DESIGN OF AN ACTIVE KNEE EXOSKELETON

MSE 420 Project

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Abstract

An anthropometrically-adjustable, ergonomic, active knee exoskeleton is designed — and its materials are selected — to compensate for the indirect effect of large backpack loads on hikers' knees. This comes at the cost of weight evenly distributed over most of the leg, and that of a power source placed in the aforementioned backpack. By damping the jarring braking motion of hiking downhill, it may also regenerate energy. A motor–drivetrain pair is sized under advisement of the load's derived speed–torque curve, and a position controller is planned to follow the knee angle curve observed over the course of the gait cycle, in a 'moving target' control scheme. These data were collected for design and development in a similar manner to how state feedback may be provided to the controller. Analysis is data-driven and our design is founded on research on hikers' knee injuries, existing knee exoskeletons, and data acquisition.

Contents

1 Background and Introduction

The adventurous activity of backcountry hiking has become an extremely popular outdoor excursion in British Columbia. With the latest developments in a vast network of interconnected backcountry trails, growing popularity and demand, as well as advancements in environment preservation through limitations of campers at any given time by BC Parks, people of all ages seek to explore BC's many beautiful destinations. Hence the nickname, Beautiful British Columbia.

1.1 Motivation

Hikes and trails throughout BC's backcountry vary in length, terrain, and elevation gain. Most hikes begin with a long ascent, to a lake or peak, or even a waterfall. More often than not, these same trails require a hike out after spending a night or two at the destination. This hike requires a descent through the treacherous terrain that has caused many knee injuries in the past. The descent is frequently the cause of injury for most hikers, as one's legs are tired from trek of the days before. However, the intense incline of the ascent also induces the most fatigue throughout the entire trek. Nonetheless, there are many aspects of a trek through the backcountry that revolve around the strength of one's legs and the knee joint. A means of reinforcing this joint would most definitely benefit all levels of hikers.

2 Literature Review

As briefly discussed in *Motivation*, steep descent over treacherous terrain is commonly a cause for injury among hikers. To better understand this biomechanical problem, extensive research was done in the area of backcountry hiking with relevance to knee injury, joint forces and backpack external loading analysis — which is the focus of our project. This section discusses said literature and study findings, to better illustrate and prove a strong motivation. Upon review of a multitude of articles and studies, the following noteworthy information was found in support of the project.[1]

2.1 Hiking Under External Loading

The most common and most utilized knee support for modern hikers — whether the hike be a moderate walk in the park or a month-long backcountry excursion — is some form of walking stick, or *trekking poles*. Trekking poles allow a hiker to transfer some of the weight purely burdened by their legs to their upper body, by allowing the arms to contact the ground. To be more specific, the physical act of hiking is summarized by "long continuous exercise of low intensity, showing positive effects on the cardiovascular system, cardiopulmonary system, and the active and passive structures of the locomotor system" [2]. Additionally, these poles can assist in bearing the weight of an external load, such as a 30-pound backpack full of hiking and overnight camping gear. According to [2],

[1] To adequately find accurate, reliable studies, we chose to utilize SFU's Web of Science database [1].

the larger the external load, the greater the contact forces on the joints, which are also amplified during downhill walking.

Drawing upon knowledge from this and previous courses, the explanation is simple. When a hiker walks downhill, their potential energy is briefly converted to kinetic energy in a fast manner due to gravity's assistance in the direction of motion — downhill. This kinetic energy builds up to a peak at the moment before the forward foot contacts the ground, at which point the energy is mostly absorbed by the contraction of leg muscles to slow the hiker's downhill speed. Thus, repetitive kinetic energy *bursts* continuously contribute to the amplified forces on the knee and ankle joints.

Why focus on the knee joint, and not the ankle? The aforementioned forces are similar to those of the act of running, however exert "much higher compressive forces at the patellofemoral and tibiofemoral joints compared with walking on level ground" [2] — the knee joint. This is why we chose to focus on a hiking solution for the knee, as the ankles are consistently trained and exercised during other activities like running.

While the study in [2] does prove "the use of hiking poles is effective in reducing many of the contributing factors to pain and overuse injuries during downhill hiking," we would like to explore another method to achieve a similar outcome. Rather than allowing the upper body to bear a portion of the stresses and exerted energy of downhill hiking, perhaps there is another method to absorb or store this energy. In addition to the article discussed in this section, many others were found that came to similar conclusions, such as [3].

2.2 Effect on Knee Joint Forces

To better understand the specifics of the knee joint during downhill hiking, a substantial study was found which elaborates on just that. The exact aim of the study was to "determine external and internal loads on the knee joint during downhill walking with and without hiking poles" [4].

Upon compiling their data collection for downhill hiking with and without hiking poles, accurate results showed that poles do in fact have a significant effect in decreasing the knee joint contact forces. The following image from the study shown in figure 1 displays said specific results — knee joint moments over three steps taken on a 25° grade decline, averaged across eight participants.

Clearly, it can be seen that hiking with poles does in fact reduce the peak flexion moments at the knee joint, ranging from 12 to 18% reductions. Even further, reductions in tibiofemoral compressive forces ranged from 14 to 18%, and 11 to 21% for shear forces. Finally, patellofemoral compression forces were reduced by 5 to 17%, but was only significant in the third step — proving that poles even make a difference for a hiker who has built up momentum [4].

These numbers are most definitely significant reductions on the overall impacts of downhill hiking, however there must be another way to do so in a more effective manner, perhaps an efficient manner that can conserve the lost potential energy during descent.

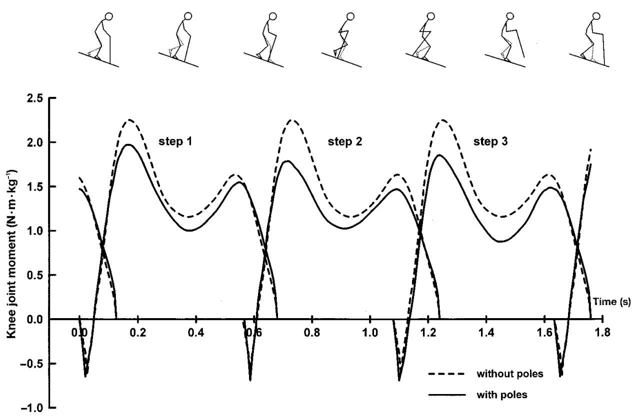


Figure 1: "Knee joint moments during downhill walking without poles (solid lines) and with poles (dashed lines). All stance phases of a cycle are time-normalized and averaged across all participants ($n = 8$)." [4]

2.3 *Hiker's Knee*

There are plenty of articles online that discuss ways to prevent Hiker's Knee, or Patellofemoral Pain Syndrome. A common article explains that "as you descend one leg at a time, the leading knee is obliged to absorb the impact of not only your body weight but also the added forces of going downhill and the weight of whatever you're packing" [5]. The syndrome is so commonly experienced by hikers that it even has its own name.

Furthermore, another study suggests the positive impacts of healthy, active hiking on one's own quality of life. It showed that following total knee arthroplasty (knee replacement surgery), under a three-month program of guided hiking activity, patients showed moderate improvement in abilities and quality of life [6].

2.4 Discussion

Clearly, there is strong evidence that proves knee joints are put under excessive stress during declined hiking — not only the accelerating weight of one's own body, but also the added weight of a heavy backpack. While there are solutions that do exist on the market, such as trekking poles, knee braces and others, there is undeniable potential for a far more effective, efficient device — such as a backdrivable bionic knee exoskeleton.

Gait Cycle Compliance

Knaepen *et al.* investigated the interaction between subjects and active knee exoskeletons (namely, *Knexo* and *AssistOn-Knee*) to get as much use out of them as possible. Exoskeletons such as this can collect accurate knee kinematics data and repeatedly follow their physically-intensive trajectory over the course of a user-specific gait cycle — and we resolve to do the same. It was found that when healthy subjects walking with the device are allowed to control their almost-natural movements, their muscle activity is slightly constrained and their kinematics are somewhat 'weighed down' by the device's inertia. The opposite is true for 'robot-in-charge' operation, which compensates for this deficit. Actuated by pneumatics, Knexo is intrinsically more mechanically compliant than our design is likely to be. However, decreasing the compliance brought out no noticeably significant differences in human interaction, at least at slow speeds and low torques. After this point, extrinsic and intrinsic compliance can be implemented by our controller and by placing a stiffness element in parallel with the device, respectively. [7]

Biomechanical Safety

Celebi *et al.* reported that much research in similar areas also focuses on backdrivability for safety in the event of power loss (but not for energy *regeneration*), as well as the design of controllers to assist only as much as is necessary (e.g., matching the torque of a backpack load such that our wearer remains an active participant). It was also found that exoskeleton and human joint axes must be as colinear as possible, which is a difficult criteria to meet at the same time as mechanical compliance, at least because these devices shift during/between uses, and given that useful imaging equipment may be unavailable. (Repetitive use of a device not honoring this requirement can cause discomfort, injury, or chronic pain.) This is less difficult for the relatively simple but still complex 3-DOF knee joint. While most lower-limb prosthetics are respecting this fact, some well-known knee exoskeletons (e.g., *Lokomat*, *Lopes*) ‘get away’ with using revolute joints, compensating with stiffness elements. In a real knee joint, the tibia rolls *on* the femur, producing noticeable anterior–posterior translation in the parasagittal plane (*rolling and sliding*). [8]

Continuing with this fact, Pratt *et al.* and Sulzer *et al.* designed exoskeletons with kinematic structures (e.g., two revolute joints in series) that approximate the desired planar motion in flexion/extension [9,10]. Similarly, Wang. *et al.* designed an adaptive exoskeleton with a ‘linear actuated cam mechanism’ [11]. Amigo *et al.* and Ergin *et al.* achieved 3-DOF kinematic structures (e.g., three revolute joints in series) [12,13]. Up to and even beyond this point, a balance must be struck between passive degrees of freedom, active (actuated) degrees of freedom, and exoskeleton weight.

2.5 Project Hypothesis and Objectives

Overall, the device concept and design is somewhat intuitive. All the potential energy that is being released and absorbed by the leg joints during decline hiking — why not capture, absorb, and maybe even store it? The idea is similar to how an electric vehicle applies regenerative braking to recharge its battery. The only difference is that instead of kinetic energy (slowing the vehicle down), the device’s source would be the hiker’s potential energy difference in each declining step (after briefly being converted to kinetic energy). It would use the hiker’s body weight and external backpack load to backdrive an actuator and store that generated energy in a battery, which could then be used for a few different applications, such as providing a power source while in the backcountry, for charging cameras, phones, accessories or even emergency items. While not only absorbing and dampening the immense forces experienced by the knee joints, the device also creates a power source.

2.5.1 Objectives

To summarize, the project contains the following objectives.

- Knee joint data collection and analysis of incline and decline walking, to enable gait cycle driven motor output torque curves for incline hiking.
- Design of an adjustable knee brace to fit various ages and genders, as well as house the actuator.
- Design of a backdrivable actuator and drivetrain to allow for *regenerative hiking*.
- A large enough battery to store enough power from hours of decline hiking — which can be used to charge/power devices or drive the motor.

For the proof of concept design for this project, the team will only be completing the data and and CAD model design of the prototype.

3 Materials and Methods

Data will be collected for the upper and lower leg angles over the course of numerous gait cycles, walking uphill and downhill. For each case, this will eventually provide us with the knee angle, angular velocity, angular acceleration, the torque for a backpack load, and the speed–torque curve to to match with that of a to-be-selected motor–drivetrain pair. This would be controlled to follow a moving reference/target — the knee angle — in user-selectable incline and decline operation modes. *Our code is available upon request.*

Note: Using a treadmill would have been advantageous to standardize speeds, gaits, and inclines. Unfortunately, we did not have access to one due to COVID-19. For reference, they are also used for testing knee exoskeletons [7].

Human subject. Only one subject (healthy male, 20 years, 1.7 m, 50 kg) was used, by having access to the to-be-discussed materials. Their height places them in 'normal stature,' albeit in the 17th male percentile, which may affect our results in ways that cannot be quantified due to lack of participants and variation therein.

3.1 Data Acquisition and Analysis

Additional research. Williamson *et al.* found that many *functional electrical stimulation (FES)* systems (including those proposed to aid walking) use knee angle and angular velocity feedback controllers, but few angular position/displacement sensors are practical: [14]

- Flexible *electrogoniometers* need frequent recalibration, are relatively delicate, draw large electrical currents (from what would be our battery), and are difficult keep aligned with the knee joint axis for accuracy. Furthermore, numerically differentiating their angular displacement amplifies pre-existing noise.
- Accelerometers are commonly used but are (slightly) temperature-sensitive and produce other (large) errors, even if they are predicted using segmental tilt via Kalman filter and somewhat corrected by feedback controller (5°+ error simulated).

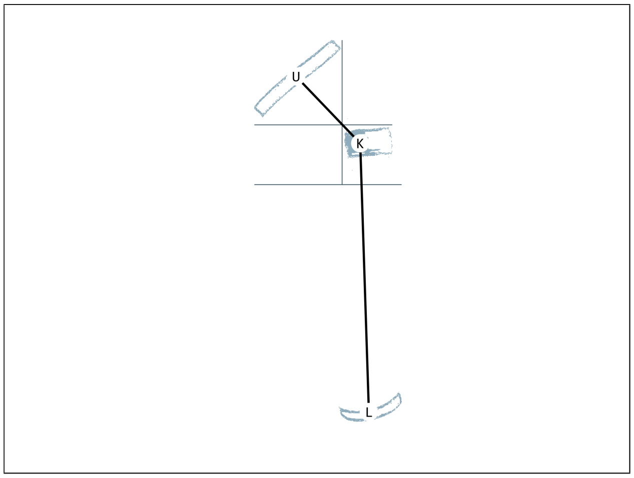


Figure 2: Scatter plot of a motion capture image's most-reflected pixels, overlaid with manually-drawn dividing lines, marker labels (U = upper leg, etc.), and leg segments.

- Relatively new rate gyroscopes may not produce zero-frequency offsets on lengthy periodic measurements (such as over the course of numerous gait cycles) thanks to auto-nulling methods taking care of calibration. Furthermore, gradually-building errors from numerically integrating their angular velocity may be partially corrected using auto-resetting methods in combination with *accelerometers* (4° error) or — better yet — high resolution optical encoders. Also, non-inertial magnetoresistors or Hall effect sensors can bring improved flexibility.

  Numerical integration eliminated the need for a low-pass filter, which can delay or introduce error to what can now be a *real-time* estimate.

3.1.1 Instrumentation — Camera

Motion capture. Sample images were taken of the lower half of the subject with reflective markers placed across their upper and lower leg, as well as mounted on a knee pad as close as possible to its axis of rotation. Various colored markers were considered but decided against in favor of maximum contrast against the surroundings, along with good lighting conditions on a foreground distant from the backdrop.

Image processing. The horizontal and vertical coordinates (x, y) of the brightest pixels were extracted from the aforementioned images (using Matlab's `imread`) and manually grouped with their respective markers. Marker locations were taken as their pixel group's 'center of mass' and spanned by leg segments. (Figure 2)

The inner knee angle is then computed as follows.

$$d_{LU}^2 = d_{LK}^2 + d_{KU}^2 - 2\, d_{LK}\, d_{KU} \cos\theta_K \quad \text{where} \quad d_{ij}^2 = \Delta x_{ij}^2 + \Delta y_{ij}^2 \tag{1}$$

This method was decided against because of (a) the aforementioned manual processing required for each frame of what *would be* a video, (b) lack of a treadmill for recording consistent/steady footage, and (c) striking a balance between number of data points and video processing time. Our slowest frame rate gives (30 frames per second) × (1 second per gait cycle) = only 30 data points per gait cycle; still desirable because generating the above image/frame took over a minute in Matlab. One might switch to Python or C++ if it were not for the following alternatives.

3.1.2 Instrumentation — Encoder

Using a quadrature encoder built into a servo motor was considered for recording both speed and direction, but decided against because of its gearbox's mechanical resistance to what would otherwise be natural movement.

3.1.3 Instrumentation — Gyroscope

See Appendix — Figures. *Our plots are vector graphics. Zoom in!*

Gyroscope pitch. The pitch angles of the gyroscopic sensor (as strapped parallel to the subject's leg segments) are *measured* with respect to whichever part of the transverse plane is closest (with some corrected-for error). For the lower leg (θ_L), these angles are *plotted* over time with respect to the plane's posterior half (relative to the knee) in which most of its time is spent. The opposite is true for the upper leg (θ_U) such that *both* segments' angles are usually (but not always) less than 90° and the knee angle is their sum. If the experiment were repeated, the gyroscopic sensor would be mounted to leg segments at an angle to avoid 90° crossings, their potential error, and the manual data preprocessing that was required to correct for it.

Gait cycle fraction. Arbitrary gait cycle fractions range from $t/T = 0$ to 1 over the course of a period T that is arbitrarily placed for a given data time series. These arbitrary periods are aligned/matched knowing that the leg is approximately straightened when the lower leg reaches maximum *extension* ($\theta_{L,\max}$) and the upper leg reaches approximately the same maximum *flexion* ($180° - \theta_{U,\max}$) at the same time t^*, as follows.

$$\theta_L(t^*) = \max_t(\theta_L) \approx \max_t(180° - \theta_U) = 180° - \theta_U(t^*) \tag{2}$$

The gait cycle period for walking uphill (1.15 s) was consistently longer than walking downhill (1 s), as expected.

Numerical methods. Local peak finding is used to partition time series data into individual gait cycles. Cubic smoothing spline interpolation is used to fit a nominal curve to these partitioned data. Global optimization is used to solve Equation 2, find the maximum motor speed ($\omega_{K,\max}$), and find the maximum motor–drivetrain torque ($T_{\max}$). Finite difference differentiation is used to find the knee angular velocity (ω_K) and acceleration (α_K). Future work might include using the expanded form of the finite differences.

Now, the minimum required motor–drivetrain torque is computed as follows.

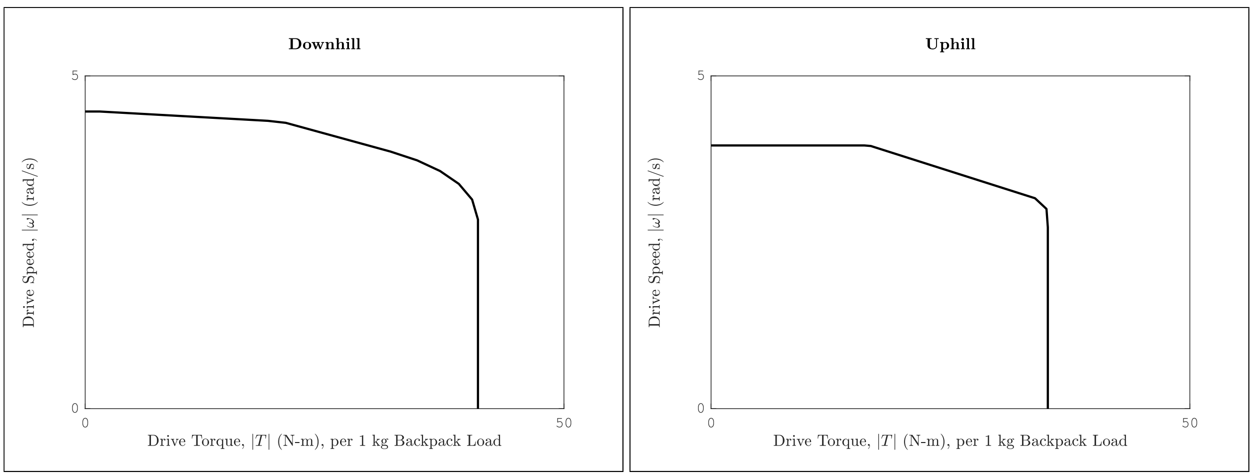
$$T(t) = I(t)\,\alpha_K(t) = m\,D^2(t)\,\frac{\theta_K(t+2\delta t) - 2\theta_K(t+\delta t) + \theta_K(t)}{(\delta t)^2}, \tag{3}$$

where $I(t)$ is the approximate rotational inertia of a backpack of mass m and the knee-to-backpack distance is as follows.

$$D(t) = \sqrt{L_T^2 + L_U^2 - 2L_T L_U \cos(\theta_U + 90°)} \approx \text{const. (modeled as variable).} \tag{4}$$

Calculated for a ~6′-tall hiker, L_T and L_U are the lengths of the trunk (below the backpack) and the upper leg, respectively.

The sign of ω_K is indicative of whether the backpack load is being raised (+) or lowered (−) and thus whether power must be delivered by or may regenerated in the motor/drivetrain, respectively. (We assume that we want to regenerate as much energy as possible.) The sign of torque T is merely indicative of the output direction thereof. The absolute value of both is used to generate parametric plots — complex speed–torque curves, around which an appropriately-simplifying 'envelope' is drawn (Figure 3), knowing motor speed–torque characteristics.



(a) Walking downhill: $\omega_{K,\max} = 4.465$ rad/s. $T_{K,\max} = 41.02$ N-m.

(b) Walking uphill: $\omega_{K,\max} = 3.955$ rad/s. $T_{K,\max} = 35.17$ N-m.

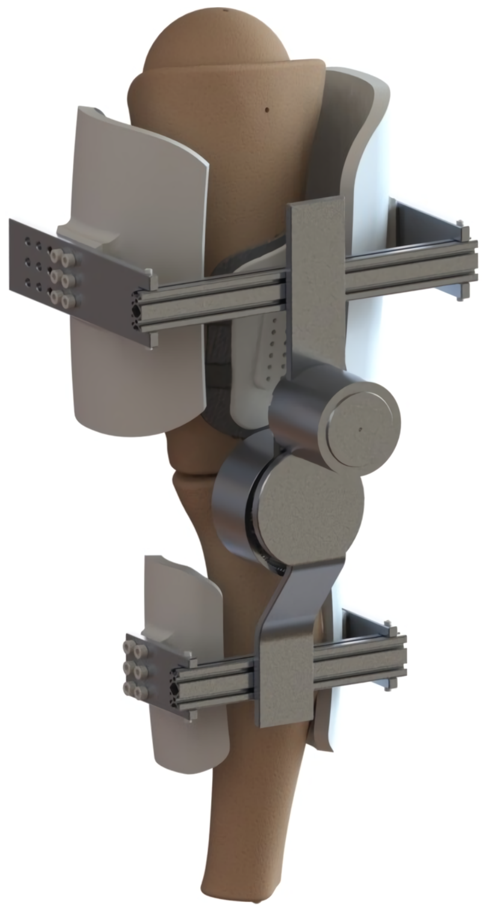
Figure 3: Required speed–torque curve envelopes *(zoom in!)*, the largest of which (a) will be matched with that of a to-be-selected motor–drivetrain pair. The envelopes scale linearly with torque (along the horizontal axis), but not with wearer anthropometry.

4 Discussion of Outcomes

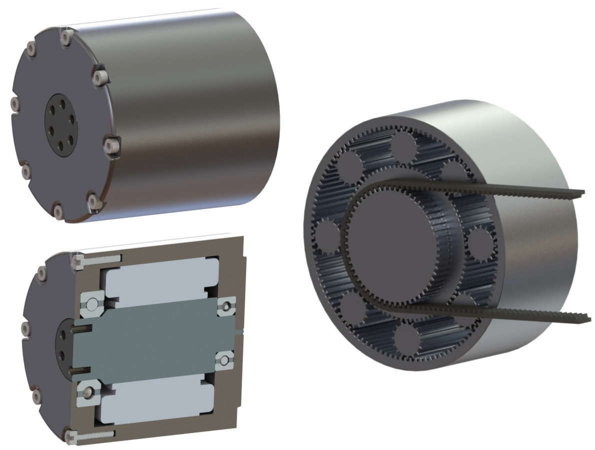
Figure 4 showcases renderings and cutaway views of our exoskeleton, its motor, and its drivetrain.

4.1 Exoskeleton Frame, Structure, and Orthotics

Frame and Structure. We referred to a 50th percentile male for the dimensions of the upper and lower leg. The frame is to be made completely out of aluminum. Complex parts (mounting plates and the structure) are to be milled from aluminum 6061 flat bars. Simple parts use 20 by 40-mm t-slot aluminum extrusions, whose purpose is to decrease the cost of manufacturing and increase design simplicity. Moreover, due to the better corrosion resistance offered by aluminum, we can expect the frame to have a longer life —



(a) Our knee exoskeleton.



(b) Actuator (left) and gearbox (right) cutaways.

Figure 4

specifically given that the exoskeleton will be used in harsh environments. Lastly, it has a considerable strength–weight ratio, making it the ideal choice of material.

Orthotics. The orthotics are the interface between the frame of the exoskeleton and the user's body. Therefore, the most important design objectives are comfort and function. To make it adaptable to more body types and shapes, we left a provision in the exoskeleton frame for modular dimensions. Thus, the orthotics are modular in the three anthropometric planes; sagittal, coronal, and transverse. The base of the orthotics are to be manufactured out of polypropylene — a lightweight and stiff thermoplastic. At the interface between the orthotic and the skin, a silicone sheet is to provide a gripping surface. A layer of polyethylene foam is also to be inserted for comfort fit and shock absorption.

4.2 Exoskeleton Actuator and Gearbox

An actuator and gearbox system was designed to achieve the target peak torque of 41.02 N-m and peak angular velocity of 42.64 rpm. A TBM 7647-B frameless motor was decided upon (made by Kollmorgen), which offers a rated torque and speed of 2.13 N-m and 1025 rpm, respectively. With this in mind, to achieve a minimum torque safety factor of 1.25 while satisfying the target output speed, the gearbox was designed to have a 24:1 effective gear ratio. After researching different gearbox designs, a compound planetary gearbox was decided upon. This decision stems from two motivators: it offers good backdrivability (for regenerative breaking) and can be put into a short housing (to maintain a relatively sleek, compact design).

	Torque	Speed
Input	2.13 N-m	1025 rpm
Output	51.21 N-m	42.64 rpm

Table 1: Input and output of the combined actuator and gearbox.

The actuator, which houses the frameless motor, comprises of a shaft, two sealed ball bearings, an encoder, and its housing. Aluminum 6061 is to be used on the housing to improve the corrosion resistance. For the shaft, low-carbon steel was decided upon for its good tensile strength. On the other hand, medium-carbon steel was decided upon for the gearbox assembly because we wanted higher tensile strength for the gear teeth. Finally, the torque is transferred from the actuator output shaft to the gearbox input 'sun' gear by a GT2-profile timing belt of 3-mm pitch. Although an actuator is conventionally connected directly to a gearbox, our gearbox is to be placed in parallel with the actuator to cut down on the total length of the actuator–gearbox system. As such, the pair does not protrude far from the side of the knee joint and allows for a relatively sleek design.

4.3 Cost Analysis

Our estimated cost for the exoskeleton is 1190 CAD per leg — 2380 CAD total. The bulk of the cost comes from the 700 CAD per motor, placing the estimated cost of the motor–drivetrain pair at around 800 CAD. The other estimated costs of the exoskeleton are as follows: orthotics = 40 CAD, structure and frame = 150 CAD, and gearbox = 200 CAD. Although the price point is quite high, it is an acceptable price relative to the prices of other hiking equipment (such as tents, backpacks, boots, and clothing). In addition, we believe that we could bring the costs down to a total of 1640 CAD by finding a cheaper motor and optimizing the designs for the actuator housing and exoskeleton frame; reducing cost of materials.

5 Conclusions

Accurate knee angle and speed–torque loading curves were generated, the latter of which advised the motor selection and drivetrain design for a 'bionic knee actuator device.' In the process, much was learned about the biomechanics of the knee joint as well as the knowledge needed to design a proper exoskeleton prototype. Although a team member had previous experience with exoskeletons, this proved to be a challenging project. The data acquired for motion of the knee joint laid a foundation for the mechanical design.

It would have been challenging to select/design an appropriate motor and drivetrain. Furthermore, the information amassed from several sources provided good design principles and requirements to guide our design stage. Overall, the team used the information available to appropriately construct an active knee exoskeleton.

5.1 Challenges and Limitations

Accommodating all sizes, genders, and body types. As always, when it comes to designing products in the biomechanics sector, this is a recurring issue. It is difficult to design our exoskeleton to perform perfectly for all types of users.

Overall cost. In summary, according to section 4.3, the prototype has an approximate value of 2380 CAD. A question was asked during our project presentation — *is this within the price range of an avid backcountry enthusiast?* In short, it is not a solution that all will seek. The target market are those who experience knee problems when hiking. Although this does not encompass the entire market, there is certainly a wide demand for such a product. In terms of price, some backcountry gear is in fact close to this price range, such as tents. A search on Mountain Equipment Co-op shows the top eight tents in terms of price ranging from $889.95 to $7,149.99 [15]. If people go this far so as to comfortably camp (or survive) the night, there is no doubt that there will be a customer segment for this product.

Exoskeleton added weight. Despite the added weight of the prototype and sizable battery bank stored in the hikers backpack, their effects are negligible when compared to the benefits of the system. It is/will be designed such that its own added weight is compensated for. Additionally, the exoskeleton acts primarily as an energy absorber, hence the added weight has effectively a beneficial impact on the hiker's strides on descent.

5.2 Future Recommendations

Further design iterations should be as a result of (a) collecting data from a wider anthropometric range of potential users (as discussed in Section 3), (b) attempts to make it more compact, and (c) making use of less expensive materials and/or components without compromising the integrity of the design.

Work Allocation

- Ibrahim Helal — research: knee injuries analysis.
- James Liu — mechanical design and SolidWorks modeling.
- Keegan Green — data collection and analysis.
- Ryan Fielding — research: motivation and design objectives.

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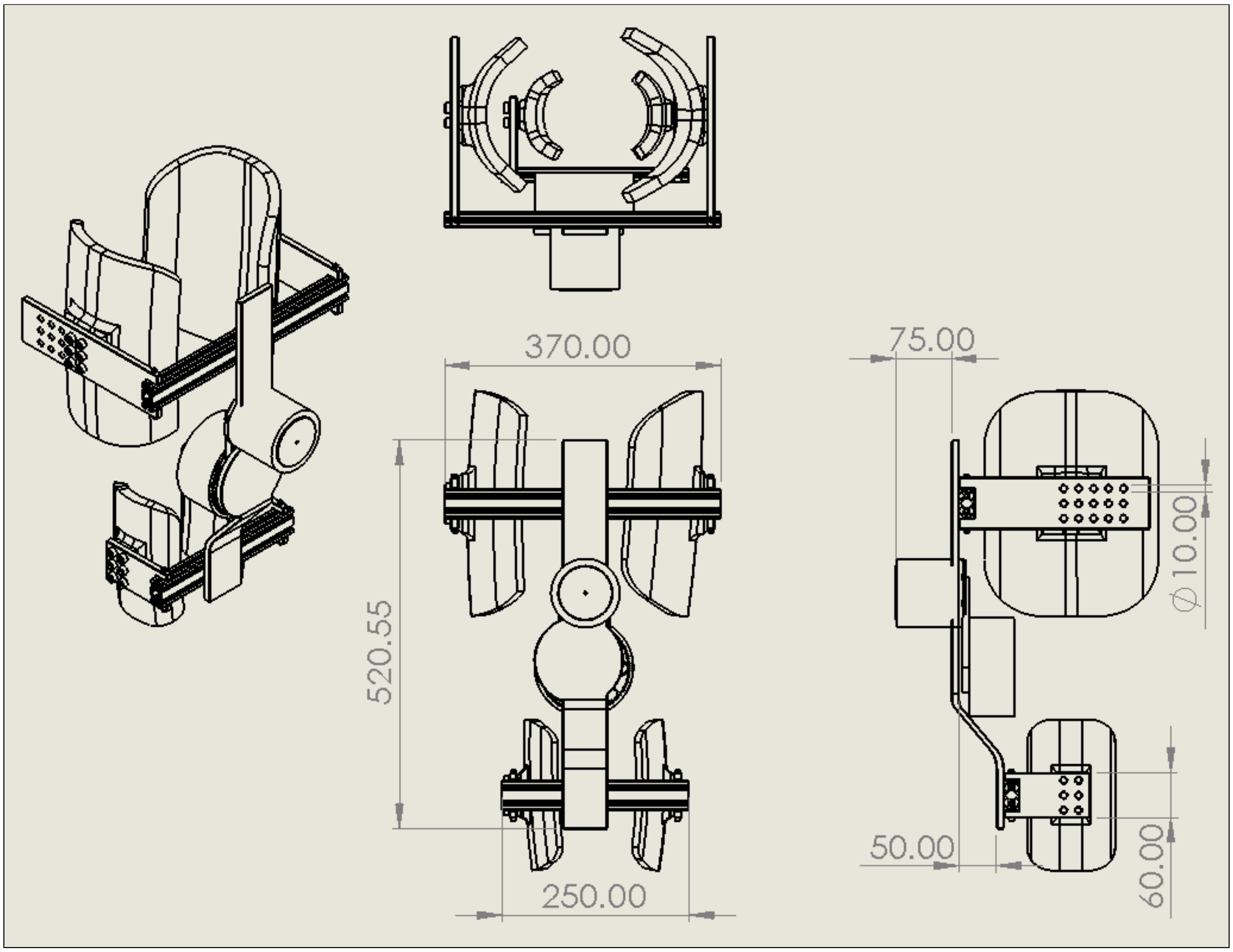
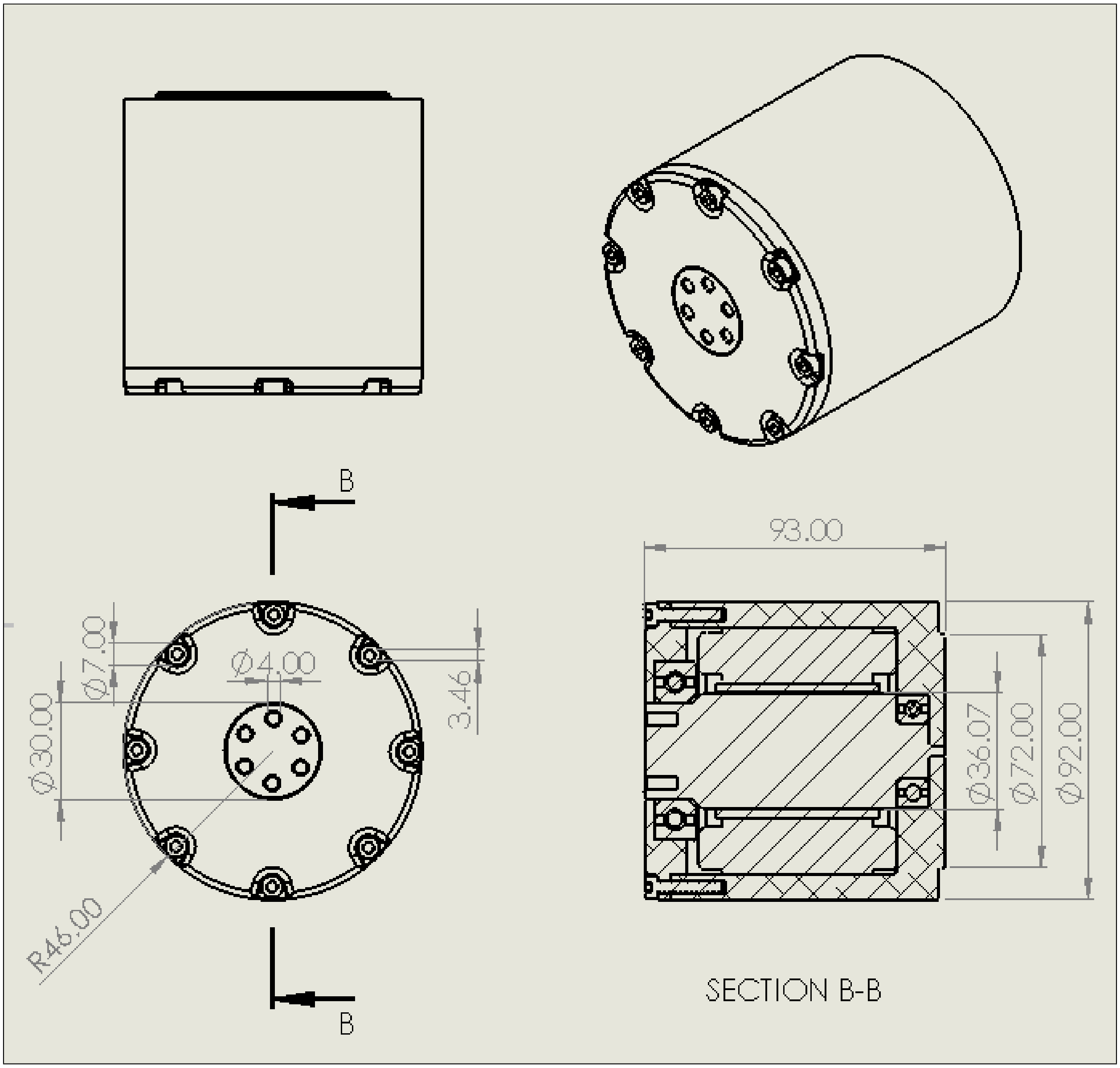
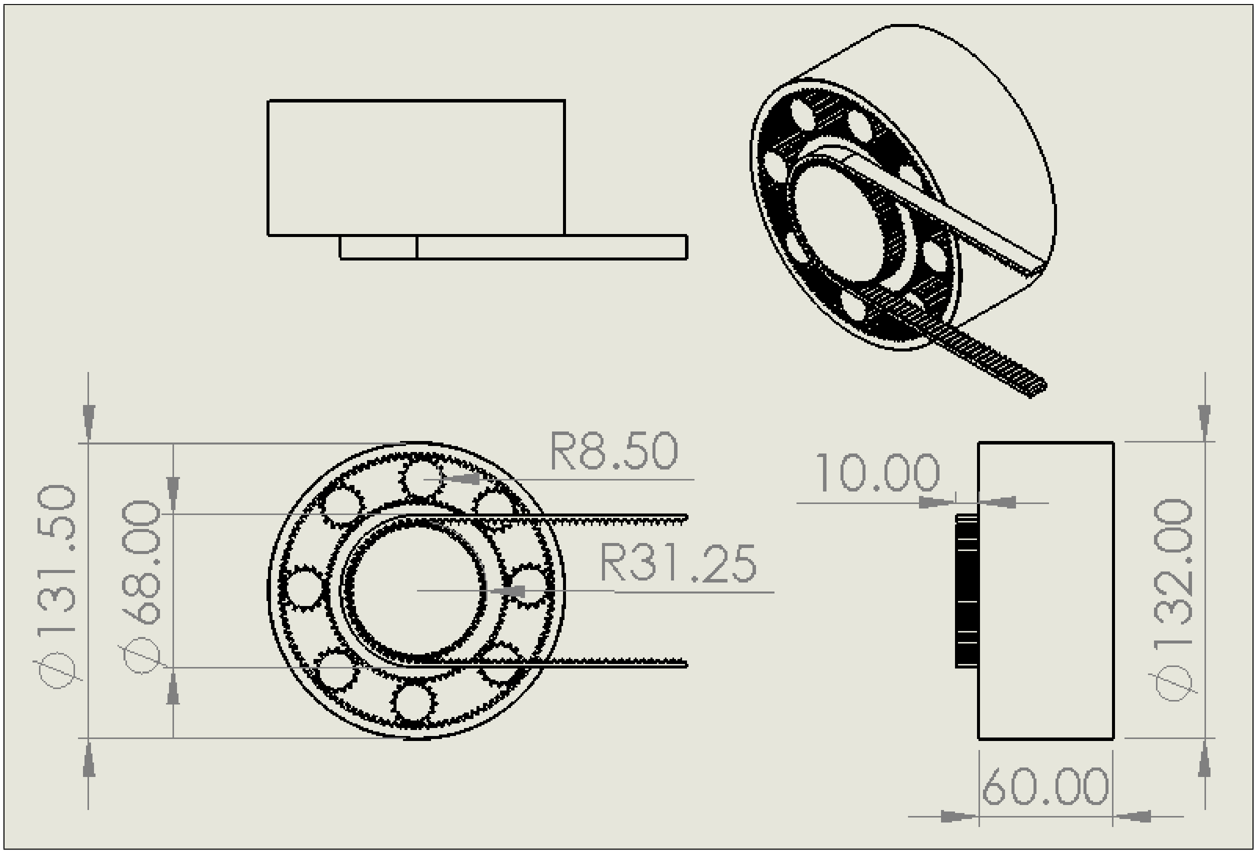
Appendix — Drawings



Exoskeleton — selected views and dimensions.



Actuator — selected views and dimensions.



Gearbox — selected views and dimensions.

Appendix — Figures

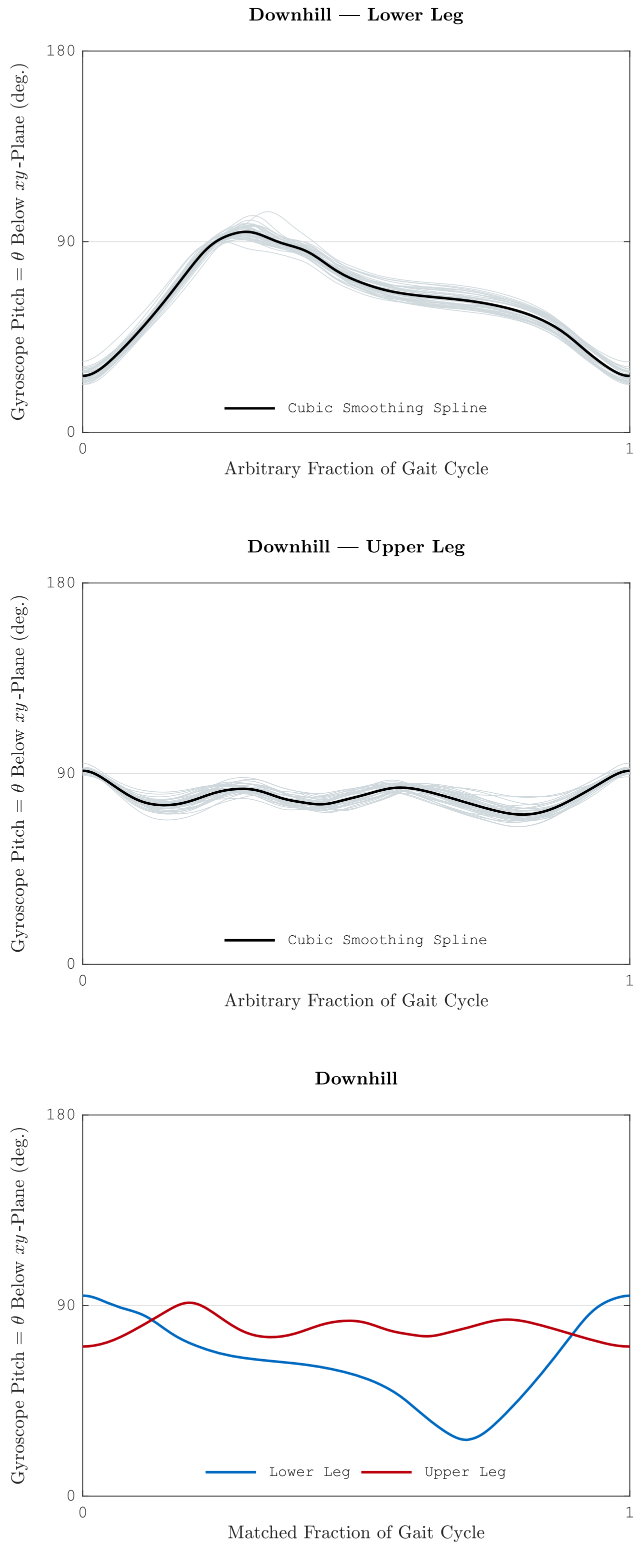


Figure A1 (above three): Walking downhill: 'misaligned' lower leg angle $\theta_L(t + \Delta t_L)$, 'misaligned' upper leg angle $\theta_U(t + \Delta t_U)$, and 'aligned' $\theta_L(t)$ *&* $\theta_U(t)$.

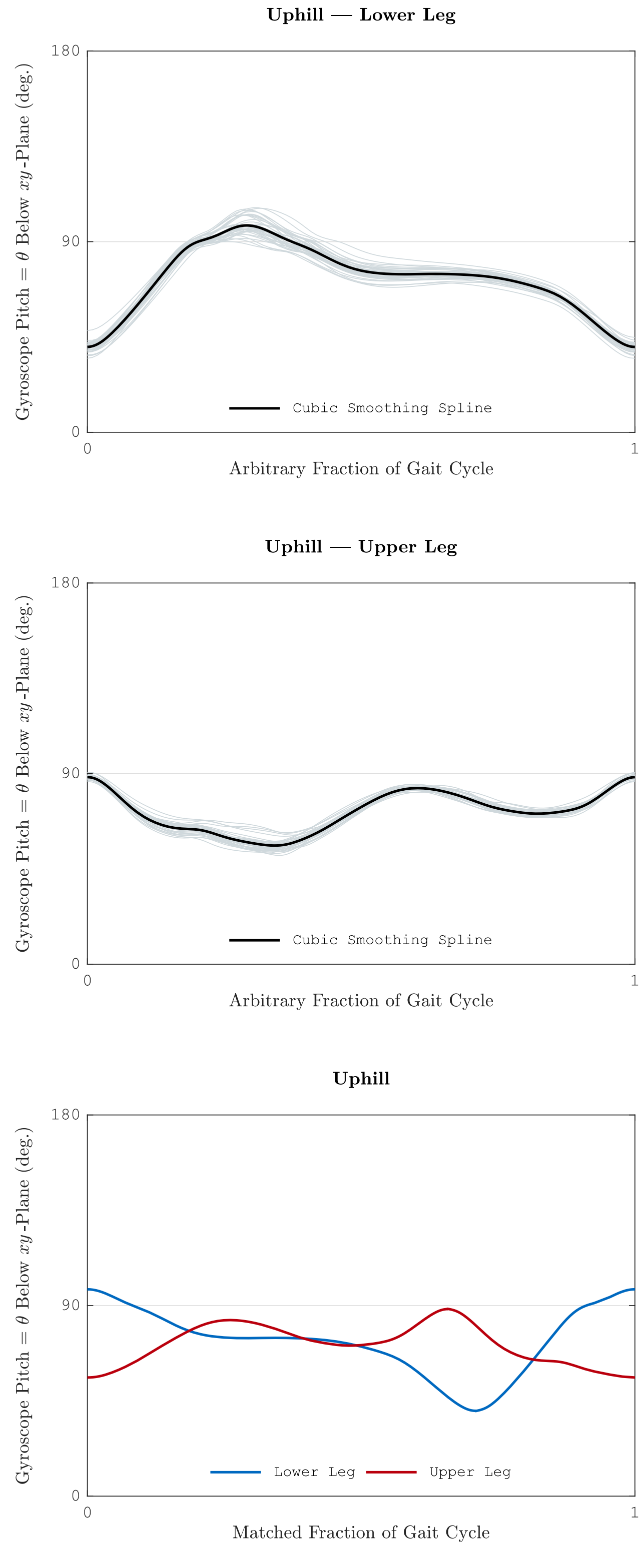


Figure A2 (above three): Walking uphill: 'misaligned' lower leg angle $\theta_L(t + \Delta t_L)$, 'misaligned' upper leg angle $\theta_U(t + \Delta t_U)$, and 'aligned' $\theta_L(t)$ *&* $\theta_U(t)$.

Appendix — Figures (Cont.)

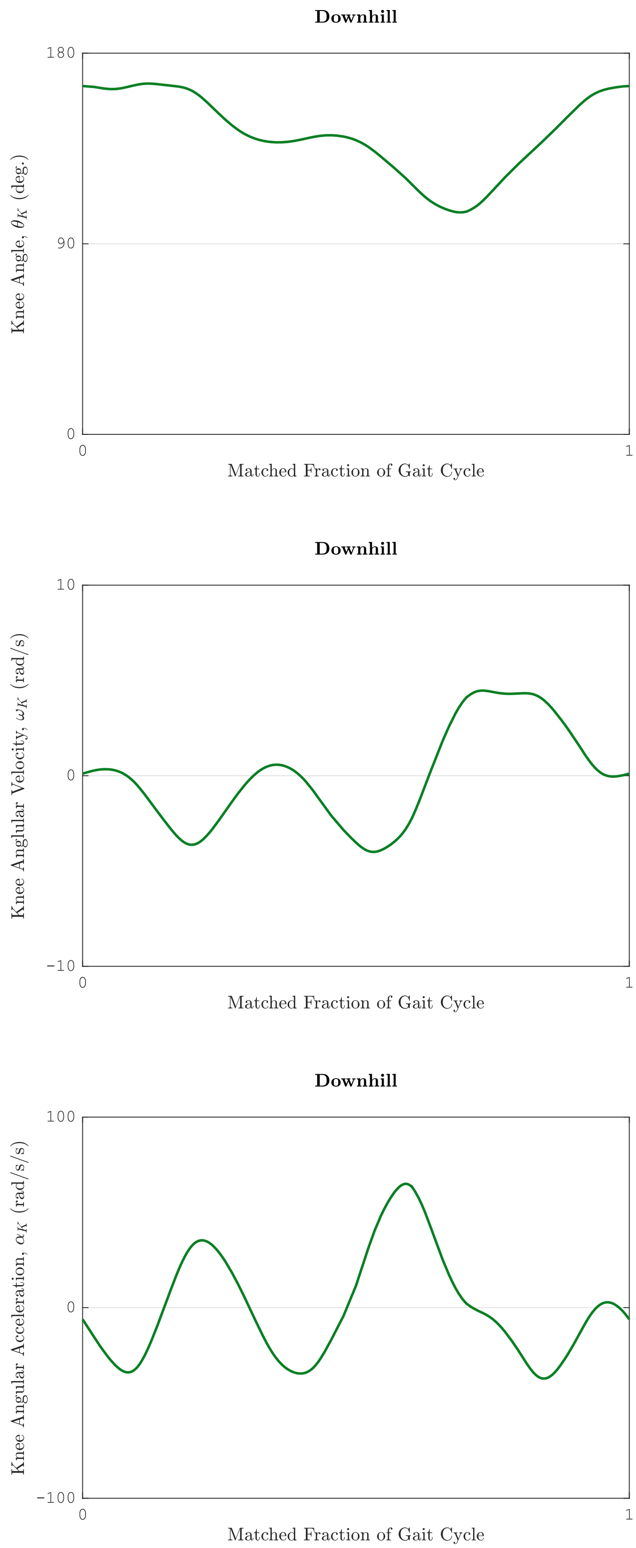


Figure A3 (above three): Walking downhill: knee angle $\theta_K(t)$, angular velocity $\omega_K(t)$, and angular acceleration $\alpha_K(t)$.

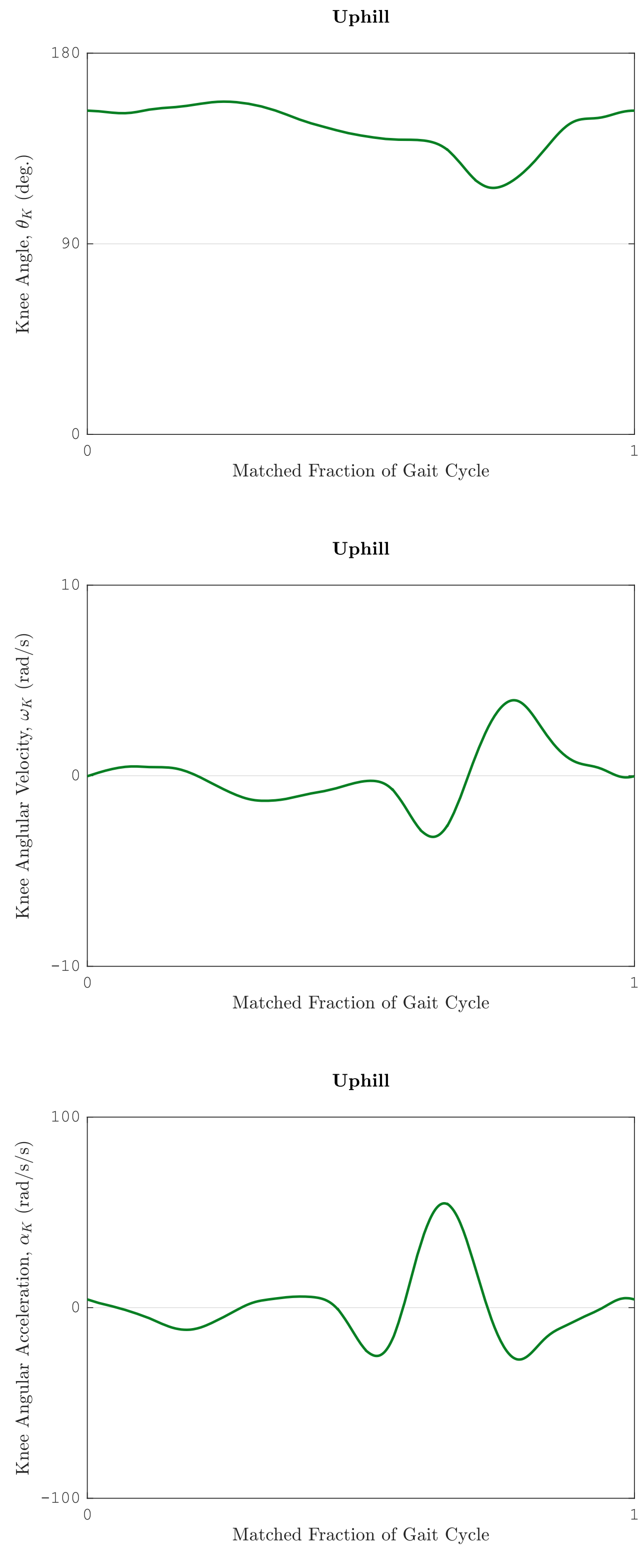


Figure A4 (above three): Walking uphill: knee angle $\theta_K(t)$, angular velocity $\omega_K(t)$, and angular acceleration $\alpha_K(t)$.