

Simulating Ideal Assistive Devices to Reduce the Metabolic Cost
of Walking in the Elderly

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ABSTRACT

As the population ages, decreased mobility and loss of independence becomes a concern for many elderly individuals. Walking assist exoskeletons offer a solution for restoring, enhancing, and maintaining mobility. However, as this technology is a growing area of study pertaining to the elderly population, there is not currently a unified standard as to the optimal strategy for assisting elderly gait. The gait patterns of elderly individuals differ from that of the younger population primarily in the ankle and hip joints. This study used musculoskeletal simulations to investigate how hypothetical ankle and hip actuators affected the metabolic cost of elderly participants during gait. Using OpenSim as a modelling tool, simulations were generated of 10 elderly participants walking at a self-selected comfortable speed. Ideal flexion and extension assistive devices were then added to the ankle and hip joints of the musculoskeletal models to simulate the subsequent metabolic savings. The resulting simulations suggest that providing hip assistance to elderly participants results in significantly greater metabolic savings compared to ankle assistance. Compared to the unassisted scenario, the use of an ideal hip actuator resulted in $25.5 \pm 7\%$ metabolic savings, whereas use of an ideal ankle actuator resulted in $14.2 \pm 2\%$ metabolic savings. Our results can help researchers determine which joint to target when developing WAE for elderly users in the future.

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CHAPTER 1: INTRODUCTION

1.1 INTRODUCTION

Over recent years, the proportion of elderly individuals within the population has been steadily increasing. According to Statistics Canada, the elderly population is expected to represent 23% to 25% of the population by 2036, and 24% to 28% by 2061 [1]. As the elderly population increases, the expected decrease in mobility associated with aging becomes a great concern, as it will inevitably impact society and the healthcare system. Impairments in mobility can lead to various negative consequences, such as decreased ability to perform daily living activities, which can lead to subsequent decreases in independence and quality of life. Without intervention, the aging population will likely lead to a burden on healthcare resources due to the increased need for daily care and support for those with reduced mobility [2]. These societal concerns have led to the increased research and development of assistive devices to enable the elderly population to remain independent.

Wearable walking assist exoskeletons (WAE) have become a promising technological solution to current mobility impairments. WAEs are mechanical devices that are designed to reduce the metabolic cost of walking, improve walking performance, increase walking speed, or restore function [3]. However, as it is still an emerging field, the design process has not yet been standardized. Current WAE designs intended towards the elderly population aim to assist random joints of the lower body, indicating that there is not currently a unified standard pertaining to gait assistance, particularly of healthy elderly users. For example, the Honda Stride

Management Assist [4] is a powered WAE that applies assistive torques to the hips of the user, whereas Keeogo by B-Temia is a powered WAE that applies complementary torques using motors aligned at the knee joints [5]. Although not commercially available, the Bilateral Ankle-Foot Exoskeleton developed by Malcolm et al. was designed to assist plantarflexion during the push-off phase of the gait cycle [6]. Additionally, in many cases, WAE are not designed specifically for healthy elderly users. This technology was originally founded for power augmentation for soldiers and recently adapted for mobility restoration for paraplegic patients [2]. On the other hand, aging is accompanied by specific gait changes, particularly in the ankle and hip joints. Therefore, assistive strategies targeted at these changes will likely lead to optimal assistance of elderly users. Assisting the ankle may be a suitable approach to repair gait, as this strategy tackles the underlying cause of gait deterioration in healthy elderly individuals. Alternatively, the hip can be assisted to facilitate the compensation strategies that are commonly observed in elderly gait.

Evaluating the performance of exoskeletons and implemented gait strategies commonly requires researchers to design and manufacture several prototypes, followed by the collection of experimental gait data, which can be costly, time consuming, and potentially invasive for the participants. These challenges have created the need for a method of modelling the musculoskeletal system, to provide researchers with valuable insights into exoskeleton design and the performance of gait strategies. Designed to complement experimental research, computational musculoskeletal modelling is an invaluable tool for developing predictive models that can accelerate the analysis process.

1.2 OBJECTIVE

The following thesis aims to study the effect of ankle and hip actuation strategies within an ideal WAE on the gait of healthy elderly participants. Using OpenSim as a modeling tool, the study will determine which of the proposed assistive strategies is optimal for healthy elderly gait in order to enhance the future design and development of WAE for elderly users. An optimum assistive strategy should result in reduced metabolic cost and ultimately, enhanced elderly mobility and independence.

1.3 CONTRIBUTIONS

This thesis has led to the following contributions.

1. A comprehensive literature review was conducted to establish the current state of walking assist devices for healthy elderly users and to investigate potential assistance strategies specific to the elderly population. Based on typical gait changes that occur with age, assistance at the ankle and hip joints were further investigated.
2. Using OpenSim as a musculoskeletal modelling tool, the effects of adding an ideal bilateral ankle and hip assistive device to elderly gait were investigated. Metabolic power was evaluated during the various assistance scenarios to compare the use of hip and ankle devices, with hip assistance resulting in a greater reduction of metabolic power compared to ankle assistance.

3. Based on results obtained from the musculoskeletal simulations, recommendations for future research and development are provided for walking assist exoskeletons for elderly users.

1.4 THESIS OUTLINE

This thesis is organized into four chapters. Chapter 1 describes the motivation behind the investigation of gait assistance strategies in the elderly population, and the objectives of this thesis.

Chapter 2 presents the results of a literature review describing elderly gait biomechanics and how they differ from the younger population, the current state of WAE and their use within the elderly population, and the use of simulation software to supplement experimental studies.

Chapter 3 discusses the chosen musculoskeletal model and data preparation process. This is followed by a description of the OpenSim workflow that was followed in this study, including Scaling, Inverse Kinematics, Residual Reduction Algorithm, and Computed Muscle Control.

Chapter 4 presents the results obtained from the simulations and a discussion of the results, as well as limitations to the study, areas for future work, and final conclusions.

CHAPTER 2: LITERATURE REVIEW

2.1 GAIT ANALYSIS

Gait is an essential component of human movement, involving the hip, knee, and ankle joints to propel the body forward in a bipedal motion. Gait analysis defines the quantitative study of human walking. It views gait as a synchronized, controlled, and cyclic sequence of limb movements. Gait analysis involves the time interval between two successive occurrences of the repetitive events of walking, referred to as one gait cycle [7]. Figure 1 illustrates the events of one gait cycle.

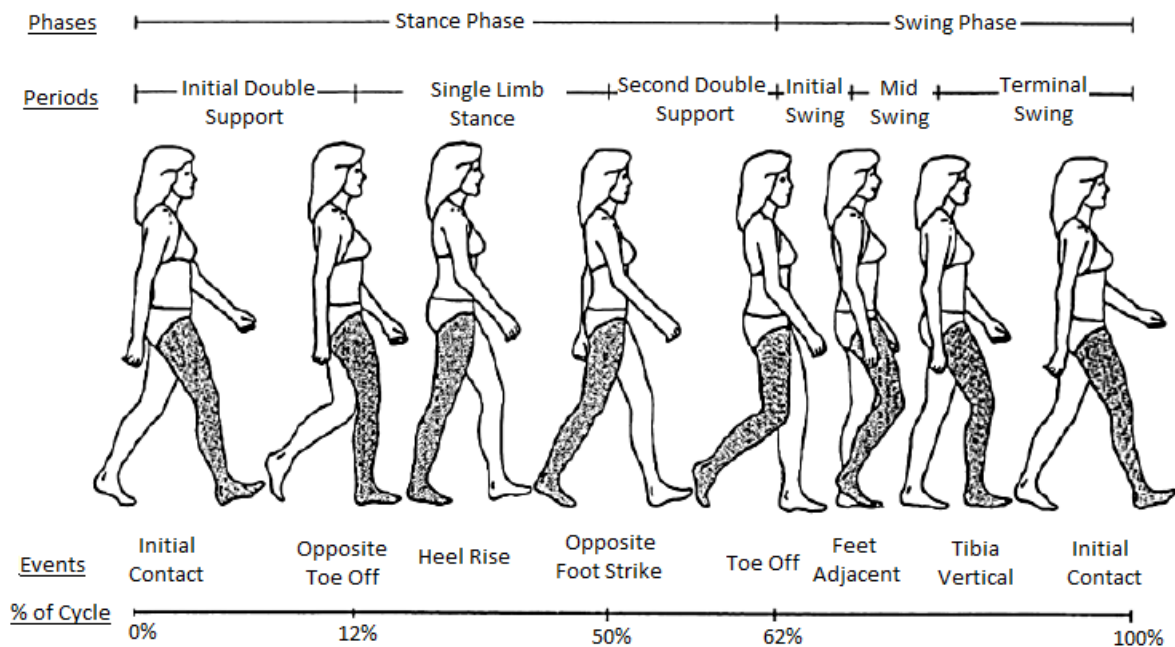


Fig 1. Illustration of gait cycle of right leg. Adapted from [8]

Heel strike, which is when the heel makes initial contact with the ground, is often used to signify the start and end of the gait cycle. One gait cycle includes both a stance and a swing phase. Stance phase is the interval when the foot is in contact with the ground, which makes up 60% of the gait cycle, whereas the swing phase is the interval when the foot is in the air, making up 40% of the gait cycle. The stance phase is divided into three periods: initial double limb support, single limb support, and second double limb support. Swing Phase is also divided into three periods: initial swing, mid swing, and terminal swing [8]. Table 1 describes the details of each phase.

Table 1. Description of Gait Cycle Phases [8]

Period	% Gait Cycle	Function
Initial Double Limb Support	0 – 12	Loading of support leg Transfer of weight between legs
Single Limb Support	12 – 50	Support of body weight Forward progression of center of mass
Second Double Limb Support	50 – 62	Unloading of support leg Preparing for swing phase
Initial Swing	62 – 75	Foot clearance
Mid Swing	75 – 85	Advancement of limb
Terminal Swing	85 – 100	Limb deceleration Preparation for weight transfer

2.2 ELDERLY GAIT BIOMECHANICS

As shown in Figure 2, the typical strategy for young, healthy gait is characterized by the concentric contraction of the ankle plantar flexor muscles. The rapid rotation of the foot pushes the leg forward, advancing it into swing phase [9]. However, as humans age, the biomechanical components of gait undergo changes due to the physiological changes that accompany the natural aging process. This often leads to loss of mobility in the elderly population. One of the most prominent physiological changes that occur in elderly individuals is loss of muscle mass and muscle strength. This loss may be attributed to loss of motor neurons, muscle fibers, or aerobic capacity [10].

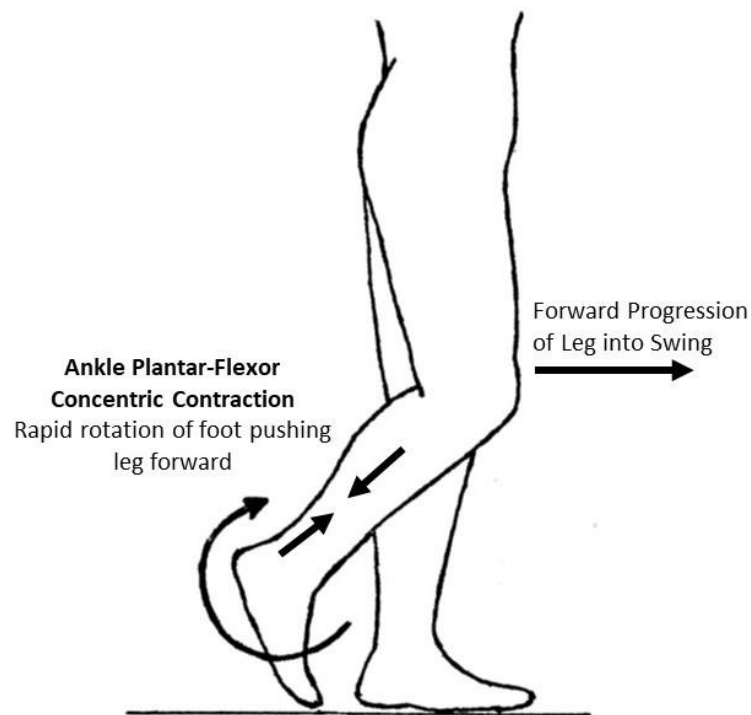


Fig 2. Ankle plantarflexion gait strategy. Adapted from [9]

Another physiological change that affects the mobility of elders is the structural weakening of lower limb joints. A lifetime of joint loading can result in degeneration of articular cartilage leading to diseases such as osteoarthritis, which reduce joint range of motion (ROM) [10]. Typically, age-related gait changes occur between 60 and 70 years old [11].

2.2.1 ELDERLY GAIT KINEMATICS

Kinematic variables are involved in the description of movement, independent of the forces that cause the movement. Kinematic variables include displacement, velocity, and acceleration [11]. Elderly gait exhibits significant kinematic changes compared to the gait of young adults, as summarized in Table 2. Elderly gait is typically characterized by slower speed, smaller stride length and increased stance duration [9] [12]. However, cadence values remain unchanged between young and old adults [13]. The joint angles and ROM in the lower limbs of elderly individuals also exhibit changes compared to young adults. Ankle ROM is lower in the elderly compared to younger adults, decreasing from 29.3° to 24.9°. Knee ROM also decreased from 59° to 55° in the elderly. Inversely, elderly hip ROM increased from 32° to 40°, along with increased anterior pelvic drop to accommodate for increased hip extension [10]. Judge et al. reported a decrease in elderly peak plantarflexion angle, with a value of $13.5 \pm 5^\circ$ compared to $17 \pm 5^\circ$ in young adults [12].

Table 2. Summary of kinematic changes between young adults and elderly adults [10] [12].

	Young Adults	Elderly
Gait Velocity (m/s)	1.16 ± 0.13	1.03 ± 0.13
Step Length (proportion of leg length)	0.74 ± 0.04	0.65 ± 0.07
Single Support Time (proportion of gait cycle)	40 ± 2	37 ± 3
Ankle ROM	29.3°	24.9°
Knee ROM	59°	55°
Hip ROM	32°	40°
Peak Plantarflexion angle	17 ± 5°	13.5 ± 5°

2.2.2 ELDERLY GAIT KINETICS

Kinetic variables are the forces that cause movement [11]. The kinetic components of both the ankle and hip exhibit age-related changes during gait. As summarized in Table 3, Winter reported a change in plantarflexion moment, decreasing from 1.63 N/m/kg in young adults to 1.44 N/m/kg in elderly [14]. Studies have reported a decrease in plantarflexion power in elderly gait, however these values vary throughout the literature [9]. According to Winter, elderly plantarflexion power was 2.48 W/kg compared to 3.27 W/kg in young adults [14]. Judge reported a similar value of 2.9 W/kg in the elderly compared to 3.5 W/kg in young adults, whereas Prince et al. reported a smaller value of 1.62 W/kg in elderly [10] [12]. The work done by ankle plantar flexors was shown to decrease in elderly gait with a value of 0.190 J/kg compared to 0.293 J/kg in young adults [10]. By age 70, Larsson et al. reported a 20-40% reduction of skeletal muscle strength [15]. Therefore, the diminished ankle power observed in elderly gait during push-off may be attributed to a reduction in skeletal muscle strength [12].

Table 3. Summary of ankle kinetic changes between young adults and elderly adults [10] [14]

	Young Adults	Elderly
Plantarflexion Moment (Nm/kg)	1.63	1.44
Plantarflexion Power (W/kg)	3.27	2.48
Plantarflexion Work (J/kg)	0.293	0.190

In addition to ankle-related gait changes, hip kinetic changes have been observed in the elderly. However, there have been inconsistent findings as to the effect of aging on the hip during gait. Judge et al. reported increased hip flexor power and moment during late stance compared to young adults [12]. However, increased hip extensor moment and power were reported by DeVita and Hortobagyi, and Kerrigan et al. [16] [17]. Figure 3 compares the ankle and hip power of elderly and young adults during the gait cycle.

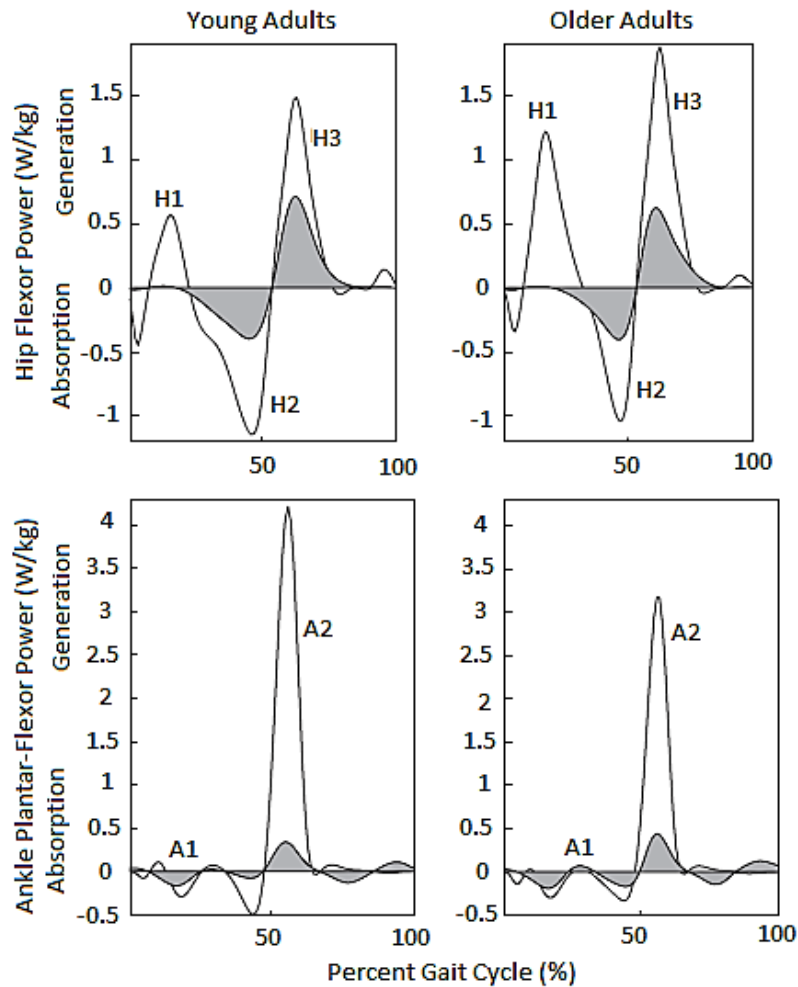


Fig 3. Comparison of hip flexor power and ankle flexor power in young and older adults. The shaded area represents passive contributions to joint power. Power absorption occurs at hip flexors (H2) and ankle dorsiflexors (A1). Power generation occurs via ankle plantar flexors (A2), hip extensors (H1), and hip flexors (H3). Adapted from [18].

2.2.3 ELDERLY GAIT STRATEGY

The gait changes that typically occur with aging lead to differing gait strategies between the young and elderly population. Elders often experience diminished musculature, particularly in the plantar flexor muscles. As a result, pelvis and trunk muscles are engaged to pull the leg into swing phase, since the plantar flexor muscles are too weak to produce a vigorous push-off

force. The hip power generation requirements during gait are generally less than that of the ankle relative to maximal capacity, suggesting that hip musculature has the capacity to compensate for the decreased ankle power [18]. The result is a switch from an ankle gait strategy to a hip gait strategy [12].

Two hip strategies have been proposed in the literature. The hip flexor strategy involves the use of the hip flexor muscles to pull the leg into swing phase. As shown in Figure 4, the hip flexor concentrically contracts during the late stance phase of the gait cycle, rotating the thigh forward and assisting with leg advancement.

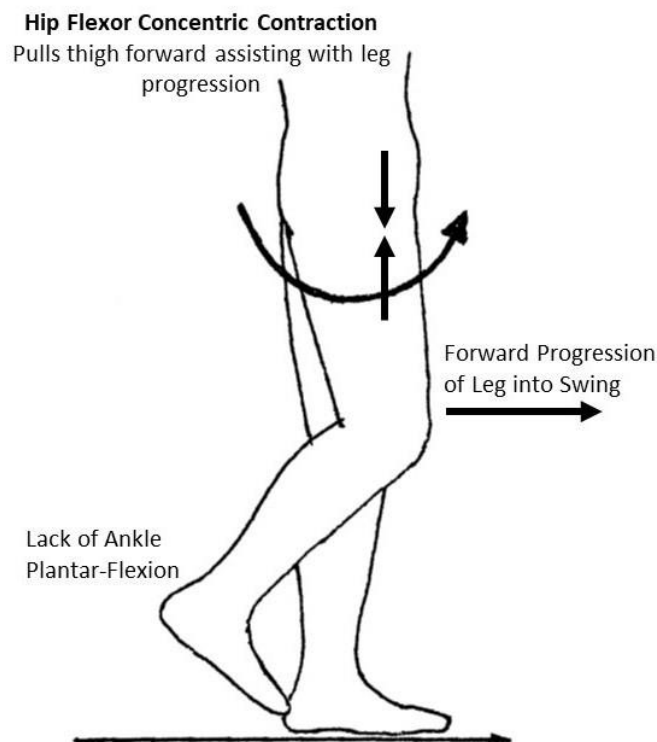


Fig 4. Hip flexor gait strategy. Adapted from [9].

Alternatively, the hip extensor strategy involves the hip extensor muscles pulling the leg into swing phase, as well as stabilizing the trunk. As shown in Figure 5, the concentric contraction of the hip extensors during late stance results in backward rotation of the pelvis, assisting with forward thigh rotation and leg advancement [9].

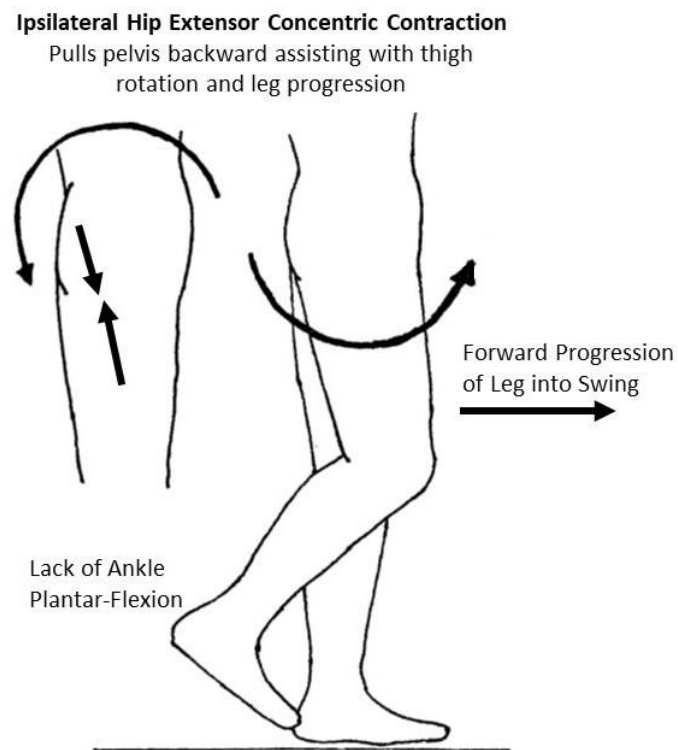


Fig 5. Hip extensor gait strategy. Adapted from [9].

2.3 ASSISTIVE DEVICES

As the biomechanical changes associated with aging often lead to decreased mobility, use of assistive devices allow the elderly to retain their independence. Assistive devices are tools that can help a user improve their mobility. This can be accomplished in several ways, such as by broadening a user's base of support or improving their stability. They can also be used to reduce lower limb loading by redistributing weight to compensate for weakness or generating movement by providing moment assistance to joints. Use of assistive devices leads to improvements in user's confidence, safety, and independence. They also result in various physiological benefits, such as improved cardiorespiratory function and enhanced circulation [19]. Assistive devices are used among a wide range of patient groups; individuals who have spinal injuries, experienced a stroke, or have conditions such as osteoarthritis are a few examples. However, as such a large proportion of the population is reaching an elderly age, assistive devices are becoming increasingly used to assist individuals who are considered healthy but are experiencing the normal gait deterioration that is associated with aging.

2.3.1 WALKING ASSIST EXOSKELETONS

Walking assist exoskeletons are mechanical devices that assist in locomotion. They are designed to be worn by the user, reducing the metabolic and physiological cost associated with walking. This type of exoskeleton is useful for individuals who can move independently but would benefit from the assistance of a WAE, making healthy elderly individuals an ideal candidate for this type of assistive device. The ultimate goal of WAE is to reduce the metabolic cost of walking,

improve walking performance, increase walking speed, or restore function. WAE are often classified as either powered or passive. Powered WAE involve the use of portable energy sources and actuators to convert electrical, pneumatic or hydraulic energy into mechanical work [20]. Examples of powered WAE targeted towards or tested on elderly users include the Honda Stride Management Assist [21], Keeogo by B-Temia [22], and Bilateral Ankle-Foot Exoskeleton by Malcolm et al [6].

2.3.2 EXISTING POWERED WAE

The Honda Stride Management Assist applies assistive torques to the hips of the user. As shown in Figure 6, the device consists of a hip frame, which incorporates the control computer and battery. The central pattern generator controls two electric motors, which generate simultaneous and complementary torques to the thighs of the user via thigh frames. The device requires walking to be initiated by the user, after which the extensor torque of the device initiates, reaching its peak just prior to mid-stance. The Stride Management Assist then switches to a flexion assist during terminal stance, with flexion torque reaching its peak around initial swing. Finally, the device switches to extension assist during terminal swing. While the device provides torque assistance exclusively in the sagittal plane, it does not restrict user movement in other directions [23]. The Honda Stride Management Assist device weighs 2.7 kg and is easily adjustable to accommodate a variety of users. Use of this device has been shown to reduce the metabolic cost of users by 7.06% while walking at a comfortable speed and 10.52% at maximum

walking speed [4]. Use of the Stride Management Assist has also resulted in an increase in user walking speed and stride length compared to baseline [23].



Fig 6. Honda Stride Management Assist [24]

Keeogo by B-Temia is a powered WAE that utilizes sensors at the knee and hip joints, as shown in Figure 7 [22]. Similar to the Honda Stride Management Assist, Keeogo requires users to initiate movements. It functions by actively assisting or resisting during key phases of movement. The sensors identify and interpret the movement that the user is trying to perform. Complementary torques are then applied by the motors aligned at the knee joints to assist movement [22]. Keeogo is designed for individuals who have mobility issues or limited walking endurance, but are still ambulatory, as it does not support or constrain the feet, pelvis or torso [25].



Fig 7. Keeogo Dermoskeleton by B-Temia [25]

The Bilateral Ankle-Foot Exoskeleton developed by Malcolm et al. was designed to assist plantarflexion during the push-off phase of the gait cycle. As shown in Figure 8, the WAE design includes a pneumatic muscle along the dorsal side of the leg, a load cell used to calculate pneumatic muscle force, a linear displacement sensor to calculate ankle joint angle and pneumatic muscle moment, and a footswitch to detect foot contact [3]. Although originally developed and tested on young adult participants, successful assistance of healthy elderly users was demonstrated in a later study with a 12% reduction in metabolic cost [3].

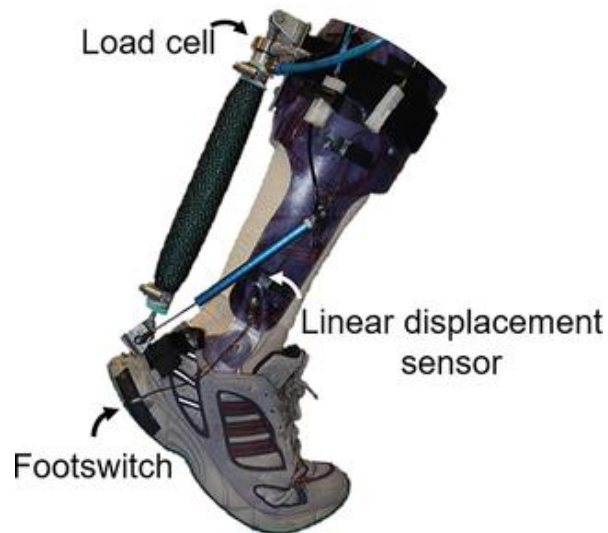


Fig 8. Bilateral Ankle-Foot Exoskeleton by Malcolm et al. [3]

2.3.3 PASSIVE WAE

While the above WAE can effectively assist a user's movement, these products tend to be expensive, making them inaccessible to many elderly users who require mobility assistance. Additionally, powered WAE designs may be bulky and complex, with a limited operating time if they rely on batteries. Alternatively, passive exoskeleton designs offer a solution that is often more simplistic and inexpensive, making them an attractive option for introduction into everyday life. Passive devices do not contain an external power source, instead using materials such as springs, dampers, and mechanical clutches to assist movement. These designs attempt to improve the energy exchange between the potential and kinetic energy of body segments, therefore reducing the energy consumed by the muscles to produce motion [20]. The design components can store energy from human movement and release the energy when needed [26]. Passive devices can be beneficial to users due to their low mass and autonomy, as they do not

contain any powered actuators or energy sources, such as battery packs [27]. However, the assist device is an added weight to the body and as there is no powered actuator, the challenge is developing a passive exoskeleton that effectively reduces metabolic cost to the user. The XPED2 and the Passive Ankle Exoskeleton by Collins et al. are examples of passive WAE.

The XPED2 is a passive WAE designed to reduce lower limb torques during gait. As shown in Figure 9, this design incorporates the use of exotendons. Exotendons are elastic cables that run parallel to the user's legs. They store and transfer elastic energy between lower limb joints, mimicking the function of tendons [28]. XPED2 was developed to test the prediction that a triarticular exotendon (hip, knee, and ankle) can reduce joint moments by 46% [29]. The exotendon extends from a lever at the pelvis to a leaf spring at the ankle, via a pulley at the knee. This configuration allows the cable to loosen during hip flexion and ankle plantarflexion and tighten during hip extension and ankle dorsiflexion. Therefore, as the user transitions between heel strike and midstance, the exotendon stretches, storing energy to be released during terminal stance [29]. The total weight of this device is 6.91 kg. When tested on young adult participants, a reduction of 12.1% in average absolute joint torque was measured when participants used the XPED2 with exotendons.

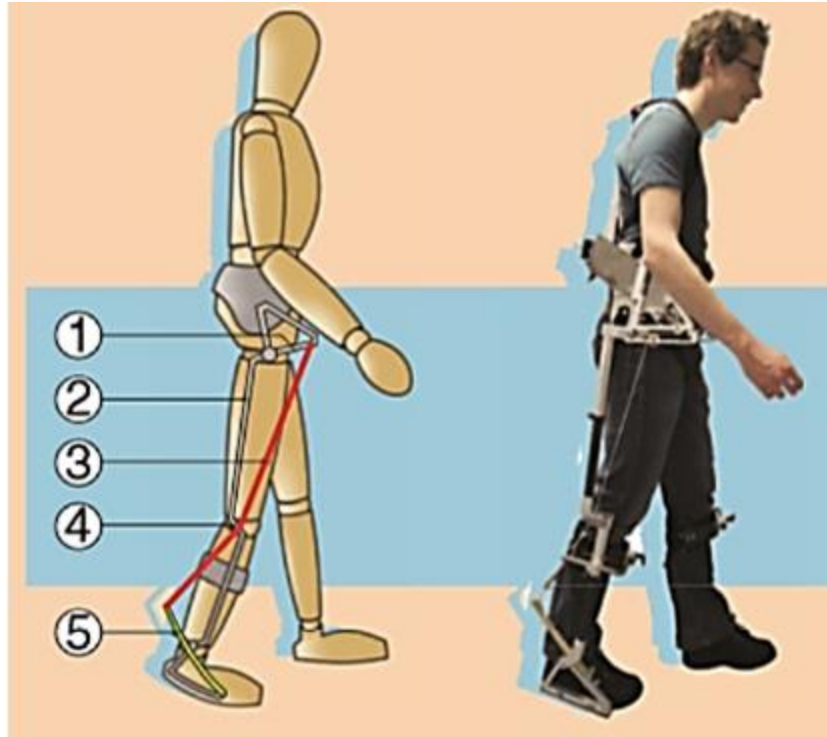


Fig 9. Schematic of the XPED2 exoskeleton (left) and a photograph of the XPED2 being worn by a user (right). 1) pelvis lever, 2) rigid frame, 3) extendon, 4) knee pulley, 5) foot leaf spring [29].

The Passive Ankle Exoskeleton shown in Figure 10 was developed by Collins et al. to reduce the energy cost of walking [30]. It is designed with a spring parallel to the Achilles tendon, connected to the leg with a lightweight frame and lever at the ankle joint. When the foot is on the ground, a mechanical clutch engages the spring. It then disengages when the foot is in the air, allowing for free movement of the foot. The Passive Ankle Exoskeleton was designed to be lightweight, with a mass ranging between 0.408 kg to 0.503 kg, depending on participant size. Testing was conducted on healthy, young participants, walking on a treadmill. When the exoskeleton was tested with various spring stiffnesses, a stiffness of 180 Nm (intermediate stiffness) resulted in a metabolic cost reduction of $7.2 \pm 2.6\%$ [30].

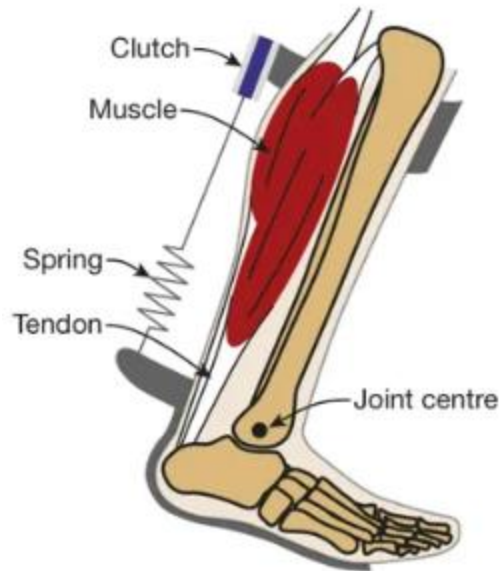


Fig 10. Schematic of Passive Ankle Exoskeleton by Collins et al. [30]

2.4 ELDERLY USE OF WAE

The above examples constitute only a few of the recent developments in the field of mobility assistive devices. As the elderly population increases, the development of technology to ensure that this demographic of the population remains independent is becoming a growing area of study and research. However, as it is still an emerging field, the research and design process has not yet been standardized. The aforementioned WAE examples target and assist different joints of the lower body, indicating that there is not currently a unified approach pertaining to gait assistance, particularly of healthy elderly users. Although recent WAE have demonstrated successful user assistance, there has not been a consistent gait strategy behind their design. Additionally, in many cases, WAE designs are not specific to healthy elderly users. As aging is accompanied by specific gait changes, particularly in the ankle and hip joints, assistive

strategies targeted at these changes will likely lead to optimal assistance. Therefore, it is likely that different design parameters must be considered to assist elderly gait.

Based on the typical gait changes associated with aging, there are two potential strategies that may be considered. Assisting the ankle may be a suitable approach to repair gait. This strategy tackles the underlying cause of gait deterioration in healthy elderly individuals. By assisting elders to increase plantar flexion power, their gait pattern may revert to the ankle strategy that is observed in the younger population. Exoskeleton assistance at the ankle joint may compensate for the reduced musculature in the plantar flexor muscles, allowing elderly users to generate enough power to walk efficiently. Alternatively, the hip can be assisted to facilitate the compensation strategies that are commonly seen in elderly gait. With the increased activation of hip flexor and extensor muscles observed in elderly gait, assisting these muscles may improve ability to maintain mobility. Assisting the hip may also reduce the stress on the hip joint, which could reduce susceptibility or prevent the worsening of common age-related joint conditions, such as osteoarthritis and bone weakness. Unlike the hip and ankle joints, changes at the knee joint are less prominent in the aforementioned differing gait strategies between young adults and the elderly [12]. Average power output contributed at the knee has been shown to be less than half of what is contributed at the ankle and hip during gait [31]. Further, at slow walking speeds, which is typical of elderly gait, less contributions to support and progression are a result of muscles at the knee (vastus muscles), with their contribution increasing dramatically with increased walking speed [32]. Therefore, assistance at the knee joint was not analyzed in this study, instead focusing exclusively on assistance at the hip and ankle.

The purpose of this study is to utilize simulation tools to determine which of the proposed assistive strategies is optimal for healthy elderly gait in order to enhance the future design and development of WAE for elderly users. An ideal assistive strategy should result in decreased metabolic cost, therefore reducing the required effort for elderly users to walk. Use of a WAE should lead to enhanced elderly mobility and independence, improving the ability to perform activities of daily living, and ultimately improving overall quality of life.

2.5 SIMULATIONS

The study of walking assist devices is a growing area of research. However, the development and testing of devices is a challenging process. The physical testing of exoskeletons requires researchers to design and manufacture several prototypes, as multiple versions are often needed to improve design parameters. Following device design and development, the collection of experimental gait data is then required to test the device on human participants. These gait trials are costly, time consuming, and potentially invasive. Additionally, obtaining data from experimental trials alone can lead to several limitations. It is difficult to measure certain variables, such as muscle forces, experimentally. Even when muscle force data is obtained, it is difficult to establish definitive cause-effect relationships. For example, electromyographic sensors (EMG) can detect when a muscle or group of muscles is activated during a specific movement [33]. However, they cannot determine whether the activation of these muscles produces the movement. When designing assistive devices, such as exoskeletons, understanding of the cause-effect relationship of muscles is imperative in order to recognize how particular muscle forces can be enhanced or compensated. These challenges have created the need for a

computational tool which can model the musculoskeletal system and provide researchers with valuable insights into exoskeleton design requirements, which they are unable to determine experimentally. Computational musculoskeletal modelling is an important tool, designed to complement experimental research by providing parametric and predictive models that can accelerate the analysis process. There are several commercially available simulation software packages that are typically used by researchers, such as *Visual 3D* (C-Motion Inc), *AnyBody* (AnyBody Technology), *BoB* (Biomechanics of Bodies) and *OpenSim* (SimTK).

Visual 3D is a dynamic simulation software, often used to analyze 3D motion capture data. It is designed to manage data from marker set systems in order to calculate joint angles and moments using inverse dynamics. However, *Visual 3D* is not able to compute muscle forces. *Visual 3D* musculoskeletal models are created by making individual body segments. These segments float in space and are able to apply forces to other segments [34].

AnyBody is a sophisticated software package that analyzes musculoskeletal systems as rigid-body systems, which allows for the calculation of multi-body dynamics. In addition to calculating inverse kinematics and inverse dynamics, it is also capable of calculating muscle forces. The anatomical properties of the default human model, such as height, weight, segment lengths, strength, and bone geometries, can also be adapted to suit the needs of the user. Another feature of this program is the ability to model the simulation environment. STL files can be imported into the program to simulate objects or assistive devices that would affect the model [35]. However, *AnyBody* must be purchased, which may be a limitation for some users.

BoB is a simulation software that contains the most simplistic interface of the tools examined. It allows users to calculate joint contact forces, joint torques, movement trajectories, and muscle activations [35]. *BoB* is written in MATLAB, which enables data to be easily exported into MATLAB, as well as allows for the use of MATLAB's analysis capabilities [36].

OpenSim is a freely available, open-source simulation software package that allows users to access and improve existing musculoskeletal models, as well as exchange and test new models. Similar to *Anybody*, *OpenSim* can compute inverse kinematics, inverse dynamics, and muscle forces. Additionally, the program can carry out optimization functions and predictive forward dynamic simulations [33]. Much of the *OpenSim* framework was adapted from *SIMM*, an older simulation platform. The purpose of *SIMM* was to analyze, edit, and develop musculoskeletal models. However, it was not able to compute muscle excitation and its source code was not transparent to users, making it difficult to improve [33]. Due to its computational functions, as well as its accessibility, *OpenSim* was chosen for this research.

CHAPTER 3: OPENSIM METHODOLOGY

3.1 OPENSIM

OpenSim is an open-source software package developed by Stanford University, which allows users to model, simulate, and analyze the neuromuscular system. As an open-source program, OpenSim acts as a public repository for musculoskeletal data, models, and computational tools [33]. This gives users the flexibility of editing existing or default models, as well as creating their own models. Newly developed models, controllers, and analyses can be uploaded to the program as plug-ins, allowing users to extend functionality. When OpenSim is downloaded, it includes these plug-ins, authored by different users, which contribute to the calculation of joint forces, muscle-induced accelerations, and muscle power [33]. Updated versions of OpenSim are periodically released, containing software improvements. In this study, OpenSim 4.0 was used. OpenSim 4.0 was downloaded from the OpenSim website [37].

3.1.1 OPENSIM MODELS

OpenSim comes equipped with several generic models, including models of the full body, upper extremities, and lower extremities. Additional models, contributed by members of the biomechanical modelling community, are available for download on the OpenSim Documentation website [38].

The model chosen for this study was Gait 2392. Gait 2392 is a musculoskeletal model that can be used to calculate muscle forces and joint loads in the lower extremities. It represents an unscaled subject with a height of 1.8 m and a mass of 75.16 kg. As shown in Figure 11, Gait 2392

consists of a head, torso, pelvis, and two legs [39]. The intended use of this model is for simulating and analyzing human movement that is dominated by the muscles of the lower extremities. Since this study focused on gait analysis, specifically related to changes in the ankle and hip, a model containing only the lower extremities was determined as adequate. The Gait 2392 model contains 23 degrees of freedom and 92 actuator forces, representing 76 muscles in the lower limbs. The model is limited in its ability to produce accurate results during motions with high degrees of knee flexion [38]. However, since this research does not simulate movements with high degrees of knee flexion, it was deemed an appropriate choice. Gait 2392 has been widely used and tested throughout literature. Use of a validated model is crucial to ensure that modelling results are representative of real-world biomechanics.

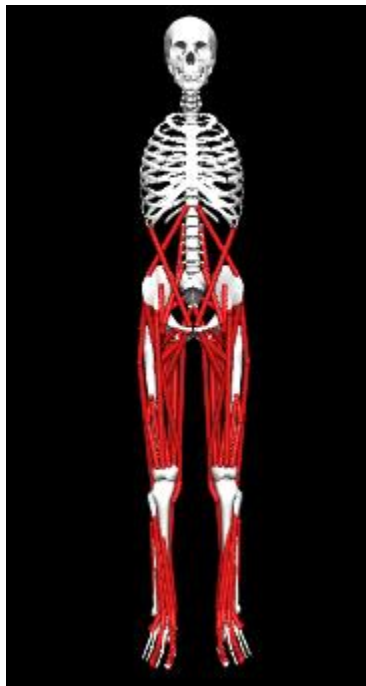


Fig 11. Unscaled Gait 2392 model

Gait 2354 is another generic model that is commonly used by researchers in lower-extremity focused simulations. It is similar to the Gait 2392 model, with identical unscaled subject dimensions and degrees of freedom. However, Gait 2354 is a simplified version of Gait 2392 and contains only 54 musculotendon forces. Its intended use is similar to that of the Gait 2392 model, however since it contains a reduced number of muscles, the simulation time is reduced, making it an ideal model for initial simulation prototyping [38]. For the purpose of this study, the difference in simulation times between the two models was acceptable. In addition, Gait 2392 contains several muscles that are of interest in this study that are not present in Gait 2354, leading to the selection of Gait 2392 [40].

Other models that were considered for this study include the Lower Limb Model 2010 by Arnold et al. and the Full Body Model 2016 by Rajagopal et al. The Lower Limb Model 2010 is similar to Gait 2392, with the addition of ellipsoidal wrapping surfaces. This allows the model to reflect muscle fiber operating lengths and muscle force generation properties more accurately. However, the addition of ellipsoidal wrapping surfaces significantly increases simulation time and is not suitable for the Computed Muscle Control (CMC) function [38]. The Full Body Model incorporates bony geometry for the full body, with 37 degrees of freedom, 80 muscle-tendon units actuating the lower body and 17 ideal torque actuators driving the upper body. This model's musculotendon parameters were derived from both the anatomical measurements collected from cadavers and magnetic resonance imaging (MRI) of young healthy adults [41]. Conversely, the aforementioned models' musculotendon parameters were derived exclusively from cadavers. However, the Full Body Model was developed more recently and as such, it has not been used as extensively throughout the literature. Gait 2392 has been widely used and cited

throughout the musculoskeletal modelling community; therefore, Gait 2392 was chosen for this study to allow for the comparison of results with existing studies.

3.2 MOTION DATA

The kinetic and kinematic data used in this study was accessed from a publicly available dataset, gathered by Fukuchi et al [42]. The dataset contains gait data from 42 individuals; 24 of whom are young adults and 18 older adults. Data was gathered using a motion-capture system (Raptor-4; Motion Analysis Corporation) and force plates while the participants performed overground and treadmill gait trials at self-selected slow, comfortable, and fast speeds. To capture motion data, 28 anatomical reflective markers were placed on the lower body of each participant. All participants were free of any lower-extremity injury in the six months prior to data collection, as well as orthopedic or neurological diseases which could affect their gait pattern [42]. As this study intended to analyze and assist the gait of healthy elderly individuals with no injuries or diseases that altered their gait, these participants met the study criteria.

For the purpose of this study, the gait data from 10 older adult participants was included. Since the majority of age-related gait changes occur after the age of 60, older adult participants younger than 60 were not included. Table 4 describes the participants whose data was used in this study. For each participant, 5 trials at the comfortable walking speed were included. Table 5 describes the comfortable gait speed used by each participant, averaged over the 5 trials. Only the overground gait trials were included in this study.

Table 4. Participant descriptions

Participant	Age (years)	Mass (kg)	Height (cm)	Gender
29	71	65.35	175.0	Male
31	63	46.05	149.2	Female
32	61	70.55	149.7	Female
33	63	72.25	172.3	Male
34	62	70.5	164.5	Male
36	63	73.1	174.3	Male
37	73	79.65	168.2	Male
39	84	66.35	155.5	Male
40	68	49.2	147.0	Female
42	63	59.85	161.2	Female

Table 5. Average comfortable gait speed over 5 trials

Participant	29	31	32	33	34	36	37	39	40	42
Average Gait Speed (m/s)	1.51	1.11	1.43	1.27	1.02	1.41	1.07	1.01	1.36	0.95
	±	±	±	±	±	±	±	±	±	±
	0.08	0.04	0.05	0.09	0.08	0.04	0.04	0.03	0.07	0.04

3.3 PREPARING MOTION DATA

The files in the dataset were provided in c3d format, which is the typical file format for experimental motion capture. However, OpenSim requires marker trajectories to be specified in marker (.trc) files and ground reaction data be specified in motion (.mot) or storage (.sto) files [43]. To use the data within OpenSim, the files were converted to the compatible file types using the c3dExport.m and the osimC3D.m Matlab codes. These Matlab functions are included with the OpenSim 4.0 download package. They read the c3d files containing the marker and force data, then write the data to .trc files and .mot files, respectively. They are also able to provide

information pertaining to the data, such as rate, number of markers, and number of forces [43]. Upon conversion to .mot file format, some force files contained undefined values (NaNs). During dynamic analysis, force data is splined [43]. Therefore, any NaN occurrences lead to the entire column becoming NaN. To prevent this, NaNs were removed from .mot files.

3.3.1 COORDINATE SYSTEM

The OpenSim coordinate system is characterized by X forward, Y up, and Z right. However, the motion data used for this study was collected using a different coordinate system, characterized by X forward, Y left, and Z up. Therefore, the data required a 90-degree rotation about the x-axis before it could be used in OpenSim. This was accomplished by the `osimC3D.m` Matlab function. To confirm that the data was oriented correctly before beginning the simulation process, the Preview Experimental Data function was used to visualize the marker and force data in the OpenSim GUI.

3.4 SCALING

The Gait 2392 model is a generic model that is included with the download of the OpenSim 4.0 package. To accurately simulate the participant's movement, the model must be anthropometrically altered to match each subject as closely as possible. Scaling allows for a generic model to be modified based on the physical dimensions of the participant, such as mass and body segment dimensions [44]. Scaling is performed through the comparison of experimental marker data to the model markers that are placed on the model. This was accomplished using the OpenSim Scale tool.

The Scale tool contains three tabs: settings, scale factors, and static pose weights. The settings pane specifies parameters related to the subject data and how the model is to be scaled. In the subject data section, the mass of the participant and the marker set file were input. The marker set was adapted from the gait2392_scale_markerset, a marker set that is included with the OpenSim package as a settings file for use with the Gait 2392 model. The scale marker set contains a list of model markers that are to be placed on the model and should include all of the markers that were placed on the participant during experimental trials. Therefore, the following markers were removed from the gait2392_scale_markerset, as they were not included in the Fukuchi et al. dataset: sternum, acromium (left and right), top head, sacral, thigh upper (left and right), thigh front (left and right), thigh rear (left and right), midfoot superior (left and right), and midfoot lateral (left and right). The following markers were then added: left and right Greater Trochanter (GTR), Iliac Crest, and Posterior Superior Iliac Spine (PSIS). In the Scale Model section, the “Preserve mass distribution during scale” box was checked to ensure that the total mass of the scaled model was equivalent to the experimentally measured mass of the subject. The participant’s static trial marker data was also input into this section.

The Scale Factors pane is used to specify the factors by which each segment will be scaled. For this study, segments were scaled based on measurements. Measurement-based scaling involves scaling the model based on the distance between a pair of experimental markers. Table 6 specifies the measurement set used for each segment.

Table 6. Measurement sets for model scaling

Measurements	Marker Pairs			
Pelvis	Right ASIS	Left ASIS		
Thigh	Right ASIS	Right Knee Lateral	Left ASIS	Left Knee Lateral
Shank	Right Knee Lateral	Right Ankle Lateral	Left Knee Lateral	Left Ankle Lateral
Foot	Right Heel	Right Toe Tip	Left Heel	Left Toe Tip

After the model was scaled, the model markers were compared to the experimental markers. When the scale tool is run, the messages window provides information regarding the overall root mean square (RMS) marker error and the maximum error. As recommended in the OpenSim documentation [44], all of the scaled models had RMS errors less than 1 cm and maximum marker errors less than 2 cm on bony landmarks. To reduce error, the model markers were moved to match the experimental marker locations or marker weightings were adjusted. The adjustment of marker weightings affects how closely the model markers match the experimental markers. A heavily weighted marker penalizes errors for that marker, meaning the virtual value will correspond more closely with the experimental value [44]. If model markers needed to be moved, the movement of markers at bony landmarks was avoided, as these markers are relied upon to accurately position and scale the model. The resulting .osim file is an updated musculoskeletal model scaled to the dimensions of the participant, as shown in Figure 12. The model scaling process was repeated for each participant included in this study.

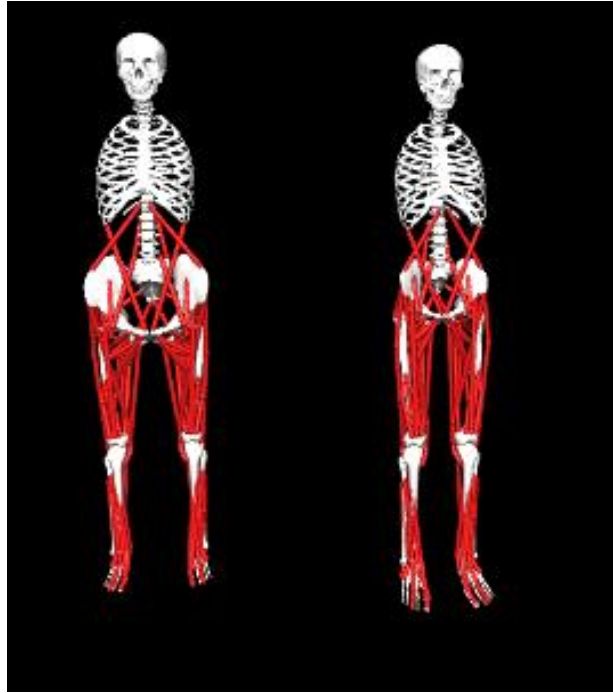


Fig 12. Comparison of scaled Gait 2392 model (left) with unscaled model (right)

3.5 ELDERLY MODEL ADJUSTMENTS

Since the OpenSim scaling tool only adjusts the model's physical dimensions, additional changes were required for Gait 2392 to accurately represent the elderly participants. Compared to young adults, elders exhibit decreased muscle strength, prolonged twitch contractions, increased passive stiffness, and reduced rate of muscle force development. To accurately represent movements of elderly participants, the elderly musculoskeletal models must account for the age-related changes in muscle properties [45]. To best represent older adults, the primary parameter adjustment is to decrease the maximum isometric muscle forces to account for their substantial loss in isometric strength. In this study, the maximum isometric strength was reduced by 30% in each lower-extremity muscle. This value is comparable to experimentally measured loss of isometric strength in the lower limbs [46].

According to Thelen [45], in addition to adjusting maximum isometric force, further parameter modifications are needed to accurately represent age-related changes. Therefore, contraction velocity, deactivation constant, and passive muscle strain were also adjusted on the elderly models. Maximum contraction velocity has been shown to decrease with age, but with less change than maximum isometric strength. Therefore, maximum contraction velocity was decreased by 20% compared to young adults [45]. To account for the increase in passive muscle stiffness in the muscles of older adults, passive muscle strain was reduced from 0.6 to 0.5 in older adults [45]. Finally, the rate of calcium uptake slows with age. As the rate of calcium uptake corresponds with the rate of muscle deactivation, the deactivation time constant was increased from 50 ms to 60 ms in elderly models [45]. These parameter adjustments were validated by Thelen through comparison of simulated and experimental ankle torque behaviour. Thelen incorporated the changes into a musculoskeletal model and determined that with the integrated parameter adjustments, the muscle-actuated model predicted age-related changes that were consistent with experimental observations [45]. In this study, each parameter change was applied to all lower-extremity muscles in the Gait 2392 model.

3.6 INVERSE KINEMATICS

The OpenSim Inverse Kinematics (IK) tool outputs a motion file containing generalized coordinate trajectories. The tool works through each time frame of experimental data and positions the model to best match the data [47]. To run the IK tool, the scaled model, marker trajectory file (.trc), and an IK setup file were input into the inverse kinematics window. When the tool is run, the Messages window reports the maximum marker error and RMS error. As

recommended in the OpenSim documentation, the maximum marker error was less than 4 cm, and the RMS error was less than 2 cm in all trials. If the error was larger, the marker weightings were adjusted, and IK was rerun until a lower error was achieved. To adjust the marker weighting, “motion” segment markers were weighted more heavily than anatomical markers. This is because anatomical markers can be affected by soft tissue during movement. The final kinematics results were saved as motion (.mot) files. To evaluate the results, the simulated kinematics were compared to the experimental kinematics gathered by Fukuchi et al [42], as well as experimentally gathered kinematics in the literature [10] [12]. Figure 13 illustrates the kinematics resulting from the OpenSim IK tool.

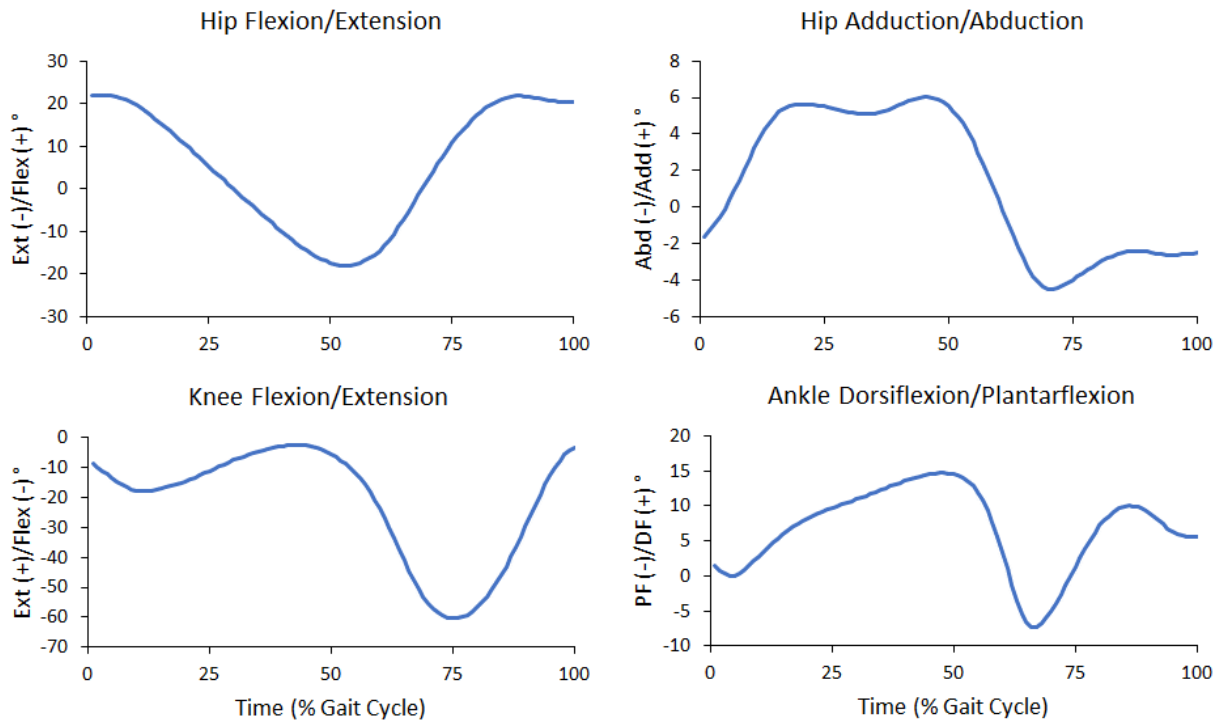


Fig 13. Average kinematics across all elderly models during one gait cycle

3.7 RESIDUAL REDUCTION ALGORITHM

Residual Reduction Algorithm (RRA) is a tool that minimizes the effects of modelling and marker data processing error that can lead to large compensatory forces called residuals [48]. The RRA tool alters the pelvis center of mass of the model and results in a new set of kinematics that are more dynamically consistent with the ground reaction force data. Prior to beginning RRA, the mtp and subtalar joints were locked in the model (osim) files. RRA first requires the input of a kinematics file, which was filtered at 6 Hz using the “filter kinematics” box. The “adjust model” box was selected, and the pelvis was selected as the body COM to adjust. This resulted in an updated model with an altered pelvis COM upon completion of RRA. An RRA Task file and RRA Actuator file were added to the RRA tool. These files were adapted from the example files provided on the RRA setup pages on the OpenSim website [48].

Prior to adding the ground reaction forces to the model, the data was filtered using a 2nd order Butterworth dual low-pass filter, with a cut-off frequency of 8 Hz. For each trial, the ground reaction data was previewed with the corresponding musculoskeletal model to determine which force plates were stepped on by the participant. An External Loads file was then created using the RRA tool. This involved uploading the ground reaction force (.mot) file and assigning the force plates to the corresponding foot used by the participant during the trial. The force measured by each force plate was then applied to either the left or right calcaneus, depending on the foot used by the participant to step on the plate. Once RRA was run, the resulting residual forces and moments, as well as the errors were analyzed. A table provided on the OpenSim RRA documentation page provided guidance on acceptable residual values [48]. If the resulting residuals were higher than the recommended values (Maximum residual force of 10-25 N and

maximum residual moment of 50-75 Nm), the process was repeated while relaxing tracking weights on coordinates with low error. Once residual values were deemed acceptable, the new model containing the updated pelvis mass and COM was saved. The average maximum and RMS residual values across the participants are shown in Table 7.

Table 7. Average maximum and RMS residual forces and moments

	F_x (N)	F_y (N)	F_z (N)	M_x (Nm)	M_y (Nm)	M_z (Nm)
Max	13.2 ± 7.0	17.8 ± 6.9	19.6 ± 6.8	54.7 ± 16.9	30.4 ± 18.8	57.1 ± 13.0
RMS	2.5 ± 2.9	1.8 ± 3.6	1.0 ± 6.9	1.8 ± 10.1	1.1 ± 8.2	1.0 ± 5.2

3.8 COMPUTED MUSCLE CONTROL

Computed Muscle Control (CMC) is a tool that computes the muscle excitations that drive the musculoskeletal model towards the desired kinematic trajectory [49]. This is accomplished using a combination of proportional-derivative control (PD) and static optimization. The CMC algorithm first computes a set of desired accelerations, which drive the model coordinates toward the experimental coordinates. CMC then computes the actuator controls that will result in the desired accelerations, followed by the use of static optimization to distribute the load across synergistic actuators. Finally, the CMC algorithm uses computed controls to conduct a forward dynamics simulation, advancing the simulation forward in time. CMC repeats these steps until the end of the desired movement interval [49].

Similar to the Scaling tool, the CMC tool is controlled by a window with three tabs: Main Settings, Actuators and External Loads, and Integrator Settings. The Main Settings tab specifies

parameters regarding the input kinematics, the time range for CMC analysis, and the output. The RRA kinematics for the current trial were first added. A CMC Tracking Task File was also uploaded in this section. The CMC task file was adapted from the example file provided with the Gait 2392 documentation. This file specifies which coordinates should follow the desired kinematic trajectory. The CMC look-ahead window was set to 0.005. This was reduced from the default value of 0.01 to reduce the resonance that was present in some CMC results.

The Actuators and External Loads tab specifies any additional actuators that are involved with the movement or external loads, such as ground reaction forces, that are applied to the model. The Additional Force Set file used was adapted from the example CMC Actuators file included with the Gait 2392 documentation. However, the pelvis center of mass (COM) was adjusted, specific to each scaled model. The actuator file was set to append rather than replace the model's force set. The ground reaction force data was added via the External Loads section in this tab.

After the completion of CMC, the results were evaluated based on suggestions provided in the OpenSim CMC documentation. Peak residual forces were less than 25 N, peak residual moments were less than 75 Nm, and peak reserve torques were less than 50 Nm in all trials. CMC was run for each participant, for all trials included in this study.

The trials used for the simulations contained ground reaction force data from only two consecutive force plates. As the CMC tool requires ground reaction force data for the entirety of the simulation time period, results were not provided for the initial double support phase of the gait cycle. To correct this, symmetry was assumed and data from the second double support

phase was mirrored to represent the initial phase of double support. This was applied across all subjects and trials to ensure a full gait cycle was analyzed.

3.9 SIMULATING ASSISTIVE DEVICES

This study aimed to investigate the effect of a hip assistive device compared to an ankle assistive device on elderly walking. The addition of assistive devices to the elderly models was based on the process by Uchida et al. in a similar study [50]. Ideal flexion/extension devices were added bilaterally to the ankle and hip separately in the elderly models. Each device was ideal in that it was modelled as a massless, lossless actuator that applied torque directly to the joint it was assisting. The magnitude and rate of assistive torque were both set to be unlimited, allowing the device to apply as much torque to the joint as required. The devices were added to the ankle and hip of the elderly models by increasing the strength of the corresponding reserve actuators. Normally, the reserve actuators offset any muscle weakness in the model by applying additional joint torques. The reserve actuators activate when an actuator is unable to generate the necessary force to produce the desired movement. This ensures that the simulation is able to run. However, reserve forces are normally minimized to prevent the model from recruiting these actuators if they are not required. The following instantaneous objective function is used by CMC to balance the recruitment of reserve actuators with the recruitment of muscles [50]:

$$J(a, \tau) = \sum_{i=1}^{nMuscles} a_i^2 + \sum_{j=1}^{nReserves} \left(\frac{\tau_j}{w_j} \right)^2 \quad (1)$$

In the above equation, n_{Muscles} and n_{Reserves} refer to the number of musculotendon and reserve actuators in the model, a_i represents the instantaneous activation of the i th muscle, τ_j is the instantaneous torque applied to the j th reserve actuator, and w_j is the constant weighting factor (optimal force) that scales the penalty of recruiting the j th reserve actuator [50]. During the unassisted CMC simulations, the weighting factors were set to 1 Nm, which penalizes the solution by the square of the torque generated by the reserve actuator, as shown in Eq (1). Therefore, the peak reserve actuator torques were very small. To simulate the ideal assistive devices, the weighting factor was increased to 1 MNm, which results in a negligible penalty when the corresponding reserve actuator is recruited. Therefore, the CMC results indicate the forces applied by the muscles, with the support of an assistive device. CMC was first run with a weighting factor of 1 MNm exclusively at the ankle joints to simulate a bilateral ideal assistive device at the ankles. Following this trial, an additional CMC trial was then run with an increased weighting factor at the hip joints to simulate a bilateral ideal assistive device at the hips. The original ground reaction forces and kinematics used during the unassisted simulations were applied during the assisted simulations, meaning that the net joint moments were unchanged. For each of the 10 elderly participants, an unassisted, ankle assisted, and hip assisted simulation were completed. This was performed for 5 trials per participant, for a total of 150 simulations.

3.10 METABOLIC CONSUMPTION

The energy consumed by each muscle was computed during the CMC simulation using a muscle energetics model. The model used was an updated version of the Umberger 2010 Metabolic Probe [51], implemented by Uchida et al. [52]. The Metabolic Probe was applied to

the elderly musculoskeletal model, adding a probe to each muscle within the model. The probes then calculated the instantaneous metabolic power consumed by each of the lower extremity muscles during the simulation. This process was repeated for each elderly participant and assistance scenario. Average metabolic power consumption for each muscle was then computed by integrating the instantaneous consumption over the gait cycle and dividing by the duration of the gait cycle. The resulting average metabolic consumption was then summed across all lower extremity muscles and divided by the participant mass, resulting in the total average metabolic power consumption. Performance of each ideal assistive device was evaluated by comparing the average metabolic power consumption in the assisted scenarios to the unassisted scenario.

WAE devices often apply assistance unidirectionally, such as applying plantarflexion assistance without dorsiflexion assistance. Therefore, metabolic power savings attributable to exclusively flexion and extension assistance were calculated for both the ankle and hip ideal assistive devices. Each simulation was segmented during which the ideal actuators were generating only flexion or only extension torque. The reduction in metabolic power during the simulation was then summed across the segments. Therefore, the total reduction in metabolic power during the period which generated flexion-only assistance was summed, then the process was repeated for periods of extension-only assistance for both the hip and ankle actuators.

CHAPTER 4: RESULTS AND DISCUSSION

4.1 RESULTS

Average metabolic power consumption was shown to decrease in both assistance scenarios compared to the unassisted scenario, as shown in Figure 14. When comparing the ankle and hip assistance conditions, the hip ideal actuator resulted in a greater reduction in average metabolic power compared to the ankle ideal actuator. The hip ideal actuator resulted in a metabolic power reduction of 25.5%, whereas the ankle ideal actuator reduced metabolic power by 14.2% compared to unassisted. The reduction in average metabolic consumption was significant in both assistance scenarios compared to the unassisted scenario ($p < 0.002$, matched pairs t-test). However, hip assistance saved significantly more metabolic power compared to ankle assistance ($p < 0.002$, matched pairs t-test).

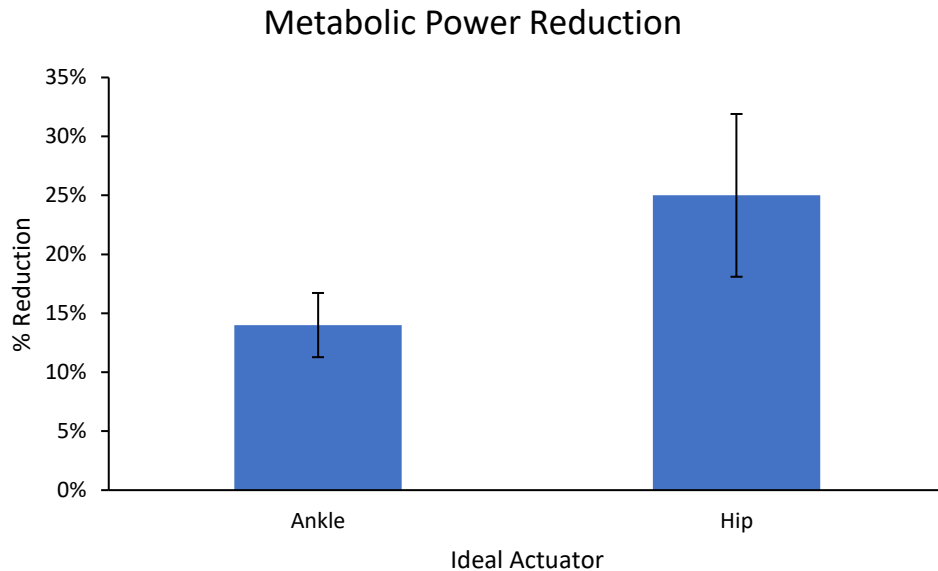


Fig 14. Reduction in average metabolic power consumed by lower-extremity muscles when assisting walking with ideal flexion/extension actuators

The reduction in average metabolic power was analyzed separately for each participant to determine if a consistent trend was observed across all participants. As shown in Figure 15, when the average reduction in metabolic power was plotted for each participant, all participants demonstrated a greater reduction in metabolic power with the assistance of an ideal hip actuator compared to an ankle actuator. However, the difference in metabolic power reduction between the ankle and hip assistance conditions varied between participants.

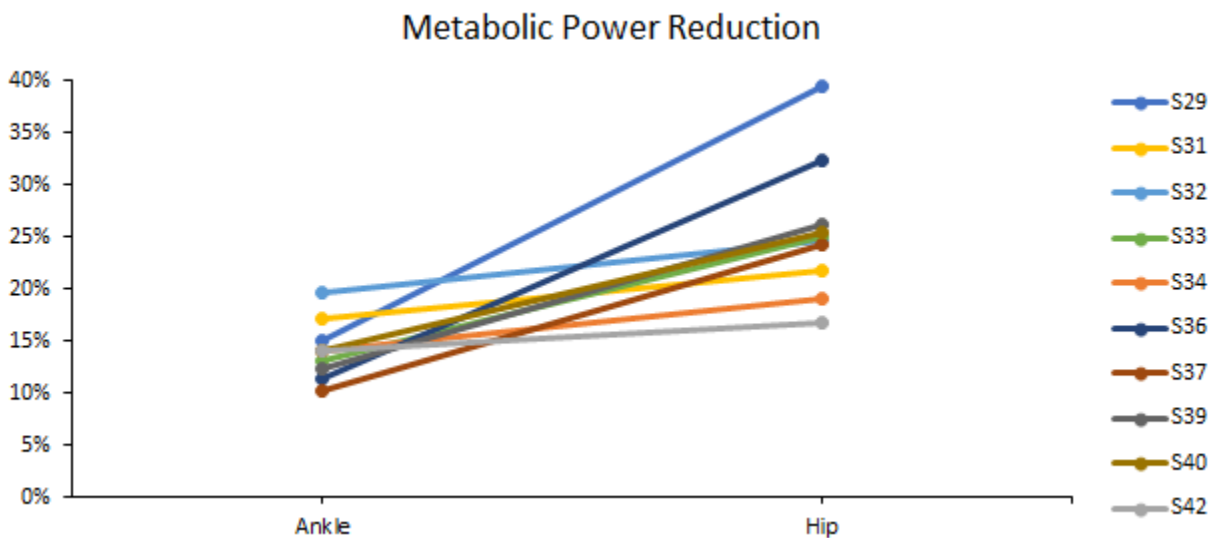


Fig 15. Reduction in average metabolic power for each participant during ankle and hip assisted walking conditions

As shown in Figure 16, if unidirectional assistance was provided (i.e. only extension or only flexion torque), flexion torque resulted in significantly more metabolic power savings than extension torque at the hip ($p < 0.002$, matched pairs t-test), with 70% of the metabolic savings attributed to flexion assistance and 30% attributed to extension. Conversely, plantarflexion torque resulted in significantly more metabolic savings than dorsiflexion torque at the ankle, with 64% of savings attributed to plantarflexion and 36% attributed to dorsiflexion.

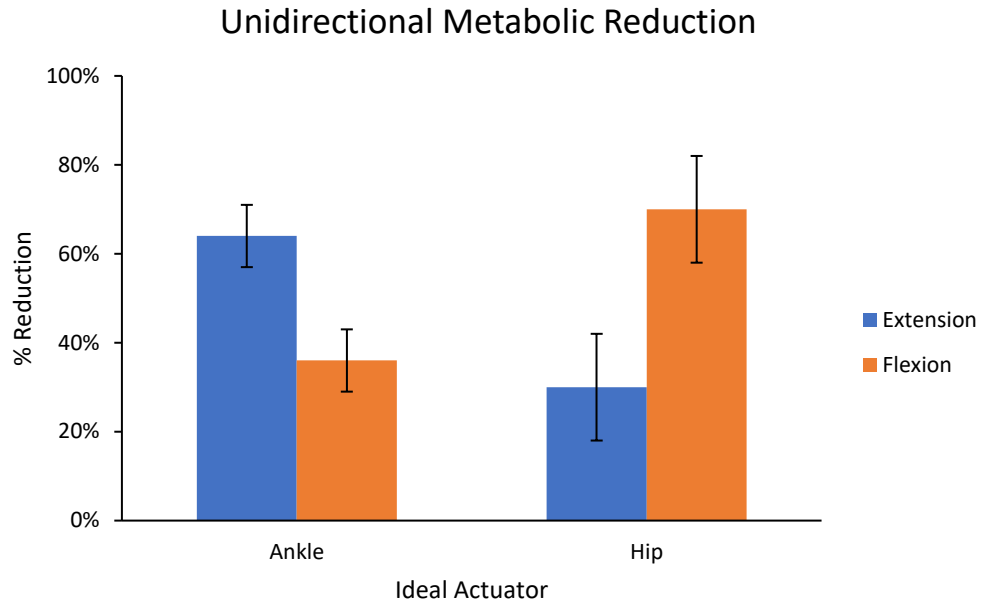


Fig 16. % metabolic reduction at the ankle and hip attributable to exclusively flexion and extension assistance

Figure 17 illustrates the changes in muscle activation that occur in 7 representative muscles during the simulated assistance scenarios. When assisting the ankle, substantial decreases in activation occurred in the soleus and tibialis anterior muscles. When assisting the hip, substantial decreases in activation occurred in the iliacus and psoas muscles. Slight decreases in activation were observed in the gluteus maximus muscle, however there is an increase in activation of gluteus maximus 3 (the most interior portion of the muscle). The activation of the rectus femoris muscle increased during hip assistance.

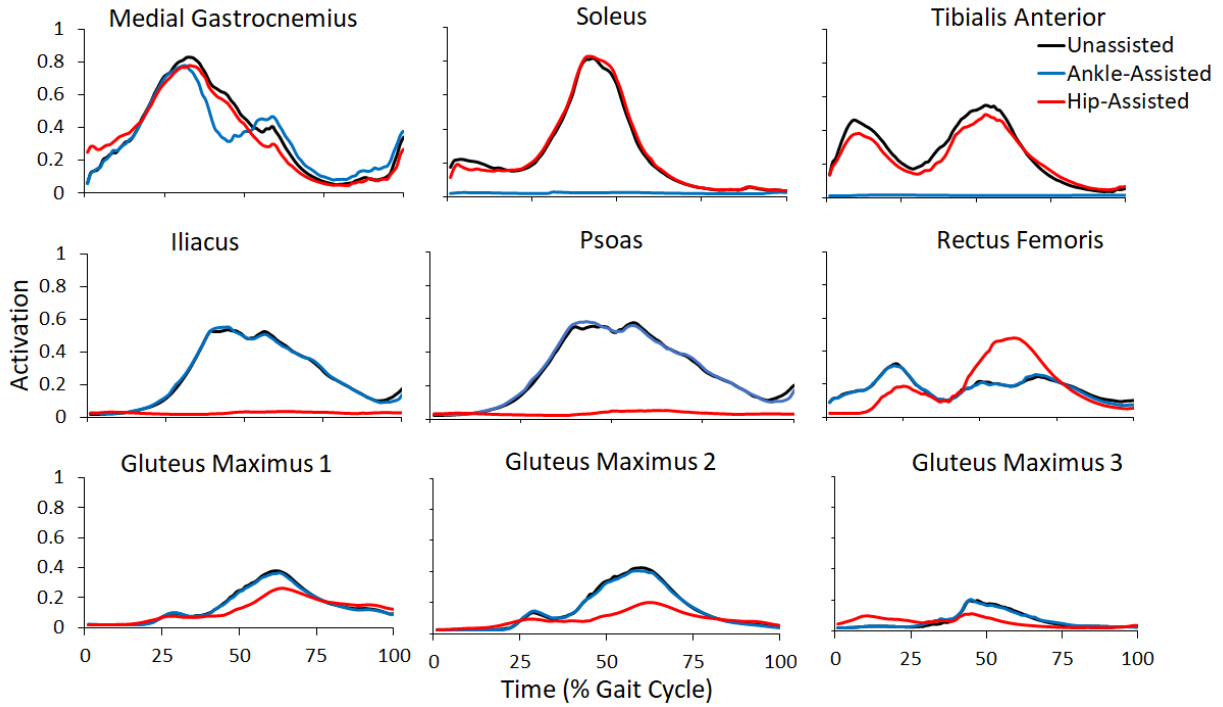


Fig 17. Muscle activations during 3 assistance scenarios: unassisted (black), ankle-actuated (blue), and hip-actuated (red).

The actuator moment is presented to analyze the mechanical feasibility of designing a physical exoskeleton device with similar values. Table 8 summarizes the actuator moment during the ankle and hip assistance scenarios. The peak positive moment of the ankle actuator was 25.0 ± 5.8 Nm or 0.39 ± 0.09 Nm/kg, while the peak negative moment was -61.8 ± 12.4 Nm or -1.00 ± 0.2 Nm/kg. The peak positive moment of the hip actuator was 47.4 ± 12.4 Nm or 0.75 ± 0.2 Nm/kg, and the peak negative moment was -58.6 ± 29.6 Nm or -0.90 ± 0.4 Nm/kg.

Table 8. Summary of average and peak device moments during ankle and hip assistance conditions, across all participants

	Actuator Moment (Nm)				Actuator Moment (Nm/kg)			
	Average positive	Average negative	Peak positive	Peak negative	Average positive	Average negative	Peak positive	Peak negative
Ankle Actuator	10.6 ± 1.9	-24.5 ± 6.4	25.0 ± 5.8	-61.8 ± 12.4	0.16 ± 0.02	-0.39 ± 0.06	0.39 ± 0.09	-1.00 ± 0.2
Hip Actuator	19.4 ± 3.6	-25.0 ± 11.6	47.4 ± 12.4	-58.6 ± 29.6	0.30 ± 0.04	-0.39 ± 0.2	0.75 ± 0.2	-0.90 ± 0.4

4.2 DISCUSSION

The primary finding in this study was that simulated hip assistance of elderly gait generated a larger reduction in metabolic power compared to ankle assistance. However, this finding differs from current WAE studies evaluating devices on elderly participants. Seo et al. recently designed a powered hip exoskeleton to assist hip flexion and extension in elderly gait. Initially, it was tested on young participants where it demonstrated a 13% reduction in metabolic cost [53]. Testing was later conducted on healthy, elderly participants who demonstrated a metabolic cost reduction of 7% [54]. Conversely, Galle et al. demonstrated that a powered ankle exoskeleton could reduce the metabolic cost of elderly participants by 12% by assisting plantarflexion.

However, while the focus on WAE development and use in the healthy elderly population has increased over the recent years, there are few studies which have tested and evaluated exoskeletons on healthy elderly participants. Conversely, there have been numerous studies conducted on healthy, young participants which have resulted in substantial metabolic savings,

particularly through hip assistance. Ding et al. demonstrated a 17.4% reduction in metabolic cost when users walked with the assistance of a hip exoskeleton, whereas Lim et al. reported a 19.4% reduction in metabolic cost with the assistance of a hip exoskeleton [55] [56]. In comparison, Sawicki et al. reported a 14% reduction in metabolic cost with the assistance of an ankle exoskeleton. Although these designs have not been tested on elderly participants, they offer promising insight into the design of exoskeletons which may also reduce the metabolic cost of elderly users. However, future experimental testing on elderly participants is needed to verify the effect of recent exoskeleton designs on elderly users.

As of December 2019, Sawicki et al. compiled peer-reviewed publications that reported improvements in user walking or running economy using exoskeletons [57]. This comparison concluded that hip exoskeletons have improved metabolic cost more than ankle exoskeletons, with a 12% greater reduction using a tethered device and an 11% greater reduction using an autonomous device. This is thought to be a result of physiological differences between hip and ankle morphology, such as differences in muscle-tendon architecture. Proximal tendons in the legs tend to have long muscle fibers, small pennation angles, and low fixed-end compliance, whereas distal tendons tend to have shorter muscle fibers with high pennation angles and long elastic tendons. Therefore, distal muscles are able to produce isometric force as a result of their long series tendons, which consumes a lower amount of metabolic energy. Conversely, proximal muscles are more likely to produce a high muscular work output, along with high metabolic energy consumption [58]. Alternatively, the difference exhibited between hip and ankle assistance may be due to the effect that the device mass has on different locations on the body. When mass is added to the extremities, greater metabolic rate has been observed when the load

is added in distal locations compared to proximal locations [59]. The above studies, whose findings reveal that hip assistance results in a greater reduction in metabolic power compared to ankle assistance, support our findings that hip assistance is more beneficial than ankle assistance for reducing metabolic cost. The finding that hip assistance contributes to greater metabolic power reduction during gait can likely be attributed to the ability for the hip musculature to compensate for decreased ankle power, since the hip power generation requirements are generally less than that of the ankle relative to maximal capacity [18].

When the average metabolic power reduction was analyzed for each participant, all participants demonstrated a greater reduction in metabolic power in the ideal hip actuator condition compared to the ankle actuator. However, the difference in metabolic power between the ankle and hip assistance conditions varied between participants. This is potentially due to differences in comfortable walking speed. Subject 29 completed their comfortable walking trials at 1.51 ± 0.08 m/s, which was the fastest gait speed among the participants, and exhibited the largest reduction in metabolic power in the hip assistance condition. Conversely, Subject 42 completed their walking trials at 0.95 ± 0.04 m/s, which was the slowest gait speed among the participants, and exhibited the smallest reduction in metabolic power in the hip assistance condition. This is supported by the findings in Uchida et al. where use of an ideal hip actuator resulted in greater metabolic savings at increased speed [50].

As reported above, a powered hip exoskeleton in the literature demonstrated a 19.4% reduction in metabolic cost during experimental testing, which is less than the 25.5% reduction resulting from our simulations [56]. However, as our simulations used ideal assistive devices that

were massless and able to provide large amounts of torque to the joint, it is likely that experimental trials would result in smaller metabolic cost reductions. These considerations may explain why the metabolic reduction observed from the simulations exceeded experimental results. However, we predict that while reduction in metabolic cost may decrease in experimental conditions, this would occur in both the hip and ankle joint and the observed trend of hip assistance resulting in greater metabolic savings compared to ankle assistance would remain unchanged. However, this prediction should be tested in experimental conditions.

Our results suggest that the most beneficial strategy for assisting elderly users is hip flexion and extension assistance. However, between flexion and extension assistance, we observed significantly more metabolic savings due to the contribution of flexion assistance. This finding is supported by the results in Dembia et al. where hip flexion assistance demonstrated a greater proportion of metabolic savings compared to hip extension assistance [60]. Our finding that ankle plantarflexion assistance resulted in significantly more metabolic savings compared to dorsiflexion assistance was also supported by Dembia et al [60]. Ankle plantar flexor muscles consume more metabolic energy than ankle dorsiflexor muscles. Therefore, there is more potential for the reduction of metabolic cost in ankle plantar flexor muscles compared to ankle dorsiflexors [61].

The analysis of muscle activations during the hip and ankle assistance scenarios is beneficial as muscle activation is a major determinant of metabolic energy [62]. Therefore, reductions in muscle activation likely contribute to the reductions in metabolic power observed during hip and ankle assistance conditions. When assisting the ankle, substantial decreases in

activation occurred in the soleus and tibialis anterior muscles, which are muscles that only generate an ankle moment. The ideal ankle actuator was able to generate these moments at negligible cost. The medial gastrocnemius decreased only partially, likely because although it was no longer recruited to generate a plantarflexion moment, it is still recruited to generate a knee flexion moment.

When assisting the hip, substantial decreases in activation occurred in the iliacus and psoas muscles. Slight decreases in activation were observed in the gluteus maximus muscle, however in gluteus maximus 3 (the most interior portion of the muscle), an increase in activation was observed, likely due to recruitment in an alternate hip degree of freedom (i.e. abduction/adduction or internal/external rotation). The activation of the rectus femoris muscle increased during hip assistance, likely to take advantage of its high force-generating capability [50].

The hip and ankle actuator moments were analyzed to determine if it is mechanically feasible to reproduce the ideal actuator torque with a physical device. When compared to literature values, the ideal actuator moments at both the hip and ankle joint were considered feasible values. Ferris et al. developed a pneumatically powered orthosis for the ankle joint which provided a peak plantarflexion torque of 70 Nm and a peak dorsiflexion torque of 38 Nm [63]. Kang et al. developed a powered bilateral hip exoskeleton which was capable of providing a peak torque of 60 Nm [64].

4.3 STUDY LIMITATIONS

One potential limitation in this study is that the kinematic data used to run the simulations was obtained from a public database. As we are unaware of any potential errors that may have occurred during the data acquisition process, such as marker placement errors, we are unable to take that into account during the scaling process in OpenSim. In addition, limited information was provided regarding the participant's muscular characteristics, such as their maximum isometric strength. Therefore, this study does not consider individual variations in muscle strength or characteristics between participants. To reflect muscular characteristics of the participants more accurately, this would need to be evaluated prior to conducting simulations. As this information was not available, the parameters suggested by Thelen were applied universally across all models during the simulations [45]. However, the parameter adjustments were only validated by Thelen pertaining to the ankle joint and have not been validated for the remainder of the lower extremities through comparison to experimental results. In addition, no sensitivity testing of the applied elderly model changes was conducted. Sensitivity testing should be conducted in future work to verify the results of this study.

Another limitation is the assumption that the kinematics and ground reaction forces would remain unchanged between the unassisted and assisted conditions. In reality, the user would likely experience changes to their kinematics with the addition of an assistive device. The extent to which they are altered varies throughout the literature, with some studies reporting relatively small changes [30] [65] [66], and others reporting large changes [67] [68]. However, kinematic changes observed experimentally likely result from both the mass of the device and

the assistive forces applied to the body [59]. Therefore, changes in kinematics do not necessarily result in substantial metabolic cost changes [69].

Another limitation is that the assistive devices were modelled as ideal torque actuators, meaning they were massless and could produce a large amount of torque precisely and instantaneously. In reality, addition of an assistive device to the body would be an additional mass with which the participant is not accustomed to moving. Therefore, the device would likely result in increased energy expenditure, decreasing the metabolic savings compared to simulation results. These considerations likely contribute to the metabolic reduction values observed in this study exceeding results observed in the literature. In addition, the weight of the assistive device has been shown to have varying effects depending on where it is located. For example, assistive devices of the same weight have demonstrated increased metabolic cost to the body when located at the foot compared to the hip [59]. Inertia was also neglected during the simulations, which would affect the gait of the user in experimental conditions. However, by excluding practical features such as device mass and inertia from this analysis, the advantageous effects of adding an assistive device were not confounded with the negative side effects that these practical features often result in within experimental conditions.

Finally, the method used to calculate unidirectional metabolic power did not consider the physiological delay that is present in the simulation from the time at which the assistance is applied to the joint, to the time at which it impacts the metabolic power. Although this physiological delay is small, it may impact the results and should be taken into consideration in future work.

4.4 FUTURE WORK

Since our simulations analyzed the effect of ideal assistive devices on the gait of elderly participants walking at comfortable speeds, in the future, we will analyze the use of the same devices while participants walk at both fast and slow speeds. This will help determine whether a different assist strategy is optimal at different walking speeds. In addition, as this study found the assistance of hip flexion and extension to be beneficial for elderly users, future work could include the analysis of assistance at additional hip DOF, such as adduction, abduction and rotation to determine how varying hip assistance strategies impact the metabolic cost of elderly users. In this study, only assistance at the hip and ankle joints were considered, as these are the locations associated with the differing gait strategies between young and elderly adults [12]. However, future work could include an analysis of knee assistance, as well as varying combinations of multi-joint assistance to investigate the effects on elderly gait.

To obtain more realistic results, simulations could be performed that model non-ideal actuators, including aspects that would affect the metabolic cost of the user such as device mass and torque limit. Additionally, in order to achieve more robust simulation results, the data could be analyzed using the Rajagopal 2016 musculoskeletal model [41]. This is an updated version of the Gait 2392 model and may improve the accuracy of results. The simulations could also be conducted using data from a larger sample size to strengthen conclusions. Sensitivity analysis should be conducted in future work to investigate the robustness of these results.

Our results predict how hypothetical assistive devices may perform and can help future researchers understand the effect of ankle and hip assistance on elderly gait. Simulations are

helpful to supplement experimental studies; however, experimental studies are needed to validate these simulation results, as well as study the practical challenges, such as device mass, that were ignored in this study.

4.5 CONCLUSION

WAE offer a solution for restoring, enhancing, and maintaining mobility in the elderly. In this study, simulations were used to analyze the effect of assistive devices on healthy elderly gait. Bilateral ankle and hip ideal actuators were modelled to gain insight into the effect that assistance at either joint would have on the metabolic cost of the participant. Based on our findings, the use of actuators at the hip joints provided a greater reduction in metabolic power compared to the use of actuators at the ankle joints to assist elderly gait. The results of this study can be used to influence the future development of exoskeletons targeted towards the elderly population. Therefore, between the two assistance strategies, it may be beneficial for researchers to focus on the design, development, and testing of hip flexion and extension devices when designing WAE for the elderly.

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