Designing an Actuated Walker for Improving User Stability

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Abstract

Figure 1: Four full size actuated walker iterations, Mark I to IV from left to right

Falling is the number one cause of death for people over 65. Despite falls often indicating the start of a functional decline, leading to painful surgeries, loss of independence, and a toll on mental health, they are often preventable accidents. This thesis proposes a new type of walker, equipped with features such as perception and actuation, and implementing a pose correction algorithm aimed at improving stability and usability. The goal is to use real-time gait and pose detection of the user to actively guide the user to a safer and more stable walk.

After multiple hardware iterations, we identified an effective combination of sensors and actuators to make the goal possible, along with novel methods of implementing user pose tracking. The final integrated prototype was able to keep walker users more stable with its correction algorithm compared to when it was turned off.

Acknowledgements

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Contents

Chapter 1

Walkers and Rollators

Falls are the leading cause of both fatal and nonfatal injuries for senior citizens over the age of 65, and as the the elderly population grows in size, the number of falls will continue to soar [1]. The likelihood that any adult over the age of 65 experiences at least one fall in a given year is 27% and one out of every five falls cause a serious injury [2, 3]. Common injuries caused by falling are fractures, bruises, and abrasions, mostly occurring at the head and neck [2]. Of the injured, one in three are hospitalized for their injuries [2].

The effects of falls extend beyond the initial injury, however. Senior citizens that experienced a fall in the previous year are 2.55 times more likely to fall again. This is partially due to falling being a marker for reduced physical ability, but also due to injuries suffered during the initial fall that increase future fall likelihood [2]. In addition, falls among older adults generally lead to subsequent hospital re-admissions, with each hospitalization cost averaging $35,000 [4]. The most common nonfatal injury from falls is bone fracture, which accounts for 61% of the cost of such injuries, or $18.8 billion [4]. Many of those injured by falls will stay in a nursing home for at least a year, with Medicaid usually covering all the costs [4].

Our research shows that an actuated smart walker has the potential to reduce falls of senior citizens, increasing their quality of life and longevity. We have investigated the danger of falls for senior citizens and the medical field's view on the dangers of conventional walkers. Our research shows that many of the falls caused by walkers could benefit from a walker that could stop or brake itself when the user is in distress, unlike 'static' conventional walkers. The below subsections expand upon the above claims in detail, including background information on the field of assistive devices.

1.1 The Demographics of Walker Users

Walker usage has increased steadily over time. In 2000, 4.55% of people over 65 used walkers; for people over 85, the percentage was 16.53% [5]. The use of walkers as a walking aid was second only to canes, with 1.82 million people in the US using walkers [5].

However, in 2011, 4.09 million people in the US used walkers [6]. Compared to 2004, there had been a 50% increase in the number of people using mobility devices in general; 24% of all people above the age of 65 used a mobility aid of some kind [6].

In the time period between 2000 and 2011, there had been a 125% increase in the number of people using walkers [5] [6]. Following this trend, with the growing size of the elderly population, and the increasing rates of diabetes and obesity, it is likely that the percentage of walker users will continue to rise. Although no large-scale survey has been performed on recent walker usage, assuming another 125% increase of walker users from 2011 to 2020, we can estimate 9.20 million walker users in the United States as of 2020.

1.2 The Benefits and Limitations of Current Walkers and Rollators

Walkers are devices that allows senior citizens to function and live more independently, and are designed to reduce risks of falls by improving the user's gait mechanics [7]. Depending on the conditions of the user, they will either use a walker with four legs, or a rollator, which is a type of walker, with two or four wheels. They reduce the weight borne by the legs to alleviate pain from weakened or impaired leg motor control or injury [8]. Also, they increase confidence and feelings of safety, which increases user's independence and comfort with activity [8]. They also enable users to be more mobile, preventing osteoporosis and cardiorespiratory deconditioning while enhancing circulation [8].

Although a rollator allows patients to walk faster and easier, not everyone can use a walker or a rollator. First, it cannot be used on patients with balance issues or cognitive impairment, as it can roll forward unexpectedly, resulting in a fall [9]. Such patients often need to use two-wheeled or no-wheeled walkers, which significantly change their gait and slow down movement [9]. In addition, individuals with a motor disability (imbalance or coordination problems) caused by tumors or strokes cannot use walkers [10]. Researchers have suggested that these limitations are related to the walker providing unstable and fragile support, and its inability to stabilize the user by adapting to their needs [10].

In addition, using any kind of walker is considerably attention demanding [11]. Even users highly trained and used to using these aids exhibited similar difficulties paying attention to other tasks as first time users [11]. Also, around 80% of walker users reported difficulty in moving uphill and downhill [12].

People using walkers are twice as likely to experience a fall than those who do not use walkers, resulting in users generally having a greater fear of falling [13]. Furthermore, walkers were associated with seven times more injuries than canes [14]. Although some studies have demonstrated that walkers are effective at improving balance and mobility, others suggest that they may increase fall risk by causing tripping or interfering with a person's balance control [14]. Although use of assistive devices has a statistical association with a higher risk of falls, walkers do not necessarily cause falls. Instead, the usage of assistive devices may be a marker for other problems and underlying causes; for example, walkers may simply be an indicator for balance impairment and functional decline [2, 8, 13]. Ultimately, walker users are a weak and vulnerable population, and a better walker is needed to help these people.

Chapter 2

Mechanical Design and Hardware Features

Our goal with this project was to build a better walker equipped with the latest available sensors and technologies with the goal of making them safer to use. As actuated walkers are a type of mobility aid that currently do not exist, it was necessary to design multiple novel mechanisms as well as the chassis to be able to integrate the multiple sensors and actuators needed. The actuated walker project necessitated the selection of features to implement, the types of parts needed to realize such features, and an iterative design process to continuously develop the hardware so that it is capable of performing its ultimate goal of reducing falls.

This chapter will first give a general overview of the different hardware iterations that has been built over the progress of this project. Then, it will introduce the variety of capabilities and features introduced and improved upon in the different iterations. After an overview of multiple different manufacturing methodologies used to build the prototypes, it will conclude with a direction for a future iteration of the walker.

2.1 Hardware Iteration Timeline

In this section, the five prototypes built for the actuated walker project is introduced.

2.1.1 Small Scale Prototype - Mark 0

The small scale prototype was built to verify the viability of using brushless DC motors (BLDC) motors as a variable braking system. In addition, it had one load cell per side to serve as force sensing sensors.

Figure 2.1: First small scale model

2.1.2 First Full Size Prototype - Mark I

The first full size prototype introduced a two-directional load cell handle assembly with the goal of measuring both forward and downward force. In addition, it integrated a timing belt driven back wheel.

Figure 2.2: Full Scale Mark I

2.1.3 Second Iteration - Mark II

The second iteration was designed to integrate the walker's robotic components better. In addition, it integrated a RealSense D435i RGB-depth camera in order to support better user detection.

Figure 2.3: Full Scale Mark II

2.1.4 Third Iteration - Mark III

The third iteration introduced a disk-brake based inline brake system, giving users access to a physical brake for the first time, along with automated control of the brake. It also updated the load cells used to be more compact and reliable.

Figure 2.5: Full Scale Mark III

2.1.5 Fourth Iteration - Mark IV

Mark IV introduced a new vertically-folding design, which gives the RealSense camera an unobstructed view of the user's lower body. In addition, it experimented with all wheel control by using a long belt that connects both the front and back wheels. It was also the first model to be constructed out of waterjet cut sheets of aluminium, significantly speeding up production.

Figure 2.6: Full Scale Mark IV

2.2 Hardware Features

The development of these prototypes were driven by the need to test, implement and assess multiple different features. In this section, the evolution of those features are presented in detail.

2.2.1 Handle Force Detection

The first implementation of load cell handles just measured how much force a user was pushing the walker with, which is why in Figure 2.7 there is just one load cell set up to measure the horizontal force.

Upon experimentation, it became apparent that being able to determine the downward force the user places on the handles as a 2-dimensional vector would be helpful for user state analysis. For example, a downward force that is too excessive could be signaling a user falling down, or a detection of no force could signal a user losing grip. In addition, the vertical load cell can be used in this case as a method of driving the walker while sitting down. The implementation of a two-loadcell-per-handle setup was in Mark I, and can be seen in Figure 2.8.

Figure 2.7: Mark 0 Horizontal Load Cells

Figure 2.8: Mark I Load Cell Design

However, the problem with this design was that the downward force, with the user's handle being attached to the upper load cell, exerts a non-trivial moment on the vertical load cell. Although there seemed to be a linear correlation of the exerted downward force to force readings in the vertical load cell, using software to account for the moment led to inaccurate readings in real life, likely due to the fact that the downward force is subject to constant vibrations.

This led to the design of a moment-constrained load cell handle design, first implemented in Mark II and further improved in Mark III.

From Figures 2.9, 2.10 and 2.11, the addition of a moment constraining bar can be seen. It is placed in such a way that the downward force is exerted fully and only to the top load cell's strain gauges, with the moment that would have been exerted on the vertical load cell constrained by the horizontal bar that physically prohibits the downward movement of the top load cell. The moment constraining bar is also not fixated to the top load cell, allowing it to function as a roller to transfer most of the force user pushes the walker with to the vertical load cell, with some potential error due to friction.

This design was able to isolate vertical and horizontal forces, even in cases where the force exerted was a combination of vertical and horizontal forces. In addition, this design integrated a user handle as well as a manual brake. One caveat is that this does require multiple custom-machined parts with a high tolerance requirement, as well as adding eight fragile cables that need to be routed through the chassis, but this design is capable of functionally achieving all necessary requirements it needs to.

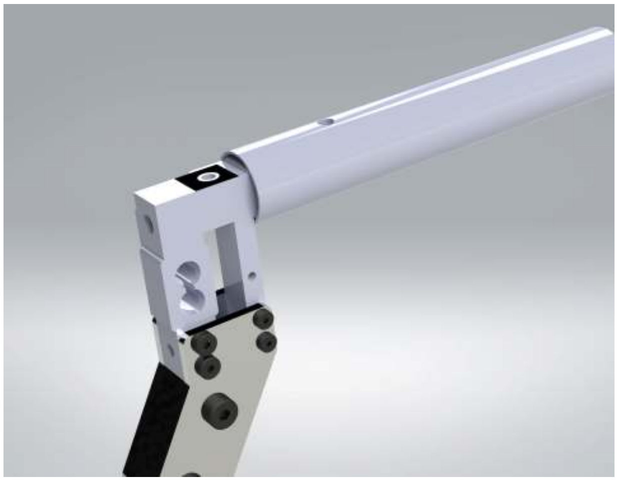

Figure 2.9: Mark III Handle Design

Figure 2.10: Dual Load Cell Handle Design

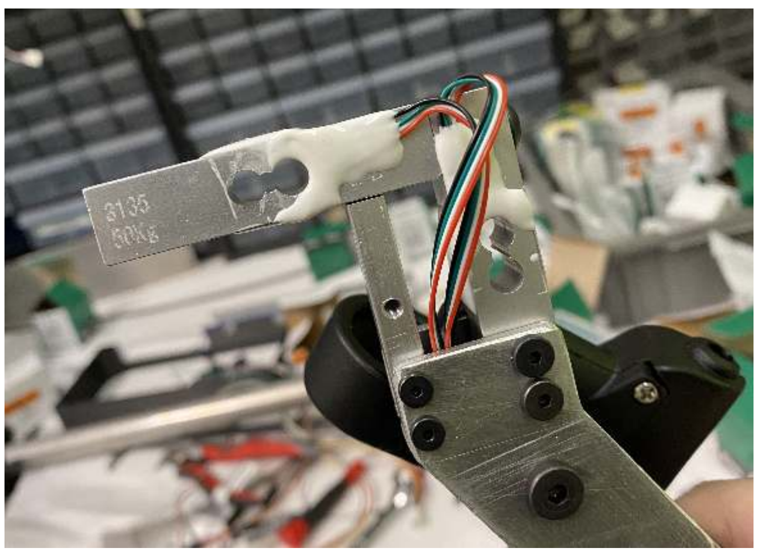

Figure 2.11: Moment-constrained dual load cell handle design

2.2.2 RGB-D Sensor Placement

When Mark I was first designed, it relied only on wheel encoder readings (for speed) and the load cell handle force readings for user emergency detection. However, one important case this sensor combination was unable to detect was the case of a user slowly drifting farther away from the walker's base of support, in which case trips or missteps are much more likely to lead to a fall. Some sort of user torso distance reading was necessary, so an ultrasonic distance sensor was retrofitted to Mark I.

However, one singular point of distance measurement led to too many potential cases of inaccuracies; for example, for the walker, a person actively falling forward, backwards, or just standing would be the same as long as a part of the user's body was at the same place for the ultrasonic sensor to read the same values. We made the decision to use an RGB-depth camera, which will give both visual and spatial awareness of the user's lower body for the correction algorithms.

Figure 2.12: Mark II RealSense placement

Figure 2.13: Mark III RealSense Placement

In Mark II, the RealSense camera was added in a horizontal fashion, as it can be seen from Figure 2.12. A large constraint introduced here was the field of view and minimum detection distance of the RealSense camera. The D435i has one of the widest field of views of DGB-D cameras rated for outside use, but it is still only 69°(H)×42°(V) FOV with a 28cm minimum depth distance. The decision to rotate it 90 degrees, with the goal of widening the FOV to capture more of the user's vertical height, was made when making Mark III, of which can be seen in Figure 2.13. However, the camera still could not get a full view of the user's lower torso, as the seat and the supporting horizontal bar blocked part of the user from the camera's view.

The goal in Mark IV was to design the chassis around the RealSense camera, with the goal of providing the widest possible view of the user in order to determine how effective the RealSense can be as a user state estimation device.

As it can be seen in Figure 2.14, the RealSense camera gets an unobstructed, full view of the user's torso. Based on this update, we were able to test the user lower torso tracking algorithm to its fullest potential, and develop a much more responsive control algorithm.

Figure 2.14: Mark IV RealSense Placement

2.2.3 Automated and Manual Braking

The most important actuation in this walker is braking, as that is the primary method of controlling the user's distance away from the walker. As the primary physics behind the actuated walker's correction algorithm is using the walker's brakes as well as the user's forward momentum to re-position the walker to be in a stable area, a reliable method of controlling brake force was necessary.

Figure 2.15 show the design of the inline braking system. The brake line is connected to a bike brake lever in the handle, which the user can control. However, the servo connected to the linear motion guide can also automatically control the braking calipers, using the same brake wire as the manual brakes.

This was initially explored as a method of providing variable braking force, but it proved difficult to accurately control the braking force exerted by the disk brakes at low speed, as even the slightest contact of the brake pads to the disk brakes was enough to bring the walker to a halt. However, the same mechanism could be implemented using a wheel-lock brake, which uses a lever to physically clamp the back wheel. The same mechanism would allow the walker to retain its functionality even when the battery runs out, as well as giving user manual brake control.

Figure 2.15: Inline braking system in Mark III

2.2.4 Wheel Actuation

Positioning the motors, and transfering power from the motor to the wheels, was an area that needed a lot of planning. As the motors are relatively bulky, if they are placed directly next to the wheels, they may reduce the range of motion for its users. In addition, they serve as the primary variable braking method, so a certain amount of gear reduction was needed to provide enough braking force.

In addition, there were a lot of constraints surrounding the design. The user needed around 24 inches of horizontal space to comfortably walk, so the back legs of the walker need to be wider than that; however, if the wheels or the motors protrude beyond that, they significantly reduce the range of motion, as it gets difficult to navigate tight spaces.

A design that was implemented from Mark I was to have the BLDC motors stick to the upper side of the back legs. The motor shaft would go through the back leg extrusion and a timing belt was connected on the outer wall of the back leg. That motor and belt had control of the back wheels.

In Mark III, in order to make space for the disk brakes, the motor was moved to the front of the walker. This had the benefit of being closer to the electronics. However, applying a braking force at the front wheels of a walker that a user pushes from the back showed to

Figure 2.16: Mark I motor-belt assembly

Figure 2.17: Mark II motor-belt assembly

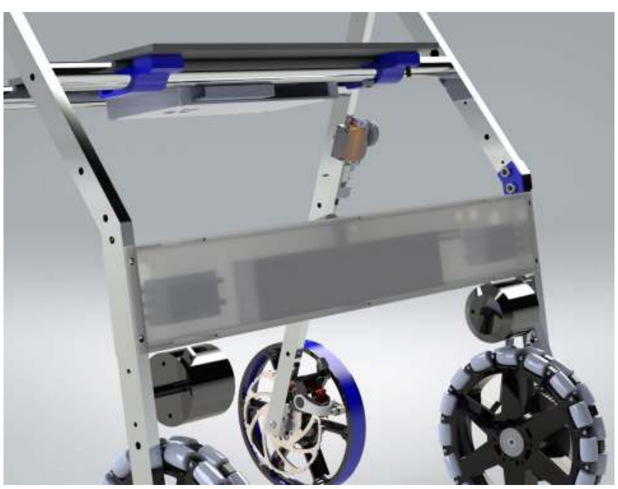

Figure 2.18: Mark III motor-belt assembly with electronics

Figure 2.19: Mark IV motor-belt all wheel drive assembly with belt tensioning

have a tipping effect; in some cases, the walker would tip slightly forward.

In Mark IV, an all-wheel drive system was experimented with; it was expected this would provide better traction when braking, without necessarily increasing the total thickness of the walker.

Figure 2.20: Mark IV belt with tensioning

Figure 2.21: Mark IV front motor

2.2.5 Foldability and Customizability

One important aspect of walkers is that they can be folded to reduce their volume when being in transport; for example, whenever someone using a walker needs to get in a car or take a taxi. All the walker prototypes were designed with the intention to be folded for transport; the first four iterations fold forward/back, while Mark IV folds vertically inwards.

Below, the vertically folded Mark IV can be seen as well. Note the folding cross-sectional beams, as well as the cam feature that allows the RealSense camera to be located at the center of the chassis.

Figure 2.22: Wheels for Mark II and III were designed to stagger to allow a fully flat folded shape.

Figure 2.23: Folded Mark IV

Figure 2.24: Mark IV in folded position

Figure 2.25: Mark IV folding structure

In addition, due to the numerous electrical wires that were inside the chassis extrusions, it was difficult to implement a height adjustable handle until Mark IV. Resting on the flat side of the hex standoffs, Mark IV's handles can easily be adjusted by the user by undoing just two screws.

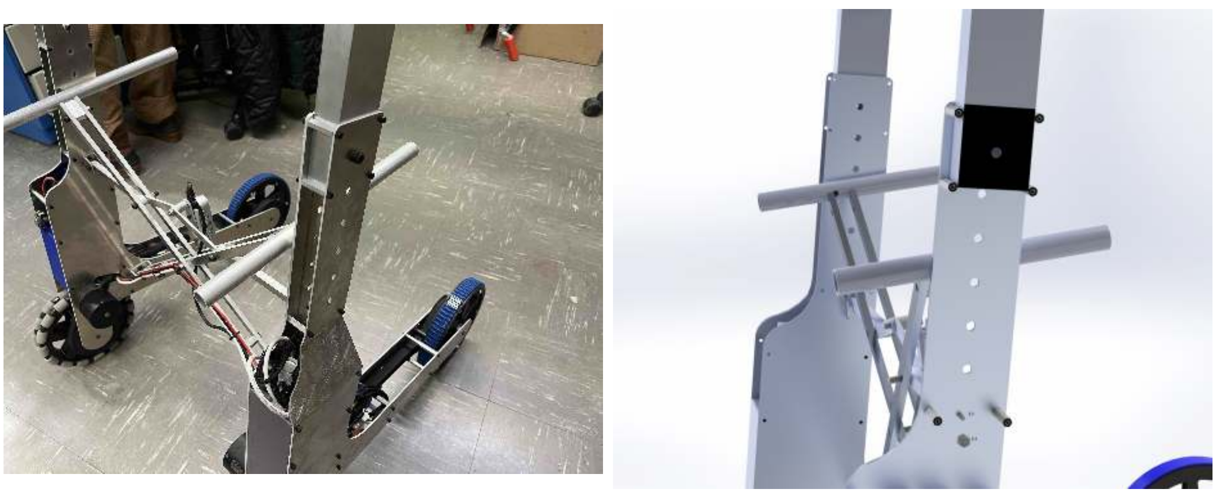

Figure 2.26: Mark IV's adjustable handles

2.3 Manufacturing and Assembly

The first three prototypes, Mark 0 through II, were hand-machined on a variety of tools such as manual mills, manual lathes, drill presses, and many others.

Figure 2.27: Machined and assembled parts for Mark II

Figure 2.28: Mark III's prime structure extrusions

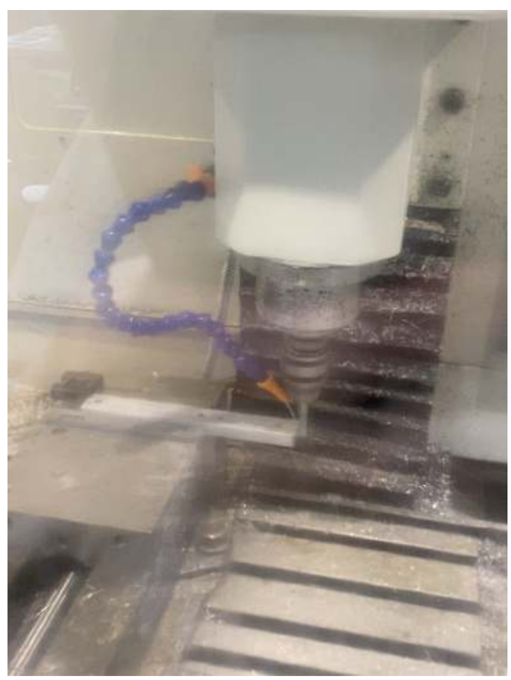

Figure 2.29: Using the Tormach to machine the Mark III legs

A large fraction of the manufacturing for Mark III has been done on the Tormach CNC mill, as it required some specific curved finishes that needed to be milled, as well as a high tolerance requirement for optimal fit.

For Mark IV, in order to simplify manufacture, most parts were cut on a waterjet and bent to generate structural rigidity.

Figure 2.30: Folded sheet metal for Mark IV exterior

Figure 2.31: Skeletal structure of Mark IV

As hardware prototypes iterated, I looked for ways to achieve the same functionality while simplifying the manufacturing process; multiple principles, such as design for manufacturability, were implemented, as well as a wide range of manufacturing methodologies.

2.3.1 Design Decisions for Future Iterations

Current mobility aids are one of two types; one that the user depends on to move around, such as motorized wheelchairs, and the other that depends on the user to use correctly, such as walkers and rollators. Starting from a small scale model, I developed four full sized prototypes, each more effective at integrating the latest technologies to help the user use a walker in a safer manner. There currently exists no such device on the market that integrates the range of sensors and actuators that can be found on this actuated walker. The seamless integration of all components lead to a one-of-its-kind robot, which has great potential to be improved further to one day become an actual device that could be used to help people. Below are some design decisions that could be used to help guide its development.

Braking

An adaptation of the inline braking system from Mark III, mixed with a simple wheel-lock style brake, should be sufficient to serve as a full wheel lock brake system.

Motor Placement and Wheel Control

The wheel-lock style brake will allow the back wheel to be connected to the motor. Either all wheel drive or back wheel drive is sufficient; however, having only the front wheels motorized makes variable braking difficult, along with having problems in case a user wants to drive around on the walker.

RealSense Placement

The RealSense currently is effective for lower body pose tracking starting from a distance between sensor and user of around 40cm. It may be beneficial to design a separate mount for the RealSense to move it as further away as possible from the user. The folding mechanism would need to be designed around sensor placement.

Handles

The moment-constrained load cell handles work great, and should be designed into a future iteration, as it provides valuable additional data.

Folding Mechanism

The brackets holding the folding scissor mechanism in Mark IV bends a little, causing a slight instability. A machined bracket, a thicker bracket, or a sturdier method of fixating the scissor mechanism is needed.

Electronics

In order to enable drive mode, a higher-voltage (32V) battery, instead of the currently used 14.8V battery, would be needed. However, in order to integrate such batteries into the chassis, a custom designed battery pack is likely to be needed.

User Interface

A subtle LED lighting that shows the state of the user (stable, slowly drifting away, in distress) that is tied to the corrective action taken by the walker could be beneficial for the users.

Chapter 3

Software and Control System Design

3.1 Control Diagram and ROS Architecture

The actuated walker has a variety of sensors and actuators that come together to sense the stability of the user and take corrective action. A Jetson Xaviar NX running NVIDIA L4T / Ubuntu 18.04 and ROS Melodic [15] serves as the processor for the actuated walker, running all algorithms locally.

The Xaviar NX pools in data from the RealSense D435i Camera, four load cells through the Phidget Whetstone Bridge, encoder speed values from the VESC motor controllers, as well as GPIO input from the two handle brake switches. These data allow it to determine the stability of the user, and send commands to its two BLDC motors for braking through the VESC motor controllers.

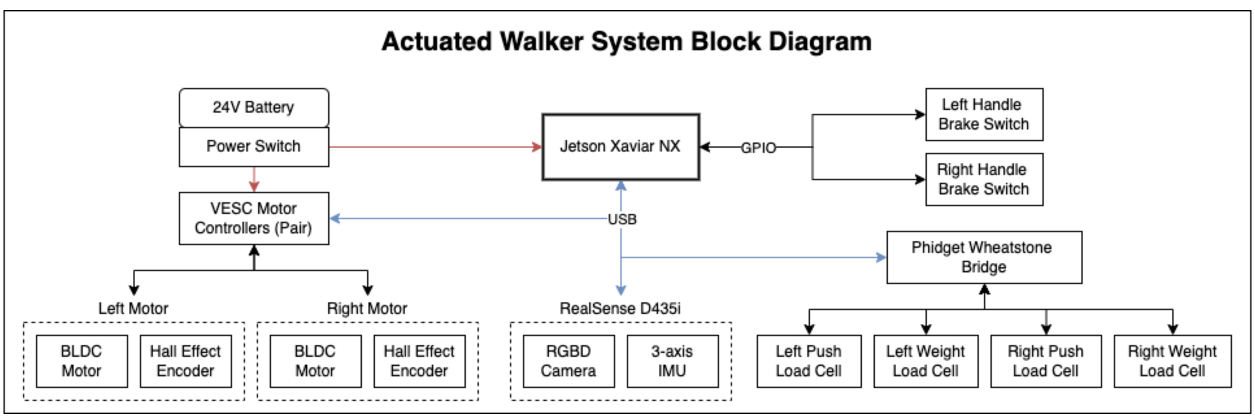

Figure 3.1: The block diagram detailing the electrical components of the walker, including all sensors and actuators.

The logic diagram below goes over the actuated walker's decision process, as well as how it switches between different modes.

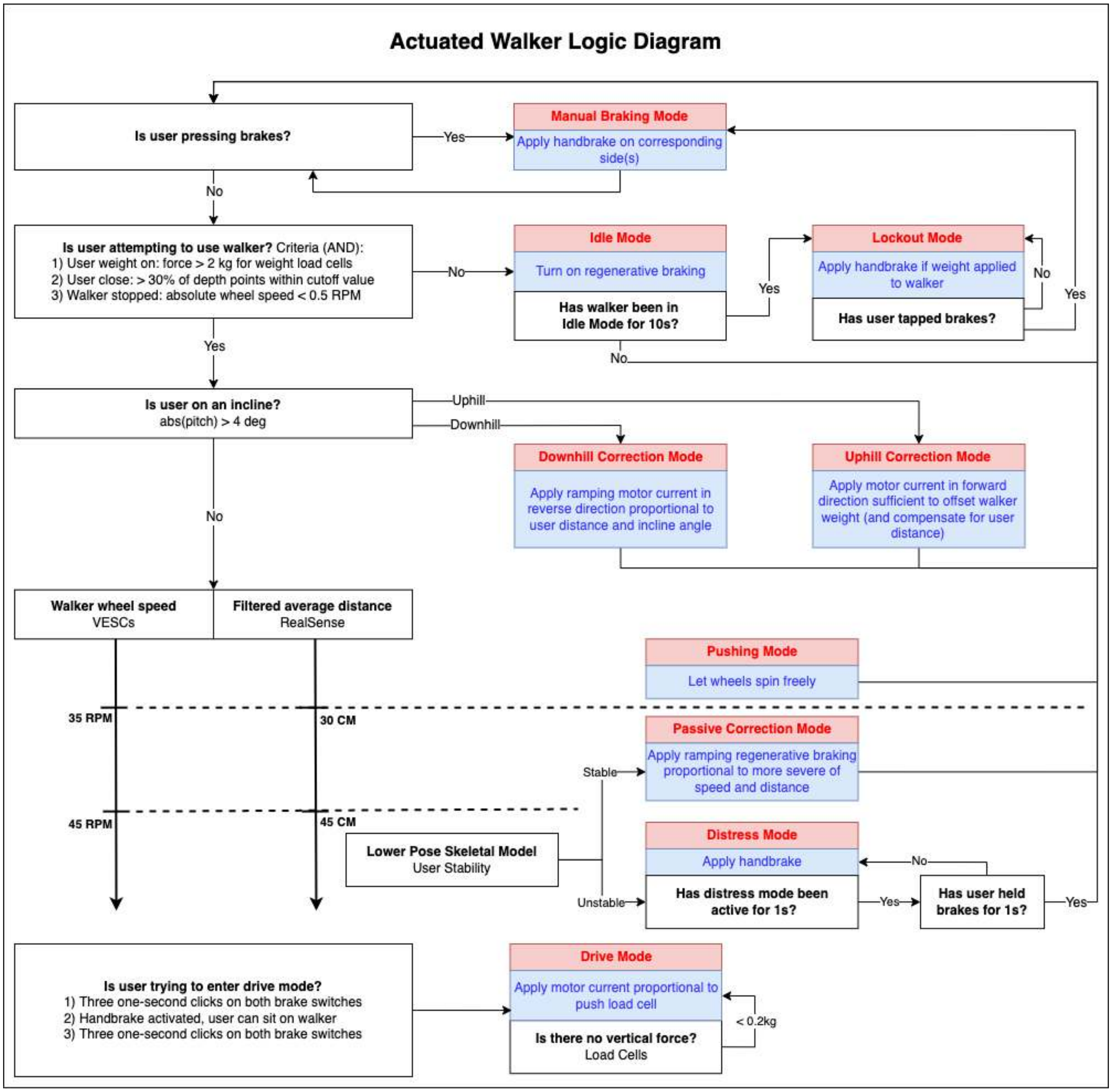

Figure 3.2: The logic diagram of the walker details the state-machine based algorithm we implemented in order to fuse multiple sensor inputs to output the best action for the safety of the user. States (modes) are headlined in red, actions taken in those modes are in blue, and white boxes represent a decision made to determine which mode to transition to.

As described above, the corrective algorithm includes both passive and active actuation, consisting of either low-resistance (regenerative) wheel braking, active full-stop braking (handbraking), or direct motor current commands. The algorithm was implemented through an approximation of a state machine; the walker performs a particular corrective actuation in each state (mode), and which mode is active is determined using an estimate of the user's state derived from sensor input. The state machine is illustrated in Figure 3.2.

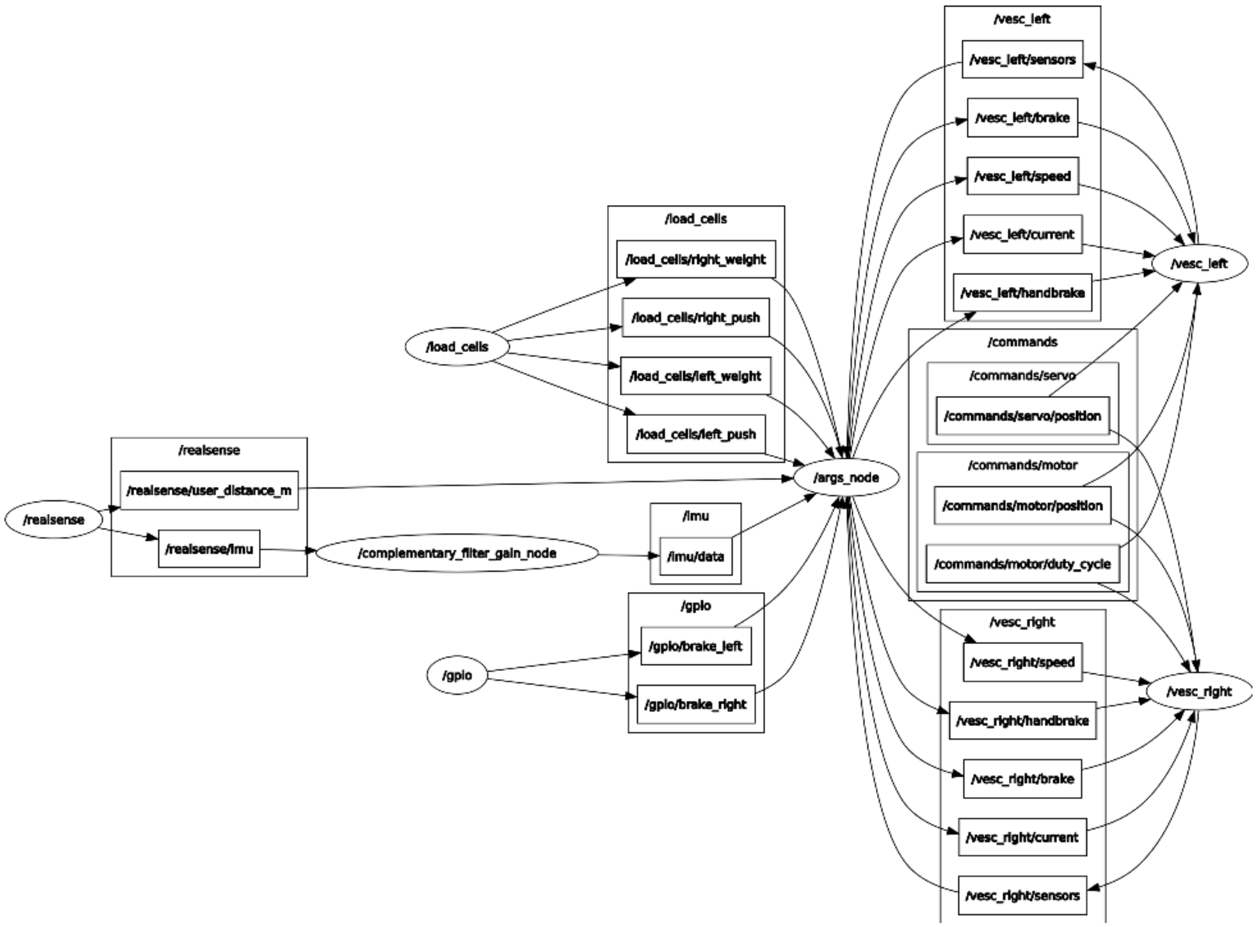

Figure 3.3: The rqt graph of the ROS setup showing the different ROS topics and nodes

The algorithm described in detail in Figure 3.2 functions as follows. A primary decision of the algorithm is whether the user is attempting to use the walker, and is determined based on a combination of nearly all sensor readings. When the walker is not in use, it brakes to prevent idle rollaways, and will "lockout" and brake aggressively if a user begins using it after being idle, in case they forget to engage the brakes themselves.

If the walker is on an uphill incline, it will compensate for its own weight to ease walking. If it is on a downhill, it will brake more aggressively to control the user's speed and posture. The raw rate gyroscope and accelerometer readings from the IMU are fused into an orientation vector using a standard complementary filter [16].

On flat ground, the algorithm integrates wheel speed and the distance of the user's center of mass from the walker's base of support to understand the user's posture. It applies increasingly aggressive corrective actions as the user approaches an unstable state, to attempt to restore stability. If the user is considered to be in distress, the walker will lock its brakes until they have recovered.

An important part of the corrective algorithm in the walker is user pose detection, as it is what dictates the corrective action to be taken by the walker. The implementation of user pose detection is expanded further in the User Pose Detection section.

3.2 Hardware Setup

3.2.1 Xaviar NX, ROS, and Linux Setup

There were multiple dependencies needed to integrate all the different software required for the actuated walker. While ROS Melodic does not support Python 3, the RealSense camera as well as the skeletal tracking model needed to be run on Python 3. An workaround for this was for the pose tracking algorithm to run separately from ROS on Python 3, exporting the data to a text file that would be read by the ROS nodes. There were also some version mismatches between matplotlib and OpenCV, necessitating building OpenCV with the right version of Qt from source.

3.2.2 Networking and Connectivity

In order to make debugging easier, the robot hardware was set up with flexibility in mind. The Xaviar NX can be connected via LAN over USB, and can be ssh'ed in or connected via VNC server. This allows easy access to the Xaviar NX in cases of network disconnect, as well as higher refresh rates for VNC screen sharing. In addition, its MAC address has been registered through Brown WiFi, allowing it to be ssh'ed into from anywhere as long as Brown WiFi is available.

3.3 User Pose Detection

User pose detection is run by two different algorithms for the actuated walker, and the algorithm to use is determined by the distance of the user from the walker body. A user is generally stable when closer to the base of the walker, and is more prone to danger when further away. In addition, when the user is close by to the walker, only part of the user's lower torso can be captured by the RealSense camera, due to its limited field of view. A simple filter-and-average algorithm was implemented as a low-compute, high-speed method to estimate user distance from the walker when the user is walking close by to the walker. When the user is further away, a lower body skeletal tracking model was implemented for distress detection with higher accuracy.

3.3.1 Nominal State: Filtered Average

For identifying the general distance of the user from the walker, an average of points within a certain threshold value was used. First, a few threshold values were set:

1. Far Cutoff Distance: distance where a depth pixel is to be thought of outside the usable range of the walker
2. Close Cutoff Distance: distance where user is assumed to be close enough to the walker to not require emergency braking under distress

For every depth frame read from the RealSense camera, the below two values are calculated:

1. Percentage of depth pixels within Far Cutoff Distance value
2. Average of depth pixels within Far Cutoff Distance value

Based on the above two values, different actions are taken by the correction algorithm:

1. Percentage of depth pixels within Far Cutoff Distance is less than 30%: Assume that there is no user present, enter Idle Mode
2. Percentage of depth pixels within Far Cutoff Distance is more than 30%: Assume that there is user present (given weight reading on load cell handles):
   (a) Average of depth pixels within threshold value is less than Close Cutoff Distance: Assume user is in stable walk, enter Pushing Mode
   (b) Average of depth pixels within threshold value is more than Close Cutoff Distance: run the *Lower Body Skeletal Tracking model*

The below subsection expands on the Lower Body Skeletal Tracking model:

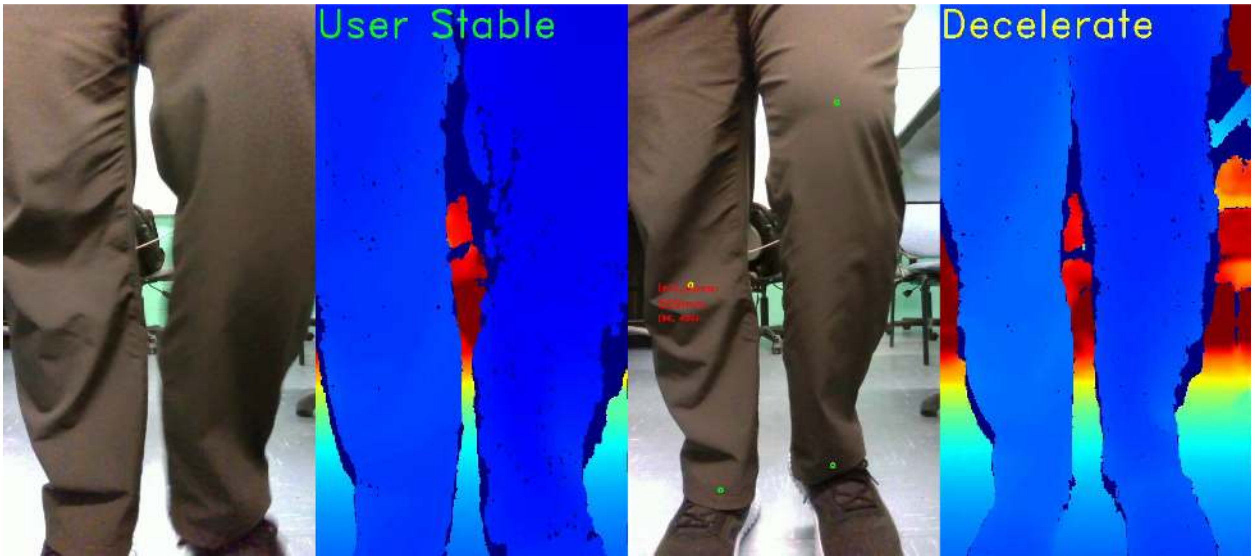

Figure 3.4: The average of depth pixels within the Far Cutoff Distance value is a good measure of general user distance from the walker

3.3.2 Distress Detection: Lower Body Skeletal Tracking

When the user is further away from the walker, it is important to identify if the user's pose is correctable or not. A user's entire lower body becomes visible to the RGB camera on the RealSense when they are around 50cm away from the camera.

Although the skeletal tracking model implemented in trt_pose [17] worked well when most of the user's body was visible, it had difficulty tracking close up shots of only the user lower body. In order to circumvent this problem, in collaboration with Seojin Jang from Kyung Hee University, a lower body pose tracking model was re-trained.

Using the pre-labeled skeletal keypoint locations in the MSCOCO dataset [18], which has been used by trt_pose in training its models, a new set of cropped images with only the lower body was generated. A retrained model from these images was effective in improving lower body skeletal tracking results.

In order to determine if the user is in distress or not, a couple additional values were defined:

1. Hip Knee Distance: RealSense to keypoint distance difference between the hip and the knee that may signal distress due to the thigh being at an unstable angle
2. Knee Ankle Distance: RealSense to keypoint distance difference between the knee and the ankle that may signal distress due to the shin being at an unstable angle

When one or both of a user's leg is in distress, the leg is considerably further from the walker,

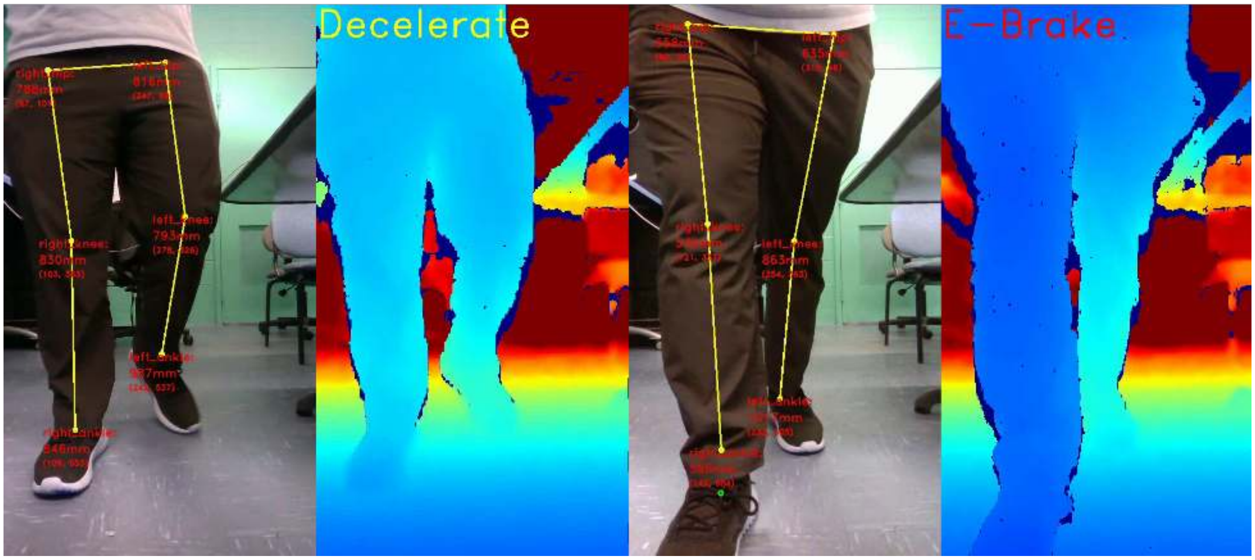

Figure 3.5: The skeletal tracking model serves as an additional data point that can differentiate between a user walking far away from the walker versus a user in active distress

as the falling motion occurs when the walker rolls away when the user needs support. Once the legs are further from the walker, it is in full view of the RealSense, which allows the skeletal tracking model to function and determine RealSense to keypoint distances of the hip, knee, and ankle.

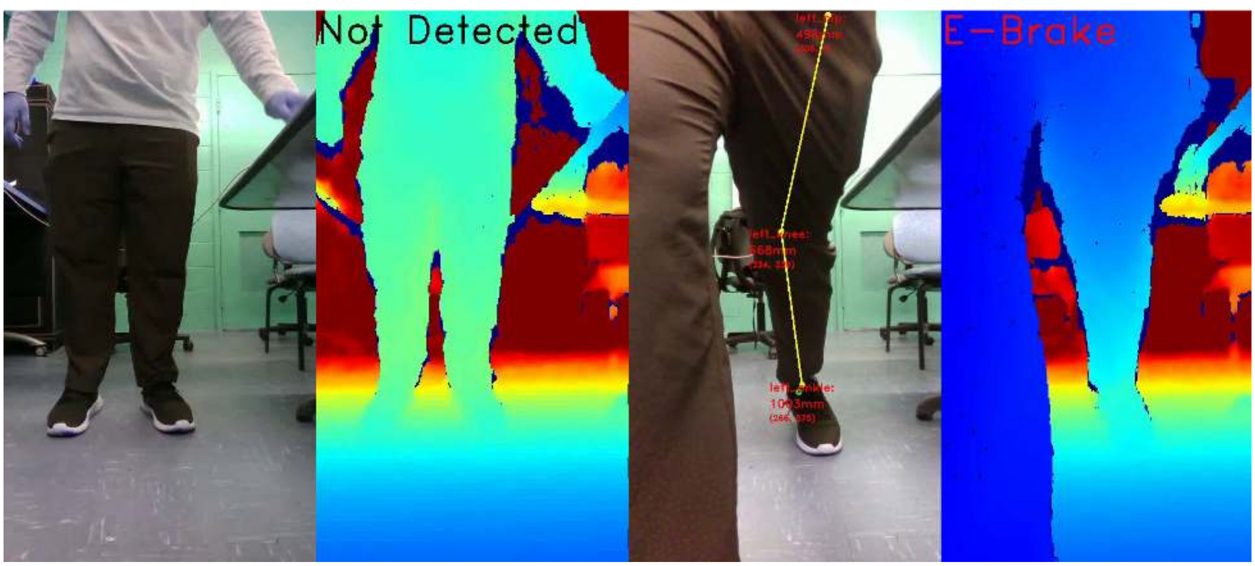

Figure 3.6: The skeletal tracking model is triggered when even one leg is close to the walker with the other leg being in distress

Therefore, when no keypoints are detected, both legs will be close enough to the walker to be correctable, but further than the Close Cutoff Value, triggering the Passive Correction Mode.

When keypoints are detected, if all the distances between the keypoints are smaller than the Hip Knee Distance and Knee Ankle Distance, this will also trigger the Passive Correction Mode. In both cases, the braking force will increase in proportion to the average distance of depth pixels within Far Cutoff Distance. When the distances between the keypoints are larger than the Hip Knee Distance and Knee Ankle Distance for three consecutive frames, the user is thought to be in distress and Distress Mode is triggered.

Chapter 4

User Study and Metrics

With the prime goal of the actuated walker to keep the user in the area where the user's center of mass is over the base of support, the user study of the pose correction algorithm was centered around its effectiveness in keeping the user distance closer to the walker compared to the normal uncorrected use case.

4.1 RealSense Camera Tracking Data Validation

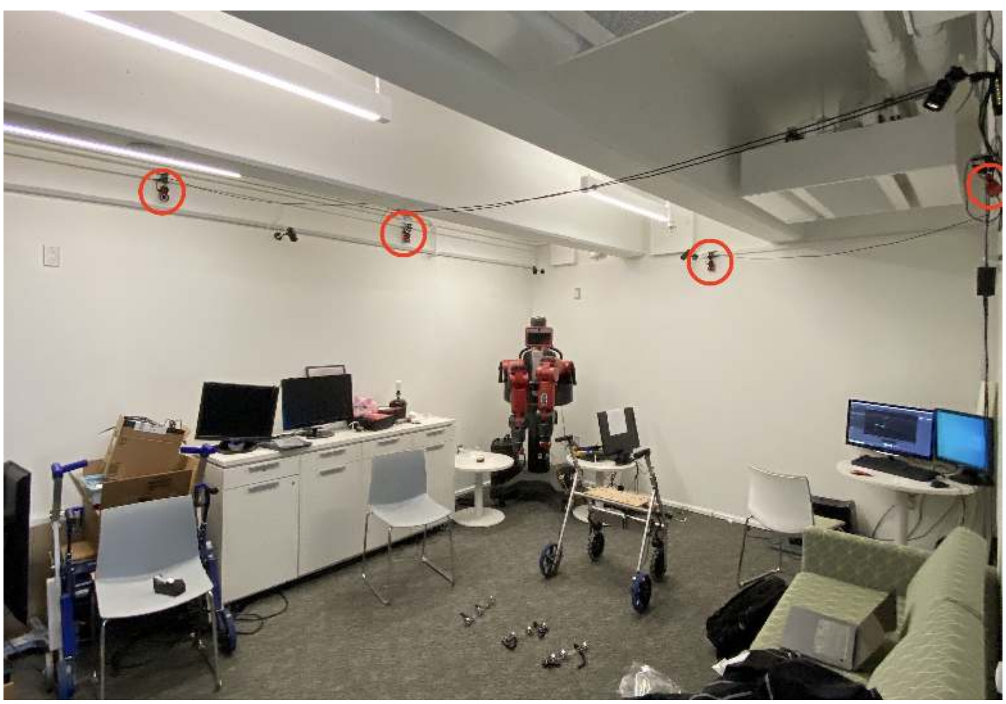

Figure 4.1: Note the ceiling mounted OptiTrack cameras used for motion capture

The first test needed for the actuated walker was to validate that the RealSense camera generated accurate data on user distance from the walker. As the current method for measuring user distance in the nominal scenario was an average of all data points within a certain cutoff value, there was a need to validate that it is a correct value compared to ground truth. In order to verify its accuracy, we used a motion capture system setup using OptiTrack cameras.

With the goal of comparing the RealSense distance values with the ground truth values from the OptiTrack system, custom tracking bars were attached to the two thighs and upper body of the user.

Figure 4.2: Custom MOCAP markers were made with the goal of tracking the location and angle of users' thighs and upper body

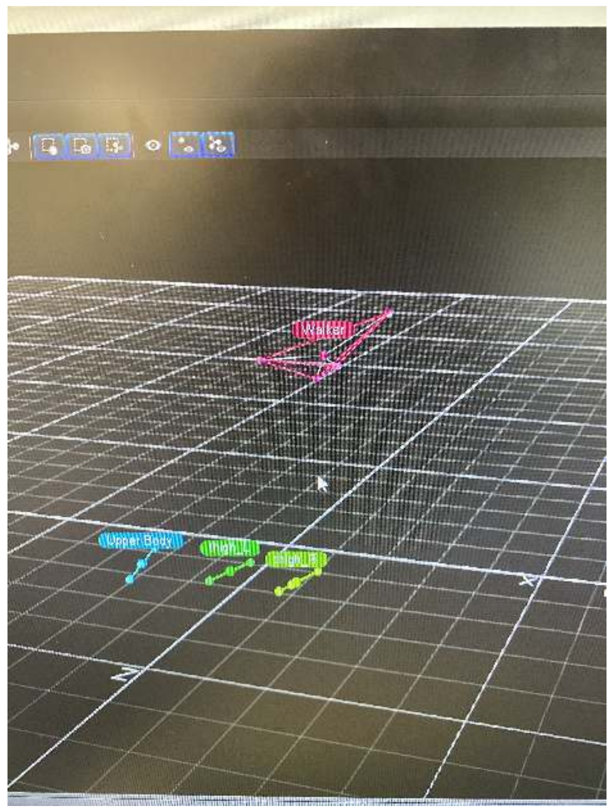

Figure 4.3: The markers on the walker's chassis and the three body markers are tracked in the OptiTrack system

By measuring the difference between the Z coordinates of the upper body tracker and the walker body, the ground truth user distance from the walker can me calculated. That ground truth data can be compared with the distance metric measured by the RealSense camera to validate its accuracy. Since the RealSense camera measures the user distance from the front of the body, and the MOCAP tracker on the upper body was at the back of the body, there is an offset between the ground truth data and the sensor data. However, it is clear that the two data are well-correlated. From this, the accuracy of the RealSense data can be verified, and be used for the pose correction algorithm.

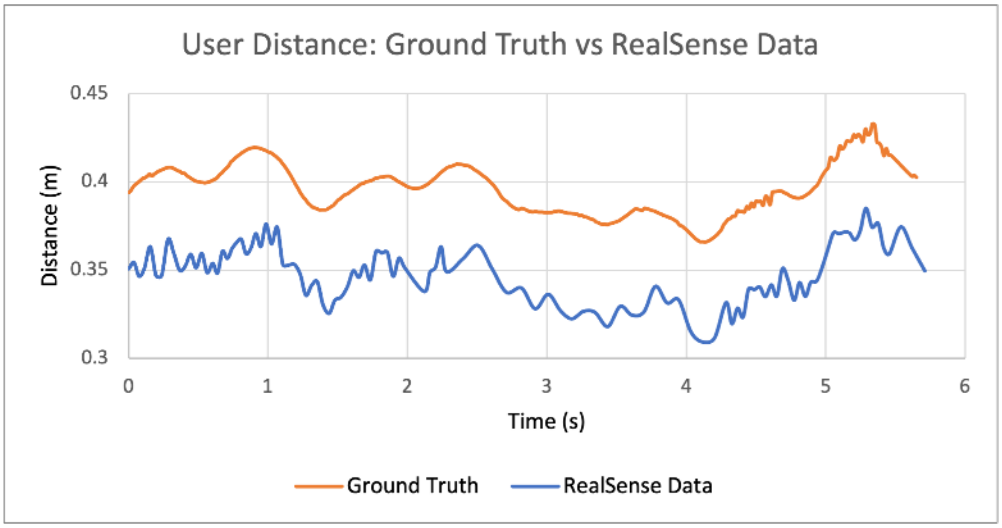

Figure 4.4: Note that the RealSense data closely matches the shape of the ground truth data, other than the 0.05m offset caused by the markers being attached to the back of the legs

4.2 Patterns from Unaware Use Case

This round of testing was designed to identify if the actuated walker's correction algorithm offers an improvement for an untrained, unaware user. By giving instructions to a user that has never used a walker before, we can identify how the pose correction algorithm is effective in improving user posture. In addition, we are able to compare the effectiveness of the algorithm with a person actively trying to stay in the correct position while using the walker.

Below was the script used on instructing testing subjects:

In this series of tests, you will be asked to walk in a straight line, using this wheeled walker in front of you, for a total of three times. Some things for you to note are:

The purpose of a walker is to 1) reduce the weight experienced by your legs by having something to lean into and 2) help you keep your balance. You will need to hold on to the handle and lean in a little so that you are placing part of your body weight on the walker.

For the first experiment, please just walk in a straight line, holding onto the walker!

[**T1: Subject walks in a straight line**]

For the second experiment, try to keep your hips inside the boundary of the walker.

[T2: Subject walks in a straight line, trying to keep hips inside the walker]

For the final experiment, the walker will try to brake so that your hips will be inside the boundary of the walker. Please walk in a straight line, but when the walker brakes, let your momentum get you closer to the walker; in other words, don't try to fight the walker, and just walk straight along with it.

[T3: Subject walks in a straight line with correction algorithm turned on]

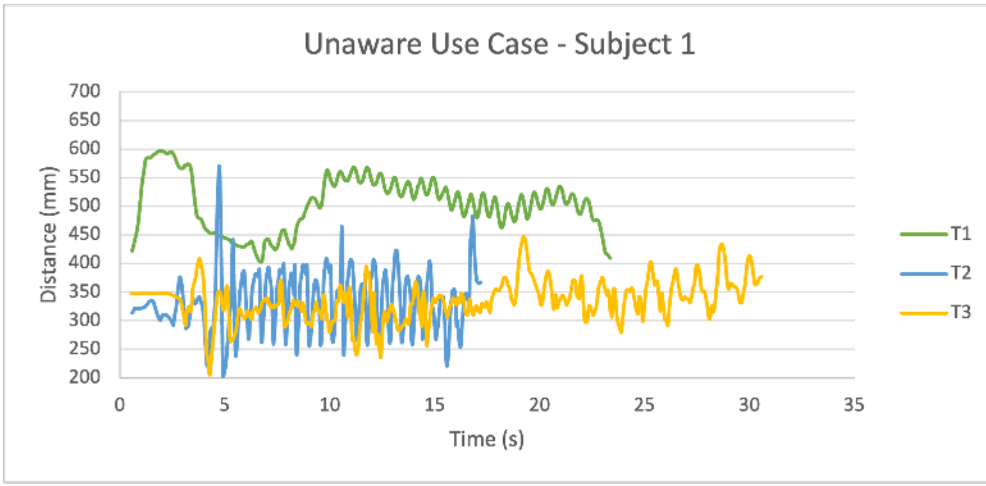

Figure 4.5: The most important metric for stability is the user distance from the walker. Note that the user walks significantly closer to the walker when the algorithm is turned on (T3) compared to the first time using the walker (T1). Also note that with the algorithm on (T3) the user walks more slowly compared to the case of being instructed to walk closer to the walker (T2), as well as having a smaller range of distance away from the walker.

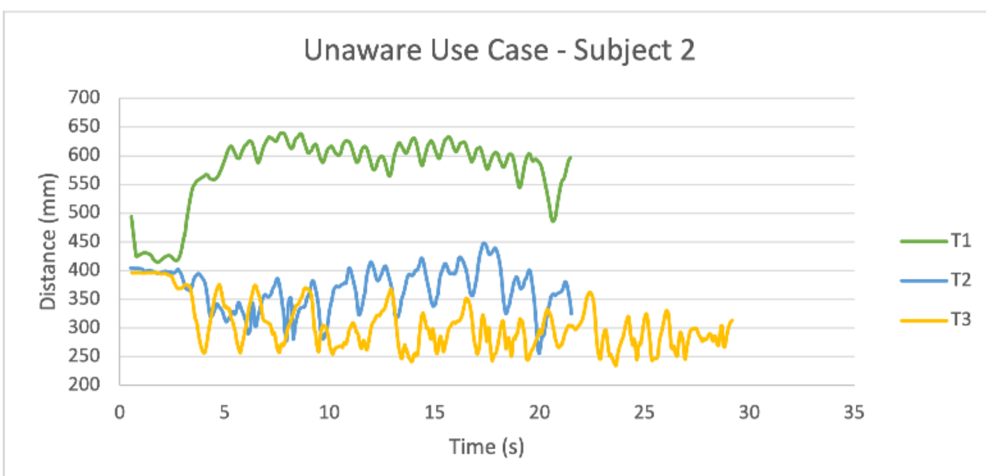

Figure 4.6: A similar reduction in user distance can be seen again between T1 vs T2 / T3

As it can be seen from the graph of the first subject, there are a few clear improvements with the pose correction algorithm compared to the unaware use case. First, the user walks around 250mm closer to the walker under both the informed use case (T2) and pose corrected case (T3). Compared to T2, the range of user distance from the walker is slightly smaller in T3, which means the user distance from the walker is more consistent with the pose correction algorithm turned on compared to the user actively making adjustments to stay within the walker boundaries.

Similar results can be seen from the graph of the second subject as well. The user walked around 300mm closer to the walker in this test case. In addition, the user distance under the pose correction algorithm is slightly closer to the walker in T3 than T2.

In both cases, the pose correction algorithm is successful at:

- Adjusting user's distance from the walker into a safer range
- Reducing the range of user distance from the walker for a more stable walk
- Slowing down the user for safer walker use

4.3 Patterns during Extended Use

This round of testing was designed to identify the prolonged effect of the pose correction algorithm over a longer use period by a user aware of the correct method of using walkers. This test was conducted in three different courses, where the same user walked each course two times, once with the pose correction algorithm turned off and once with it turned on. With the goal of identifying long-term usage patterns, the data was graphed along with a moving average trendline, each with 25 samples.

From all three tests, the pose correction algorithm showed that it reduced user distance even for a trained user. Not only that, it was able to control the user distance consistently over longer periods of time. The distance measurements for the case with correction on also followed a periodic pattern, while the case with pose correction off showed more unexpected changes.

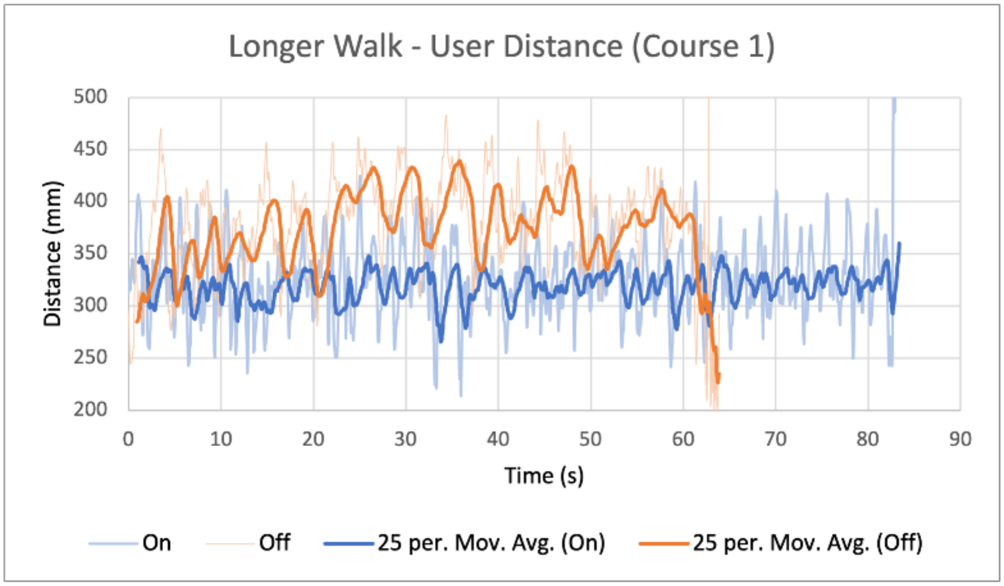

Figure 4.7: Note how the user distance with pose correction enabled stays consistent for over a minute

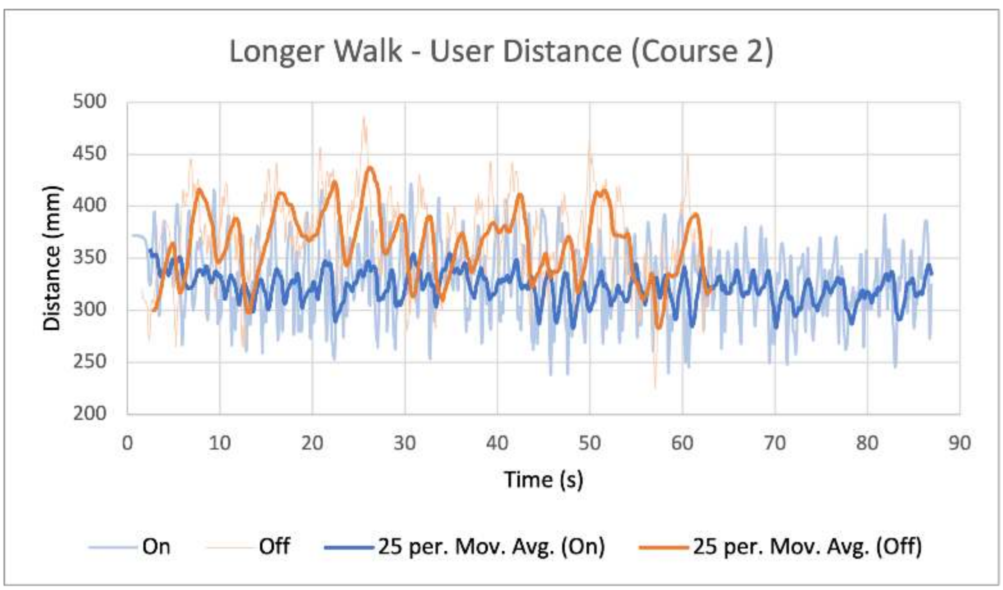

Figure 4.8: The variability of measured distances is much lower for the case with pose correction on

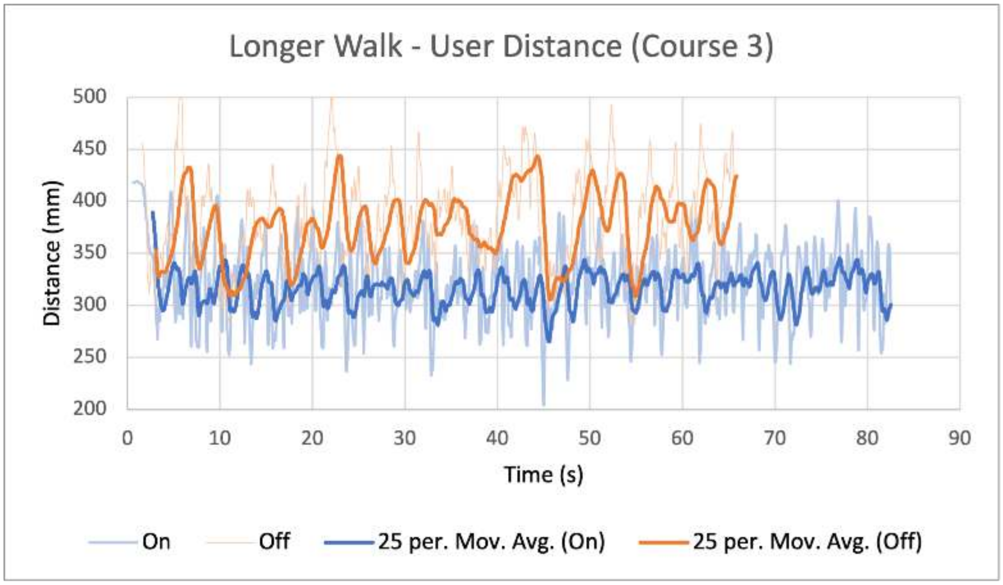

Figure 4.9: The pose correction algorithm also follows a periodic pattern, while the case with it off is more random

Similar to results from the unaware use case, the three graphs show that the pose correction algorithm is capable of achieving the below benefits under extended use:

- Keeping user distance within a preset range
- Reducing the range of user distance from the walker for a more stable walk
- Slowing down the user for safer walker use

Conclusion

Better mobility for senior citizens is an important social problem that will only be exacerbated as our senior population rises. Giving people the ability to move around in a safe and easy manner is crucial for their physical health, mental health, and ability to continue being a member of our society.

However, our current approaches to mobility aids are very limited. People have to give up walking around in order to use the safest mobility aids, such as wheelchairs, or they have to rely on walkers, rollators or canes, which only serve as a mobile handrail for people to hold on when they walk.

In this research project, I have integrated multiple sensors and actuators, with the prime goal of integrating the latest technologies available to make walker use easy and safe. Brushless DC motors were used to provide adjustable soft braking, while dual load cells on the handles allow an accurate detection of user intent and stability. A robust pose-tracking algorithm along with a RealSense depth camera allows real-time user state estimation, and all these integrate to provide the user a seamless experience where the walker adjusts itself to be in a stable position in relation to the user.

We have shown that with a relatively low-cost system, additional safety features, which are highly effective at reducing user distance from the walker, could be implemented. In addition, automatic braking under emergency situations have potential to greatly reduce the severity of falls, as a walker could roll away when the user forgets to brake in distress. A further development of the actuated walker concept has great promise to reduce the number of falls, increase the quality of living for senior citizens, and enable them with the mobility needed to help them continue being an included member of our society.

Bibliography

[1] C. for Disease Prevention, "Take a Stand on Falls." Available at `https://www.cdc.gov/features/older-adult-falls/index.html` (2017/09/22).

[2] D. A. Ganz, Y. Bao, P. G. Shekelle, and L. Z. Rubenstein, "Will My Patient Fall?," *JAMA*, vol. 297, pp. 77–86, 01 2007.

[3] C. for Disease Prevention, "Keep on Your Feet—Preventing Older Adult Falls." Available at `https://www.cdc.gov/injury/features/older-adult-falls/index.html` (2019/12/19).

[4] F. F. Coalition, "Issue brief: Funding for elder falls prevention." Available at `https://www.ncoa.org/wp-content/uploads/Falls-Funding-Issue-Brief-8-15.pdf` (2015/08/12).

[5] H. Kaye, T. Kang, and M. Laplante, "Mobility device use in the united states," *Disability Statistics Report*, 01 2000.

[6] N. M. Gell, R. B. Wallace, A. Z. LaCroix, T. M. Mroz, and K. V. Patel, "Mobility device use in older adults and incidence of falls and worry about falling: Findings from the 2011–2012 national health and aging trends study," *Journal of the American Geriatrics Society*, vol. 63, no. 5, pp. 853–859, 2015.

[7] M. Mundt, J. P. Batista, B. Markert, C. Bollheimer, and T. Laurentius, "Walking with rollator: a systematic review of gait parameters in older persons," *European Review of Aging and Physical Activity*, vol. 16, no. 1, p. 15, 2019.

[8] H. Bateni and B. E. Maki, "Assistive devices for balance and mobility: Benefits, demands, and adverse consequences," *Archives of Physical Medicine and Rehabilitation*, vol. 86, no. 1, pp. 134 – 145, 2005.

[9] G. M. Bradley and C. R. Hernandez, "Geriatric Assistive Devices," *American Family Physician*, vol. 84, pp. 405–411, 08 2011.

[10] M. Martins, C. Santos, A. Frizera, and C. Ramon, "A Review of the Functionalities of Smart Walkers," *Medical Engineering and Physics*, vol. 37, pp. 917–928, 05 2015.

[11] D. L. Wright and T. L. Kemp, "The Dual-Task Methodology and Assessing the Attentional Demands of Ambulation with Walking Devices," *Physical Therapy*, vol. 72, pp. 306–312, 04 1992.

[12] U. Lindemann, M. Schwenk, J. Klenk, M. Kessler, M. Weyrich, F. Kurz, and C. Becker, "Problems of older persons using a wheeled walker," *Aging clinical and experimental research*, vol. 28, 07 2015.

[13] T. Roman de Mettelinge and D. Cambier, "Understanding the relationship between walking aids and falls in older adults: A prospective cohort study," *Journal of Geriatric Physical Therapy*, vol. 38, no. 3, 2015.

[14] J. A. Stevens, K. Thomas, L. Teh, and A. I. Greenspan, "Unintentional fall injuries associated with walkers and canes in older adults treated in u.s. emergency departments," *Journal of the American Geriatrics Society*, vol. 57, no. 8, pp. 1464–1469, 2009.

[15] M. Quigley, K. Conley, B. Gerkey, J. Faust, T. Foote, J. Leibs, R. Wheeler, and A. Ng, "Ros: an open-source robot operating system," vol. 3, 01 2009.

[16] R. Valenti, I. Dryanovski, and J. Xiao, "Keeping a good attitude: A quaternion-based orientation filter for imus and margs," *Sensors*, vol. 15, pp. 19302–19330, 08 2015.

[17] NVIDIA-AI-IOT, "trt-pose." `https://github.com/NVIDIA-AI-IOT/trt_pose`, 2020.

[18] T.-Y. Lin, M. Maire, S. Belongie, L. Bourdev, R. Girshick, J. Hays, P. Perona, D. Ramanan, C. L. Zitnick, and P. Dollár, "Microsoft coco: Common objects in context," 2014.