HANDLE-HEIGHT ADJUSTMENT EFFECTS ON GAIT KINETICS IN OLDER ADULTS WHILE WALKING WITH A ROLLATOR

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DEDICATION

I dedicate this dissertation to my mother and father who have been my inspiration of strength and perseverance. To my siblings who have been always my biggest critics. Finally, to Noelle, I could not express how much you are part of this, without you this would have been a much harder process.

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ABSTRACT

MARCO ANTONIO AVALOS

EFFECTS OF HANDLE-HEIGHT ON GAIT KINETICS IN OLDER ADULTS WHILE WALKING WITH A ROLLATOR

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An increasing geriatric population has increased the demand for walking-aids among this population. The use of walking-aids requires proper adjustment to maintain adequate shared weight-bearing while maintaining correct posture. Current guidelines leave users with vague instructions not allowing for proper adjustment of these devices. The purpose of this investigation was to determine the effects of handle-height adjustment on gait kinematics and kinetics in able-bodied older adults when walking with an all-terrain rollator. Participants were tested while walking at 80 and 100% of their normal cadence. Thirty-three participants were recruited from the Denton, TX, area. Every participant signed an IRB approved Informed Consent Form. Participants were asked to complete 5 successful trials for each of the different handle adjustments (i.e., 48% of the user's height, 55% of the user's height, and wrist crease height) on two cadences (i.e., normal cadence and slow cadence); the participant was also asked to walk without the rollator as a control condition for each walking cadence condition. Participants were instructed to walk as naturally as possible while applying force over the rollator. Kinetic variables measured include ground reaction force, lower body resultant joint moment, postural back angles and walking velocity were extracted in order to see the effects of the rollator. Significant decrease in vertical, forward, backward, and inward GRF from 11.9-26.5% was seen, and, hip flexor, hip abductor from, and ankle plantar flexor and invertor moments up to 16%

reduction, while an increased L4/L5 flexor moment by 25% were found while using the rollator. These effects were accompanied by a reduction of kyphosis from 1.3 to 1.4% and lordosis up to 1%, but with an increase in thorax flexion by 8-11% and shoulder elevation by 7-10%. The use of a rollator reduced the forces received by the lower limb joints, but a more stooped posture was acquired, increasing the stress over the L4/L5 joint. Rollator adjustment depends on the clinical goals set for the use of the device as well as the user's need for mobility.

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CHAPTER I

INTRODUCTION

Advances in medicine have increased the expected lifespan and the number of older adults from 562 million in 2012 (8% of the total population) to 617 million in 2015 (8.5% of the total population), as reported in the 2015 International Population Reports (He et al., 2016). With a growing number of older adults, caregivers must address factors that may impact their well-being, such as the decline in sensory and perceptual organs due to aging (i.e., sight and vestibular integration), which may affect the ability to maintain posture and balance while walking ((Lacour et al., 2008; Winter, 1995; Winter et al., 1990)). Consequently, the decrease in postural and balance ability makes this population a high risk for falls, decreasing the ability to carry out activities of daily living (ADLs). A fall incident in older adults can be severe enough to require medical attention, creating a fear of future falling events and causing them to adopt sedentary lifestyles (Patel et al., 2014). This sedentarism leads to physical changes, such as osteoporosis and muscle weakness, which affect their strength, endurance, and aerobic capacity. All this creates a vicious cycle of health problems due to lack of activity and inability to perform activities to improve their health, reducing their ability and confidence to walk (Fiatarone Singh, 2004).

According to the Centers for Disease Control (CDC), 14.2% of the population in the US has some type of physical difficulty, of which 73% belong to the geriatric population (Blackwell & Villarroel, 2018). From this group of older adults, 25% have trouble walking a quarter of a mile and require the use of a walking-aids (Gell et al., 2015). Although 25% of the disabled geriatric population requires a walking-aid, only 14.0% have access to these devices (Kaye et al., 2000). Even more, half of the geriatric population requiring walking aids do not use them due to the cost of the device, social factors (i.e., social stigma of disability), and bad experiences due to improper use of a previous device. Every year approximately 800,000 people seek hospital treatment due to injuries caused by falls; these injuries vary from simple bruises to broken bones and concussions. The use of walking-aid can reduce the incidence of falling by improving stability while ambulating.

After these concerning statistics and knowing that physical activity can improve quality of life (QoL) and general health (Fiatarone Singh, 2004; Mandel et al., 2016; Schafer et al., 2018), practitioners should prescribe an easy, practical and accepted activity among this population. Walking is considered an activity of moderate intensity that can improve QoL, social environment, and health (Pollock et al., 1971); this older population can easily perform this activity once they can be warrantied their safety and appease their fear for falls. The introduction of walking-aids while performing this activity may allow users to overcome their physical impairments or at least to reduce the impact of those impairments over their QoL (Bennett et al., 2017). Also, walking-aid may assist ambulation by increasing their base of support (BOS), improving balance and assuring the safety of the user. Therefore, it is imperative to understand the advantages and disadvantages of prescribing different walking-aids that can comply with the needs of the user.

There are several walking-aid devices on the market. Between motorized and non-motorized devices, the latter is the most used due to costs and accessibility (Iezzoni et al., 2010; Kaye et al., 2000). Non-motorized walking aids, such as walking sticks, canes, crutches, and walkers, are devices that allow users to maintain independent mobility. Canes are the most commonly used among the geriatric population, followed by walkers (Bennett et al., 2017; Gell et al., 2015; Iezzoni et al., 2010; Kaye et al., 2000). Canes are usually prescribed for individuals who have moderate mobility but need assistance to maintain their balance. As a single-point-of-contact device, canes have the advantage of being portable and easy to store; nevertheless, this single-point mechanism can become a risk for the user if the device does not properly maintain contact with the floor, resulting in slips.

Walkers are recommended for people with poor balance or lower body weakness. These devices are intended to aid the user by increasing the BOS and allowing shared weight-bearing (Bradley & Hernandez, 2011). Three types of walkers are available in the market: the standard walker, the front-wheeled walker, and the rollator (three- or four-wheeled). The standard walker provides excellent support but limits the gait since the device must be lifted and moved forward every other step. The front-wheeled walker provides a fluent motion, improving gait velocity of the user (H. Choi et al., 2015; Kegelmeyer et al., 2013; Schülein et al., 2017), but the wheels are generally small, and outdoor usage can be limited (Cetin et al., 2010; Government of South Australia, 2013; Härdi et al., 2014). The rollator is a device that provides a more continuous motion since it has 3-4 wheels. Health practitioners recommend the use of rollators to individuals with higher functional capabilities but who are unable to lift the walker every other step. The three-wheeled rollator (a.k.a. the delta rollator) allows good maneuverability in sharp turns and narrow spaces. However, the triangular design of this device can become a risk for falls if one of the wheels gets stopped by an object in its path; swiveling of the front wheel can also easily lead to a loss of balance. In contrast, the four-wheeled rollator, with four-point contact, increases the user’s BOS and gait double support phase, which studies have shown to have an impact on stabilization adaptations that are related to dynamic balance control (Härdi et al., 2014; Schwenk et al., 2011). Due to the squared design, however, it has a disadvantage of poorer maneuverability, mostly in sharp turns (Orrenius & Rose, 2007). Another design of the rollator is the all-terrain rollator, an innovative design that combines the maneuverability and stability of the three-wheeled and four-wheeled rollators, respectively. This was achieved by using only three wheels but increasing the distance between the front and back wheels.

Independent of the use of the walking-aid, a correct posture while walking allows for proper distribution of the weight over the spine and lower limbs (Lacour et al., 2008; Rosário, 2014; Winter, 1995; Woollacott & Shumway-Cook, 2002). The user, therefore, should maintain a correct posture while using a walking device and a key aspect in this is the handle-height adjustment. Most users adjust the handles too high or low to provide comfort while ambulating, disregarding the function of the device. If the handle is positioned too high, the user will not be able to share the weight properly with the device, limiting the functionality of the device. Conversely, if the handle is positioned too low, it will force the user to use a forward bent posture, not allowing proper weight alignment of the upper body over the lower body while ambulating.

Incorrect handle adjustment also occurs since the recommendation for this adjustment is challenging to follow due to a duality of information; one recommendation indicates the adjustment to be at the wrist-crease height and another states that the elbow should be flexed between 15 to 30° while holding the device (Bradley & Hernandez, 2011; Mayo Clinic, 2016; Van Hook et al., 2003). The former is hard to measure without assistance, while the latter is hard to establish without the use of a goniometer. Nevertheless, users tend to focus more on the parameter of elbow flexion, which most times is only estimated. Moreover, some webpages include incorrect graphical explanation of how to measure this angle (Preferred Health Choice, n.d.), creating confusion for the user. The schematics in these webpages shows the angle measured regarding the longitudinal axis of the device handle and not regarding the user's upper arm as it should be measured (Mayo Foundation for Medical Education and Research, 2014); creating an elbow flexion above 30°, which is considered a high handle adjustment. High handle adjustments do not allow for proper shared weight-bearing from the user to the device.

Handle-height adjustment affects the distance between the body and the device, which may alter gait spatiotemporal parameters and the position of the user-device center of mass (COM). Regarding spatiotemporal parameters, the distance between the user and the walking-aid may alter walking step length and walking posture. Changes in step length and/or step frequency modify walking velocity; therefore, changes in step length due to the rollator combined with the drag produced by the device tend to decrease walking velocity (H. Choi et al., 2015; Levangie et al., 1989). Walking posture is dependent of the user-rollator interaction. Too high or too low handle-height may introduce more distance between the user and the device provoking a stooped posture from the user. A proper adjustment would allow the user to have an adequate distance to the device, allowing for proper posture while ambulating. A study focused on gait kinematics reported that adjusting the handles to 55% of the user's height as well as increased walking velocity and stride length with improved posture (H. Choi et al., 2015).

Walking velocity affects the force exerted by a person while ambulating. Force is a relation of mass times acceleration, and since acceleration is a derivative of velocity over time, changes in the latter will affect force generation (Hall, 2014b). Reduction in force is essential to reduce joint moments over the lower limbs. Nevertheless, reduction of walking velocity can increase the mediolateral displacement of the COM (Orendurff et al., 2004), which in older adults can be detrimental to their balance. Therefore, the only way to reduce the force over the lower limb joints is to reduce the mass loaded over them. Walking-aids allow for the reduction of forces over the lower limbs due to a shared weight-bearing between the user and the device. To be able to perform a proper shared weight-bearing over the device, the user's arms should be aligned with the COM. A mathematical modeling study considered how handle adjustment affected muscle load while using a four-wheeled rollator and concluded that adjusting the handles to a height of 48% of the user's height allows for a reduction in muscular load on the lower limbs compared to a higher handle-height (Takanokura, 2010). Nevertheless, this adjustment demanded an increase load in the upper extremities and a stooped posture while walking. Additional studies performed on these devices have manifested that rollators assist with the shared weight-bearing of the user by increasing the BOS and reducing the GRF while ambulating (Alkjaer et al., 2006; Eggermont et al., 2006).

As presented, appropriate handle-height adjustment has demonstrated advantages, but no research has presented a complete understanding of gait parameters associated with various adjustments. The wrist-crease adjustment is a recommendation extrapolated from other walking-aids (i.e., cane and crutches), but does not have strong evidence in rollator users. The 55% height adjustment shows an appropriate comfortable posture of the user but does not allow for adequate shared weight-bearing with the device. Finally, the 48% height adjustment allows for good shared weight-bearing, but consequently leads to a stooped posture, as seen through mathematical modeling.

To clearly understand the force effects of the rollator on gait, either the mass or acceleration component of the force must be controlled. User's mass is a constant factor that cannot be changed without the use of a device, which in this case would be the rollator. In relation to acceleration, while walking, gravitational and horizontal acceleration affect the generation of force in the participant; and although the gravitational acceleration cannot be altered, the horizontal acceleration can be altered through changes in walking velocity components. Walking velocity is the product of cadence (number of steps given in a lapse of time) and step length (Hall, 2014a; Winter, 1980). Altering step length would require the participants to focus on the foot distance placement, which would alter the normal gait pattern. However, cadence is a factor that can be altered through the change of rhythm or with the use of a metronome (Buhmann et al., 2017; Van Dyck et al., 2015), without the need for the participants to focus on their stepping patterns. This way, using controlled cadences in this study will allow for the isolation of the mechanical effects of the use of a rollator with different handle adjustments.

Purpose of the Study

The purpose of the study was to investigate the effects of rollator handle-height on walking velocity, posture (i.e., spinal posture angles, elbow flexion), and gait kinetics (i.e., GRF, lower body RJM) in two different cadence conditions in able-bodied older adults.

Research Hypotheses

It was hypothesized that:

1. The use of a rollator and the rollator handle-height would affect the walking velocity.
2. The use of a rollator and the rollator handle-height would affect the spinal posture angles.
3. The use of a rollator and the rollator handle-height would affect the lower extremity kinetics (i.e., normalized GRF and RJM).

Significance of the Study

Despite the recommendations for the proper adjustment of the rollator, no clear evidence exists to support their application, which is why many users interpret them only as guidelines. For this reason, it is essential to acquire evidence about proper device adjustments. It is not only important to understand which handle adjustment is beneficial for the user in matter of force reduction, but also to determine if the user can maintain correct posture while ambulating with the device. Previous studies have described the benefits of rollator adjustments on gait kinematics, postural effects, and minimally on kinetics, but no study has provided a complete description of the changes produced on joint forces while describing correct posture by the user with handle adjustment. Additionally, this study intended to examine if these changes are maintained across various cadences. Findings from this investigation can provide health practitioners with evidence and clear expectations of the benefits of using these devices.

Assumptions

1. The health information provided by the participants was accurate.
2. Participants were free of any pathology that could affect their normal gait patterns at the time of data collection.
3. Bilateral gait patterns were symmetric.
4. Each body segment was rigid, and the mass and moments of inertia remained constant throughout the entire motion.
5. The body segments were linked through frictionless pin joints.
6. Participants walked at a consistent cadence and step length through the capture area, allowing the researcher to analyze two steps (i.e., one right and one left) as representative of their gait pattern.

7. Participants demonstrated maximal effort of shared weight-bearing over the device.

Delimitations

1. Participants were considered able-bodied if they could walk without any type of walking aid for at least 100 m without requiring rest.
2. Normal cadence was established for each participant and slow cadence was 80% of the normal cadence.
3. Participants were offered rest periods between trials and between conditions in order to prevent fatigue.

Limitations

1. Due to the size of the force plates, the participant often shared two force plates for one step. Because of this, sometimes the front tire of the rollator did not clear the force plate before the participant stepped on to it.
2. The distance between the front tire and the lead leg, created an interference with the detection of the heel contact (HC).
3. Participants were observed during data collection, which may have led to alterations in their normal gait pattern (Hawthorne effect).

Definition of Terms

Cadence: number of steps per minute.

Center of mass: the balance point of a body; this point represents the location at which gravity is assumed to act.

Gait cycle: the period between the foot strike of one leg and the following ipsilateral foot strike.

Ground reaction force (GRF): the force exerted by the floor in reaction to the force applied to the ground.

Inverse dynamics: indirect method to determine the resultant joint forces and moments based on the kinematic, inertial, and external force/moment data of the joints and segments.

Kinematics: the branch of mechanics dealing with description of motion. Positions, velocities, and accelerations are the key variables in kinematics.

Kinetics: the branch of mechanics dealing with forces and moments. The key aspect is explanation of motion, focusing on the causes.

Rollator: a modern type of walker equipped with 3-4 wheels that allow continuous ambulation.

Resultant joint moment (RJM): sum of the moments generated by the muscles around a joint about the joint center.

Stance phase: the period of the gait cycle in which the foot is in contact with the ground. In a normal gait cycle, this corresponds to approximately 60% of the cycle time. The stance is further broken down to two sub-phases, the deceleration phase and the acceleration phase.

Step-length: the distance between the two consecutive foot strike positions on the ground measured in the direction of walk.

Swing phase: the period of the gait cycle in which the foot is not in contact with the ground. In a normal gait cycle, this corresponds to approximately 40% of the cycle time.

CHAPTER II

REVIEW OF LITERATURE

This section has been divided into two main areas. (1) Older adult population and gait, where the demographics of older adults and the need for walking aids, factors of gait and posture, and aging effects related to gait will be explored. (2) Walking-aids, where the types of walking-aids and the recommendations for handle-adjustment will be presented. Finally, a conclusion and importance of this study will be presented.

For this chapter, potentially relevant articles were sought through Google Scholar and papers identified from cited references using a combination of the keywords 'older adult', walking-aid', 'rollator', 'elderly', 'posture', 'walking', and 'gait'. All articles abstracts were reviewed and included if this review if they referenced older adults using walking-aids and the biomechanical effects of using them. Inclusion criteria for the articles were: (1) articles was written in English, (2) the article contained information about non-pathological older adults, (3) the study contained information about walking biomechanical changes due to aging, (4) the article contained information about walking-aids, (5) the article analyzed postural changes while walking with a walking-aid, (6) the article analyzed kinetics of walking, (7) the article contained specific information about walkers and/or rollators, and (8) the article contained information about handle adjustment for walking-aids. Article were excluded if did not reported means and standard deviations of variables of interest (i.e., postural variables, kinematics, walking velocity). A total of 63 articles comply with these criteria and were included in this review.

Older Adult Population and Gait

Demography

An increasing older adult population has been seen in the past decade, with baby boomers being a significant contribution to this group. It is expected that by the year 2020, most of this group will be retired (Jones, 1980). Even more so, by the year 2050, the older population in the United States (US) is expected to increase to 88 million older adults from 48 million in 2015 (He et al., 2016). Advances in the medical field have increased the expected lifespan and thus have increased the number of older adults worldwide, from 562 million in 2012 (8% of the total population) to 617 million in 2015 (8.5% of the total population). This tendency is expected to continue, with an expected triplication of the current number of the older population by 2050 (He et al., 2016).

Data from the Centers for Disease Control (CDC) shows that 14.2% of the population in the US have some type of physical difficulty, 73% of which belong to the geriatric population. These disabilities range from physical impairments to physiological limitations. From this group of older adults, 25% have difficulty walking a quarter of a mile and require the use of a walking-aid (Blackwell & Villarroel, 2018; Gell et al., 2015); this can be due to neuromuscular disabilities (e.g., Parkinson, hip dislocation, etc.) or physical impairments (i.e., chronic obstructive pulmonary disease). Although 25% of the disabled geriatric population needs walking aids, only 14% have access to these devices (Kaye et al., 2000). Approximately half of the geriatric population requiring walking aids do not use them due to bad experiences, resulting from their improper use or the inappropriate prescription of a previous device. Other reasons for not using these walking aids involve social stigmas of not wanting to be labeled as disabled or as an old person who needs help; additionally, the cost of the devices and insurance coverage limitations contribute to their lack of use.

Every year, approximately 800,000 people seek hospital treatment due to injuries caused by falls; these injuries vary from simple bruises to broken bones and concussions. Some risk factors of these falls include impaired strength and balance, which can be improved with the use of an assistive device (Bennett et al., 2017). The annual cost of medical attention due to fall-related injuries is estimated to be more than $19 billion in the US (Gell et al., 2015; Sleet et al., 2008), an amount which the insurance system may not be equipped to handle. For these reasons, older adults must be provided with proper instruments to maintain ambulation without a high risk of falling.

Gait

Gait is the process of ambulation in a living animal, which requires a sequence of motions that allow maintenance of balance (Winter, 1995). Gait is carried out with one and two-point support in humans, as bipeds must alternate the position of their lower limbs to be able to mobilize from one place to another. While walking, a person alternates between single support, while one leg swings to the front, and double support when the two legs are on the ground. This alternating of single to double-support creates a change in a person's BOS. The BOS is the area formed by the outline of the parts in contact with the supporting surface (Hall, 2014b). To be able to maintain balance while ambulating, a person's COM must be kept within the boundaries of the BOS; this way, when a person is in a single support phase, the trunk has to be shifted onto the supporting leg in order for the COM to be aligned within the area of the BOS (MacKinnon & Winter, 1993; Winter, 1995, 2009). This change of attitude of the trunk about the BOS generates angular forces (torque) over the torso and lower limbs to allow for continuous ambulation (Hall, 2014b; Pollock et al., 1971).

Gait can be assessed by a series of parameters. Gait characteristics like step length, cadence, and walking velocity are known as spatiotemporal parameters. Other aspects that describe body segment positions or orientations (e.g., joint angle, and displacement) are known as *kinematic variables*, while variables related to forces generated and/or applied to the body (e.g., ground reaction force [GRF], and resultant joint moment [RJM]) during the motion are known as *kinetic variables*. All these variables have patterns that have been established according to age, gender, and even to specific disabilities (Ferber et al., 2003; S.-U. Ko et al., 2009; Whittle, 2007a; Winter et al., 1990). Changes that occur due to aging will be explained further on in this chapter.

Posture

While ambulating, people must support the weight of their body with their lower limbs; this creates forces on the joints of the torso and lower body. To be able to maintain a proper distribution of forces throughout the spine and over the lower limbs, correct posture while walking is needed (Government of South Australia, 2013; Rosário, 2014; Winter et al., 1990). Proper posture, while ambulating has been defined as an erect torso over the pelvis, parallel to the vertical axis of the global frame (Whittle, 2007b; Winter, 1995). Incorrect posture has been linked to pain and dysfunction of the spine and lower limb joints (Rosário, 2014). Posture and body positioning are continuously updated to adapt to multisensorial feedback within the body (Lacour et al., 2008). An inadequate distribution of forces, due to a disability (i.e., leg length discrepancy), generates inappropriate feedback that provokes incorrect posture while ambulating; consequently, the person develops pain in other regions (i.e., back pain), which becomes a vicious cycle of inappropriate feedback and bad posture (MacKinnon & Winter, 1993; Winter, 1995).

Aging

Aging is a process that has an adverse effect on sensory and perceptual organs (i.e., sight and vestibular integration) requiring more cognitive demand to maintain correct posture and balance while walking (Lacour et al., 2008; Li & Lindenberger, 2002; Nigam et al., 2012; Winter, 1995; Winter et al., 1990). Other physiological and physical changes that occur in older adults include a decrease in muscle and bone mass (Fiatarone Singh, 2004). These changes are natural and should not impair functionality in older adults who maintain a “healthy and active” lifestyle. In the absence of an active lifestyle, there is a decline in the ability to maintain balance while walking, making it difficult for older adults to carry out ADLs due to the disuse of proprioception and muscular systems. The combination of disuse of systems and typical aging factors result in a loss of homeostatic balance, leading to what has been termed the ‘syndrome of disuse and aging’ (Bortz, 1989). Every year at least one-third of older adults suffer a fall with the need for medical attention in emergency services (Lacour et al., 2008). As the geriatric population has the highest risk for falls, due to tripping or physical limitations (Lockhart et al., 2003), and that falling is the leading cause of death among this population group (Sattin, 1992), some older adults develop what is called ‘post-fall syndrome.’ Murphy and Isaacs (1982) have described the post-fall syndrome as a change in lifestyle, a decrease in mobility, and an increase in anxiety as a result of disturbed balance and falls.

Due to the aforementioned factors, most of the elderly population have adopted a sedentary lifestyle to avoid any possible injuries. This sedentarism leads to physical changes that affect health, strength, endurance, and aerobic capacity (Fiatarone Singh, 2004), becoming detrimental to the ability and confidence to walk. The decline in physical activity creates a cascade of physical and emotional deterioration in this population. It is through a person's mobility that social life, the ability to perform ADLs, and QoL is maintained in older adults (Hoang et al., 2014; Hortobagyi et al., 2003; Roman de Mettelinge & Cambier, 2015). Therefore, this population should maintain mobility to preserve QoL, and those who have declined in this activity should be encouraged to maintain it with the assistance of a walking aid.

Additionally, Pollock et al. (1971) have manifested that walking is considered an activity of moderate-intensity or even high intensity at speeds of more than four mph. Fiatarone Singh (2004) expressed that moderate-intensity activities, such as walking, can improve QoL, social environment, and health in older adults. Furthermore, maintaining ambulation has proven to reduce the progression of functional limitations and disabilities, and to be an effective antidepressant (Blumenthal et al., 1999). Therefore, although it is essential to prescribe walking time as an exercise to this population, it is necessary first to ensure their safety while performing this activity.

Gait Changes in Older Adults

Several studies have determined changes in gait patterns due to aging in older adults. One of the aspects clearly seen and demonstrated is that of the reduction of step length in order to have better control of COM while the body is supported by only one leg (single support phase), thus increasing double support time and generating a feeling of safety (Lockhart et al., 2003; Winter et al., 1990). The reduction in step length is also a consequence of decreased muscle strength, causing the legs to be unable to generate active push-off of the foot (Winter et al., 1990). Furthermore, shorter step length has been associated with lower foot clearance and a higher risk for tripping (Pavol et al., 1999).

Cadence, which is the number of steps taken during a specified time period, is a characteristic that changes during aging. Since walking velocity is the result of the product of cadence and step length, some older adults will try to increase cadence to compensate for the natural reduction in step length to maintain walking velocity; the problem here is that walking with increased velocity becomes unstable due to muscle weakness. An alternative approach is to maintain or reduce the cadence in order to control ambulation and avoid falls (Frimenko et al., 2015; S.-U. Ko et al., 2011; Lockhart et al., 2003; White & Lage, 1993), with the latter being the most common change in this population.

Postural control while ambulating is an integration of visual, vestibular, and proprioceptive system feedback (Lacour et al., 2008), which becomes an automatically learned aspect on which most of the able-bodied population does not focus its attention on during ambulation; but it is because of a functionality decline of these systems while aging that posture becomes the source of attentional demand in this older adult population (Li & Lindenberger, 2002). It is common to see older adults assume a stooped posture due to muscle weakness. Therefore, older adults should maintain an active lifestyle to avoid a decline in muscle strength and, consequently, improper posture, which does not allow for proper distribution of forces along the spine and lower limbs.

The RJM is the total rotational effect generated by all the internal structures (i.e., muscles, bones, ligaments) between two segments (e.g., upper arm and forearm) and external forces (i.e., GRF) producing moments of force (torques) across a joint (i.e., elbow). Changes in walking velocity, the presence of the syndrome of disuse, and changes in posture provoke alterations in the body and body segment positional attitudes affecting the way RJMs are generated in the joints of the lower limbs. Muscles must be able to support the body and maintain balance with proper posture (Messier et al., 2005; Whittle, 2007b; Winter, 1980) in the presence of the aforementioned factors. Otherwise, in the long term, this could cause joint stress, disability, and pain. To be able to reduce this dysfunctionality, assistive devices can be used to help the ambulation of this population.

Walking-Aids

A walking aid is a device created to assist a person with ambulation. The use of assistant walking devices goes back to the Egyptian era, where they were not only used as an assistive device but also as an authority statement (Loebl & Nunn, 1997). There are several walking aid devices on the market; these devices can be categorized into two main groups: motorized devices (i.e., motorized scooters and wheelchairs) and nonmotorized devices (e.g., canes, walkers, etc.). Between motorized and non-motorized devices, the latter is the most used due to costs and accessibility (Iezzoni et al., 2010; Kaye et al., 2000). Nonmotorized walking aids, such as walking sticks, canes, crutches, and walkers, are devices that allow users to maintain independent mobility. Canes are the most commonly used among the geriatric population followed by walkers (Bennett et al., 2017; Gell et al., 2015; Iezzoni et al., 2010; Kaye et al., 2000). This section will present the advantages and disadvantages of the most common prescribed walking-aids

Canes

Canes are usually prescribed for individuals who have reduced mobility but need assistance to maintain their balance. The main benefit for users is to widen their BOS, thus increasing stability during single support while walking (Kuan et al., 1999). Even when a cane is not recommended for shared weight-bearing, it has been reported that a cane can support a quarter of a person's body weight (BW) when used appropriately (Faruqui & Jaeblon, 2010), but other studies mention that they can only support 12.7 - 16% of the BW (Laufer, 2003). As a single-point-of-contact device, canes have the advantage of being portable and easy to store. Furthermore, canes have high acceptance in the general population due to fashion statements that remount back to the Egyptian era (Gell et al., 2015). A standard cane is very inexpensive and lightweight, which makes it easy to transport and handle. These canes can be made of wood (which must be custom fitted) or aluminum (which can be adjusted). This device must be fitted appropriately or adjusted for the user to avoid improper postural effects. There are different cane designs, but most of them have a single-point mechanism, which can become a risk factor for the user if the device does not properly contact the floor, resulting in slips.

The quadripod-cane has an adaptation of the points of contact in comparison to that of the standard cane, thereby increasing the BOS of the device and preventing slipping due to a single-point of contact. The quadripod-cane may also allow for more shared weight-bearing (Laufer, 2003). The adjustment of this device is easily carried out by the user and does not require customization. The problem with these devices is that all legs must be in contact with the floor, which can be cumbersome for users (Van Hook et al., 2003). Additionally, in opposition to what Laufer (2003) presented, Dickstein et al. (1993) reported that the quadripod device does not allow for more BW unloading than the standard cane.

The use of a cane with the hand ipsilateral to the involved injured limb allowed for a reduction of the users' mediolateral and anteroposterior sway, while using a cane in the hand contralateral to the involved limb allowed for the reduction of hip abductor moments (Faruqui & Jaeblon, 2010). Controversial aspects of the use of canes in rehabilitation have been presented by investigators, assuring that these devices may increase gait asymmetry by increasing single support of the non-involved limb while decreasing it on the involved limb (Laufer, 2003).

Crutches

Crutches are like canes with the difference that they allow for more shared weight-bearing since the area of the device interacting with the user is larger. Two types of crutches are available; the first being the axillary crutch that allows the user to unload from 80% (single crutch) to 100% (double crutch) of BW (Faruqui & Jaeblon, 2010), which allows for faster walking than when using walkers (Cetin et al., 2010). However, this device requires significant energy expenditure and upper limb strength (Bradley & Hernandez, 2011). Also, this type of crutch can create compression of the axillary nerve plexus; thus, it is only recommended for short-term assistance. The second type of crutch is the forearm crutch, which has a cuff that goes around the proximal third of the forearm. This device allows users to unload up to 50% of their BW and, in comparison to all other devices, these crutches are the simplest to use when climbing stairs (Faruqui & Jaeblon, 2010). Forearm crutches are recommended for participants needing long-term bilateral support. This device is convenient for users because of the cuff around the forearm, which allows the hands to be free while the device is not in contact with the floor. The adjustment of crutches depends on the type of crutch, and this will be discussed further on in this chapter.

Walkers

Walkers are devices that improve balance when ambulating by increasing BOS and increasing the user's lateral stability (Faruqui & Jaeblon, 2010). These devices are meant to support more shared weight-bearing while providing more comfort to the user in comparison to crutches (Bradley & Hernandez, 2011). These devices are recommended for use by individuals having poor balance or lower body weakness, but these devices are difficult to maneuver and are not suited for climbing stairs. There are three types of walkers: the standard walker, the front-wheeled walker, and the rollator (three or four-wheeled walkers). All these devices limit the user's arm swing and if not properly adjusted can lead to an abnormal flexion of the back.

The standard walker provides good balance assistance and is the most stable device for supporting the user's weight (Faruqui & Jaeblon, 2010). The walker had been shown to control the lateral sway of the trunk in patients with hemiplegia, allowing them to have a more symmetrical gait (Tyson, 1999). The problem with this device is that it limits the user's gait velocity, since it must be lifted and moved forward every other step, and the user must make sure that all four legs are in contact with the floor before taking another step. A consequence of this is the need for proper coordination between walking and moving the device, as well as the demand for a high level of attentional focus when walking while performing double-task activities. This can become cumbersome for the user and creates a high risk for falls (Cetin et al., 2010; Wright & Kemp, 1992).

The front-wheeled walker (FWW) has two small wheels attached to the front legs of the device, which provide a faster and more fluid motion, improving a user's gait. This device is recommended for users that cannot lift a standard walker or require a more fluid movement while performing ADLs (Van Hook et al., 2003). Studies have shown that using a FWW requires less energy use than that required when using a standard walker (Cetin et al., 2010; H. Choi et al., 2015), but the incorporation of the two front wheels makes this device less stable than the standard walker. More so, the wheels on this device are small and limit outdoor usage, thus restricting the user to specific areas (Cetin et al., 2010; Government of South Australia, 2013; Härdi et al., 2014).

Rollators

Rollators are three- or four-wheeled walking aids that allow a more continuous motion, which may decrease attentional demand by enhancing control over the device during locomotion (Miyasike-daSilva et al., 2013). Rollators are recommended for individuals having higher functional capabilities and do not require shared weight-bearing. Studies have shown that users can unload around 36% of their BW when using a rollator. However, if a large amount of force is applied, it may cause the device to tip or rollover, increasing the risk for falls or injuries (Alkjaer et al., 2006; Van Hook et al., 2003; Youdas et al., 2005). The rollator is the only walking-aid that complies with the three functional safety requirement standards for elderly users established by the Consumer Product Safety Association in Japan, which are: (1) a walker should assist with walking, allowing partial weight-bearing by the user; (2) the walker should have a space to place the user's belongings; and (3) the device should provide users with the possibility of resting and allow them to sit when necessary. The rollator is the only device that provides a basket in which to put belongings and offers a seat to allow the user to rest if needed (Takanokura, 2010). Additionally, these rollators have an integrated locking brake system that allows users to be able to sit safely without the device rolling under them (Miyasike-daSilva et al., 2013).

Between the two types of rollators mentioned above, the three-wheeled rollator, better known as the delta rollator, allows good maneuverability in sharp turns and narrow spaces. The triangular design of this device, however, can become a risk for falls if one of the wheels gets stopped by an object in its path, especially if the front wheel swivels, leading to a loss of balance. In contrast to the three-wheeled rollator, the four-wheeled rollator, having a four-point contact, improves balance by increasing the user's base of support. Due to its rectangular design, the four-point rollator has a disadvantage of poor maneuverability, mostly on sharp turns (Orrenius & Rose, 2007). In a study that investigated user satisfaction with rollators, Brandt et al. manifested that handling this device was difficult, especially in the case of female users (Brandt et al., 2003). An innovative rollator is the all-terrain rollator which combines the maneuverability and stability of the two previous rollators. This has been achieved using a three-point rollator with an increased distance between the front and back wheels, displacing the COM of the device further away from the handles than on traditional three or four-point rollators (Orrenius & Rose, 2007). The all-terrain rollator has been incorporated with 12 to 14-in. pneumatic wheels (normal rollators have up to 9-in. wheels), which allows it to cruise over obstacles with less force exerted by the user.

Based on all its characteristics and properties, the rollator has become popular in Europe and is increasing its popularity in the US; however, little research about the biomechanics of this type of rollator has been done. Most of the research on this type of rollator has been focused on user satisfaction or has been carried out with a small number of subjects. Among the studies done on this device, few have explored the biomechanical effects of its use; Alkjaer et al. (2006) explored the force effects of the use of these rollators and found that these devices may reduce plantar flexor, knee extensor, and hip abductor joint moments compared to walking without any device. Orrenius and Rose (2007) found a reduction of activity of the trapezius when using the all-terrain rollator compared to when using a four-point rollator. Schwenk et al. (2011) described that the use of rollators may limit the ability to detect mobility deficit assessments using spatiotemporal gait parameters. Finally, Takanokura (2010) introduced what he called a critical height adjustment (48% of the user’s height) to reduce muscular load while walking up or down a steep hill, while H. Choi et al. (2015) found that better spatiotemporal parameters where obtained when the handle-height was adjusted to 55% of the user's height. All these studies have agreed that a correct adjustment of the device is needed for the user to benefit from the use of a rollator.

Handle-height Adjustment

Handle-height adjustment of walking-aids has been described, but little evidence-based information on the validity of these measurements has been presented. Most of the studies have been done on canes and crutches, and only a few articles have explored the biomechanical effects of handle adjustment on rollator devices. An Incorrect adjustment of a cane can either generate improper posture for the user, making him/her lean forward in the case that the device is too long, or cause pain in the lumbosacral region in case that the device is too short (Van Hook et al., 2003). On crutches, an improper adjustment can lead to axillary nerve plexus compression or stooped posture with lower back pain if the crutch is too high or too low, respectively (Bauer et al., 1991). Incorrect adjustment of a rollator handle-height may generate improper posture, which can lead to pain in the long term. Either the participant acquires a stooped posture (low adjustment) to be able to share weight-bearing with the device or pushes the device forward (high adjustment), not allowing proper shared weight-bearing with the device, thus disregarding its purpose.

Three measurements for rollator handle-height adjustment have been presented in the literature, two related to user's total height (48 and 55% of the user's height) and one relating to the attitude of the elbow flexion while using the walking aid. From the few articles concerning handle-height, no study has compared all three adjustments or has indicated a complete description of gait parameters; they have either focused on kinetic or kinematic parameters. The recommendation from physical therapists is that while using a walking-aid the elbow should be flexed between 15 to 30° in relation to the longitudinal axis of the torso. To acquire this position, a measurement of the height from the floor to the ulnar styloid process is the most recommended method (Faruqui & Jaeblon, 2010; Government of South Australia, 2013; Laufer, 2003; Mayo Foundation for Medical Education and Research, 2014; Van Hook et al., 2003). Although, to the author's knowledge, this is the most recommended handle adjustment, there is no evidence of the biomechanical benefits of this adjustment on rollators. As to the other two measurements, Takanokura (2010) manifested that adjusting the handle-height at 48% of the user's height allows for a reduction in the muscular load calculated through mathematical modeling. H. Choi et al. (2015) compared two height adjustments (48 and 55% of the user's height) and found that at 48% the user acquires a stooped posture, but the user has a better sense of security, while at 55% the user has a better posture and larger step length, but a sense of insecurity and fear of falling is created. No kinetic parameters were tested in the study.

Summary

The growing numbers in the older adult population is accompanied by an increase of ambulatory limitations in the general population, creating a burden on the healthcare system. Falls constitute the main reason for older adults to attend to emergency rooms and annually the government spends millions of dollars in expenses on preventable events. Being that older adults have the highest risk for falling, which is increased by fear of falling created by previous fall events or reduced musculoskeletal mass due to aging, it is imperative to ensure security while ambulating in this population. Walking is a medium intensity activity well accepted among older adults, which can improve their overall health. Improving ambulatory conditions for this population will allow them to improve their health, muscle strength, and consequently quality of life. Walking-aids can assist ambulation by increasing the base of support of the person and providing increased stability. Between the many walking-aids, rollators have acquired popularity due to the design of the wheels, allowing a more continuous motion, a seat, allowing the user to rest when needed, and a basket, allowing the user to unload belongings onto the device. However, to be able to serve their purpose, a walking-aid must be properly adjusted. The lack of complete information about the device's adjustment has led to improper user interaction with the device, leading to bad posture while walking or the inability to properly share weight-bearing over the device. The results of this investigation may provide information to clinicians regarding the benefits and proper adjustment of rollators, with the intention of proving better adjustment to accommodate the user's needs to increase their ambulation.

CHAPTER III

METHOD

This chapter is divided into the following sections: participants, trial conditions, experimental setup, data reduction and processing, data analysis, and statistical analysis.

Participants

A total of 33 able-bodied older adults (16 females and 17 males) participated in this study (see Table 1). Participants were recruited from the Denton area using emails and flyers posted in community centers. Participants complied with the following requirements: (a) between 60 and 80 years old, (b) height between 160 and 189 cm, (c) ambulatory without the need of any walking aid, (d) not having had any surgical procedures 6 months prior to the study, (e) not having any orthopedic or health problems that would have altered their gait, and (f) being comfortable wearing spandex clothing. Any participant registering below 75 cm or above 104 cm in the handle-height adjustment was excluded from the study. A power analysis was conducted using G-Power software (Faul et al., 2007) and a minimum sample size of 34 participants (effect size f = .3067, significance level α = .025, power = .81) was obtained using the effect size from a previous study with young adults.

Prior to initiating participation in the study, every participant was explained the study protocol and notified of the purpose of the study. Every participant had the chance to ask questions before and during data collection; after that, participants were requested to consent on their participation and asked to sign an informed consent form that was approved by Texas Woman's University Institutional Review Board (see Appendix BA and Appendix B).

Table 1

Demographics and anthropometrics of female, male and combined participants

	Female (*n* = 16)	**Male (*n* = 17)**	**Combined (*n* = 33)**
Age (years)	66.9 ± 5.7	74.2 ± 6.2	68.6 ± 6.1
Mass (kg)	65.0 ± 10.1	82.4 ± 14.4	74.0 ± 15.1
Height (cm)	167.0 ± 5.1	174.3 ± 6.0	170.7 ± 6.6
Leg Length (cm)	88.1 ± 4.6	90.6 ± 4.5	89.4 ± 4.6
Wrist Crease Height (cm)	80.3 ± 3.1	83.5 ± 4.3	82.0 ± 4.1
48%-height (cm)	80.2 ± 2.4	83.7 ± 2.9	82.0 ± 3.2
55%-height (cm)	91.8 ± 2.8	95.9 ± 3.3	93.9 ± 3.6
Normal c**adence (steps/min)**	115 ± 7	112 ± 9	114 ± 8
Slow c**adence (steps/min)**	92 ± 6	90 ± 7	91 ± 6

Note. Data is presented as M ± SD.

Trial Conditions

Each participant was assessed in eight walking conditions based on two cadences and four walking-aid/handle-height conditions. Cadence conditions, slow (CS) and normal (CN), were established during an initial familiarization session. In this session, the participants were instructed on how the rollator worked and were asked to walk around the laboratory with the device. Anthropometric measurements (i.e., weight, height, limb length, and wrist crease height) were taken with a stadiometer and measuring tape, and cadences were established by having the participants walk without the rollator with one reflective marker on each lateral malleolus. For cadence (steps per minute), two walking trials using their preferred speed were carried out. Cadence was calculated from these trials and the CN condition was established. The CS condition was set as 80% of the CN condition.

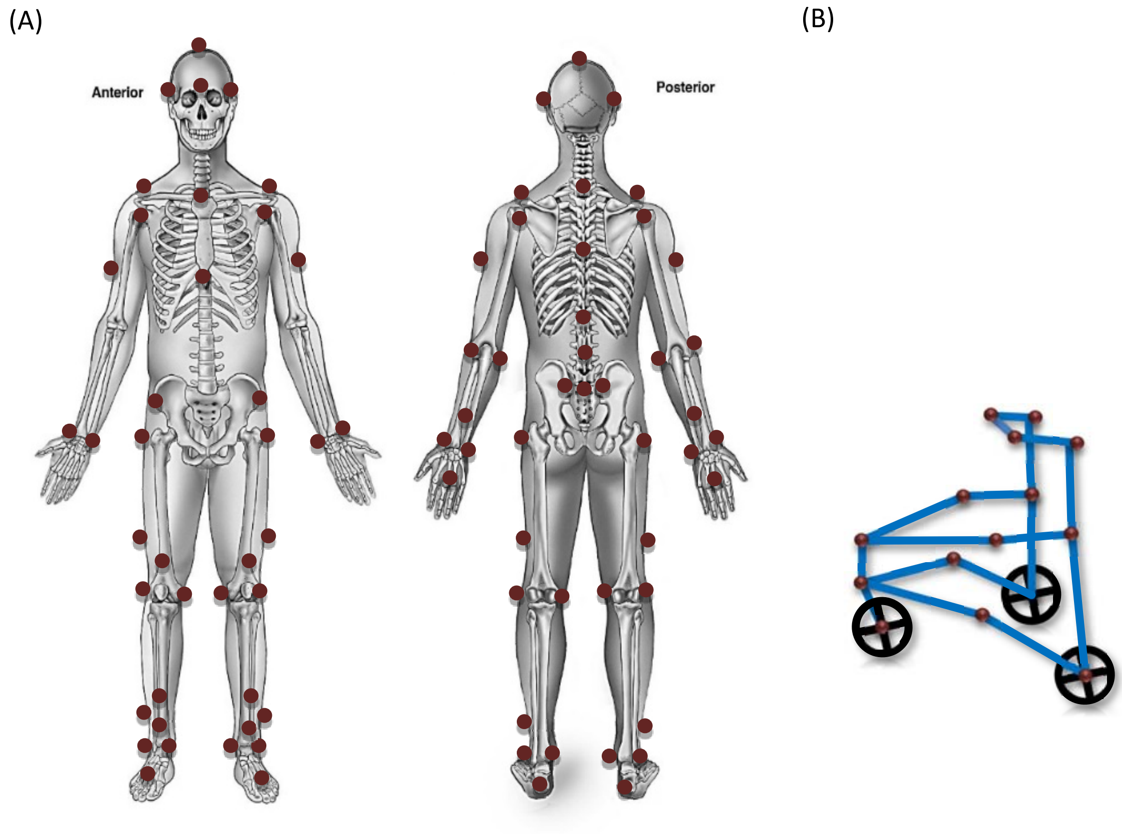
The walking-aid conditions were: (a) walking without a rollator (no rollator; NR), (b) walking with a rollator with handle-height adjusted to participant's wrist crease (HW), (c) walking with a rollator with handle-height adjusted at 48% of participant's height (H48), and (d) walking with a rollator with handle-height adjusted at 55% of participant's height (H55). The order of the conditions was randomized for each participant to prevent order effects, and each participant performed five successful walking trials on a 10-m path for each condition. Each participant performed a total of 40 walking trials (8 cadence-height conditions x 5 trials). To eliminate effects of the shoes over GRF, participants were requested to walk barefoot for every condition.

Experimental Setup

A 10-camera infrared motion capture system (Oqus, Qualisys AB, Gothenburg, Sweden) sampling at 300 Hz was used to capture the trajectories of 74 reflective markers attached to the participants and the rollator (see Figure 1). The 80-point 'Walker' body model (see Figure 1A, Table 3) consisting of 59 markers (49 dynamic and 10 static-only) and 21 computed points (including 13 joint centers) were used to analyze the gait motion. The 15-point 'Rollator' body model was used to visualize the rollator (see Figure 1B). Cameras were calibrated boefore each data collection session.

Figure 1

Body Models



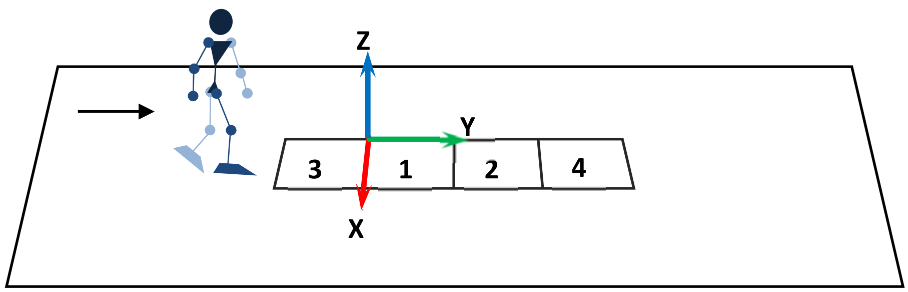
Note. Panel A: Participant body model. Panel B: Rollator body model.

Four force-plates (AMTI OR65, Advanced Mechanical Technology, Inc., Watertown, Massachusetts, USA) sampling at 1,000 Hz were used to capture the GRF for two steps (left and right). The force-plates were set one next to another in a straight formation in the walking direction (see Figure 2). The Trionic Veloped rollator (Trionic Sverige AB, Uppsala, Sweden) was used in the walking-aid conditions because the distance between the rear tires was wider than the force-plates. The origin of the global reference frame was set at a corner of force-plate #1 with the Y-axis and Z-axis aligned along the walking direction and vertically, respectively (see

Figure 2). Qualisys Track Manager (QTM 2019.3, Qualisys AB, Gothenburg, Sweden) was used for capturing trials, and tracked maker coordinates and force data were stored in the C3D file format (http://www.c3d.org).

Figure 2

Force plate and global reference frame setup



Note. Four plates (#1-4) were used and the Y-axis was aligned along the walking direction with the Z-axis being the vertical axis.

Experimental Procedures

Reflective markers were placed on select anatomical landmarks, using hypoallergenic double-sided tape. The participants wore spandex attire to minimize marker motion artifacts. A T-pose static trial with arms abducted 90° was captured before the gait trials and was used to locate joint centers (see Table 2). Ten static-only markers (see Table 3) were removed after the static trial was captured. Between conditions, a 5-min resting period was offered to avoid fatigue. During that period, the principal investigator adjusted the rollator for the next height condition.

To avoid force plate targeting, participants were asked to walk as naturally as possible, while gazing on a “T” marked on the opposite wall at eye level. Participants walked 2 m before and after the capture area to avoid any acceleration/deceleration in the capture area.

Participants were also instructed to walk consistently in terms of velocity (maintaining a natural step length) and weight-bearing over the rollator without losing their rhythm of cadence. Once all the conditions were collected, the participant was asked to complete an exit survey about their perceptions of the handle adjustments (Appendix C).

A trial was considered successful if the cadence was maintained without a reduction in step length. Cadence was controlled using a smart-phone metronome application (Pro Metronome, EUMLab, Berlin, Germany) set to the cadences established during the familiarization session. Trials were discarded if both feet landed on the same force plate, or if the front tire and a foot were over the same force plate.

Table 2

'Walker' Body Model Degrees of Freedom (Joints)

Joint	**Proximal Segment**	**Distal Segment**	**Axis**	**Joint Motion**
C7/T1	Chest	Head	X	Flexion (-) / Extension (+)
			Y	Right flexion (+) / Left flexion (-)
			Z	Right rotation (-) / Left rotation (+)
T12/L1	Abdomen	Chest	X	Flexion (-) / Extension (+)
			Y	Right flexion (+) / Left flexion (-)
			Z	Right rotation (-) / Left rotation (+)
L4/L5*	Pelvis	Abdomen	X	Flexion (-) / Extension (+)
			Y	Right flexion (+) / Left flexion (-)
			Z	Right rotation (-) / Left rotation (+)

Joint	Proximal Segment	Distal Segment	Axis	Joint Motion
Right Shoulder	Chest	Right Upper Arm	X	Flexion (+) / Extension (-)
			Y	Abduction (-) / Adduction (+)
			Z	Internal rotation (+) / External rotation (-)
Right Elbow	Right Upper Arm	Right Forearm	X	Flexion (+) / Extension (-)
Right Wrist	Right Forearm	Right Hand	X	Flexion (+) / Extension (-)
			Y	Radial deviation (-) / Ulnar deviation (+)
Left Shoulder	Chest	Left Upper Arm	X	Flexion (+) / Extension (-)
			Y	Abduction (+) / Adduction (-)
			Z	Internal rotation (-) / External rotation (+)
Left Elbow	Left Upper Arm	Left Forearm	X	Flexion (+) / Extension (-)
Left Wrist	Left Forearm	Left Hand	X	Flexion (+) / Extension (-)
			Y	Radial deviation (+) / Ulnar deviation (-)
Right Hip*	Pelvis	Right Thigh	X	Flexion (+) / Extension (-)
			Y	Abduction (-) / Adduction (+)
			Z	Internal rotation (+) / External rotation (-)
Right Knee*	Right Thigh	Right Shank	X	Flexion (-) / Extension (+)
Right Ankle*	Right Shank	Right Foot	X	Dorsi-flexion (+) / Plantar-flexion (-)
			Y	Inversion (+) / Eversion (-)

Joint	Proximal Segment	Distal Segment	Axis	Joint Motion
Left Hip*	Pelvis	Left Thigh	X	Flexion (+) / Extension (-)
			Y	Abduction (+) / Adduction (-)
			Z	Internal rotation (-) / External rotation (+)
Left Knee*	Left Thigh	Left Shank	X	Flexion (-) / Extension (+)
Left Ankle*	Left Shank	Left Foot	X	Dorsi-flexion (+) / Plantar-flexion (-)
			Z	Inversion (-) / Eversion (+)

Note: *Joints of interest.

Table 3

The 80-point (59 marker) 'Walker' Body Model

Segment	Marker/Computed points	Acronym	Description
Pelvis	Markers (7)	(R/L)ASIS	right/left anterior superior iliac spine
		(R/L)PSIS	right/left posterior superior iliac spine
		Sacrum	mid-point between the two PSIS markers
		(R/L)GT*	right/left greater trochanter
	Computed (8)	JL45	L4/L5 joint, computed using the 'MacKinnon Method'
		J(R/L)Hip	right/left hip joint, computed using the 'Tylkowski-Andriacchi Method.'
		MHip	mid-Hip - the mid-point between the two hip joints
		M(R/L)Pel	mid-right/left pelvis - the mid-point of RASIS and RPSIS
		MPel	mid-pelvis - the mid-point between MRPel and MLPel
		MASIS	the mid-point between RASIS and LASIS

Segment	Marker/Computed points	Acronym	Description
Thighs	Markers (8)	(R/L)Quad	right/left quadriceps tendon above the patella
		(R/L)LThigh	right/left iliotibial band on the lower third
		(R/L)LKnee	right/left lateral femoral epicondyle
		(R/L)MKnee*	right/left medial femoral epicondyle
	Computed (2)	J(R/L)Knee	right/left knee joint computed using the 'Mid-Point Method'
Shanks	Markers (10)	(R/L)Fib	right/left fibula distal third 3 inches above the lateral malleolus
		(R/L)UTib	right/left tibial anterior border 1 inch above the fibular marker
		(R/L)LTib	right/left tibial anterior border 1 inch below the fibular marker
		(R/L)LAnkle	right/left lateral malleolus prominence
		(R/L)MAnkle*	right/left medial malleolus prominence
	Computed (2)	J(R/L)Ankle	right/left ankle joint, computed using the 'Mid-Point Method'
Feet	Markers (4)	(R/L)Toe	right/left second metatarsophalangeal joint
		(R/L)Heel	right/left calcaneus bone

Segment	Marker/Computed points	Acronym	Description
Abdomen	Markers (3)	L3	third lumbar spine (one vertebra higher than the Inter-Iliac crest line [Snider et al., 2011])
		T12	twelfth thoracic spine (insertion of the last floating rib)
		XP	xiphoid process
	Computed (1)	Mab	mid-abdomen - the mid-point between XP and T12
Chest	Markers (9)	T7	seventh thoracic spine (inferior border of the scapula [Cooperstein & Haneline, 2007])
		C7	seventh cervical spine (Chakraverty et al., 2007)
		SN	sternal notch
		(R/L)Acro	right/left shoulder over the acromion
		(R/L)ASho	right/left anterior glenohumeral joint
		(R/L)PSho	right/left anterior glenohumeral joint
	Computed (3)	MTh	mid-thorax - the mid-point between SN and C7
		J(R/L)Sho	right/left shoulder joint computed using the 'Mid-Point Method'

Segment	Marker/Computed points	Acronym	Description
Head	Markers (4)	(R/L)Head	right/left temporal bone above the ear
		Ahead	frontal bone over the glabella
		THead	head vertex
	Computed (1)	MHead	mid-head - the mid-point between the RHead and LHead
Upper Arms	Markers (6)	(R/L)Arm	right/left proximal part of the triceps tendon
		(R/L)LElbow	right/left lateral humeral epicondyle
		(R/L)MElbow*	right/left medial humeral epicondyle
	Computed (2)	J(R/L)Elbow	right/left elbow joint computed using the 'Mid-Point Method'
Forearms	Markers (6)	(R/L)Farm	right/left dorsal lower third of the forearm
		(R/L)LWrist*	right/left radial styloid process
		(R/L)MWrist	right/left ulnar styloid process
	Computed (2)	J(R/L)Wrist	right/left wrist joint, computed using the 'Mid-Point Method'
Hands	Markers (2)	(R/L)Hand	right/left third metacarpophalangeal joint

Note. * Static-only markers.

Data Reduction and Processing

Captured motion and force plate data were imported into Kwon3D (Visol Inc., Seoul, Korea; Version 5.1), were data analysis were performed. Marker coordinates were filtered with a Butterworth low-pass 4th-order zero phase lag filter with a cut-off frequency of 6 Hz.

Sixteen body segments were defined (i.e., pelvis, abdomen, chest, head, two upper arms, two forearms, two hands, two thighs, two legs, and two feet [see Table 3]), linked through 13 joints. From the 13 joint centers, the L4/L5 joint was computed using the MacKinnon method (1993), hip joint centers were computed using the Tylkowski-Andriacchi hybrid method (Bell et al., 1989), and the last 10 were computed using the mid-point method (see Table 4).

Table 4

Method Used to Locate Computed Joint Centers

Method	Joint computed	Points needed	Reference
Tylkowski-Andriacchi	Hip center	RASIS, LASIS, RGT, LGT, Sacrum	Bell et al. (1989)
MacKinnon	L4/L5	MASIS, Sacrum	MacKinnon & Winter (1993)
Mid-Point	Knees, ankles, shoulders, elbows, wrists	Two markers (e.g., right lateral knee and right medial knee in the case of the right knee)	

The segmental COM positions were determined by using the body segment parameters reported by Zatsiorsky and Seluyanov (1983) and corrected by De Leva (1996). The local (segmental) reference frames were defined based on the marker and joint coordinates with the

X-, Y-, and Z-axis aligned to the mediolateral, anteroposterior, and longitudinal axis, respectively (see Table 3).

For this study, only the lower body segments (i.e., pelvis, thighs, shanks, and feet) and joints were included in the kinetic analysis. The joint coordinate system (JCS; Grood & Suntay, 1983) was used to calculate the RJM of the joints. Resultant joint moments were calculated through an inverse dynamics procedure based on Newton's equations of motion:

$$\mathbf{F}_{\mathrm{j}} = \sum_{\mathrm{s}} \frac{\mathrm{d}\mathbf{P}_{\mathrm{s}}}{\mathrm{dt}} - \sum_{\mathrm{s}} \mathbf{W}_{\mathrm{s}} - \mathbf{F}_{\mathrm{E}} \quad [1]$$

$$\mathbf{N} = \sum_{\mathrm{s}} \left(\frac{\mathrm{d}\mathbf{L}_{\mathrm{s}}}{\mathrm{dt}} + \mathbf{r}_{\mathrm{js}} \times \frac{\mathrm{d}\mathbf{P}_{\mathrm{s}}}{\mathrm{dt}} \right) - \sum_{\mathrm{s}} \left(\mathbf{r}_{\mathrm{js}} \times \mathbf{W}_{\mathrm{s}} \right) - \left(\mathbf{r}_{\mathrm{jE}} \times \mathbf{F}_{\mathrm{E}} + \mathbf{N}_{\mathrm{E}} \right) \quad [2]$$

where $\mathbf{F}$ is the resultant joint force acting at the joint, $\frac{d\mathbf{P}_s}{dt}$ is the rate of change in linear momentum of a segment, $\mathbf{W}_s$ is the weight of a segment, $\mathbf{F}_E$ is the ground reaction force acting as an additional external force on the foot, $\mathbf{N}$ is the resultant joint moment acting at the joint, $\frac{d\mathbf{L}_s}{dt}$ is the rate of change in local angular momentum of a segment due to the rotation of the segment about its own COM, $\mathbf{r}_{js}$ is the position vector drawn from the joint center to a segment's COM, $\mathbf{r}_{jE}$ is the position vector drawn from the joint center to the point at which the ground reaction force acts, and $\mathbf{N}_E$ is the ground reaction moment.

Pelvis segment was related to the global reference frame, given by the laboratory, and orientation was calculated using the ZXY (longitudinal-mediolateral-anteroposterior) rotation sequence; while the relative orientation angles of the segments to their respective proximal segments were calculated using the XYZ (mediolateral-anteroposterior-longitudinal) rotation sequence. The JCS was used to extract the RJM acting on the L4/L5, hip, knee, and ankle joints. Flexor/extensor (L4/L5, hip, and knee) and plantar/dorsi-flexor (ankle) moments were about the first axis of rotation. The abductor/adductor (Hip and knee) and lateral flexor (L4/L5) moments were about the second axis of rotation, while the inverter/everter (ankle) moment was about the third axis of rotation. Finally, the internal/external rotator (hip, and knee) and left/right rotator (L4/L5) moments were about the third axis of rotation, while the internal/external rotator (ankle) moments were about the second axis of rotation (see Table 2).

Data Analysis

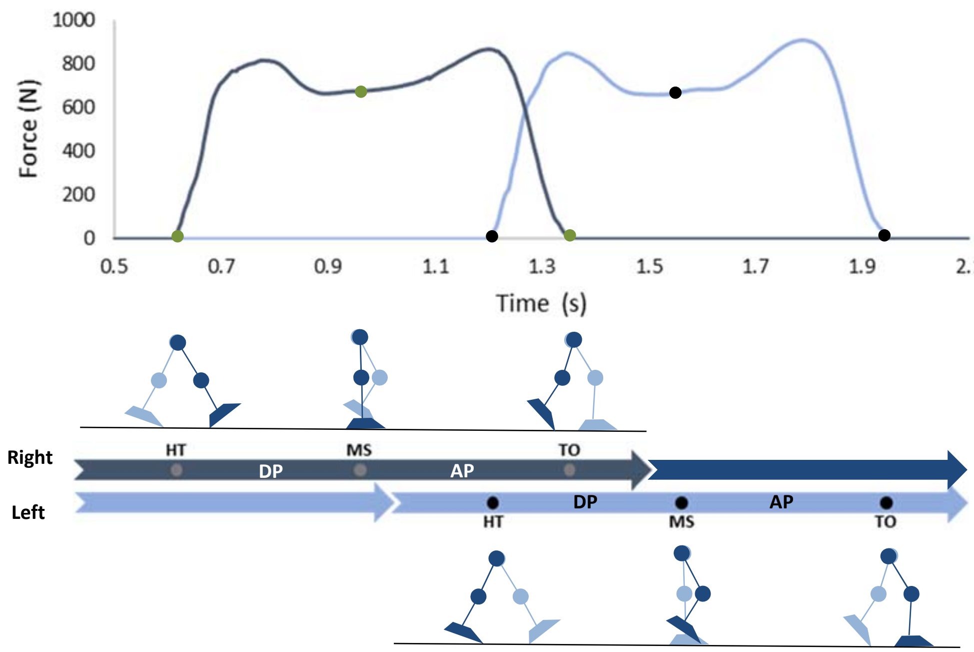
For data analysis, only the stance phase of each limb (from the HC to the toe-off [TO]) was analyzed from the gait cycle. Two consecutive steps (i.e., one right and one left) were identified using vertical GRF. Since the two steps overlapped, the raw data was reduced from the HC of the first step to the TO of the second step (i.e., right HC – left TO [see Figure 3]). Finally, in order to perform kinetic analysis on each leg, the stance phase was subdivided into two phases (Whittle, 2007a) using three events defined based on the vertical GRF of each step and the position of ankle markers towards the walking direction (see Figure 3). The events were defined as follows:

- Heel Contact (HC): the instant the foot touches the force-plate (vertical GRF > 100 N).
- Mid-Stance (MS): the instant both medial ankle markers show the same Y-coordinate.
- Toe-Off (TO): the instant the toe leaves the ground (vertical GRF < 100 N).

The two phases were defined as follows: Deceleration Phase (DP) between HC and MS events and Acceleration Phase (AP) between MS and TO events. These events and phases were identified for the right and the left limbs. Assuming symmetry between the limbs, the left limb inward-outward data was mirrored to match the right limb data, and both limbs were averaged as representative of that trial.

Figure 3

Defined gait stance events and phases



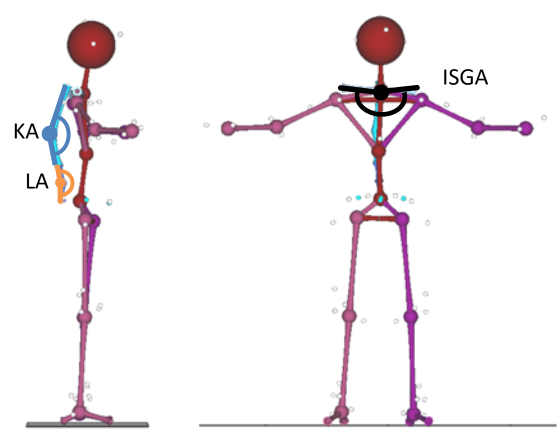
Note: HC = Heel Contact; MS = Mid-Stance; TO = Toe-Off; DP = Deceleration Phase; AP = Acceleration Phase.

For kinematic data, three joint angles (i.e., right elbow, left elbow, L4/L5 joint) were computed in the sagittal plane in order to describe postural changes resulting from different rollator handle-heights; assuming symmetry between the two arms, angles for both arms were averaged and computed as elbow flexion data for the trial. In addition, the kyphotic angle (KA) was calculated using the lines created by the T7 to T12 and T7 to C7 markers, where a decrease of the angle between these lines, implies an increase of the kyphosis (see Figure 4). The lordotic angle (LA) was calculated using the lines created from the L2 to Sacrum and L2 to T12 markers, where an increase of the angle created between these lines, implies an increase of lordosis (see Figure 4). The inter-shoulder girdle angle (ISGA) was calculated from the mid-thorax point to the right acromium marker and mid-thorax point to the left acromium marker in the frontal plane, where an increase of the angle implies elevation of the shoulders (see Figure 4). The KA and LA were projected to the sagittal plane of the chest and abdomen, respectively, while the ISGA was projected to the frontal plane of the chest.

For the kinetic data, three GRF components (vertical, forward-backward, and inward-outward) and the RJM of the L4/L5 joint (sagittal plane), hip joint (sagittal and frontal planes), knee joint (sagittal plane), and the ankle joint (sagittal and frontal planes) were used in the data analysis. The GRF variables were normalized by BW and the RJM was normalized by body weight and limb length (BW*LL).

Figure 4

Schematic representation of the postural user angles



Note: KA = kyphotic angle; LA = lordotic angle; ISGA = inter-shoulder girdle angle.

The data from each trial was time-normalized for the stance phase from HC until TO for each limb (0 – 100). Mean values of the five trials of each condition were used for statistical analysis to develop each participant's representative trial. Finally, ensemble average patterns (see Figure 5 and Figure 6) for each condition were developed from the representative data of all participants to create a population representative pattern. Left inward-outward patterns were adjusted to maintain the same orientation as the right leg patterns (i.e., positive trend for outward and negative for inward).

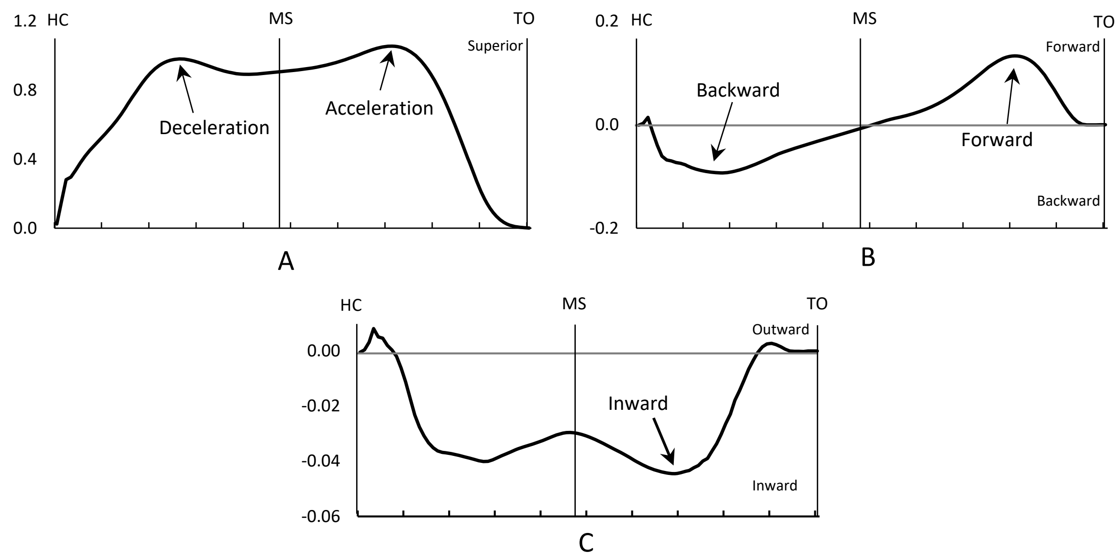
Kinetic (GRF and RJM) and kinematic (COM velocity, and postural angles) parameters were utilized for statistical analysis. Peak COM velocity normalized by participant's height, five peak postural angles, five normalized peaks forces for GRF, and six normalized peak RJMs were extracted as follows:

- Peak velocity: the maximum COM velocity during the stance phase.
- Peak elbow flexion: the maximum elbow angle value in the sagittal plane during the stance phase.
- Peak kyphosis: the lowest value of the kyphotic angle projected in the sagittal plane of the thorax during the stance phase.
- Peak lordosis: the highest value of the lordotic angle projected in the sagittal plane of the thorax during the stance phase.
- Peak L4/L5 flexion: maximum angle value on the L4/L5 joint in the sagittal plane during the stance phase.
- Peak ISGA: minimum value of the ISGA angle projected in the frontal plane of the thorax.
- Deceleration force: the maximum vertical GRF between HC and MS events (see Figure 5A).
- Propulsion force: the maximum vertical GRF between MS and TO events (see Figure 5A).
- Backward force: the maximum backward GRF (negative Fy [see Figure 5B]).
- Forward force: the maximum forward GRF (positive Fy [see Figure 5B]).

- Inward force: the maximum inward GRF (negative Fx [see Figure 5C]).

Figure 5

Ground reaction force patterns:



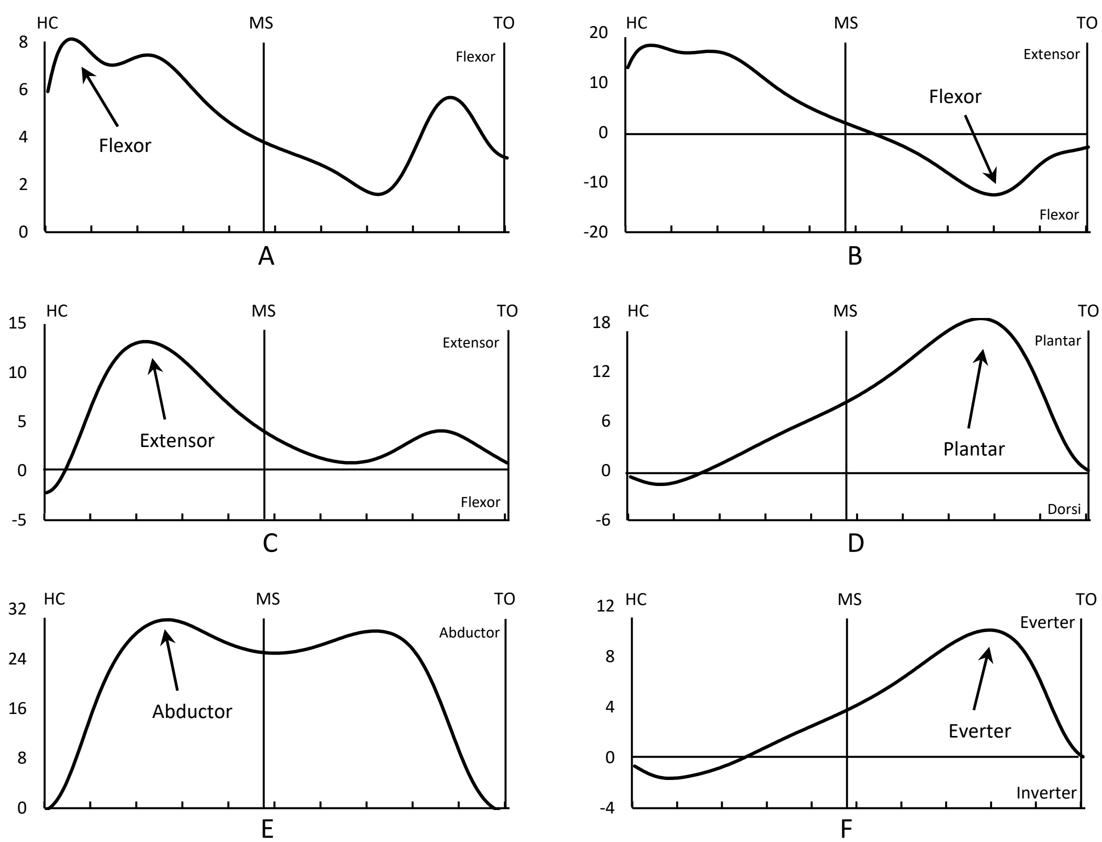
Note: Panel A: Vertical force. Panel B: Forward/backward force . Panel C: Inward/outward force. Arrows indicate the location of respective labeled peak force location.

- Ankle plantar flexor moment: the maximum RJM value on the ankle's sagittal plane (see Figure 6D).
- Ankle everter moment: the maximum RJM value in the ankle's frontal plane (see Figure 6F).
- Knee extensor moment: the maximum RJM value in the knee's sagittal plane (see Figure 6C).
- Hip flexor moment: the minimum RJM value in the hip's sagittal plane (see Figure 6B).
- Hip abductor moment: the maximum RJM value in the hip's frontal plane between HC and MS events (see Figure 6E).

- L4/L5 flexor moment: the minimum RJM value in the L4/L5's sagittal plane between HC and MS events (see Figure 6A).

Figure 6

Resultant joint moment patterns



Note: Panel A: L4/L5 flexor moment. Panel B: Hip extensor/flexor moment. Panel C: Knee extensor/flexor moment. Panel D: Ankle plantar/dorsiflexor moment. Panel E: Hip abductor moment. Panel F: Ankle everter/inverter moment. Arrows indicate the location of respective labeled peak moment locations.

Statistical Analysis

Three separate repeated measures, 2 x 4 (cadence x height) factorial Multivariate Analysis of Variance (MANOVA) were conducted to compare select kinematic variables (i.e., elbow flexion, kyphosis, lordosis, L4/L5 flexion, walking velocity), normalized GRF peaks (i.e., acceleration, deceleration, forward, backward, inward), and normalized RJM peaks (i.e., hip flexor, hip abductor, knee extensor, ankle plantar flexor, ankle everter, L4/L5 flexor), respectively. Preliminary assumptions testing was conducted and no serious violations of normality, linearity, univariate and multivariate outliers were noted. Significant multivariate analysis interaction or multivariate cadence and walking-aid condition significance was explored. Univariate analysis significance was conducted and, in the case, that univariate sphericity was not found ($p < .05$ in the Mauchly's test of sphericity), Greenhouse-Geisser correction was used. Bonferroni post-hoc tests were used to follow up on significant condition differences. Results were analyzed using SPSS v.22 (IBM Inc., Armonk, NY) with a .017 significance level.

CHAPTER IV

RESULTS

This chapter is divided into the following sections: Kinematics, Ground Reaction Force, and Resultant Joint Moment.

Kinematics

No significant cadence x walking-aid interaction [*Wilks'* λ = .591, *F* (15, 18) = 1.97, *p* = .096] was found in peak postural angle variables. Significant main effects were observed in walking-aid factors [*Wilks'* λ = .06, *F* (18, 15) = 13.24, *p* < .001].

Table 5

Summary of average kinematic variables of the four walking-aid conditions (N = 33)

	Walking-aid Conditions				
	NR	**HW**	**H48**	**H55**	***p***
Elbow flexion	39.3 ± 1.5	48.5 ± 1.6§*	48.5 ± 1.8§*	77.3 ± 1.8§	< .001
Kyphosis	154.0 ± 0.7	156.0 ± 0.8	154.1 ± 0.8	156.2 ± 0.8§	.001
Lordosis	196.5 ± 1.0	194.5 ± 1.0§	194.7 ± 1.0§	195.6 ± 1.2§	< .001
ISGA	178.3 ± 1.2	190.4 ± 2.6§*	190.2 ± 2.4 §*	196.0 ± 3.0§	< .001
L4/L5 flexion	17.1 ± 1.2	18.9 ± 1.8§	18.4 ± 1.8§	18.6 ± 1.7§	< .001
Walking Velocity	0.8 ± 0.02	0.8 ± 0.02	0.8 ± 0.02	0.8 ± 0.02	.667

Note: Data is presented in *M ± SE* format. Postural angles are presented in degrees (deg); while normalized walking velocity, by limb length, is presented in limb length per second (LL/s).

§ significantly different from the matching Normal condition ($p < .017$); * significantly different from the matching H55 ($p < .017$). Abbreviations: ISGA = Inter-shoulder girdle angle; NR = No Rollator (Normal); HW = Wrist Crease adjustment; H48 = 48%-height adjustment; H55 = 55%-height adjustment.

The follow-up univariate analysis showed that the use of the rollator resulted in a significant change in postural angles, but no significant changes were seen in walking velocity. Compared to the NR condition, the H55 condition showed the most increase in elbow flexion and ISGA, the most decrease in kyphosis, and the least decrease in lordosis. The H48 condition showed the most lordosis decrease and most L4/L5 torso flexion increase, while the HW condition resulted in the least ISGA changes compared to the NR condition. Between the rollator conditions, the H48 and HW conditions showed similar reduced ISGA and elbow flexion compared to the H55 condition (see Table 5).

Ground Reaction Force

No significant cadence x walking-aid interaction [*Wilks'* $\lambda = .495$, $F\,(15, 18) = 1.22$, $p = .339$] was found in peak GRF variables. There were significant main effects observed in walking-aid factors [*Wilks'* $\lambda = .072$, $F\,(15, 18) = 15.56$, $p < .001$].

The univariate analysis showed that the use of the all-terrain rollator with any handle height adjustment can significantly decrease deceleration, acceleration, forward, and inward peak forces ($p < 0.001$); while in the backward peak force, only the 48%-height adjustment showed significant force reduction ($p = 0.014$) in comparison to the N condition (see Table 6). A 48% handle height adjustment allowed the greatest force reduction of any other handle adjustment, while the 55% handle height adjustment produced the least force reduction in comparison with that of the N condition. No significant differences were found between the rollator conditions in any of the five peak forces.

Table 6

Summary of average normalized peak GRF (in BW) of the four walking-aid conditions (N = 33)

	Walking-aid Condition				
	NR	**HW**	**H48**	**H55**	***p***
Deceleration	1.02 ± 0.02	0.93 ± 0.03§	0.89 ± 0.02§	0.93 ± 0.03§	< .001
Acceleration	1.06 ± 0.02	0.94 ± 0.03§	0.90 ± 0.02§	0.97 ± 0.03§	< .001
Backward	0.16 ± 0.01	0.14 ± 0.01	0.14 ± 0.01§	0.15 ± 0.01	.014
Forward	0.17 ± 0.01	0.15 ± 0.01§	0.15 ± 0.01§	0.16 ± 0.01§	< .001
Inward	0.07 ± 0.01	0.05 ± 0.01§	0.05 ± 0.01§	0.05 ± 0.01§	< .001

Note: Data is presented as proportions of body weights (BW) in *M ± SE* format; § significantly different from the matching NR condition ($p < .017$). Abbreviations: NR = No Rollator (Normal); HW = Wrist Crease adjustment; H48 = 48%-height adjustment; H55 = 55%-height adjustment.

Resultant Joint Moment

No significant cadence x walking-aid interaction [*Wilks'* λ = .337, F (18, 15) = 1.64, p = .663] was found in peak RJM variables. There were significant main effects observed in walking-aid factors [*Wilks'* λ = .088, F (18, 15) = 8.66, $p < .001$].

The univariate analysis showed significant differences in the lower limbs' joint moments due to the use of the rollator compared to the NR condition. In the hip flexor moment, only the HW condition showed a moment decrease compared to the NR condition. For the rest of the lower limb joint moments, the use of the rollator showed significant increase in L4/L5 flexor moment and significant decreases in hip abductor, ankle plantar flexor, and ankle everter joint moments. The H55 condition showed the least difference, while the H48 and HW showed similar differences when compared to the NR condition. Between the rollator conditions, when compared to the H55 condition, the H48 condition showed significant decrease in ankle everter moment (see Table 7).

Table 7

*Summary of average normalized peak RJM (in BW*LL) of the four walking-aid conditions (N = 33)*

	Walking-aid Conditions				
	NR	**HW**	**H48**	**H55**	***p***
L4/L5 Flexor	9.13 ± 0.34	11.42 ± 0.53§	11.22 ± 0.45§	11.19 ± 0.47§	< .001
Hip Flexor	14.50 ± 1.44	12.29 ± 1.24§	12.44 ± 1.23	13.43 ± 1.16	.001
Hip Abductor	33.57 ± 0.89	28.28 ± 0.76§	28.36 ± 0.84§	29.09 ± 0.88§	< .001
Knee extensor	14.91 ± 0.43	12.38 ± 0.41	12.70 ± 0.41	13.31 ± 0.49	.631
Ankle plantar flexor	19.99 ± 0.69	17.62 ± 0.64§	17.47 ± 0.62§*	18.07 ± 0.62§	< .001
Ankle Invertor	11.39 ± 0.94	10.16 ± 0.86§	10.04 ± 0.82§*	10.44 ± 0.85§	< .001

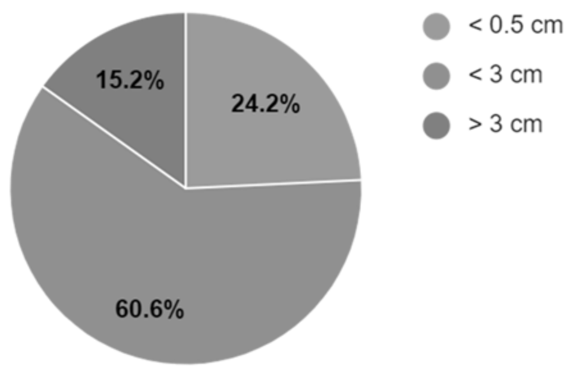
Data is presented as 10^3 body weight * limb length (10^3BW*LL) in *M ± SE* format; § significantly different from the matching Normal condition (*p* < .017); * Significantly different from the matching 55 condition (*p* < .017). Abbreviations: NR = No Rollator (Normal); HW = Wrist Crease adjustment; H48 = 48%-height adjustment; H55 = 55%-height adjustment.

Participants Perceptions

It was expected that, between the three adjustments, the H55 condition would be the highest handle adjustment; what was not expected was that the HW and the H48 adjustments were very close in height. Among all the participants, only 15.2% of them had a difference of more than 3 cm between the H48 and HW conditions (see Figure 7); more so, among those in which the difference was less than 3 cm, 24.2% were less than 0.5 cm. From this difference, almost 40% of the time the H48 condition was higher than the HW condition. This partially rejects our hypothesis about GRF and RJM in two ways, (1) H48 was not the lowest adjustment for all participants, and (2) on the RJM only the ankle moments showed to be the lowest in the H48 condition, with the rest of the moments being lower in the HW condition.

Figure 7

Percentage of the absolute difference between the HW and H48 conditions



To have an input of the perception from the users about the rollator and the comfort and functionality of the different handle adjustments, an exit survey was conducted (see Appendix C) and the following was determined:

1. 54.5% of the participants felt that the HW adjustment allowed them to better unload their weight over the device, followed by the H48 and H55 adjustment (39.4% and 6.1% respectively).
2. Participants felt that they were able to ambulate more naturally at the HW condition, followed by the H55 and the H48 (39.4 %, 36.4%, and 24.2% respectively).

3. Lastly, the H55 and the HW conditions showed to be equally comfortable positions for walking (at 42.4% each), and only 15.2% of the participants felt that the H48 condition was comfortable when ambulating.

CHAPTER V

DISCUSSION AND CONCLUSION

The purpose of the study was to investigate the effects of rollator handle-height on walking velocity, posture, and gait kinetics in two different cadence conditions in able-bodied older adults. To asses these changes, participants were tested using three different handle adjustments and data was compared to walking without a rollator. To isolate the effect of the rollator over force, participants were tested while walking at their normal cadence and walking at 80% of their normal cadence.

Kinematics

There was no significant change in walking velocity across the walking-aid conditions with the cadence controlled. This lack of significance in walking velocity with different handle-heights contradicts the results presented in other studies, where they reported that the change in handle-height of the device produced significant changes in walking velocity (C. Ko et al., 2014; Levangie et al., 1989). Previous studies have shown that the use of walking-aids reduced the walking velocity (Kegelmeyer et al., 2013; Tyson, 1999), which was not seen in the current study. While evaluating balance ability with different handle-heights, Choi et al. (2015) found that people with good balance did not show walking velocity differences between handle adjustments similar to what were used in the present study (i.e., H48 and H55). Additionally, the group of participants in the present study were relatively physically active, participating in group exercises at least once a week, which may improve balance ability and maintain spatiotemporal parameters related to walking velocity (i.e. step length). Alkjaer et al. (2006), Kegelmeyer et al.

(2013), and Levangie et al. (1989) have manifested that other walking devices showed a decrease in step length due to the lack of space between the device and the user. The results of the present study showed that at a controlled cadence the participants using the all-terrain rollator had enough space to maintain normal step length without having to acquire improper hunched posture by pushing the rollator far in front of the body, which is commonly seen with the use of the traditional walker. The ability to maintain spatiotemporal parameters in the older adult has been shown to improve their ambulation and QoL (Boss & Seegmiller, 1981; Devita et al., 2016; S.-U. Ko et al., 2009).

In order to understand the direct interaction of the rollator and the user, the elbow flexion angle was examined. The use of the rollator resulted in a significant increase in elbow flexion of 38° from the NR condition to the H55 condition, but only a marginal significant increase of 9° to the HW and H48 conditions. Between rollator conditions, the H55 condition showed a significant 29° elbow flexion increase compared to the HW and H48 conditions, but no differences were found between HW and H48 conditions. The present study had contradictory results to the handle-height adjustment recommendations where one of the guideline parameters is to maintain elbow flexion between 15 to 30° while using the walking-aid device (Faruqui & Jaeblon, 2010; Government of South Australia, 2013; Laufer, 2003; Mayo Clinic, 2016; Van Hook et al., 2003). At any handle-height adjustment, the participants had a minimum elbow flexion close to 40° at the two lowest adjustments (i.e., H48 and HW) and an elbow flexion around 70° at the H55 adjustment. The increased elbow flexion decreases the ability for a proper shared weight-bearing over the device and affects the ability of the device to catch the user in case of tripping. Even when participants were instructed to maintain the rollator close to them, the handle adjustments may change this distance between the user and the rollator.

Changes in posture seen in the current study reflect the user-rollator interaction. For analysis, the increase in postural kyphosis and lordosis were defined as the decrease in the measured kyphotic angle and increase of the measured lordotic angle, respectively, and the increase in ISGA represented shoulder elevation. The use of the rollator resulted in a significant decrease in kyphosis of 2.2° for the H55 adjustments, and a marginal significant decrease of 2° for the HW adjustment; while a significant decrease in lordosis up to 2° in all rollator conditions in comparison to the NR condition. The ISGA parameter showed that interacting with the rollator creates a shoulder elevation of 12° when walking with the H48 and HW adjustment, and of 18° when walking with the H55 adjustment. This interaction was possible through an increased torso flexion of 1.3° for the H48, 1.5° for the H55, and 1.8° for the HW condition.

Between rollator conditions, only a significant 6° shoulder elevation was seen from the H55 adjustment to the other two adjustments, but no significant changes in spine curvatures were found. The H48 and HW conditions maintained a similar flatter spine attitude and a more relaxed shoulder attitude by the participants, while the H55 condition maintained a more natural kyphosis and lordosis attitude, but an elevated shoulder posture. Maintaining raised shoulders provokes an uncomfortable posture which may lead to rounded shoulder posture, which has been identified as part of cervicogenic disorders (Kim & Kim, 2016). And this adjustment may be limiting the ability to properly shared weight-bearing over the device, this adjustment may be better to maintain balance and stability while ambulating. The different adjustments did not show changes in torso flexion, as seen in H. Choi et al.'s (2015) study, where they found significant lower torso flexion in people with good balance. The current study recruited male and female participants who were instructed to apply force over the device while walking at a controlled cadence, which could compel the participant to hold a posture to allow a better shared weight-bearing; the former study only had female participants who were just instructed to ambulate at a self-selected speed.

Ground Reaction Force

The use of the all-terrain rollator, as expected, produced a decrease in the GRF compared to walking without any walking-aid. Significant decreases in the deceleration, acceleration, forward, and inward peak forces were seen with the use of any handle adjustment; while the backward peak force decreased only when the device was adjusted to the H48 condition. The H48 condition allowed the greatest force reduction of any other handle adjustment in comparison to the NR condition, reducing 0.13 BW in the deceleration peak, 0.16 BW of the acceleration peak, 0.02 BW of the forward (14%), backward peak (12%), and inward peak (26%). These results corroborate the results presented by Takanokura (2010) where he manifested that this adjustment is the critical height, which allows for a relief in muscular load.

No significant difference in GRF between the three different handle adjustments was found in each cadence condition, indicating that the user-rollator interaction did not change significantly regarding these adjustments. Even more, HW and H48 handle adjustments showed very little differences between each other, allowing the most shared weight-bearing with the device. The H55 handle adjustment produced the least significant force reduction in comparison with that of the NR condition, reducing 0.09 BW of force for the deceleration, 0.16 BW for the acceleration, and 0.02 BW for the forward peaks, which is almost 9% of reduction for all these peaks. Nevertheless, the differences to the other two rollator conditions were not significant, contrasting with the results obtained by Takanokura (2010), were he manifested that higher adjustments would not allow for the same shared weight-bearing ability. Also, the decrease of the load seen in the current study of up to 26% contrasts the values found in a study performed on participants with lower extremity surgery, where the rollator device allowed unloading up to 50% of the BW (Youdas et al., 2005) in a static stand; it is important to point out that rollators are not recommended for user's with the high shared weight-bearing need, due to the chance of tipping the device if too much force is applied.

Resultant Joint Moment

The findings of the present study also demonstrated that the use of a rollator had an effect of reducing most of the moments measured, except for the L4/L5 moment, which had a significant increase, and the knee extensor moment, which did not have any significant difference. The L4/L5 flexor moment showed a significant increase of 25% for the HW (2.29 BW*LL) and about 23% for the H48 (2.09 BW*L) and H55 (2.06 BW*LL) adjustment compared to the NR condition. This stooped posture is expected to allow for the user-rollator interaction and is needed for the user to apply force over the device; but this stooped posture generates increased muscular load on the lower back (Takanokura, 2010; Winter, 2009). This is an interesting parameter that, to the knowledge of the researcher, has not been explored extensively enough, as lower back pain is one of the most common complaints among wheeled walking-aid users (Brandt et al., 2003; Kherad et al., 2016).

The hip flexor moment only showed a significant reduction of 15% from the NR to the HW (2.21 BW*LL) condition, while the hip abductor moment showed a significant decrease from the NR condition to the H48 (5.21 BW*LL) and HW (5.29 BW*LL) by almost 16%, and to the H55 (4.48 BW*LL) by 13%. These results are consistent with the ones presented in younger adults (Alkjaer et al., 2006). The lack of significance in the knee extensor moment implies that the user-rollator interaction has no effects over the knee flexion changes; even more, our results showed that only the HW condition showed an effect over the hip, with a 15% reduction in the flexor moment. These two factors demonstrate that balance activity while ambulating is mostly performed maintaining an ankle strategy, which had been proven to be the strategy to maintain balance in the sagittal plane (J. H. Choi & Kim, 2015; Gatev et al., 1999).

The ankle plantar flexor moment showed a significant decrease of force of 12.6% on the H48 (2.52 BW*LL), 9.6% on the H55 (1.92 BW*LL), and 11.8% on the HW (2.37 BW*LL), when compared to the NR condition. Studies focused in perturbances in the direction of the walk have demonstrated that it is the ankle joint that shows changes in the plantar flexor moment to catch the person and avoid falling (Blenkinsop et al., 2017; Spink et al., 2011). Therefore, since the rollator increased the BOS in the anteroposterior direction, it is the ankle joint moment that would mostly be affected by this type of walking-aid, as seen in these results. Additionally, revising the double pendulum model (Winter, 1995) at the stance phase, the hip flexor moments do not intend to decelerate or accelerate the support leg. The function of the hip flexor moment is to sustain the leg over the ground and avoid slipping forward while the body is decelerating; while the hip extensor moment avoids the backwards slipping of leg (Winter, 1980) before the leg starts the swing phase. This is done while controlling the trunk momentum over the pelvis. The ankle plantar flexor moment occurs in the acceleration phase and pushes the body to transfer the body weight to the lead leg; the use of a rollator facilitates this transition since the rollator shares the weight-bearing of the user's body, thus reducing the moment needed in the ankle.

The ankle invertor moment was significantly reduced by 11.8% for the H48 (1.35 BW*LL), 8.4% for the H55 (0.95 BW*LL), and 10.8% for the HW (1.23 BW*LL) condition when compared to the NR condition. The rollator device allows the user to maintain a more stable side to side sway of the body by increasing the points of support, and thus stabilizing the ankle during the transition between single to double stance (MacKinnon & Winter, 1993). Between the rollator conditions, the H48 adjustment showed significant reduction in the ankle plantar flexor moment by 3.3% (0.60 BW*LL) and inverter moment by 3.8% (0.40 BW*LL) when compared to the H55 condition. This confirms that the lower adjustment allows the user to decrease the internal moments by increasing the ability to share weight-bearing with the device (Hall, 2014b). Additionally, the use of a rollator allows the user to increase the bilateral BOS by using the connection between the arms and the rollator as points of support (Messier et al., 2005).

Participants Perceptions

From the participant's perceptions, the wrist-crease height adjustment was perceived to be the best adjustment, allowing good shared weight-bearing assistance while having a comfortable posture when ambulating. Nevertheless, comments from the participants introduce the factor of the user's need, which should be taken into consideration for these adjustments: "the H55 does not provide a good support but can catch you better in case of tripping, the H48 allowed better shared weight-bearing" and "If I used this device, I would work normally pushing it in front of me, and unloading weight on it only if I needed to in order to regain my balance". This feedback and the presented results reaffirm that a proper adjustment of the rollator should allow adequate shared weight-bearing over the device, while allowing a comfortable walk without any development of pain due to the use of the device.

The goal of the current guidelines for rollator handle adjustment is to maintain a neutral attitude of the arm flexion (15 to 30°) while ambulating with the device, but this was not observed in the results presented. One of the reasons for this elbow flexion discrepancy is that an increased stooped posture will demand an increased elbow flexion. Before prescribing the use of a walking-aid, a clinician must educate the user about the clinical objectives of the device and when doing so, a clinician must understand the level of activity and needs that the user intends to maintain. Finally, even though the all-terrain rollator was easy to adjust, it may not allow participants with heights under 160 cm to properly adjust according to these recommendations. Therefore, walking-aids should be created for this group, since a good amount of Hispanic and Asian populations are below this height. Also, we must take into consideration that as an aging effect a person tends to reduce their height due to compaction of vertebrae or even an acquired hunched posture due to muscular weakness.

Limitations

This study intended to see the effects of the all-terrain rollator on gait variables in older adults. Healthy participants were recruited to avoid any additional gait adaptations due to other morbidities (i.e., arthritis and diabetic foot) in an aging population. These results gave a view of the effects of the device; it may not be possible to apply these results to a population that requires the use of a walking-aid, but they may be applied to people that are going through post-surgery rehabilitation in an otherwise healthy older population (i.e., hip surgery). Another limitation was the controlled cadence which did not allow for the assessment of walking velocity changes while using the device. However, the user does not need to distance the device from the lower limbs to maintain walking velocity and acquire inappropriate postures (i.e., stooped posture).

Conclusions

The use of rollator improved spinal posture by reducing kyphosis and lordosis, but with increased shoulder elevation and torso flexion. The results of this study showed that all three adjustments allow shared weight-bearing. However, the body posture while using the device was not the same, generating a slight stooped posture at H48 and HW conditions, which can create lower back pain.

Overall, the adjustments that were more beneficial to the user were either the H48 or the HW conditions, but only the HW adjustment showed significant reduction in all joint moments. Even when 40% of the time the H48 was higher than the HW, the former is much easier to measure and adjust than the latter. However, this is only true if the purpose of using a rollator is to reduce the GRF and RJM; if the need for a rollator is as an assistant device for balance and stability while ambulating, the H55 condition showed to be preferred among the participants and they were able to maintain a more natural posture. With regards to walking velocity, the device did not limit step length, but velocity differences were seen due to changes in cadence.

Recommendations

This study developed important information about handle-height adjustment when using a rollator. It is important to remark that clinical comparisons of posture cannot be done with the current measurements. This study used external markers, which may not reflect the clinical measurement of these curvatures. A validated positioning of the markers may allow for the development of a clinical dynamic evaluation of posture in motion analysis, which currently is not possible. Finally, follow-up studies should focus on the following:

1) To be able to evaluate the muscle ability to perform shared weight-bearing over the device, studies should be done on participants after provoking fatigue in the upper extremities. This may correlate with populations with muscle weakness due to neuromuscular disabilities.

2) A specific disability population should be used to understand the effects of the disease on the ability to use rollator.
3) Studies performed related to user-device interaction should include user’s perception input.

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APPENDIX A

Institutional Review Board Letter of Approval – Texas Woman's University

Texas Woman's University
Institutional Review Board (IRB)
irb@twu.edu
https://www.twu.edu/institutional-review-board-irb/

April 26, 2019

Marco Avalos
Health Promotion & Kinesiology

Re: Initial - IRB-FY2019-29 Handle Height Adjustment Effects on Gait Kinetics in Older Adults while Walking with a Rollator

Dear Marco Avalos,

The above referenced study has been reviewed at a fully convened meeting by the TWU IRB - Denton operating under FWA00000178 and approved on April 16, 2019. If you are using a signed informed consent form, the approved form has been stamped by the IRB and uploaded to the Attachments tab under the Study Details section. This stamped version of the consent must be used when enrolling subjects in your study.

Note that any modifications to this study must be submitted for IRB review prior to their implementation, including the submission of any agency approval letters, changes in research personnel, and any changes in study procedures or instruments. Additionally, the IRB must be notified immediately of any adverse events or unanticipated problems. All modification requests, incident reports, and requests to close the file must be submitted through Cayuse.

Approval for this study will expire on April 15, 2020. A reminder of the study expiration will be sent 45 days prior to the expiration. If the study is ongoing, you will be required to submit a renewal request. When the study is complete, a close request may be submitted to close the study file.

If you have any questions or need additional information, please contact the IRB analyst indicated on your application in Cayuse or refer to the IRB website at http://www.twu.edu/institutional-review-board-irb/.

Sincerely,

TWU IRB - Denton

APPENDIX B

Institutional Review Board Consent Form – Texas Woman's University

Biomechanics Laboratory
304 Administration Dr.
Denton, TX 76204-5647

BIOMECHANICS
TEXAS WOMAN'S UNIVERSITY

TEXAS WOMAN'S UNIVERSITY CONSENT TO PARTICIPATE IN RESEARCH

for a Research Study entitled

"Handle Height Adjustment Effects on Gait Kinetics in Older Adults while Walking with a Rollator"

Principal Investigator: Marco Avalos, M.D. .... mavalos1@twu.edu 940/898-2618
Faculty Advisor: Young-Hoo Kwon, Ph.D. .... ykwon@twu.edu 940/898-2598

Summary and Key Information about the Study

You are being asked to participate in a research study conducted by Marco Avalos, a Texas Woman's University doctoral student in the Biomechanics and Motor Behavior Laboratory, as a part of his dissertation. The purpose of the study is to investigate the effects of different rollator handle heights on gait kinetic factors in able-bodied older adults and how consistent these changes are when walking at two different cadences. The rollator is a wheeled-walker which provides a more continuous motion than a normal walker since it has either three or four tires; also having a seat and a basket which allows the user to rest in case of fatigue, and put their belongings in the device, respectively. You have been invited to participate in this study because you are male or female between 60 and 80 years old. As a participant, you will be asked to walk 40 trials with and without a wheeled-walker in a path of 10 m (33 feet) in the biomechanics laboratory at Texas Woman's University in Denton, Texas. Total estimated time commitment is 1.5 hour for two visits. Following the completion of the study you will receive a $10 gift card for your participation. The risks of this study include the potential loss of confidentiality and anonymity, emotional discomfort, embarrassment, fatigue, coercion, skin irritation, injury, and falling. We will discuss these risks and the rest of the study procedures in greater detail below.

Your participation in this study is completely voluntary. If you are interested in learning more about this study, please review this consent form carefully and take your time deciding whether or not you want to participate. Please feel free to ask the researcher any questions you have about the study at any time.

Description of Procedures

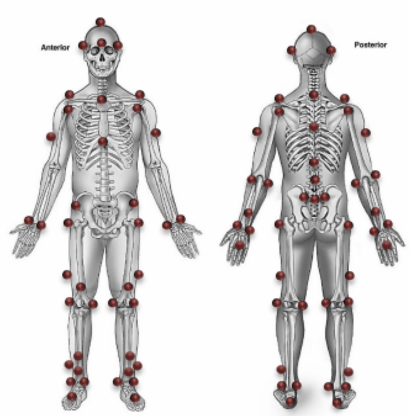
You will be assessed on eight walking conditions based on two cadences and four walking-aid conditions. Cadences include slow cadence (SC) and normal cadence (NC), which will be established in the familiarization session. The walking-aid conditions are: a) walking without a rollator, normal (N); b) walking with a rollator with handle height adjusted to your wrist crease height (WC); c) walking with a rollator with handle height adjusted at 48% of your height (48); and d) walking with a rollator with handle height adjusted at 55% of your height (55). The order of the conditions will be randomized to prevent order effects, and you will perform five successful walking trials on a ten-meter path for each condition, totaling 40 walking trials (8 conditions x 5 trials per condition).

You will attend a familiarization session, where you will be explained of the intention of the study and informed consent will be described and reviewed, and you will be asked to sign it. Once you have agreed to participate, parametric measurements (i.e., weight, height, limb length, and wrist crease height) will be taken, and you will be introduced to the rollator and requested to walk around the laboratory with the device. Cadences will be established by having you walk without the rollator with one reflective marker on each ankle. Two gait trials using your preferred speed will be carried out. Cadence (step per minute) will be calculated from these trials, thus establishing the NC condition's cadence. The SC condition will be set as 80% of the NC condition.

During the data acquisition session, you will be verbally reminded about the protocol and your participation in the study. The principal investigator will place the reflective markers on anatomical landmarks of your body, defining body segments and joints (see diagram below). You will walk barefoot and wearing spandex attire, thus eliminating the effects of shoes on the force generated while you are walking over the floor and avoiding marker movement from the anatomical landmarks.

A static trial will be captured, with you standing still in the center of the lab with arms raised at 90-degrees, before the gait trials and will be used to locate joint centers. Ten markers will be removed after the static trial is captured. During dynamic trials, you will be asked to perform five trials of the four conditions, through the two cadences. Between conditions, a five minutes resting period will be given to avoid fatigue. During this period the principal investigator will adjust the rollator for the following condition. You will be asked to walk as naturally as possible, focusing your gaze on a "T" marked on the opposite wall at eye level, avoiding any concern about how you are stepping on the force-plates. Also, you will be requested to walk consistently maintaining your natural step length and weight-bearing over the rollator without losing their rhythm of cadence.

A trial will be considered successful if the cadence is maintained without a reduction in step length. Cadence will be controlled by using a smart-phone metronome application (Pro Metronome, EUMLab, Berlin, Germany) set to cadences established in the familiarization session. A successful trial must also include a precise measurement of two steps over the force-plates (i.e., one right and one left) without any artifact that may alter the forces of each limb over the force-plate (i.e., two steps over the same force plate or a rollator tire and a step on the same force-plate). In case of detecting an unsuccessful trial, you will be asked to repeat the trial a maximum of five more times for each condition. After the completion of all conditions, you will be asked to answer some questions about your experience in the study.



For you to complete all requirements in this study, the total time commitment will be:

Familiarization session: 0.5 hour

Approved by the
Texas Woman's University
Institutional Review Board
Approved: April 16, 2019
Modifications Approved: June 4, 2019

_________ Initials

Testing session:	**1 hour**
TOTAL TIME COMMITMENT:	**1.5 hour**

Potential Risks

RISK	STEPS TO MINIMIZE RISK
Loss of confidentiality	It is possible that there might be a loss of your confidentiality with data stored offline. To minimize this risk, all data forms collected will be coded using alphanumeric IDs. A single identification form linking names with their respective IDs will be kept in a separate folder from the other data. Persons not associated with the study will have no access to the folders. Data collection sheets will be locked in a file cabinet in Pioneer Hall, Room 123F. There is also a potential risk of loss of confidentiality in all email, downloading, and internet transactions. In case of existing any electronic data that could identify you (i.e., photos), this will be stored in a password protected flash drive. All data inputted in Vicon will be alphanumerically codified to identify you. After three years, when the information is no longer needed, it will be shredded or otherwise appropriately destroyed to be unreadable. Confidentiality will be protected to the extent that is allowed by law.
RISK	STEPS TO MINIMIZE RISK
Loss of anonymity	It is possible that multiple participants may be tested at one time, or that testing may take place such that you are exposed to the general public; because of this, you will be informed (before the study begins) that loss of anonymity may occur. You may withdraw from the study at any time without penalty.
RISK	STEPS TO MINIMIZE RISK
Embarrassment	Words of encouragement and motivational language will be used by the investigators in the event you are embarrassed due to your performance during the testing sessions. Before the placement of the marker, the investigator will request your permission. If you get embarrassed based on your appearance in spandex clothing or marker placement, the investigators will remind you that participation is voluntary, and you may withdraw from the study at any time.
RISK	STEPS TO MINIMIZE RISK
Emotional Discomfort	A research team member of the same sex will be available if the participant prefers, for marker application. Participation is voluntary, and the participants may withdraw from the study at any time.
RISK	STEPS TO MINIMIZE RISK
Fatigue	The principal investigator will verbally check for fatigue with you while performing the trials in every condition. If you feel fatigued, the principal investigator will give you a resting period. Participation is voluntary, and you may withdraw from the study at any time.

RISK	STEPS TO MINIMIZE RISK
Coercion	Services provided to you will not be affected by participation/non-participation in the study. Your relationship with TWU and the School of Health Promotion and Kinesiology will not be affected by your participation or non-participation in the study. Participation is voluntary, and you may withdraw from the study at any time.
RISK	STEPS TO MINIMIZE RISK
Skin irritation	Participation requires the use of double-sided tape. The tape used will be non-irritable hypoallergenic tape to minimize possible skin irritation.
RISK	STEPS TO MINIMIZE RISK
Injury	Every precaution will be taken by the researchers to prevent any injury or problem that could happen during the research study. If an injury should occur, all proper and necessary medical and/or first aid procedures will be followed as dictated by the type or extent of the injury.
RISK	STEP TO MINIMIZE RISK
Falling	Every precaution will be taken by the researchers to prevent any injury or problem that could happen during the research study. A researcher/assistant will always be standing close by or in your line to be able to catch you in case of a loss of balance and to ensure your safety from falling while walking the 10 m (33 feet) path. If an injury should occur, all proper and necessary medical and/or first aid procedures will be followed as dictated by the type or extent of the injury.

The data collected and the survey will be stored in a locked cabinet in the researcher's office. Only the researcher and his advisor will have access to these data. All identifiable data will be destroyed within three years after the study is finished. The signed consent form will be stored separately from all collected information and will be destroyed three years after the study is closed. The results of the study may be reported in scientific journals, but your name or any other identifying information will not be included. There is a potential risk of loss of confidentiality in all email, downloading, electronic meetings and internet transactions.

The researchers will remove all your personal or identifiable information (e.g., your name, date of birth, contact information) from the collected data and/or any study information. After all identifiable information is removed, your personal information collected for this study may be used for future research or be given to another researcher for future research without additional informed consent.

If you would like to participate in the current study but not allow your de-identified data to be used for future research, please initial here _____.

The researchers will try to prevent any problem that could happen because of this research. You should let the researchers know at once if there is a problem and they will try to help you. However, TWU does not provide medical services or financial assistance for injuries that might happen because you are taking part in this research.

Approved by the
Texas Woman's University
Institutional Review Board
Approved: April 16, 2019
Modifications Approved: June 4, 2019

____
Initials

AUTHORIZATION TO USE PHOTOGRAPHS AND/OR AUDIO-VISUAL

In the data collection for this study, we might request to take a photo or video. This material will not be published in any journal, manuscript or webpage. It will in no way be used by Texas Woman's University to promote/publicize ongoing research. This authorization is continuous and may only be withdrawn by your specific rescission of this authorization. You may refuse to take any photo and/or video that may pertain to you– including your image, likeness, and/or voice without compensation. If you authorize, please sign the following statement:

I ___________________________ authorize Marco Avalos to use or reproduce photographs and/or video that may pertain me.

Participation and Benefits

Your involvement in this study is entirely voluntary, and you may withdraw from the study at any time. Following the completion of the study you will receive a $10 gift card for your participation, and at the completion of the study a summary of the results will be mailed to you on request. On receiving the results of the research, if you happen to have any further questions, you are welcome to contact the principal investigator and set-up an appointment for a private consultation to discuss your individual results. The time, date and location will be determined at the time of contact.*

Questions Regarding the Study

You will be given a copy of this signed and dated consent form to keep. If you have any questions about the research study you should ask the researchers; their contact information is at the top of this form. If you have questions about your rights as a participant in this research or the way this study has been conducted, you may contact the TWU Office of Research and Sponsored Programs at 940-898-3378 or via e-mail at IRB@twu.edu.

____ ______________________

Signature of Participant Date

*If you would like to know the results of this study tell us where you want them to be sent:

Email: ___________________________ or Address: ____

Approved by the
Texas Woman's University
Institutional Review Board
Approved: April 16, 2019
Modifications Approved: June 4, 2019

APPENDIX C

Exit Survey

Exit Survey

While walking with a rollator, which position felt that you unload better your weight?

- ☐ 55% height
- ☐ Wrist crease
- ☐ 48% height

While walking with a rollator, which position felt more like your normal walk?

- ☐ 55% height
- ☐ Wrist crease
- ☐ 48% height

While walking with a rollator, which position felt more comfortable to walk?

- ☐ 55% height
- ☐ Wrist crease
- ☐ 48% height

Comments: