Gait Adjustments to Assistive Forces from a Smart Walker

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Gait Adjustments to Assistive Forces from a Smart Walker

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Supervised by,

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Abstract

Smart Walkers are often viewed as the next-generation mobility aid; context-aware and equipped with actuators, these devices can be controlled in real-time to move and provide external forces to assist users in ways that were previously not possible. One potential smart feature is improved physical assistance; that is, to use its actuators to reduce the metabolic cost of walking, increase walking speed, and help maintain a healthy gait. While development thus far have relied on the researcher's or developer's intuition and experience, effectively implementing a physical assistance feature requires a better understanding of the interaction with users, due to the inherent ergonomic challenges involved. The user is physically coupled with the device as they use it for balance and partial body weight support while still mainly relying on their legs to walk. Hence, the external force generated by the Smart Walker may lead to gait adjustments that can counteract any potential benefits afforded by the actuators. This work aims to investigate the user's gait adjustments to the forces generated by a Smart Walker.

In the first study, it was demonstrated that when a constant force is applied, assistive forces increased the walking speed of its users while resistive forces had the opposite effect. More importantly, the perceived exertions reported by users followed a quadratic trend as the force increases from around −20 N to 30 N with minimum exertion occurring at around 1.5% of body weight. While a relatively weak constant force may be helpful, stronger forces may lead to upper body strain or difficulty using the device, which increases exertion.

The second study investigated how users adjust to forces from a Smart Walker when it was used to control the forward speed. This meant that a user only

had to walk at the speed targeted by the Smart Walker and they would not have to push the device forward. However, a substantial proportion of users chose to work against the device and overpower it to walk at a more comfortable speed. Furthermore, even when the users matched the speed targeted, they push or pull on the device although this did not result in any observable change in speed.

Study 3 investigated the biomechanical effects of a constant assistive force from a Smart Walker on its users at a range of speeds. Although walking at higher speeds were found to increase the work done by the users on their center of mass in all phases, the assistive force applied decreased the positive work done in the push-off and rebound phases. Additionally, the assistive force supplied reduced the ankle push-off joint power while increasing the hip pull-off flexion power.

In conclusion, a constant assistive force can be used elicit a higher walking speed and can reduce perceived exertions when applied at low magnitudes. However, this also leads to gait changes such as increased knee extensor loading power and hip pull-off flexion power. Furthermore, it was shown that users will choose to push/pull on the device when walking at speeds different from their preferred walking speed. These results can contribute to providing a basis for proposing effective uses for the assistive force from a Smart Walkers.

Acknowledgements

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Yeoh, W.L., Choi, J., Loh, P.Y., Saito, S., Muraki, S. (2020) The effect of horizontal forces from a Smart Walker on gait and perceived exertion. *Assistive Technology*. 1–9. Available from: doi:10.1080/10400435.2020.1744771.

Chapter 1

General Introduction

1.1 Background

1.1.1 Mobility Disability

Mobility, the ability to move around freely within community environments, is fundamental to a person's ability to live independently. People need to be able to move independently from one point to another in order to perform many of the basic activities of daily living (BADLs) such as dressing and toileting, as well as many of the instrumental activities of daily living (IADLs) such as shopping, housekeeping, and meal preparation (Katz, 1983) in order to live in a community without additional care. The central role that mobility plays in our lives means that any impairment to this ability would have spillover effects to other parts of our life. Loss of mobility can lead to decreased physical activity (Baker, Bodner & Allman, 2003), reduced social engagement (Rosso *et al.*, 2013; Groessl *et al.*, 2007; Yeom, Fleury & Keller, 2008), and have been shown to increase to risk of mortality (Hirvensalo, Rantanen & Heikkinen, 2000).

However, mobility disability have been reported to be the most common type of disability, affecting about 6.5% of the population (Kraus, 2017). Furthermore, this disability disproportionately affects older adults and is one of the more prevalent manifestations of functional decline in old age. As many as one in five older adults have been reported to experience walking difficulty (Kraus, 2017; Sagardui-Villamor *et al.*, 2005) and usual walking speeds, on average, have been found to decline sharply from around the age of 65 (Ferrucci *et al.*, 2016; Schrack *et al.*, 2012). Therefore, the number of mobility disability suffers are expected to grow substantially as the elderly population increases (United Nations, 2019).

In today's ageing world, it is increasingly important that mobility of individuals can be improved in order for them to maintain their wellbeing and independence and thereby, decreasing the burden on caregiving by the society.

1.1.2 Improving Mobility

Mobility limitations arise from the gaps between the environmental/task demands and an individual's functional capacity (Webber, Porter & Menec, 2010). Patla & Shumway-Cook (1999) proposed that these gaps occur along the eight dimensions shown in Figure 1.1. A deterioration in the capacity to meet any one of the demands these dimensions will lead to decreased mobility. Hence, different approaches and strategies will need to be used to improve mobility.

There are, in general, three ways to improve an individual's mobility along these dimensions. Namely, (1) modify the environment to reduce the physical and psychological demands placed on the individual, (2) improve the mobility capabilities of the individual, and (3) bridge the gap between the demands of the environment and the capability of the individual using an assistive device.

Improvements can be made to the environment (approach 1), for instance, by adding curb cuts and having barrier-free pavements, while therapeutic methods (approach 2) can increase a person's walking capability, for example, through various treadmill-training programs and ambulatory training devices. Even though both these approaches are crucial to increasing mobility, it is not always possible to close the gap between environment/task demands and a person's capabilities in this way. In these cases, an assistive device is required to bridge the gap and the type and capabilities of the device can significantly affect mobility as well as the long-term health and capability of its user. Hence, this thesis will focus on approach (3), the use of an assistive device, to improve mobility.

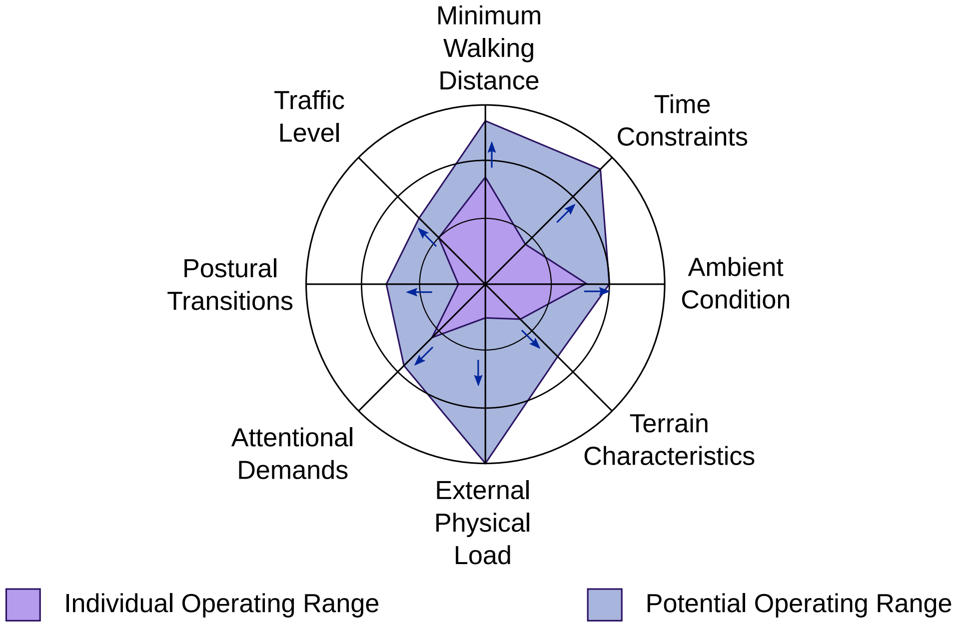


Figure 1.1. The eight dimensions of mobility (Adapted from Patla & Shumway-Cook (1999)).

These assistive devices for mobility, known as mobility aids, can be categorized into alternative and augmentative devices (Martins *et al.*, 2012). Alternative devices such as wheelchairs replace the use of a user's legs with wheels in order to move around. This can cause underutilizations of the user's legs and decreases his or her physical activity, which in turn can accelerate the decline in walking ability and can lead to obesity, loss of muscle mass (sarcopenia) and loss of bone density (osteoporosis) (DiPietro, 2001; Frank *et al.*, 2010). Furthermore, seating for extended periods when using a wheelchair can lead to pressure ulcers in the buttocks (Brienza *et al.*, 2001). Augmentative devices, on the other hand, retain the use of the user's legs when moving around. Examples include canes, four-legged walkers, and four-wheeled walkers. Instead of completely replacing the use of the user's legs to move around, they augment the way in which an individual moves by providing functional compensation, which can increase a user's physical activity (Bertrand *et al.*, 2017) and may slow down the deterioration in walking ability.

As indicated in Figure 1.2, the type of mobility aid to use depends on a person's current levels of function and their overall health and mobility goals. When a person is still able to use his or her legs, it is important to maintain that capability and prevent further decline for as long as possible (A). On the other hand, for people who are no longer able to walk, there is a need to provide assistance so that he or she can move around (B). If the goal were purely to improve mobility, an alternative device would suffice. However, for people who are still able to walk but require assistance (C), the overlapping goals of preventing further decline and providing assistance suggest an augmentative device is preferable.

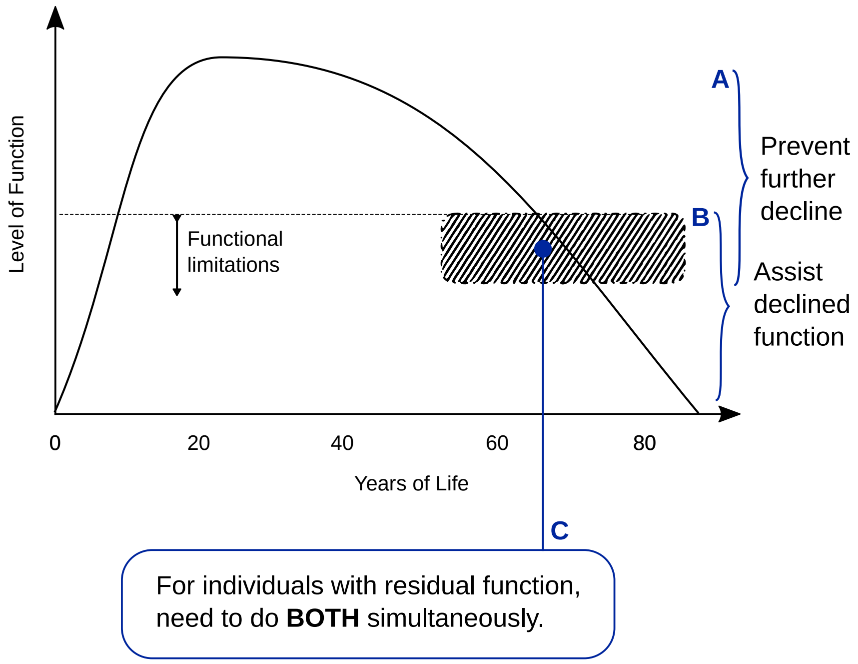


Figure 1.2. Life-course functional trajectories (Adapted from Kuh et al., 2014).

Today, the most common augmentative devices are walking aids such as canes and walkers. These are basically support frames that the users can push on the frames for partial body weight support and to produce balancing forces (Bradley & Hernandez, 2011; Faruqui & Jaeblon, 2010; Van Hook, Demonbreun & Weiss, 2003). However, this means the user is fully responsible for the coordination and generation of forces involved in walking aid use and must be able to meet the strength, endurance, vestibular and cognitive demands that come with it at all times. If at any instant a user is unable to meet any of these demands, he or she will fall and particularly for older adults, the injuries inflicted are likely to be severe. The passive nature of these walking aids limits the assistance it can provide and the groups of users who are able to use them safely.

1.2 Smart Walkers

Technological progress has led to the decreasing cost of sensors, actuators, and controllers (processing power) as well as the decreasing physical size and weight of various sensors and actuators. These trends are leading to the increasing viability of personal walking aids equipped with robotic components that have various smart features. Therefore, we have been seeing increasing interest in applying these robotic technologies to walking aids by researches in recent years as shown in Figure 1.3. Today, there are already a number of commercially available walkers that incorporate robotic technologies (RT.WORKS co., 2020; Robot Care System, 2020; Kowa co., 2020; BEMOTEC GmbH, 2020; Kawamura Cycle LTD, 2020).

These robotic technologies have been incorporated into various kinds of walking aids. An example is the intelligent cane robot developed by Wakita *et al.* (2013) which uses its sensors and motorized Omni-wheels to provide navigational guidance to its user, prevent falls and perform rehabilitation training. However, by far the most common are upgraded four-wheeled walkers known as Smart Walkers or Robotic Rollators (Martins *et al.*, 2012, 2015; Page *et al.*, 2016; Werner *et al.*, 2016, 2018). A selection of Smart Walkers developed in academia is shown in Figure 1.4.

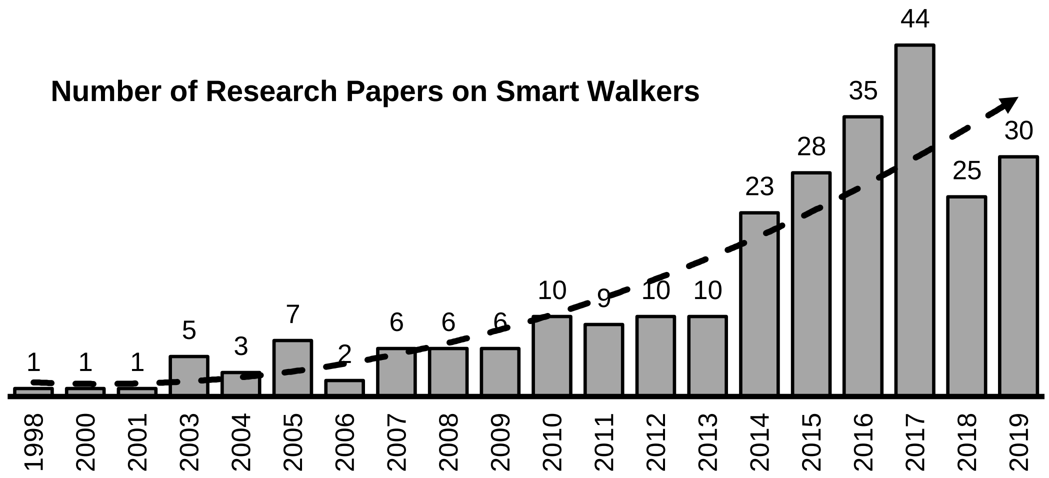


Figure 1.3. Number of research papers on Smart Walkers, by year (records identified through Web of Science*).

1.2.1 Characteristics

Although no two Smart Walkers are perfectly alike, they all share certain characteristics and some functionalities are present in the vast majority of Smart Walkers. Firstly, the basic structure of Smart Walkers is the same as that of four-wheeled walkers. Secondly, they are equipped with sensors that provide information about the user and its surroundings. Thirdly, they have actuated wheels.

Because Smart Walkers shared the same structure as four-wheeled walkers, they inherit their strengths as well as their weaknesses. Unlike canes, four-wheeled

* Search field used: ALL=("Smart Walker" OR "Robotic Rollator" OR "Smart Rollator" OR "Robotic Walker" OR "Walking Assist-Robot" OR "Walking Assistant Robot" OR "Power-Assisted Walking" OR "Smart Walkers" OR "Robotic Rollators" OR "Smart Rollators" OR "Robotic Walkers")

walkers provide bilateral support to its users, preventing asymmetric gait patterns. In addition, the wheels on four-wheeled walkers relieve their users from having to lift the device when moving around, preventing the unnatural “step-to” gait associated with four-footed walker use. Hence, among the various mobility aids, four-wheeled walkers require the least amount of effort to use (Priebe & Kram, 2011) and the gait during their use most resemble normal walking (Van Hook, Demonbreun & Weiss, 2003). However, there are some drawbacks to the use of four-wheeled walkers. They are less stable than other mobility aids; the wheels may roll forward unexpectedly and the lack of kinematic constraints from the front casters can lead to directional instability (Anslow *et al.*, 2001). Therefore, four-wheeled walkers are typically only recommended for people with mild impairment (Bradley & Hernandez, 2011; Faruqui & Jaeblon, 2010).

Smart Walkers are typically equipped with sensors that provide information about its surroundings, the state of the user, and the interaction forces between it and the user. Common sensors include laser range sensors that provide information about the surroundings (Lee *et al.*, 2014; Wachaja *et al.*, 2015; Jiang *et al.*, 2017; Lu, Huang & Lee, 2015; Faria *et al.*, 2014) and the user (Cifuentes *et al.*, 2016; Papageorgiou *et al.*, 2016; Martins *et al.*, 2014a; Jiménez *et al.*, 2018; Fotinea *et al.*, 2016) and embedded handle force sensors (Cifuentes *et al.*, 2014; Jiang *et al.*, 2017; Xu, Huang & Yan, 2015; Tan *et al.*, 2013; Ko *et al.*, 2013).

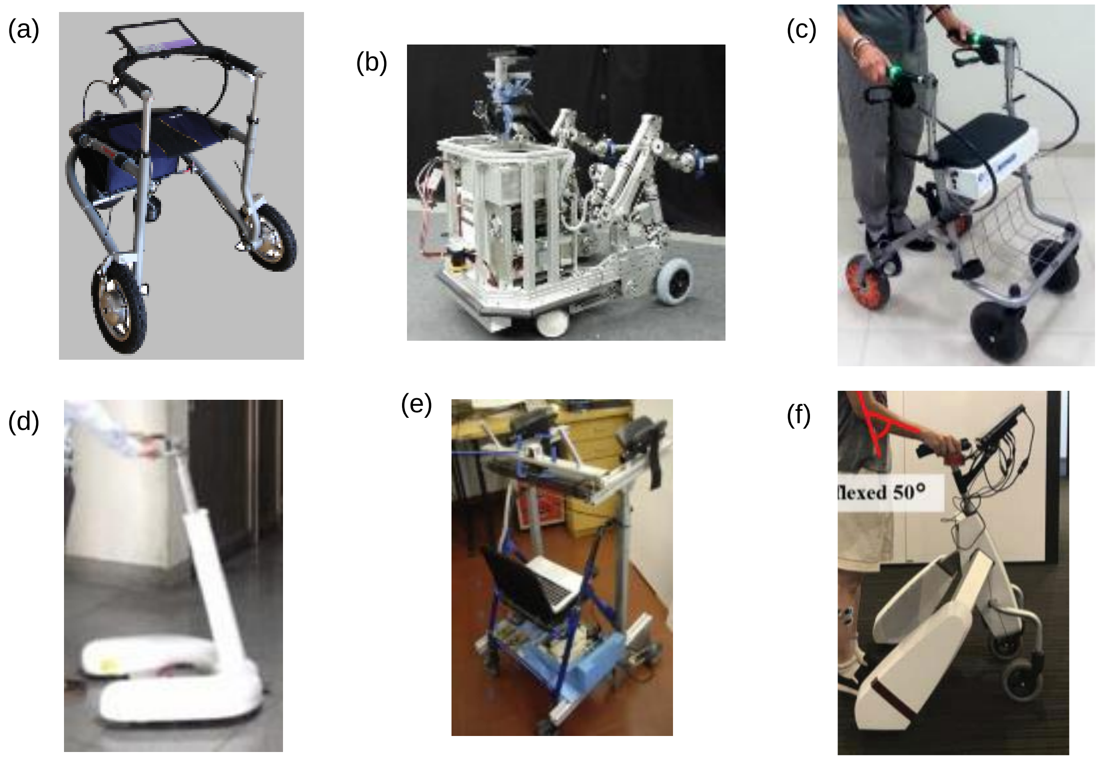


Figure 1.4. Examples of Smart Walker in Research. (a) SMARTWALKER (Shin, Steinmann & Meyer, 2015), (b) MOBOT (Moustris & Tzafestas, 2016), (c) i-Walker (Morone *et al.*, 2016), (d) i-Go (Hsieh *et al.*, 2016), (e) ASBGo (Martins *et al.*, 2014b), (f) Smart Walker (Sato *et al.*, 2019).

The typical actuator found on Smart Walkers are electric motors (Martins *et al.*, 2015) that are connected its wheels. These can be used to generate forces that move or rotate the Smart Walker or result in an interaction force with the user through its handles. In situations where external power is not required, some Smart Walker would use passive electromechanical brakes to produce purely resistive forces (Hsieh, Young & Ko, 2015; Lu, Huang & Lee, 2015; Chen *et al.*, 2013).

1.2.2 Features and Focus

These sensors and actuators are used to implement various features that enable their users to meet the different demands imposed by their environment. Examples of the features commonly implemented are physical assistance, navigation assistance (Sierra *et al.*, 2018; Jiang *et al.*, 2017; Aggravi *et al.*, 2015; Faria *et al.*, 2014), sit-to-stand assistance (Hirata, Muraki & Kosuge, 2006; Geravand *et al.*, 2017; Chugo *et al.*, 2016), and fall prevention (Geravand, Rampeltshammer & Peer, 2015; Dune, Gorce & Merlet, 2012; Nakagawa *et al.*, 2015). These features can improve an individual's mobility in all the dimensions shown in Figure 1.1. That said, the main focus of this thesis is in the potential to extend the operating range of the user in the physical dimensions while encouraging healthy gait. Specifically, the minimum walking distance and time constraints dimensions.

Priebe & Kram (2011) found that the metabolic cost of walking with a four-wheeled walker was 4% higher than that for free walking. Although this is substantially lower than other walking aids, it nonetheless does not decrease the effort required to walk and does not change the minimum walking distance of the user. However, the electric motors in Smart Walkers can be used supply additional energy to reduce the effort required for walking. Previous studies have shown that the energy required for propulsion make up almost half of that required for walking (Gottschall & Kram, 2003; Zirker, Bennett & Abel, 2013). Hence, Smart Walkers has the potential to improve mobility by increasing the minimum walking distance of an individual by supplying additional energy in the form of assistive forces.

Walking at slow speeds can also hinder an individual's mobility due to time constraints imposed by the environment, such as by traffic light on street crossings. This is an issue particularly because walking with four-wheeled walkers have been shown to decrease the walking speed of an individual (Liu *et al.*, 2009). In contrast, external assistive forces have been found to reduce the mechanical effort required to walk faster (Dionisio, Hurt & Brown, 2018). Therefore, assistive force from Smart

Walkers may make it easier to walk faster and enable its user to complete tasks within the time constraints imposed.

In addition, the motorized wheels on Smart Walkers can be used to illicit more natural gait patterns from the user. Recall from Section 1.1.2 that one of the benefits of using an augmentative device is that the user still walks using his or her legs and this may prevent further decline. However, solely continuing to use his or her legs may not be sufficient. Although this can help maintain or even increase physiological capacities such as muscle strength and stamina, effective walking requires the motor skill to coordinate and time gait movements (Kuo & Donelan, 2010). Loss of this skill can lead to inefficient gait (high metabolic cost and low speed) and instability (Brach & VanSwearingen, 2013). Smart Walkers may be actuated to lead to more effective walking and avoid the unnatural gait patterns usually associated with mobility aids.

Although these smart features have significant potential to improve the mobility of an individual, there are substantial ergonomic challenges that make the effective implementation of these features difficult. These difficulties are discussed in Section 1.3. Implementing these features without considering the ergonomic challenges can make the features ineffective and may even be detrimental to a user's gait and mobility.

1.3 Ergonomic Challenges

Despite the benefits of the features discussed previously, effectively implementing these features come with significant ergonomic challenges. In this section, four ergonomic challenges faced when attempting to control a Smart Walker is presented.

1.3.1 Intuitive Control

Even though it is possible for the actuators on a Smart Walker to be controlled mainly through explicit commands by the user, the mental load required to simultaneously walk, maintain balance, and control the Smart Walker is extremely high and will most likely exceed the capabilities of most frail elderly users. As an illustration, if the speed and direction of a Smart Walker were to be controlled using a joystick similar to many powered wheelchairs, the user would have to match the command provided through the joystick with his or her walking speed and direction while simultaneously using the device for partial body weight support and to balancing forces.

Hence, it is important for the Smart Walker to be controlled through implicit commands. In other words, the Smart Walker needs to sense and infer the intention of the user and move the Smart Walker appropriately without explicit instructions from the user. Recognizing this, most Smart Walkers developed by researchers are equipped with sensors that detect the leg position of the user and the handle forces applied by the user to provide this feature (Section 1.2.2). While studies have shown that it is possible to detect the desire movement direction, walking speeds, and pushing force of the user, how best to actuate the Smart Walker based on the interpretation of this intention is still not known.

Some researchers have suggested that Smart Walker should behave passively (only changing the apparent dynamics) (Chuy *et al.*, 2007) or that assistance can be provided simply by amplifying the force applied by the user (Fei Shi *et al.*, 2010). However, most of the smart features that a Smart Walker can provide require that the Smart Walker be controlled actively. That is to say, the Smart Walker need to be able to act against the forces applied by the user, as opposed to passive walking aids whose motion and reaction forces are completely dependent on the user. Hence, these approaches would not be able to provide many of the physical assist features discussed in Section 1.2.2. More research is needed to understand how best to actively control the Smart Walker to provide the various smart features discussed. This involves understanding how the users adjust their gait to motor forces from the Smart Walker.

1.3.2 Physical Interaction

Section 1.1.2 discussed the benefits of using augmentative devices as mobility aids instead of alternative devices. Smart Walkers, being augmentative devices, have the same benefits. They allow their users to continue using their legs to walk while improving their mobility, thereby maintaining their walking ability or at least decreasing the rate of decline. This, however, complicates the control of the Smart Walker and the implementations of the various features discussed in Section 1.2.2.

In a general human-machine system, the device receives commands, either implicitly or explicitly, from the user using their sensors or input devices. This command is then decoded based on information about its environment. Based on this decoded command, the device performs an action using its actuators and provides feedback to its user if appropriate (Figure 1.5). For example, for a motorized wheelchair, the user instructs the device to move forward at a desired speed using a joystick and its electric motors will move the device forward at the speed selected.

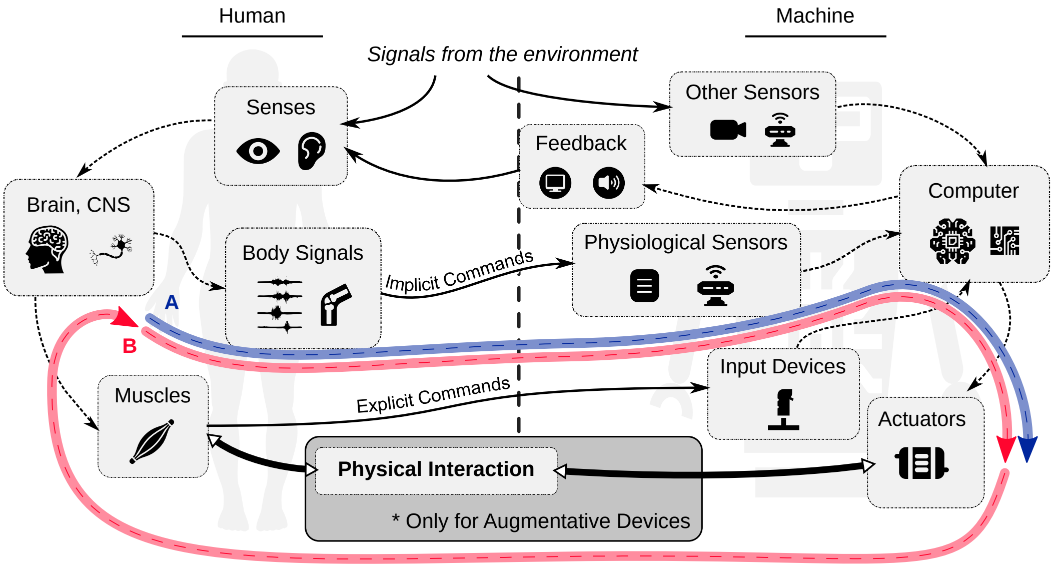


Figure 1.5. Schematic of a general human-machine system showing the added dimension of physical interaction for augmentative devices.

For alternative devices like motorized wheelchairs, the action can be performed without considering the effects on the user (A in Figure 1.5). However, for augmentative devices such as Smart Walkers, there is an added dimension of physical interaction. When using a Smart Walker, the user is still mainly supported by his or her legs and is still relying on those legs to walk and move around. At the same time, the user will be holding on to the handles of the Smart Walker, using it for partial body weight support and to produce balancing forces. This connection at the handles leads to the exchange of forces between the user and the Smart Walker. Hence, due to this physical interaction, the user will have to respond and adjust to the forces generated by the Smart Walker and vice versa. This inadvertently creates a feedback loop where the user response or adjusts to the movement and forces of the Smart Walker, changing his or her body signals and the forces he or she applies to the handles, which in turn changes the motion and the forces generated by the Smart Walker (B in Figure 1.5).

This feedback loop makes it difficult to design an appropriate control method or algorithm without understanding how the user reacts or responds to the forces and movements from a Smart Walker. While it is possible for the Smart Walker to treat this physical interaction as an external disturbance to be rejected, thereby minimizing the effect of the feedback loop, the main goal of a Smart Walker is to improve the mobility of its user without causing further decline to his or her walking ability. Thus, the physical interaction should be used to decrease the user’s effort, increase their walking speed, improve their balance etc.

In order to do so, there is need to understand how users respond and adjust to these forces. This understanding can then act as a basis to control Smart Walkers in a way that elicits the responses and adjustments that best benefit the user and achieve the goals stated.

1.3.3 Dynamics of Bipedal Gait

As mentioned previously, when an individual uses a Smart Walker, he or she is still mainly using his or her legs to walk. Hence, the gait pattern adopted by the user can significantly affect the metabolic cost and stability of the user, as well as his or her long term walking ability. Healthy gait, which is both efficient and stable, depends on certain biomechanical features that exploit the natural dynamics present in bipedal walking. Because actuating a Smart Walkers produces movements and forces that can interfere with these biomechanical features, there is a need to understand how the control of a Smart Walker affects its user’s gait.

More than three decades ago, Mochon & McMahon (1980) and McGeer (1990a, 1990b) demonstrated that a purely mechanical structure, without actuation or control, could walk stably down a shallow slope while displaying surprisingly human-like gait (Figure 1.6). This testifies to bipedal walking as an efficient means of transport that can be achieved using very little energy input (only the small amount of energy gained from gravity when going down a shallow slope was needed in the case of a passive walker). It also suggests that inefficient and unstable walking may be cause by control or actuation characteristics that cancel out the natural dynamics of bipedal walking.

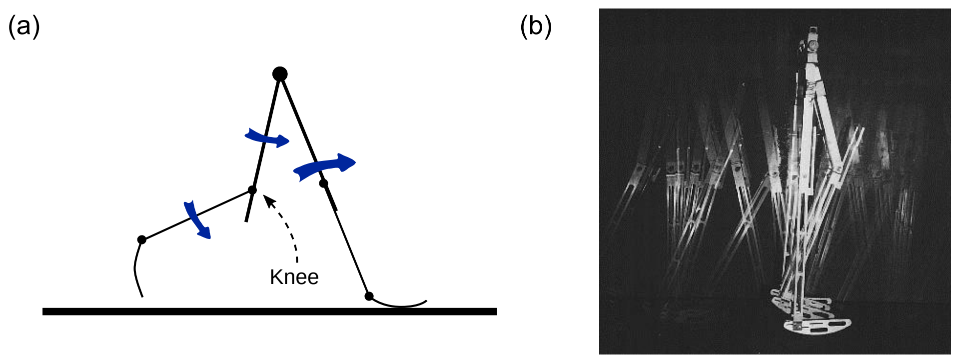


Figure 1.6. Passive walking with knees. (a) Model (Adapted from McGeer (1990b)), (b) Strobe photo by Garcia, Chatterjee & Ruina (1999).

Passive walking is achieved by taking advantage of the pendulum-like natural dynamics of bipedal walking. In the single support phases, which make up around 80% of a gait cycle, the stance leg behaves like an inverted pendulum, pivoting around the ankle joint. The swing leg, meanwhile, swings like a regular double pendulum around the hip. Both these movements conserve mechanical energy; no work is required to move the body's center of mass (CoM) forward in this phase other than the work done due to friction and other dissipative forces as illustrated in Figure 1.7.

Most of the work performed for walking occurs during the double stance phase, when transitioning from one single support step to the next (Donelan, Kram & Kuo, 2002a). Here, the downward velocity of the CoM at the end of the inverted pendular motion of one single support step is redirected upwards to begin the next single support step. The leading leg performs negative work† to slow down and redirect the CoM upwards starting from the heel-contact event. At the same time, the trailing leg performs positive work‡ that generates additional forward velocity while also redirecting the CoM upwards (Figure 1.8).

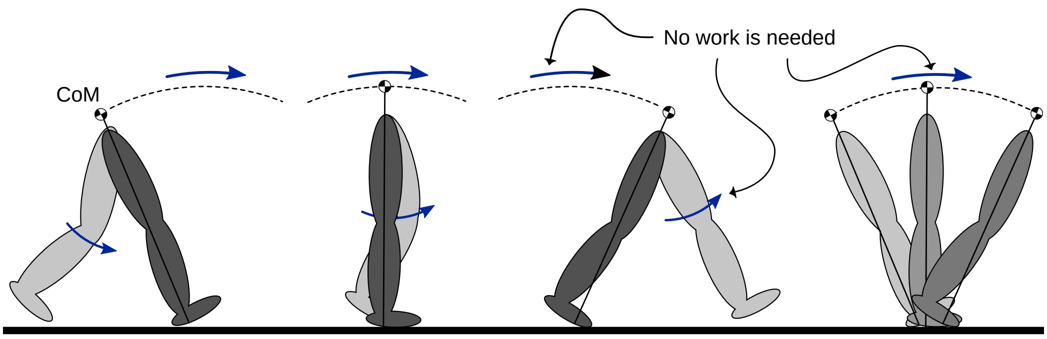


Figure 1.7. Conservation of mechanical energy during the single support phase due to the pendulum-like motions of the stance and swing legs (Adapted from Kuo, Donelan & Ruina (2005)).

If the actuation of a Smart Walker interferes or works against the natural dynamics of walking, the metabolic costs will likely be increased and stability decreased. Furthermore, it may cause further deterioration in the user's walking ability. Brach & VanSwearingen (2013) have argued that one of the major factors for the decline in walking ability in older adults is inefficient walking due to a loss of skill. Extended periods spent walking unnaturally using Smart Walkers can lead

† Negative work: Direction of force is opposite to the direction of displacement (Energy is removed from the system)

‡ Positive work: Direction of force is the same as the direction of displacement (Energy is added from the system)

to loss of ability to utilize momentum for walking and correctly time the generation of muscular forces.

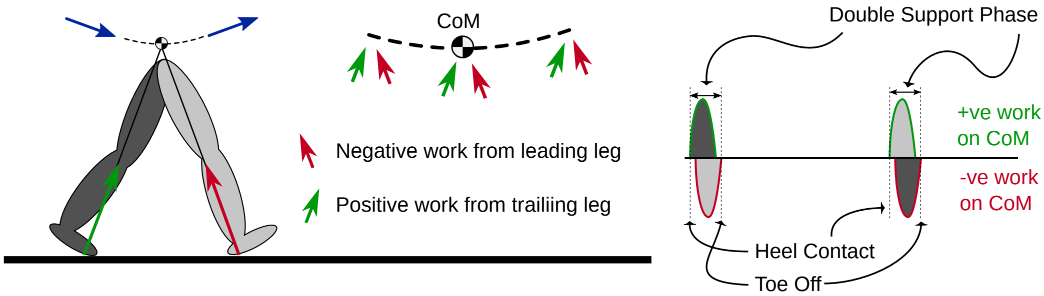


Figure 1.8. Work done by trailing and leading legs to redirect the center of mass (CoM) during the double support phase (Adapted from Kuo, Donelan & Ruina (2005)).

Unlike the cyclic pendulum-like motions of gait, wheeled robots such as Smart Walkers are typically controlled to move at a steady or almost constant speed during steady state. If the external dissipative forces acting on the device does not change, the work rate on its CoM will also remain constant. The user will have to move together with the Smart Walker during use but how he or she adjusts their gait to the difference in motion is still not known. Furthermore, the user may be required to walk at a speed that differs from their preferred walking speed. The effects of these adjustments on the user's skill and ability to leverage the natural dynamics of bipedal gait have yet to be investigated but are crucial to providing effective physical assistance for walking.

Additionally, one consequence of actuating a Smart Walker is the application of an external horizontal force onto the CoM through the device handles and the user's upper limbs. Although bipedal walking can be very efficient, work is nonetheless required to move around and this external force provides a way for the work to be done by the Smart Walker instead of the user. Zirker, Bennett & Abel,

(2013) and Gottschall & Kram (2003) have found that an constant external horizontal force can reduce the metabolic cost of walking by 35% and 47% respectively while results from Dionisio, Hurt & Brown's study (2018) showed that external force can reduce the mechanical effort required to walk faster.

However, these studies were performed with the assistive forces applied close to the participants' CoM (at his or her waist) while the participants walk freely (without bodyweight support). For a Smart Walker, the forces need to be transmitted through its handles and the upper limbs of its user while the user pushes down on the handles for body weight and balance support. This means that the user can have movements or actions that mediate the transmission of forces from a Smart Walker to its CoM and its influence of the user's gait may vary due to amount partial bodyweight support used. For example, the user can pull with different strengths during different phases of the gait cycle, varying the force experienced at the CoM. In addition, the user can push down harder on the Smart Walker to increase the rolling friction and decrease the force experienced. More research is needed to investigate whether the benefits to metabolic cost from assistive force shown in previous studies is still present when applied through a Smart Walker.

1.3.4 Upper Body Exertions

As mentioned in the previous section, assistive forces from a Smart Walker need to be transmitted through the upper body of the user. In other words, the user will need to activate his or her shoulder, elbow, and trunk extensor muscles to pull the Smart Walker and to keep their upper body upright before an assistive force can be transmitted to their CoM (Figure 1.9). This exertion by the upper body may negate the benefits gained from the assistive forces in terms of the mechanical effort required for walking.

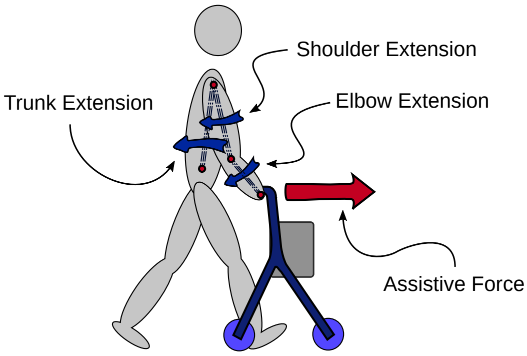


Figure 1.9. Muscle activations in response to assistive forces from a Smart Walker that need to occur in order for the force to be transmitted to the user's CoM.

Furthermore, as discussed in the previous section, there may be upper body adjustments when transmitting the force to the CoM. The user might pull/push with different strengths depending on the gait phase and this can produce additional oscillations in the trunk, shoulders, and elbow when walking as well as in the Smart Walker. This may increase the difficulty to use the Smart Walker when an assistive force is applied.

It has been pointed out that four-wheeled walkers are often used with the incorrect posture; that is, walking in a stooped posture with the four-wheeled walker far in front of the body (van Riel *et al.*, 2014). There is a possibility that these upper body posture adjustments may be used to elicit a better posture from the users,

1.4 Approach & Aim

There are significant ergonomic challenges that need to be overcome before Smart Walkers can be used to safely and effectively improve an individual's mobility. Previous studies have demonstrated that it is possible to integrate sensors that can effectively detect the user's leg position and the forces applied by the user on the handles. In addition, many Smart Walkers developed are capable of identifying obstacles in the environment and even plan paths that can be used to guide a user to their desired destination. This information can then be used to actuate the Smart Walker to assist its user by implementing various features. However, it is still not clear how to actuate the Smart Walker in a way that is beneficial to the user due to the ergonomic challenges presented in the previous section (Page *et al.*, 2016).

Today, researches and developers rely purely on their intuition and experience to conceptualize solutions to these ergonomics challenges. Their solutions are then implemented and evaluated for effectiveness. However, the way to achieve intuitive control and the effects of physical interaction, gait, and handle forces are all tightly linked. This makes it difficult to design or modify solutions that can effectively deal with the ergonomic challenges discussed based only on intuition and experience. Furthermore, there is often a need to adjust or tune implementations to the different needs of individual users. The result of adjusting different parameters for new application without empirical evidence can be difficult to predict. In addition, their efforts may be compromised if their intuition or experience is inaccurate.

Therefore, a human-centered approach to complement the engineering approach described above is needed. This approach focusses on improving our understanding the user's gait adjustments and responses when interacting with a Smart Walker. The results can then be used to ground the intuition of developers with empirical data and provide guidelines and heuristics to help conceptualize solutions to the ergonomics challenges presented.

The aim of this thesis is to improve our understanding of the interaction between a user and a Smart Walker in order to determine how best to provide assistance using its actuators (motorized wheels). As a start, a series of open-loop experiments were performed to describe the gait adjustments to forces from a Smart Walker.

1.4.1 Thesis Outline

Chapter 2 – The Effects of Constant Forces on Gait and Perceived Exertion

To start-off, the simplest case of applying constant forces through a Smart Walker was investigated. I studied the effects of different magnitudes of constant assistive and resistive horizontal forces on the user's gait adjustments and their perceived exertion.

Chapter 3 – The Effects of Speed Control on Gait and Perceived Exertion

Often, the actuators in Smart Walker are not used to produce constant forces but are regulated to achieve a certain desired motion. The simplest instance of this is in controlling the Smart Walker speed during steady-state. In this study, the force from a Smart Walker is controlled using a proportional speed controller to regulate its speed to different target speeds. Similar to the previous chapter, the user's spatiotemporal gait adjustments and perceived exertion were investigated.

Chapter 4 –The Biomechanical Effects of Constant Forces on Gait

While Chapter 2 and 3 provided insights about the adjustments in terms of spatio-temporal gait parameters and perceived exertion, a more detailed analysis of the effect of constant force is provided in this chapter. Building on the previous chapters, the biomechanical effects of different magnitudes of assistive force while walking at different speeds were investigated.

Chapter 2

The Effects of Constant Forces on Gait and Perceived Exertion

2.1 Introduction

Walking is essential to our ability to perform basic everyday tasks and to live independently (Hirvensalo, Rantanen & Heikkinen, 2000; Rosso *et al.*, 2013). However, walking ability is known to decline with age and as many as one in five older adult experiences walking disability (Kraus, 2017; Sagardui-Villamor *et al.*, 2005). Today, older adults who experience walking difficulty often have to resort to walking aids such as four-wheeled walkers (also known as rollators) for assistance (Bradley & Hernandez, 2011; Haines, Brown & Morrison, 2008). These are essentially passive support frames that enable their users to compensate for weaknesses in their lower extremities and deterioration in their postural control with their upper extremities.

As technology progresses, walking aids are increasingly being equipped with sensors and actuators, providing them with various smart features. The most common of which are modified four-wheeled walkers known as Smart Walkers or Robotic Rollators (M. M. Martins, Santos, Frizera-Neto, & Ceres, 2012; M. Martins, Santos, Frizera, & Ceres, 2015; Werner, Ullrich, Geravand, Peer, & Hauer, 2016). These devices are typically equipped with sensors that provide information about the surroundings and the current state of its user. Typical sensors include force or haptic sensors embedded into the handles, laser range finders, wheel encoders and cameras (Martins *et al.*, 2015). While information from these sensors can be used for purely monitoring or assessment purposes (Henry & Aharonson, 2010; Papageorgiou *et al.*, 2016; Chan & Green, 2008), most Smart Walkers are designed to use the information to provide additional supporting functionalities to the user. The typical actuators added to Smart Walkers to implement these features are electric motors (Frizera-Neto *et al.*, 2011) or electromechanical brakes (Hsieh, Young & Ko, 2015) connected to the wheels of the device. These can be used to provide improved physical assistance (Sierra *et al.*, 2018; Sabatini, Genovese & Pacchierotti, 2002; Song & Lin, 2009; Hyeon-Min Shim *et al.*, 2005; Martins *et al.*, 2014b; Geunho Lee *et al.*, 2014), prevent falls (Hirata, Komatsuda & Kosuge, 2008), and provide navigation assistance (Jiang *et al.*, 2017; Sierra *et al.*, 2018; Geravand *et al.*, 2016; Andreetto *et al.*, 2016; Ko *et al.*, 2013).

The utilization of forces generated by actuators connected to the wheels to implement these functionalities has generally been based on the developers' or researchers' intuition and experience of what the user will find helpful. For example, for improved physical assistance, Ohnuma, Lee, & Chong (2014) proposed using a motion controller that keeps the center of the device aligned to the center of the user's body and Shi, Cao, Leng, & Tan (2010) proposed setting the forward velocity or acceleration to be proportional to the handle forces applied by the user. However, as Page, Saint-Bauzel, Rumeau, & Pasqui (2016) acknowledges in their review, there is currently no way of selecting between different strategies and it is unclear how to adjust the implementations to the specific needs of individual users. Despite the potential benefits of adding actuators to the wheels, knowing how best to beneficially apply the force is not straightforward.

Because the forces from the motorized wheels do not directly affect the center of mass of the user, he or she will have to respond and adjusts to these forces, which, in turn can disrupt the other supporting functionalities of the Smart Walker. First, the force is transmitted through the handles of the device to the user's hands. To utilize this force to move his or her body, the user will have to pull or push on the device handles horizontally. However, he or she is simultaneously using the same handles for partial body weight support and balance. This act of pushing or pulling on the handles while using it for support places additional strain on the upper body and can create postural perturbations, which may lead to a loss of balance. This can make the device harder to use and may even contribute to falls. Besides this, the user is still mainly supported by his or her own legs. For the user to move around, he or she is still required to voluntarily move his or her legs to step forward. Likewise, upright stability still depends considerably on the user's ability to adjust the base of support of the legs. Hence, assisting strategies will need to consider the upper body strain and unbalancing effects of the force as well as the gait changes or leg movements responding to or adjusting to the horizontal force.

These suggest that instead of relying purely on the intuition and experience of the developers, a complementary approach based on understanding how users respond or adjust to forces from motorized wheels on a Smart Walker could be beneficial to the development and evaluation of Smart Walkers. To start off, this study aims to investigate a user's preferred speed, gait adjustments, and perceived exertion in the simplest case of applying constant horizontal forces on the steady-state walking of healthy users.

The purpose of this study is to provide answers to basic questions with regards to the effects of magnitude and direction of constant horizontal forces applied from a Smart Walker on its user. Namely: “Does the preferred walking speed change?”, “What are the effects on spatio-temporal gait parameters and gait phases?”, and “What are the effects on the perceived exertion of the user?” I hypothesized that assistive forces will lead to higher walking speeds and vice versa. In addition, the users are expected to walk with higher cadence and lower stride length for both high assistive and resistive forces. The perceived exertion of the user is hypothesized to increase with higher resistive force and decrease with higher assistive forces. Because this study is mainly concerned with the user’s response or adjustments to the force applied, the use of elderly subjects would complicate matters by introducing the effects of aging to the study. Hence, young users were used to isolate the effects of horizontal forces on walking and allow us to focus on answering these basic questions.

2.2 Methods

2.2.1 The Experiment Walker

A custom experiment walker (EW) was developed by modifying a standard four-wheeled walker (Symphony, Shima Seisakusyo, Japan). The original four-wheeled walker weighed 6.4 kg and its width and length measured at 52.5 cm and 55.0 cm. Its handle height was adjustable, with a height range of 77 cm to 87 cm. The main modifications made were the removal of its braking system and the replacement of its rear wheels with a drive wheel assembly. All other additional components were placed in the basket component of the four-wheeled walker (Figure 2.1).

As shown in Figure 2.1, each drive wheel assembly contains an iron core polyurethane drive wheel (UB100, Chubo Sangyo, Nagoya, Japan) connected to a shaft supported by ball bearings whose housing is attached to a custom-made aluminium base plate. This base plate was clamped onto the leg of the four-wheeled walker and was supported by a steel dowel pin pressed into the existing axle hole in the leg. The wheel is driven by a motor supported by the base plate and the shaft of the drive wheel was connected to the motor shaft using a flexible coupling (MJT-40CK-BL-12-18, Nabeya Bi-tech Kaisha, Japan).

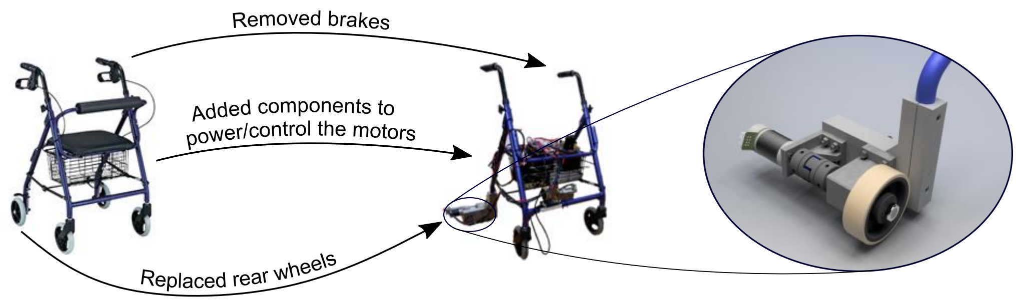


Figure 2.1. Modifications made to the standard four-wheeled walker

The motors used were Maxon EC 45 flat brushless DC motors with integrated encoders and 15:1 planetary gearhead (Maxon Motor, Switzerland). The motors were controlled using a TMS320F28379D Launchpad microcontroller and two BOOSTXL-DRV8301 (Texas Instruments, USA) motor drivers. Power was supplied through a 24 V switched-mode power supply, the RWS600B-24 (TDK, Japan), connected to the mains using extension cables. Additionally, external braking resistors were connected parallel to the power supply and regulated using a MOSFET switch to dissipate the regenerative energy generated when braking.

The brushless DC motors were controlled using a field-oriented control scheme (Toliyat & Campbell, 2003). The angular positions of the motors were measured using its integrated encoders and angular velocity was calculated by numerically differentiating its angular position. The three-phase motor currents were measured using the BOOSTXL-DRV8301 motor drivers. In this control scheme, the motor current is decomposed into two components, one parallel to the motor rotor and one orthogonal to it. The component orthogonal to the rotor is the torque generating component, referred to as i_q . This current component is proportional to torque produced and can be estimated using equation (2.1), where τ is the motor torque generated and k_t is the torque constant of the motor. The motor torque is controlled by regulating motor currents such that i_q is maintained at the desired torque generating level and the component parallel to the rotor, i_d is held at zero.

$$\tau = k_t . t_q \tag{2.1}$$

The developed EW weighed 17 kg and is able to produce horizontal assistive and impeding forces of up to 40 N at an operating speed of 1.5 m s^{-1}. Each drive wheel assembly weighed 4.3 kg and the other additional components weighed 4.7 kg. A pushbutton was used to activate the motors and the motors will deactivate when the pushbutton is released.

2.2.2 Participants

Based on power analysis using G*Power (Faul *et al.*, 2007), to be able to detect a significant difference for a standardized large effect size of 0.4 (Cohen J., 1988) with alpha probability of 0.80 and at a significance level of 0.05, a sample size of eighteen was required. 18 young, healthy, male adults were recruited from the university community to participate in this study. None of the participants had prior experience of using a four-wheeled walker. Due to the limited range of the EW handle height, only people whose wrist height is within the range of the handle height were recruited to participate in this study. The mean age, height and weight of the participants were 24.6 years (SD=2.70), 169 cm (SD=3.2), and 61 kg (SD=7.5) respectively. This study was approved by the Ethics Committee of the Faculty of Design at Kyushu University (Approval No. 249). All participants provided written informed consent before the experiment.

2.2.3 Experimental Conditions and Task

A total of six force level conditions were investigated using a repeated measures experimental design. Specifically, three levels of propulsive force (9.23 N, 18.47 N, and 27.70 N), two levels of braking force (-9.23 N and −18.47 N) and a zero force level were applied. For each condition, both motors were set to maintain the same constant torque output based on equation (2.3) throughout the trial as the participant walks using the EW. The force stated here refers to the force generated by the motor to 'push' or 'pull' the EW and does not take into account the friction and momentum of the EW. The order in which the participants performed the trials for each condition was randomized.

The participants were instructed to walk along a 17 m long straight and level path at a self-selected comfortable walking speed for all conditions. A straight line floor marking was drawn to help guide the participants. In addition, the participants were asked to try their best to maintain an upright posture and place 30% of their body weight on the EW to simulate partial weight-bearing when using the EW.

2.2.4 Experiment Protocol

The height of the EW handles was adjusted to the level of the participant's wrist, similar to previous studies involving wheeled-walkers (Alkjaer *et al.*, 2006; Liu *et al.*, 2009; Tung *et al.*, 2011; Schülein *et al.*, 2017; Suica *et al.*, 2016) and recommended for proper use (Bradley & Hernandez, 2011). To reduce the influence of clothing and footwear, the participants wore the same type of loose-fitting clothing and walking shoes prepared by the experimenter. Footswitches were attached to the toe and heel part on the sole of each shoe to identify gait events. These footswitches were connected to the microcontroller in the EW with electrical wires. As shown in Figure 2.2, these electrical wires were attached to the body of the participant up to his waist using adhesive tapes before the final connection to the microcontroller on the EW. This connection remained slack throughout to prevent restrictions to the motion of the participant.

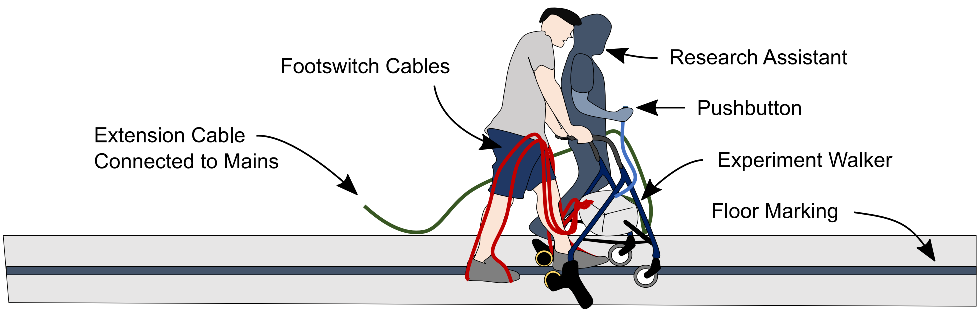


Figure 2.2. Experimental setup.

To begin, a training session was held during which each participant learned the desired 'push down' force for partial weight-bearing using the EW for this experiment. The participant stood on a weighing scale as he adjusted the 'push down' force on the EW to achieve the desired target. Training continued until the participant was able to apply 30% of his body weight on the EW without feedback.

Throughout the experiment, a research assistant would walk alongside the participant while the EW was being used. He or she ensured that the participant did not fall when using the EW, was responsible for activating the motors using the pushbutton, and making sure the extension cable did not impede the motion of the participant or of the EW.

2.2.5 Measurement

Immediately following the completion of each trial, the participants were asked to rate their perceived exertion according to Borg's RPE scale (Borg, 1990) during steady-state walking. The angular velocity and torque for both motors together with signals from the footswitches were sampled at 100 Hz using the microcontroller on the EW and transferred wirelessly to a laptop for recording through the local area network. Three successful trials were performed for each condition. For each trial, four consecutive steady-state gait cycles were chosen for further analysis.

The forward velocity of the EW, v_{EW} was calculated using equation (2.2) where r is the drive wheel radius, G is the gear ratio, and ω_l and ω_r are the 'forward' angular velocities of the left and right motors.

$$v_{EW} = \frac{r}{2G}(\omega_l + \omega_r) \tag{2.2}$$

The horizontal forward force generated by the EW, F is calculated using equation (2.3) where τ_l and τ_r are the 'forward' torques generated by the left and right motors.

$$F = \frac{G}{r}(\tau_l + \tau_r) \tag{2.3}$$

The rising edge of the heel footswitches and the falling edge of the toe footswitches were used to identify gait-partitioning events. Single support time (s), swing time (s), and double support time (s) per gait cycle were calculated using the timing of these events. The right heel-contact event was used to separate gait cycles.

Because the participants and the EW moved forward together, the average walking speed (m s^{-1}) is assumed to be the average forward speed of the EW. Cadence (steps min^{-1}) was calculated using the time between successive right heel-contact events (stride period). Average stride length was calculated by dividing the walking speed by cadence. Additionally, walk ratios of the participants were calculated as the ratio between their step length and step cadence.

To determine if the parameters were affected by the addition of horizontal EW forces, one-way repeated measures analysis of variance (ANOVA) tests were performed. If a significant effect was found, polynomial contrasts were used to identify the trends present in the parameter. In addition, Dunnett's tests were used to compare the parameters at zero force condition (0 N) with that of the others. The significance level used to assess the statistical tests was set at $p < .05$. All statistical analysis was performed using IBM SPSS 25 software.

2.3 Results

The results are presented in detail in Table 2.1. Walking speed was significantly affected by the different horizontal force applied ($F [5, 85] = 38.54$, $p < .05$). In addition, polynomial contrasts analysis revealed significant linear and quadratic components. Dunnett's tests showed significant differences in selected walking speed for comparisons between 0 N and the other force conditions except for that between 0 N and 9.23 N (Figure 2.3a). For the gait parameters that were investigated, significant overall effects of horizontal force ($p < .05$) were found for cadence ($F [5, 85] = 27.00$), stride length ($F [5, 85] = 21.75$), and double support phase (F [3.29, 55.88] = 12.93). An analysis of the polynomial contrasts showed significant linear components for all of the gait parameters that were significantly affected by the horizontal force. However, significant quadratic components were only present for stride length (Figure 2.3c, d). Even though a significant quadratic trend was present for stride length but not for cadence, no significant overall effect was found for walk ratio (Figure 2.3b).

Dunnett's tests showed no significant difference between 0 N and 9.23 N for all gait parameters. Cadence was significantly different from 0 N for all other conditions. However, only relatively high assistive forces (18.47 N and 27.70 N) produced significantly different stride lengths compared to stride length at 0 N (Figure 2.3b). Additionally, the tests also indicated that the effects of assistive and resistive forces were different for the gait phases. Only double support time was significantly different from 0 N for resistive forces. Whereas, all of single support, swing and double support time for 27.70 N was significantly different from that at 0 N (Figure 2.4).

The different horizontal forces had a significant overall effect on the ratings of perceived exertion (F [2.72, 46.21] = 7.21, $p < .05$). A significant quadratic trend present in the effect indicated that both high assistive and resistive forces increased the perceived effort required to walk using the Experiment Walker (Figure 2.3f). Although the rating for 9.23 N was not significantly different to that for 0 N, the mean rating was lower. In fact, the rating for 9.23 N was the lowest among all of the conditions.

Table 2.1. Walking speed, gait parameters and ratings of perceived exertion under various force conditions.

	Force Settings (N)						
	-18.47 N	-9.23 N	0.00 N	9.23 N	18.47 N	27.70 N	*p* <
Walking Speed (m s^{-1})	0.93 ± 0.14	0.96 ± 0.13	1.05 ± 0.15	1.04 ± 0.17	1.20 ± 0.15	1.30 ± 0.15	.001
Cadence (steps min^{-1})	92.4 ± 11.0	95.6 ± 10.6	100.7 ± 9.1	99.9 ± 10.8	107.2 ± 8.0	111.7 ± 8.5	.001
Stride Length (m)	1.20 ± 0.10	1.21 ± 0.08	1.25 ± 0.11	1.24 ± 0.11	1.34 ± 0.12	1.40 ± 0.12	.001
Walk Ratio (mm / (steps min^{-1}))	6.58 ± 0.88	6.38 ± 0.81	6.26 ± 0.65	6.27 ± 0.64	6.28 ± 0.69	6.30 ± 0.66	.520
Double Support Phase (%)	33.5 ± 4.8	31.5 ± 4.6	30.4 ± 4.8	29.7 ± 4.3	28.5 ± 5.5	28.3 ± 5.2	.001
Double Support Time (s)	0.44 ± 0.08	0.40 ± 0.06	0.37 ± 0.06	0.36 ± 0.08	0.32 ± 0.05	0.30 ± 0.05	.001
Single Support Time (s)	0.44 ± 0.07	0.43 ± 0.06	0.42 ± 0.05	0.43 ± 0.06	0.41 ± 0.05	0.39 ± 0.05	.128
Swing Time (s)	0.44 ± 0.07	0.44 ± 0.07	0.42 ± 0.05	0.43 ± 0.07	0.40 ± 0.05	0.39 ± 0.05	.042
Borg Rating (6-20)	12.1 ± 2.9	10.6 ± 2.0	10.1 ± 2.4	8.7 ± 1.7	10.2 ± 2.0	10.6 ± 2.2	.004

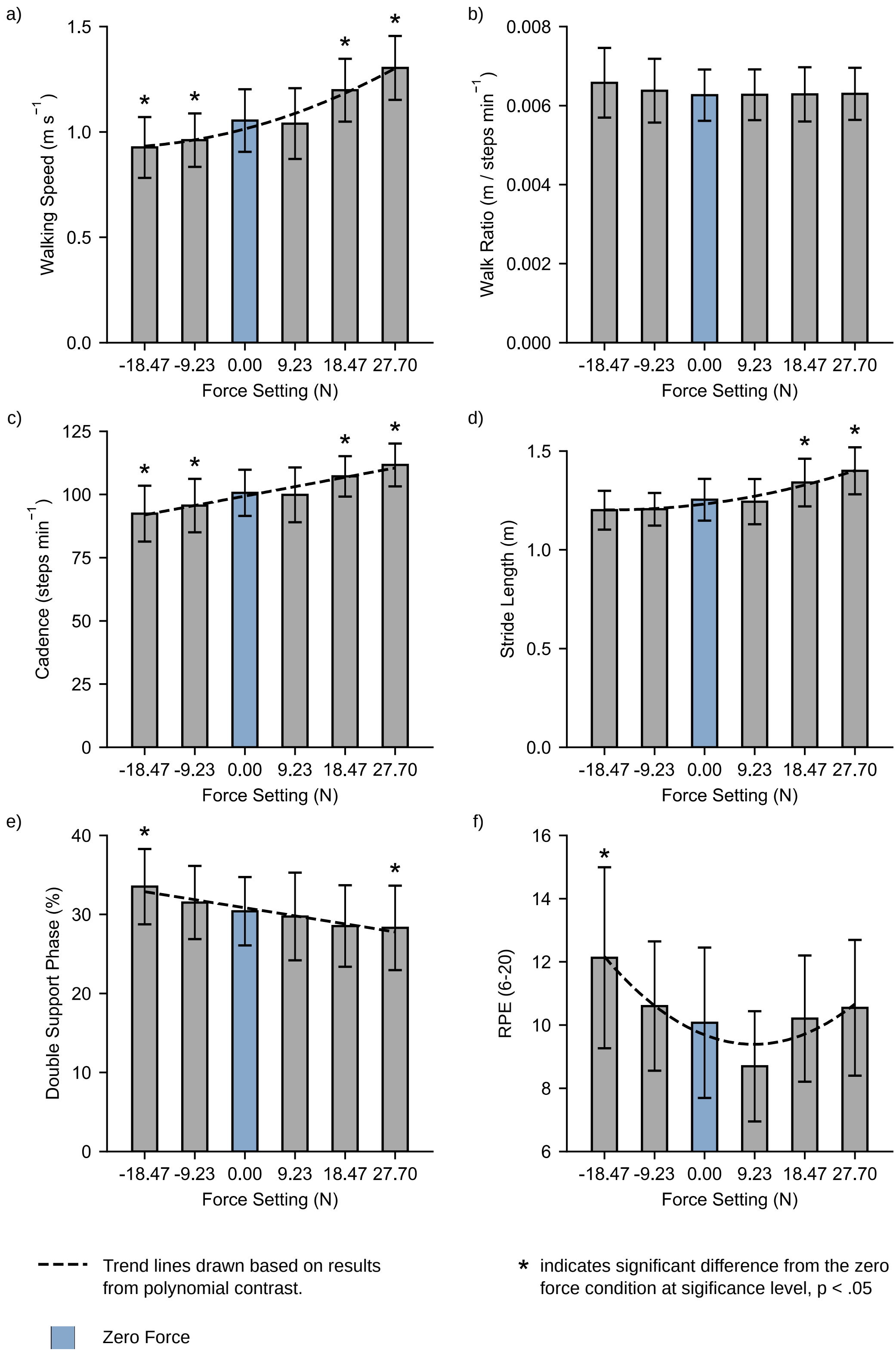


Figure 2.3. The bar charts show the mean values (± standard deviation, n = 18) of (a) self-selected comfortable walking speed, (b) walk ratio, (c) cadence, (d) stride length, (e) double support phase, and (f) ratings of perceived exertion (RPE).

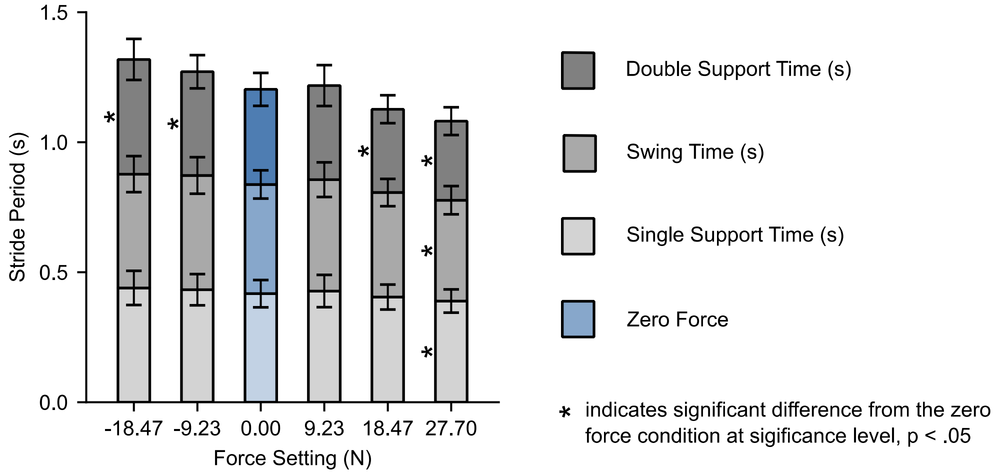


Figure 2.4. The stacked bar chart illustrates the distribution of single support time, swing time, and double support time in the stride period (mean ± standard deviation, n = 18).

Based on the trend lines and comparisons between 0 N and the other forces, I could discern three distinct ranges in which the gait responses were different, and these ranges corresponded to the trend in the ratings of perceived exertion. Namely, (1) from −18.47 N to 0 N, where increasing resistive force corresponded to an increase in perceived exertion; (2) from 0 N to 9.23 N, where exertion initially decreased with the presence of the assistive force; and (3) from 9.23 N to 27.70 N, where further increasing the assistive force to a rise in perceived exertion.

2.4 Discussion

I investigated the effects of the magnitude of assistive and resistive horizontal forces from a Smart Walker on the preferred walking speed, gait and perceived exertion of the user. Significant overall effect on self-selected walking speed, all gait parameters except walk ratio, and the reported ratings of perceived exertion were found. However, the trends in the data suggest that different effects were present for the three ranges of (1) resistive forces, (2) low assistive forces, and

(3) high assistive forces identified in the results section. Because different mechanisms may be at play in these three distinct ranges, they will be discussed separately in this section.

2.4.1 Resistive Force

In range (1), as resistive force increased from 0 N to −18.47 N, walking speed decreased significantly from 1.05 m s^{-1} to 0.93 m s^{-1}. The gait parameters measured suggest that this change in walking speed, however, was largely due to an increase in double support time. With no significant decrease in stride length, the majority of the effect on walking speed can be attributed to the decrease in cadence. The decrease in cadence (or increase in stride period) can, in turn, be traced to the increase in double support time from 0.37 s to 0.44 s as shown in Figure 2.4. Only double support time was significantly different from 0 N in range (1).

The increase in double support time here suggests adjustments were made by the user to generate additional pushing force. The resultant force from a person pushing while walking has previously been shown to oscillate in tandem with the gait cycle, suggesting that the ability to generate a propulsive force varies depending on the gait phase (De Looze *et al.*, 2000). Following from that, a previous study has indicated that the average pushing force can be increased without increasing the force applied by each leg while walking by extending the double support phase of walking (Suzuki & Uchiyama, 2009). The user might have altered his or her gait in order to minimize the effort required to push the EW. Studies into the effects of walking up an incline (working against the resistive force of gravity) on gait also provide supporting evidence for this. Similar to the effect of resistive force, walking speed and cadence decrease while the percentage time in stance phase increases when ascending an incline (Kawamura, Tokuhiro & Takechi, 1991; Sun *et al.*, 1996).

The increase in perceived exertion with increasing resistive force in range (1) was as expected. The user will have to generate more force to overcome a higher resistive force from the EW. This is in agreement with a study by Gottschall & Kram (2003) that showed that increasing the impeding force applied at the waist of the participant increased the metabolic cost of walking. Additionally, the upper body strain from the user having to push harder may have also contributed to the higher perceived exertion found.

2.4.2 Low Assistive Forces

In range (2), the self-selected walking speed was not significantly different from 0 N when the force is relatively low at 9.23 N. In addition, there were no significant changes in gait parameters. However, the average perceived exertion of the users decreased from 10.1 to 8.7 and the minimum point of the trendline is around 9.23 N.

The lower exertion at 9.23 N may be due to the lower force required to push the EW. That is, the assistive force is not sufficiently high to overcome the frictional resistance, but the user is able to more easily push the EW forwards. However, while the effect of this should be similar to reducing the resistive force applied, the gait adjustments observed in range (1), such as increasing cadence and walking speed with decreasing resistive force, does not occur here.

Another possibility is that the 9.23 N assistive force provided was more than that required to move the EW, and the user experiences a small pulling force from the handles. The users may have been able to utilize the pulling force to reduce the positive work required for walking, decreasing their exertion. Human walking involves both the dissipation of mechanical energy (negative or eccentric work done when decelerating at the start of the stance phase) and the generation of mechanical energy (positive or concentric work done to accelerate the center of mass towards the end of the stance phase). While the need to pull the EW may strain the upper body of the user and increase the negative work of walking, the user can use it to offset some of the positive work of walking. Because the metabolic cost of negative work is less than that of positive work (Abbott, Bigland & Ritchie, 1952), it has been shown that assistive forces applied at the waist can decrease the metabolic cost of walking up to a point (Gottschall & Kram, 2003; Zirker, Bennett & Abel, 2013)

Unlike as hypothesized, the ratings of perceived exertion began to rise with increasing assistive force from around 9.23 N (~1.5% bodyweight). This is not consistent with previous studies where assistive forces up to 8% of their bodyweights were found to decrease the metabolic cost required for walking (Gottschall & Kram, 2003; Zirker, Bennett & Abel, 2013). However, in these experiments, the assistive force was applied at the waist, which is close to the center of mass of the participants. Although the experimental conditions differed and only a subjective measure of exertion used in this study, the large difference in maximum beneficial assistive force (1.5% and 8%) suggests that other factors may be involved.

This may be due to the indirect application of the force through the handles of the EW. Firstly, the users will need to activate their extensor muscles of the shoulder and elbow joints to pull the EW and activate their trunk extensor muscles to keep their upper body upright. While the assistive force may have reduced the user's effort to walk, this is negated by the effort exerted by the upper body of the user. Secondly, the user is pushing down on the handles for partial bodyweight support and balance. The force applied through the handles may have made it more difficult for the users to place their weight on the EW. Thirdly, the EW may have become more difficult to use or control due to the assistive force. It has been shown that the downward force applied at each handle naturally oscillates as the user walks with a wheeled walker, which changes the rolling friction on each side (Abellanas *et al.*, 2010; Alwan *et al.*, 2007). Because equal torque was provided by each motor in the EW, a resulting moment may inadvertently have been generated. The need for the user to counteract this moment and prevent the EW from turning may have also contributed to an increase in perceived exertion.

2.4.3 High Assistive Forces

In range (3), the users walked at significantly higher speeds when higher assistive forces were applied. The users preferred to walk at 1.30 m s^{-1} when a 27.70 N assistive force was applied compared to 1.05 m s^{-1} at 0 N. A previous study has demonstrated that an assistive force applied at the waist reduces the mechanical effort required to walk faster (Dionisio, Hurt & Brown, 2018). In addition, another study suggested that while a comfortable pushing force decrease with increasing walking speed, a comfortable pulling force does not seem to be affected by walking speed (Suzuki *et al.*, 2015). These support the view that the users chose to walk at a higher speed because it minimized their effort despite the higher rating of perceived exertion reported.

Increasing assistive force in range (3) also led to higher cadence, stride length, double support time and a lower swing phase. Nonetheless, in free walking, when people with healthy gait increase their speed, they have been found to exhibit similar changes in their gait (Kirtley, Whittle & Jefferson, 1985; Latt *et al.*, 2008; Sekiya & Nagasaki, 1998; van Hedel, Tomatis & Müller, 2006). These suggest that the user did not increase their walking speed in an unusual matter such as by increasing cadence but maintaining the same stride length.

The rating of perceived exertion increased as the magnitude of assistive force increased. As discussed earlier, this may be due to the need to pull the EW and the increased difficulty in using the EW when higher assistive forces are applied.

2.4.4 Overall

In general, the participants chose to walk at higher speeds as the applied force increases from −18.47 N to 27.70 N except at 9.23 N where walking speed remained the same. A similar pattern was observed in the gait parameters when significant linear components are present but no significant difference was observed between 0 N and 9.23 N. In addition, it is only at this level of force that the perceived exertion of the users is lower than 0 N. These suggest that when a low assistive force is applied, the users walk with a similar gait pattern to using a four-wheeled walker but with lower physical exertion. If the developer aims to use Smart Walkers to alter the preferred walking speed and spatio-temporal gait parameters of its users, a resistive force or a relatively high assistive force is required.

2.4.5 Limitations

I did not measure the interaction forces between the user and the EW in this experiment. Hence, although the participants were instructed to place 30% of their bodyweight on the EW as they walked using it, there was no way to verify the downward force applied on the handles. In addition, I was not able to determine the pulling and pushing forces experienced by the user from the EW handles. The participants were free to select their walking speed in this experiment. While this allowed us to identify the comfortable walking speed for each force setting, it also made it difficult to compare the effects on the other gait parameters because walking speed significantly alters the gait of a person.

2.5 Conclusion

In conclusion, different ranges of horizontal force from the EW were found to lead to different outcomes. For resistive forces, the higher the magnitude of the force applied, the higher the perceived exertion and the higher the double stance period when walking, leading to lower cadence and walking speed. When the assistive force is relatively high (above 9.23 N), increasing force magnitude also increases perceived exertion but the participants chose to walk at higher speeds at higher force magnitudes. However, when only a small assistive force is applied, the participants' gait did not change while perceived exertion decreased. Of all the conditions investigated, only one showed a reduction in perceived exertion relative to the zero-force condition. This validated our view that applying horizontal forces for assistance is not straightforward as well as showing the inappropriateness of the simple constant force used in this experiment in real-world applications.

Chapter 3 The Effects of Speed Control on Gait and Perceived Exertion

The Effects of Speed Control on Gait and Perceived Exertion

3.1 Introduction

Maintaining our ability to walk as we age is increasingly important in today's aging world. Around 20% of older adults have been reported to experience walking difficulty (Kraus, 2017; Sagardui-Villamor *et al.*, 2005), and preferred walking speeds, on average, have been found to decline sharply from around age 65 (Ferrucci *et al.*, 2016; Schrack *et al.*, 2012). In recent years, four-wheeled walkers equipped with electric motors are increasingly used to help address this problem (Kawamura Cycle LTD, 2020; RT.WORKS co., 2020; Robot Care System, 2020; Kowa co., 2020; BEMOTEC GmbH, 2020). Although the assistance provided by these electric motors is currently limited (i.e., only providing assistance in challenging conditions, such as on slopes), the scope of utilizing motorized wheels for better assistance is expected to expand to include dynamic assistance in normal situations. This will likely alter the way we interact with the devices as demonstrated by the recent research on the use of smart walkers or robotic rollators (Frizera-Neto *et al.*, 2011; Hsieh, Young & Ko, 2015; Geunho Lee *et al.*, 2014; Martins *et al.*, 2014c; Sierra *et al.*, 2018; Song & Lin, 2009; Andreetto *et al.*, 2016; Geravand *et al.*, 2016; Jiang *et al.*, 2017; Ko *et al.*, 2013; Fei Shi *et al.*, 2010).

In Chapter 2, the effect of different magnitudes of constant force applied through a Smart Walker on the walking speed, gait patterns, and perceived exertion of the user were investigated. It was found that while a small assistive force can reduce the perceived exertion during steady-state walking, applying a relatively high assistive or resistive force would increase perceived exertion. This is likely due to the need to continuously produce pulling or pushing force to respond to the constant assistive force applied by the EW. In situations where the assistance provided is variable, users may not have to respond continuously to the force applied, and this may subsequently decrease the exertion experienced by them.

A common and basic method to provide variable assistance through a Smart Walker is by moving it without the user pushing it. The simplest case of moving forward steadily can be done by having the walker move forward at a fixed speed. However, this will likely alter the gait pattern and perceived exertion of a user while walking. Commercial Smart Walkers usually have settings that allow its users to adjust the speed of the walkers. Previous research has shown that it is possible to infer the desired speed of the user by sensing his or her leg movements. However, this inferred value may not be perfectly accurate, and understanding the effects of this deviation on the actual desired walking speed of the users is also important. In addition, an arbitrary speed limit is often applied to walkers for safety reasons, and this speed may be lower than the desired walking speed of the users. Hence, this study aims to investigate the effects of different Smart Walker speed settings on the gait patterns and perceived exertion of the users.

3.2 Methods

3.2.1 Participants

Twenty young healthy male subjects participated in this study (age = 24.3 ± 2.76 years, height = 1.69 ± 0.03 m, weight = 60.3 ± 7.40 kg). None of the participants were familiar with the use of wheeled walkers, and they provided written informed consent prior to the experiment. The experimental protocols were approved by the Ethics Committee of the Faculty of Design at Kyushu University (Approval No. 249). Healthy young male participants were selected for this experiment to allow the researchers to focus first on the effects of the use of a Smart Walker on walking before including the effects of age and various impairments.

3.2.2 Experiment Walker

A pair of Maxon EC 45 flat brushless DC motors with integrated planetary gearheads (GP42 C, 15:1) and encoders (MILE) were added to a standard four-wheeled walker (Symphony, Shima Seisakusyo, Japan) to control the speed of the device. This was done by replacing the rear wheels of the walker with drive-wheel assemblies (wheel diameter = 0.1 m) that enabled connection with the motors, as shown in Figure 3.1. The motors were connected to the mains using an extension cord, and all components used to control the motors were placed in the basket compartment of the walker. The resultant Experiment Walker (EW) weighed 17 kg.

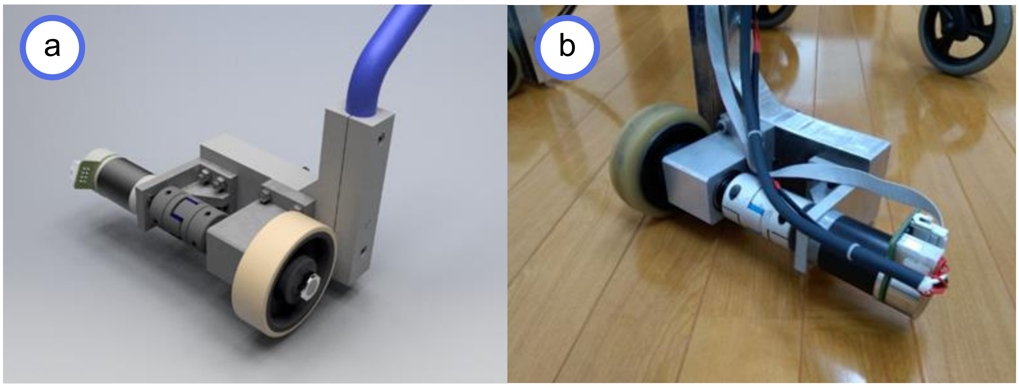


Figure 3.1. Drive-wheel assembly used to replace the rear wheels: a) rendered and b) photo.

A field-oriented control scheme was used to independently control the speed of each motor. The integrated encoders were used to measure rotor speed and rotor position. Figure 3.2 shows the comparison of the measured rotor speed and the desired rotor speed. The motor phase currents were regulated based on the speed difference to produce the desired wheel torque. The generated torque was estimated using the phase currents measured. Proportional control with gain 0.557 N (rad s^{-1})$^{-1}$ and saturation limits at 0.693 N m (equivalent to a total force of 27.7 N for both wheels) were used in this study to control the rotational speed of the wheels. Proportional control was selected due to its simplicity and predictability.

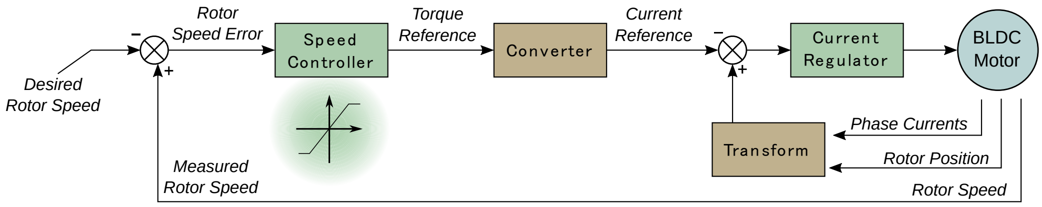


Figure 3.2. Control loop used to control the brushless motors.

3.2.3 Experimental Conditions

The desired rotor speed that corresponded to the desired forward speed was set for each experimental condition. Six experimental conditions were investigated in this study: five speed settings evenly spaced between 0.6 m s^{-1} and 1.4 m s^{-1}, and one “No Assist” control condition where the motors were not used to produce any assistive force. The torque–angular velocity characteristic for each speed setting is shown in Figure 3.3.

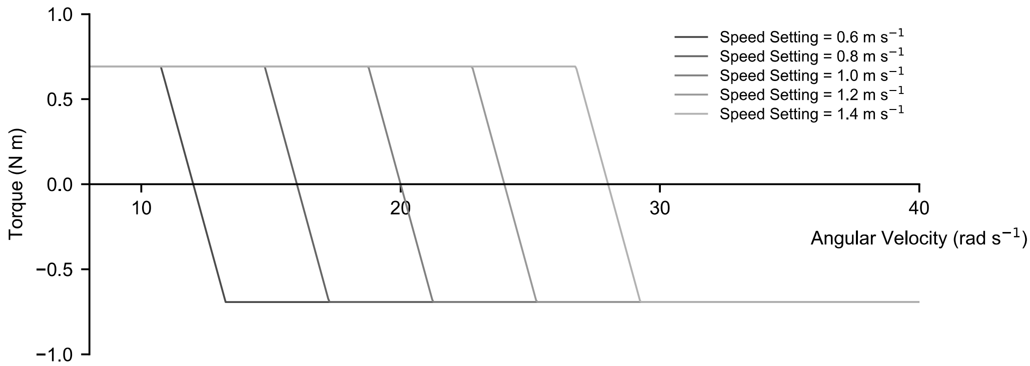


Figure 3.3. Torque–angular velocity for different conditions.

3.2.4 Experimental Task

The participants were instructed to walk at a self-selected, comfortable walking speed in a straight line for 17 m while placing around 30% of their bodyweight on the EW. Weight-bearing training using the EW was performed before the start of the experiment where the participants had to learn how hard to push down on the EW while walking. In addition, the participants performed two practice trials for each condition to familiarize themselves with the speed settings. At least three successful trials were performed for each condition. The experimental conditions were randomly applied.

3.2.5 Measurement

Four footswitches were attached to the toe and heel parts of the participants' shoes, as shown in Figure 3.4. The angular velocity of the drive wheels was calculated using the integrated encoders in the brushless motors, and the torque generated by the motors was estimated using the measured phase currents. The footswitch signal, the wheel angular velocities, and motor torques were sampled at 100 Hz. Four consecutive steady-state gait cycles were selected for further analysis.

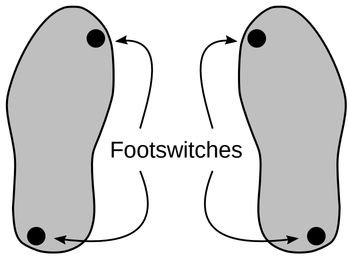


Figure 3.4. Positions of the footswitches.

The right-heel contact footswitch was used to calculate cadence (steps min^{-1}) and to separate the gait cycles. The mean walking speed (m s^{-1}) was assumed to be the same as the mean EW speed, and it was calculated from the mean rotational velocity of the drive wheels. The assistive force provided by the EW was calculated as the sum of the force generated by each drive wheel. The mean force (N) was calculated from the assistive force applied during the four consecutive steady-state gait cycles.

Mean stride length (m) was calculated from the mean walking speed and cadence. The participants were asked to report their rates of perceived exertion (RPE) based on Borg's RPE (Borg, 1990) scale during steady-state walking immediately after every trial.

Friedman's analysis of variance tests were performed to determine if a measured parameter was significantly affected by the different speed settings. Wilcoxon signed-rank tests were used to compare walking speed, mean force, RPE, cadence, stride length, and gait phases for the "No Assist" condition with different speed settings. The familywise error rates for the Wilcoxon signed-rank tests were controlled using Bonferroni corrections. All statistical analyses were performed using IBM SPSS 25 at $p < .05$ significance level.

3.3 Results

The different speed settings significantly affected the walking speed selected by the participants ($\chi_F^2(4) = 73.96$, $p < .001$). Except for when the speed setting was set at 1.00 m s^{-1}, the Wilcoxon signed-rank test found that walking speed was significantly different from that in the "No Assist" condition. The selected walking speed was similar to the speed setting when the speed was set close to the "No Assist" speed (1.05 m s^{-1}), specifically at 0.8 m s^{-1}, 1.0 m s^{-1}, and 1.2 m s^{-1}. However, when the speed setting was set very low or very high at 0.60 m s^{-1} and 1.40 m s^{-1}, the participants chose to walk faster at 0.68 m s^{-1} and slower at 1.26 m s^{-1} (Figure 3.5a).

The speed setting also had a significant effect on the mean force applied by the EW ($\chi_F^2(4) = 71.88$, $p < .001$). Except for the 1.00 m s^{-1} speed setting, the force applied was significantly different from the "No Assist" condition. In addition, the mean force applied by the EW increased from −15 N to 24 N as the speed increased from 0.60 m s^{-1} to 1.4 m s^{-1} (Figure 3.5b).

The ratings of perceived exertion reported by the participants were significantly affected by the speed setting ($\chi_F^2(4) = 11.70$, $p = .020$). Although the Wilcoxon signed-rank tests did not find any significant difference between the perceived exertion for the "No Assist" condition and those for the different speed setting conditions, only the 1.00 m s^{-1} speed setting showed a decrease from the "No Assist" condition. The ratings of perceived exertion seemed to follow a quadratic pattern with the minimum point around the 1.00 m s^{-1} speed setting, indicating that speed settings higher and lower than 1.00 m s^{-1} increased the perceived exertion when using the walker (Figure 3.5e).

Significant overall effects were found for stride length ($\chi_F^2(4) = 72.12$, $p < .001$) and cadence ($\chi_F^2(4) = 71.60$, $p < .001$) (Figure 3.5c, d). The double support phase ($\chi_F^2(4) = 45.60$, $p < .001$), the swing phase ($\chi_F^2(4) = 46.28$, $p < .001$), and the single support phase ($\chi_F^2(4) = 33.52$, $p < .001$) were all significantly affected by the speed setting. However, the swing phase was only significantly different from the "No Assist" condition when the speed was set to 0.60 m s^{-1} and 0.80 m s^{-1}, while the single and the double support phase were only significantly different when the speed was set to 0.60 m s^{-1} (Figure 3.6).

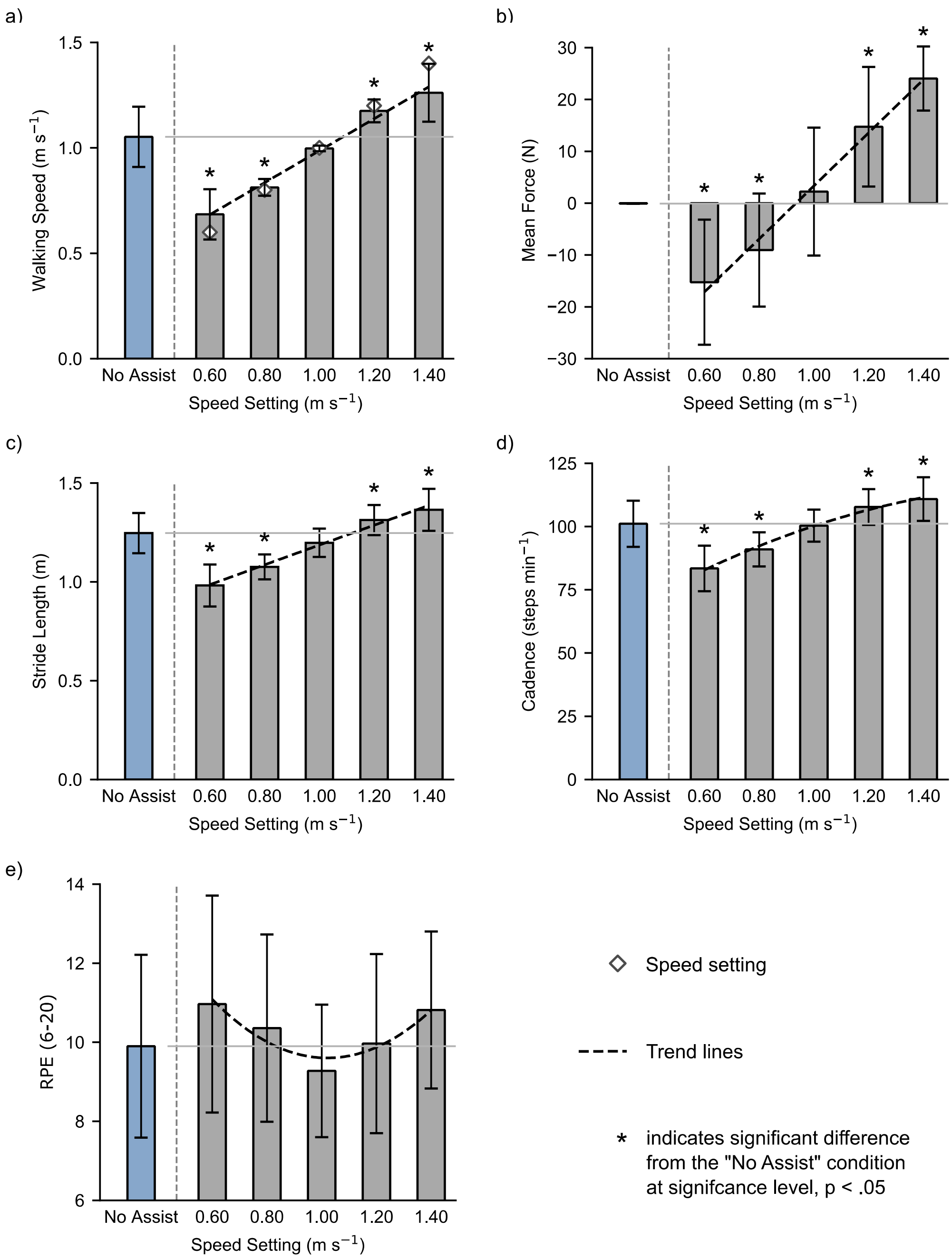


Figure 3.5. Mean values (± standard deviation, n = 20) of a) self-selected walking speed, b) mean force applied by the EW, (c) stride length, (d) cadence, and (e) rating of perceived exertion (RPE).

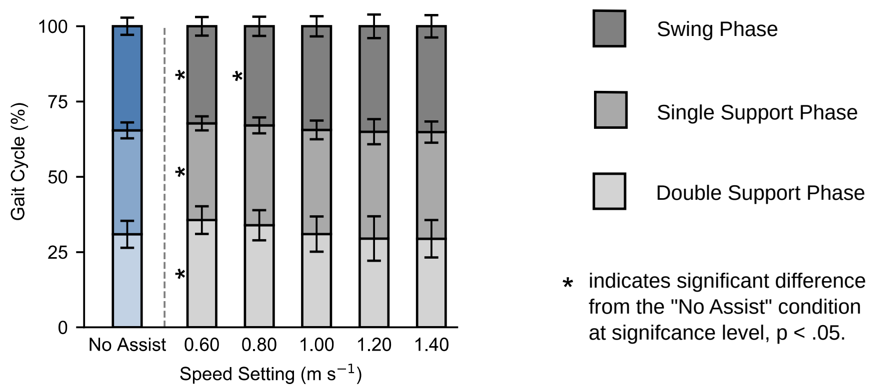


Figure 3.6. Gait phases as a percentage of gait cycle (mean values ± standard deviation, n = 20).

However, looking more closely at the data, there are a large number of instances (76%) where participants matched the speed setting of the EW. This suggests that data points could be classified into compliant and non-compliant instances. The percentage difference in the walking speed from the set speed was calculated using equation (3.1). If a percentage difference was less than ±5% a data point was classified as compliant and vice versa (Figure 3.7a).

$$\%\,\Delta\left(v, v_{Setting}\right) = \frac{v - v_{Setting}}{v_{Setting}} \tag{3.1}$$

Figure 3.7b shows that the further away a setting was from the mean of the "No Assist" speed, the more participants walked at speeds different from those set for the EW.

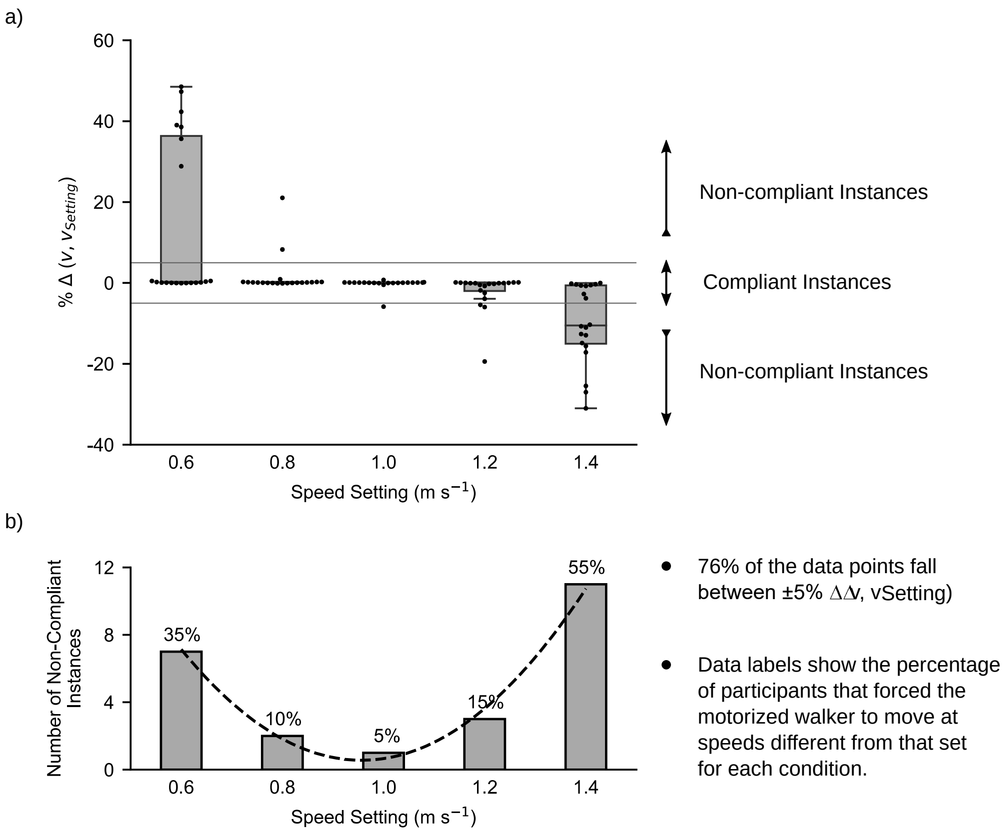


Figure 3.7. Classification into compliant and non-compliant instances (a) A swarm plot (showing individual data points) overlaid over a box-and-whiskers plot of the percentage difference of the walking speed from each speed setting and (b) number of non-compliant instances for each speed setting.

In addition, the data suggest that the difference between the "No Assist" speed and the selected walking speed plays an important role in the resultant force generated by the EW and the perceived exertion of the user. Hence, equation (3.2) was used to calculate the percentage difference from the "No Assist" speed. Figure 3.8 shows the relationships between the percentage difference, the mean force, and ratings of perceived exertion for both compliant and non-compliant instances.

$$\% \, \Delta \, (v, v_{NoAssist}) = \frac{v - v_{NoAssist}}{v_{NoAssist}} \tag{3.2}$$

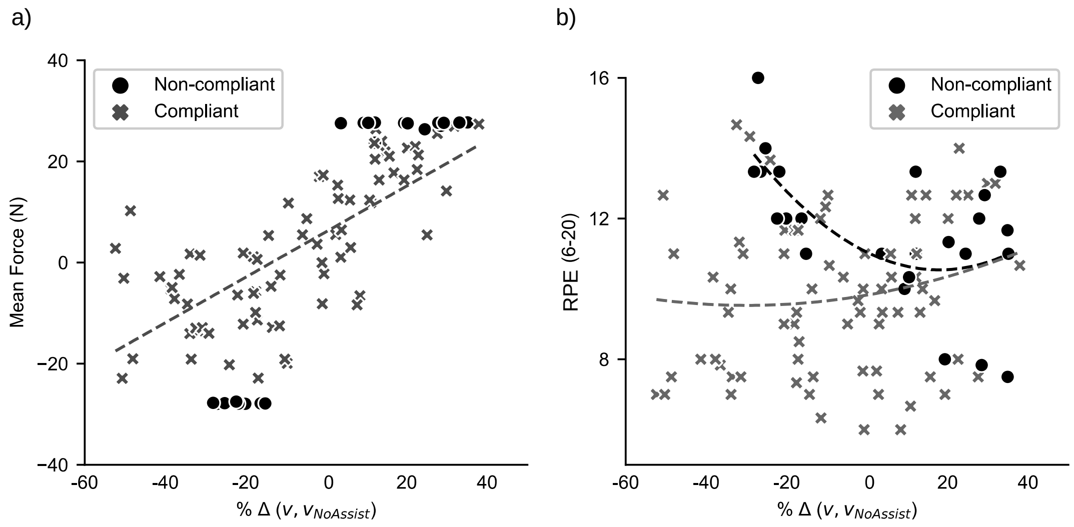


Figure 3.8. Compliant and non-compliant instances for (a) mean force and (b) rating of perceived exertion (RPE) against the percentage difference of the walking speed from the "No Assist" speed of each participant.

In non-compliant instances, participants had to overpower the EW and therefore experienced the maximum force that the EW could apply. However, in compliant instances, the forces experienced appeared to vary linearly with the percentage difference from the "No Assist" speed.

The ratings of perceived exertion for non-compliant instances followed a parabolic pattern with the minimum point occurring around 18%. Conversely, for compliant instances, the percentage difference from the "No Assist" speed only showed a small increasing trend after 0%.

3.4 Discussion

It has long been known that the energy efficiency–speed relationship for human walking is parabolic, and there is an optimal speed where the energy expenditure for a given distance travelled is minimized (Ralston, 1958; Martin, Rothstein & Larish, 1992). When a person is asked to walk at a "comfortable" speed, a walking speed near that minimum point is typically selected (Ralston, 1958; Browning *et al.*, 2006). In this experiment, the participants were instructed to walk at a “comfortable” speed with the EW equipped with electric motors that attempted to regulate the EW speed by generating a propulsive or braking force based on proportional-only control. This force, while substantial (only saturating at 27.7 N), is not sufficient to overpower the user. However, to move forward at speeds not set by the EW at any point in time, the participants had to work against the EW by either pushing or pulling on it. To walk with the EW, the participants had to move forward at the same average speed. If the speed setting was different from the optimal walking speed, the participants had to decide between complying with the EW by walking at a suboptimal speed and working against the EW by pushing or pulling it to move at their optimal walking speed.

In the “No Assist” condition, the participants walked with the EW at their preferred speeds without being influenced by forces from its electric motors. The “No Assist” speed chosen by the participants, with a mean of 1.05 m s^{-1}, was presumed to be close to their optimal walking speeds. In our experiment, the larger the difference between this optimal walking speed and the speed settings, the larger the number of participants who chose to walk at speeds different from those set for the EW. This is in agreement with the parabolic relationship between energy expenditure per distance travelled and speed (Ralston, 1958; Martin, Rothstein & Larish, 1992; Browning *et al.*, 2006). Close to the minimum point, the rate of change of energy expenditure per distance with respect to speed is relatively low. Therefore, the participants may have been content to deviate from their optimal walking speed and to match the speed of the EW when the energetic penalty for walking at a suboptimal speed is fairly low. In contrast, when required to walk very slowly or very fast, the participants may have preferred to work against the EW to deviate less from their optimal walking speed.

If we assume that acceleration in the steady state is negligible and the force required to overcome rolling friction does not change with the speed setting, the differences in the force applied by the EW are caused by the different forces applied by its user (Equation (3.3)).

$$F_{EW} - F_{User} = ma + F_{Friction} \tag{3.3}$$

We would expect participants to either push or pull on the EW to affect a change in speed or to minimize the force applied on the EW when matching its speed to optimize their effort. However, despite there being only a small difference between the speed setting and the average walking speeds for the 0.8 m and 1.2 m s^{-1} conditions, the mean forces applied by the EW in these two conditions were significantly different from 0 N, while that for the 1.0 m s^{-1} condition was not. In addition, even when the participants matched the speed of the walker, the mean force applied by the EW varied linearly with the percentage difference between the walking speed and the “No Assist” speed (Figure 3.8a). Instead of being content to match the speed of the walker as suggested previously for these conditions, which would result in the mean force being close to 0 N, the force data suggest that the participants attempted to change their walking speeds but were simply unable to do so regardless of the force they exerted. This indicates that, rather than setting the force to apply to achieve a target speed, the force exerted by the participants was dependent on the difference between the optimal and the current walking speeds.

The ratings of perceived exertion reported by the users for the various speed settings followed a parabolic curve with the minimum point of the curve occurring near the optimal walking speed. This suggests that the further away the speed setting from the preferred speed, the higher the perceived exertion. In addition, although not significantly different, only the mean rating of perceived exertion reported at speed setting 1.00 m s^{-1} was lower than that in the "No Assist" condition. Hence, we can infer that the speed setting of the EW has to be similar to the optimal walking speed of its user to reduce exertion. However, Figure 3.8b shows that whether participants choose to match the speed of the EW can affect the perceived exertion (compliance or non-compliance). The increase in perceived exertion for speed settings 0.6 m s^{-1} and 0.8 m s^{-1} can almost completely be attributed to non-compliant cases.

In non-compliant instances, users had to overpower the EWs by saturating the speed controllers to walk at speeds more than 5% different from the speed settings. In all these cases, the participants experienced the maximum pushing or pulling force of 27.7 N. Although walking at suboptimal speed could have contributed to higher ratings of perceived exertion, the high forces experienced were likely to have a larger effect on perceived exertion. Walking at lower than the optimal speeds increased perceived exertion, and this indicates the need for the participants to generate the forces to push the EW. This is in agreement with the results in Chapter 2 where higher resistive forces led to higher reported ratings of perceived exertion. A study Gottschall & Kram (2003) where an impeding force increased the metabolic cost required for walking further supports our data. When walking at a speed faster than the optimal speeds, the trend is not as clear-cut with some participants reporting lower ratings of perceived exertion. This may be due to the 27.7 N assistive force generated by the EW in these instances. Previous studies have shown that walking with a constant assistive force up to 8% of a person's bodyweight (47 N on average for our participants) can reduce the metabolic cost required for walking (Zirker, Bennett & Abel, 2013; Gottschall & Kram, 2003). However, in our previous study, a constant force of 27.7 N was found to increase perceived exertion on average, but a low assistive force around 9.23 N decreased the average perceived exertion. In this study, the participants who reported lower ratings of perceived exertion may have found the assistive forces helpful. Dionisio, Hurt & Brown (2018) also showed that assistive forces reduce the mechanical effort required to walk faster. This suggests that it may be possible to change the energy expenditure in the distance-walking speed relationship with the addition of an assistive force. The optimal walking speed of a participant could have increased and may have contributed to the lower perceived exertion reported.

In compliant cases, the perceived exertion does not increase when walking at speeds lower than the optimal walking speeds. This is despite the fact that the participants pushed harder on the EWs the lower their current walking speeds were relative to their preferred speed as shown in Figure 3.8a. This may be because the increase in exertion required to overcome the resistive forces may have been offset by the decreased exertion required to walk at lower speeds. Although preferred walking speeds are typically selected based on the energy expenditure per distance travelled, walking at lower speeds decreases both metabolic rate and oxygen consumption (Martin, Rothstein & Larish, 1992; Browning *et al.*, 2006). When walking at higher than optimal speeds for compliant instances, only a slight trend could be observed as walking speed increased. In this case, the benefits of the assistive force from the EW (Zirker, Bennett & Abel, 2013; Gottschall & Kram, 2003) may have been counteracted by the need to walk at higher speeds that increase metabolic rate and perceived exertion (Martin, Rothstein & Larish, 1992; Browning *et al.*, 2006).

Both stride length and cadence increased linearly as the speed setting increased from 0.6 m s^{-1} to 1.4 m s^{-1}. However, the rate of increase of cadence with increasing walking speeds appears to level off at high speeds. In addition, the Wilcoxon signed-rank tests showed that walking at very low speeds (0.6 m s^{-1} condition) significantly increased the double support phase and decreased the swing phase and single support phase relative to the "No Assist" condition. Walking at higher speeds did not significantly change the distribution of the gait phases. The linear relationship for stride length and cadence with speed was consistent with previous studies (van Hedel, Tomatis & Müller, 2006; Kirtley, Whittle & Jefferson, 1985; Latt *et al.*, 2008; Sekiya & Nagasaki, 1998). The levelling off of cadence at high speeds and the pattern observed for the gait phases are in agreement with the results obtained by van Hedel, Tomatis & Müller (2006) for free walking. This suggests that the participants changed their stride length, cadence, and gait phases similar to free walking to achieve different walking speeds and the speed control of the EW did not cause its users to walk with an unnatural gait pattern.

In summary, when a EW is controlled to move at speeds faster or slower than the optimal walking speed, some users will match the speed of the EW while others will force the device to move at a speed closer to the optimal walking speed. The more different the speed at which the EW is set to move at from the optimal walking speed, the higher the proportion of users that will work against the EW and force it to move at a different speed. Generating the forces to overpower the EW and to cause it to move at different speeds leads to increased perceived exertion by the user.

Even when the users matched the speed of the EW, it was found that when they pushed or pulled harder on the EW, the more different their current walking speed was from their optimal speed. This pushing or pulling of the EW, together with walking at suboptimal speeds, can increase perceived exertion and decrease the effectiveness of the EW.

It is important to consider a user's preferred or optimal walking speed when controlling the speed of an EW. The user and the EW are likely to work against each other if the user has to walk at suboptimal speeds, increasing both the effort exerted by the user and the energy used by the device. In addition, if a speed limit is implemented on an EW to improve safety, developers should ensure that the speed limit is above that of the optimal walking speeds of typical users.

3.5 Conclusion

In summary, a Smart Walkers must be controlled to move at a speed close to the optimal walking speed of its user. More research is needed to understand how users choose whether or not to comply with the speed setting of a Smart Walker and why they push or pull on the device despite it not leading to any consequential change in speed.

Chapter 4 The Biomechanical Effects of Constant Forces on Gait

The Biomechanical Effects of Constant Forces on Gait

4.1 Introduction

From smartwatches to robotic wheelchairs, our lives are being improved every day by technologies that extend the capabilities of familiar everyday devices. In the same vein, traditional four-wheeled walkers are being upgraded with motorized wheels that can provide assistive forces to its users as well as sensors that can detect obstacles and sense its user’s motion. These Smart Walkers or Robotic Rollators (Martins *et al.*, 2015; Werner *et al.*, 2018), as they are often known, have the potential to substantially improve the lives of people who experience walking difficulty by providing additional features while also overcoming some the shortcomings of four-wheeled walkers.

Four-wheeled walkers are popular among older adults (Liu, 2009; Brandt, Iwarsson & Stahl, 2003) and are often prescribed to people who experience walking difficulty (Schwenk *et al.*, 2011; Haines, Brown & Morrison, 2008; Bradley & Hernandez, 2011). This is because using it resembles normal walking (Bradley & Hernandez, 2011), more so than other traditional mobility aids, and requires the least amount of effort to use (Priebe & Kram, 2011). However, previous researches have shown that its users experience a higher risk of more severe injuries (van Riel *et al.*, 2014; Stevens *et al.*, 2009) and have an altered gait pattern associated with older adults with decreased walking capability (Liu *et al.*, 2009).

There is an increasing amount of research and development being done on Smart Walkers. Research in this domain, however, has so far been largely focused on the development of technical functionalities and studies involving users have been mainly to evaluate the effectiveness of the functionalities developed (Martins *et al.*, 2012, 2015; Page *et al.*, 2016; Werner *et al.*, 2016, 2018). This study looks to take a step back and try to investigate a user's response or adjustments to the additional actuation capability present in Smart Walker to better understand how it can be used to improve and extend on the physical assistance provided by four-wheeled walkers.

In Chapter 2, it was shown that assistive forces from a Smart Walker could be used to increase the walking speed of the user and although a relatively small assistive force could decrease the perceived exertion of the user, stronger forces led to increased perceived exertion. However, because walking speed was one of the responding variables, the effect of walking speed and the assistive force on the parameters measured could not be distinguished. Walking speed can substantially affect the user's gait adjustments and perceived exertion; hence, it is also important to consider the effects of walking speed. Furthermore, as demonstrated in Chapter 3, when a Smart Walker is controlled to move at a non-preferred speed, users would push or pull on the device even when this action does not result in any observable change in speed. This further highlights the need to understand the effects of and the relationships between walking speed and assistive forces on a user's gait and posture responses or adjustments and their physical exertion.

This study looks to build on the results obtained in previous studies and investigate in more detail the user's gait and posture adjustments to assistive forces from a Smart Walker and walking speed as well as their interactions. Firstly, in this study, I aim to control for the effects of walking speed on the parameters measured by performing the experiment at different walking speeds for each force level. Secondly, in addition to the basic parameters measured in our previous studies, I aim to investigate the effects on the mechanical work performed by the user, their joint kinematics and dynamics, and their posture.

4.2 Methods

4.2.1 Participants

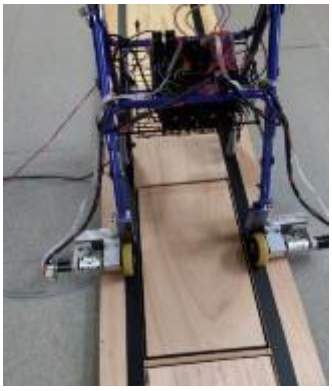
Nineteen male participants from the university community were recruited to participate in this study (age: 23.8 ± 2.1 years, height: 169.8 ± 3.4 cm, weight: 60.7 ± 6.2 kg). All the participants were healthy and did not experience any difficulty walking. This study was approved by the Ethics Committee of the Faculty of Design, Kyushu University. Written informed consent was provided by the participants before the experiment was performed.

4.2.2 Experiment Setup

The walker used is this experiment is a typical four-wheeled walker that was custom equipped with electric motors such that it can be controlled to produce forward assistive forces of different magnitudes (Figure 1a). The experiment walker (EW) used in this experiment is the same as that used in Chapter 2 and Chapter 3.

A custom-made 12.8 m long walkway with three embedded force platforms was used in this experiment. The width of the force platform's measurement area was made narrower to allow the EW wheels to pass next to it without loading the force platform as shown in Figure 4.1

a)



b)

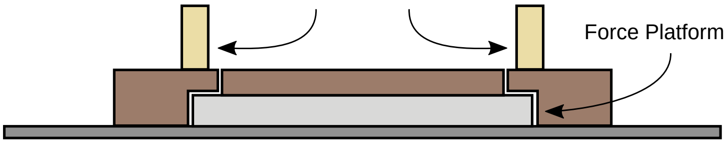


Figure 4.1. (a) Photograph showing the EW and the walkway with embedded force platforms, (b) Schematic showing how the force platforms have been embedded to allow the EW's wheels to pass next to it without loading the force platform.

A light-emitting diode (LED) strip was placed next to the walking path to help the participants walk at the targeted speeds (Figure 4.2). The lights on the LED strip were turned on consecutively at constant time intervals to create a "running" effect (Huang, Zhuang & Zhang, 2013). These lights were programmed to move at the target speed and the users could match their walking speed to the speed of the "running" LEDs. All participants wore tight, form-fitting clothing and the same model of lightweight flexible shoes prepared by the experimenter throughout the experiment.

4.2.3 Experiment Conditions and Tasks

Three target walking speeds and three force settings were investigated in a 3×3 factorial study design. The participants were instructed to walk at 0.8 m s^{-1}, 1.0 m s^{-1}, and 1.2 m s^{-1}. For each walking speed, the effects of constant assistive horizontal forces with magnitudes 0 N, 15 N, and 30 N was investigated. The order of the experimental trials were randomized.

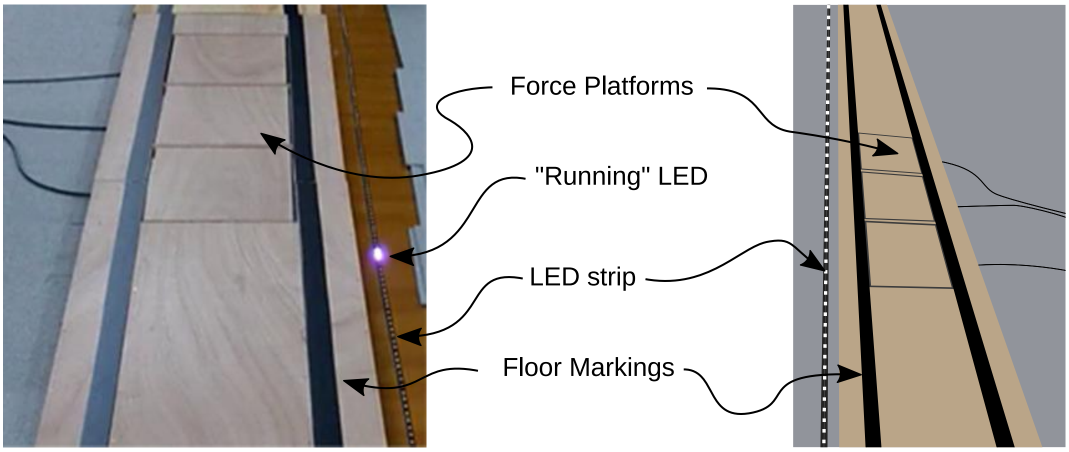


Figure 4.2. Experiment setup.

The participants were instructed to try to walk along the 12.8 m long walkway with the EW while trying to keep an upright posture and placing 30% of their body weight on the EW. A research assistant would walk alongside the participant when performing the experiment trials to activate the EW and to prevent falls.

In addition, to prevent the EW's wheels from going onto the force platforms, the participants were asked to keep the EW wheels within the width of the floor markings shown in Figure 4.2. The participant was not informed about the presence of the force platforms to prevent them from walking unnaturally in order to step on the force platforms. Despite the need to keep the wheels within the floor markings and the use of the LED strip to help control their speed, participants were asked to look straight ahead when performing the experiment.

4.2.4 Experiment Protocol

Participants were given training before the start of the experiment to be able to place 30% of their body weight on the EW as described in Section 2.2.1. Before starting experiment trials for each condition, participants were allowed several practice trials to familiarize themselves to the walking speed and the force applied.

During the experiment, the participant's speed was measured using embedded encoders in the motors of the EW. If the speed in a trial differed more than 5% from the target speed, that trial was discarded. In addition, the participants need to step on the force platforms with at least two clean separate contiguous steps (one on each force platform). In other words, if the EW's wheel moved outside the width of the floor marking and touched the force platform or the participant's feet was not fully on the measurement area, the trial was also discarded. Trials for a condition were repeated until at least three successful trials were obtained.

4.2.5 Data Acquisition

The participants' kinematic data were collected using a 3D motion-capture system (Cortex 7.0; Motion Analysis Corporation, Santa Rosa, CA, USA) comprising of 11 infrared cameras and reflective markers. Thirty-nine of these passive retro-reflective markers were attached to the bony landmarks on the participant's body as shown in Figure 4.3. The trajectories of these markers were tracked and recorded at 100 Hz. Only motion performed near the center of the walkway was recorded.

A modified Helen Hayes marker-set protocol was used in this study. Additional markers were placed at the 7th cervical vertebrae, greater trochanter, proximal tip of the fibula head, and the most anterior border of the tibial tuberosity to allow for reconstruction if a marker was obscured from the camera view by the EW or by the research assistant that walked alongside the participant during trials. The medial markers (colored red in Figure 4.3) were only used for static calibration and were removed for dynamic trials.

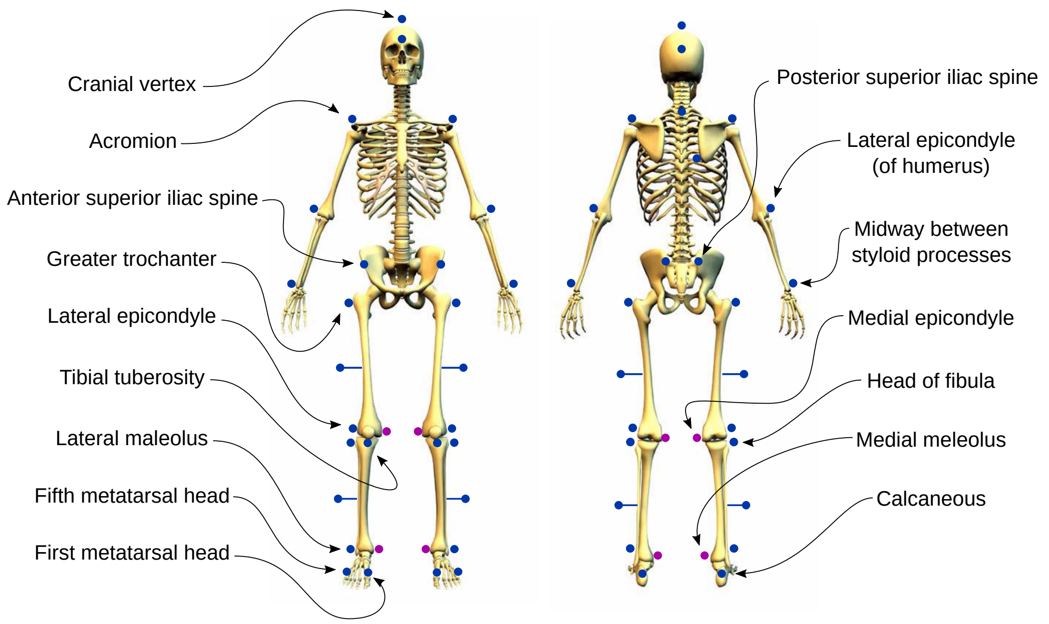


Figure 4.3. Attachment locations of reflective markers.

The three force platforms (9286A; Kistler, Winterthur, Switzerland) embedded in the walkway were sampled at 1000 Hz in sync with the kinematic data through the 3D motion-capture system. The participants were asked to rate their perceived exertion based on the Borg's RPE scale (Borg, 1990) after every trial.

4.2.6 Data Processing

The motion data was smoothed using a fourth-order low pass Butterworth filter with cutoff frequency at 6 Hz. One full gait (heel contact to the subsequent heel contact of the same foot) was used from each trial.

Cadence was calculated from the period of the gait cycle. Stride length was calculated as the distance between the calcaneus marker on the side of the heel contact at the start of the gait cycle and the same calcaneus marker at the end of the gait cycle. Walking speed was calculated by dividing stride length with cadence. Gait phases were calculated using the time of the gait events identified.

A 15-segment rigid body segment model consisting of a head, trunk, and pelvis, and pairs of upper arm, forearm, hand, thigh, shank, and feet was used in this study. The location of the hip joint center was estimated using the "prediction" approach laid out by Bell, Brand & Pedersen (1989) where its location was defined as 30% distal, 14% medial, and 22% posterior to the anterior superior iliac spines (ASIS) marker (expressed as a percentage of the ASIS to ASIS distance). The knee and ankle joint centers were defined as the midpoint between the lateral and medial epicondyle markers, and the lateral and medial malleolus markers respectively. The distal endpoint of the foot segment (ball of foot) was defined to be 2 cm below the first metatarsal head marker when standing upright. The location of the joint centers for the shoulder, elbow, and wrist were approximated as the location of the shoulder, elbow, and wrist markers. The distal endpoint of the hand segment (dactylion) was defined to be the point extended from the wrist marker distally by 40% of the distance between the elbow and wrist markers. The cervicale was approximated as the midpoint between the acromion markers while the Lumbo-Scarum was approximated as midway between the ASIS markers and 44% of the distance from the ASIS markers to the posterior superior iliac spine (PSIS) markers. The proximal and distal endpoints of the segments used are shown in Table 4.1. The primary axis for each segment runs from its distal to the proximal endpoints. The inertial parameters (relative segment masses, center of mass positions, and radii of gyrations) for these segments were obtained from de Leva (1996).

The total center of mass (CoM) for the participant is calculated as the weighted average of the CoM of the body segments as shown in equation (4.1) where $\vec{\boldsymbol{r}}_{CoM}$ refers to the total CoM of the participants, P_s is each segment's mass proportion, and $\vec{\boldsymbol{r}}_{segment}$ is the segment's CoM. The CoM velocity and acceleration were calculated by taking the first and second order derivatives of $\vec{\boldsymbol{r}}_{CoM}$.

$$\vec{\boldsymbol{r}}_{CoM} = \sum_{s=1}^{15} P_s \, \vec{\boldsymbol{r}}_{segment} \tag{4.1}$$

Table 4.1. Proximal and distal endpoints of body segments (Adapted from Ho Hoang & Mombaur (2015))

Segment	Proximal endpoint	Distal endpoint
Head	Cervicale	Cranial vertex
Trunk	Lumbo-Sacrum	Cervicale
Pelvis	Midpoint of hip joint centers	Lumbo-Sacrum
Upper arm	Shoulder joint center	Elbow joint center
Forearm	Elbow joint center	Wrist joint center
Hand	Wrist joint center	Dactylion
Thigh	Hip joint center	Knee joint center
Shank	Knee joint center	Ankle joint center
Foot	Ankle joint center	Ball of foot

The work rate and mechanical work performed on the participant's CoM by the leading leg was calculated using the procedure described in Donelan, Kram, & Kuo (2002). The work rate, P was calculated using equation (4.2) where $\vec{\boldsymbol{F}}_{GRF}$ is the ground reaction force (GRF) acting on the leading leg and $\vec{\boldsymbol{v}}_{CoM}$ is the velocity of the CoM. The mechanical work, W_{CoM}^{T} performed by the leading leg in a certain period, T was calculated as the cumulative time-integral of the CoM work rate, P_{CoM} as shown in equation (4.3).

$$P_{CoM} = \vec{\boldsymbol{F}}_{GRF}\, \vec{\boldsymbol{v}}_{CoM} \tag{4.2}$$

$$W_{CoM}^{T} = \int_{T} P_{CoM}\ dt \tag{4.3}$$

The stance phase was further divided into four sub-phases found in Kuo, Donelan & Ruina (2005). These sub-phases, which are collision, rebound, pre-load, and push-off, are as shown as banded regions (1) to (4) in the gait graphs in Figures 4.7 to 4.12 based on the transitions between single support and double support phases and that between positive and negative work rate on the CoM.

Joint angles, moments, and powers for the lower limb segments and coordinates of the center of mass (CoM) in the sagittal plane were calculated using KinTools RT. A python script was used to calculate the angles of the upper limbs and trunk as well as the work rate on the CoM. The flexion/extension joint angles for the lower and upper limbs were calculated as the angle between the primary axes of its proximal and distal segments. The trunk angle refers to the angle between the primary axis of the trunk segment and the global vertical axis.

The joint moments for the segments, were calculated from the distal to proximal segments separately by solving the net joint force and moment equations shown in equations (4.4) and (4.5) for each segment i, where $\vec{\boldsymbol{F}}_{ip}$ and $\vec{\boldsymbol{F}}_{id}$ are the proximal and distal force vectors, $\vec{\boldsymbol{M}}_{ip}^{s}$ and $\vec{\boldsymbol{M}}_{id}^{s}$ are the proximal and distal moment vectors, m_i is the segment mass, $\vec{\boldsymbol{a}}_i$ is the linear acceleration vector of the segment, $\dot{\boldsymbol{H}}_{i}^{s}$ is the time derivative of the angular momentum, and $\vec{\boldsymbol{k}}_{i}^{s}$and $\vec{\boldsymbol{l}}_{i}^{s}$ are the proximal and distal lever arm vectors (Dumas, Aissaoui & de Guise, 2004).

$$\vec{\boldsymbol{F}}_{ip} = m_i\vec{\boldsymbol{a}}_i - m_i\vec{\boldsymbol{g}} - \vec{\boldsymbol{F}}_{id} \tag{4.4}$$

$$\vec{M}_{ip}^{s} = \dot{H}_{i}^{s} - \vec{M}_{id}^{s} - \left(\vec{k}_{i}^{s} \times \vec{F}_{ip}^{s}\right) - \left(\vec{l}_{i}^{s} \times \vec{F}_{id}^{s}\right) \tag{4.5}$$

The joint were calculated using the joint moment and joint angular velocity as shown in equation (4.6) where $\vec{P}_i$ is the joint power vector, $\vec{M}_i$ is the joint moment vector, and $\vec{\omega}_i$ is the joint angular velocity vector.

$$\vec{P}_i = \vec{M}_i \, \vec{\omega}_i \tag{4.6}$$

The results from each trial were time normalized to the gait cycle by interpolating the data points for averaging purposes and expressed as a percentage of the gait cycle. The calculation were perform on a cycle-by-cycle basis. The peak joint moments and powers of the lower limb were calculated for specific parts of the gait graphs keeping with the previous convention (Winter, 1983). These were labelled as A1, A2, K1, K2. K3. H1, H2, and H3 in Figure 4.11 for joint powers while those for joint moments were simply labelled as the maximum flexion/extension.

4.2.7 Statistical Analysis

Two-way repeated-measures analysis of variance (ANOVA) tests were used to determine the effects of different magnitudes of assistive force and the different target walking speeds of the participants. All results are reported as means ± standard deviation, n=19.

4.3 Results

4.3.1 Walking Speed

The participants walked at 0.84 ± 0.03 m s^{-1}, 1.04 ± 0.03 m s^{-1}, and 1.22 ± 0.03 m s^{-1} when the target speed was at 0.8 m s^{-1}, 1.0 m s^{-1}, and 1.2 m s^{-1} respectively (Figure 4.4). The target speed significantly affected the measured walking speeds of the participants, F[2, 36] = 3008.9, $p < .05$, $\omega^2 = .483$. However, there was also significant main effect of force setting on walking speed, F[2, 36] = 6.1, $p < .05$, $\omega^2 = .0006$. There was no significant interaction between speed target and force setting on the walking speed of the participant, F[4, 72] = 0.85, $p = .50$.

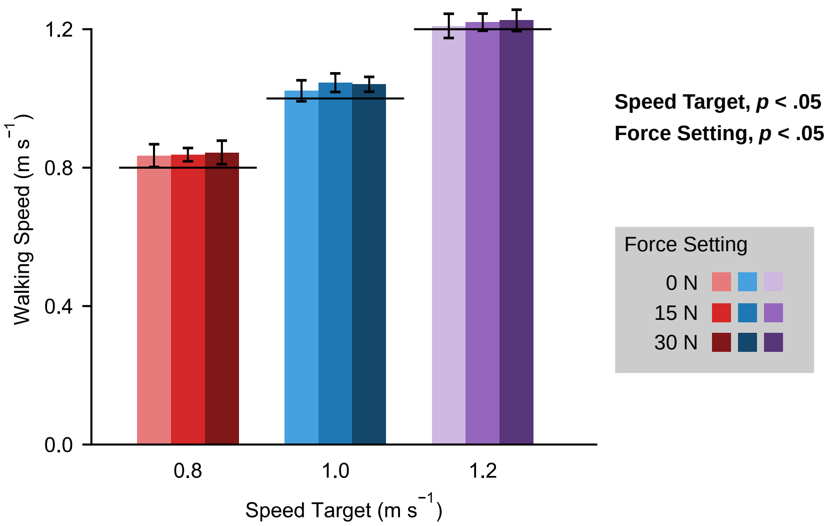


Figure 4.4. Walking speed for each speed target and force setting condition with horizontal lines showing the speed target.

4.3.2 Basic Gait Parameters and Perceived Exertion

Significant main effect of speed target was present for stride length (F[2, 36] = 191.6, $p < .05$), cadence (F[2, 36] = 227.7, $p < .05$), and walk ratio (F[2, 36] = 3.548, $p < .05$) but there were no significant main effects of force setting on these spatiotemporal gait parameters. There was also no significant interaction effects between target speed and force setting for the spatiotemporal gait parameters (Figure 4.5).

Although the ratings of perceived exertion reported were not significantly affected by both force setting and speed target there was a significant interaction effect between force setting and target speed, F[4, 72] = 7.9, $p < .05$ (Figure 4.5).

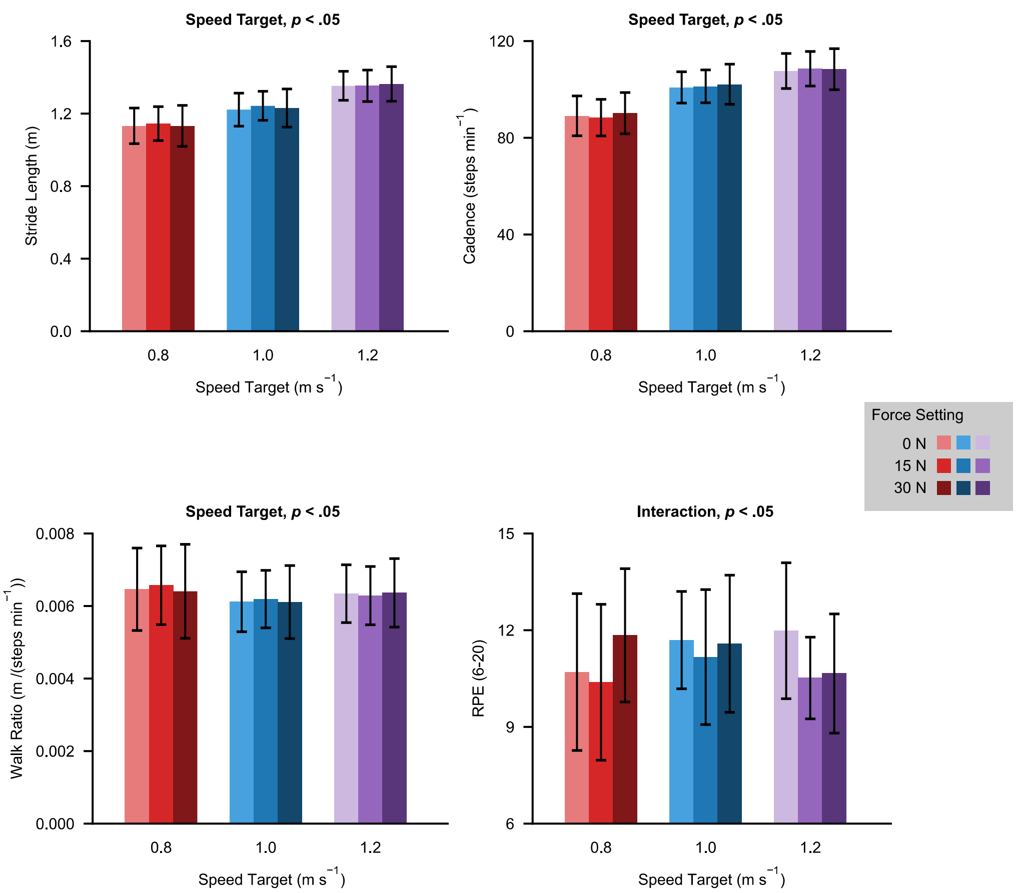


Figure 4.5. The effects on stride length, cadence, walk ratio, and ratings of perceived exertion (RPE).

4.3.3 Gait Phases

Both double support phase and stance phase were significantly affected by target speed (P[2, 36] = 37.0, p < .05 and P[2, 36] = 27.5, $p < .05$), and force setting (P[2, 36] = 11.0, p < .05 and P[2, 36] = 4.1, $p < .05$). No significant interaction effects were found for the gait phases measured (Figure 4.6).

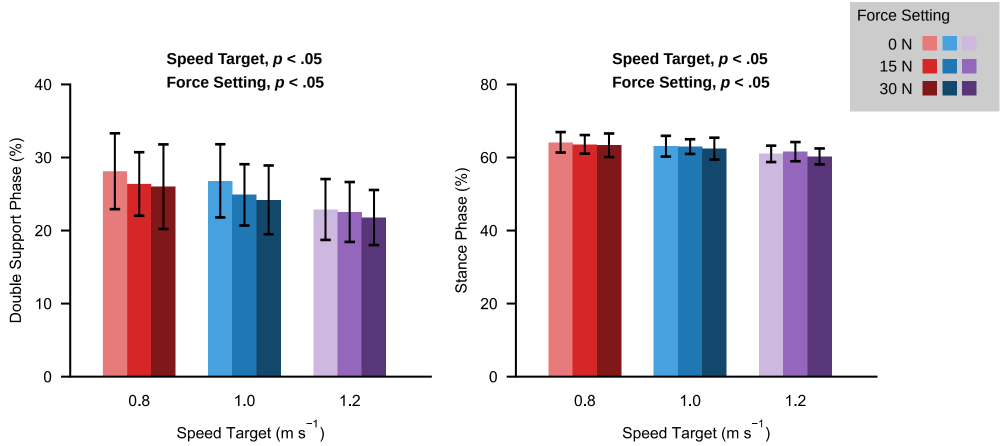


Figure 4.6. The effects on the percentage of time spent in each gait phase.

4.3.4 Work and Accelerations of the Center of Mass

The different target speed had a significant main effect on the amount of external work on the participant's center of mass (CoM) produced by the ground reaction force (GRF) during collision (F[2, 36] = 30.5, $p < .05$, $\omega^2 = .067$), rebound (F[2, 36] = 16.6, $p < .05$, $\omega^2 = .046$), pre-load (F[2, 36] = 4.8, $p < .05$, $\omega^2 = .009$), and push-off (F[2, 36] = 43.0, $p < .05$, $\omega^2 = .066$). Similarly, there were significant main effects of force settings on the work produced during collision (F[2, 36] = 65.5, $p < .05$, $\omega^2 = .105$), rebound (F[2, 36] = 79.2, p < .05, $\omega^2 = .149$), pre-load (F[2, 36] = 50.4, $p < .05$, $\omega^2 = .081$), and push-off (F[2, 36] = 127.2, p < .05, $\omega^2 = .098$). The only significant interaction effect present for the external work on CoM produced by the GRF was the negative work during collision (F[4, 72] = 3.5, $p < .05$, , $\omega^2 = -.001$) (Figure 4.7).

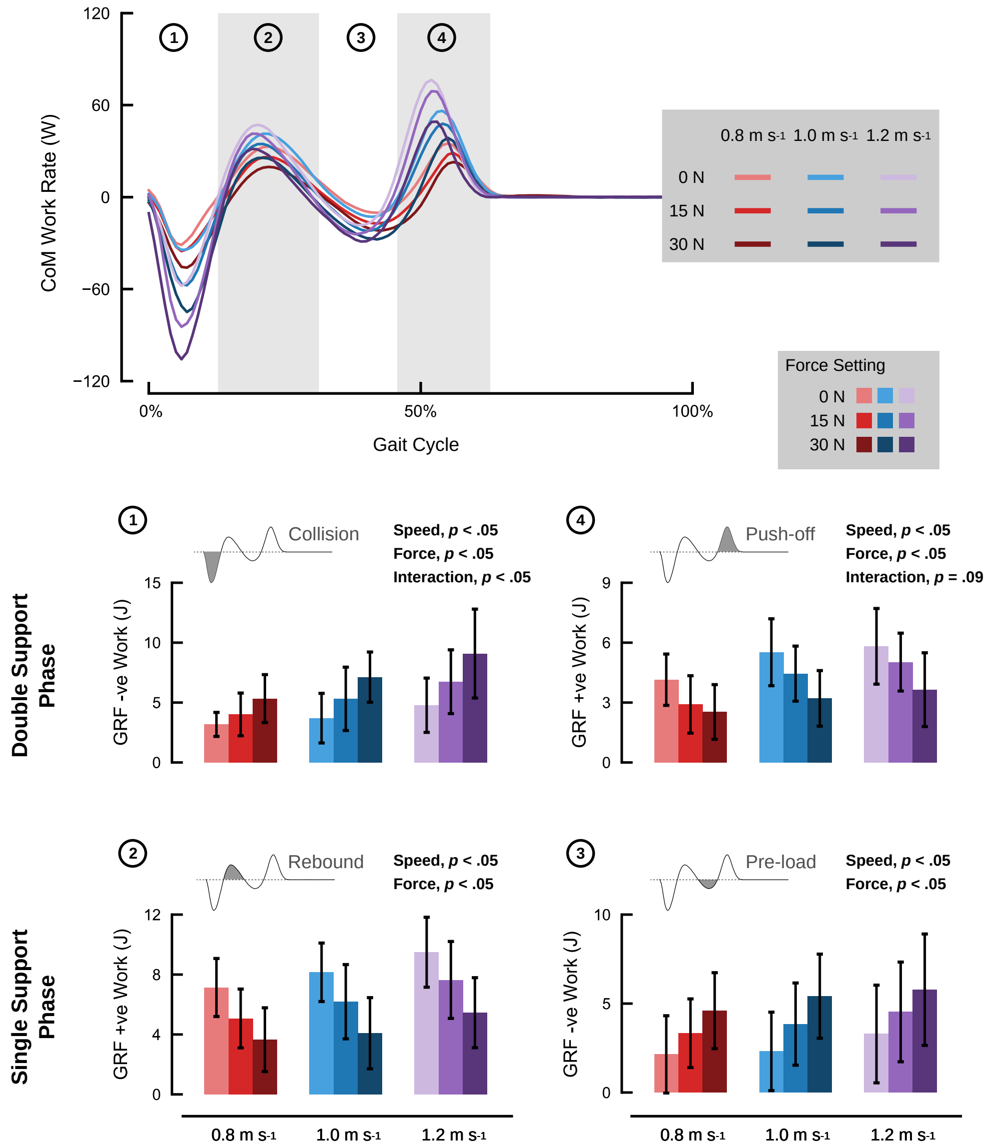


Figure 4.7. The effect on work rate on the center of mass (CoM) produced by the ground reaction force (GRF) of one limb, normalized to the gait cycle and the corresponding work produced during (1) collision (negative work during double support phase), (2) rebound (positive work during the single support phase), (3) pre-load (negative work during the single support phase), and (4) push-off (positive work during the double support phase).

Maximum forward and vertical accelerations and decelerations of the CoM were all significantly affected by the speed target. (F[2, 36] = 45.2, $p < .05$, F[2, 36] = 47.4, $p < .05$, F[2, 36] = 53.3, $p < .05$, and F[2, 36] = 96.1, $p < .05$) were significantly affected by the speed target. However, only the maximum vertical acceleration was significantly affected by the force setting (F[2, 36] = 35.4, $p < .05$) and had significant interaction effect between speed target and force setting (F[4, 72] = 2.6, $p < .05$) (Figure 4.8).

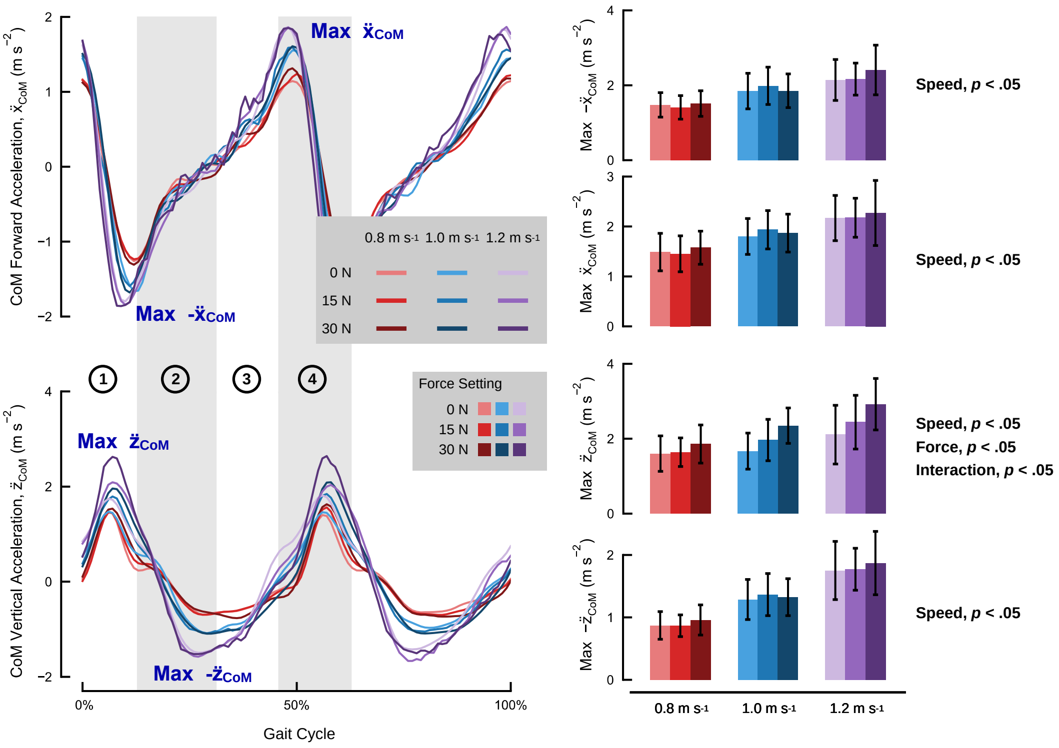


Figure 4.8. The effect on the forward and vertical acceleration of the center of mass (CoM), normalized to the gait cycle. The bar charts show the maximum acceleration and decelerations for the vertical and forward directions in the stance phase.

4.3.5 Joint Angles, Moments, and Powers

No interaction effects were found for target speed and force setting on the maximum and minimum sagittal plane joint angles of the lower limb. The target walking speed had significant main effects on the maximum plantarflexion angle (F[2, 36] = 45.9, $p < .05$), the maximum knee flexion angle (F[2, 36] = 15.5, $p < .05$), and the maximum hip flexion (F[2, 36] = 14.65, $p < .05$) and extension angles (F[2, 36] = 14.78, $p < .05$). The force setting significantly affected the maximum dorsi- and plantarflexion angles of the ankle (F[2, 36] = 10.97, $p < .05$; F[2, 36] =

15.45, $p < .05$), maximum knee flexion angle (F[2, 36] = 4.24, $p < .05$), and maximum hip flexion angle (F[2, 36] = 18.18, $p < .05$) (Figure 4.9).

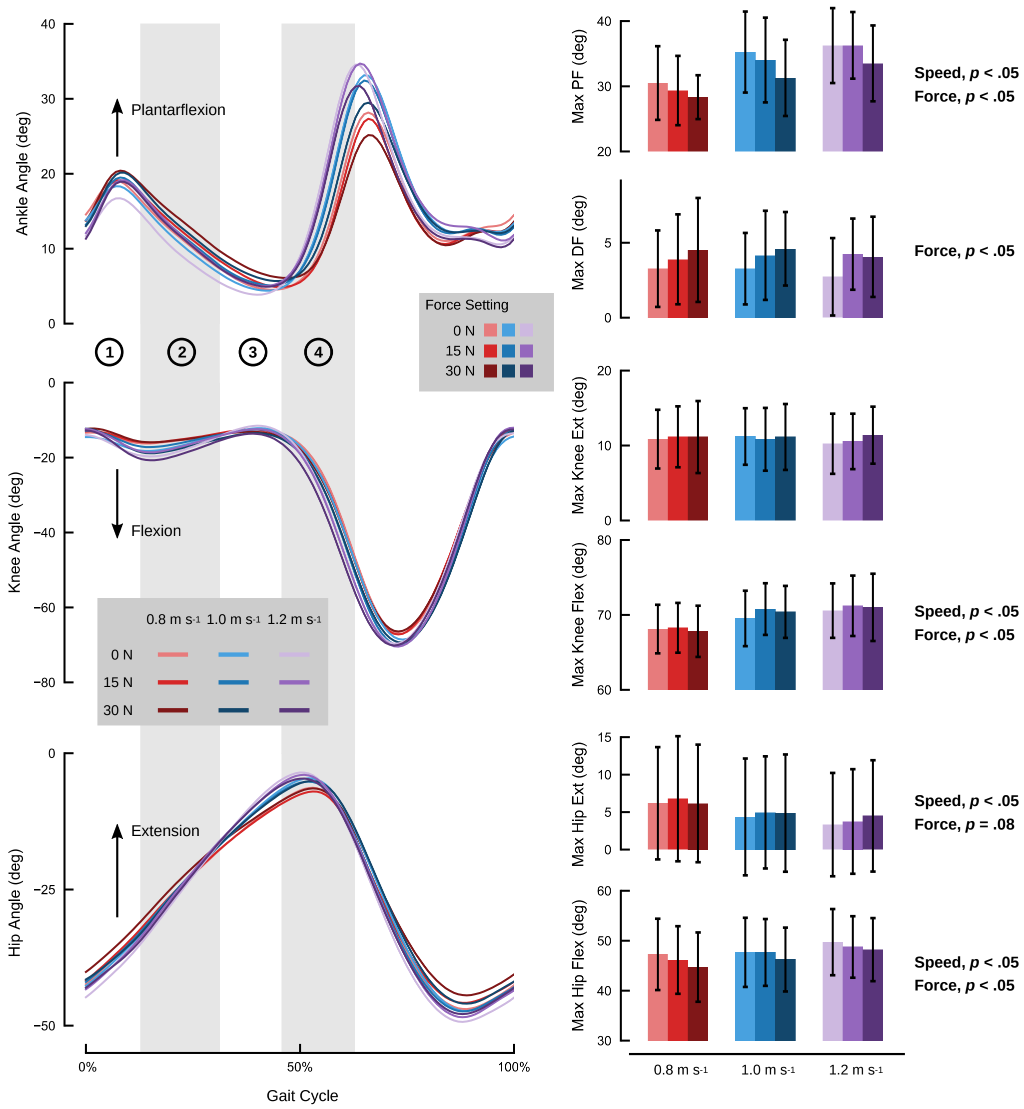


Figure 4.9. The effect on lower limb joint angles in the sagittal plane, normalized to the gait cycle. Angles are defined as positive in extension. The bar charts show the maximum extension and flexion angles in the sagittal plane.

The speed target has significant main effects on maximum plantarflexion moment (F[2, 36] = 18.9, $p < .05$), maximum knee flexion moment (F[2, 36] = 15.6, $p < .05$), maximum hip extension moment (F[2, 36] = 85.9, $p < .05$), and maximum hip flexion moment (F[2, 36] = 37.2, $p < .05$). No significant main effect of force setting were present for joint moments except for the knee joint (F[2, 36] = 4.35, $p < .05$). There was significant interaction for maximum hip flexion moment (F[4, 72] = 3.1, $p < .05$) (Figure 4.10).

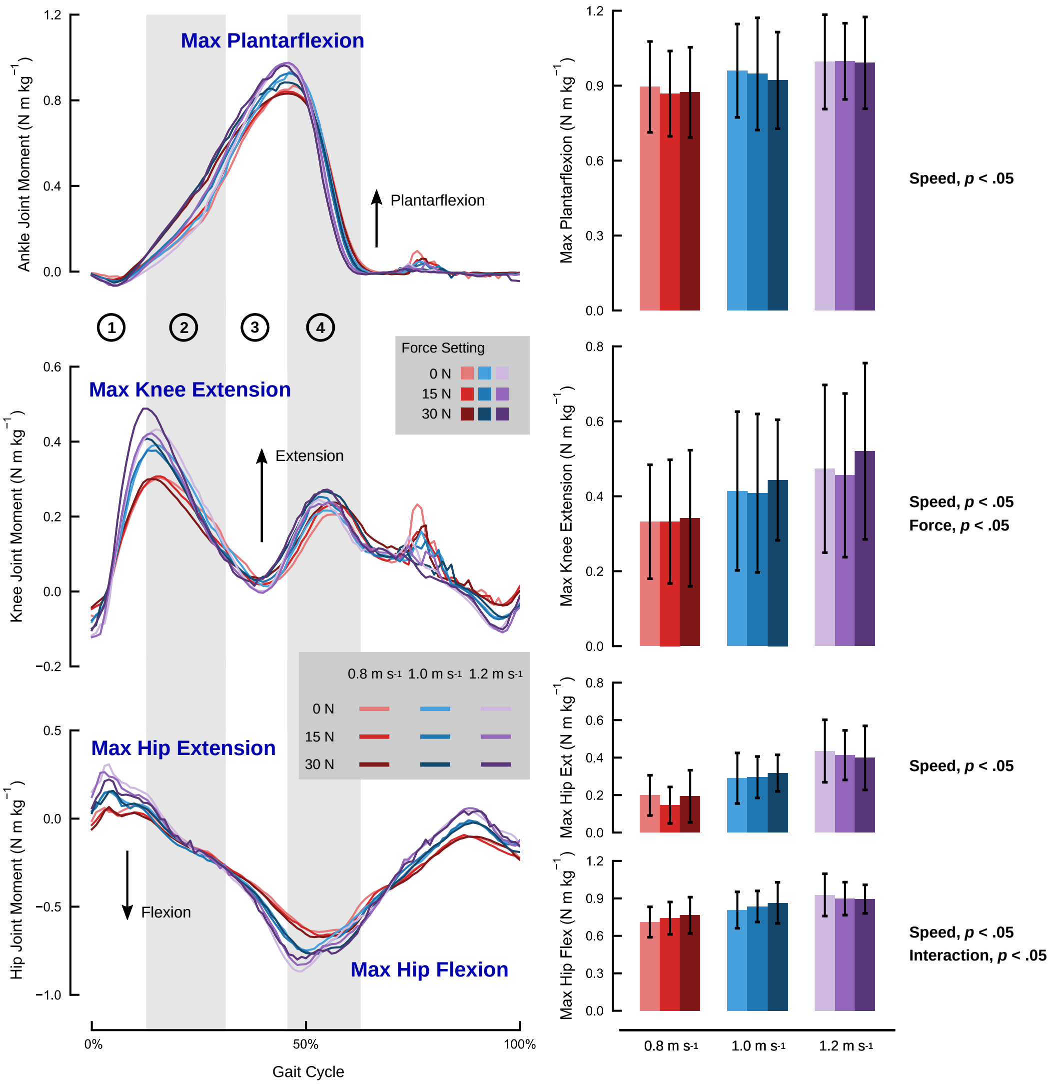


Figure 4.10. The effect on joint moments in the sagittal plane, normalized to the gait cycle. Moments are defined as positive in extension. The bar charts show the maximum plantarflexion moment of the ankle, maximum knee extension moment of the knee, and maximum flexion and extension moments of the hip.

All peak powers shown in Figure 4.11 were significantly affected by the targeted walking speed, specifically, A1 (F[2, 36] = 25.8, $p < .05$), A2 (F[2, 36] = 179.8, $p < .05$), K1 (F[2, 36] = 35.9, $p < .05$), K3 (F[2, 36] = 24.4, $p < .05$), H2 (F[2, 36] = 124.6, *p* < .05), and H3 (F[2, 36] = 88.4, *p* < .05). All peak powers except H2 were significantly affected by the force setting. Namely, A1 (F[2, 36] = 8.1, *p* < .05), A2 (F[2, 36] = 12.9, *p* < .05), K1 (F[2, 36] = 20.3, *p* < .05), K3 (F[2, 36] = 18.8, *p* < .05), and H3 (F[2, 36] = 5.9, *p* < .05). There were no significant interaction effects for the peak powers investigated (Figure 11).

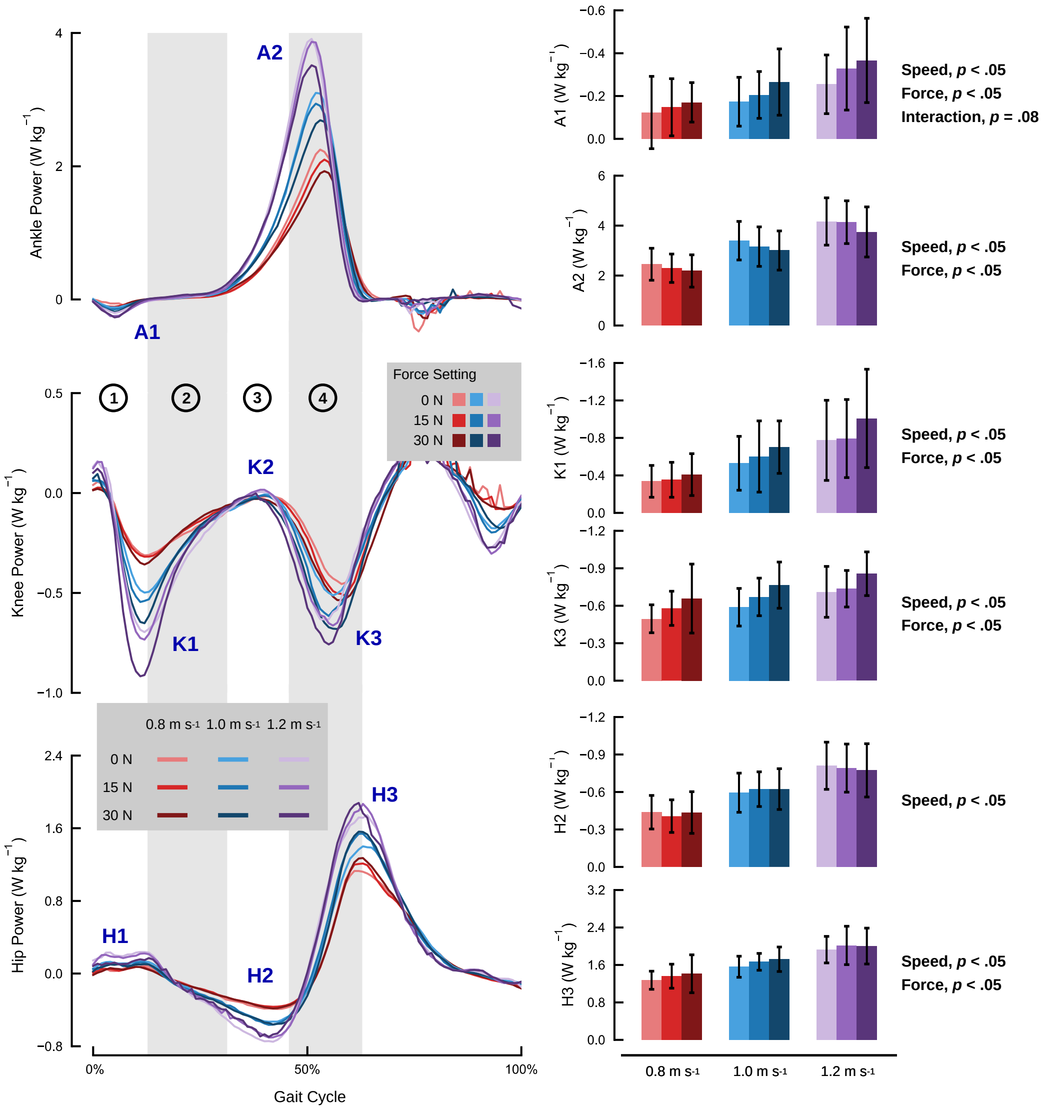


Figure 4.11. The effect on joint powers in the sagittal plane, normalized to the gait cycle. A1 refers to the peak power absorbed during collision. A2 refers to the peak power produced by plantar flexor during push-off. K1 refers to the peak power absorbed by the knee extensor during collision. K3 refers to the peak power absorbed by the knee extensor during push-off. H2 refers to the peak power absorbed by the hip flexor. H3 refers to the peak power generated by the hip flexor during pre-swing.

4.3.6 Posture

The mean elbow angle was significantly affected by the speed target ($F[2, 36] = 8.2$, $p < .05$) while the mean shoulder angle was significantly affected by the force setting ($F[2, 36] = 5.1$, $p < .05$). The RoM of the trunk was significantly affected by both the speed target (F[2, 36] = 3.8, $p < .05$] and the force setting (F[2, 36] = 18.0, p < .05). Likewise, the RoM of the shoulder was also significantly affected by speed target (F[2, 36] = 9.4, $p < .05$) and the force setting (F[2, 36] = 5.2, p < .05). The elbow RoM, on the other hand, was only significantly affected by the speed target (F[2, 36] = 15.3, $p < .05$).

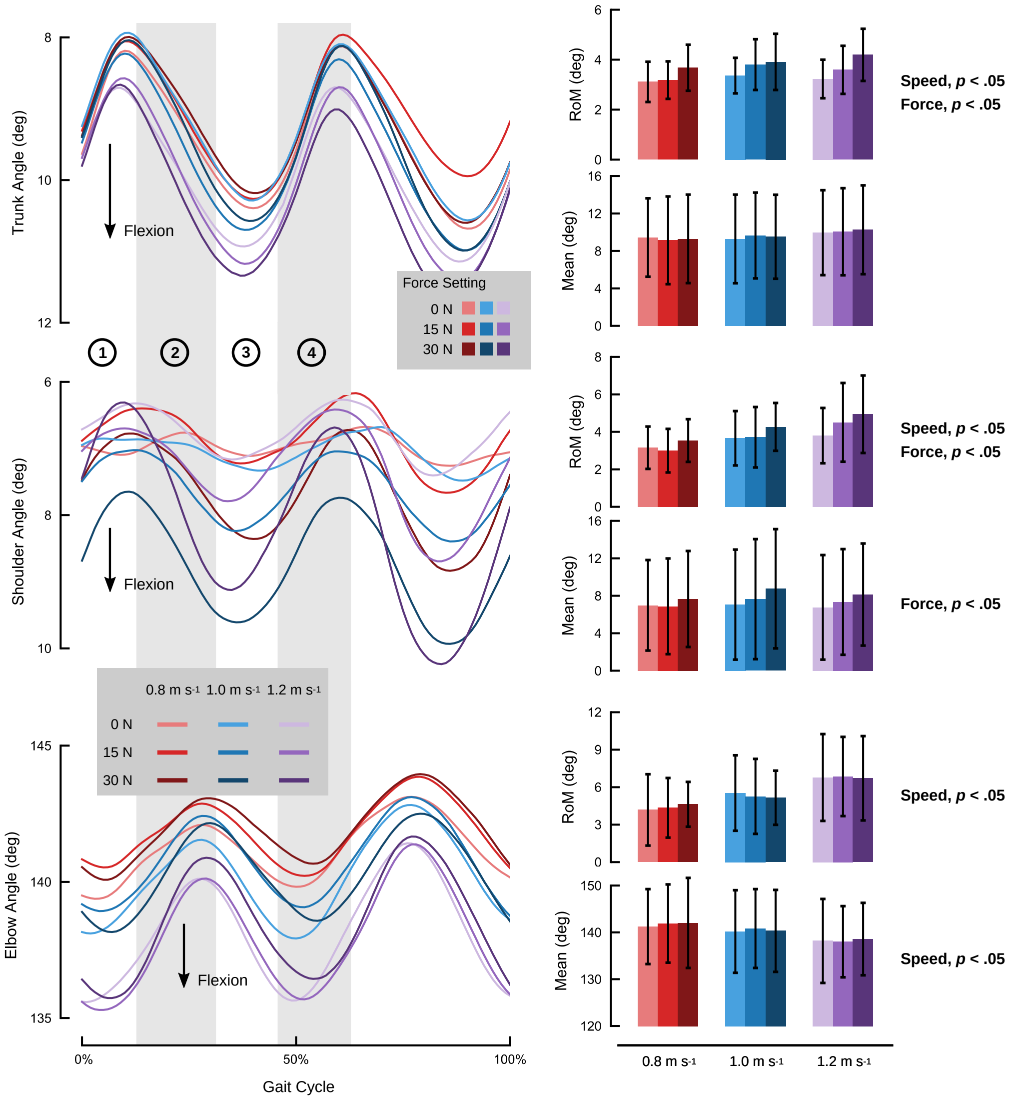


Figure 4.12. The effect on upper body joint angles in the sagittal plane, normalized to the gait cycle. The bar charts show the range of motion (RoM) and the mean angle in the sagittal plane.

4.4 Discussion

4.4.1 Walking Speed

Three distinct walking speeds corresponding to the speed targets were successfully achieved for all the force settings. There is at least a 0.18 m s^{-1} difference in walking speed on average between each speed target condition and the two-way ANOVA test performed shows that speed target had a significant main effect on walking speed (Figure 4.4).

However, the mean walking speeds were consistently slightly above the speed targets set in this experiment. This can be attributed to the method used to measure walking speed when performing the experiment in order to decide whether to discard a trial. During the experiment, the mean EW speed over the measurement area was measured in real-time and this was the speed that was ensured to be within 5% of the target speed. Although mean walking speed should be the same as the EW speed on average (the user and the EW need to move together), there may be a slight difference in speed between the user and the EW in certain gait phases. The period used to calculate the average speed may not correspond to a full gait cycle and this may have led to the discrepancy between the target speed and the actual walking speed. This is exacerbated by the odd number of steps (3 steps or 1.5 cycles) taken inside the measurement area.

There was also a significant main effect of force setting on the walking speed of the user. Despite my best efforts to ensure that walking speeds remained the same for each speed target condition, higher force settings led to significantly higher walking speeds. Nevertheless, this effect is very small; the effect size for the effect of force setting is orders of magnitude smaller than that for speed target ($\omega^2_{\text{speed target}}$

= .483, $\omega^2_{\text{force setting}} = .0006$). This small change in walking speed would likely not have a noticeably effect on the other parameters investigated.

4.4.2 Spatiotemporal Gait Parameters

In Chapter 2, because the speed increased as the assistive force increased, there was difficulty distinguishing between the effect of force and the effect on cadence, stride length, walk ratio. In this study, the assistive force was shown not have affected these gait parameters significantly (Figure 4.5). Hence, it can be concluded that the significant effect on cadence and stride length observed in Chapter 2 was due to the change in walking speed and not because of the magnitude of the assistive force applied.

The increase in stride length and cadence with increasing walking speed observed in this study is in line with previous studies (Kirtley, Whittle & Jefferson, 1985; Latt *et al.*, 2008; Sekiya & Nagasaki, 1998; van Hedel, Tomatis & Müller, 2006). However, surprisingly, the walking speed also had a significant main effect on walk ratio. Walk ratio for free walking has been previously shown to remain the same for each person in the usual walking speed range of between 0.43 m s^{-1} and 1.67 m s^{-1} (Rota *et al.*, 2011). Therefore, this may be an adjustments to walking using a wheeled walker.

Previous work on assistive forces have suggested that applying an assistive force on a user's waist decreases their stride length while increasing their cadence (Zirker, Bennett & Abel, 2013), similar to walking down a slope (Kawamura, Tokuhiro & Takechi, 1991). However, the assistive forces provided did not significantly affect cadence, stride length, and walk ratio. One possible reason is that, when using the EW, the user can adjust the downward/support force applied on the handle. Pushing down harder on the handles (increased partial body weight loading) has the effect of decreasing the load supported by the user's legs and increases the friction in the wheels of the EW. The users may have adjusted this downward force in response to the assistive force produce by the EW.

4.4.3 Gait Phases

As expected, higher walking speed led to a significantly lower relative duration of double support and stance phase with respect to the gait cycle (Hebenstreit *et al.*, 2015). In addition, increasing assistive forces significantly decreased the percentage of time spent in the double support and stance phases. This is in agreement with the study by Zirker, Bennett, & Abel (2013) that showed that an assistive force applied on a participants' waists decreased their percentage of stride spent in the double support phase. This shows that assistive forces can be used to decrease the percentage of the cycle spent in the double support and stance phases (Figure 4.6).

The decrease in the relative durations in the stance and double support phases with increasing assistive force may be due to the significant effect of assistive force on the vertical acceleration of the user's CoM (Figure 4.8). As will be discussed in Section 4.4.5, the user may have been able to utilize the assistive force provided to accelerate their CoM.

4.4.4 Ratings of Perceived Exertion

There were significant interaction effects between walking speed and the magnitude of force applied by the EW on the ratings of perceived exertion (RPE), suggesting that walking speed changes the effect of force magnitude on the exertion of the user.

From Figure 4.5, we can see that as the magnitude of the force is increased from 0 N to 30 N, the Borg rating reported follows the same pattern. That is, perceived exertion decreased from 0 N to 15 N but increased from 15 N to 30 N. However, there were substantial differences in the magnitude of change in RPE. At a low walking speed (0.8 m s^{-1}), increasing the assistive force from 0 N to 15 N only decreased the RPE slightly but increasing from 15 N to 30 N led to a large increased in RPE. The opposite effect was observed for walking at a high speed (1.2 m s^{-1}); RPE decreases considerably when going from 0 N to 15 N but only increases slightly when going from 15 N to 30 N. At a moderate walking speed (1.0 m s^{-1}), however, the changes from 0 N to 15 N and from 15 N to 30 N were similar.

These indicate that a higher assistive force decreases the exertion required by the user to walk faster but increase the exertion required to walk slower. This result agrees with the study by Dionisio, Hurt & Brown (2018) that demonstrated that a forward assistive force reduces the mechanical work required to walk faster. Furthermore, this validates the assumption in Chapter 2 that the users were minimizing their exertion by walking faster as the assistive force is increased and provides support to the view that these forces can be used to elicit a higher walking speed from users. The general pattern of a low assistive force decreasing RPE and a high assistive force increasing RPE is in agreement with Chapter 2.

4.4.5 Joint Dynamics and the CoM

The results for the CoM accelerations and lower limb work on the CoM as well as that for joint kinematics suggest that the effects differ in the four sub-phases of the stance phase as defined by Kuo, Donelan, & Ruina (2005). Hence, these parameters will be discussed separately in each of the collision, rebound, pre-load, and push-off phases. Because most of the work done for walking is performed in the collision and push-off phases, this discussion will focus on these two phases (the step-to-step transition phase).

Collision

During the collision phase, the user's leg performs negative work on the CoM, first using the ankle and then the knee, to redirect the CoM from the downward motion at the end of one pendula arc to the upward motion at the start of the next pendula arc.

Cadence is higher when walking at a higher speed (Figure 4.5), which leads to a shorter duration for each stride and a shorter double support phase (Figure 4.6). Because the relative duration of the double support phase is defined as a percentage of the duration of each stride, this has the effect of substantially decreasing the absolute duration of the double support phase. The collision phase refers to the phase of the leading leg during the first double support phase, hence, the absolute duration of the collision phase also decreases substantially. This means that the CoM needs to transition from a downward motion to an upward motion at a faster rate. Therefore, higher CoM accelerations are expected when walking at a higher speed. In line with this, both the maximum forward deceleration and the maximum vertical acceleration examined in this study increased significantly with increasing walking speed (Max $\ddot{z}_{CoM}$ and Max $-\ddot{x}_{CoM}$ in Figure 4.8).

More work needs to be done and at a higher rate to produce these higher accelerations. As expected, the negative work done on the CoM by the user's leg in this phase was found to increase significantly with increasing walking speed ((1) in Figure 4.7). As mentioned previously, this work was done mostly by the ankle and the knee. The higher rate of work done at higher walking speeds led to increased peak absorption powers at the ankle (A1 in Figure 4.11) and the knee (K1 in Figure

4.11). Increasing walking speed also increases the maximum hip extension moment and the maximum knee extension moment during the collision phase (Figure 4.10). Higher forces are needed to produce these higher rates of work when redirecting the CoM as revealed by the higher ground reaction forces that occur when walking at higher speeds. These are in agreement with previous free walking studies that investigated the effect of speed on the dynamics and kinematics of gait (Zelik & Kuo, 2010; Fukuchi, Fukuchi & Duarte, 2018).

The amount of collision work done by the user's limb on their CoM increases significantly as the magnitude of the assistive force increases. This is to be expected as additional work is done against the assistive force when decelerating the CoM in the forward direction. However, there was also a significant interaction effect between walking speed and magnitude of the assistive force on the amount of collision work done ((1) in Figure 4.7). Furthermore, while the peak CoM forward deceleration was not significantly affected by force, the peak CoM vertical acceleration was not only significantly affected by the magnitude of the assistive force, but was also significantly affected by the interaction with walking speed, similar to the amount of collision work done (Max $\ddot{z}_{CoM}$ in Figure 4.8). This suggests that the additional collision work done was not solely to overcome the assistive force from the EW.

Zirker et al. (2013) pointed out that a horizontal assistive force may be able to contribute to elevating the CoM in this phase. During unassisted walking, muscular work is needed to raise the CoM's position. However, if an external horizontal force is provided, a person's foot can act as a pivot around which the CoM can rotate and the external horizontal force will generate a resulting moment that provides the energy to raise the CoM. A higher assistive force will generate a higher rotating movement that increases the vertical acceleration of the user. This may have decreased the total positive work that needed to be generated by the user. Higher walking speeds may have facilitated this pivoting motion, increasing the negative work produced by the user's leg at higher assistive force conditions but also leading to an increase in the vertical acceleration of the CoM. This mechanism, along with the lower physiological cost of negative work (Abbott, Bigland & Ritchie, 1952), may have contributed to the smaller increase in RPE from 15 N to 30 N at higher speeds as discussed in Section 4.4.4. This also explains the higher knee extension moment (Max Knee Moment in Figure 4.10) and higher peak negative power observed in the knee and ankle (A1 and K1 in Figure 4.11) with increasing assistive force. The knee joint extensors have to generate more force and absorb more of the power generated by the assistive force of the EW to keep the leg straight and acting as an inverted pendulum. The higher peak negative power in the ankle joint may be due to the need to absorb more power to maintain control of the pivoting movement.

Rebound & Pre-Load

In these two phases, if push-off work is equal to collision work, the stance leg could theoretically act as a rigid inverted pendulum and no additional work would need to be performed. In human walking, however, a relatively small amount of positive work is performed on the CoM by straightening the knee as the leg move towards mid-stance and negative work is performed by the ankle joint in the pre-load phase to store elastic energy in the Achilles tendon (Kuo, Donelan & Ruina, 2005; Kuo & Donelan, 2010).

The work done on the CoM by the user's limb significantly increased both in the rebound and the pre-load phase with increasing walking speed ((2) and (3) in Figure 4.7). Additionally, in the pre-load phase, peak hip joint negative power (H2 in Figure 4.11) was significantly affected by walking speed. These increases are in agreement with previous studies on free walking (Kuo, Donelan & Ruina, 2005; Zelik & Kuo, 2010). Increasing assistive force significantly decreases the work done on the CoM in the rebound phase but significantly increases the work done in the pre-load phase. Straightening of the knee in the rebound phase has the effect of raising the CoM. The user may have been able to utilize the assistive force to offset some of the work required to raise the CoM with a pivot mechanism similar to the one described previously in the collision phase. The increased negative work in the pre-load phase may be due to the storing of energy from the assistive force of the EW in the Achilles tendon to reduce the muscular work needed in the push-off phase.

Push-Off

In the push-off phase, the user's limb performs positive work on the CoM, mainly through plantarflexion of the ankle and this provides most of the energy used to move forward.

The push-off phase occurs simultaneously with the collision phase of the opposite foot. Hence, similar to the collision phase, higher walking speeds require more work to be done on the CoM and at a higher rate. This leads to the higher maximum CoM forward acceleration (Max $\ddot{x}_{CoM}$ in Figure 4.8) and the higher positive work done on the CoM ((4) in Figure 4.7) measured at higher speeds. This additional positive work at higher walking speeds was performed by the ankle and the hip as demonstrated by the significant increase in the positive ankle and hip powers (A2 and H3 in Figure 4.11) and the higher maximum plantarflexion moment and hip flexion moment at higher walking speeds (Max Plantarflexion and Max Hip Flexion in Figure 4.10).

Increasing the magnitude of the assistive force provided has the effect of decreasing the work done by the user's limb on the CoM ((4) in Figure 4.7). The increase in work done during collision and decrease work down during push-off is in agreement with studies that investigated the effect of an assistive force applied at the waist when walking on a treadmill, which showed higher peak breaking and lower propulsive impulses (Zirker, Bennett & Abel, 2013; Gottschall & Kram, 2003). This decrease in work done on the CoM in the push-off phase corresponded with a significant decrease in peak positive ankle power (A2 in Figure 4.11) as the magnitude of the assistive force increases. This suggests that the assistive force from the EW is used to perform the work on the CoM, partially relieving the ankle plantar flexor from having to propel the CoM forward and upwards.

However, the positive hip flexion power (H3 in Figure 4.11) increases significantly with increasing assistive force magnitude. In addition, there is a significant interaction effect for the maximum hip flexion moment. The significantly higher hip flexion power may be an adjustment to compensate for the lower plantarflexion power. Although the assistive force from the EW reduced the need to propel the CoM using the plantar flexor, it does not provide the energy required to swing the leg. Gottschall & Kram (2005) demonstrated that the energy required to swing the leg is not negligible and estimated that it makes up around 10% of the metabolic cost required for walking. Furthermore, it has been shown that hip flexion power is used to compensate for reduced ankle push-off (Lewis & Ferris, 2008).

4.4.6 Joint Angles and Posture

An increase in walking speed was found to lead to a significant increase in the maximum plantarflexion angle, the maximum knee flexion angle, and the maximum hip flexion angle and lead to a significant decrease in the maximum hip extensions angle (Figure 4.9). These changes with walking speed are consistent with previous studies on free walking (Fukuchi, Fukuchi & Duarte, 2018). The increased stride length with increasing walking speed (Figure 4.5) may have led the adjustments observed. The increase in the maximum hip flexion angle allowed for a longer step. Also, the increase in maximum plantarflexion angle and knee flexion angle in the swing phase just after ankle push-off may have been an adjustments to maintain ground clearance. When taking longer steps, the height of the CoM at the start of the swing phase is lower due to the longer arc length of the inverse pendulum trajectory of the stance leg. Hence, these may be adjustments to avoid scuffing the floor.

For the upper body, the mean angles were the same for all speed conditions except for the elbow joint. The mean elbow angle was found to be lower as walking speed increased. This suggest that the EW was positioned slightly further from the user's body when walking faster. The trunk, shoulder, and elbow angle in the sagittal plane were also found to oscillate in phase with the gait cycle. The timing or phase of the trunk oscillation were found to be consistent with that in free walking (THORSTENSSON *et al.*, 1984). The RoM, which is related to the amplitude of the oscillation, increased for the elbow, shoulder, and trunk angles as the walking speed increased. The result for trunk angle differed with that found by previous studies for free walking where the peak-to-peak displacements decreased as walking speed increased (Bruijn *et al.*, 2008; THORSTENSSON *et al.*, 1984). It may be possible that this, along with the oscillations of the elbow and shoulder, are the user's adjustments to allow their CoM to oscillate independent of the EW by varying the distance between the EW and their body. This increase in RoM may also have been due to the lower CoM height at the start of the stance phase (Phase (1)) when taking longer steps as mentioned in the previous paragraph. Because the handle height is constant and the shoulder would be at a lower height (corresponding to the lower CoM height) at this instant, elbow flexion would be at its maximum. This is shown in Figure 4.12 and higher elbow flexion was observed for higher walking speeds.

An increase in the magnitude of the assistive force also has a significant effect on the maximum flexion/extension angles of the lower limb. Maximum dorsiflexion angle increased significantly with increasing force but that for plantarflexion decreased significantly. These may be due to the decreased joint power of the ankle push-off (A2 in Figure 4.11). The increased dorsiflexion angle may have reflected the reduce work performed by the ankle while the decreased plantarflexion angle may be due to the lower momentum generated due to the lower ankle push-off power. Increasing force also led to decreased maximum hip flexion angle during the swing phase. This may also be due to the decrease ankle power during push-off that lead to lower momentum to swing the leg. Although higher power is observed for hip pull-off, it may generated less momentum than when lower assistive force is provided.

Increasing the magnitude of assistive force significantly increased the RoM for shoulder and trunk but not the elbow. In response to the assistive force from the EW, the user will need to flex their trunk and shoulders. The increased RoM may be an adjustments that enables the user to pre-emptively pull the EW at certain gait phases. In addition, the only mean angle affected by the increase in force was the shoulder angle. The users walked with increased shoulder flexion with increasing force magnitude. This suggests the user is pulling the EW closer to their body when an assistive force is applied.

4.5 Conclusion

In conclusion, an assistive force significantly decreases the positive work done on the CoM in the push-off and rebound phase but significantly increases the negative work done on the CoM in the collision and pre-load phase. The significantly higher vertical acceleration of the CoM at higher magnitude assistive force when walking faster suggests that users utilizes the assistive force to raise their CoM in the collision phase by using their standing leg as a pivot. This leads to increased knee extension power during collision, decreased ankle push-off power and increase hip flexion power during pre-swing. Additionally, the effect of increasing the magnitude of the assistive force on the perceived exertion of the users can be affected by their walking speed.

Chapter 5 General Discussions

General Discussions

5.1 Main Findings

This research investigated the gait adjustments to assistive forces from a Smart Walker through three open-loop experiments as presented in Chapter 2, Chapter 3, and Chapter 4 where participants walk with a custom-developed Smart Walker as different forces were applied and various measurements were made. The main contributions of this research are summaries as answers to the research questions shown below.

Q1 Can a constant force be used to increase walking speed?

The results from Chapter 2 showed that increasing the assistive force applied by a Smart Walker would increase its user’s walking speed. Furthermore, in Chapter 4, it was shown that there was a significant interaction effect between the magnitude of the force applied and walking speed. This indicates that an assistive force can make it easier to walk at a higher speed.

Q2 Can a constant force be used to decrease exertion?

Chapter 2 showed that when a relatively low magnitude assistive force is applied, the perceived exertion of the user decreases. However, when stronger assistive force is applied, exertion increases. Although no significant effect was found in Chapter 4, the same pattern where a relatively weak assistive force reduces the perceived exertion could be observed. These indicate that it may be possible to decrease exertion using a constant horizontal assist force.

Q3 Do users adjust their spatiotemporal gait parameters to constant forces?

The results from Chapter 4 indicate that they do not. Although results from Chapter 2 may have suggested that these parameters varied with force, the experiment in Chapter 4 demonstrated that this was due to the increased in walking speed.

Q4 When a force is used to control speed, would the users match the speed of the device to minimize exertion?

Chapter 3 showed that when the speed target of a Smart Walker is similar to a user's preferred walking speed, most users would match the speed of the device instead of working against the device. However, if the speed target differs from their preferred walking speed, more users will choose to overpower the device to make it move at a more comfortable speed the more different it is from their preferred walking speed. Furthermore, even when they matched the speed of the device, they would push/pull on the device with a pushing/pulling force proportional to the difference between their preferred speed and the target speed.

Q5 Can forces be used to improve posture?

The mean trunk angle when using a Smart Walker was not affected by assistive forces as shown in Chapter 4.

Q6 How are the joint moments affected by a constant force?

An assistive force increased the maximum knee extension moment during loading but had no significant effect on ankle plantarflexion and hip flexion/extension.

Q7 How are the joint powers affected by a constant force?

An assistive force will decrease ankle push-off power, increase knee extension powers in the collision and push-off phases, and increase the hip flexion power during pre-swing.

5.2 Gait Adjustments to Assistive Force

During the double support phase, the leading leg performs negative work on the CoM in its collision phase while the trailing leg performs positive work on the CoM its push-off phase to redirect the CoM. When an assistive force is applied, the negative work in the collision phase increases while positive work in the push-off phase decrease. However, in the collision phase, this increase in negative work is more pronounced when walking at higher speeds. The same pattern occurs in the vertical CoM acceleration in the collision phase where higher walking speeds led to more pronounced increases when an assistive force is applied. This suggests the user is adjusting his or her gait to utilize the force applied to perform work to raise its CoM. In this phase, the leading leg acts like a rigid pendulum that pivot around the ankle, converting the horizontal assistive force into a rotating moment about the ankle. This rotating moment performs work to accelerate the CoM vertically as well as horizontally. The increased work done by the leading leg to counteract the assistive force leads to increased vertical acceleration of the CoM. On the other hand, the work performed by the assistive force to accelerate the CoM horizontally does not increase the acceleration, but instead reduces the work done by the ankle of the trailing leg during the push-off phase.

To keep the leading leg rigid while maintaining upright posture, the knee extension joint moment and hip joint moment in the collision phase increases. In addition, more power is absorbed by this knee extension joint moment as well as by the ankle joint. On the other hand, the reduced ankle push-off power leads to increased power generated during hip pull-off. This increased power from the hip flexors compensates for the loss in energy from ankle push-off that was used to swing the leg.

The increased vertical acceleration of the CoM means that less time is needed to redirect to CoM during the double support phase. Hence, the duration of the double support phase decreases with increasing assistive force. Given that cadence is not affected by the magnitude of assistive force if walking speed does not increase, the relative duration of the double support phase (as a percentage of the gait cycle) also decreases. If the relative duration of the single support phase is assumed to remain constant, this decrease in double support phase leads to a decrease in the relative duration of the stance phase.

While walking at higher speeds lead to higher increases in negative work due to the assistive force as mentioned earlier, the data also suggest it may also lead to higher decreases in positive work during push-off (p = .09 in Figure 4.7). Furthermore, performing negative work have been known to be more metabolically efficient than that for positive work (Abbott, Bigland & Ritchie, 1952). These indicate that it may be more metabolically efficient to walk at higher speeds when an assistive force is provided. Therefore, users would choose to walk at higher speeds when an assistive force is provided by a Smart Walker.

5.3 Implications

It has been demonstrated that even when using a simple constant assistive force, users will adjust their gait to walk faster. This can help overcome mobility difficulty in the time constraints dimension discussed in Section 1.1.2. However, Chapter 4 shows that this comes at the price of higher negative work done by the user. In addition, the user may have to make undesirable adjustments to perform this negative work such as increased knee extension during the loading phase.

The user's perceived exertion during use was also shown to decrease when a relatively low magnitude of force was applied. The reduced exertion enables the user extends the range in which the user is able to walk, thereby improving mobility is another dimension; the minimum walking distance shown in Figure 1.1. Similar to the increase in walking speed, this was achieved using only a constant force. This indicates that if the assistive force is only applied at appropriate times, such as during push-off, the effectiveness might be much higher.

Another approach to reduce the effort or change the walking speed of a user is by controlling the speed of device. Chapter 3 showed that this is a viable method to increase the users walking speed. Three quarters of users were found to match the speed set to the device. Even when the users did not match the speed, their walking speeds would increase when the speed setting was higher than their preferred walking speed. However, this can come at the cost of exertion. The minimum ratings of perceived exertion reported was found to occur at the preferred walking speed and the further away the speed set is from their preferred walking speed, the harder the users will push/pull on the device.

In addition, understanding the user's gait adjustments to the assistive forces can provide the opportunity to use these assistive forces to improve the gait of the user. Alkjaer *et al.* (2006) demonstrated that when walking with a four-wheeled walker, a user would have lower ankle and knee joint moments compared to free walking. They suggested this might have long-term functional consequences. In Chapter 4, the ankle plantarflexion moment was shown to increase with speed but not change with force. Hence, the assistive force may be used, for example, to encourage a higher walking that increases ankle plantarflexion. Furthermore, knee joint moment during the loading phase was found to increase with increasing assistive force. This can be used to counteract the effect of decreased knee joint moment caused by the user of four-wheeled walkers. Both the use a four-wheeled walker and ageing has been found to lead to increase in double support phase (Herssens *et al.*, 2018; Liu *et al.*, 2009). Assistive force have been shown to decrease the double support phase and may be used to counteract this effect.

5.4 Limitations

This research focused on the gait characteristics of the users. However, another important aspect that needs to be considered is the interaction between a Smart Walker and its user. In this research, we were not able to measure the interaction forces between user and the Smart Walker. Hence, we did not know whether the user was pulling or pushing on the device. Furthermore, the magnitude of the assistive force used to in this research refers to force generated by the motor. The actual force experience by the user may vary based on the device (weight, rolling friction etc.) and the instantaneous momentum of the device.

In this research, only young healthy male adults participated in the experiments. This had a number of benefits. Firstly, it allowed us to first isolate the effects on healthy gait before layering in subject-specific impairments. Secondly, use of young and healthy participants allowed us to perform the experiment using higher forces that may have been considered too risky or overly strenuous for the actual target users of walkers. Nonetheless, the young adults are not the target audience of these mobility aids. Therefore, future studies should also include participant who are the actual target users for Smart Walkers.

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