2007, *Journal Articles*, paper 150.

Impulse. The area under the GRF curves represents the impulse or the time integral of force. Impulse is an important factor to consider when examining AP GRFs. In this force-time curve, it can be observed that the positive and negative forces are approximately symmetrical, which can be explained by the change in impulse. The area under the force-time curve represents the impulse, which can also be referred to as the change in momentum. If an individual is walking at a constant speed then there should be no change in momentum, and the total impulse in the AP direction should equal to zero. This means that the breaking impulse is approximately equal to the propulsion impulse in normal gait. This fact is particularly important for the analysis and diagnosis of pathological gait patterns (Marasovic et al., 2009).

Muscle activity during gait. Muscle activity during gait is typically studied using electromyography (EMG; Criswell & Cram, 2011). Electromyography is a diagnostic technique for recording the electrical activity produced by skeletal muscles (Robertson, 2004). In the case of Delsys EMG systems, wireless hybrid sensors can be used to detect the electrical potential generated by muscle fibers when they are neurologically active (Robertson, 2004). Generally, EMG systems should not record any electrical activity when the muscle is at rest. Electromyographic recordings differ between individuals, and within individuals according to variables such as velocity; however, as with joint positions, typical patterns of muscle activation during normal gait have been identified (Criswell & Cram, 2011). Muscle activity can also be defined as the actions that take place in each of the aforementioned stages of gait. Since initial contact is identified as an instance (rather than a phase), it will be grouped with the loading response phase in order to simplify the explanation of muscle activation.

The loading response phase is a period of extensive muscle activity. Hip flexion is controlled through isometric action of the hamstrings and the lower portion of gluteus maximus. The quadriceps contract eccentrically to control knee flexion. The ankle dorsiflexors also act eccentrically to prevent slapping of the foot on the ground. In the frontal plane, activity in the tensor fascia latae, hip abductors, and gluteus maximus control drop of the contralateral pelvis. The erector spinae muscles are also active during the loading response. This muscle group acts to stabilize the trunk during weight transfer (Ivanenko, Poppele, & Lacquaniti, 2004).

During the midstance phase, the hip abductors continue their activity isometrically in order to control and halt pelvic drop. Knee extension is initiated as the quadriceps contract concentrically. Plantarflexors of the foot act eccentrically to control ankle dorsiflexion (Ivanenko et al., 2004).

Moving into terminal stance, the hip abductors change roles from working eccentriccally to concentriccally in order to elevate the ipsilateral pelvis in preparation for swing. The quadriceps remain inactive during this phase as the plantarflexors, coupled with GRFs, maintain extension of the knee. Ankle plantarflexors continue to function and contract isometrically as the heel begins to rise from the floor (Ivanenko et al., 2004).

Much like the loading response, the pre-swing phase is a period of large muscle activity. At typical walking speeds (5.0 km/h), the rectus femoris acts to limit knee flexion. It is only at speeds that are slower than typical (when GRFs are too small to initiate knee flexion) that the knee flexors work to flex the knee directly. The plantarflexors act concentrically in order to produce a propulsive pushoff (Ivanenko et al., 2004).

During the initial swing phase, the hip flexors and knee extensors (rectus femoris) continue their activity similar to the pre-swing phase activity pattern. The dorsiflexors begin to act concentrically to permit the forefoot to clear the ground. Muscle activity virtually ceases during the midswing phase except for the dorsiflexors of the ankle as inertia carries the limb through much like a pendulum (Ivanenko et al., 2004).

During the final phase of the gait cycle, terminal swing, the hamstrings contract eccentrically to decelerate the swinging limb, while the dorsiflexors hold the ankle joint in position for initial contact. In preparation for initial contact, the quadriceps and hip abductors initiate activity (Ivanenko et al., 2004). Refer to Figure 6 for an illustration of muscle activity during normal walking.

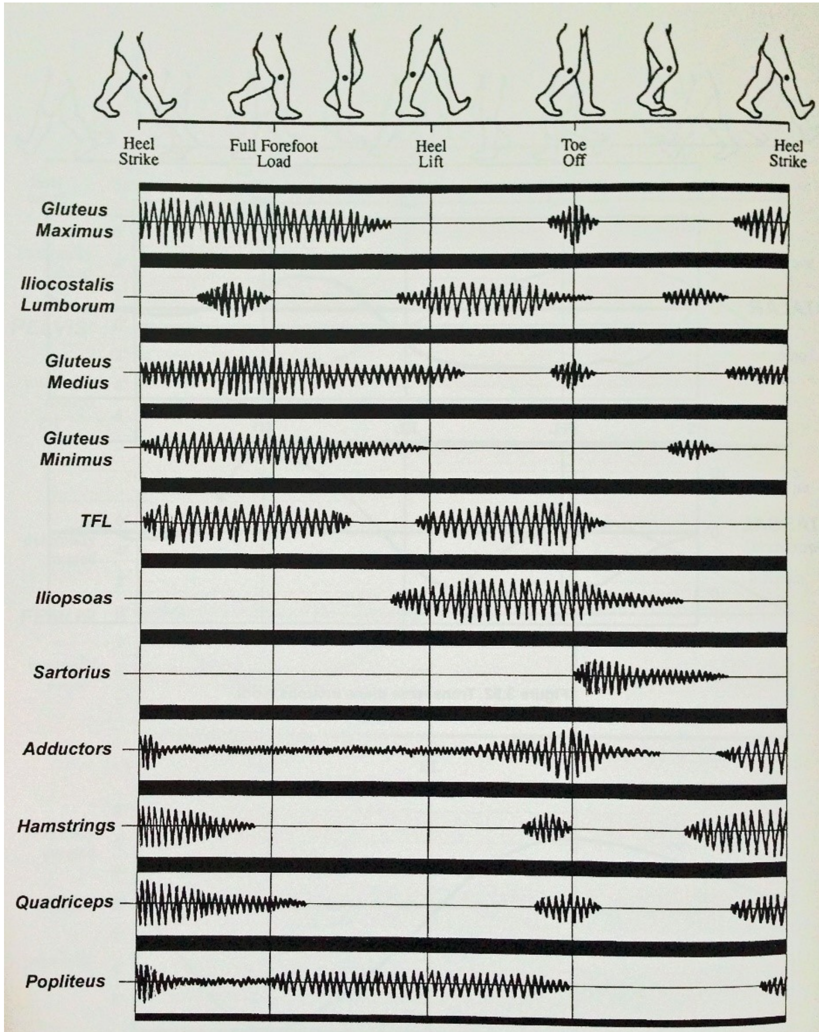

Figure 6. Muscle activation during normal gait. Reprinted from Muscular contribution to gait, from Running Reform, January 5, 2014, retrieved December 15, 2015, from http://runningreform.com/muscular-contribution-to-gait-what-do-we-really-know/.

Pathological Gait Patterns

Pathological gait patterns deviate from the typical pattern described above (Saunders, Inman, & Eberhart, 1953). There are numerous causes of pathological gait and there can be great variation depending on the severity of the problem (Kirtley, 2006).

Causes of a pathological gait. Several factors can contribute to a deviation from typical gait patterns. Some of these factors include weakness, pain, hypomobility, hypermobility, neurological involvement, or leg length discrepancy (Clave, Galland, & Cagny, 1939). The following sections will define these factors and indicate their relevance to pathological gait.

Weakness. Muscle weakness may result from disuse, primary muscle disease, or neurologic impairment (Page, Frank, & Lardner, 2015). Weakness in the leg/postural muscles may contribute to the presentation of a pathological gait pattern. For example, uncompensated calf weakness results in diminished midstance control of the anteriorly rotating tibia (Nadeau, Gravel, Arsenault, & Bourbonnais, 1999). Another example can be seen as a consequence of weakness of the muscles located in the anterior compartment. With mild weakness, foot slap can be observed during the loading response stage, with drop foot and toe drag occurring in more extensive weakness (Lamontagne, Malouin, Richards, & Dumas, 2002). Weakness in the anterior compartmental muscles results in increased knee and hip flexion (to allow the dropped foot to clear the walking surface) as well as circumduction of the hip. Quadriceps and hamstring muscle weakness diminish knee control, and deficits in stance are most pronounced (Mikesky, Meyer, & Thompson, 2005). Finally, hip adductor weakness results in pelvic instability during the stance phase of the gait cycle, which causes the individual to present with a tilt towards the unaffected side.

Hypermobility/hypomobility. The amount of motion that an individual is capable of producing at each joint can be moderated by several factors. Hypermobility refers to joints that are capable of stretching further than normal (Kirk, Ansell, & Bywaters, 1967). Hypermobility can be caused by misaligned joints, abnormally shaped articular surfaces, connective tissue defects, abnormal joint proprioception (an impaired ability to locate body parts in space and/or monitor a joint exceeding normal range of motion), or congenital issues such as Down’s Syndrome and Ehlers Danelos Syndrome (Castori, 2012; Gabbey, 2015; Uno, Kataoka, & Shiba, 1996). Individuals with hypermobility may be easily injured and develop problems as the muscles in the region fatigue; the muscles need to work harder to compensate for laxity in the ligaments that support the joints. Both muscle weakness and injury can contribute to the development of pathological gait patterns.

Hypomobility, on the other hand, refers to a limitation in the amount of motion possible at a joint (Fernández-de-las-Peñas, 2009). Hypomobility can result from pain, surgery, fractures of the surrounding bone structure, or extended periods of immobilization (i.e., casting) (Active Thai Stretch, 2013). For example, hypomobility of the knee joint may result if the individual holds the knee in a position that unloads the painful or swollen joint. Sometimes the position chosen may be the resting position of the knee often limiting the joint to -30 degrees of extension as this is the position that correlates with the resting position of the knee (Bogey, 2015). This type of hypomobility diminishes limb advancement in the early swing phase and shortens step length as a result of decreased knee extension in the terminal swing stage (Bogey, 2015).

Pain. Pain does not directly alter the normal gait cycle; however, changes in the normal walking pattern may occur when an individual attempts to attenuate pain through gait modifications (Graven-Nielsen, Svensson, & Arendt-Nielsen, 1997). Generally, GRFs at the level of the joints are magnified with increased muscle forces crossing the joint. These increases in joint reaction forces are typically associated with increased discomfort. One way to reduce joint pain is to limit the muscle force output at the painful joint. Thus, individuals experiencing pain in the lower limb joints tend to present with a pathological gait pattern with decreased stride length, decreased velocity, and decreased time spent in the stance phase (Powers, Heino, Rao, & Perry, 1999).

Kinematics of Pathological Gait

Pathological gait, depending on the abnormality, can have kinematic deviations in all three planes and all phases of the gait cycle (Hsu, Michael, & Fisk, 2008). In the transverse plane, there can be either excessive or insufficient rotations at all joints of the lower extremity (Hsu et al., 2008). In the coronal plane, there can be excessive or insufficient hip adduction or abduction knee varus or valgus, or ankle inversion or eversion. In the sagittal plane, there can be excessive or insufficient plantar or dorsiflexion of the ankle (Hsu et al., 2008). At the knee and hip there can be excessive or insufficient flexion or extension (Hsu et al., 2008).

Kinetics of Pathological Gait

Deviations from normal gait can also create deviations in the point of application, magnitude, and line of action of the GRFs (Hsu et al., 2008). When considering vertical GRFs, a common indication of abnormal gait is an excessively high first peak in early stance and an insufficient second peak in terminal stance (less than body weight). This pattern of vertical forces indicates that the limb is not supporting the body weight sufficiently to remain fully functional as a support (unless an external support is being utilized). As a consequence, the contralateral limb may generate an excessive first peak in the vertical GRF (McCrory, White, & Lifeso, 2001). In conditions where an antalgic gait pattern is present, propulsive GRFs are decreased. This decrease can be attributed to the lack of muscular strength, pain avoidance, or lack of mobility (Zeni & Higginson, 2009).

Assessment of Pathological Gait

In clinical settings, the assessment of the aforementioned signs and symptoms and pathological gait is often performed through simple observation (Saunders et al., 1953). The clinician examines the patient and makes note of any significant observations with regards to his/her walking pattern (Saleh & Murdoch, 1985). For example, if the patient has difficulty rising from a chair, this may suggest proximal muscle weakness, balance problems, or difficulty initiating movements (Hughes, Weiner, Schenkman, Long, & Studenski, 1994). The speed of walking can also be indicative of certain pathologies such as degenerative joint disease or weakness in the case of slowed walking (Lowth, 2015). The way that a patient walks is also important to observe as weakness in one area may lead to compensation in another area (i.e., increased flexion at the hip or knee due to a drop foot or loss of dorsiflexion). Problems changing direction while ambulating are also common with many gait disorders as turning is generally more difficult than walking Finally, if the clinician observes difficulty with balance and a widened BOS compensatory pattern is evident, it may indicate cerebral dysfunction or neurological involvement (Lowth, 2015). The assessment and examination of gait and balance must always be supplemented with an appropriate subjective patient history and objective examination of all body systems. The end result of the clinical gait assessment procedure, is the determination of the appropriate treatment plan that may include a specific orthotic design and recommended walking support (if it is a pathology that can be managed with an assistive device).

Rehabilitation and Treatment for Pathological Gait

The patient's type of deviation, primary functional deficit, and pathology influence the type of orthotic/assistive device recommended (Stewart & Shortland, 2010). The treatment of pathological gait often begins with the identification of the problem and determination of the underlying cause. In the case of pathological gait, the individual may be attempting to reduce or eliminate pain in one limb, maintain balance, or compensate for a weak or injured limb. In most cases, once the underlying cause or pathology is eliminated, the pathological gait pattern also diminishes and normal gait is restored (Stewart & Shortland, 2010). Additional treatments that can be prescribed and/or provided by a clinician to aid the patient in reducing pain and returning to normal gait include: the prescription of a cane, walker, or other type of ambulatory device; medications to reduce pain and swelling; and modified activity/exercises (Grabli et al., 2012). The goal of these interventions is to reduce symptoms and increase balance, strength, and mobility. For the purposes of this research, we will focus on the prescription of canes as a method of treating pathological gait.

Cane Prescription

Canes are prescribed for a wide variety of conditions and can be used depending on the patient's level of balance. Canes are useful for patients who have pain, vestibular dysfunction, visual impairment, sensory deficit, or an antalgic gait pattern (Dean & Ross, 1993); however, they are most commonly used when treating hip and knee osteoarthritis (Aragaki et al., 2008). Antalgic gait patterns may be the result of many pathologies including fractures, muscle strains, ligament sprains, or following surgical interventions.

The use of a cane and the method by which this is done may also have an effect on the user's gait pattern. The effect that is seen can be beneficial or detrimental to the user's condition depending on whether proper cane use techniques are followed or not (Gitlin & Burgh, 1995; Mann et al., 1993; Schemm & Gitlin, 1998). It is for this reason that canes must be properly fitted and proper walking techniques must be instructed by a practitioner before a patient is allowed to use a cane.

Cane fitting. Fitting a cane to its user involves determination of height and angle at the level of the elbow. The most accepted method for determining the correct height of a cane is to set it equal to the distance between the greater trochanter of the patient's femur and the floor, measured when the patient is wearing walking shoes (Teodoro, Tomazini, Galera, &

Nascimento, 2012). This length is defined as the vertical distance from the most prominent part of the greater trochanter to the ground. When the cane height is correctly determined, the patient should be able to maintain the elbow in 15 to 30 degrees of flexion while the cane is in contact with the ground (Teodoro et al., 2012). Once the cane is properly fitted, the patient is given instructions regarding proper cane-assisted locomotion.

Proper cane-assisted ambulation technique. When standing with a cane, patients are instructed to hold the cane in the hand that is on the same side as the uninjured limb. During normal locomotion, humans swing the left hand with the right foot and vice versa (Cifu, 2016). Therefore, orientating the cane in such a manner is suggested as it maintains natural arm movement. This manner of holding the cane is also beneficial in terms of decreasing GRFs on the hip (Edwards, 1986).When walking, patients are instructed to step forward with the injured leg, while simultaneously moving the cane forward and distributing the weight between these two points of support (Au, Wu, Batalin, & Kaiser 2008; WikiHow, 2015).

Types of Canes

The specific type of cane that is prescribed depends on the injury or clinical pathology that the individual has. Different types of canes currently on the market include the folding, forearm, quad, and tripod cane as depicted in Figure 7 (Inverarity, 2015). The folding cane has several joints that are linked by an internal elastic cord enabling them to be folded when not in use (Inverarity, 2015). The forearm cane differs from the standard single point cane with the addition of a forearm support, enabling a shift in the load from the wrist to the forearm (Inverarity, 2015). The quad cane has four ferrules at the base, allowing it to be more stable (Inverarity, 2015). The last type of cane, the tripod cane, has a three pronged base (Inverarity, 2015). Most of these canes are adjustable allowing for a specific degree of elbow flexion when the user is standing.

Figure 7. Types of canes (left to right): folding cane; forearm cane; quad cane; and tripod cane. Reprinted from 5 Types of Canes and Their Uses, from Access Ability Home Medical Products, n.d., retrieved December 15, 2015, from http://www.hmestore.net/5-types-of-canes-and-their-uses/.

Nolen et al. (2010) compared single-tip, tripod, and quad canes and suggested that cane type can have an effect on velocity, cadence, stance, and swing time. In the order of walking without a cane, walking with a single-tip cane, walking with a tripod cane, and walking with a quad cane, subjects demonstrated a significantly decreased velocity, cadence, and an increased stance and swing time. There was no significant difference in stride and step lengths between any of the canes. Furthermore, using a quad cane resulted in a much slower velocity and decreased cadence with longer stance time than using a single-tip cane or tripod cane. There was also no difference reported between a single-tip cane and tripod cane (Nolen et al., 2010).

Handle type. As with variations in base type, canes can also differ in terms of the handle type. There are many different types of cane handles that are currently available for purchase. The type of handle used depends on whether it is meant for weight bearing, balance, aesthetic, or fashion. The different types of cane handles include: Hook, Derby, Fritz, Anatomical, and T-handle as depicted in Figure 8; the different types of handles can have an effect on the biomechanics of cane-assisted locomotion.

Figure 8. Different types of cane handles (left to right): Derby; Fritz; Anatomical; Hook; and T-handle. Reprinted from Cane Handle Guide, in CanesCanada, n.d, Retrieved April 4, 2015, from http://canescanada.com/Cane-Handle-Guide_ep_45-1.html.

Researchers Chiou-Tan, Magee, and Krouskop (1999) have investigated differences in upper limb muscle activity among different handle types (traditional Fritz handle and two prototype handles). Muscle activity was measured through root mean square (RMS) voltage muscle output. The findings of this research suggested that the two prototype cane handles significantly decreased RMS voltage muscle output in the upper limb (Chiou-Tan et al., 1999). The first prototype (illustrated in Figure 9) positioned the wrist in extension and was based on an infantile crawling pattern. In the infancy stage, humans crawl on their hands with the wrists extended. This allows the infant to bear weight predominantly through the wrist rather than the hand. The second prototype was based on the walking pattern of gorillas, who bare weight through their knuckles (Chiou-Tan et al., 1999). It is proposed that this walking pattern assists in maintaining neutral wrist alignment and allows for weight bearing through the long bone axis.

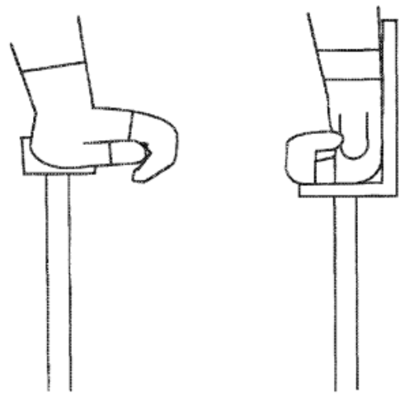

Figure 9. Two prototype canes used in the aforementioned study. These prototypes differ from the standard cane by their handle type. Reprinted from "Comparison of Upper Limb Muscle Activity in Four Walking Canes: A Preliminary Study," by Chiou-Tan et al., 1999, *Journal of Rehabilitation Research and Development, 36*(2), 94-99.

When using a standard cane, individuals hyperextend their wrist causing a large magnitude of the force to be placed through the metacarpal bones. Using the cane in this manner creates torques at the wrist joint and increases the probability for injury (Chiou-Tan et al., 1999). The findings from this study demonstrated the need for further research and to possibly develop an alternate handle type in order to facilitate better rehabilitation minimizing the negative effects of using a cane. Decreasing the forces on the upper limb and the muscle activity could lead to decreased discomfort and pain; however, more research is required in this field.

Advantages of Cane Use

The improvement of mobility is the greatest known reason for which canes are used and prescribed (Joyce & Kirby, 1991); however, these assistive devices can be used for a variety of purposes. Bradley and Hernandez (2011) suggested that canes are most effective in improving stability by increasing the size of the support base, redistributing weight from a lower limb that is either weak or painful, aiding in propulsion and breaking during gait, and providing tactile information about the ground. Tactile information is not only useful for blind cane users, but also for those individuals who have difficulty maintaining balance. Furthermore, the ability of a cane to decrease weight bearing on one leg can result in decreased pain (Jones et al., 2011). Generally, canes are prescribed for individuals who are exhibiting various levels of impairment (Joyce & Kirby, 1991).

Canes have also been directly linked with both physical and psychological benefits for users. The psychological benefits of cane use can be observed in older adults who, with the use of a cane, have reported increased subjective levels of confidence and feelings of safety (Aminzadeh & Edwards, 2011). This improvement in confidence could in turn lead to increased levels of independence and activity (Dean & Ross, 1993; Tinetti & Powell, 1993). The physiological benefits of cane use are a direct result of enabling the user to ambulate (Jaeger, Yarkony, & Roth, 1989). Studies have shown that the mere continuation of ambulatory practices can lead to the prevention of osteoporosis, cardiorespiratory deconditioning, and enhanced circulation (Jaeger et al., 1989).

Increased stability. One of the major clinical uses of a cane is the improvement of stability by increasing the BOS which is defined as the area that lies within an outline surrounding all ground contact points (Joyce & Kirby, 1991; King, Judge, & Wolfson, 1994). Balance is often thought of as a regulation of an individual's COM within his/her BOS (MedicineNet, 2012). To achieve postural equilibrium in a static position (i.e., reducing the amount of net forces acting on the body), the individual must be capable of positioning his/her COM over his/her BOS. Postural instability, or loss of balance, can result when the COM is displaced from its location over the BOS by a sudden movement or external perturbation (e.g., slips, trips, pushes; Winter, 1995).

When standing, humans usually have two feet in contact with the ground. When stability is challenged, the general reaction is to spread the feet apart in order to regain stability (van Dieën, Pijnappels, & Bobbert, 2005). By doing this, the individual is increasing the size of the BOS. Older or injured individuals often use canes to add additional ground contact points to the system, thereby, increasing their BOS as depicted in Figure 10 (Bateni & Maki, 2005). The effect of this increase on the BOS is particularly noticeable during the single support (swing) phase of gait. The mobility aid allows the user to keep the COM within the BOS limits for a greater proportion of the gait cycle (Bennett, Murray, Murphy, & Sowell, 1979).

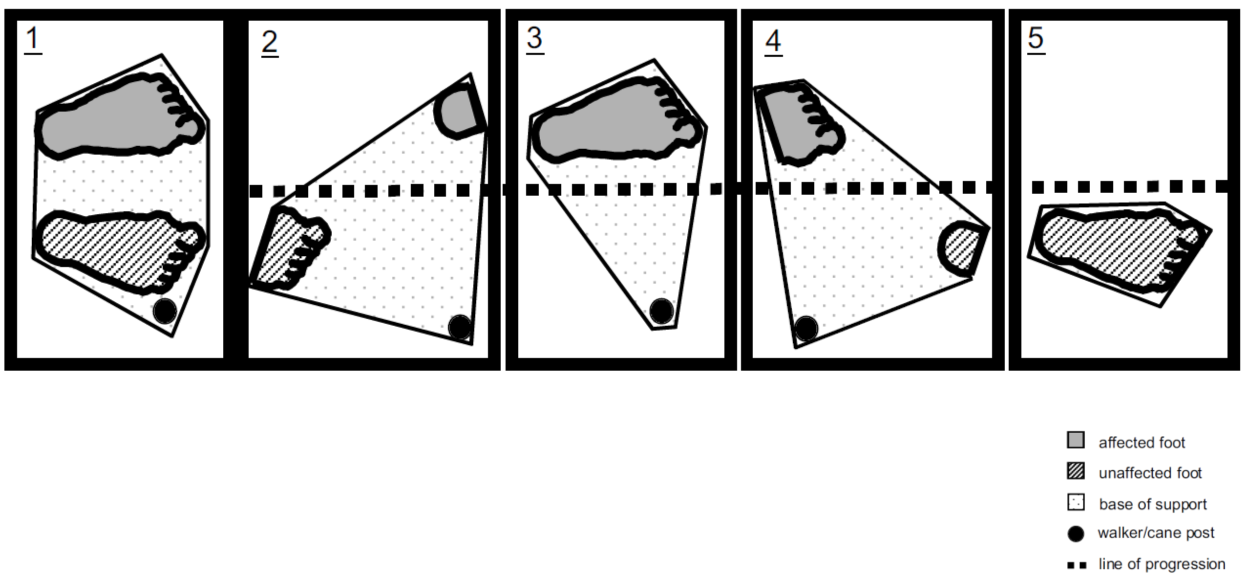

Figure 10. Increase of BOS with the addition of a cane; a third ground contact point is achieved and the BOS of the user is significantly increased. Reprinted from "Assistive Devices for Balance and Mobility: Benefits, Demands, and Adverse Consequences," by H. Betani and B. Maki, 2005, *Archives of Physical Medicine and Rehabilitation, 86*(1), 134-145.

Other than increasing the BOS, it is proposed that canes may be able to improve stability by providing additional stabilizing forces at the level of the hand (Bennett et al., 1979). The vertical and horizontal GRFs produced at the level of the wrist (as denoted by F_{cv} and F_{ch} in

Figure 11) act to oppose the downward and lateral motion of the COM that occurs during single-leg support (Bennett et al., 1979).

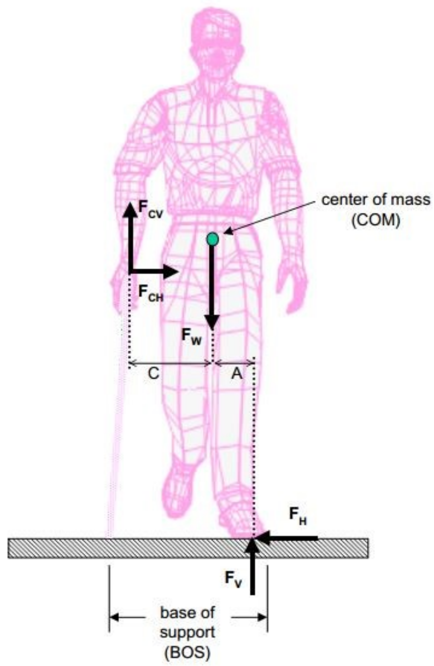

Figure 11. Ground reaction forces during cane-assisted locomotion. Reprinted from “Assistive Devices for Balance and Mobility: Benefits, Demands, and Adverse Consequences,” by H. Betani and B. Maki, 2005, *Archives of Physical Medicine and Rehabilitation, 86*(1), 134-145.

In terms of static balance, cane use has been reported to reduce the displacement of the COM in a study involving 24 stroke patients (Lu, Yu, Basford, Johnson, & An, 1997; Milczarek, Kirby, Harrison, & Macleod, 1993). Ashton-Miller et al. (1996) found that with the use of a cane, patients who had peripheral neuropathy in the lower extremity were able to maintain equilibrium as they transferred from a double-leg stance to a single-leg stance on an unsteady surface. Kuan, Tsou, and Su (1999) reported a stabilizing effect with the use of a cane in 15 stroke patients who exhibited increased step length and decreased step width in comparison to a control group; however, interpretation of these results may be affected by other factors such as the slower cadence in stroke patients.

Lower-limb load reduction. The ability to reduce loads on the lower limb may be beneficial for individuals who have sustained an injury or are presenting with weakness or joint pain in this area. By supporting a percentage of an individual's body weight, a cane can reduce the vertical GRF exerted on the supporting leg. Research completed on individuals with a variety of hip disorders determined that peak GRFs on the lower limb were reduced in both static and dynamic (ambulation) conditions when a cane was used (Aragaki et al., 2008). This concept is also illustrated in Figure 11 as the vertical GRF acting on the lower limb (F_v) is equal to the body weight minus the vertical force produced at the wrist (F_{cv}). This means that the loading of the cane can reduce the vertical GRF acting on the supporting limb; however, decreased limb loading does not equate with a reduction of loads placed on the hip joint. This is because the amount of load on the hip is highly dependent on hip abductor muscle activity (Neumann, 2015). In summary, the level of abductor muscle activity is dependent on the side of the body on which the cane is held (Neuman, 1999; Nordin & Frankel, 1991; Röhrle, Scholten, Sigolotto, Sollbach, & Kellner, 1984). This is an important point to consider since most clinical research is concerned with decreasing forces on the hip and, thereby, subsequent pain.

Some evidence suggests that a cane's ability to lower loads on the lower-limb is highly dependent on the orientation of the cane to the user. That is, the benefits associated with cane use depend on whether the cane is held on the contralateral or ipsilateral side of the injured limb. Harrison (2004) examined the best method of using a cane in order to shift the GRFs away from the injured leg. Harrison concluded that the use of a cane on the contralateral side produced less force on the femoral epiphysis and decreased force produced by the abductor muscles on the same leg. Harrison also suggested that an individual with a lower limb injury would be able to decrease the force placed on his/her ankle by almost half by using a cane on the contralateral side of the injury. Holding a cane ipsilateral to the side of the injured limb has actually been shown to increase the amount of force placed on the affected hip joint (Vargo, Robinson, & Nicholas, 1992). Conversely, holding the cane contralaterally can reduce the forces acting on the hip by up to 60% when compared with walking without a cane (Radin, 1979). More recently, the influence of cane orientation on hip abductor muscle activity was recorded using integrated rectified surface EMG activity from the various muscles surrounding the knee during a variety of standing manoeuvres. The findings of this study showed that hip abductor muscle activity was the lowest when maximal weight was placed through a cane held on the contralateral side and highest with maximal weight placed through the ipsilaterally positioned cane (Vargo et al., 1992).

Propulsion and breaking during gait. Using a mobility aid to generate horizontal GRFs can help to provide propulsive and/or breaking forces during gait (Bennett, Murray, Murphy, & Sowell, 1979). This component of cane use is particularly beneficial for individuals who have difficulty initiating or terminating a movement due to pain, muscle weakness, or impaired motor control in the lower limbs. Additional horizontal GRFs could also help an individual achieve smoother and more efficient movement of the body during gait (Bennett et al., 1979; Chen, Chen, Wong, Tang, & Chen, 2001; Melis, Torres-Moreno, Barbeau, & Lemaire, 1999).

Bennett et al. (1979) studied the AP cane impulse generated by nine subjects with hip pain. The results of this study found that subjects with hip pain tended to apply larger propulsive impulses rather than braking impulses (Figure 12). In contrast, Chen et al. (2001) found that 20 stroke patients tended to generate larger breaking impulses (Figure 13). Chen concluded that the difference between these two studies was attributable to the fact that stroke patients relied primarily on the unaffected limb to generate propulsive forces and used the cane to aid in decelerating the motion; whereas patients with hip pain tended to use the cane to reduce the required joint forces when pushing forward with the painful limb.

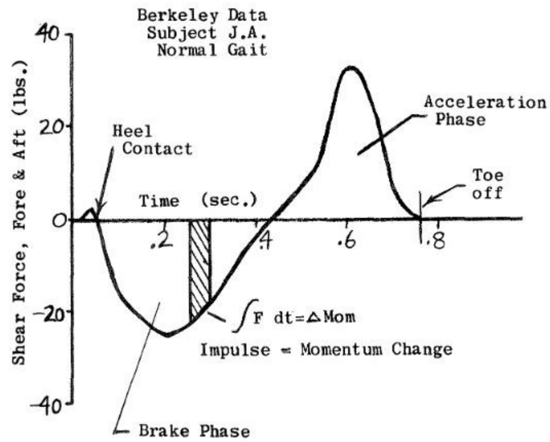

Figure 12. Anterior-posterior GRF in patients with hip pain. Reprinted from “Locomotion assistance through cane impulse,” by L. Bennett, M. Murray, E. Murphy, and T. Sowell, 1979, *Bulletin of Prosthetic Research, 10*(31), 38-47.

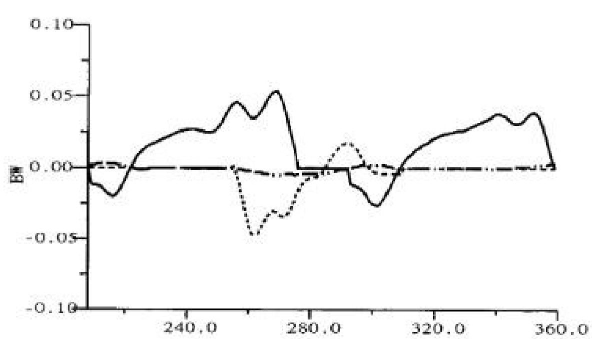

Figure 13. Anterior-posterior GRF in patients with stroke. Reprinted from “Temporal stride and force analysis of cane-assisted gait in people with hemiplegic stroke,” by C. Chen, H. Chen, M. Wong, F. Tang, and R. Chen, 2001, *Archives of Physical Medicine and Rehabilitation, 82*(1), 43-48.

It is important to note that the capacity to produce these breaking and propulsive forces is highly dependent on the user’s ability to hold the cane at an appropriate angle. Ely and Shmidt

(1977) determined that the horizontal component of the axial force made a much larger contribution to propulsion, provided that the cane is tilted forward. Similarly, the horizontal component of the axial force can help in braking if the cane is tilted backward (Ely & Shmidt, 1977); however, it is unclear as to whether the subjects in the aforementioned studies were taught to use the cane in a particular manner or use was learned through experience. This should be addressed in future studies.

Tactile benefits. The central nervous system maintains stability by processing external information regarding the position and the movement of the body's segments. The methods by which this data is collected are through the visual, vestibular, and somatosensory systems (Massion, 1998). Jeka (1997) found that tactile information from the hand contributed to postural stabilization. Additionally, light touch of the fingertip to any external surface (i.e., wall) has also been shown to significantly reduce COM displacement associated with the control of postural sway in five adults aged 20 to 50 years (Jeka & Lackner, 1994). These effects were observed in both open eyes and closed eyes conditions, although they were more pronounced when vision was deprived. Jeka et al. (1996) studied the effect of tactile cues derived from a cane, and required that their subjects maintain a static posture. Their results indicated that touch contact of a cane was equally as effective at low force as higher force conditions in terms of controlling postural sway when vision was deprived (Jeka et al., 1996). This study indicated that the contribution of tactile cues may be useful not only in creating biomechanical advantages, but also in providing additional information for the central nervous system to control of balance (Jeka, 1997). However, there are limitations to most studies in this area as sample size is often limited and the focus is geared towards static stance on stationary surfaces. Further research is required in this field simulating more complex, dynamic situations in order to assess the effect of ongoing changes in the position and orientation of the body on the capacity of the central nervous system to utilize the tactile cues provided by mobility devices.

Disadvantages of Cane Use

Despite the proposed advantages associated with cane use, research has shown that there is a high rate of abandonment and disuse of mobility aids. These high rates of disuse are a cause for concern and raise questions regarding the device's effectiveness and design. The most common complaints that cane users report include increased pain and development of new injuries in compensatory structures such as the supporting upper limb (Chen et al., 2001). The reported pain is often associated with the development of certain pathologies that are caused by repetitive stresses placed on the joints of the upper extremity. Chiou-Tan et al., (1999) suggested that chronic cane use is linked with the development of pathologies such as tendonitis, osteoarthritis, and carpal tunnel syndrome. Individuals with arthritis, who often use canes to reduce weight bearing on their lower limbs, are at particularly high risk of developing joint inflammation in the upper limb resulting from repetitive forces (Florack, Miller, Pellegrini, Burton, & Dunn, 1992; Yosipovitch & Yosipovitch, 1993). In a study of long-term poliomyelitis patients, 64% of individuals reported upper limb pain associated with the use of mobility aids (Koh et al., 2002). Upper limb loading can even lead to fracture in some cases, as evidenced by an anecdotal report of scapular-body stress fracture that occurred with extensive cane use (Parr & Faillace, 1999). It is important to note that inappropriate device prescription, inadequate user training, or use of unprescribed devices may exacerbate the problems listed above (Gitlin & Burgh, 1995; Mann et al., 1993; Schemm & Gitlin, 1998). Other disadvantages resulting from cane use include decreased stability, increased demands placed on the upper limb, and increased demands placed on the contralateral (uninjured) lower limb which are discussed in greater detail below.

Decreased stability. Although it has been stated that cane use is associated with increased stability, recent studies indicated that canes may also hinder stability through different mechanisms (Loram, Maganaris, & Lakie, 2004). The proposed mechanism by which this occurs is a consequence of the combined inertia and weight of the individual's arm and mobility device. The act of lifting the cane from the ground and advancing it anteriorly creates reaction forces that may perturb the COM. Under normal conditions, the human body is capable of anticipating these perturbations in the COM and make the necessary adaptations to maintain balance (Loram et al., 2004). The use of a cane can alter the body's ability to anticipate perturbations and maintain balance. That is, as the cane is removed from the walking surface, the BOS is suddenly reduced and a state of imbalance is created. This state occurs as the COM is quickly forced outside of the limits of the BOS (Bouisset & Zattara, 1981). The body is then less equipped to adjust to this sudden state of imbalance, possibly resulting in a fall or stumble (Bouisset & Zattara, 1981).

Unanticipated protuberances in the walking surface could also result in a large horizontal force applied to the cane causing the device to suddenly slip. Accidental contact between the cane and objects in the environment can be another source of perturbation to the individual's postural control. Many studies have reported that canes and environmental obstacles may be associated with falls (Campbell, Borrie, & Spears, 1989; Campbell, Reinken, Allan, & Martinez, 1981); however, the link between these two factors has not been established within the literature.

When balance is interrupted, stabilizing joint movements are generated by postural reactions at the ankle, hip, lumbar spine, and cervical spine (Nashner & McCollum, 1985). In some situations (if the perturbation is large, or weakness or impaired neuromotor control is present), these stabilizing movements may not be sufficient to recover equilibrium. In these situations, the individual may step forward or rapidly reach and grasp an object within the environment for support. In the case of compensatory stepping, the cane has the potential to impede the movement, resulting in a fall (Nashner & McCollum, 1985). Bateni et al. (2004) studied the effect of a cane on an individual's capacity to recover from a perturbation by stepping laterally. Ten healthy young adults were tested using lateral platform perturbations. This study found that collisions of the cane with the surrounding surfaces led to a significant reduction in lateral step length (26-37%) when compared to the no collision trials.

Upper limb demands. The current available literature examining upper limb joint loading and strength demands often infers these factors from measurements of force applied to the device. There have been several published studies, which investigated the amount of loading that is applied to the device during cane-assisted locomotion (Anglin & Wyss, 2000; Bennett et al., 1979; Chen et al., 2001; Edwards, 1986; Ely & Smidt, 1977). Most of these studies indicated that users rarely placed greater than 15-20% of their bodyweight on the cane during normal assisted locomotion; however, it is pertinent to note that the amount of loading placed on the upper limb is dependent on the type of disability present. For example, the highest loads placed axially through a cane were reported in individuals who were in the postoperative stage of knee or hip replacement. These individuals, on average, placed 31% of their body weight on the mobility device (Edwards, 1986). Another factor that may provide an explanation for the variation in the amount of cane loading between studies is the walking speed. In a study examining 20 stroke patients, walking speed was very low and associated with relatively low amounts of axial loading on the cane when compared to other studies (Chen et al., 2001).

A limited number of studies directly addressed upper limb joint loading and strength requirements associated with cane use. The data reported in these trials suggests that joint forces may be very high when loading the cane. Anglin et al. (2000) tested muscle forces acting at the shoulder in six healthy adults and found that the glenohumeral joint contact force reached up to three times one's body weight during cane-assisted locomotion. External moments at the shoulder were also quite high and comparable to lifting a 10 kg object. Kinematic analysis of cane-assisted locomotion revealed that the elbow is typically flexed and the wrist extended, suggesting that significant demands are placed on the elbow extensors and wrist flexors (Anglin & Wyss, 2000; Bachschmidt, Harris, & Simoneau, 2001). Few studies have directly measured EMG muscle activation during cane use, but one such study reported that activation levels could be reduced by changing the design of the cane handle (Chiou-Tan et al., 1999).

Demands on the contralateral lower limb. When a clinician recommends a cane to his/her patient, he/she often instructs the patient to place a greater amount of the body weight on the unaffected limb (Hoeman, 2008). Some individuals believe that this uneven distribution of loading can cause new symptoms to appear in the unaffected limb. There is no scientific basis for such reasoning. In fact, research shows that lessening the force and load on one leg does not necessarily equate to an increase of loading on the other (Youdas, Kotajarvi, Padgett, & Kaufman, 2005). Harrison and Harris (1994) examined patients who had a paralysed and shortened limb from poliomyelitis and confirmed that the force transmitted to the affected leg was reduced, but the force in the opposite leg was the same as that generated in normal individuals. These findings were similar in individuals with an antalgic gait pattern resulting from arthritis (Harrison & Harris, 1994).

Recognizing the proven disadvantages of cane use, cane designers have attempted to modify the traditional single-tip design. The most novel modification is the addition of a spring loading device to the base of the shaft. The goal of this modification was to remedy some of the associated disadvantages.

Spring Loading

A spring is an elastic object used to store mechanical energy (Xie, Ko, & Du, 2013). When a spring is compressed or slightly stretched from rest, the force it exerts is approximately proportional to its change in length (Xie et al., 2013). The goal of adding a spring mechanism to a cane is to store the energy of the impact from cane strike and use this elastic energy to provide propulsion after the midstance stage of ambulation (Liu et al., 2011). This cycle of storing and releasing mechanical energy is thought to reduce the magnitude of the impact during the initial contact phase and propel the body after midstance. Furthermore, the spring mechanism is hypothesized to reduce extra push-off being exerted by the upper extremities after midstance, thus, reducing the incidence of upper extremity injuries. There are several canes that are currently on the market with spring loaded shafts; however, the research supporting these types of canes is limited to testimonials and case studies (StanderCane, 2016).

Studies performed on such mechanisms in axillary crutches report that these crutches are a more comfortable alternative to the standard axillary crutch (Seeley et al., 2011). The addition of in-line springs to these assistive devices has been shown to reduce the impulse and rate of GRF rise during ambulation by 13-26% (Segura & Piazza, 2007). These decreases are thought to significantly limit the jarring movements seen with standard crutch use and, thereby, decrease the likelihood of overuse injuries. Further literature also suggests that handgrip force and stride length are decreased when spring loaded crutches are used (Parziale & Daniels, 1989). In summary, the use of spring loaded crutches has been shown to alter the mechanics of crutch gait in ways that are likely to reduce injury in crutch users.

The propulsion effect of these spring mechanisms is highly dependent on the property of the spring. If the spring mechanism is too stiff, the initial ground contact yields a large impulse that is projected back onto the body (Segura & Piazza, 2007). On the other hand, if the spring is too soft, the stored elastic energy would be insufficient for propulsion. Stoer and Bulirsch (1980) attempted to design an axillary crutch with optimal spring properties. The goal was to design a crutch that could store the energy of the impact from the crutch strike as elastic energy and use this stored energy to provide propulsion after the midstance of ambulation. Simulation results of this study found that the optimal spring stiffness for a user of normal weight (58-88 kg) was about 4-4.5 kN/m; this result was experimentally validated by Liu et al., (2011), who compared GRFs produced during traditional and springy crutch use.

Liu et al., (2011), demonstrated that the spring loaded crutches provided an effective propulsive mechanism. This mechanism was evidenced by the difference in vertical GRFs. Specifically, two peaked profiles of vertical GRFs were observed during ambulation with standard crutches, similar to the GRFs observed during normal human walking. In contrast, only one peak was observed in the GRF profile for the optimal crutch during the midstance phase of gait. The single peak profile suggested that the spring reduced the magnitude of the impact during the initial contact phase. Furthermore, it also indicated that the stored elastic energy during the impact stage was converted to mechanical energy to propel the body after midstance, with reduced extra push-off being exerted by the upper extremities after midstance. The propulsion effect observed in this study was also thought to reduce the total metabolic energy expenditure in crutch walking (Liu et al., 2011).

To determine if spring loaded crutches have an effect on metabolic outcome, Seeley et al. (2011) examined metabolic energy expenditure during spring loaded crutch ambulation. The purpose of this study was to determine whether the novel spring loaded crutches reduced oxygen consumption during crutch ambulation, relative to traditional crutch ambulation. A secondary purpose was to evaluate the design for subject-perceived comfort and ease of use. The findings indicated that compared with traditional axillary crutches, the spring loaded crutch was more comfortable but did not appear to benefit subjects via reduced metabolic energy expenditure (Seeley et al., 2011).

The mechanical advantages provided by the addition of a spring loading mechanism to crutches suggest possibilities for improvements of other rehabilitative devices. These improvements may be particularly beneficial in the case of single-tip support canes as the associated disadvantages (demands of the upper limb) could be reduced via the addition of a spring loading mechanism.

Research Problem

Current single-tip support cane designs are associated with pain and development of certain pathologies in compensatory structures such as the upper limb. A large portion of the related literature suggests that these upper limb pathologies occur due to repetitive stresses placed on this limb during cane-assisted locomotion (Chiou-Tan et al., 1999; Florack et al., 1992; Gitlin & Burgh, 1995; Koh et al., 2002; Mann et al., 1993; Parr & Faillace, 1999; Schemm & Gitlin, 1998; Yosipovitch & Yosipovitch, 1993). Specifically, chronic cane use is linked with the development of certain pathologies such as tendonitis, osteoarthritis, and carpal tunnel syndrome. Individuals with arthritis, who often use canes to reduce weight bearing on their lower limbs, are at particularly higher risk of developing joint inflammation in the upper limb resulting from the repetitive forces.

Creators of spring loaded canes and users anecdotally claim that the addition of spring mechanisms enable the canes to decrease the amount of force transmitted to the upper limb thereby reducing pain and the likelihood of developing pathologies (StanderCane, 2016). These claims are substantiated by previous research performed on spring loaded crutches; however, no research has been completed examining the relationship between spring loaded canes and force on the upper extremity and upper limb EMG muscle activation. Furthermore, previous investigations of the mechanics of spring loaded crutches are limited in their scope. For example, Parziale and Daniels (1989) did not measure ground reaction forces during ambulation with spring loaded crutches and Shoup (1980) evaluated the ground reaction forces in only a single subject. Segura and Piazza (2007) built upon previous research in this field by studying the differences in ground reaction force, rate of force rise, impulse, and spatiotemporal gait variables between standard and spring loaded crutches. The researchers involved in the proposed current study planned on adapting and expanding upon the research performed by Segura and Piazza (2007). This was done by determining if impulse, force, and upper extremity muscle activation differ between traditional and spring loaded canes, while also assessing the differences (in terms of ease-of-use) subjectively in order to address clinical significance.

Establishing a relationship between spring loaded canes, upper limb ground reaction forces, and EMG muscle activation is an important first step that will set the stage for subsequent studies intending to show how these canes reduce pain and pathologies empirically. The findings of such research may have implications on the rates of abandonment and disuse such that users would be more likely to continue using their assistive devices as negative feelings associated with upper limb pain may be minimized or eliminated.

Purpose of the Research

The primary purpose of this study was to determine the effect of commercial spring loaded single-tip support canes on minimizing GRFs, impulse, and muscle EMG activation in the upper limb during ambulation. Ground reaction forces and impulse were also assessed for the injured lower limb in order to compare the off-loading capabilities of each cane. A secondary purpose was to assess both traditional and spring loaded cane designs for subject-perceived ease of use in order to determine if any differences exist.

Research Questions

The following research questions were used to guide the study:

1) Is there an interaction effect between type of cane (traditional, Miracle Cane®, and Stander Cane®) and muscle (flexor carpi radialis, extensor carpi radialis longus, brachioradialis, triceps brachii, infraspinatus, and pectoralis major) on upper extremity EMG muscle activity?

2) Is there an interaction effect between type of cane and extremity (upper or lower) when measuring forces in the vertical, AP, and medial-lateral planes?

3) Is there an interaction effect between type of cane and extremity when measuring impulse in the vertical and AP planes?

4) Is there a difference between cane types in terms of subject-perceived ease of use?

Chapter 2 - Method

Participants

Twenty one healthy participants were recruited to partake in this study to examine and compare different types of canes in relation to upper extremity muscle activity and ease of use. Upper and lower extremity measures of force and impulse across different types of cane were also examined. The lower extremity injury was simulated by placing a knee brace on a healthy participant's knee causing him or her to limp.

Inclusion criteria. Healthy males and females aged 18-45 years with the ability to ambulate unaided without gross deviation; the ability to understand verbal and written instructions; and the capacity to give informed consent were considered eligible for this study. Participants were selected on the basis of their bodily dimensions, such that the stature of each participant was suited to the size of the modified and standard canes (Shortell, Kucer, Neeley, & LeBlanc, 2001). Specifications for the canes being used in this study recommend a maximum weight of 300 lbs and a height range of 4' 6" to 6' 6" (StanderCane, 2016). The spring/weight relationship protocol stated by Shortell et al. (2001) was also used to determine weight parameters for the spring loaded canes (Appendix A). This spring/weight relationship protocol outlined a range of weights that are appropriate for certain spring constants.

Exclusion criteria. Individuals with a history of knee trauma or surgery within the past six months; the presence of an active inflammatory rheumatological condition; injury or amputation of the lower extremity; injury or condition of the upper extremity affecting the ability to use a cane; spinal or lower quadrant pain impeding gait; the presence of a neurological condition impeding gait; or poor health interfering with a gait assessment were excluded from the study. Potential participants were screened for these criteria during a preliminary meeting preceding the testing session. The demographic questionnaire (used to screen for the exclusion criteria) was administered to the participants once they read the letter of recruitment and provided consent.

Participant Demographics. Males (n=9) and females (n=12) were recruited for the purposes of this study. These participants fell between the ages of 19-45 years. Participant heights ranged from 157.5-188 cm. Participant weights ranged from 54-113 kg.

Instrumentation

For this study, the following instruments were used:

Types of canes. The three types of canes used were the traditional, Miracle Cane®, and Stander Cane®. These types of canes are very similar in handle and base styles. That is, all of these canes possess a Fritz handle type and a single-point base. The Miracle Cane® and Stander Cane® differ from the traditional cane by way of possessing a spring loaded mechanism within the shaft.

T-scope knee brace. This type of knee brace is designed to provide controlled range of motion for patients recovering from knee surgery or those who have knee injuries or instabilities. The brace controls the range of motion of the knee through a hinge mechanism. Range of motion can be limited in both flexion and extension at the knee via a locking mechanism limiting the range of motion between 0-120 degrees. For the purposes of this study, this knee brace was used to simulate the presence of a unilateral 30 degree knee flexion contracture and antalgic gait pattern. This injury simulation was done by limiting the amount of extension possible at the knee joint.

Advance Mechanics Technologies Incorporated force plate and Biosoft software. The Advance Mechanics Technologies Incorporated (AMTI) force platform measures the GRFs that are generated by a body standing or moving across the plate. The AMTI force plate is also capable of measuring torques about each axis (vertical, AP, and medial-lateral) by sensing the position of the foot on the platform. There are many different kinds of force platforms available such as strain gauge, piezoelectric, piezoresistive, or capacitive (Marasovic, Cecic & Zanchi, 2009). All of these varieties operate based on the principle that the applied force causes a certain amount of strain within the transducer. Using the appropriate technology, this strain is translated into a signal that is proportional to the applied force (Marasovic et al., 2009). Advance Mechanics Technologies Incorporated force platforms are often used to quantify gait, balance, and other parameters of biomechanics. These force plates have been shown to have a high inter-rater reliability of .90 and a high intra-rater reliability of .95 when measuring GRFs during locomotion and jumping tasks (Hansen, Cronin, & Newton, 2011). The Biosoft software is used in conjunction with the force platform and allows for the analysis of gait and balance. This program allows the researcher to view and analyze their data post collection. The data is graphically displayed in a force-time axis, which indicates the magnitude of force produced throughout the movement of interest.

Electromyography. Surface EMG can be recorded by inserting electrodes directly into the muscle (termed indwelling electrodes or fine wire EMG), or alternatively using electrodes placed on the surface of the skin (Criswell & Cram, 2011). The EMG signal is recorded by these electrodes in response to muscle activity. During muscular activation, each nerve becomes excited, signalling and enabling the muscle to stretch or contract (Criswell & Cram, 2011). The activation of the muscle fiber by nerve endings induces waves of depolarization. The electrical signals related to depolarization of the muscle fiber can be recorded by electrodes on the skin; however, the muscle contraction itself is much slower than the cycle of depolarization.

Consequently, it cannot be said that there is a direct relationship between EMG and force (Criswell & Cram, 2011). Rather, the EMG signal infers force parameters by measuring changes in the number of recruited muscle units and frequency of recruitment.

For the purposes of this study, Delsys hybrid wireless surface EMG sensors were used in conjunction with EMGworks software for data acquisition and analysis. Hybrid sensors have the ability to collect EMG and triaxial accelerometry data simultaneously. The parallel bar design used in these sensors allows for high fidelity signals. Surface EMG measures have demonstrated strong evidence of reliability in a number of studies. Spector (1979) used surface electrodes to assess paraspinal muscle activity that yielded correlation coefficients ranging from .73 to .97. Other researchers have examined the reliability of surface EMG electrodes in the leg, torso, and arm muscles during running and have determined that this method displayed strong reliability measures (intra class coefficient (ICC) > .80) for all parameters studied (Smoliga, Myers, Redfern, & Lephart, 2010).

PowerLab data acquisition system. The PowerLab data acquisition system is often used for signal processing, data recording, display, and analysis features for a wide variety of research applications. In conjunction with LabChart software, this system can be used to collect up to 32 channels of data in real time. For the purposes of this study, the PowerLab and LabChart systems were used to simultaneously collect and synchronize the EMG data and data from the force plate. Figure 14 illustrates the configuration of the systems that allowed for this method of data collection. The synchronization of these systems was performed with the use of an interface board. The wireless EMG sensors were connected to the interface board via the Trigno Base Station Receiver, which was in turn connected to the PowerLab system. The force plate was then directly connected to the PowerLab equipment, allowing for the integration of all systems into the LabChart program.

The EMG and force data continued to be collected through EMGworks and Biosoft separately from the LabChart program. The data from these external programs was used to calibrate the data obtained from the LabChart program. Calibration is a necessary procedure as the LabChart program assumes all input data is measured in volts. In order to convert volt measures into their correct respective measures, a ratio must be created using the original raw data from EMGworks and Biosoft. That is, a ratio of volts/Newtons (in the case of force and impulse calculations) was inputted into the LabChart software in order to make the conversion from volts to Newtons.

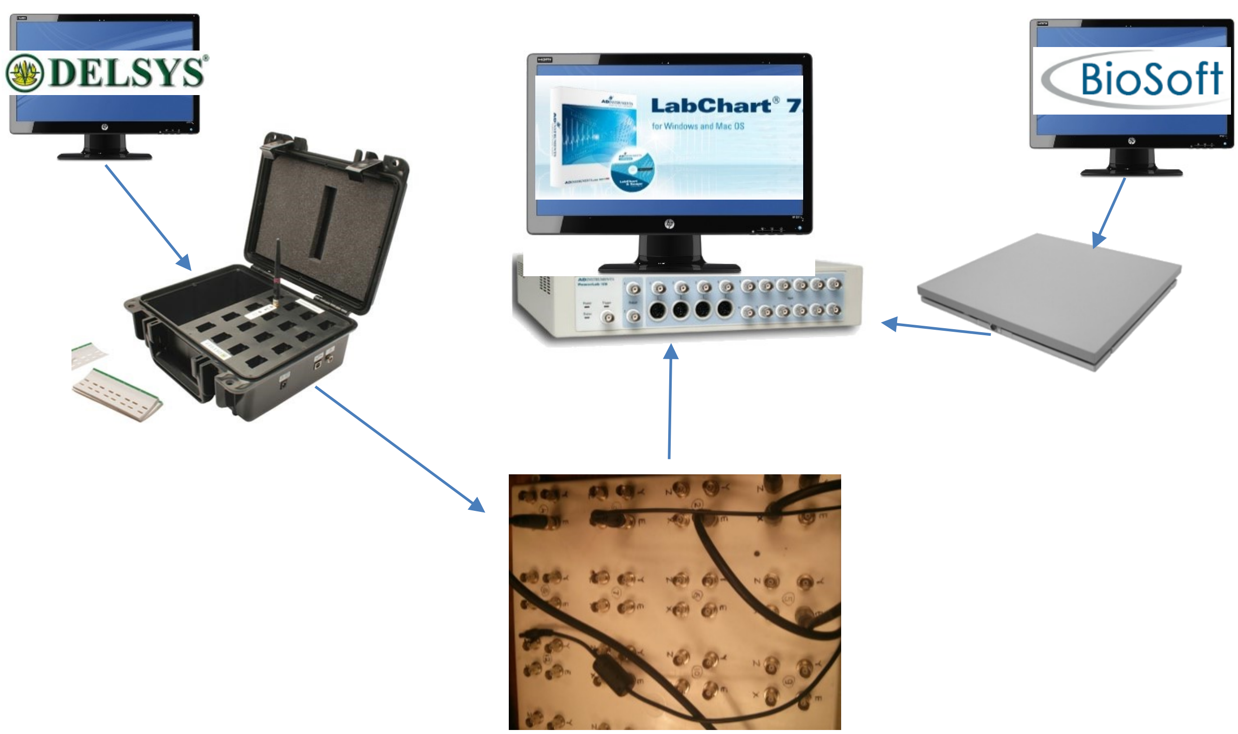

Figure 14. Synchronization of force plate and electromyography data. Reprinted from Trigno Wireless Systems and Smart Sensors, in Delsys, 2016, Retrieved April 4, 2015, from http://www.delsys.com/products/wireless-emg/.

Video analysis. Video data were collected during this research study in order to assess if proper technique was used consistently for all trials. For the purposes of this research, video data was analyzed through Kinovea software. Instances of interest through observation included: cane-tip strike and foot strike. Kinovea software analysis included the measurement of cane angle and distance of cane from the simulated injured foot. The angle of the cane was measured by positioning the vertex of the angle over the cane tip and placing the reference lines along the ground and the cane shaft. The distance of the cane from the toe was measured by using the distance tool in Kinovea.

Brower timing gates. Brower timing gates use infra-red signals and detectors to determine when the beam is broken. A researcher can use these gates to determine the length of time a movement takes by recording the time at which the movement begins and ends. Van Loo et al. (2003) found that the inter- and intra-rater reliabilities of using timing gates to measure walking speed were very high. The reliability of this measurement had an ICC of .998 for both comfortable and fast-paced tests (Van Loo, Moseley, Bosman, De Bie, & Hassett, 2003). Similarly, Waldron, Worsfold, Twist, and Lamb (2011) investigated the concurrent validity and test-retest reliability of timing gates and a global positioning system (GPS) when assessing sprint performance variables. Timing gate measures were found to be more reliable and valid when compared to the GPS measurements of distance and speed (Waldron et al., 2011).

Self-perceived ease of use questionnaire. The questionnaire included two questions to assess the overall usability issues related to the three types of canes. The first question required participants to rate the canes in the order of preference. The second question was open ended and allowed the participants to elaborate on reasons for liking or disliking any of the canes. Data acquired from this questionnaire was used to make inferences regarding the clinical significance of the study's findings.

Procedures

Recruitment. Healthy participants (n=21) were recruited through convenience sampling. Convenience sampling is a statistical method of drawing data by selecting people because of their availability or easy access (Farrokhi & Mahmoudi-Hamidabad, 2012). The healthy participants were sampled from the student population at Lakehead University. Convenience sampling was implemented through the use of posters, miniature presentations in various classes around campus, and through word-of-mouth. Posters were placed in multiple populated areas on the Lakehead University campus, such as the Chancellor Paterson Library, Outpost, and ATAC (Appendix B). The researchers did not hand out individual posters to potential participants. Participants were instructed to contact the researcher via email. If the participant was interested in partaking in the study then he/she was made aware of the inclusion and exclusion criteria. A preliminary meeting took place before the testing session where the participants were given a letter of recruitment and informed consent followed by a demographic questionnaire and a Physical Activity Readiness Questionnaire (Par-Q form), which ensured that the participants met the inclusion and exclusion criteria.

Data Collection. The preliminary meeting and data collection sessions were conducted at Lakehead University in the Sanders Building, room SB-1028, and took approximately 90 minutes of time for each participant. Each participant was tested individually. Participants took part in two testing sessions. The purpose of the initial session was to ensure that individuals met the inclusion and exclusion criteria and to familiarize themselves with the equipment and the testing procedure. This preliminary session was approximately 30 minutes in duration. The second session involved formal data collection and was approximately 60 minutes in duration.

Preliminary testing session. During the preliminary meeting preceding the testing session, participants were given a letter of recruitment (Appendix C) and a letter of informed consent (Appendix D) to fill out and return to the researchers. Once participants read and signed the informed consent, they were provided with a Par-Q (Appendix E) and a general demographic questionnaire (Appendix F). The Par-Q was used to determine if the participants were physically capable of participating in the study (CSEP, 2015). The general demographic questionnaire included data regarding the participants' height, weight, gender, program of study, and presence of any condition listed in the exclusion criteria. Height and weight of the participants was also measured in order to ensure accuracy. This demographic questionnaire and Par-Q were used as a method of screening the participants for exclusion criteria.

During this preliminary session, participants were also given a chance to familiarize themselves with the equipment. Participants were fitted with each type of cane (traditional cane, Miracle Cane®, and Stander Cane®) and the T-Scope knee brace and were allowed to practice cane-assisted ambulation (walking with the brace alone and with the brace and each type of cane). In order to fit the canes to each participant, the participant was positioned with the elbow flexed to 15 to 30 degrees while the cane was in contact with the ground (Teodoro et al., 2012). This measurement was taken when the participant was wearing his/her shoes. The degree of elbow flexion was measured with a 10-inch goniometer and was performed by the same researcher for all participants. To do this, the center fulcrum of the goniometer was placed over the lateral epicondyle of the humerus. The proximal arm was aligned with the lateral midline of the humerus, using the center of the acromion process for reference. Finally, the distal arm was aligned with the lateral midline of the radius, using the radial head and radial styloid process for reference (White, 2009). Participants were also fitted with a T-scope knee range of motion limiting brace, which was placed on the dominant leg (Appendix G). For the purpose of this study, bilateral symmetry was assumed and the dominant leg was considered as the simulated injured leg. The dominant leg was determined by having each participant run up to and kick a soccer ball with the assumption that he/she would strike the ball with his/her dominant leg (Velotta, Weyer, Ramirez, Winstead, & Bahamonde, 2011). For the purposes of this study, the T-scope knee brace was set at an angle of 30 degrees and only knee extension was limited in order to simulate a unilateral knee flexion contracture and an antalgic gait pattern (Harato et al., 2008).

Participants were also instructed on proper cane-assisted ambulation in which he/she held the cane contralateral to the dominant leg and advanced the cane simultaneously with the simulated injured limb (Chiou-Tan et al., 1999). The participants were allowed 10 practice trials for each type of cane. This practice session was recorded using a digital camcorder and the videos were analyzed using Kinovea software. This software enables the user to view, edit, and analyze videos. The researchers were specifically looking at consistency of the angle of the cane shaft in reference to the ground and the distance of the cane from the foot with the simulated injury as the individual stepped forward. The angle of the cane was measured by positioning the vertex of the angle over the cane tip and placing the reference lines along the ground and along the cane shaft. The distance of the cane from the toe was measured by using the distance tool in Kinovea.

Second testing session. Once participants successfully completed the first session, they were given the option to take part in a second testing session. When participants attended the second session, wireless EMG electrodes were attached to the surface of his/her skin to measure muscle activity and determine if muscle activation in the upper limb differed among cane types. The EMG signals from six main muscles of the arm, forearm, and shoulder were recorded. The anatomical localization of the muscles was accomplished by palpating the muscles as the participants isometrically contracted each respective muscle (Seniam, 2014). Electrodes were attached along the direction of the muscle fibers in the bulky central part of the muscle (Roman-Liu, & Tokarski, 2002). The electrodes were secured by way of adhesive pads overlying the following muscles: flexor carpi radialis, extensor carpi radialis longus, brachioradialis, triceps brachii, infraspinatus, and pectoralis major (Chiou-Tan et al., 1999; Delsys, 2015). These muscles were chosen because they all have a significant role with this movement (Chiou-Tan et al., 1999). Specific placement of the EMG electrodes followed procedures outlined in Cram's Introduction to Surface Electromyography (Criswell & Cram, 2011; Appendix H). Prior to the collection of any EMG data, participants were instructed to perform maximal contractions of each of the aforementioned muscles using isometric manual muscle testing techniques (Appendix I). This was done to enable the researchers to measure muscle activity as a percentage of maximum EMG, a method known as normalization. This normalization process allowed the data to be compared among participants (Halaki & Ginn, 2012). The EMG data was collected for each subsequent trial during this testing session, which provided a measure of muscle activation during cane-assisted ambulation for each different type of cane. The EMG data was filtered using a low pass filter with a cut-off frequency of 10 Hz implemented in LabChart7 to remove high frequency noise.

Before further data collection took place, the participants were given another chance to practice the proper cane walking technique and become comfortable with the use of these devices. This re-familiarization period took approximately 10 minutes. Subjects were allowed to practice walking approximately 10 steps with each assistive device prior to data collection to minimize hesitancy in maneuvering and to become familiar with the device. Participants were also given a chance to walk with the knee brace alone to familiarize themselves with the change in gait pattern. Once this was complete, the canes were randomly allocated in order to limit ordering effects. Randomization of the order for each cane used was performed through the Latin Square method. This method of randomization involves the use of an *n* x *n* matrix which contains *n* different conditions, each occurring exactly once in each row and once in each column. In this study, a 3 x 3 matrix was used, which contained 3 different cane conditions (traditional cane, Miracle Cane®, and Stander Cane®). Refer to Appendix J for a visual representation of the randomization process.

Once the participants were comfortable with cane use, they were instructed to walk over the force platforms. Two force platforms were used to measure the ground reaction forces and calculate impulse under the cane tips and the simulated injured limb. The first force platform collected simulated injured lower limb data, which included measures of maximum ground reaction forces (vertical, AP, and medial-lateral) and impulse (vertical and AP). The second force platform was used to collect upper limb data through the contact of the cane with force platform, which also included measures of maximum ground reaction forces (vertical, AP, and medial-lateral) and impulse (vertical and AP). Participants were instructed to perform the cane-assisted walking technique using the three different types of canes. For each cane type (traditional, Miracle Cane®, and Stander Cane®), the participants were asked to perform 5 trials (Segura & Piazza, 2007). A trial was considered valid if the participant hit the force plate with the complete base of the cane and no secondary impacts were present. Secondary impacts referred to any other impacts of the base of the cane or adjacent foot to the force plate. Furthermore, if the participant hesitated or broke natural stride in any manner, the trial was not considered for analysis and the participant was instructed to repeat the trial for a maximum of 10 trials per condition.

Each trial was performed at self-selected walking speeds; however, these speeds were within normal walking speeds. Normal walking speeds fall within a range of 1.25-1.50 m/s, which encompasses both older and younger individuals (Usroads, 2015). The speed of walking was monitored by setting up timing gates parallel to the force plates (Appendix K). Rest periods of three minutes were given between conditions to minimize any carry over effects that might have occurred from repeated walking and to avoid fatigue (De Salles et al., 2009).

Participants' trials were also recorded in this testing session and the data was analyzed using the same procedures described in the preliminary session. That is, the data was analyzed using Kinovea software to assess the consistency of the participants' ambulation techniques across trials. Following data collection, participants were given a questionnaire in order to assess subject-perceived ease of use for each type of cane (Appendix L).

Data Analysis

Reliability analysis. Impulse, EMG, walking speed, force plate data, and kinematic variables (angle and distance) were collected for each of the 5 trials for each cane condition. Intra-class correlation (ICC) values were calculated prior to performing any tests of significance for all dependent variables across all types of walking canes using SPSS for Windows. A one-way random effects model was used. Comparing values obtained across trials provided a measure of test-retest reliability.

All collected data was grouped based on cane condition (traditional, Miracle Cane®, and Stander Cane®). Force and impulse in the vertical (Fz), AP (Fx), and medial-lateral (Fy) directions were also grouped based on the extremity used (upper or lower). Electromyographic data was grouped based on muscles used (flexor carpi radialis, extensor carpi radialis longus, brachioradialis, biceps brachii, triceps brachii, pectoralis major, and infraspinatus). Intra-class correlations were also run for the cane angle and distance of the cane from the toe (acquired through kinematic analysis) to assess the consistency of the participants' ambulation techniques.

If strong correlations were found explaining over 50% of the variance, an average trial score was calculated for each dependent variable to answer the research questions.

Walking speed analysis. Prior to performing tests of significance, a one-way analysis of variance (ANOVA) was performed to determine whether any differences existed in walking speed between cane types. Since walking speed was expected to influence EMG, impulse, and force measures, any differences in walking speed would have affected the interpretation of the results. If walking speeds between cane types were found to be significantly different, ANCOVAs were used to address the effects of this covariant. If no differences were found, ANOVAs were used to answer the research questions.

The following statistical analyses were conducted to address each research question:

Question 1. What is the interaction effect between cane and muscle on upper extremity muscle activity?

Descriptive Statistics. Descriptive statistics were produced as part of the one-way ANCOVA procedure. Specifically, the "Descriptive Statistics" and "Estimates" tables were used to note any trend/apparent changes in means between cane types with regards to muscle activation; however, statistical significance could not be inferred from these tables. These tables were also used to ensure "cleanliness" of the data, such that there appeared to be no errors committed during data entry.

Inferential Statistics. A two-way ANOVA with repeated measures on one factor, cane type (traditional cane, Miracle Cane®, and Stander Cane®) was used to analyze the data. In this analysis, the two independent factors were type of cane (traditional cane, Miracle Cane®, and Stander Cane®) muscle (flexor carpi radialis, extensor carpi radialis longus, brachioradialis, triceps brachii, infraspinatus, and pectoralis major) with repeated measures on the first factor. The dependent variable was EMG as a percentage of maximum EMG. The two-way ANOVA procedure was analyzed for statistically significant interaction or main effects. If a statistically significant interaction was found, the simple main effects were used to help explain the interaction. The level of statistical significance was set at $\alpha < .05$ for all tests.

Question 2. Is there an interaction effect between type of cane and extremity (upper or lower) when measuring forces in the vertical, AP, and medial-lateral planes?

Descriptive Statistics. Descriptive statistics were calculated in order to provide the mean and standard deviation for each combination of the groups of the independent variables. These results were useful if there was no statistically significant interaction.

Inferential Statistics. A two-way ANOVA was run to determine the effect of different types of canes over upper and lower extremities on maximum force. This procedure was run for vertical, AP, and medial-lateral force maximums separately. In this analysis, the two independent factors were type of cane (traditional cane, Miracle Cane®, and Stander Cane®) and type of extremity (upper and lower) with repeated measures on the first factor. The dependent variable was maximum force in the vertical, AP, and medial-lateral directions (separate analyses were performed for each direction as the two-way ANOVA procedure is a univariate analysis). The two-way ANOVA procedure was analyzed for statistically significant interaction or main effects.

If a statistically significant interaction was found, the simple main effects were used to help explain the interaction.

Question 3. Is there an interaction effect between type of cane and extremity (upper or lower) when measuring impulse in the vertical and AP planes?

Descriptive Statistics. Descriptive statistics were calculated in order to provide the mean and standard deviation for each combination of the groups of the independent variables. These results were useful if there was no statistically significant interaction to help explain the main effect.

Inferential Statistics. A two-way ANOVA was run to determine the effect of different types of canes over upper and lower extremities on impulse. This procedure was run for both vertical and AP impulse maximums separately. In this analysis, the two independent factors were type of cane (traditional cane, Miracle Cane®, and Stander Cane®) and type of extremity (upper and lower) with repeated measures on the first factor. The dependent variables were maximum impulse in the vertical, anterior, and posterior directions. Separate analyses were run for each direction. The two-way ANOVA procedure was analyzed for statistically significant interaction or main effects. If a statistically significant interaction was found, the simple main effects were used to help explain the interaction.

Question 4. Is there a difference between cane types in terms of subject-perceived ease of use?

Data regarding subject-perceived ease of use was acquired from the ease of use questionnaire, which allowed the participants to rate the canes in order of preference and provide a subjective statement indicating the reasoning behind these choices. Numerical data acquired from this questionnaire was analyzed through descriptive statistics, including means and standard deviations. Higher means indicated higher levels of preference. The subjective data acquired from this questionnaire was used to provide further evidence of the validity of these ratings.

Chapter 3 – Results

The result of this study provide evidence of reliability measures across replications of the protocol for each dependent variable. Descriptive and inferential statistical analysis techniques were used to help explain the interaction and main effects of cane type, muscle type, upper and lower extremity on measures of EMG, force, and impulse.

Reliability Results

Table 1 shows the results from the ICC procedure for all dependent variables, grouped accordingly. Intra-class correlation values ranged from .55 to .99, indicating strong correlations between trials for all measured variables.

Table 1

Intra-class correlations for all dependent variables

Dependent Variable	Traditional Cane	Miracle Cane®	Stander Cane®
	Force		
	Force Plate 1 (Force Data for Simulated Injured Limb Impact)		
Fz Force (Vertical)	.902	.946	.917
Fx Force (Anterior)	.820	.924	.803
Fx Force (Posterior)	.870	.861	.873
Fy Force (Lateral)	.866	.863	.939
Fy Force (Medial)	.821	.785	.769
	Force Plate 2 (Force Data for Cane Impact)		
Fz Force (Vertical)	.873	.862	.875
Fx Force (Anterior)	.742	.785	.758
Fx Force (Posterior)	.779	.733	.887
Fy Force (Medial)	.689	.678	.558
Fy Force (Lateral)	.698	.637	.616
	Impulse		
	Force Plate 1 Impulse (Impulse Data for Simulated Injured Limb)		
Fz Impulse (Vertical)	.938	.731	.918
Fx Impulse (Anterior)	.809	.741	.708
Fx Impulse (Posterior)	.931	.946	.915
	Force Plate 2 Impulse (Impulse Data for Cane)		
Fz Impulse (Vertical)	.928	.845	.891
Fx Impulse (Anterior)	.831	.768	.836
Fx Impulse (Posterior)	.743	.709	.834
	Electromyography		
Flexor	.867	.839	.925
Extensor	.938	.952	.982
Brachioradialis	.908	.744	.742
Biceps	.912	.851	.871
Triceps	.988	.992	.999
Pectoralis	.914	.861	.900
Infraspinatus	.997	.935	.807
	Kinematics		
Angle	.765	.715	.810
Distance	.8	.768	.710
	Time		
Timing-Gates	.868	.806	.879

Walking Speed Analysis

A one-way ANOVA was conducted to determine if walking speeds were different in individuals who used different cane types. Participants were classified into three groups: traditional cane (n = 21), Miracle Cane® (n = 21), and Stander Cane® (n = 21). The differences between cane types were not statistically significant, $F(2,60) = .02$, $p > .05$.

Question 1

A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and muscle (flexor carpi radialis, extensor carpi radialis longus, brachioradialis, triceps brachii, infraspinatus, and pectoralis major) on upper extremity muscle activity.

Assumptions.

Outliers. There were 18 outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. These values remained in the analysis as they were determined to be genuinely unusual values. That is, the researchers did not believe that these values were due to data entry or measurement errors; rather, they were attributed to natural differences in muscle activation between participants. Furthermore, the analysis was run with and without the inclusion of these outliers and yielded similar conclusions, deeming the effect of the outliers on the results to be minimal.

Normality. Electromyographic measures were normally distributed, as assessed by the Shapiro-Wilk's test ($p > .05$), with the exception of the following muscles: triceps and infraspinatus in the traditional cane condition; pectoralis and infraspinatus in the Miracle Cane® condition; extensor carpi radialis longus, biceps, triceps, and infraspinatus in the Stander Cane® condition.

Sphericity. Mauchly's Test of Sphericity indicated that the assumption of sphericity was met for the two-way interaction, $\chi^2(2) = 5.683$, $p = .058$.

Descriptive statistics. Descriptive statistics, shown in Table 2, indicate that EMG activation decreased in spring loaded cane conditions when compared to the traditional cane condition (28.84 ± 16.15 %), but more so in the Miracle Cane® condition (24.49 ± 13.35 %) than the Stander Cane® condition (27.54 ± 15.41 %). This relationship was true for all muscles.

Table 2

*Descriptive statistics for cane type * muscle*

	Muscle	Mean	Std. Deviation	N
TraditionalCane	Flexor	46.73	6.84	21
	Extensor	27.59	6.80	21
	Brachioradialis	23.03	6.20	21
	Biceps	13.63	2.95	21
	Triceps	50.97	13.47	21
	Pectoralis	24.26	2.72	21
	Infraspinatus	15.72	15.57	21
	Total	28.85	16.15	147
MiracleCane®	Flexor	40.11	7.10	21
	Extensor	21.34	3.27	21
	Brachioradialis	20.09	4.41	21
	Biceps	11.34	2.07	21
	Triceps	43.97	9.66	21
	Pectoralis	22.13	4.76	21
	Infraspinatus	12.50	8.75	21
	Total	24.50	13.35	147
StanderCane®	Flexor	45.54	7.09	21
	Extensor	25.94	6.48	21
	Brachioradialis	20.49	3.85	21
	Biceps	12.96	3.00	21
	Triceps	49.82	12.21	21
	Pectoralis	23.72	4.58	21
	Infraspinatus	14.32	10.54	21
	Total	27.54	15.42	147

Inferential statistics. As indicated in Table 3, there was no statistically significant interaction between the type of cane and muscle on EMG activation, $F(12, 280) = 0.87$, $p > .05$, partial $\eta^2 = .036$.

Table 3

*Interaction effect of cane type * muscle*

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
CaneType	Sphericity Assumed	1464.96	2.00	732.48	18.45	0.00	0.12
	Greenhouse-Geisser	1464.96	1.92	761.82	18.45	0.00	0.12
	Huynh-Feldt	1464.96	2.00	732.48	18.45	0.00	0.12
	Lower-bound	1464.96	1.00	1464.96	18.45	0.00	0.12
CaneType * MuscleType	Sphericity Assumed	416.13	12.00	34.68	0.87	0.58	0.04
	Greenhouse-Geisser	416.13	11.54	36.07	0.87	0.57	0.04
	Huynh-Feldt	416.13	12.00	34.68	0.87	0.58	0.04
	Lower-bound	416.13	6.00	69.36	0.87	0.52	0.04
Error(CaneType)	Sphericity Assumed	11119.03	280.00	39.71			
	Greenhouse-Geisser	11119.03	269.22	41.30			
	Huynh-Feldt	11119.03	280.00	39.71			
	Lower-bound	11119.03	140.00	79.42			

The main effect of muscle, determined through interpretation of Table 4, showed that there was a statistically significant difference in EMG activation between muscles, $F(6, 140) = 124.28$, $p < .05$, partial $\eta^2 = 0.84$. Pairwise comparisons among all muscles indicated significant differences between all muscles, with the exception of the flexor and triceps muscles and the extensor and brachioradialis muscles.

Table 4

Main effect of muscle on EMG activation

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	320560.421	1	320560.421	3252.238	.000	.959
Muscle	73497.870	6	12249.645	124.278	.000	.842
Error	13799.256	140	98.566			

The main effect of cane type, determined through interpretation of Table 3, showed that there was a statistically significant difference in EMG activation among cane types, $F(2, 280) =$

732.48, $p < .05$, partial $\eta^2 = 0.11$. Pairwise comparisons indicated significant differences in EMG activation between the traditional cane and the Miracle Cane®; the Miracle Cane® and Stander Cane®. No significant differences were found between the traditional cane and Stander Cane®. These differences indicate that EMG activation is significantly decreased when using the Miracle Cane® as compared to the traditional cane and Stander Cane®. A visualization of these results can be found in Figure 15.

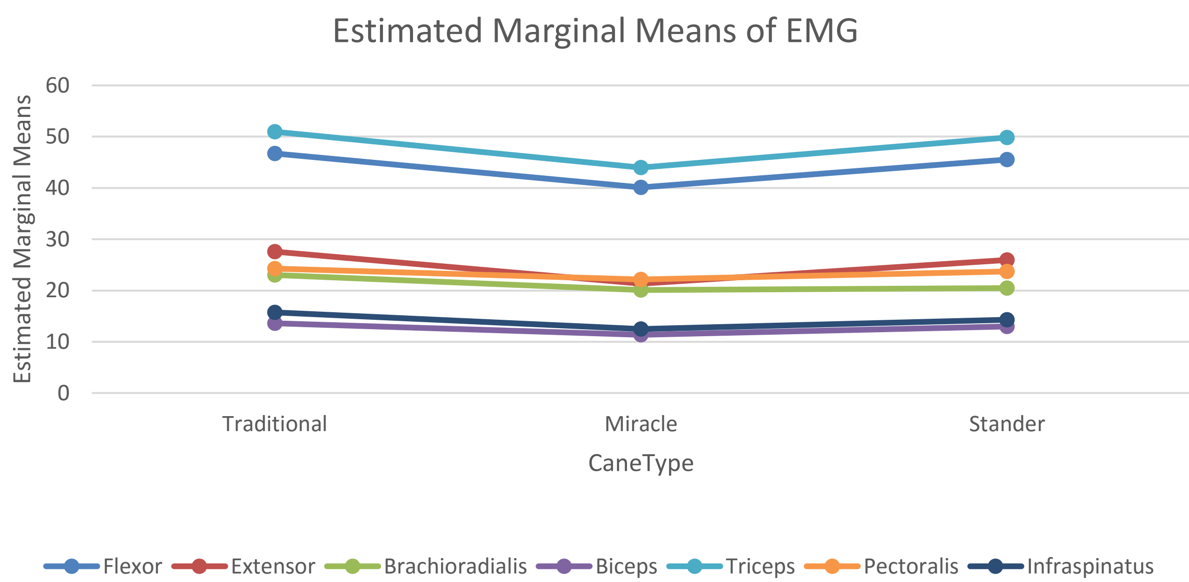

Figure 15. Estimated Marginal Means of EMG.

Question 2

Vertical force. A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and limb type (upper and lower) on vertical GRF.

Assumptions.

Outliers. There were two outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. These values remained in the analysis as they were determined to be genuinely unusual values. That is, the researchers did not believe that these values were due to data entry or measurement errors; rather, they were attributed to natural differences in force production between participants. These differences may be attributable to differences in cane-ambulation techniques. Furthermore, the analysis was run with and without the inclusion of these outliers and yielded similar conclusions, deeming the effect of the outliers on the results to be minimal. Although they were not considered outliers in this specific analysis, lower limb vertical force data for participant 9 was removed from the analysis as the data was thought to include measurement error, which could influence the internal validity of the vertical GRFs. That is, the force plate may not have been normalized for body weight when collecting data for this participant, which could create extreme outliers in subsequent analyses. To ensure that the removal of this data did not influence the results greatly, the ANOVA procedure was run with and without the inclusion of the data and yielded similar results.

Normality. Vertical force was normally distributed, as assessed by a Shapiro-Wilk's Test ($p > .05$).

Sphericity. Mauchly's test of sphericity indicated that the assumption of sphericity was met for the two-way interaction, $\chi^2(2) = 0.027$, $p = .986$.

Descriptive statistics. Descriptive statistics, shown in Table 5, indicate that mean vertical GRF in the upper limb decreased in spring loaded cane conditions when compared to the traditional cane condition (184.84 ± 46.04 Newtons), but more so in the Miracle Cane® condition (138.21 ± 40.08 Newtons) than the Stander Cane® condition (172.4 ± 51.85 Newtons). Mean vertical GRF in the lower limb increased in the spring loaded cane conditions when compared to the traditional cane condition (650.7 ± 116.72 Newtons), but more so in the Miracle

Cane® condition (681.35 ± 109.4 Newtons) than the Stander Cane® condition (664.92 ± 116.23 Newtons).

Table 5

*Descriptive statistics for cane type * limb type on vertical force*

	Type of Limb (Upper or Lower)	Mean	Std. Deviation	N
Vertical force measured when ambulating with the traditional cane	Lower Limb	650.70	116.72	20
	Upper Limb	184.85	46.05	21
	Total	412.09	251.22	41
Vertical force measured when ambulating with the Miracle Cane®	Lower Limb	681.35	109.40	20
	Upper Limb	138.21	40.09	21
	Total	403.16	286.42	41
Vertical force measured when ambulating with the Stander Cane®	Lower Limb	664.92	116.23	20
	Upper Limb	172.41	51.86	21
	Total	412.66	264.36	41

Inferential statistics. As indicated in Table 6, There was a statistically significant interaction between the type of cane and type of limb on vertical GRF, $F(2,78) = 35.16$, $p < .05$, partial $\eta^2 = .47$. Given the descriptive statistics in Table 5 and the illustration of the interaction in Figure 16, it appears that the interaction lies in the change in vertical force produced during the Miracle Cane® condition on the upper and lower limbs. This data indicates that as vertical force is increased on the lower limb during the Miracle Cane® condition, vertical force is decreased on the upper limb.

Table 6

*Interaction effect for cane type * limb type on vertical force*

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
CaneType	Sphericity Assumed	1961.84	2.00	980.92	2.19	0.12	0.05
	Greenhouse-Geisser	1961.84	2.00	981.62	2.19	0.12	0.05
	Huynh-Feldt	1961.84	2.00	980.92	2.19	0.12	0.05
	Lower-bound	1961.84	1.00	1961.84	2.19	0.15	0.05
CaneType * LimbType	Sphericity Assumed	31573.14	2.00	15786.57	35.16	0.00	0.47
	Greenhouse-Geisser	31573.14	2.00	15797.87	35.16	0.00	0.47
	Huynh-Feldt	31573.14	2.00	15786.57	35.16	0.00	0.47
	Lower-bound	31573.14	1.00	31573.14	35.16	0.00	0.47
Error(Cane Type)	Sphericity Assumed	35020.01	78.00	448.97			
	Greenhouse-Geisser	35020.01	77.94	449.30			
	Huynh-Feldt	35020.01	78.00	448.97			
	Lower-bound	35020.01	39.00	897.95			

Simple main effects can be used to further describe the interaction effect. To test for the simple main effect of limb type, three separate univariate ANOVAs were performed. Each of these statistical tests was used to analyze the differences in vertical GRF between limbs (upper and lower) at each category of the within-subjects factor, cane type. The first of these ANOVAs tested for the difference between upper limb and lower limb vertical GRF while ambulating with the traditional cane. There was a statistically significant difference in vertical GRFs between the upper limb and lower limb while ambulating with the traditional cane $F(1, 39) = 287.79$, $p < .05$, partial $\eta^2 = .88$. The second ANOVA tested for the difference between upper and lower limb vertical GRF while ambulating with the Miracle Cane®. There was a statistically significant difference in vertical GRFs between the upper limb and lower limb while ambulating with the Miracle Cane®, $F(1, 39) = 454.12$, $p < .05$, partial $\eta^2 = .92$. The third ANOVA tested for the difference between upper and lower limb vertical GRF while ambulating with the Stander

Cane®. There was a statistically significant difference in vertical GRFs between the upper limb and lower limb while ambulating with the Stander Cane®, $F(1, 39) = 312.15$, $p < .05$, partial $\eta^2 = .89$.

To test for the simple main effect of cane type, two separate ANOVAs were performed. Each statistical test examined the differences in vertical GRF between cane types for each category of the between-subject factor, limb type. The first ANOVA tested for the difference between vertical forces created by each cane type on the lower limb. There was a statistically significant effect of cane type on vertical GRF for the lower limb, $F(2, 38) = 16.03$, $p < .05$. Further examination of pairwise comparisons indicated that there were significant differences between vertical forces produced during ambulation with all cane types in the lower limb. That is, all cane types produced statistically significantly different vertical GRFs on the lower limb. In combination with descriptive statistics, the results show that the Miracle Cane® produced significantly greater vertical force on the lower limb than the traditional cane and Stander Cane®. The Stander Cane® produced significantly greater vertical force on the lower limb than the traditional cane and significantly lower force on the lower limb than the Miracle Cane®. The second ANOVA tested for the difference between vertical forces created by each cane type on the upper limb. There was a statistically significant effect of cane type on vertical GRF for the upper limb, $F(2, 40) = 20.52$, $p < .05$. Further examination of pairwise comparisons revealed significant differences in vertical force production between the Miracle Cane® and the traditional cane and the Miracle Cane® and Stander Cane® on the upper limb. There was no significant difference in vertical force production between the traditional and Stander Cane® on the upper limb. In combination with descriptive statistics, the results indicate that the Miracle

Cane® produced significantly lower vertical GRF on the upper limb when compared to the traditional and Stander Cane®. A visualization of these results can be found in Figure 16.

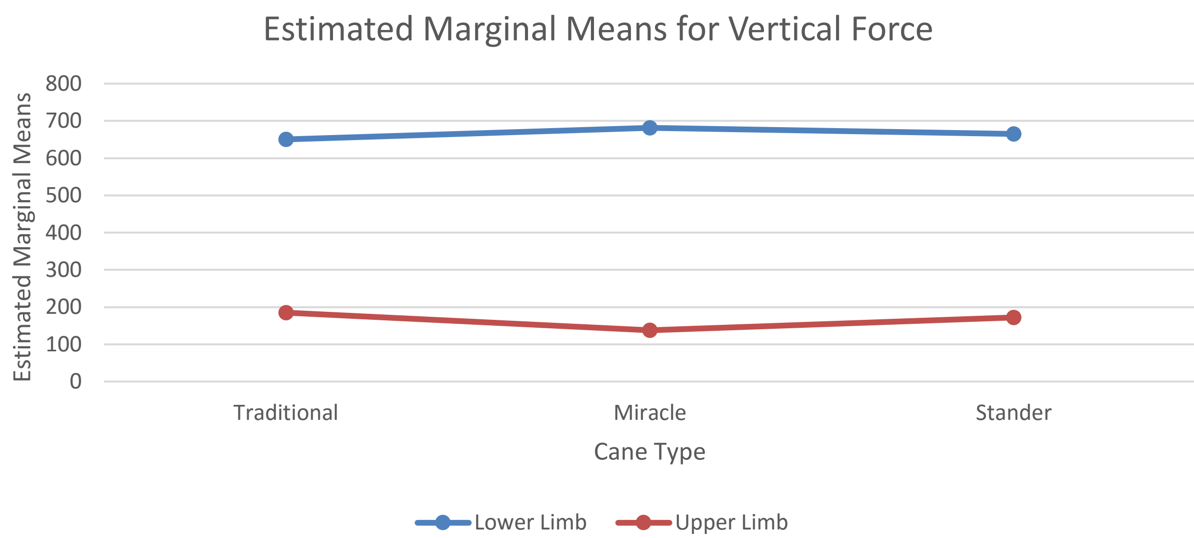

Figure 16. Estimated Marginal Means for Vertical Force.

Anterior force. A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and limb type (upper and lower) on anterior GRF.

Assumptions.

Outliers. There were 5 outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. One of these values was removed as per the previous analysis. Participant 9 lower limb anterior force data was once again removed due to measurement error and validity concerns. The remainder of the outliers remained in the analysis as they were determined to be genuinely unusual values. That is, the researchers did not believe that these values were due to data entry or measurement errors; rather, they were attributed to natural differences in force production between participants. These differences may be attributable to differences in cane-ambulation techniques. Furthermore, the analysis was run with and without the inclusion of these outliers and yielded similar conclusions, deeming the effect of the outliers on the results to be minimal.

Normality. Anterior force was normally distributed with the exception of anterior force produced by the lower limb while ambulating with the traditional cane and Miracle Cane®. Normality was assessed by a Shapiro-Wilk's Test ($p > .05$).

Sphericity. Mauchly's Test of Sphericity indicated that the assumption of sphericity was not met for the two-way interaction, $\chi^2(2) = 9.381$, $p < .05$. The Greenhouse-Geisser adjustment was used instead to ensure that the results of the analysis were not biased, minimizing the possibility of committing a type I measurement error.

Descriptive statistics. Descriptive statistics, shown in Table 6, indicate that mean anterior GRF in the upper limb increased in spring loaded cane conditions when compared to the traditional cane condition (17.02 ± 6.99 Newtons), but more so in the Miracle Cane® condition (24.33 ± 11.79 Newtons) than the Stander Cane® condition (18.09 ± 8.77 Newtons). Mean anterior GRF in the lower limb increased in the spring loaded cane conditions when compared to the traditional cane condition (69.25 ± 25.35 Newtons), but more so in the Miracle Cane® condition (78.64 ± 36.43 Newtons) than the Stander Cane® condition (72.74 ± 27.24 Newtons).

Table 6

*Descriptive statistics for cane type * limb type on anterior force*

	Type of Limb (Upper or Lower)	Mean	Std. Deviation	N
Anterior force measured when ambulating with the traditional cane	Lower Limb	69.25	25.35	20
	Upper Limb	17.02	6.99	21
	Total	42.49	32.07	41
Anterior force measured when ambulating with the Miracle Cane®	Lower Limb	78.64	36.43	20
	Upper Limb	24.33	11.79	21
	Total	50.82	38.15	41
Anterior force measured when ambulating with the Stander Cane®	Lower Limb	72.74	27.24	20
	Upper Limb	18.09	8.77	21
	Total	44.75	34.00	41

Inferential statistics. As indicated in Table 7, there was no statistically significant interaction between the type of cane and type of limb on anterior GRF, $F(1.645,64.164) = 0.17$, $p > .05$, partial $\eta^2 = .005$. Greenhouse-Geisser values were used to adjust any bias created by violations of the assumption of sphericity.

Table 7

*Interaction effect for cane type * limb type on anterior force*

Source		Type III SOS	df	Mean Square	F	Sig.	partial η^2
CaneType	Sphericity Assumed	1527.49	2.00	763.75	7.75	0.00	.166
	Greenhouse-Geisser	1527.49	1.65	928.44	7.75	0.00	.166
	Huynh-Feldt	1527.49	1.75	871.74	7.75	0.00	.166
	Lower-bound	1527.49	1.00	1527.49	7.75	0.01	.166
CaneType * LimbType	Sphericity Assumed	35.05	2.00	17.53	0.18	0.84	.005
	Greenhouse-Geisser	35.05	1.65	21.31	0.18	0.80	.005
	Huynh-Feldt	35.05	1.75	20.00	0.18	0.81	.005
	Lower-bound	35.05	1.00	35.05	0.18	0.68	.005
Error(Cane Type)	Sphericity Assumed	7691.52	78.00	98.61			
	Greenhouse-Geisser	7691.52	64.16	119.87			
	Huynh-Feldt	7691.52	68.34	112.55			
	Lower-bound	7691.52	39.00	197.22			

The main effect of limb type, determined through interpretation of Table 8, showed that there was a statistically significant difference in anterior force production between the upper and lower limb, $F(1,39) = 70.42$, $p < .05$, partial $\eta^2 = 0.64$.

Table 8

Main effect of limb type on anterior force

Source	**Type III Sum of Squares**	**df**	**Mean Square**	**F**	**Sig.**	**partial η^2**
Intercept	267823.17	1	267823.17	212.58	.00	.84
LimbType	88717.75	1	88717.74	70.41	.00	.64
Error	49134.46	39	1259.85			

The main effect of cane type, determined through interpretation of Table 7, showed that there was a statistically significant difference in anterior force production between cane types, $F(1.645, 64.164) = 7.74$, $p < .05$, partial $\eta^2 = 0.16$. Once again, Greenhouse-Geisser values were used to adjust any bias created by violations of the assumption of sphericity. Pairwise comparisons between cane types indicated significant differences in anterior force production between the traditional cane and the Miracle Cane®. No significant differences were found between the traditional cane and Stander Cane® or the Miracle Cane® and Stander Cane®. In combination with descriptive statistics, the results indicate that the Miracle Cane® produced greater anterior force than the traditional and Stander Cane®. A visualization of these result can be found in Figure 17.

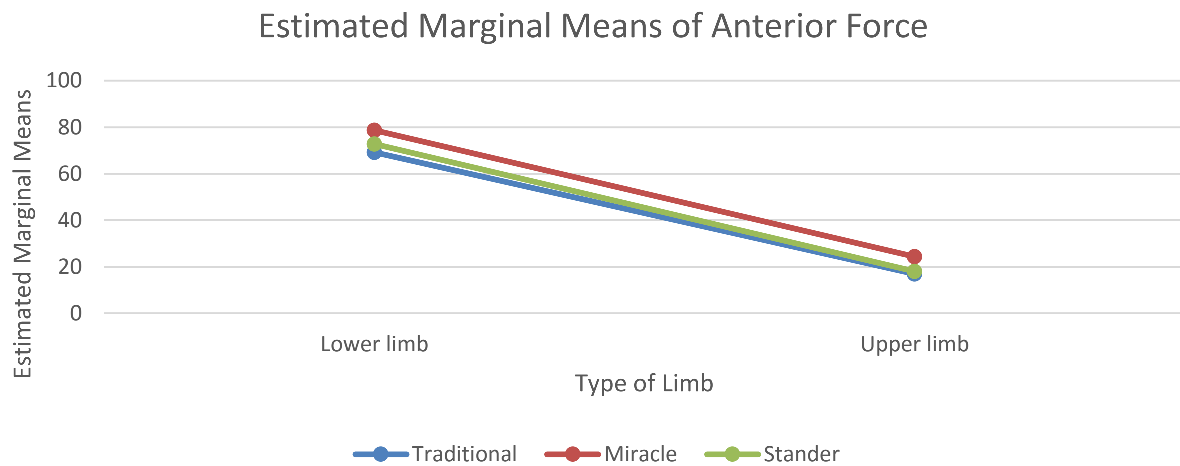

Figure 17. Estimated Marginal Means for Anterior Force.

Posterior force. A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and limb type (upper and lower) on posterior GRF.

Assumptions.

Outliers. There were four outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. These values remained in the analysis as they were determined to be genuinely unusual values. That is, the researchers did not believe that these values were due to data entry or measurement errors; rather, they were attributed to natural differences in force production between participants. These differences may be attributable to differences in cane-ambulation techniques. Furthermore, the analysis was run with and without the inclusion of these outliers and yielded similar conclusions, deeming the effect of the outliers on the results to be minimal. Although they were not considered outliers in this specific analysis, lower limb posterior force data for participant 9 was removed from the analysis as the data was thought to include measurement error, which could affect the internal validity of the data on posterior force measures. That is, the force plate may not have been normalized for body weight, which could create extreme outliers in subsequent analyses. To ensure that the removal of this data did not influence the results greatly, the ANOVA procedure was run with and without the inclusion of the data and yielded similar results. Once this data was removed, only two outliers remained.

Normality. Posterior force was normally distributed with the exception of posterior force produced by the upper limb while ambulating with the traditional cane and Stander Cane®. Normality was assessed by a Shapiro-Wilk's Test ($p > .05$).

Sphericity. Mauchly's Test of Sphericity indicated that the assumption of sphericity was met for the two-way interaction, $\chi^2(2) = 4.810$, $p > .05$.

Descriptive statistics. Descriptive statistics, shown in Table 9, indicate that mean posterior GRF in the upper limb increased in the Stander Cane® condition (-29.42 ± 17.14 Newtons) and decreased in the Miracle Cane® condition (-28.67 ± 9.63 Newtons) when compared to the traditional cane condition (-29.1 ± 13.23 Newtons); however, it is important to note that these differences are minimal. Mean posterior GRF in the lower limb increased in the spring loaded cane conditions when compared to the traditional cane condition (-99.1037 ± 34.14 Newtons), but more so in the Miracle Cane® condition (-105.79 ± 37.42 Newtons) than the Stander Cane® condition (-101.16 ± 36.1 Newtons).

Table 9

*Descriptive statistics for cane type * limb type on posterior force*

	Type of Limb (Upper or Lower)	Mean	Std. Deviation	N
Posterior force measured when ambulating with the traditional cane	Lower Limb	-99.10	34.14	20
	Upper Limb	-29.10	13.23	21
	Total	-63.25	43.54	41
Posterior force measured when ambulating with the Miracle Cane®	Lower Limb	-105.80	37.42	20
	Upper Limb	-28.67	9.63	21
	Total	-66.29	47.28	41
Posterior force measured when ambulating with the Stander Cane®	Lower Limb	-101.17	36.10	20
	Upper Limb	-29.42	17.14	21
	Total	-64.42	45.65	41

Inferential statistics. As indicated in Table 10, there was no statistically significant interaction between the type of cane and type of limb on posterior GRF, $F(2,78) = 2.13$, $p > .05$, partial $\eta^2 = .05$.

Table 10

*Interaction effect for cane type * limb type on posterior force*

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	partial η^2
CaneType	Sphericity Assumed	204.79	2.00	102.40	1.55	0.22	0.04
	Greenhouse-Geisser	204.79	1.79	114.57	1.55	0.22	0.04
	Huynh-Feldt	204.79	1.92	106.90	1.55	0.22	0.04
	Lower-bound	204.79	1.00	204.79	1.55	0.22	0.04
CaneType * LimbType	Sphericity Assumed	282.45	2.00	141.23	2.13	0.13	0.05
	Greenhouse-Geisser	282.45	1.79	158.02	2.13	0.13	0.05
	Huynh-Feldt	282.45	1.92	147.44	2.13	0.13	0.05
	Lower-bound	282.45	1.00	282.45	2.13	0.15	0.05
Error(Cane Type)	Sphericity Assumed	5167.93	78.00	66.26			
	Greenhouse-Geisser	5167.927	69.712	74.133			
	Huynh-Feldt	5167.927	74.711	69.173			
	Lower-bound	5167.927	39.000	132.511			

The main effect of limb type, determined through interpretation of Table 11, showed that there was a statistically significant difference in posterior force production between the upper and lower limb, $F(1,39) = 80.17$, $p < .05$, partial $\eta^2 = 0.67$.

Table 11

Main effect of limb type on posterior force

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	partial η^2
Intercept	528065.19	1	528065.19	258.8	.00	.86
LimbType	163589.58	1	163589.58	80.17	.00	.67
Error	79574.53	39	2040.37			

The main effect of cane type, determined through interpretation of Table 10, showed that there was no statistically significant difference in posterior force production between cane types, $F(2, 78) = 1.54$, $p > .05$, partial $\eta^2 = 0.03$. A visualization of these result can be found in Figure 18.

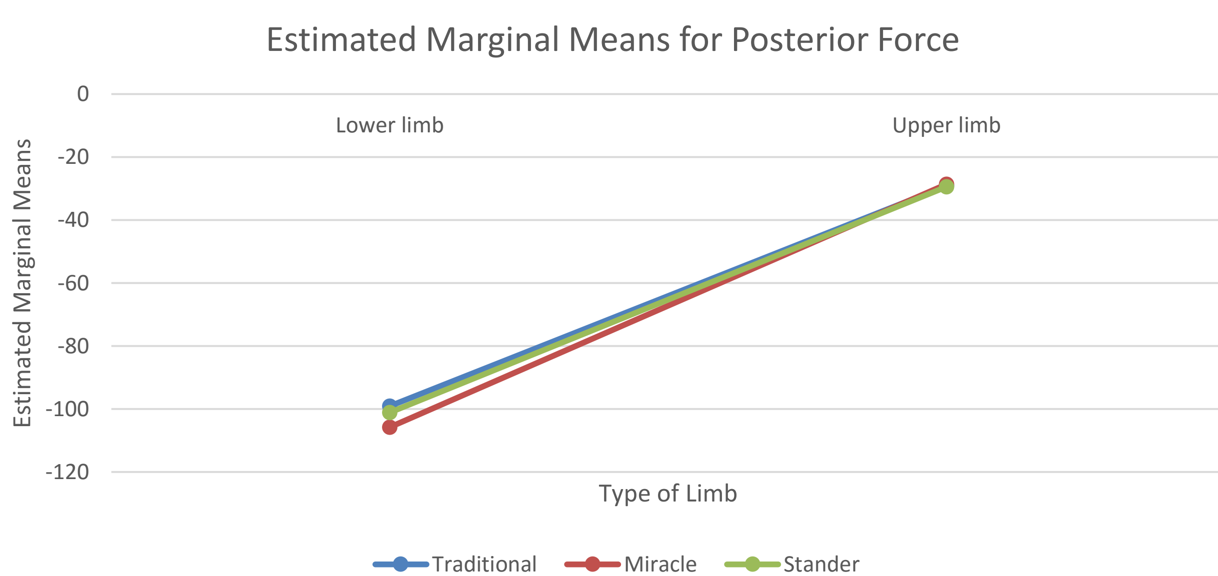

Figure 18. Estimated Marginal Means for Posterior Force.

Medial force. A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and limb type (upper and lower) on medial GRF.

Assumptions.

Outliers. There were 12 outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. Two of these values were removed as per the previous analysis. Participant 9 lower limb medial force data was once again removed due to measurement error and validity concerns. The remainder of the outliers remained in the analysis as they were determined to be genuinely unusual values. That is, the researchers did not believe that these values were due to data entry or measurement errors; rather, they were attributed to natural differences in medial force production between participants. These differences may be attributable to differences in cane-ambulation techniques, foot placement, or cane placement. Furthermore, the analysis was run with and without the inclusion of these outliers and yielded similar conclusions, deeming the effect of the outliers on the results to be minimal.

Normality. Medial force was normally distributed with the exception of medial force produced by the upper limb while ambulating with the traditional cane and Miracle Cane®. The assumption of normality was also not met in the lower limb/traditional cane condition. Normality was assessed by a Shapiro-Wilk's Test ($p > .05$).

Sphericity. Mauchly's test of sphericity indicated that the assumption of sphericity was met for the two-way interaction, $\chi^2(2) = 2.903$, $p > .05$.

Descriptive statistics. Descriptive statistics, shown in Table 12, indicate that mean medial GRF in the upper limb increased in spring loaded cane conditions when compared to the traditional cane condition (7.39 ± 5.3 Newtons), but more so in the Miracle Cane® condition (16.88 ± 11.75 Newtons) than the Stander Cane® condition (8.7068 ± 5.29 Newtons). Mean medial GRF in the lower limb decreased in the spring loaded cane conditions when compared to the traditional cane condition (36.08 ± 20.14 Newtons), but more so in the Stander Cane® condition (32.87 ± 16.69 Newtons) than the Miracle Cane® condition (35.46 ± 18.58 Newtons).

Table 12

*Descriptive statistics for cane type * limb type on medial force*

	Type of Limb (Upper or Lower)	Mean	Std. Deviation	N
Medial force measured when ambulating with the traditional cane	Lower Limb	36.08	20.14	20
	Upper Limb	7.39	5.30	21
	Total	21.39	20.43	41
Medial force measured when ambulating with the Miracle Cane®	Lower Limb	35.46	18.58	20
	Upper Limb	16.88	11.75	21
	Total	25.94	17.93	41
Medial force measured when ambulating with the Stander Cane®	Lower Limb	32.87	16.69	20
	Upper Limb	8.71	5.29	21
	Total	20.49	17.20	41

Inferential statistics. As indicated in Table 13, there was a statistically significant interaction between the type of cane and type of limb on medial GRF, $F(2,78) = 4.07$, $p < .05$, partial $\eta^2 = .095$. Given the descriptive statistics in Table 12 and illustration of the interaction in Figure 19, the interaction appears to lie in the difference between medial force production during cane conditions on the upper limb. That is, medial force production on the upper limb appears to be greater in the Miracle Cane® condition when compared to the traditional and Stander Cane®.

Table 13

*Interaction effect for cane type * limb type on medial force*

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	partial η^2
CaneType	Sphericity Assumed	676.436	2	338.218	5.248	.007	.119
	Greenhouse-Geisser	676.436	1.863	363.097	5.248	.009	.119
	Huynh-Feldt	676.436	2.000	338.218	5.248	.007	.119
	Lower-bound	676.436	1.000	676.436	5.248	.027	.119
CaneType * LimbType	Sphericity Assumed	525.205	2	262.603	4.075	.021	.095
	Greenhouse-Geisser	525.205	1.863	281.919	4.075	.023	.095
	Huynh-Feldt	525.205	2.000	262.603	4.075	.021	.095
	Lower-bound	525.205	1.000	525.205	4.075	.050	.095
Error(Cane Type)	Sphericity Assumed	5026.975	78	64.448			
	Greenhouse-Geisser	5026.975	72.656	69.189			
	Huynh-Feldt	5026.975	78.000	64.448			
	Lower-bound	5026.975	39.000	128.897			

Simple main effects can be used to further explain the interaction effect. To test for the simple main effect of limb type, three separate univariate ANOVAs were performed. Each of these statistical tests was used to analyze the differences in medial GRF between limbs (upper and lower) at each category of the within-subjects factor, cane type. The first of these ANOVAs tested for the difference between upper limb and lower limb medial GRF while ambulating with the traditional cane. There was a statistically significant difference in medial GRFs between the upper limb and lower limb while ambulating with the traditional cane $F(1, 39) = 39.76$, $p < .05$, partial $\eta^2 = .5$. The second ANOVA tested for the difference between upper and lower limb medial GRF while ambulating with the Miracle Cane®. There was a statistically significant difference in medial GRFs between the upper limb and lower limb while ambulating with the Miracle Cane®, $F(1, 39) = 14.79$, $p < .05$, partial $\eta^2 = .27$. The third ANOVA tested for the difference between upper and lower limb medial GRF while ambulating with the Stander Cane®.

There was a statistically significant difference in medial GRFs between the upper limb and lower limb while ambulating with the Stander Cane®, $F(1, 39) = 39.87$, $p < .05$, partial $\eta^2 = .5$.

To test for the simple main effect of cane type, two separate ANOVAs were performed. Each statistical test examined the differences in medial GRF between cane types for each category of the between-subject factor (limb type). The first ANOVA tested for the difference between medial forces created by each cane type on the lower limb. There was no statistically significant effect of cane type on medial GRF for the lower limb, $F(2,38) = 0.6$, $p > .05$. The second ANOVA tested for the difference between medial forces created by each cane type on the upper limb. There was a statistically significant effect of cane type on medial GRF for the upper limb, $F(1.507, 30.134) = 16.06$, $p < .05$. Further examination of pairwise comparisons revealed that there were significant differences in medial force production between the Miracle Cane® and the traditional and Stander Cane® on the upper limb. There was no significant difference in medial force production between the traditional and Stander Cane® on the upper limb. In combination with descriptive statistics, the results show that the Miracle Cane® produced greater medial forces on the upper limb when compared to the traditional and Stander Cane®. A visualization of these results can be found in Figure 19.

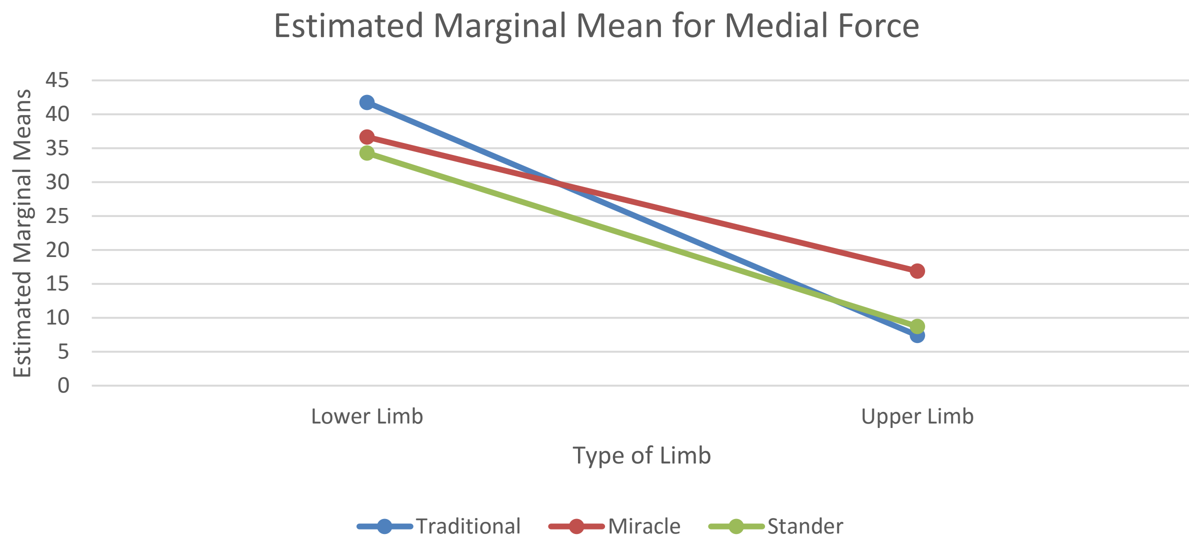

Figure 19. Estimated Marginal Means for Medial Force.

Lateral force. A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and limb type (upper and lower) on lateral GRF.

Assumptions.

Outliers. There were six outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. Three of these values were removed as per the previous analysis. Specifically, participant 9 lower limb lateral force data was once again removed due to measurement error and validity concerns. The remainder of the outliers remained in the analysis as they were determined to be genuinely unusual values. That is, the researchers did not believe that these values were due to data entry or measurement errors; rather, they were attributed to natural differences in lateral force production between participants. These differences may be attributable to differences in cane-ambulation techniques, foot placement, or cane placement. Furthermore, the analysis was run with and without the inclusion of these outliers and yielded similar conclusions, deeming the effect of the outliers on the results to be minimal.

Normality. Lateral force was normally distributed, as assessed by a Shapiro-Wilk's Test ($p > .05$).

Sphericity. Mauchly's Test of Sphericity indicated that the assumption of sphericity was met for the two-way interaction, $\chi^2(2) = 0.044$, $p > .05$.

Descriptive statistics. Descriptive statistics, shown in Table 14, indicate that mean lateral GRF in the upper limb decreased in spring loaded cane conditions when compared to the traditional cane condition (23.52 ± 14.41 Newtons), but more so in the Stander Cane® condition (18.37 ± 8.69 Newtons) than the Miracle Cane® condition (21.24 ± 11.29 Newtons). Mean lateral GRF in the lower limb increased in the spring loaded cane conditions when compared to the traditional cane condition (80.2275 ± 26.95 Newtons), but more so in the Miracle Cane condition® (89.64 ± 26.57 Newtons) than the Stander Cane® condition (82.5 ± 25.23 Newtons).

Table 14

*Descriptive statistics for cane type * limb type on lateral force*

	Type of Limb (Upper or Lower)	Mean	Std. Deviation	N
Lateral force measured when ambulating with the traditional cane	Lower Limb	80.23	26.95	20
	Upper Limb	23.52	14.41	21
	Total	51.18	35.67	41
Lateral force measured when ambulating with the Miracle Cane®	Lower Limb	89.64	26.57	20
	Upper Limb	21.24	11.29	21
	Total	54.60	39.97	41
Lateral force measured when ambulating with the Stander Cane®	Lower Limb	82.50	25.23	20
	Upper Limb	18.37	8.69	21
	Total	49.66	37.33	41

Inferential statistics. As indicated in Table 15, there was a statistically significant interaction between the type of cane and type of limb on lateral GRF, $F(2,78) = 5.29$, $p < .05$, partial $\eta^2 = .12$. Given the descriptive statistics in Table 14 and illustration of the interaction in Figure 20, the interaction appears to lie in the difference between lateral force production during cane conditions on the lower limb. That is, lateral force production on the lower limb appeared to be greatest in the Miracle Cane® condition when compared to the traditional and Stander Cane®.

Table 15

*Interaction effect for cane type * limb type on lateral force*

Source		**Type III Sum of Squares**	**df**	**Mean Square**	**F**	**Sig.**	**partial η^2**
CaneType	Sphericity Assumed	543.318	2	271.659	4.011	.022	.093
	Greenhouse-Geisser	543.318	1.998	271.971	4.011	.022	.093
	Huynh-Feldt	543.318	2.000	271.659	4.011	.022	.093
	Lower-bound	543.318	1.000	543.318	4.011	.052	.093
CaneType * LimbType	Sphericity Assumed	717.071	2	358.536	5.294	.007	.120
	Greenhouse-Geisser	717.071	1.998	358.947	5.294	.007	.120
	Huynh-Feldt	717.071	2.000	358.536	5.294	.007	.120
	Lower-bound	717.071	1.000	717.071	5.294	.027	.120
Error(Cane Type)	Sphericity Assumed	5282.174	78	67.720			
	Greenhouse-Geisser	5282.174	77.911	67.798			
	Huynh-Feldt	5282.174	78.000	67.720			
	Lower-bound	5282.174	39.000	135.440			

Simple main effects can be used to further explain the interaction effect. To test for the simple main effect of limb type, three separate univariate ANOVAs were performed. Each of these statistical tests were used to analyze the differences in lateral GRF between limbs (upper and lower) at each category of the within-subjects factor, cane type. The first of these ANOVAs tested for the difference between upper limb and lower limb lateral GRF while ambulating with the traditional cane. There was a statistically significant difference in lateral GRFs between the upper limb and lower limb while ambulating with the traditional cane $F(1, 39) = 71.57$, $p < .05$, partial $\eta^2 = .64$. The second ANOVA tested for the difference between upper and lower limb lateral GRF while ambulating with the Miracle Cane®. There was a statistically significant difference in lateral GRFs between the upper limb and lower limb while ambulating with the Miracle Cane®, $F(1, 39) = 117.07$, $p < .05$, partial $\eta^2 = .75$. The third ANOVA tested for the difference between upper and lower limb lateral GRF while ambulating with the Stander Cane®. There was a statistically significant difference in lateral GRFs between the upper limb and lower limb while ambulating with the Stander Cane®, $F(1, 39) = 120.73$, $p < .05$, partial $\eta^2 = .75$.

To test for the simple main effect of cane type, two separate ANOVAs were performed. Each statistical test examined the differences in lateral GRF between cane types for each category of the between-subject factor, limb type. The first ANOVA tested for the difference between lateral forces created by each cane type on the lower limb. There was a statistically significant effect of cane type on lateral GRF for the lower limb, $F(2,38) = 5.44$, $p < .05$. Further examination of pairwise comparisons reveal that there are significant differences in lateral force production between the Miracle Cane® and the traditional cane on the lower limb. There was no significant difference in lateral force production between the traditional and Stander Cane® or the Miracle Cane® and Stander Cane® on the lower limb. In combination with descriptive statistics, the results show that the Miracle Cane® produced greater lateral forces on the lower limb when compared to the traditional cane. The second ANOVA tested for the difference between lateral forces created by each cane type on the upper limb. There was no statistically significant effect of cane type on lateral GRF for the upper limb, $F(2, 40) = 2.91$, $p > .05$. Visualization of these results can be found in Figure 20.

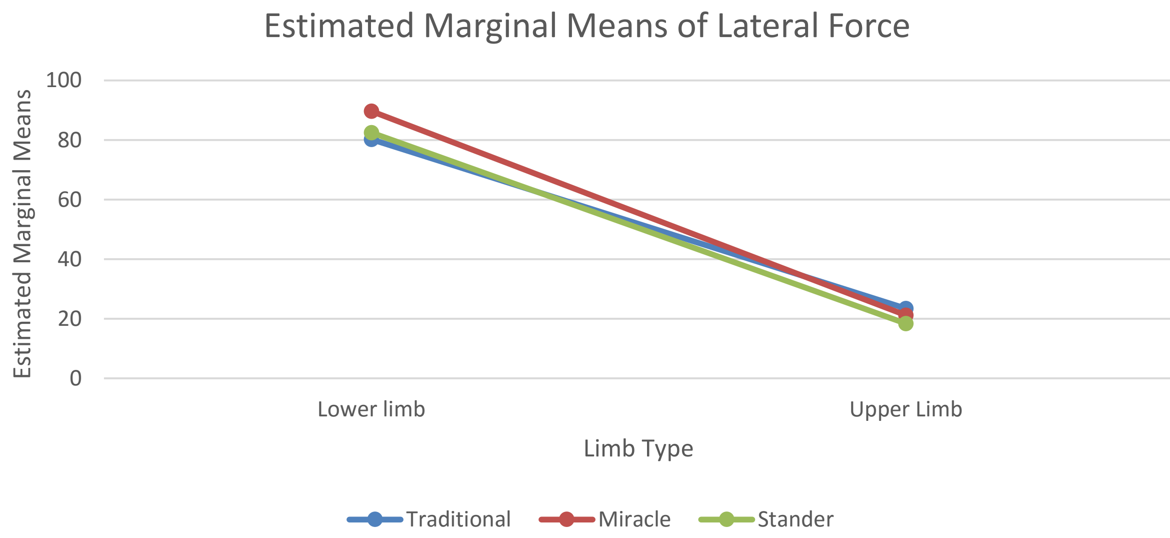

Figure 20. Estimated Marginal Means of Lateral Force.

Question 3

Vertical impulse. A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and limb type (upper and lower) on vertical impulse.

Assumptions.

Outliers. There were three outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. These values remained in the analysis as they were determined to be genuinely unusual values. That is, the researchers did not believe that these values were due to data entry or measurement errors; rather, they were attributed to natural differences in impulse production between participants. These differences may be attributable to differences in cane-ambulation techniques. Furthermore, the analysis was ran with and without the inclusion of these outliers and yielded similar conclusions, deeming the effect of the outliers on the results to be minimal. Although they were not considered outliers in this specific analysis, lower limb vertical impulse data for participant 9 was removed from the analysis as the data was thought to include measurement error and lack validity. That is, the force plate may not have been normalized for body weight, which may have created extreme outliers in subsequent analyses. To ensure that the removal of this data did not influence the results greatly, the ANOVA procedure was run with and without the inclusion of the data and yielded similar results.

Normality. Vertical impulse was normally distributed with the exception of the lower limb/traditional cane condition. Normality was assessed by a Shapiro-Wilk's Test ($p > .05$).

Sphericity. Mauchly's Test of Sphericity indicated that the assumption of sphericity was met for the two-way interaction, $\chi^2(2) = 5.547$, $p > .05$.

Descriptive statistics. Descriptive statistics, shown in Table 16, indicate that mean vertical impulse in the upper limb decreased in spring loaded cane conditions when compared to the traditional cane condition (99.31 ± 38.16 Newtons), but more so in the Miracle Cane® condition (92.63 ± 25.52 Newtons) than the Stander Cane® condition (98.38 ± 35.29 Newtons). Mean vertical impulse in the lower limb increased in the spring loaded cane conditions when compared to the traditional cane condition (432.04 ± 108.89 Newtons), but more so in the Miracle Cane® condition (472.78 ± 99.5 Newtons) than the Stander Cane® condition (446.26 ± 97.91 Newtons).

Table 16

*Descriptive statistics for cane type * limb type on vertical impulse*

	Type of Limb (Upper or Lower)	Mean	Std. Deviation	N
Vertical Impulse measured when ambulating with the traditional cane	Lower Limb	432.04	108.89	20
	Upper Limb	99.31	38.16	21
	Total	261.62	186.31	41
Vertical Impulse measured when ambulating with the Miracle Cane®	Lower Limb	472.78	99.49	20
	Upper Limb	92.63	25.52	21
	Total	278.07	205.03	41
Vertical Impulse measured when ambulating with the Stander Cane®	Lower Limb	446.26	97.91	20
	Upper Limb	98.38	35.29	21
	Total	268.08	190.18	41

Inferential statistics. As indicated in Table 17, there was a statistically significant interaction between the type of cane and type of limb on vertical impulse, $F(2,78) = 9.93$, $p < .05$, partial $\eta^2 = .2$. Given the descriptive statistics in Table 16 and illustration of the interaction in Figure 21, the interaction appears to lie in the difference between vertical impulse production during cane conditions on the lower limb. That is, vertical impulse production on the lower limb appears to be greater in the Miracle Cane® condition when compared to the traditional and Stander Cane®.

Table 17

*Interaction effect for cane type * limb type on vertical impulse*

Source		**Type III Sum of Squares**	**df**	**Mean Square**	**F**	**Sig.**	**partial η^2**
CaneType	Sphericity Assumed	6036.202	2	3018.101	4.987	.009	.113
	Greenhouse-Geisser	6036.202	1.761	3428.037	4.987	.012	.113
	Huynh-Feldt	6036.202	1.885	3202.305	4.987	.011	.113
	Lower-bound	6036.202	1.000	6036.202	4.987	.031	.113
CaneType * LimbType	Sphericity Assumed	12019.166	2	6009.583	9.930	.000	.203
	Greenhouse-Geisser	12019.166	1.761	6825.840	9.930	.000	.203
	Huynh-Feldt	12019.166	1.885	6376.367	9.930	.000	.203
	Lower-bound	12019.166	1.000	12019.166	9.930	.003	.203
Error(Cane Type)	Sphericity Assumed	47204.659	78	605.188			
	Greenhouse-Geisser	47204.659	68.672	687.388			
	Huynh-Feldt	47204.659	73.513	642.124			
	Lower-bound	47204.659	39.000	1210.376			

Simple main effects can be used to further explain the interaction effect. To test for the simple main effect of limb type, three separate univariate ANOVAs were performed. Each of these statistical tests was used to analyze the differences in vertical impulse between limbs (upper and lower) at each category of the within-subjects factor, cane type. The first of these ANOVAs tested for the difference between upper limb and lower limb vertical impulse while ambulating with the traditional cane. There was a statistically significant difference in vertical impulse between the upper limb and lower limb while ambulating with the traditional cane $F(1, 39) = 173.85$, $p < .05$, partial $\eta^2 = .81$. The second ANOVA tested for the difference between upper and lower limb vertical impulse while ambulating with the Miracle Cane®. There was a statistically significant difference in impulse between the upper limb and lower limb while ambulating with the Miracle Cane®, $F(1, 39) = 287.1$, $p < .05$, partial $\eta^2 = .88$. The third ANOVA tested for the difference between upper and lower limb vertical impulse while ambulating with the Stander Cane®. There was a statistically significant difference in vertical impulse between the upper limb and lower limb while ambulating with the Stander Cane®, $F(1, 39) = 233.52$, $p < .05$, partial $\eta^2 = .85$.

To test for the simple main effect of cane type, two separate ANOVAs were performed. Each statistical test examined the differences in vertical impulse between cane types for each category of the between-subject factor, limb type. The first ANOVA tested for the difference between vertical impulses created by each cane type on the lower limb. There was a statistically significant effect of cane type on impulse for the lower limb, $F(2, 38) = 7.72$, $p < .05$. Further examination of pairwise comparisons indicated that there were significant differences between vertical impulses produced during ambulation with the traditional and Miracle Cane® in the lower limb. No significant differences were found in vertical impulse production between the traditional and Stander Cane® or the Miracle Cane® and Stander Cane® in the lower limb. In combination with descriptive statistics, the results show that the Miracle Cane® produced greater vertical impulse on the lower limb when compared to the traditional cane. The second ANOVA tested for the difference between vertical impulses created by each cane type on the upper limb. There was no statistically significant effect of cane type on vertical impulse for the upper limb, $F(2, 40) = 2.14$, $p > .05$. A visualization of these result can be found in Figure 21.

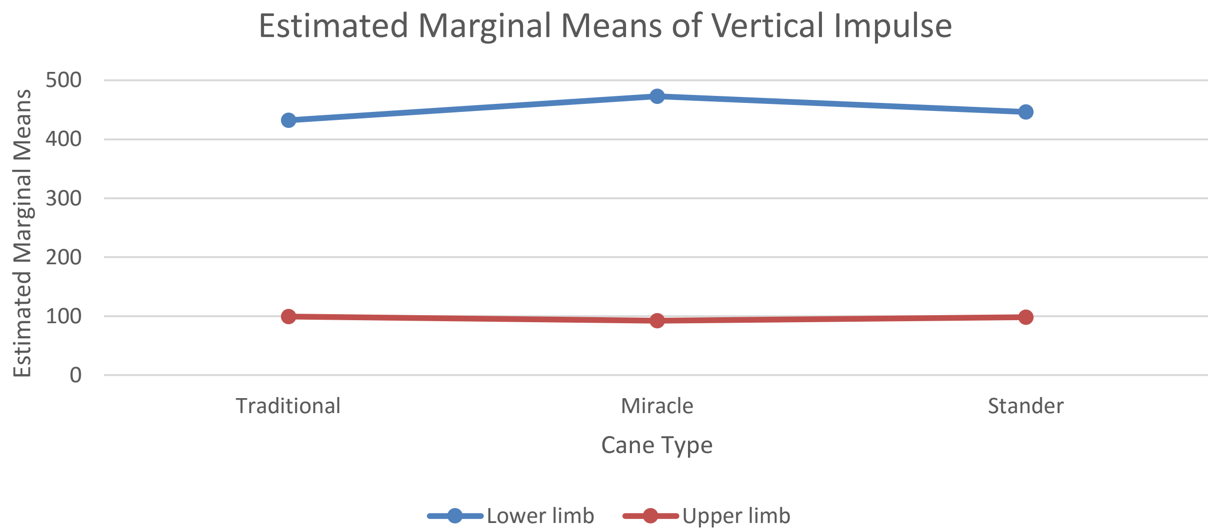

Figure 21. Estimated Marginal Means of Vertical Impulse.

Anterior impulse. A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and limb type (upper and lower) on anterior impulse.

Assumptions.

Outliers. There were no outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. Although they were not considered outliers in this specific analysis, lower limb anterior impulse data for participant 9 was removed from the analysis as the data was thought to include measurement error and lack validity. That is, the force plate may not have been normalized for body weight, which may have created extreme outliers in subsequent analyses. To ensure that the removal of this data did not influence the results greatly, the ANOVA procedure was run with and without the inclusion of the data and yielded similar results.

Normality. Anterior impulse was normally distributed as assessed by a Shapiro-Wilk's Test ($p > .05$).

Sphericity. Mauchly's Test of Sphericity indicated that the assumption of sphericity was met for the two-way interaction, $\chi^2(2) = 4.852$, $p > .05$.

Descriptive statistics. Descriptive statistics, shown in Table 18, indicate that there was minimal differences in mean anterior impulse in the upper limb between the traditional cane (5.31 ± 3.13 Newtons), Miracle Cane® (4.74 ± 2.77 Newtons), and Stander Cane® (5.55 ± 3.12 Newtons). Minimal differences were also seen in mean anterior impulse in the lower limb between the traditional cane (19.28 ± 5.46 Newtons), Miracle Cane® (19.66 ± 4.84 Newtons), and Stander Cane® (20.13 ± 6.23 Newtons).

Table 18

*Descriptive statistics for cane type * limb type on anterior impulse*

	Type of Limb (Upper or Lower)	Mean	Std. Deviation	N
Anterior Impulse measured when ambulating with the traditional cane	Lower Limb	19.28	5.46	20
	Upper Limb	5.31	3.13	21
	Total	12.13	8.31	41
Anterior Impulse measured when ambulating with the Miracle Cane®	Lower Limb	19.66	4.84	20
	Upper Limb	4.74	2.77	21
	Total	12.02	8.48	41
Anterior Impulse measured when ambulating with the Stander Cane®	Lower Limb	20.13	6.23	20
	Upper Limb	5.55	3.12	21
	Total	12.66	8.82	41

Inferential statistics. As indicated in Table 19, there was no statistically significant interaction between the type of cane and type of limb on anterior impulse, $F(2,78) = 0.71$, $p > .05$, partial $\eta^2 = .01$.

Table 19

*Interaction effect for cane type * limb type on anterior impulse*

Source		**Type III Sum of Squares**	**df**	**Mean Square**	**F**	**Sig.**	**partial η²**
CaneType	Sphericity Assumed	9.75	2.00	4.87	1.47	0.24	.036
	Greenhouse-Geisser	9.75	1.79	5.46	1.47	0.24	.036
	Huynh-Feldt	9.75	1.91	5.09	1.47	0.24	.036
	Lower-bound	9.75	1.00	9.75	1.47	0.23	.036
CaneType * LimbType	Sphericity Assumed	4.77	2.00	2.39	0.72	0.49	.018
	Greenhouse-Geisser	4.77	1.79	2.67	0.72	0.48	.018
	Huynh-Feldt	4.77	1.91	2.49	0.72	0.49	.018
	Lower-bound	4.77	1.00	4.77	0.72	0.40	.018
Error(Cane Type)	Sphericity Assumed	259.43	78.00	3.33			
	Greenhouse-Geisser	259.43	69.65	3.73			
	Huynh-Feldt	259.43	74.64	3.48			
	Lower-bound	259.43	39.00	6.65			

The main effect of limb type, determined through interpretation of Table 20, showed that there was a statistically significant difference in anterior impulse production between the upper and lower limb, $F(1,39) = 123.73$, $p < .05$, partial $\eta^2 = 0.76$.

Table 20

Main effect of limb type on anterior impulse

Source	**Type III Sum of Squares**	**df**	**Mean Square**	**F**	**Sig.**	**Partial Eta Squared**
Intercept	19047.698	1	19047.698	365.378	.000	.904
LimbType	6450.280	1	6450.280	123.731	.000	.760
Error	2033.129	39	52.132			

The main effect of cane type, determined through interpretation of Table 19, showed that there was no statistically significant difference in anterior impulse production between cane types, $F(2, 78) = 1.46$, $p > .05$, partial $\eta^2 = 0.03$. A visualization of these result can be found in Figure 22.

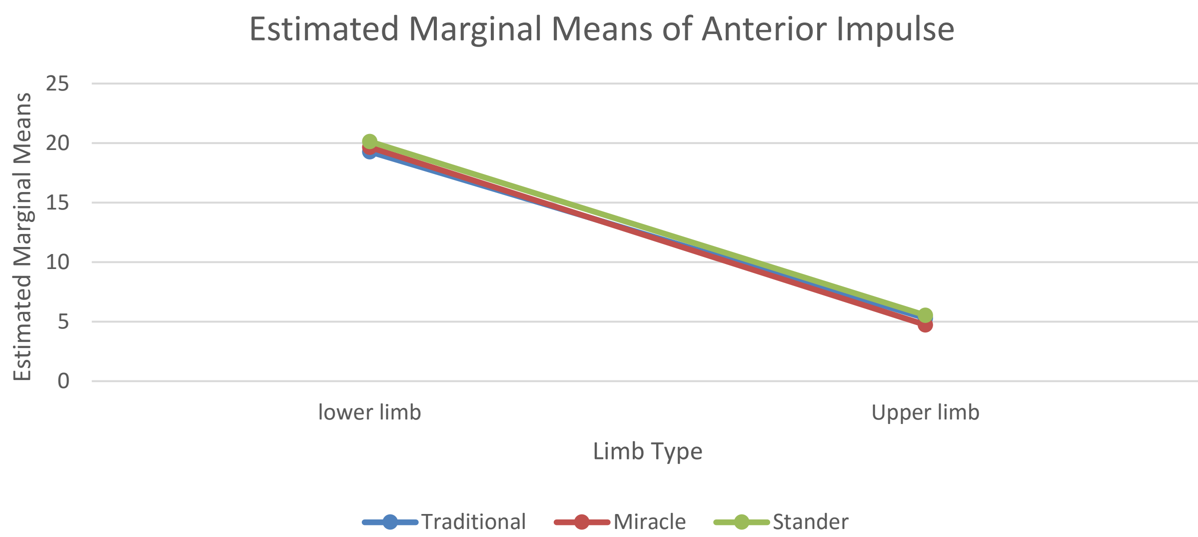

Figure 22. Estimated Marginal Means of Anterior Impulse.

Posterior impulse. A two-way mixed factorial ANOVA was performed to determine if there was a significant interaction between cane type (traditional cane, Miracle Cane®, and Stander Cane®) and limb type (upper and lower) on posterior impulse.

Assumptions.

Outliers. There were eight outliers in the data, as assessed by inspection of a box plot for values greater than 1.5 box-lengths from the edge of the box. These values remained in the analysis as they were determined to be genuinely unusual values. That is, the researchers did not believe that these values were due to data entry or measurement errors; rather, they were attributed to natural differences in impulse production between participants. These differences may be attributable to differences in cane-ambulation techniques. Furthermore, the analysis was run with and without the inclusion of these outliers and yielded similar conclusions, deeming the effect of the outliers on the results to be minimal. Although they were not considered outliers in this specific analysis, lower limb posterior impulse data for participant 9 were removed from the analysis as the data was thought to include measurement error and lack validity. That is, the force plate may not have been normalized for body weight, which may have created extreme outliers in subsequent analyses. To ensure that the removal of this data did not influence the results greatly, the ANOVA procedure was run with and without the inclusion of the data and yielded similar results.

Normality. Posterior impulse was not normally distributed in most conditions as assessed by a Shapiro-Wilk's Test ($p < .05$). Posterior impulse was normally in the upper limb/traditional cane condition.

Sphericity. Mauchly's Test of Sphericity indicated that the assumption of sphericity was not met for the two-way interaction, $\chi^2(2) = 12.07$, $p < .05$. The Greenhouse-Geisser adjustment was used to ensure that the results of the analysis were not biased and to minimize the possibility of committing a type I error.

Descriptive statistics. Descriptive statistics, shown in Table 21, indicate that there was minimal differences in mean posterior impulse in the upper limb between the traditional cane (7.44 ± 3.52 Newtons), Miracle Cane® (7.12 ± 3.68 Newtons), and Stander Cane® (7.58 ± 4.19 Newtons). Minimal differences were also seen in mean posterior impulse in the lower limb between the traditional cane (24.46 ± 19.93 Newtons), Miracle Cane® (21.64 ± 18.02 Newtons), and Stander Cane® (24.76 ± 19.39 Newtons).

Table 21

*Descriptive statistics for cane type * limb type on posterior impulse*

	Type of Limb (Upper or Lower)	Mean	Std. Deviation	N
Posterior Impulse measured when ambulating with the traditional cane	Lower Limb	24.46	19.93	20
	Upper Limb	7.44	3.53	21
	Total	15.74	16.40	41
Posterior Impulse measured when ambulating with the Miracle Cane®	Lower Limb	21.64	18.02	20
	Upper Limb	7.12	3.68	21
	Total	14.21	14.66	41
Posterior Impulse measured when ambulating with the Stander Cane®	Lower Limb	24.76	19.39	20
	Upper Limb	7.58	4.19	21
	Total	15.96	16.22	41

Inferential statistics. As indicated in Table 22, there was no statistically significant interaction between the type of cane and type of limb on posterior impulse, $F(1.572,61.315) = 1.9$, $p > .05$, partial $\eta^2 = .04$.

Table 22

*Interaction effect for cane type * limb type on posterior impulse*

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	partial η^2
CaneType	Sphericity Assumed	77.48	2.00	38.74	3.24	0.05	0.08
	Greenhouse-Geisser	77.48	1.57	49.28	3.24	0.06	0.08
	Huynh-Feldt	77.48	1.67	46.43	3.24	0.05	0.08
	Lower-bound	77.48	1.00	77.48	3.24	0.08	0.08
CaneType * LimbType	Sphericity Assumed	45.64	2.00	22.82	1.91	0.16	0.05
	Greenhouse-Geisser	45.64	1.57	29.03	1.91	0.17	0.05
	Huynh-Feldt	45.64	1.67	27.35	1.91	0.16	0.05
	Lower-bound	45.64	1.00	45.64	1.91	0.18	0.05
Error(Cane Type)	Sphericity Assumed	933.65	78.00	11.97			
	Greenhouse-Geisser	933.65	61.32	15.23			
	Huynh-Feldt	933.65	65.08	14.35			
	Lower-bound	933.65	39.00	23.94			

The main effect of limb type, determined through interpretation of Table 23, showed that there was a statistically significant difference in posterior impulse production between the upper and lower limb, $F(1,39) = 15.19$, $p < .05$, partial $\eta^2 = 0.28$.

Table 23

Main effect of limb type on posterior impulse

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	29528.777	1	29528.777	55.362	.000	.587
LimbType	8103.869	1	8103.869	15.194	.000	.280
Error	20801.651	39	533.376			

The main effect of cane type, determined through interpretation of Table 22, showed that there was no statistically significant difference in posterior impulse production between cane types, $F(1.572,61.315) = 3.23$, $p > .05$, partial $\eta^2 = 0.07$. A visualization of these results can be found in Figure 23.

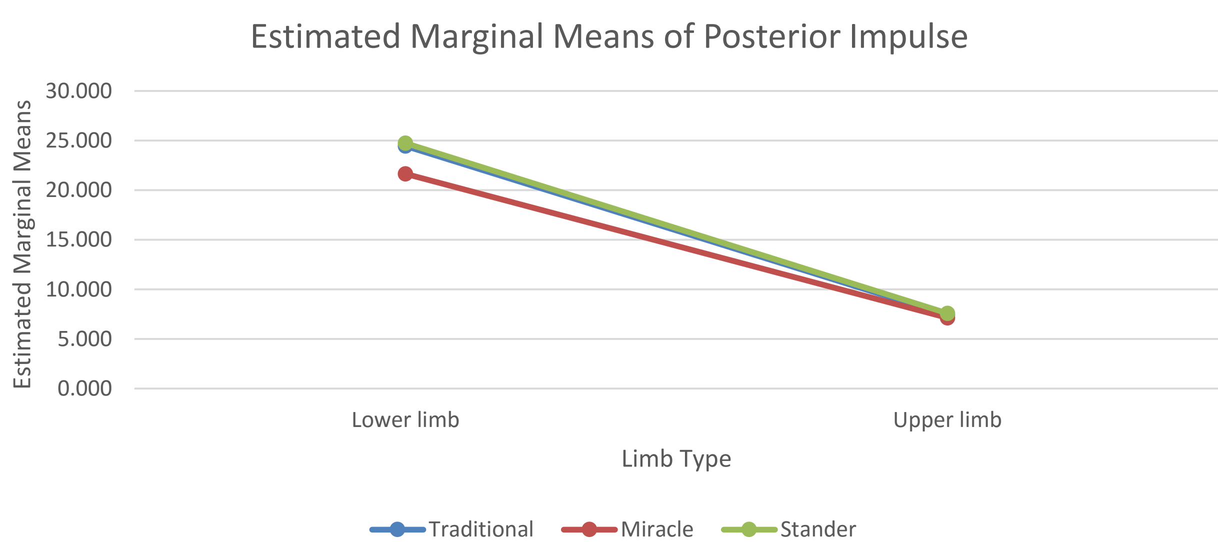

Figure 23. Estimated Marginal Means of Posterior Impulse.

Question 4

A descriptive statistics procedure was performed in order to examine participants' subjective preferences in terms of cane types. This procedure allowed the researchers to understand which cane was preferred, overall.

Descriptive statistics. Descriptive statistics, shown in Table 24, indicate that individuals preferred the spring loaded canes over the traditional cane (1.71 ± 0.64 points); however, participants preferred the Stander Cane® (2.33 ± 0.86) over the Miracle Cane® (1.95 ± 0.86).

Table 24

Descriptive Statistics for Ease of Use Questionnaire

	N	Mean	Std. Deviation
TradintionalCaneScore	21	1.7143	.64365
MiracleCane®Score	21	1.9524	.86465
StanderCane®Score	21	2.3333	.85635
Valid N (listwise)	21		

Participants were also asked to provide a qualitative assessment of the canes by stating the reasoning behind their ranking. Table 25 provides a summary of the most common responses and the percentage of individuals that provided these responses.

Table 25

Common Responses for Ease of Use Questionnaire

Common Response	Percentage of Individuals
Stander Cane® handle more comfortable than Miracle Cane® and traditional	28.5%
Stander Cane® not as stiff as traditional cane	23.8%
Too much spring in Miracle Cane®/hard to control and balance	57.1%
Miracle Cane® spring and Stander Cane® handle ideal	19%
Miracle Cane® has more support for you to lean on/others are hard on the joints	14.3%

Chapter 4 - Discussion

Reliability

It is well documented in measurement literature that to test a hypothesis and make inferences from test result interpretations, it is critical to assess and provide evidence of reliability of the measures obtained from the instruments and testing protocols (Kane, 2006). The outcome of the reliability analysis revealed that all ICC values were greater than .558 and demonstrated strong correlations between values collected from the force platforms, EMG sensors, kinematic variables, and Brower timing gates. Since all participants were healthy individuals, the variability between ICC values (.558-.999) for various measures may be explained by the participants' ability to ambulate with a cane-assisted device. As the results indicate, upper extremity measures often resulted in greater variability than lower extremity measures due to greater variation in the pressure applied to each cane; however, it is important to note that these measures were still highly reliable in healthy participants. Reliability results of kinematic variables indicated that participants were consistent in their ambulation techniques.

The findings of this reliability assessment provided evidence that these measures were an accurate and precise method of assessing force, impulse, EMG, walking speed, and kinematic variables during cane-assisted locomotion. This method can be used in future research to assess differences between traditional and spring loaded canes with regards to the aforementioned kinetic and kinematic variables.

Walking Speed

It is known that as the speed of force application changes, there is a change in the complex interaction between vertical and horizontal forces needed for propulsion and equilibrium during human locomotion (Nilsson & Thorstensson, 1989). This interaction illustrates the importance of assessing the differences in walking speed between cane conditions prior to reporting the results of this study. Since there was no difference in mean walking speed between cane conditions, it is safe to report the results without concern of the interaction with walking speed.

Electromyography

Although no studies had examined differences in upper limb muscle activation between traditional and spring loaded canes, Chiou-Tan et al. (1999) examined these differences in canes with varying handle types. The researchers of this study found that changing the handle type into a more ergonomic form led to significant decreases in muscle activation in 12 muscles of the forearm, arm, shoulder, and back. These findings were in line with the findings of the present study, which showed that muscle activation was significantly decreased in seven muscles of the forearm, arm, chest, and shoulder when participants used the Miracle Cane® in comparison to the traditional cane. Previous kinematic analyses indicated that during the support phase, the cane produced vertical ground reaction forces on the wrist, which created a wrist extension moment (Ter, Parasuraman, Khan, & Elamvazuthi, 2015). The total flexor force must then balance the wrist extension moment. Changes in the location of the cane's ground reaction force creates changes in muscle and joint reaction forces. Although no changes were made in the location of the ground reaction force on the wrist in the present study, significant changes were found in the magnitude of vertical ground reaction forces placed on the upper limb between the traditional and Miracle Cane®. These changes may explain the significant differences found in upper limb muscle activation between the aforementioned canes. That is, the decreases in vertical force placed on the wrist may have led to decreases in joint muscle and joint reaction forces. An alternative explanation for the change in muscle activation between cane types is that participants may have used different ambulation techniques for each cane type.

It is well documented in the cane literature that chronic cane use leads to repetitive stress disorders such as tendinitis, carpal tunnel syndrome, and osteoarthritis; however, little has been done to augment current cane designs to address these issues (Chiou-Tan et al., 1999; Florack, et al., 1992; Gitlin & Burgh, 1995; Koh et al., 2002; Mann et al., 1993; Parr & Faillace, 1999; Schemm & Gitlin, 1998; Yosipovitch & Yosipovitch, 1993). The findings of the present study indicate that spring loaded canes, specifically the Miracle Cane®, may decrease the likelihood of these injuries by decreasing upper arm muscle activation and, thereby, joint loading; however, further research needs to be completed assessing the differences between these canes in injured individuals who regularly use canes. Furthermore, kinematic analyses need to be completed to examine if any differences exist in ambulation techniques between traditional and spring loaded canes.

Force and Impulse

Based on the findings of Shoup (1980) and Parziale and Daniels (1989), who assessed forces and impulses in spring loaded crutches, it was hypothesized that peak vertical ground reaction forces and vertical impulse would decrease in the upper limb when participants used spring loaded canes than when they used the traditional cane. As expected, the maximum vertical ground reaction force decreased in the upper limb in the Miracle Cane® condition when compared to the traditional cane. Contrary to our hypothesis, however, were the findings that the impulse of the vertical ground reaction force were not significantly different on the upper limb for spring loaded canes.

Parziale and Daniels (1989) added springs to the shafts of standard crutches but measured uniaxial forces at the crutch handles rather than the ground reaction forces. The authors reported handle forces that were 24% lower for the spring loaded crutches than the standard crutches. These findings were similar to the differences found in vertical force production between the Miracle Cane® and Traditional Cane®, in which the spring loaded cane produced vertical ground reaction forces that were 25% lower than the traditional cane. The second spring loaded cane condition in the present study, Stander Cane®, did not produce similar results as no differences were found between the Stander Cane® and the traditional cane in terms of vertical force production on the upper limb. The differences in methodology (forces measured at the handle versus crutch tip and using only one spring stiffness) and transmission of torque at the crutch handle may be the cause of this discrepancy. Future studies should consider simultaneous measurement of these forces as both are likely to have implications for injury. Studies examining vertical forces at the cane handle and cane tip may have implications on the rates of cane abandonment and disuse. If these measures should agree with the findings of the present study, the likelihood of upper limb injuries may decrease and individuals may be more likely to continue using these assistive devices.

Shoup (1980) reported a reduction of the initial force transient for a single subject, which was contrary to the lack of significant differences in vertical impulse on the upper limb found in 21 participants in the present study. These findings were also contrary to the findings of Segura and Piazza (2007), who reported that the average impulse produced on the upper limb was significantly lower with spring loaded crutches than with standard crutches. The unexpected lack of differences in vertical impulse on the upper limb between spring loaded and traditional canes may have been caused by "bottoming out" that might have occurred if the spring became fully compressed during cane stance. Although calculations were performed using subjects' body weights and spring constants to ensure that bottoming out did not occur, incorporation of a spring deflection measurement into the methodology would aid in determining if this is true. Some participants described experiencing this sensation during Stander Cane® trials. Future research examining weight limits for each of these spring constants would aid clinicians in making informed decisions regarding cane prescription. An alternative explanation for the discrepancy in vertical impulse is that the subjects may have used different ambulation techniques for each cane type. Although it is known that vertical ground reaction forces differ between the traditional cane and the Miracle Cane®, when using the Miracle Cane® condition, the participants' hand fell through a greater distance before the spring arrested the fall of participants' hand, causing cane contact time to be lengthened. Knowing that impulse is the product of force and time, a decrease in force accompanied by an increase in the time that the cane is in contact with the ground could result in similar impulses between cane types in situations when the impulse is given by an increase in force and decrease in time.

Although no previous research has examined the differences between traditional and spring loaded canes on forces and impulses on the lower limb, mediolateral forces, or AP forces, the findings of the present study support the aforementioned theory that ambulation techniques may differ between cane types (Guild et al., 2012). A significant increase in vertical force production on the lower limb in spring loaded cane conditions when compared to traditional cane conditions indicates that the participants shifted more of their weight onto the simulated injured limb during spring loaded cane conditions, but more so in the Miracle Cane® condition than the Stander Cane® condition. Significant increases in medial force on the upper limb between spring loaded cane conditions and traditional cane conditions accompanied by significant increases in lateral force on the lower limb between spring loaded and traditional cane conditions further support the theory that participants may be shifting their weight towards the side of the simulated injury. A significant increase in anterior force produced on the upper limb when using the Miracle Cane® in comparison to the traditional cane supports the aforementioned theory that participants may have fallen through a greater distance before the spring arrested their falls (Segura & Piazza, 2007). A significant increase in vertical impulse produced by the lower limb during the Miracle Cane® condition when compared to the traditional cane condition supports the aforementioned theory that participants spent a greater amount of time with the cane and not the foot in contact with the ground (Bennett et al., 1979). Since vertical force below the lower limb varied at a similar rate as upper limb vertical force between cane conditions, the varying factor between the upper and lower limb is the time spent in contact with the plate; however, it is important to note that walking speed was consistent across cane types and did not have an impact on the overall results of the study.

The results indicating that vertical ground reaction forces produced on the upper limb were lower when individuals used the Miracle Cane® have important implications on the likelihood of overuse injuries. Because injuries such as tendinitis, carpal tunnel syndrome, and osteoarthritis are caused by repetitive stresses on the upper limb, it is likely that walking with spring loaded canes lessens the risk of injury to cane users. These findings, coupled with the EMG findings, may indicate that spring loaded canes decrease skeletal impact loading, thus, decreasing the likelihood of the aforementioned overuse injuries. A reduction in vertical forces on the upper limb indicates that the Miracle Cane® may contribute to a reduction in the likelihood of upper limb injuries; however, the findings that the Miracle Cane® also contributes to a shift in weight onto the limb with the simulated injury may also be useful to patients and practitioners. In the rehabilitation process for a multitude of lower limb injuries, patients are often instructed to maintain minimal weight bearing on the side of the injury until they are cleared to ambulate without an assistive device (Chinn & Hertel, 2010). Weight bearing status is often dependent on the physiological healing that must take place. In these cases, weight bearing is often an essential part of the healing process and is implemented through progressive exercises using free weights, resistance bands, and other forms of progressive strengthening (Chinn & Hertel, 2010). The Miracle Cane® may offer a mid-way point between the use of a standard assistive device and clearance for ambulation without an assistive device. That is, canes such as the Miracle Cane® can aide in the rehabilitative process for lower limb injuries by gradually returning weight onto the site of the injury. Early active muscle re-education and early weight bearing that follows the specific instructions provided by the primary healthcare provider and falls within the pain tolerance of the patient has been shown to encourage tone, blood flow, callus formation within the healing bone, faster fracture healing rate, and reduced swelling and stiffness (Bunker, Colton, & Webb, 1989). Therefore, gradual and early weight bearing through the use of the Miracle Cane® may lead to shortened healing times for lower limb injuries. It is difficult to make these conclusions, however, as further research needs to be completed examining the differences between spring loaded canes and traditional canes in habitual cane users. Further research also needs to address the kinematic and kinetic differences in walking patterns between cane types. Examination of kinematic and kinetic differences would aid in understanding whether the differences in force are truly attributable to the spring loaded device or changes in ambulation techniques.

Ease of Use Questionnaire

When participants (n=21) were asked to rate which cane they preferred after having used all three cane types, the Stander Cane® was the preferred choice; the Miracle Cane® was rated as their second choice and the traditional cane was the least preferred. One theme that seemed to be consistent among most participants was that the spring loaded canes felt more comfortable than the traditional cane; however, the soft spring of the Miracle Cane® caused feelings of imbalance for 57% of the participants. These findings are in line with the findings of Segura and Piazza (2007), who indicated that participants found spring loaded crutches to be more comfortable than their standard alternatives. The results of this questionnaire support the inferential analysis conducted previously as both spring loaded canes often produced lower forces, impulses, and EMG on the upper limb than the traditional cane, indicating that these canes may be perceived as more comfortable to use. It is important to note, however, that the ergonomic handle of the Stander Cane® may have contributed to the increased feelings of comfort when compared to the traditional cane and Miracle Cane®. It is difficult to know whether these feelings would be similar to those of habitual users as this was the participants' first encounter with the canes; however, if spring loaded canes, such as the Miracle Cane®, offer a trade of stability for comfort, they may not be suitable for use by patients with certain disorders such as lower quadrant hypermobility or neuromuscular disorders. The findings of this questionnaire illustrate the need for further research examining the effect on balance between these canes.

Conclusion

Current single-tip support cane designs are associated with pain and development of pathologies in compensatory structures such as the upper limb. Creators of spring loaded canes and cane users anecdotally claimed that the addition of spring mechanisms enabled the canes to decrease the amount of force translated to the upper limb and, thereby, reduce the pain and the likelihood of developing pathologies. Current research in the field is limited to anecdotal reports on spring loaded canes and preliminary studies on spring loaded crutches. The purpose of this study was to fill this gap in the literature by determining the effect of commercially available spring loaded single-tip support canes on minimizing ground reaction forces, impulse, and muscle EMG activation in the upper limb during ambulation while increasing ground reaction forces and impulse at the simulated injured lower limb as an approach for rehabilitation. The findings of the present study indicate that vertical force and EMG activation in the upper limb were decreased when the Miracle Cane® was used in comparison to the traditional cane. These findings may have implications on the likelihood of overuse injuries by way of decreased repetitive stresses placed on the upper limb and decreased skeletal loading. An increase in vertical forces and vertical impulses on the lower limb when using the Miracle Cane® when compared to the traditional cane suggests that these canes may be useful in lower limb injury rehabilitation via a progress and gradual return to weight bearing. Participants' subjective accounts also indicated that spring loaded canes are a more comfortable, yet less stable alternative to traditional canes. The findings of this research may have implications for the design of standard single-tip support canes and suggest avenues for future research. The findings of this research may also have implications on the rates of cane abandonment such that users may be more likely to continue using their assistive devices as negative feelings associated with upper limb pain may be minimized or eliminated. The clinical utility of these findings and the reported differences between cane types may assist clinicians in choosing the most appropriate cane for their patients.

Limitations of the Study

A number of limitations were present in the study. One of the largest limitations was that healthy participants were used rather than habitual cane users. The researchers chose to test healthy participants as they were easy to recruit in a short period and had minimal risk of falls or injury when ambulating with a cane. To minimize the effect that these healthy participants had on the results, participants were fitted with a knee brace to simulate injury and were given approximately 15 minutes to practice cane-assisted ambulation. Another limitation of the study was the use of surface electrodes to collect EMG values. Use of surface electrodes rather than indwelling fine wire EMG allows for surface movement, increased crosstalk between sensors, and less accurate measurements of deep muscles; however, indwelling electrodes were not available to the researchers at the time the study was performed.

Future Directions and Recommendations

Future work should examine the differences between these canes in habitual crutch users in terms of ground reaction forces, upper limb EMG, loads applied to the body, kinematic and kinetic variables, and balance constructs. Although the use of a knee brace to create a unilateral knee flexion contracture was an effective method of producing an antalgic gait pattern, the use of actual habitual cane users in such a study would be beneficial to most accurately represent this patient population and provide information on the effectiveness of these canes. Habitual cane users would produce a more consistent ambulatory pattern (given that the conditions being assessed remain consistent) with all types of canes and would produce a more accurate account of cane preference. Future research should also examine the differences between ground reaction forces and forces applied to the body (i.e., forces applied at the cane handle) when using traditional and spring loaded canes as both of these force types have implications on the likelihood for injury. Kinematics of cane ambulation techniques should also be addressed to determine whether differences in force are attributable to spring mechanisms or kinematic differences. Given that 57% of the participants in the present study reported decreases in feelings of balance when the Miracle Cane® was used, future research should also address differences in balance constructs between cane types.

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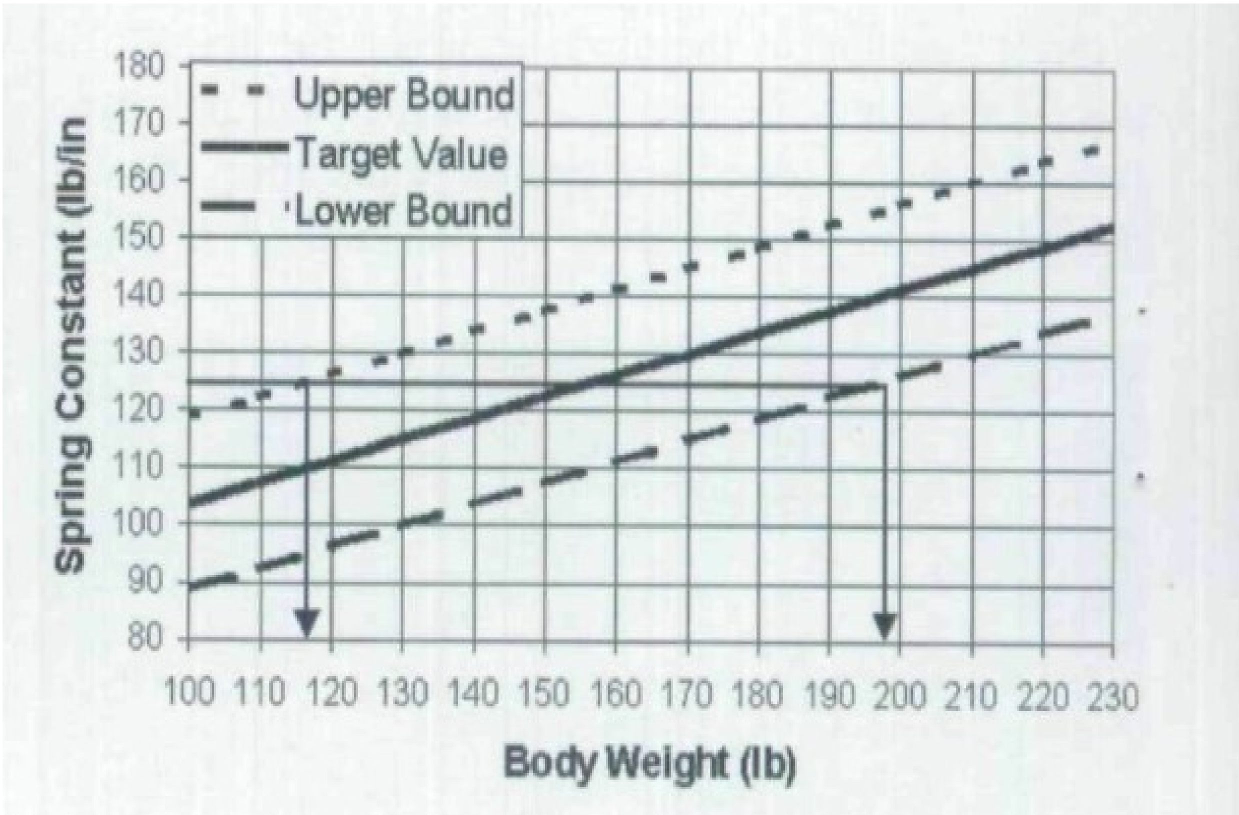
Appendix A

Protocol for Determining Weight Parameters



(Shortell et al., 2001)

Appendix B

Poster

Lakehead UNIVERSITY

Are you interested in Research?
Join our research study comparing traditional and spring-loaded

CANES

Testing involves an analysis of cane-assisted walking biomechanics!

Participants:

You are welcomed to participate if you are a healthy male/female age 18-45 with the ability to walk in a straight line, the ability to understand spoken and written instruction, and the capacity to give informed consent.

You also welcomed to participate if you have none of the following:

history of knee trauma or surgery within the past six months; the presence of an active inflammatory rheumatological condition; injury or amputation of the lower extremity; injury or condition of the upper extremity affecting the ability to use a cane; spinal or lower quadrant pain impeding gait; the presence of a neurological condition impeding gait; or poor health interfering with a gait assessment

You will be attending two meetings. The first meeting will take only 30 minutes of your time. During this time, you will be given an information letter, consent form, physical activity readiness questionnaire (PAR-Q) and demographic questionnaire to ensure you are eligible to participate. You will also be practice cane walking. If you qualify, you will be attending a second meeting, which will include a 60 minutes testing session that will take place at Lakehead University, Sanders Fieldhouse, room Sb1028. Testing will be completed under 3 different conditions: Traditional cane, Stander cane, and Miracle Cane.

Aya Mohammed asmohamm@lakeheadu.ca	Aya Mohammed asmohamm@lakeheadu.ca	Aya Mohammed asmohamm@lakeheadu.ca	Aya Mohammed asmohamm@lakeheadu.ca	Aya Mohammed asmohamm@lakeheadu.ca	Aya Mohammed asmohamm@lakeheadu.ca	Aya Mohammed asmohamm@lakeheadu.ca	Aya Mohammed asmohamm@lakeheadu.ca

Appendix C

Letter of Recruitment

School of Kinesiology
e: asmohamm@lakeheadu.ca

Dear Potential Participant,

We would like to invite you to participate in our research study. The title of our study is: "The Effect of a Spring Loaded Single-Tip Support Cane Mechanism on Upper and Affected Lower Limb Ground Reaction Forces, Muscle Activity, and Self-Perceived Ease of Use." Before you decide to participate in this study, we encourage you to read all the information provided in this letter, consent form, and health assessment questionnaires. This study is being conducted by Dr. Carlos Zerpa and Ms. Aya Mohammed, a graduate student from the School of Kinesiology at Lakehead University.

Our research aims to examine and compare different types of canes in terms of upper extremity muscle activity, force, and ease of use. For this preliminary study, only healthy individuals who do not necessarily require a cane will be recruited and a knee injury will be simulated by using a knee brace.

You are being invited to participate in this study because you are a healthy individual aged 18-45 with the ability to walk in a straight line, understand spoken and written instruction, and the capacity to give informed consent. Your participation in this study is voluntary; you may refuse to participate in any parts of the study, decline to answer any questions, and may withdraw from the study at any time.

If you are interested in participating in this study, you will be attending this initial meeting, which will take approximately 30 minutes of your time and a second meeting on a different date and time. During this initial meeting, the researchers will answer any questions or concerns you may have about participating in this study. If you agree to participate, you will be given a letter of consent to sign and return to the researchers. Once you have given consent, you will complete a Physical Activity Readiness Questionnaire and a demographic questionnaire to get a better understanding of your health status. You will also be given a chance to become familiar with the equipment. If you meet the criteria and are willing to continue with the study, you will attend a second meeting, which will take place in room SB 1028, Sanders Building at Lakehead University.

The second meeting will include the testing session and will take approximately 60 minutes of your time. This does not include the time that is needed to get to and from the test facility. This testing session will take place at a different time than the initial meeting. When you arrive to the testing session, you will be introduced to the researchers as well as the testing procedures and equipment. You will then be fitted for three types of canes and a knee range of motion limiting brace. This brace simply straps around the knee of your dominant leg and temporarily limits the amount of motion that is possible at that knee to simulate a knee injury. Your dominant leg will be determined by having you kick a soccer ball. This is done under the assumption that you will kick the ball with your dominant leg. Your dominant leg will be considered as the injured leg. Electrodes, which study the activity of your muscles, will then be attached to several locations on your upper arm. The electrodes simply sit on the surface of your skin. You then will be given some time to practice walking with each type of cane. Once you are comfortable with the equipment, you will walk over the force plate (used to measure ground reaction forces). Each walking trial will be videotaped. A total of 10 trials will be performed for each type of cane. You will be given a perceived ease of use questionnaire after 5 trials and then again at the end of 10 trials. You will be tested individually. You will not be rewarded for taking part in this study.

As with any type of physical activity, there are possible risks. The risks associated with any walking activity include: muscle soreness and muscle fatigue, strains, and sprains. These risks are being minimized by giving you a chance to practice cane use, providing proper instruction, and proper fitting of the equipment.

There will be no direct benefit to you for participating in this study; however, the information gained from this study could be beneficial to the affected populations (e.g., those with hip/knee/ankle injuries and require the use of a cane). The findings of this pilot study may also have implications in expanding this research to further examine the rates of abandonment and disuse of canes with a larger population. The results of this study and future research may provide an avenue for health practitioners in making the correct choice when it comes to cane prescription to minimize negative feelings associated with upper limb pain so that users would be more likely to continue using their devices.

Upon the completion of this research study, only the researchers Dr. Carlos Zerpa and Ms. Aya Mohammed will have access to the data. You will be assigned a code and the data will be kept confidential in a locked cabinet in Dr. Zerpa's office during and after completion of the study. Electronic files will be stored on a password-protected external computer hard drive. These records will be kept for a minimum period of five years. The data will be published in an aggregate form. You will not be identified in published results.

The results of the study will be made available to you via conference presentations, journal publications or upon request. If you have further questions regarding the study, please do not hesitate to contact any of the researchers listed below. If you have any questions related to the ethics of the research and would like to talk to someone other than the researchers, please contact Sue Wright at the Research Ethics Board at (807) 343-8283 or research@lakeheadu.ca.

Yours truly,

Dr. Carlos Zerpa
Email: carlos.zerpa@lakeheadu.ca
Phone: 807-3438940 .

Aya Mohammed
Email: asmohamm@lakeheadu.ca

Appendix D

Letter of Informed Consent

Lakehead UNIVERSITY

School of Kinesiology
e: asmohamm@lakeheadu.ca

By signing this form, I agree that I have read and understood the letter of information for this research study “The Effect of Spring Loaded Single-Tip Support Cane Mechanisms on Upper and Affected Lower Limb Ground Reaction Forces, Muscle Activity, and Self-Perceived Ease of Use.” I also agree to participate in this research study with the understanding of the following conditions:

1. My participation in this research study is voluntary and I may withdraw at any time. I may also choose not to answer any questions presented to me throughout the study.
2. I understand that this study will require 90 minutes of my time.
3. I understand the potential risks and/or benefits of the study and what they are.
4. The information I provide will be securely stored in Dr. Carlos Zerpa's office for a minimum of five years at Lakehead University. If Dr. Carlos Zerpa should leave Lakehead University, the information will continue to be securely stored with the School of Kinesiology at Lakehead University.
5. I will receive copies of any publications in which the research is discussed if I so wish.
6. I will remain anonymous in any publication of the research findings and will not be identified in published results.
7. I understand that my trials will be videotaped.

If you wish to receive a copy of your personal results or information about the results of the study as a whole in aggregate form, please provide your contact information below:

____ I would like to receive a copy of my personal results
____ I would like to receive information regarding the study as a whole

Mailing address:
Email Address:

All questions have been answered to my satisfaction. I understand and agree to the above statements.

____
Name of Participant (please print)

______________________ ______________________
Signature of Participant Date

____
Name of Researcher (please print)

______________________ ______________________
Signature of Researcher Date

______________________ ______________________
Witness Signature Date

Appendix E

Par-Q

Physical Activity Readiness
Questionnaire - PAR-Q
(revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

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

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

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO	
☐	☐	**1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	**2. Do you feel pain in your chest when you do physical activity?**
☐	☐	**3. In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	**4. Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	**5. Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	**6. Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	**7. Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

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

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

NO to all questions

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

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

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

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

Informed Use of the PAR-Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ________________________________

SIGNATURE ________________________________ DATE ________________

SIGNATURE OF PARENT
or GUARDIAN (for participants under the age of majority) ________________________________ WITNESS ________________

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

continued on other side...

Appendix F

Demographic Questionnaire

Demographic Information Questionnaire

(Please note, your information will ***not*** *be sold or given to outside entities. It is for internal use only.)*

1. Name: ____
2. Educational Program Currently Enrolled in:
3. Grade Level:
4. Age:
5. Gender: Female Male
6. Height:
7. Weight:

8. Do you possess any of the following *(Check all that apply)*

____ history of knee trauma or surgery within the past 6 months

____ history of rheumatologic condition

____ injury or amputation of the lower extremity

____ injury or condition of the upper extremity effecting ability to use cane

____ spine/foot/hip pain impeding gait

____ neurological condition impeding gait

____ poor health interfering with gait assessment

Appendix G

ROM Limiting Knee Brace

Appendix H

EMG Electrode Placement

BRACHIORADIALIS PLACEMENT

Type of Placement: Quasi-specific.

Action: Elbow flexion.

Clinical Uses: Rehabilitation.

Muscle Insertions: This muscle arises from the lateral supracondylar ridge of the humerus and the related intermuscular septum and inserts on the tendinous attachments of the styloid process of the wrist.

Innervation: The radial nerve from the posterior cord and spinal nerves C-5 and C-6.

Location: Palpate the muscle mass just distal to the elbow while resisting elbow flexion with the wrist in the neutral position (thumb up). Two active electrodes, 2 cm apart, are placed approximately 4 cm distally from the lateral epicondyle of the elbow on the medial fleshy mass that covers that area, so that they run parallel to the muscle fibers (**Figure 17–35A**).

Behavioral Test: Flex the forearm.

Tracing Comment: In **Figure 17–35B**, three muscle sites are monitored: extensor carpi ulnaris, extensor

Figure 17–35A Electrode placement for the brachioradialis site.

Source: Copyright © Clinical Resources, Inc.

EXTENSOR CARPI RADIALIS (LONGUS AND BREVIS) PLACEMENT

Type of Placement: Quasi-specific.

Action: Wrist extension, abduction, radial deviation.

Clinical Uses: Hand rehabilitation, industrial ergonomics.

Muscle Insertions: The brevis component arises from the common head of the lateral epicondyle of the humerus and related ligaments and inserts on the base of the third metacarpal. The longus component arises from the margin of the humerus and related septum and inserts on the base of the second metacarpal.

Innervation: The radial nerve via the spinal nerves at C-6 and C-7.

Location: Ask the patient to flex the wrist and palpate the muscle mass approximately 5 cm distal from the lateral epicondyle of the elbow, on the dorsal side of the arm just lateral to the brachioradialis. Place two active electrodes 2 cm apart over the muscle mass that emerges, with the electrodes running in the direction of the muscle fibers (**Figure 17–38A**).

Behavioral Test: Wrist extension and radial deviation.

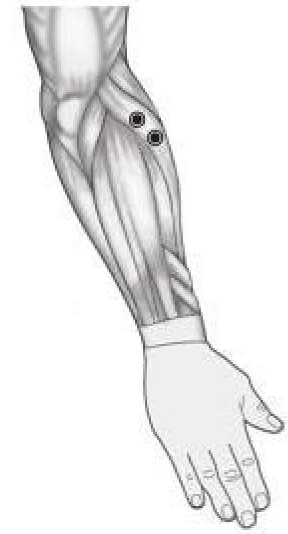


Figure 17–38A Electrode placement for the extensor carpi radialis (longus and brevis) site.

Source: Copyright © Clinical Resources, Inc.

Tracing Comment: In **Figure 17–38B**, SEMG recordings from the extensor carpi radialis, extensor indicis proprius, and abductor pollicis sites while the wrist was supported in the neutral position (thumb up) are presented. Three movements were studied in isolation:

1. During wrist extension with the fingers remaining relaxed, the extensor carpi radialis shows clear isolation.
2. During extension of the first finger with the wrist in the neutral position, a clear burst pattern is seen at the extensor indicis proprius site not involving abductor pollicis. Some activity is also noted in the extensor carpi radialis. It is uncertain as to whether this is a synergy pattern or cross-talk.
3. During thumb and finger extension (spreading the palm), a clear burst pattern is seen from the abductor pollicis site, along with strong activity from extensor indicis proprius. Extensor carpi radialis shows some minor activity (raw SEMG).

Also see the tracings for the brachioradialis (Figure 17–35B), the extensor carpi ulnaris (Figure 17–37B), and the extensor digitorum (Figure 17–39B).

Clinical Considerations: The way in which the wrist is supported may affect SEMG values associated with resting tone. Because finger extension activity is noted at this site (Figure 17–38B), finger position may also affect recording levels. This site is known to play a role in the power grip.[42]

Volume Conduction: Brachioradialis and extensor digitorum.

Other Sites of Interest: Extensor carpi ulnaris and finger extensors during extension; flexor carpi radialis during ulnar deviation; flexor carpi ulnaris during flexion.

Referred Pain Considerations: Trigger points in these muscles refer pain primarily to the lateral epicondyle, lightly over the dorsum of the arm, and the dorsal aspect of the web of the thumb.[17]

pattern of recruitment (raw SEMG).

Also see the tracings for the extensor carpi radialis (Figure 17–38B).

Clinical Considerations: The level of support for the arm, wrist, and fingers can affect the resting tone. Wrist position (deviation) can affect the magnitude of recruitment during finger movement.

FLEXOR CARPI RADIALIS AND PALMARIS LONGUS PLACEMENT

Type of Placement: Quasi-specific.

Action: Wrist flexion and radial deviation.

Clinical Uses: Hand rehabilitation.

Muscle Insertions: The flexor carpi radialis muscle arises from the medial epicondyle of the humerus and related superficial fascia, and inserts on the palmar surface of the base of the second metacarpal. The palmaris longus muscle arises from the medial epicondyle of the humerus and radiates into the palmar aponeurosis.

Innervation: The median nerve via the spinal nerves C-6 and C-7.

Location: Support the arm with the fingers while palpating the ventral aspect of the forearm near the elbow on the medial (little finger) side of the arm. Ask the patient to flex the wrist. Place two active electrodes, 2 cm apart, over that muscle mass so that they run in the direction of the muscle fibers (**Figure 17–40A**).

Behavioral Test: Wrist flexion.

Tracing Comment: In **Figure 17–40B**, the flexor carpi radialis and brachioradialis sites are examined while the elbow is flexed to 90 degrees and the thumb is up. First, the wrist is in radial deviation with slight flexion. A very clear separation of recordings is noted between the two sites, with the flexor carpi radialis showing a strong burst of activity. Next, the elbow flexion is resisted at

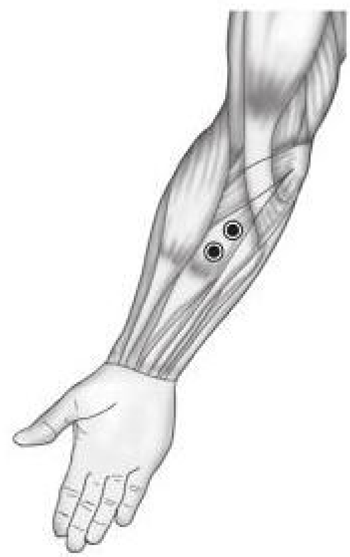


Figure 17–40A Electrode placement for the flexor carpi radialis and palmaris longus site.

Source: Copyright © Clinical Resources, Inc.

Other Sites of Interest: Upper trapezius, middle trapezius, lower trapezius, serratus anterior, biceps, deltoid, infraspinatus, teres major, and pectoralis major during movements of the upper extremities and shoulder girdle.

Referred Pain Considerations: Trigger points at this site refer pain to the middle deltoid region and may include the lateral epicondyle region.[17]

Artifacts: ECG.

INFRASPINATUS PLACEMENT

Type of Placement: Specific.

Action: Lateral rotation of the shoulder joint, along with stabilization of the head of the humerus in the glenoid cavity.

Clinical Uses: Treatment of stroke patients to facilitate use of the upper extremities; treatment of shoulder instability and orthopedic impingement syndromes.

Muscle Insertions: The fibers arise from the infraspinatus fossa, below the spine of the scapula, and insert on the greater tubercle of the humerus.

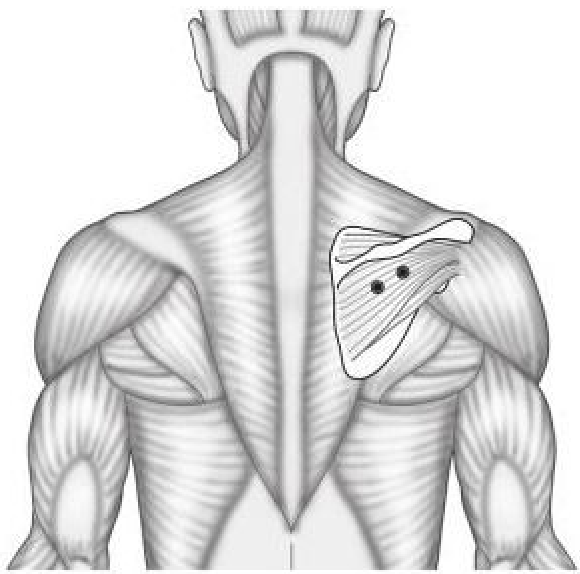
Innervation: The superior cord of the brachial plexus, from the spinal nerves of segments C-4, C-5, and C-6.

Joint Considerations: The joints of the cervical spine related to the muscle (C-4, C-5, and C-6) may affect the SEMG resting or recruitment patterns, along with the glenohumeral joint (particularly the anterior glide stability and the inferior glide capacity) and the acromioclavicular (AC) joint.

Location: Palpate the spine of the scapula. Two closely spaced electrodes (2 cm apart) are placed parallel to and approximately 4 cm below the spine of the scapula, on the lateral aspect, over the infrascapular fossa of the scapula. Avoid placement over the posterior deltoid (**Figure 17–24A**).

Behavioral Test: Elbow bent to 90 degrees with lateral (external) rotation of the bent arm out to the side; abduction of the arm.

Tracing Comment: **Figure 17–24B** shows SEMG tracings from infraspinatus and posterior deltoid during lateral rotation with the right arm flexed at the elbow, and external rotation and extension of the right arm with the elbow extended. During lateral rotation, the activity at the infraspinatus site is greater. During lateral rotation



Pectoralis Major*

Placement

To assess the pectoralis major, the scanning electrodes are placed so that they run parallel to the middle fibers of this rather large and flat muscle group. The electrodes are placed below the clavicle and above the breast and nipple, with the scanning electrodes running almost parallel to the ground.

Interpretation

The primary function of this muscle group is the adduction and medial rotation of the humerus or upper arm. Activation patterns have also been noted in post-myocardial infarction patients who continue to have a fear of a future myocardial infarction.

Triceps*

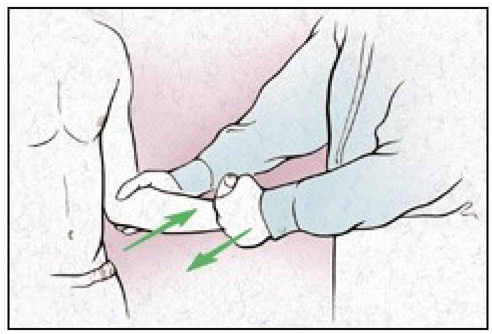
Placement

To assess the triceps, the scanning electrodes are placed on the posterior lateral surface of the upper arm, with the patient's arms hanging to the sides in a standing posture. The scanning electrodes are placed over the belly of the muscle, approximately half the distance between the shoulder and the elbow. The electrodes are oriented in a vertical plane so as to follow the fibers of the muscle.

(Criswell & Cram, 2011)

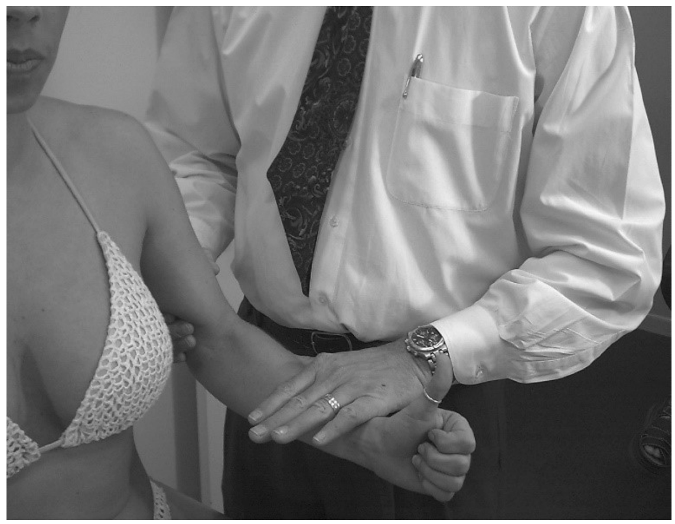
Appendix I

Maximal muscle contraction tests

Muscle	Test being used	Description of test
Infraspinatus	Test of external rotation against resistance in adduction	The main function of the infraspinatus muscle is to externally rotate the humerus and stabilize the shoulder joint. In order to isolate this muscle and allow the participant to perform a maximal contraction, a resisted isometric muscle testing technique will be used and the participant will be instructed to perform the movement maximally. The test of external rotation against resistance in adduction requires the participant to sit in front of the examiner with shoulder adducted, in neutral position, and with the elbow flexed at 90 degrees. The participant is then asked to push in external rotation against resistance.  (Fusco, 2008)
Pectoralis Major	Resisted adduction	The pectoralis major muscle adducts and medially rotates the arm at the shoulder joint. The clavicular head flexes the arm and the sternocostal head extends the flexed arm to side of trunk. To test this muscle, the participant will be in a seated position with the shoulder adducted, in neutral position, and with the elbow at 90 degrees. With one hand, the examiner will stabilize the anterior-inferior acromion and resists against the medial aspect of the forearm. The participant will be instructed to push in internal rotation against resistance.

		 (Hammer, 2007)
Triceps	Elbow extension	The triceps brachii muscle is responsible for extending the forearm at the elbow joint and extending the arm at the shoulder joint. To allow for maximal contraction of this muscle, the participant will be asked to be seated with elbow flexed at 45 degrees and shoulder abducted to 90 degrees. The examiner will then resist against the forearm toward flexion and the participant will be asked to maximally extend their elbow. Testing this motion at 90 degrees of shoulder abduction prevents the inferior-superior compression of the subacromial area.  (Hammer, 2007)
Flexor Carpi Radialis	Wrist flexion	The flexor carpi radialis is responsible for flexion and radial deviation at the wrist joint. To allow for maximal contraction of this muscle, the individual is seated with his/her elbow flexed to 90 degrees and wrist in neutral position. The practitioner is then seated with their fingers placed over the radial side of the individual's hand (thenar

		eminence). The participant will then be instructed to flex their wrist against the practitioner's resistance.  (Hammer, 2007)
Extensor Carpi Radialis longus	Wrist extension	The extensor carpi radialis muscle is responsible for extension and radial deviation of the wrist. The participant will be asked to be seated with elbow at 90 degrees of flexion and forearm placed on a flat surface for support. The practitioner will place resistance against the dorsum of the hand along the second and third metacarpals in the direction of flexion toward ulnar deviation. The participant will be asked to extend and radially deviate the wrist maximally. Resistance against the 2nd metacarpal will also be added in order to further isolate this muscle from brachioradialis (a concern due to proximity). 

		(Hammer, 2007)
Brachioradialis	Elbow flexion	The brachioradialis muscle is responsible for flexion of the forearm at the elbow joint as well as supination and pronation of the forearm at radioulnar joints to neutral position. To test maximal contraction of this muscle, the participant will be asked to be seated with his/her arm in neutral position. The practitioner will then stand beside the individual with his/her hand on the individual's wrist. The participant will then be asked to maximally flex his/her forearm at the elbow joint against the practitioner's resistance.  (Hammer, 2007)

Appendix J

Latin Square Method of Randomization

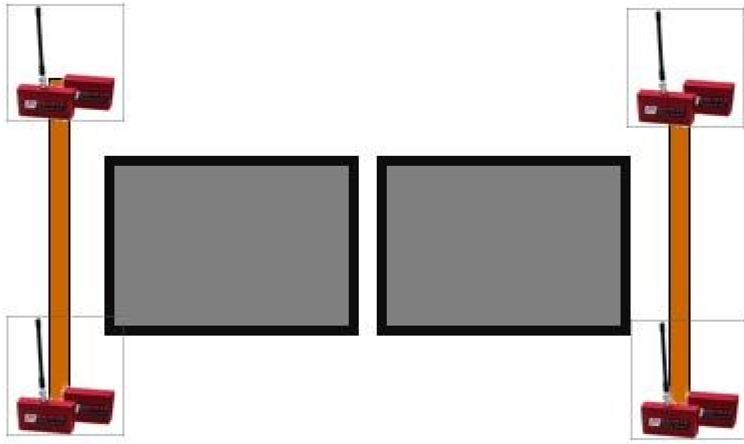
	Traditional Cane	Miracle Cane®	Stander Cane®
Participant 1	1	2	3
Participant 2	2	3	1
Participant 3	3	1	2

The numbers within the Latin Square table represent the order that each cane will be presented to the participants. The number 1 represents the cane that will be tested first. The number 3 represents the cane that will be tested last.

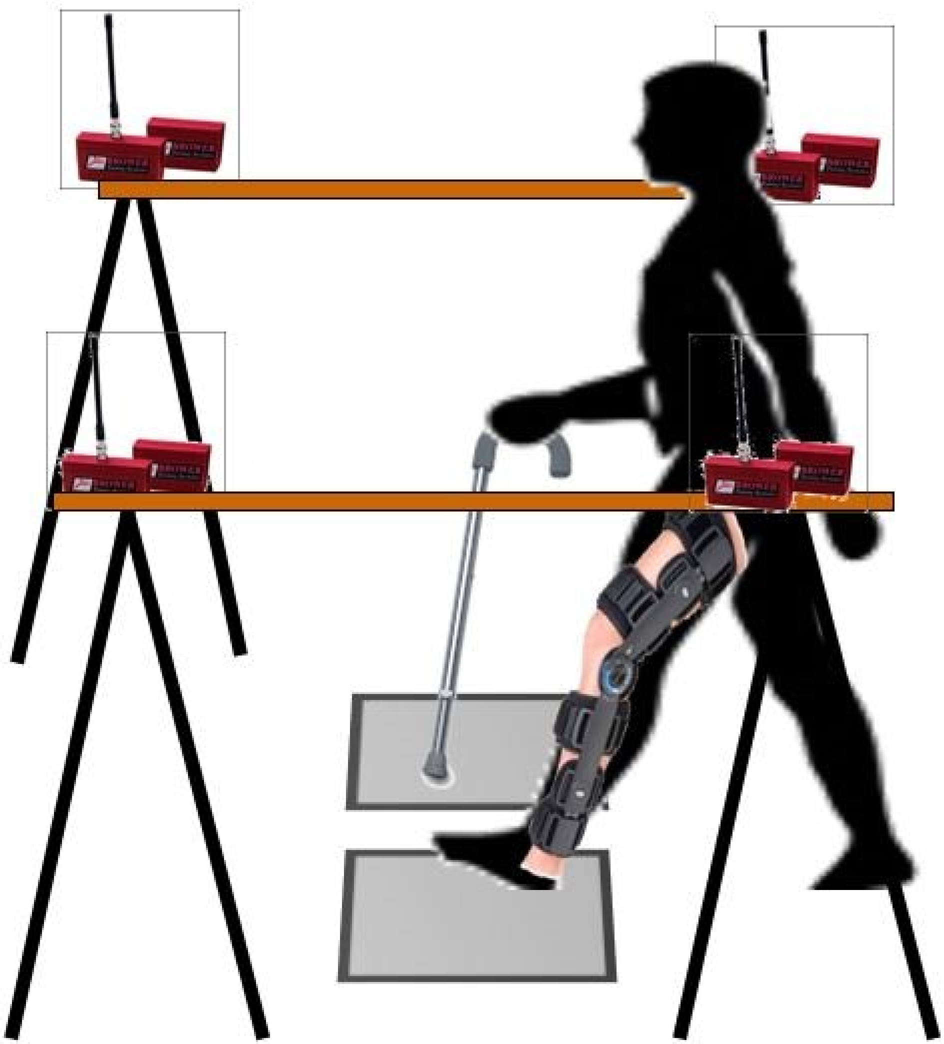
Appendix K

Illustration of Timing Gate Setup

Top-down view



Sagittal View



Appendix L

Self-Perceived Ease of Use Questionnaire

Please indicate your order of preference for each cane by placing a number ranging from 1-3 in the box next to each cane name, with 3 being the cane you preferred the most to 1 being the cane you preferred the least.

Recall,

Traditional Cane — **No Spring-Loading**

Miracle Cane® — **Soft Spring**

Stander Cane® — **Stiff Spring**

Cane Name	Rating
Traditional Cane	
Miracle Cane®	
Stander Cane®	

Please indicate any reasons why you may or may not have liked each type of cane. Feel free to add any additional comments in this section.