DESIGN AND CONTROL OF A LOWER-LIMB EXOSKELETON EMULATOR FOR ACCELERATED DEVELOPMENT OF GAIT EXOSKELETONS

by

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ABSTRACT

Robotic exoskeletons for gait assistance carry the potential to dramatically improve the quality of life for individuals with hemiparesis resulting from a stroke. This thesis presents the design and implementation of a modular lower-limb exoskeleton emulator: a new research platform intended for accelerated research into and development of lower-limb exoskeletons. The device features five lightweight, modular braces for human interface with ten total lower-limb degrees of freedom. Braces are actuated by four off-board motors via Bowden-cable transmission. A closed-loop controller utilizing high-frequency real-time measurements provides accurate and responsive torque application to the wearer. A versatile control software model featuring real-time gait event detection has been developed and verified with preliminary experimentation conducted with the emulator. Future investigations with the device will inform the design of a novel assistive, multi-joint body-powered leg exoskeleton for hemiparetic gait assistance.
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LIST OF SYMBOLS

Motor Winding Temperature .................................................. $T_w$
Ambient Temperature .......................................................... $T_a$
Motor Winding Temperature with respect to Ambient ................. $\Delta T_w$
Motor Winding-Housing Thermal Resistance ............................ $R_{th1}$
Motor Housing-Ambient Thermal Resistance ............................ $R_{th1}$
Motor Winding Electrical Resistance at 25°C .......................... $R_{25}$
Electrical Current Applied to Motor ..................................... $I_m$
Thermal Resistance of Copper ............................................. $\alpha_{Cu}$
Motor Torque Constant ....................................................... $K_m$
Motor Efficiency ............................................................... $\eta_m$
Motor Torque ................................................................. $\tau_m$
Motor Speed ................................................................. $\omega_m$
Motor Speed/Torque Gradient ............................................ $\frac{\Delta \omega}{\Delta \tau}$
Motor Pulley Diameter ....................................................... $\phi p_m$
Motor Angle ................................................................. $\theta_m$
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Integral Gain ................................. $K_i$
Derivative Gain ................................. $K_d$
Sampling Period .............................. $T_s$
Low-Pass Filter Coefficient ................. $N$
Emulated Torsional Spring Stiffness ........... $\kappa$
LIST OF ABBREVIATIONS

Assistive Multi-joint Body-powered Leg Exoskeleton . . . . . . . . . . . . . . . . AMBLE
Central Nervous System ..................................................... CNS
Center of Mass ................................................................. COM
Ankle Foot Orthosis ............................................................. AFO
Range of Motion ............................................................... ROM
Degree of Freedom ............................................................. DOF
Flexion/Extension .............................................................. F/E
Abduction/Adduction ......................................................... A/A
Dorsiflexion/Plantarflexion .................................................... D/P
Inversion/Eversion .............................................................. I/E
Medial/Lateral ................................................................. M/L
Anterior/Posterior ............................................................. A/P
Superior/Inferior ............................................................... S/I
Position-Integral-Derivative .................................................. PID
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Robotic exoskeletons are defined as an anthropomorphic mechanical devices which can be worn by a human user and operate in concert with the user’s movements [1]. To date, robotic exoskeletons have been serving three primary functions in society: performance augmentation of an able-bodied wearer (e.g. increased strength or endurance), rehabilitation of motor deficiencies through repeated movements, and assisting with disabilities (e.g. stroke or spinal cord injury). Recent advances in robotic technology and control have made exoskeletons increasingly viable for an increasing number of military, industrial, and medicinal applications, leading to a surge in development and commercialization of new devices.

1.1 Motivation

This research is motivated by a desire to provide assistance to people with hemiparesis due to a stroke. Stroke is the leading cause of adult disability in the United States, with prevalence expected to continue to grow in the foreseeable future. Stroke often results in hemiparesis, in which motor function is lost or severely limited on one side of the body; recovery from this deficit is the main goal of stroke rehabilitation. One of the most desired specific outcomes of rehabilitation is improved walking ability, which has important implications for functional independence, cardiovascular health, and community reintegration [2].

The importance of dosage and repetition to successful post-stroke motor recovery has driven research and development into exoskeletons specifically targeting gait rehabilitation for stroke survivors. Widespread integration of such devices in stroke rehabilitation clinics and other improvements in rehabilitation science have led to a growing proportion of stroke survivors who are able to regain independent ambulatory function during therapy. However,
post-stroke gait is characteristically asymmetric, with reduced speed, cadence, and stride length [3].

At the optimal stride length, cadence and step width, able-bodied gait has been modeled as an inverted pendulum requiring *external* energy only to transition from step to step [4]. In addition, elastic tissues spanning the ankle perform considerable positive and negative work on the center of mass [5], reducing the *internal* work required to attain the dynamic benefits associated with pendulum motion. Hemiparetic gait, due in part to weakness and poor timing control in the affected leg, does not share the same benefit from inverted pendulum dynamics or ankle energy storage and return [6]. Gait in chronic stroke survivors is typified by higher metabolic cost, slower speed, and increased fall rate; all of which contribute to a significant deficit in walking endurance and community reintegration [2].

Recently, several commercially-available assistive exoskeletons with the ability to improve gait in people with mobility impairments (including hemiparesis) have been developed. However, the high cost of such devices (driven by high-powered motors, and cutting-edge battery technology) limits their accessibility and impact in the general post-stroke population. An affordable, body-powered device capable of providing similar benefits to post-stroke mobility could provide widespread improvements to the quality of life of stroke survivors.

1.2 Purpose

The long-term goal of this research is to inform the design of *AMBLE*: an assistive, multi-joint, body-powered leg exoskeleton to improve hemiparetic gait by providing an external path for energy transfer between legs, allowing storage of mechanical energy from the non-paretic limb and delivery to the impaired limb at advantageous times during the gait cycle.

Many questions need to be answered and multiple parameters need to be identified before such a device can be realized. In order to efficiently address these questions and identify ideal device parameters, a versatile research tool capable of a broad range of exoskeleton functionalities was envisioned to separate the challenges associated with desirable parameter identification from their physical realization.
1.3 Scope of Research

This thesis presents the conceptualization, design, and implementation of a lower-limb exoskeleton emulator for investigation into and accelerated development of gait exoskeletons (Figure 1.1).

Prior to conceptualization, a literature review was conducted on existing lower-limb exoskeleton designs and experimentation, stiffness estimation of lower-limb joints during gait, variable-stiffness actuators, and existing exoskeleton research tools. Following the review, design constraints and specifications were developed for a device capable of real-time torque-generation about sagittal-plane lower-limb joints (hip flexion/extension, knee flexion/extension, ankle dorsiflexion/plantarflexion) during gait.

The exoskeleton incorporates a modular design including 5 braces capable of adjustment to 5th and 95th percentile ranges of relevant anthropometric dimensions. Prototypes were fabricated and assembled in-house and used to assess and revise the mechanical design. Motor specifications were developed to maximize torque bandwidth and response, and the torque generation capabilities of the selected motors were enhanced by implementation of custom heat sinks. An actuation cart was designed and fabricated to house all actuation, instrumentation, and data acquisition hardware. Dynamics of the mechanical transmission were characterized and used to calibrate a real-time position-based feedback controller for closed-loop torque control of the actuated brace joints.

To inform the feasibility of AMBLE, a study was designed to investigate the amount of feasible mechanical energy storage during the swing phase of able-bodied gait. A control model was developed to emulate clutched torsional springs of varying stiffness to resist knee extension during the swing phase of gait. Pilot data was collected for 6 discrete spring stiffness values at two gait velocities with the emulator generating torques proportional to knee flexion angle.
Figure 1.1: Lower-Limb Exoskeleton Emulator
1.4 Outline

The remainder of this thesis is organized as follows: Chapter 2 presents a discussion of background topics, including:

- Stroke and the importance of functional gait recovery
- Mechanics and energetics of gait (able-bodied and hemiparetic)
- Exoskeletons for gait assistance

The mechanical design of the emulator braces is detailed in Chapter 3. Actuation details are outlined in Chapter 4. Chapter 5 describes the design and implementation of the position-based torque controller, and a preliminary experiment conducted with the emulator is presented in Chapter 6. Chapter 7 concludes this thesis with a summary of the work presented in this document and recommendations for future work.
CHAPTER 2
BACKGROUND

2.1 Stroke

Stroke affects nearly 800,000 people annually in the United States, making it the leading cause of long-term adult disability [7]. Direct and indirect cost of stroke has been estimated to be over $100 billion [7]. Due in part to an aging population, stroke prevalence and related expenses are both projected to grow in the future, placing a large financial strain on national health care [8].

A stroke occurs when the blood flow to an area of the brain is obstructed, resulting in the death of brain cells. As a result, abilities controlled by that area of the brain (such as motor control) are lost or severely damaged. Over 70% of stroke victims experience hemiparesis, in which motor function is lost or severely limited on one side [8]. The debilitating effects of stroke, however, are mitigated by neural plasticity. Neural plasticity - a property of the central nervous system (CNS) allowing it to rebuild neural connections - is the basic mechanism for motor recovery following stroke [9], but gains in functional recovery typically reach a plateau after 3 to 4 months [10]. Physical rehabilitation is therefore most effective during this time period in order to to maximize neuroplastic functional improvement [11].

Successful post-stroke rehabilitation depends on several principles, including patient engagement and motivation, dosage, repetition, and task and context-specific training, making it a labor-intensive process [9][12][13]. In recent decades, rehabilitation exoskeletons and robotics have grown substantially in prevalence thanks to their ability to provide highly repeatable, high-dosage training [9][14]. Gait recovery is one of the most frequently-stated goals of post-stroke individuals, and more rehabilitation time is spent on gait recovery than all other activities [2]. Thanks to years of existing and ongoing research in gait rehabilitation, the majority of initially non-ambulatory stroke patients regain the ability to walk unassisted.
However, gait patterns typical to chronic hemiparetic stroke victims preclude the dynamic and energetic advantages enjoyed by able-bodied gait. These limitations lead to a higher fall rate, fear of falling, and a decrease in community ambulation among chronic stroke patients [18].

2.2 Mechanics and Energetics of Gait

2.2.1 Able-Bodied Gait

Human walking has been optimized by millions of years of evolution. Studies show that self-selected walking pace (1.2 - 1.4 m s$^{-1}$), when compared to speeds both faster and slower, correlates with a minimum in metabolic cost of transport [6][19]. Additionally, while joint moment patterns for the hip and knee during gait vary from person to person, the resulting ground reaction forces and kinematic patterns are largely consistent [20]. This phenomenon has been partially explained by a simple model representing the swing phase of gait as a pendulum, and the single-leg stance phase as a rigid inverted pendulum [21]. Pendulum models, although controversial among biomechanists due to their inability to describe/account for important nuances of gait (e.g. stability, postural sway, internal work requirements, roles of individual muscles, etc.), are capable of coarsely reproducing the energy and movement patterns of human walking [22] and have motivated a substantial amount of gait research in the literature. Within such a framework, the core energy cost of walking is associated with step-to-step transitions from one inverted pendulum arc to the next (Figure 2.1) [4].

Dynamic consistency of the inverted pendulum model with human gait has been dramatically improved by the addition of linear elasticity to the legs, allowing a spring-like storage and return of mechanical energy to the center of mass (COM) during stance [25]. Further examination into elastic energy storage and return in gait has led to the discovery that passive elastic elements spanning lower limb joints contribute a substantial amount of work during gait, reducing the metabolic energy load on lower-limb musculature [23]. This effect is especially pronounced in the Achilles tendon, which aids positive push-off work gen-
Figure 2.1: Inverted Pendulum Model of Gait

Between steps, center-of-mass (COM) velocity must be redirected from a downward trajectory (terminating one pendular arc) into an upward trajectory (initiating a new arc). This is accomplished by negative collision work performed by the leading leg at heel strike [23][24]. In order to maintain a constant velocity, energy lost in this collision must be replaced with positive work performed on the center of mass. This work (push-off of the trailing leg) is ideally performed during the same transition. (Adapted from from [4])

Importantly, however, this benefit is only realized when push-off occurs immediately before contralateral heel strike [26][27]. Improper temporal coordination of these two events can increase energy cost of COM redirection by up to 400% [24][26].

2.2.2 Hemiparetic Gait

Following stroke, motor function is often severely limited on one side of the body. Deterioration in strength and control of muscle contraction timing and magnitude are present immediately following stroke; after a few weeks, spasticity and other mechanical changes to muscle tissue may also develop [28]. The combination of these symptoms may hinder or even eliminate the ability of the stroke survivor to walk. Although modern rehabilitation has a high (nearly 80%) and growing rate of success in restoring independent gait function [10], post-stroke gait is characteristically asymmetrical [29], and can be generally described with
several spatio-temporal, kinematic, and kinetic deviations from able-bodied gait, including:

- Decreased gait velocity [3][6][30][31][32]
- Decreased stride length and cadence [3][31]
- Increased double support duration [3][30]
- Increased stance duration for affected limb [30][33]
- Reduced affected hip range of motion [31][32]
- Reduced affected knee flexion at toe-off and mid-swing [32][33]
- Reduced affected knee extension at heel strike [34][31]
- Reduced affected ankle plantar-flexion at toe-off [31][32]
- Circumduction of affected limb [32][33]
- Reduced peak knee extension and ankle plantarflexion moments [34]
- Decreased affected limb medial ground reaction force prior to single support [35]
- Increased vertical ground reaction force variability [35]
- Increased metabolic cost of gait [6][29][36][37]

Figure 2.2 shows the range of kinematic and energetic variability among a group of 30 ambulatory post-stroke subjects [3], and helps to illustrate one of the challenges in addressing deficits in hemiparetic gait: severity and nature of motor impairment are unique to each individual, making generally-applicable, specific interventions difficult to pinpoint. However, literature has identified a number of common problem areas.

Ankle push-off, as discussed above, is a critical component in energetically efficient gait. However, sufficient ankle plantarflexion strength and the precise coordination in timing necessary to properly execute push-off are commonly lost after stroke. A clear deficit in hemiparetic positive ankle power timing and amplitude during push-off (regardless of self-selected
Figure 2.2: Hemiparetic Gait: Joint Kinematics and Kinetics
Mean profiles of sagittal-plane (a) joint kinematics and (b) joint powers during gait for slow (S) (0.25 ± 0.05 m/s), medium (M) (0.41 ± 0.08 m/s), and fast (F) (0.63 ± 0.08 m/s) hemiparetic walkers as well as unimpaired subjects (N) walking at a slow speed. (Adapted from [34])
walking speed) can be observed in Figure 2.2, and hemiparetic push-off has been generally accepted as an important target to improve post-stroke gait [2][3][6][10][23].

In comparison to able-bodied gait, hemiparetic leg swing is typically shortened, accompanied by a decrease in stride length (a direct correlate of gait velocity in speeds over 0.33 m s\(^{-1}\) [38]). Reductions in early- and mid-swing knee flexion obstruct toe clearance during this phase, necessitating compensatory measures such as hip circumduction and pelvic hike [33], the combination of which may contribute to the substantially increased metabolic cost of hemiparetic gait. While intuition may attribute this reduction in knee flexion to commonly-observed post-stroke symptoms such as spasticity in knee extensor muscles [31] or deficient knee flexor strength [28], recent studies have cited the lack of adequate pre-swing propulsive work from the hip flexors as a primary cause of insufficient knee flexion during swing [33][39].

Passive, spring-like ankle-foot orthoses (AFOs) are commonly prescribed to assist with hemiparetic gait [40][41]. AFOs help to ensure toe clearance by preventing excessive ankle plantarflexion during swing. Furthermore, their elasticity allows them to store mechanical energy at the ankle joint during stance, naturally returning it in pre-swing to assist ankle push-off [42]. However, the level of dorsiflexion required for toe clearance during gait (enforced by a hard stop) constrains the plantarflexion range of motion (ROM) during stance, thereby limiting positive mechanical work capacity at the ankle. Theoretically, if toe clearance could reliably be achieved without such a kinematic constraint on ankle dorsiflexion (e.g., by ensuring proper knee flexion in early- and mid-swing), ankle push-off power could be further improved.

### 2.3 Exoskeletons for Gait Assistance

Exoskeletons and rehabilitation robots have been widely implemented in post-stroke gait rehabilitation [9][13][43]. In addition to reducing physical burden on therapists, rehabilitation robots have been found to increase duration, consistency, and frequency of rehabilitation sessions [14][43]. Additionally, they have the ability to provide real-time quantitative functional assessment measures [14]. Despite these tangible benefits however, no consistent
improvement in quality of training outcomes with rehabilitation robotics has been observed when compared with traditional therapy protocols [44].

Advances in control algorithms and actuator technology continue to drive improvements in human intention detection, device transparency and bandwidth, and a number of powered exoskeletons have been developed with the goal of gait assistance in impaired populations (Table 2.1). However, prevalence of assistive exoskeletons in the post-rehabilitation chronic stroke population is currently limited. The complexity of state-of-the-art exoskeleton designs and control systems often requires expert knowledge and/or supervision for proper operation. Device weight, driven by heavy motors, can necessitate crutches for additional support and limit battery life (and therefore portability). Finally, high prices constrain access to these devices to a minority of the stroke-impaired population.

While numbers grow and quality of rehabilitation exoskeletons continues to improve, there is currently little innovation targeting portable, passive assistive devices for the chronic stroke patient [53]. Canes, walkers, and passive orthoses are common among chronic stroke survivors, but offer no improvement to hemiparetic gait mechanics and energetics [54].

Recently, the field of body-powered (quasi-passive) exoskeletons has generated renewed research interest [55][56][57][58][59], and proven capable of improving able-bodied gait energetics [60]. Body-powered devices, while lacking an external power source and high-fidelity

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<td>6</td>
<td>15</td>
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<td>14k - 29k</td>
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<td>12</td>
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</table>
control, are advantageous in their simplicity, low cost and portability, and have proven capable of improving energetics of able-bodied gait [60]. The traditional mechanism behind body-powered exoskeletons is controlled mechanical energy storage and return, typically accomplished by springs. Rather than injecting energy into the gait cycle (as is done with powered exoskeletons), body-powered exoskeletons provide an effective anatomical augmentation, allowing CNS to find a new optimal coordination pattern [60]. Developments in lightweight, low-power clutches [61][62][63] and variable stiffness springs [64][65] offer increasing control authority without sacrificing device autonomy or mobility, opening new possibilities for future design of body-powered exoskeletons.

The design of body-powered exoskeletons presents a substantial practical challenge for a number of reasons. For instance, spring stiffness should be selected based on human joint mechanical impedance (resistance to perturbation), a parameter which humans are known to modulate during gait [66]. Variable stiffness springs provide a means to track human joint stiffness, but quantitative knowledge of human stiffness modulation in lower-limb joints only exists for the ankle. Today, gait exoskeletons are designed and actuated typically under the assumption of a normal or nominal kinematic pattern. Under this assumption, actuator parameters (e.g. spring stiffness) can be informed by the quasi-stiffness of lower-limb joints during normal gait in order to replace existing function (such as energy storage and return in the Achilles tendon) at a discounted metabolic cost. The validity of such an assumption is limited, however, as humans are known to adapt muscle recruitment strategy based on a re-calibration of CNS’s internal representation of that body segment in the presence of a modification to body segment mechanical properties (e.g. mass, inertia, stiffness) [67]. Gait exoskeletons, however, present much more than a static modification to segment mass and inertia; interaction with such a device constitutes a continuous dynamic perturbation to gait energetics, mechanics, and kinematics. While the nature of human adaptation has been studied for over 30 years, accurate prediction of adaptation to interactions with a gait exoskeleton (regardless of impairment level) presents a substantial challenge in the
field of biomechanics [68][69][70][71]. Experimental human studies are therefore critical in the evaluation of device parameter combinations, but are limited by the requirement of a physical device possessing the parameters to be evaluated. For this reason, a haptic device capable of displaying a wide range of physical parameters to its user (emulator) is an ideal research tool for the development of new wearable robots [70][72][73].
CHAPTER 3
MECHANICAL DESIGN OF EMULATOR BRACES

This chapter will outline the process by which the exoskeleton emulator braces (human attachment interfaces allowing controlled torque application to joints) were designed and implemented.

3.1 Desired Design Specifications

Formulation of design requirements and specifications for an exoskeleton emulator revolves around maximizing the range of exoskeleton parameters that can be displayed to the user [70]. In the design of any haptic device, one critical design requirement is device transparency (a measure that quantifies the performance of the device by comparing the desired environment to be displayed to the actual environment displayed), with device/manipulator dynamics completely hidden from the user in an ideally transparent haptic interaction [74]. In many haptic devices, compensation algorithms are required to artificially “hide” device mass/inertia from the user. For an emulator (in which actuation/instrumentation hardware can be housed elsewhere), a more practical approach to maximizing transparency is to simply minimize manipulator mass and inertia. This allows actuator forces/torques to be designated primarily to the generation of haptic effects, rather than compensation for undesirable device dynamics.

Braces for human interface were designated to be as lightweight as possible while maintaining structural rigidity, and to have sufficient adjustability so as to minimize kinematic constraints experienced by a wide range of body types/sizes while maintaining a tight fit with the wearer. 10 total rotational degrees of freedom were selected based on a literature review of existing gait exoskeletons, in order for the braces to passively track the wearer’s movements during gait. Human attachment points for the braces needed to conform to soft
tissue of the wearer’s torso, thigh, and shank while maintaining sufficient structural rigidity to allow torque transmission from the brace to the wearer.

Adjustment range specifications for the braces were determined by 5\textsuperscript{th} and 95\textsuperscript{th} percentiles of various published anthropometric measurements [75][76][77] where available. For dimensions not found in published anthropometric data, measurements were taken from members of the research laboratory, with adjustment ranges extrapolated from the ranges found in literature.

3.2 Design Features

In order to maximize the potential combinations of actuated lower-limb joints while maintaining minimal device mass and inertia, a modular design was chosen for the braces. An additional benefit to this choice is the elimination of the requirements for segment-length adjustability and multi-joint static alignment. For instance, a design with multiple rigidly-connected lower-limb joints must allow sufficient adjustability to account for both segment length and simultaneous joint axis alignment. Braces for hip, knee, and ankle joints are modular, stand-alone units and can be worn either individually or together. In total, five DOF are included per leg (Table 3.1), with ranges of motion chosen to match their anatomical human counterparts. The most notable DOF omission was hip internal/external rotation.

<table>
<thead>
<tr>
<th>Brace</th>
<th>Degree of Freedom</th>
<th>Range of Motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td>Flexion</td>
<td>120°</td>
</tr>
<tr>
<td></td>
<td>Extension</td>
<td>15°</td>
</tr>
<tr>
<td></td>
<td>Abduction</td>
<td>40°</td>
</tr>
<tr>
<td></td>
<td>Adduction</td>
<td>30°</td>
</tr>
<tr>
<td>Knee</td>
<td>Flexion</td>
<td>135°</td>
</tr>
<tr>
<td></td>
<td>Extension</td>
<td>0°</td>
</tr>
<tr>
<td>Ankle</td>
<td>Dorsiflexion</td>
<td>50°</td>
</tr>
<tr>
<td></td>
<td>Plantarflexion</td>
<td>20°</td>
</tr>
<tr>
<td></td>
<td>Inversion</td>
<td>15°</td>
</tr>
<tr>
<td></td>
<td>Eversion</td>
<td>35°</td>
</tr>
</tbody>
</table>
This decision was made for three practical reasons:

1. Aligning an external center of rotation with the long axis of the femur would require a substantial amount of added mass and complexity

2. Range of motion for hip internal/external rotation is small (∼10°) during able-bodied gait [78]

3. Soft tissue at the relevant human/device interface (thigh) will likely allow sufficient range of motion for this DOF during able-bodied gait

To meet the specifications regarding rigidity and mass of the braces, all structural components of the braces were manufactured from 6061 aluminum alloy due to its low cost and high strength-to-weight ratio. Specific design features will be described in the following sections.

3.2.1 Degrees of Freedom

The braces contain 10 total functional joint axes, four passive and six powered - each corresponding to a rotational DOF listed in Table 3.1. For both types, two lubricated radial bearings are housed in proximal plates and support a single 1/4″ stainless steel shaft on both sides of a single distally-mounted plate allowing low-friction rotation (Figure 3.1). Powered axes feature pulleys for torque transmission (described in more detail in Chapter 4) and rotary encoders to measure joint angle. Pulleys are fixed with respect to the distal segment; rotary encoders are fixed with respect to the proximal segment. Mechanical “hard stops” are implemented for all joints to prevent the device from applying torque outside the wearer’s range of motion.
3.2.2 Human Interface

Three mechanisms are used to interface the exoskeleton braces with the wearer. Aluminum brackets (Figure 3.2(a)) lined with 1" memory foam and secured with nylon cinching straps are used to fix the hip brace to the torso. Flexible cuffs comprising thin-wall PVC pipe, medical foam, hook-and-loop straps (Figure 3.2(b)) form the attachment points to the wearer's thigh and shank for all three brace types. Foot attachment is accomplished with modified commercially-available athletic sandals (Figure 3.2(c)). $\frac{3}{16}$" stainless steel rods penetrate the sole of the sandals and are rigidly connected on either side by an aluminum bracket.
Figure 3.2: Human Interface Mechanisms

(a) Torso Bracket

(b) Leg Cuffs

(c) Foot Attachment
3.2.3 Adjustment Points

Segment links of the exoskeleton braces are composed of $\varnothing 0.54''$ standard wall aluminum pipe. Slotted clamp-style fittings with clearance holes sized for a slip fit with the pipe allow for quick adjustment of segment lengths (Figure 3.2(a)). At the desired position, the fitting clamp can be tightened to rigidly connect the two parts. Because the circular cross-section allows relative rotation between the parts during adjustment, engravings on both pipe and fitting were added to help ensure proper alignment after length adjustment.

3.3 First Generation Emulator Braces

This section will describe the first-generation implementation of individual hip, knee, and ankle braces for the exoskeleton emulator.

3.3.1 Bilateral Hip Brace

![Figure 3.3: Bilateral Hip Brace](image-url)
The bilateral hip brace (Figure 3.3) has a mass of 2.33 kg and four functional joints: two actuated flexion/extension (F/E) and two passive abduction/adduction (A/A). It is secured to the wearer's torso with two torso brackets, which are adjustable to translate/rotate along/about their medial/lateral (M/L) horizontal segments. The horizontal segments can be adjusted individually along the length of the lumbar segment, which terminates at the posterior T-joint. From the T-joint, A/A joint locations can be adjusted along their proximal segments so that the axes pass through the functional hip joint center of the wearer. F/E joint axes are located downstream of the A/A joints, and can be individually adjusted in the anterior/posterior (A/P) direction to intersect the functional hip joint, and along the F/E joint axes to ensure hip clearance for the wearer. Finally, distal F/E segments are secured to the wearer's thigh with two attachment cuffs per leg. Orthogonality and intersection of A/A and F/E axes are ensured by alignment engravings on relevant pipe segments and slotted fittings. In total, the hip brace has 23 adjustment points for a completely customizable fit. Notable dimensions are specified in Figure 3.6 and Table 3.2.

3.3.2 Knee Braces

Each of the two knee braces (Figure 3.4) has a mass of 0.54 kg and a single actuated degree of freedom corresponding to F/E. The brace is secured to the wearer's thigh and shank with two attachment cuffs per segment. Distal cuffs for each segment have additional medical foam padding in order to better conform to the shape of the thigh and shank (an example is shown in Figure 3.2(b)). Distal segments can also be adjusted in the M/L direction if needed to ensure a concentric fit.
3.3.3 Ankle Braces

Each ankle brace (Figure 3.5) has a mass of 1.05 kg and two degrees of freedom: actuated dorsi/plantarflexion (D/P) and passive inversion/eversion (I/E). Three hook-and-loop straps on the athletic sandal provide a highly-adjustable and secure fit to the wearer’s foot, and the height of the I/E joint is adjusted near the heel of the foot. The D/P joint can be adjusted in both A/P and M/L directions with respect to the I/E joint in order to best align with the wearer’s ankle joint. I/E and D/P axes are, by design, perpendicular and intersecting (similar to A/A and F/E joints of the hip), and attachment to the shank is accomplished with two leg attachment cuffs. Ankle braces have a total of 10 adjustment points, with segment lengths detailed in Figure 3.6 and Table 3.2.
3.3.4 Brace Summary

All 181 structural brace components were manufactured in-house by either manual mill, lathe, or CNC. For the first implementation, no comprehensive strength analysis was performed for further mass reduction - a step that is recommended for improved future brace designs. Figure 3.6 and Table 3.2 show the minimum and maximum segment lengths for each brace.
### Figure 3.6: Emulator Brace Dimensions

### Table 3.2: Emulator Brace Dimensions

<table>
<thead>
<tr>
<th>Brace</th>
<th>Dimension</th>
<th>Min. Length (cm)</th>
<th>Max. Length (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td>A</td>
<td>19.7</td>
<td>35.6</td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>2.7</td>
<td>26.8</td>
</tr>
<tr>
<td></td>
<td>C</td>
<td>16.2</td>
<td>20.0</td>
</tr>
<tr>
<td></td>
<td>D</td>
<td>28.5</td>
<td>45.8</td>
</tr>
<tr>
<td></td>
<td>E</td>
<td>5.9</td>
<td>18.9</td>
</tr>
<tr>
<td></td>
<td>F</td>
<td>5.6</td>
<td>28.0</td>
</tr>
<tr>
<td>Knee</td>
<td>G</td>
<td>5.9</td>
<td>24.0</td>
</tr>
<tr>
<td></td>
<td>H</td>
<td>5.8</td>
<td>23.1</td>
</tr>
<tr>
<td>Ankle</td>
<td>I</td>
<td>7.8</td>
<td>25.9</td>
</tr>
<tr>
<td></td>
<td>J</td>
<td>6.6</td>
<td>13.7</td>
</tr>
<tr>
<td></td>
<td>K</td>
<td>5.0</td>
<td>12.1</td>
</tr>
<tr>
<td></td>
<td>L</td>
<td>9.4</td>
<td>12.1</td>
</tr>
</tbody>
</table>
CHAPTER 4
ACTUATION AND TORQUE TRANSMISSION

As discussed in the previous chapter, high-performance haptic interfaces require a high level of device transparency. For the emulator, transparency is primarily accomplished by the implementation of lightweight, low-inertia braces with sufficient adjustment to provide minimal restriction to the movements of the wearer. However, while a sufficient level transparency is necessary, the core determinants of performance in a haptic device are its specific actuation and control characteristics.

In order to study human-device interactions during gait resulting from a broad range of theoretical lower-limb exoskeleton designs, the emulator must be capable of accurately generating torque magnitudes similar to those experienced in unaided human gait. Perhaps more important, however, is the responsiveness of the system – its ability to rapidly change the magnitude of applied torques. System responsiveness would be theoretically maximized with motors rigidly connected to the braces - however, such an arrangement would add considerable mass to the braces themselves. Instead, a Bowden-cable transmission was implemented, similar to those described in [70] and [73], allowing torque to be transmitted from motor to brace via tensile forces conducted by a lightweight, flexible tether. Pulleys at the motor and brace convert motor torque to cable tension, and cable tension to brace torque, respectively.

The remainder of this chapter will describe motor selection, implementation, and performance optimization via modification of thermal (heat dissipation) properties, as well as details of the mechanical transmission and torque-generation capabilities of the exoskeleton emulator.

4.1 Motor Selection

Published values for peak torques and angular velocities experienced at hip, knee, and ankle in the sagittal plane during able-bodied gait (Table 4.1) guided motor selection criteria.
Table 4.1: Peak Sagittal-Plane Joint Torques & Angular Velocities During Able-Bodied Gait

<table>
<thead>
<tr>
<th>Joint Flexion</th>
<th>Peak Angular Velocity (rad s$^{-1}$) [79]$^a$</th>
<th>Peak Joint Torque (N m) [80]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip Flexion</td>
<td>2.8</td>
<td>75.0</td>
</tr>
<tr>
<td>Hip Extension</td>
<td>1.7</td>
<td>52.2</td>
</tr>
<tr>
<td>Knee Flexion</td>
<td>4.5</td>
<td>38.4</td>
</tr>
<tr>
<td>Knee Extension</td>
<td>5.2</td>
<td>45.0</td>
</tr>
<tr>
<td>Ankle Dorsiflexion</td>
<td>2.0</td>
<td>n/a</td>
</tr>
<tr>
<td>Ankle Plantarflexion</td>
<td>2.9</td>
<td>132.6</td>
</tr>
</tbody>
</table>

$^a$Angular velocities reported in [79] were normalized with respect to gait velocity, and reported in units of degrees/% gait cycle. As stride times were not explicitly reported an assumed stride time of 1.2 s was used to calculate the values reported here.

Motor specifications were set with a benchmark of 200 N nominal cable tension (as reported in [73]) at a minimum joint angular velocity of 5 rad s$^{-1}$. Calculations were performed based on these benchmarks assuming a brace pulley radius of 3 cm, and a candidate list of capable motors was compiled. To further narrow the selection, a ratio of motor torque to inertia was calculated for each motor and used as an additional consideration in order to maximize responsiveness. Finally, practical considerations such as availability (lead time) and price narrowed the selection to four 200 W $\varnothing$ 30 mm brushless DC motor assemblies with 74:1 planetary gearheads and 500-CPT rotary encoders (30515 with 203123 and 110514, Maxon Motor, Sachseln, Switzerland). While constraints on pulley size at the emulator braces limit moment arms to $\sim$6 cm (12 N m torque at 200 N cable tension), it was anticipated that the nominal current of the motors could be increased with the addition of heat sinks to the motor housing. In addition, intermittent operation above nominal current was deemed likely during the study of human gait due to its cyclical nature.

4.2 Motor Implementation

Motors are mounted on an off-board actuation cart, for which additional detail will be provided in Chapter 5. Each motor assembly (Figure 4.1) is mounted in a vertical orientation with adjustable position to accommodate various brace actuation configurations.
Heat sinks for the motors were implemented to improve the effective thermal conductivity of the motor housing and increase the nominal motor current (see Section 4.3). After a failed exhaustive search for commercially-available φ30 mm heat sinks, custom heat-sinks were designed and manufactured in-house. Fin thickness and spacing were determined so as to maximize effective heat transfer along the 6.6 cm length of exposed motor housing. The final design contains a total of 32 fins per motor with $\frac{1}{32}$" thickness, 4.38" surface area, and $\frac{1}{8}$" spacing.
The heat sinks also serve a key structural role in securing the motor assemblies to the actuation cart, and provide mounting accommodations for miniature temperature sensors (LM35, Texas Instruments, Dallas, TX) to monitor the housing temperature of each motor. Heat sinks were manufactured from 6061 aluminum alloy because of its low cost, ease of manufacturability, and high thermal conductivity.

4.3 Thermal Characterization

The maximum continuous current tolerable by the motors is determined by the manufacturer based on the maximum tolerable motor winding temperature. When a current is applied to the motor, the electrical resistance of the motor windings causes a thermal power loss dependent on the current amplitude, which heats the motor. This heat must be dissipated by the winding surfaces and motor housing, for which thermal resistance values are given by the manufacturer, and the relationship between applied motor current and motor winding temperature is given by:

$$\Delta T_w = T_w - T_a = \frac{(R_{th1} + R_{th2}) \cdot R_{25} \cdot I_m^2}{1 - \alpha_{Cu} \cdot (R_{th1} + R_{th2}) \cdot R_{25} \cdot I_m^2}$$  \hspace{1cm} (4.1)

Where $\Delta T_w$ is the increase of the winding temperature, $T_w$, with regard to the ambient temperature $T_a$; $R_{th1}$ and $R_{th2}$ are respectively the housing-ambient and winding-housing thermal resistance; $R_{25}$ is the electrical resistance of the motor windings at 25°C; $I_m$ is the current applied to the motor; and $\alpha_{Cu}$ is the thermal resistance coefficient for the copper motor windings. All manufacturer specifications can be found in Appendix A.

The desired effect of adding heat sinks to the motor housing is to decrease $R_{th2}$, thereby decreasing the winding temperature realized for any given current. Modified housing-ambient thermal resistance, $R_{th2,mod}$ was determined experimentally by applying a constant current to two motors: one with a heat sink and one bare. Ambient temperature as well as the temperature of each motor housing was recorded for 100 minutes so that temperatures reached steady-state (thermal time constant ~ 20 minutes). Trials were conducted with motor currents of 1 A, 2 A, and 3 A, data for which are shown in Figure 4.2.
Ambient and motor housing temperature were collected for motors with and without heatsink with constant applied currents of 1 A, 2 A, and 3 A.

Steady-state temperatures were substituted into Equation 4.1 for $T_w$ in order to experimentally determine $R_{th2,mod}$ for each trial. Values for $R_{th2,mod}$ from each trial were averaged and again substituted into a rearrangement of Equation 4.1 using the maximum winding temperature reported by the manufacturer to solve for the modified nominal motor current:

$$I_{\text{max}} = \sqrt{\frac{\Delta T_{w,\text{max}}}{R_{25} \cdot (\alpha_{Cu} \Delta T_{w,\text{max}} + 1) \cdot (R_{th1} + R_{th2,mod})}}$$ (4.2)
As shown in Table 4.2, the addition of heat sinks to the motors resulted in a substantial (~80%) reduction in housing-ambient thermal resistance, and (~117%) increase in nominal current rating of the motors compared to the values reported by the manufacturer.

<table>
<thead>
<tr>
<th></th>
<th>( R_{th2} ) (K W(^{-1}))</th>
<th>( I_{max} ) (A)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Manufacturer Spec Sheet</td>
<td>7.4</td>
<td>5.40(^a)</td>
</tr>
<tr>
<td>Experimental - With Heat Sink</td>
<td>1.5</td>
<td>11.70</td>
</tr>
</tbody>
</table>

Changes in motor housing-ambient thermal resistance \((R_{th2})\) and nominal motor current \((I_{max})\) as a result of heat sink implementation.

\(^a\)The nominal current directly reported by the manufacturer (3.68 A) is assumed to correspond to the reported nominal motor speed of 16000 rpm. The value reported in this table for the manufacturer value was calculated using Equation 4.2 with the reported value for housing-ambient thermal resistance, and appears to agree with the graphic depicting maximum continuous current at 0 rpm (Appendix A).

### 4.4 Mechanical Transmission

Torque is transmitted from the motor to the emulator brace using a Bowden-cable configuration (Figure 4.3). Torques generated by the motor (A) are converted by the \( \varnothing 96 \text{ mm} \) motor pulley (B) to tensile forces in the 0.9 mm synthetic cable (Braided Vectran\(^\circledR\)200, TwinLine, Boulder, CO) (C). The proximal Bowden sheath termination block (D) is rigidly mounted to the actuation cart (along with the motor assembly) and is the proximal support for the \(~3 \text{ m}\) transmission tether (E), containing a lined Bowden conduit (415187-01, Lexco Cable Mfg., Norridge, IL) and brace sensor wiring. The far end of the tether is secured to the distal Bowden sheath termination block (F), which is mounted to the proximal segment of the actuated emulator brace (G). Tensile forces in the cable are transmitted through the brace pulley (H), resulting in a torque about the powered axis of the brace. Brace angle and cable tension are measured with 1250-CPR rotary encoders (E2-1250-250-IE-H-D-3, US Digital, Vancouver, WA) (I) and miniature S-beam load cells (LSB200, FUTEK, Irvine, CA) (J), respectively.
While each transmission assembly can provide only unidirectional actuation (i.e. knee flexion OR extension, but not both), combinations of actuated joints/directions are easily re-configured. Modular design of the transmission attachment allows rapid and straightforward installation to and removal from individual emulator braces:
• Distal Bowden sheath termination blocks are permanently affixed to each transmission tether and mount to blocks on the proximal ends of emulator braces with two easily-accessible hex-head machine screws. For each actuated brace, two sets of tapped holes in the attachment block align the Bowden termination block so that the cable is tangent to the brace pulley.

• Rotary encoders (6 total) are permanently installed on functional axes of each emulator brace. Encoders feature easily-accessible 5-pin single-ended male connectors; cabling within the transmission assembly terminates with the mating female connector to power the encoder and relay the data back to the cart.

• Load cells (4 total), unlike the rotary encoders, remain with their respective transmission assemblies and are connected to the end of the synthetic cable leading back to the motor. At each brace, short synthetic cable lengths are permanently affixed to brace pulleys, terminated with threaded eye-bolts to easily connect to the distal end of the load cell.

In order to safely explore the torque generation capabilities of the emulator, the modified nominal current rating $I_{\text{max}} = 11.70$ A (resulting from the heat sink implementation) was multiplied by the reported motor torque constant $K_m = 27.6 \text{ mN m A}^{-1}$ and maximum efficiency $\eta_m = 0.90$, resulting in a modified nominal motor torque of $\tau_m = 291 \text{ mN m}$ (compared to 92.9 mN m reported by the manufacturer at 3.68 A) at nominal motor speed $\omega_{m,\text{nom}} = 16000$ rpm. A gear reduction of 74:1 at 70% efficiency through the planetary gearhead yields 15.1 N m at the gearhead output resulting in a cable tension of 314 N at nominal speed. Motor operation at speeds below nominal will yield proportionally increasing torques and tensions due to the speed/torque gradient of the motor, $\frac{\Delta \omega}{\Delta \tau} = 4.83 \text{ rpm/mNm}$. However, operation at nominal motor currents below nominal motor speeds would place the gearhead at risk of mechanical failure, as it is only rated to 15.0 N m continuous and 22.5 N m intermittent (see Appendix A). For this reason, peak torque generation capability of the
emulator (Table 4.3) was calculated directly from the gearhead rating, and not explicitly
tested.

Table 4.3: Exoskeleton Emulator Transmission and Torque-Generation Specifications

<table>
<thead>
<tr>
<th>Joint</th>
<th>$\varphi_m$ (mm)</th>
<th>$\varphi_b$ (mm)</th>
<th>$\tau_{b,\text{cont}}$ (N m)</th>
<th>$\tau_{b,\text{int}}$ (N m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip F/E</td>
<td>96</td>
<td>83.3</td>
<td>10.4</td>
<td>15.6</td>
</tr>
<tr>
<td>Knee F/E</td>
<td></td>
<td>83.3</td>
<td>10.4</td>
<td>15.6</td>
</tr>
<tr>
<td>Ankle D/P</td>
<td></td>
<td>121.4</td>
<td>15.1</td>
<td>22.8</td>
</tr>
</tbody>
</table>

Motor pulley diameter ($\varphi_m$), brace pulley diameter ($\varphi_b$), maximum continuous brace
torque ($\tau_{b,\text{cont}}$), and maximum intermittent brace torque ($\tau_{b,\text{int}}$) for actuated hip, knee, and
ankle joints of exoskeleton emulator. All reported torque values assume a 20% transmission
loss.
CHAPTER 5
SOFTWARE AND CONTROLLER DESIGN

The final design requirement for the emulator is a control algorithm for responsive and accurate torque delivery to the emulator braces. This chapter will summarize the emulator’s control architecture, system model, and controller implementation and verification.

5.1 Control Architecture

Motors, transmission assemblies, and all actuation/instrumentation hardware are housed in an offboard actuation cart (Figure 5.1). Motor current is regulated with four 4-quadrant PWM servo controllers (409510, Maxon Motor, Sachseln, Switzerland), for which a constant 48 V DC is supplied by two 480 W power supplies (DRP-480S-48, MEAN WELL USA, Fremont, CA). A single 75 W power supply (DR-75-24, MEAN WELL USA, Fremont, CA) powers four analog amplifiers (IAA100, FUTEK, Irvine, CA) to amplify the low-voltage signals from the load cells. A 32-bit 33 MHz PCI-based hardware-in-the-loop (HIL) control board (Q8 Data Acquisition Board, Quanser, Markham, Ontario, Canada) is used to collect and simultaneously sample all sensor signals and deliver analog output signals to the motor amplifiers corresponding to desired motor current. The data acquisition board communicates via PCI bus with a desktop “target” computer equipped with a real-time operating system (QNX Neutrino RTOS, QNX, Ottawa, Ontario, Canada) capable of real-time digital I/O at 10 kHz. Real-time control algorithms are developed and compiled using MATLAB/Simulink (Mathworks, Natick, MA) and QUARC software (Quanser, Markham, Ontario, Canada) on a “host” computer in a Windows environment, and uploaded to the target computer for real-time operation via ethernet bus. A schematic detailing the emulator hardware and flow of data is shown in Figure 5.2.
Figure 5.1: Actuation Cart
A) Transmission Tethers B) Motor Assemblies C) Load Cell Amplifiers D) Load Cell Power Supply E) Data Acquisition Board
F) Data Acquisition Computer G) Motor Amplifiers H) Motor Power Supplies
Figure 5.2: Schematic: Hardware Control Architecture
5.2 Software Model

A generic model was developed in Simulink (Mathworks, Natick, MA) for torque control and data collection with the exoskeleton emulator. The program converts signals from all 16 input signals to their physical variables: Encoder inputs from the data acquisition board are read directly by Simulink in encoder counts, and converted to angular position by the ratio of counts per revolution. Analog inputs from temperature sensors and load cells are read by Simulink as voltages, filtered by digital low-pass filters (Butterworth, 2nd order, 10 Hz and 20 Hz cutoff, respectively) and converted to motor housing temperature and brace torques according to the output mapping of the sensors.

Brace torque set points\(^1\) for each actuated joint are fed into the controller (described in Section 5.3), by which a motor current output reference is calculated and sent to the data acquisition board, where it is converted to an analog signal and sent to the motor amplifiers. Torque set point and motor current output references are saturated to safely limit brace torques. Model sample time is 0.0001 s to match the sampling rate of the target computer (10 kHz).

5.2.1 Gait Event Detection

To apply torques at specific times during gait, the software model requires accurate real-time information regarding the wearer’s gait phase at any given point in time. Therefore, a custom gait event detection algorithm was implemented in the generic system model, using only brace encoder readings. The core mechanism behind the algorithm is an observable impact peak in angular time-series data collected by the emulator occurring immediately after heel strike. This peak is assumed to coincide with the heel-strike event for real-time control purposes (Figure 5.3).

Brace angular velocity \(\omega_b\) is approximated from real-time brace angle \(\theta_b\) data using a discrete derivative and digital low-pass filter (Butterworth, 2nd order, 6 Hz). The algorithm

\(^1\)Details of torque set point are unique to the scenario under investigation with the emulator, and are therefore not included in the generic model.
Figure 5.3: Representative Angle Time-Series Data during Treadmill Gait. Brace angle and angular velocity values computed by the emulator software model are shown in blue and red, respectively for (a) hip and (b) knee flexion. Green arrows denote heel strike events.
utilizes angular thresholds informed by typical kinematic profiles of the joint angle of interest\(^2\) in combination with changes in sign of the angular velocity to detect local maxima corresponding to peak flexion/extension angles. Boolean flags corresponding to detection of specific gait events within the model are used as additional logical operators to increase reliability of subsequent event detection. Table 5.1 outlines the logic implemented by the gait event detection algorithm using angular data from either hip or knee:

<table>
<thead>
<tr>
<th></th>
<th>Peak Flexion</th>
<th>Peak Extension</th>
<th>Heel Strike</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Hip</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>if:</td>
<td>EXT = 1</td>
<td>if: ( \theta_{b,i} &lt; 0 )</td>
<td>if: FLX = 1</td>
</tr>
<tr>
<td>( \theta_{b,i} &gt; 20 )</td>
<td></td>
<td>( \omega_{b,i} &gt; 0 )</td>
<td>( \omega_{b,i} &gt; \omega_{b,(i-1)} )</td>
</tr>
<tr>
<td>( \omega_{b,i} &lt; 0 )</td>
<td></td>
<td>( \omega_{b,(i-1)} \leq 0 )</td>
<td>FLX = 0</td>
</tr>
<tr>
<td>( \omega_{b,(i-1)} \geq 0 )</td>
<td></td>
<td>then: EXT = 1</td>
<td>then: EXT = 0</td>
</tr>
<tr>
<td>then: FLX = 1</td>
<td></td>
<td>HS = 0</td>
<td>HS = 1</td>
</tr>
<tr>
<td><strong>Knee</strong></td>
<td>( \theta_{b,i} &gt; 40 )</td>
<td>if: FLX = 1</td>
<td>if: EXT = 1</td>
</tr>
<tr>
<td>if:</td>
<td>( \omega_{b,i} &lt; 0 )</td>
<td>if: ( \theta_{b,i} &lt; 20 )</td>
<td>( \omega_{b,i} &lt; \omega_{b,(i-1)} )</td>
</tr>
<tr>
<td>( \omega_{b,(i-1)} \geq 0 )</td>
<td>( \omega_{b,i} &gt; 0 )</td>
<td>( \omega_{b,(i-1)} \leq 0 )</td>
<td>FLX = 0</td>
</tr>
<tr>
<td>then: FLX = 1</td>
<td>( \omega_{b,i} &gt; \omega_{b,(i-1)} )</td>
<td>then: EXT = 0</td>
<td>HS = 1</td>
</tr>
<tr>
<td>then: HS = 0</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Real-time brace angles (\(\theta_b\)) and angular velocities (\(\omega_b\)) are compared to find peaks in the brace angles, which are then used to identify heel strike. Boolean flags (FLX, EXT, HS) are used to both track prior gait events and inform subsequent event detection.

5.3 Controller Design

This section will describe the design and implementation of the torque controller for the Simulink model. Four main goals were considered while designing the torque controller:

1. **Accuracy**: Applied torques should closely match desired torques fed to the controller.

2. **Responsiveness**: Rise and settling times should be minimized.

3. **Low overshoot**: Applied torques should quickly converge to their desired value rather than oscillate around it.

\(^2\)Angular thresholds are tuned according to magnitudes of peak angles in nominal gait patterns, and should be adjusted to the individual user.
4. Transparency: Magnitude of torques should be as close to zero as possible when no torque is desired.

A PID (proportional-integral-derivative) control algorithm was chosen for the controller due to its ease of implementation and proven ability to meet the first three stated controller goals when properly tuned. The operation of a PID controller is based upon the error between the controller set point and a measured plant value. The output of the controller attempts to minimize the error:

$$u(t) = K_p \cdot e(t) + K_i \cdot \int_0^t e(t) dt + K_d \cdot \frac{d}{dt} e(t) \quad (5.1)$$

Where $u(t)$ is the controller output, $e(t)$ is the error between the set point $r(t)$ and measured plant value $y(t)$, $K_p$ is the proportional gain, $K_i$ is the integral gain, and $K_d$ is the derivative gain. The transfer function of a PID controller is given by the Laplace transform of Equation 5.1:

$$G(s) = K_p + K_i \frac{1}{s} + K_d s \quad (5.2)$$

Digital implementation of a PID controller requires $u(t)$, $e(t)$, $r(t)$ and $y(t)$ to be discretized. Using a Forward Euler method for numerical integration and low-pass filter (a less-noisy alternative to numerical differentiation), the transfer function becomes:

$$G(z) = K_p + K_i \frac{T_s}{z - 1} + \frac{K_d N}{1 + NT_s z^{-1}} \quad (5.3)$$

Where $T_s$ is the sampling period and $N$ is the low-pass filter coefficient. Controller tuning is accomplished by adjusting the values of $K_p$, $K_i$, $K_d$, and $N$ to minimize rise time, overshoot, and steady-state error.

### 5.3.1 Direct Torque Control

The first controller implementation (Figure 5.4) utilized a basic PID controller using the measured torque at the brace as the plant value and a commanded motor current as the controller output.
Preliminary controller tuning and evaluation were conducted with a single motor/transmission assembly actuating the knee flexion DOF for a single brace. The brace was placed against the maximum flexion hard stop, and data was collected to evaluate step response over a range of reference torque values (manually entered during model operation). Tuning parameters were iteratively adjusted to optimize controller performance (Table 5.2).

Table 5.2: Direct PID Torque Controller Parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$K_p$</td>
<td>0.4</td>
</tr>
<tr>
<td>$K_i$</td>
<td>0</td>
</tr>
<tr>
<td>$K_d$</td>
<td>0.0001</td>
</tr>
<tr>
<td>$N$</td>
<td>$40 \cdot 2 \cdot \pi$</td>
</tr>
<tr>
<td>$T_s$</td>
<td>0.0001</td>
</tr>
</tbody>
</table>

The performance of the direct PID torque controller was found to be extremely limited. The proportional gain required to eliminate steady-state error at low torque values resulted in highly oscillatory response and instability at higher commanded torque values. Additionally,
the noise from the load cell reading presented another challenge, as filtering of the signal caused a measurement delay, further contributing to system instability. The prioritization of controller stability at all commanded torque magnitudes resulted in unacceptably large steady-state errors and response times.

5.3.2 Position-Based Torque Control

To address the issues associated with direct torque control, the second controller implementation aimed to use the current motor angle $\theta_m$ as the plant variable, with a set point informed by two variables: the desired brace torque magnitude $\tau_{b,\text{des}}$ and the current brace angle $\theta_b$.

5.3.2.1 Basic Control Model

Initial conceptualization of the controller was motivated by a desire to optimize controller performance with zero commanded torque (transparent mode). The idea was that in dynamic transparent operation, the motor should act to maintain the minimum zero-torque transmission cable length - in other words, the motor should let out and take in slack depending on the brace angle to maintain tangency between cable and pulleys:

$$\theta_m = A\theta_b + b$$

Where $A = \frac{p_m}{p_b}$ is the ratio of pulley diameters, and $b = \theta_{m0} - \theta_{b0}$ is the angle differential at the minimum zero-torque cable length.

Using this simple physical model, the PID controller set point is calculated in real time as a function of brace angle and cable length. While a known challenge associated with this approach is the characterization of the synthetic transmission cable’s compliance (in order to expand the model beyond transparent-only operation to a larger range of commanded torques), the manually-tuned PID controller resulted in nearly instantaneous response times with no observable overshoot or steady-state error (likely due to the fast response time of the motor and absence of noise in the encoder input signals). However, the torques measured
at the brace experienced substantial error even at $\tau_{b,des} = 0$, indicating a need to improve fidelity of the physical model used to generate the PID set point.

Friction within the Bowden-cable transmission was identified as a primary contributor to the limited performance of the controller. This is due to the fact that friction acts to assist the motor in generating torque at the brace in one direction of brace motion, but counteracts the motor in the other. For example, when the motor is configured to deliver flexion torques during increasing knee extension, the frictional force inflates the measured knee flexion torque. Conversely, during increasing knee flexion, the frictional force causes a decrease in measured knee flexion torque. Frictional forces within the transmission tether are also highly dependent upon the tether’s routing geometry.

### 5.3.2.2 Physical Transmission Characterization

To account for both cable compliance and the path- and direction-dependent effects of friction, a custom Simulink model (utilizing direct PID torque control\(^3\)) and transmission characterization procedure were developed, with the goal of broadly mapping the conditions under which specific brace torques are realized. The characterization procedure and model functionality are as follows:

The brace of interest and transmission tether are orientated and routed, respectively, in a manner representative of the conditions under which they will be used for experimentation. The brace DOF is placed against the mechanical hard stop corresponding to the direction of torque generation, and slack is removed from the cable via manual back-driving of the motor. The model is then initialized with a constant desired torque of 0.4 N m and the initial brace angle (against the hard stop) is recorded by the model. At this point, desired torque is reduced to 0.15 N m, the approximate minimum value for which the controller is able to overcome static friction within the Bowden-cable. The brace is then manually manipulated through its entire range of motion and back for three trials. Desired torque is increased by

---

\(^3\)Although the direct PID torque controller as described above is unable to match its set point, it is dynamically stable for a large range of torque set points.
0.1 N·m and three more trials are conducted. This step is repeated until smooth manual manipulation becomes too difficult. During the characterization procedure, brace angle, motor angle, and brace torque are recorded by the model and written to a data file. After the final trials (typically around $\tau_{b,\text{des}} = 9$ N·m) are completed, the model is terminated.

Next, the data from the characterization procedure is read into a custom MATLAB script. All data points for which the brace angle is within $10^\circ$ of a hard stop are discarded, and the remaining data are grouped according to direction of motion (i.e. flexion or extension). Within each direction group, the minimum and maximum brace torques are extracted and used to define 100 discrete torque ranges. The script then iterates through the trimmed datasets and allocates all data points into one of the 100 group definitions according to the brace torque. For each group, the average measured torque value is calculated and a first order polynomial is fit to the angular data in the form of Equation 5.4.

Polynomial coefficients $A$ and $b$ from each of the 100 groups per direction of motion are then plotted against respective average torque values for easy removal of any outliers resulting from poor characterization trials. As can be seen in Figure 5.5, a clear nonlinear relationship exists between the motor angle offset $b$ and the brace torque $\tau_b$, and is well-described by a second-order polynomial (also shown in Figure 5.5).

Incorporation of the observed relationship between motor angle offset and brace torque into Equation 5.4 yields a refined model for controller set point determination:

$$\theta_m = A\theta_b + B\tau_{b,\text{des}}^2 + C\tau_{b,\text{des}} + D$$

(5.5)

Where $B$, $C$, and $D$ are the second-order polynomial coefficients defining motor angle offset $b$ as a function of brace torque $\tau_b$. 

44
Figure 5.5: Representative Transmission Characterization
Data points with inconsistent values for pulley diameter ratios are easily identified as outliers. A clear relationship exists between motor angle offset and brace torque, which is well-defined by a second-order polynomial.
5.3.2.3 Model Calibration

With the nonlinear cable compliance and path- and direction-dependent friction behavior of the transmission characterized, the polynomial coefficients from Equation 5.5 can be employed in the control model. However, because the modularity of the emulator system allows numerous transmission assembly/brace combinations, a short model calibration procedure was developed. The goal of the model calibration is to allow the control model to account for changes in effective transmission cable length from the time of transmission characterization (due to altered transmission tether routing, cable failure and replacement, unexpected encoder zero-count reset, etc.) without the need to perform an exhaustive transmission re-characterization.

The model calibration is essentially an abbreviated transmission characterization. Bidirectional trials are collected as described above for three torque set-points spanning the range of torque generation of the direct torque controller (i.e. 0.5 N m, 1.5 N m, and 2.5 N m), and again read into a custom MATLAB script, which computes first-order polynomial fits relating brace angle to motor angle for individual brace torque values. Data points are plotted along with the characterization data in order to identify outliers and determine if a recalibration is necessary. Finally, an optimization is performed in MATLAB to compute a new value for the polynomial coefficient $D$ which minimizes RMS error between the motor angle offset data points and a second-order polynomial defined by the coefficients $B$, $C$, and $D$ (values for $B$ and $C$ are fixed at the values calculated during characterization). Separate polynomial coefficients are calculated for each direction of motion. Results of a successful model calibration are shown in Figure 5.6).
Figure 5.6: Successful Model Calibration (Right Knee Brace)
5.3.2.4 Controller Implementation

The final controller implementation (Table 5.3, Figure 5.7) utilizes a calibrated PID controller with set point $\theta_{m,des}$ defined as a function of two input variables: $\tau_{b,des}$ and $\theta_b$ for indirect real-time control of the brace torque $\tau_b$.

Table 5.3: Position-Based PID Torque Controller Parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$K_p$</td>
<td>0.012</td>
</tr>
<tr>
<td>$K_i$</td>
<td>0.03</td>
</tr>
<tr>
<td>$K_d$</td>
<td>0.0002</td>
</tr>
<tr>
<td>$N$</td>
<td>$60 \cdot 2 \cdot \pi$</td>
</tr>
<tr>
<td>$T_s$</td>
<td>0.0001</td>
</tr>
</tbody>
</table>

Figure 5.7: Schematic: Position-Based PID Torque Controller

Model calibration results (Figure 5.8) for each direction of brace motion are hard-coded into the control algorithm to map the current encoder zero-position and transmission cable length to an experimentally-determined model-based characterization of the transmission properties.
Output references for desired motor position are calculated for each model calibration direction \((\theta_{m,des,flx}, \theta_{m,des,ext})\) at each model time step. The controller “chooses” the appropriate output reference based on the current direction of motion at the brace (determined by a real-time measurement of brace angular velocity, \(\omega_b\)). For small angular velocities (\(|\omega_b| \leq \frac{25^\circ}{s}\)), a linear relationship is used to smoothly transition between output references. Therefore, (assuming flexion is positive) the output reference can be generally represented by Equations 5.6 - 5.8:

\[
\theta_{m,des,ext} = A \cdot \theta_b + B_{ext} \cdot \tau_{b,des}^2 + C_{ext} \cdot \tau_{b,des} + D_{ext} \tag{5.6}
\]

\[
\theta_{m,des,flx} = A \cdot \theta_b + B_{flx} \cdot \tau_{b,des}^2 + C_{flx} \cdot \tau_{b,des} + D_{flx} \tag{5.7}
\]

\[
\theta_{m,des} = \begin{cases} 
\frac{\theta_{m,des,ext}}{50} \cdot \omega_b + \frac{\theta_{m,des,flx}-\theta_{m,des,ext}}{2} & \omega_b < -25 \\
\theta_{m,des,flx} & -25 \leq \omega_b \leq 25 \\
\theta_{m,des,flx} + \frac{\theta_{m,des,ext}-\theta_{m,des,flx}}{2} & \omega_b > 25 
\end{cases} \tag{5.8}
\]
5.4 Controller Evaluation & Model Verification

To evaluate the performance of the controller, an experiment was performed in which a single healthy subject participated in two short treadmill walking trials at 1.2 m s\(^{-1}\) while wearing the right knee brace configured to deliver knee flexion torques. The first trial was conducted in transparent mode (\(\tau_{b, des} = 0\)). In the second trial, desired torque patterns were generated as a function of knee flexion angle to emulate a clutched torsional spring (\(\kappa = 3.18\, \text{N m rad}^{-1}\)) to resist knee extension during swing.

Results for both trials are presented in Figure 5.9 for three representative gait cycles. In transparent mode, motor position closely tracked its desired trajectory, and interaction torques remained at or below 0.1 N m for the entire trial. Motor position tracking was also consistent during spring emulation mode, resulting in torque profiles closely matching their desired trajectories with minimal delay.
Figure 5.9: Controller Performance During Gait
Knee flexion angle, desired vs. actual motor position, and desired vs. actual joint torque during gait at 1.2 m/s with emulator: (a) operating in transparent operation mode with zero desired torque, and (b) delivering torques as a function of knee flexion angle to emulate a clutched torsional spring resisting knee extension during swing.
CHAPTER 6
PRELIMINARY EXPERIMENT: ENERGY STORAGE DURING GAIT

With the performance of the torque controller verified, the emulator is ready to be utilized for human subject experimentation. This chapter describes a preliminary experiment conducted with the knee braces of the emulator.

6.1 Background

As discussed in Chapters 1 and 2, the motivation behind the design and implementation of the exoskeleton emulator was to inform the design of an assistive multi-joint body-powered leg exoskeleton (AMBLE) to aid hemiparetic gait. Such a device carries the potential to broadly impact the post-stroke population, as the absence of expensive motors and batteries (core drivers of exoskeleton cost) may substantially drive down final device cost.

Literature has identified ankle plantarflexion and hip flexion at terminal stance as desirable targets for hemiparetic gait assistance. However, a specific challenge associated with the design of a body-powered device capable of providing such assistance is that all energy delivered by the exoskeleton must originate from the wearer. Variable stiffness springs and lightweight, low-power clutches afford versatility to the manner in which energy may be stored, but research specifically targeting energy storage during gait is limited. Whether a sufficient amount of mechanical energy can be stored during a single gait cycle to meaningfully improve hemiparetic gait is an important question that must be answered if such a device is to be realized.

One strategy for storing energy during gait is to replace the existing negative work requirement of the wearer, which exists for all lower-limb joints at some point in time during gait. Theoretically, a device with optimally-tuned and properly-clutched springs could offload a large portion of the negative work requirement from the lower-limb musculature without adversely altering gait patterns. This theory has been lent experimental support by recent
research involving a passive ankle exoskeleton [60]. In the study, a substantial amount of mechanical energy was stored in a spring during ankle dorsiflexion in the stance phase of gait and returned to the ankle to assist push-off, resulting in a decrease of metabolic expenditure of the wearer. While this study did not explicitly isolate energy storage, it does provide a foundation for future studies investigating energy storage during gait.

Rather than storing and returning energy at a single joint, AMBLE seeks to provide an external path for energy stored by joints in the healthy leg of a post-stroke individual to assist the impaired limb at advantageous times during gait. The effect of such a strategy is to increase the work requirement of the healthy limb in order to aid impaired function of the impaired limb. Therefore, studies investigating isolated energy storage in this context should not be limited to parts of the gait cycle where negative work is typically performed at any given joint, and neither should their success be directly tied to the commonly-observed desire to reduce metabolic expenditure. Instead, they should seek to quantify the amount of energy storage tolerable by humans at a given joint during gait, in order to truly identify an upper limit on available energy for AMBLE.

The goals of the preliminary study described below were as follows:

1. To develop and validate an experimental protocol for quantitative investigation of mechanical energy storage during gait using the exoskeleton emulator to mimic the behavior of clutched springs.

2. To quantify the maximum tolerable amount of mechanical energy storage using an emulated clutched torsional spring to resist knee extension during the swing phase of able-bodied gait.

6.2 Methods

6.2.1 Participant

One healthy subject (female, 20 years; 57.6 kg; 1.70 m) volunteered to participate in this experiment. The subject was familiar with the operation of the exoskeleton emulator, and
was provided a detailed description of the experimental methods, including the magnitude of torques that were to be applied.

6.2.2 Measurements

In addition to the real-time data collected by the emulator (brace angle and applied brace torque), lower-limb kinematics and electromyography (EMG) data were collected using a 7-camera motion capture system (Oqus 300, Qualisys, Gothenburg, Sweden) and wireless EMG system (Trigno, Delsys, Natick, MA), respectively.

*Motion Capture:* Reflective markers were placed as shown in Figure 6.1 for measurement of bilateral 3-D hip, knee, and ankle angles. Marker position data was recorded at 100 Hz.

![Figure 6.1: Motion Capture Marker Placement](image)

36 total markers were used during data collection. Markers at the blue locations (10) were used for static and functional joint location trials only. Markers at the red locations (26) were left in place for the duration of the experiment.
EMG data from 5 muscles per leg were collected: vastus medialis, rectus femoris, biceps femoris long head, tibialis anterior, and medial gastrocnemius.

*Electromyography:* EMG sensors were placed as shown in Figure 6.2 in accordance with published guidelines [81]. Skin was shaved and cleaned with an alcohol pad at the site of each sensor. Test contractions were conducted prior to sensor fixation in order to ensure adequate signal strength. Sampling frequency was set to 1000 Hz.

### 6.2.3 Exoskeleton Control Model

The control model for this experiment was adapted from the generic Simulink model described in Chapter 5.

*Spring:* Values for desired torque fed into the position controller are calculated using the following framework for the implementation of a clutched torsional spring. The zero-angle of each knee brace is determined with a user-enabled “static trial” subsystem which calculates a running average of brace encoder readings while active. This value is used as an offset to the raw encoder reading, yielding the anatomical knee angle of the wearer in real-time. When active, desired spring torque is calculated as a function of this anatomical angle as
well as the spring equilibrium angle and spring constant (both of which are hard-coded into the model).

**Clutch:** The clutch subsystem is implemented immediately upstream of the spring torque calculation to determine whether or not the calculated torque is fed to the position controller (default desired torque is 0 N m). Predetermined clutch rules (Table 6.1) are hard-coded into the subsystem, comprising a simplified version of the gait event detection algorithm described in Chapter 5:

<table>
<thead>
<tr>
<th>Engage Spring</th>
<th>Disengage Spring</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\theta_{b,i} &gt; 40$</td>
<td>SPR = 1</td>
</tr>
<tr>
<td>if: $\omega_{b,(i-1)} &gt; 0$</td>
<td>if: $\omega_{b,(i-1)} &lt; 0$</td>
</tr>
<tr>
<td>$\omega_{b,i} \leq 0$</td>
<td>$\omega_{b,i} \geq 0$</td>
</tr>
<tr>
<td>then: SPR = 1</td>
<td>then: SPR = 0</td>
</tr>
</tbody>
</table>

Brace angle ($\theta_b$) and angular velocity ($\omega_b$) readings at current ($i$) and previous ($i-1$) simulation loops are compared to numerical constants to determine spring engagement and disengagement. A boolean flag (SPR) is the output of the clutch subsystem.

**User Interface:** For ease of use during experimentation, a number of useful control model functions are mapped to user interface “buttons” on the front panel of the model, including:

- Static trial (1 = active, 0 = inactive).
- Spring constant determination (0 = no spring, 1-5 = predetermined spring constants of increasing magnitude).
- Motor Disable: Overrides output of position controller to send 0 A to the motor.
- Record Data: Because data is continuously written to a file for the duration of the model, this button writes a binary to the output file indicating whether an experimental trial is underway.
6.2.4 Procedure

After the subject was fully instrumented with the knee braces of the exoskeleton emulator, EMG sensors and reflective markers, the control model was initialized. A brief familiarization session was conducted during which the subject walked for several minutes at both designated treadmill speeds with the emulator operating in transparent mode to ensure proper joint alignment of the emulator braces. This was followed by a 5-second static standing trial to determine nominal joint angles and a 10-second seated trial to measure baseline EMG activity. With the nominal joint angles defined in the control model, a second familiarization session was conducted with the emulator actively delivering knee flexion torques according to the control model described above. This session was used to both verify proper device performance and to evaluate the maximum emulated spring stiffness ($\kappa$) at which the braces would remain stable during gait. At $\kappa = 4 \text{ N m rad}^{-1}$ the subject reported a noticeable “slipping” of the brace cuffs on her leg, so five distinct emulated spring stiffnesses were selected ranging between 0 and 3.5 N m rad$^{-1}$.

The experimental protocol consisted of 12 2-minute experimental trials with speeds and emulated spring stiffnesses as described in Table 6.2. During each trial, 90 seconds were allocated for adaptation with 30 seconds of data collection immediately following. A mandatory 5-minute break followed the first 6 trials.

6.3 Data Processing

Motion capture data were post-processed in Qualisys Track Manager (Qualisys, Gothenburg, Sweden) and Visual3D (C-Motion, Germantown, MD) to generate 3-D joint angle time series data for the hip and knee. The resulting angles, along with time-series data for exoskeleton brace angle and torque, were post-processed in MATLAB (Mathworks, Natick, MA). Data were segmented into individual steps based on the heel-strike events detected by the emulator control model algorithm and spline-fit into arrays of equal length to facilitate step-to-step comparison. For each step, the negative work performed by the exoskeleton
Table 6.2: Experimental Conditions

<table>
<thead>
<tr>
<th></th>
<th>( v ) (m s(^{-1}))</th>
<th>( \kappa ) (N m rad(^{-1}))</th>
</tr>
</thead>
<tbody>
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<td>0</td>
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</tr>
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</tr>
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<td></td>
<td>1.91</td>
</tr>
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<td>5</td>
<td></td>
<td>2.55</td>
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<td>12</td>
<td></td>
<td>3.18</td>
</tr>
</tbody>
</table>

brace (energy stored in the emulated spring) was calculated via numerical integration of the measured torque with respect to brace angle while the spring was active.

6.4 Results

Calculated values for energy storage per step are shown in Figure 6.3. Knee flexion angles and applied torques (as collected by the emulator control model) are presented in Figure 6.4. Figure 6.5 shows a comparison of knee flexion angles as measured by the brace encoders and motion capture, and Figure 6.6 presents the 3D hip angles for all conditions.\(^4\)

6.5 Discussion

6.5.1 Energy Storage

Emulated energy storage per step, as shown in Figure 6.3, was quite small (under 1 J) for all conditions and is summarized in Table 6.3.

---

\(^4\)For brevity, results are only included for the right leg (left leg results are similar for all plots shown).
Figure 6.3: Results: Emulated Energy Storage per Step at the Right Knee

Average energy storage per step at the right knee (calculated by numerically integrating exoskeleton torque with respect to brace angle while the spring was active) for all individual gait cycles in the 30 seconds of data collection are shown for each emulated spring constant for (a) conditions 1-6 and (b) conditions 7-12. Error bars denote standard deviation across gait cycles.
Figure 6.4: Results: Right Knee Flexion Angles and Exoskeleton Torques
Mean (solid) ± standard deviation (shaded) values for all individual gait cycles in the 30 seconds of data collection are shown for exoskeleton torque and knee flexion angle (as measured by the emulator brace instrumentation) for (a) conditions 1-6 and (b) conditions 7-12. Each color represents a different emulated spring constant, as indicated in the legend. Desired torques are indicated by dashed lines.
Figure 6.5: Results: Right Knee Flexion Angles (Brace Encoder vs. Motion Capture)
Mean encoder (solid) and mean ± std motion capture (shaded) right knee flexion angles for all gait cycles in the 30 seconds of data collection are shown for (a) conditions 1-6 and (b) conditions 7-12.
Figure 6.6: Results: 3-D Right Hip Joint Angles

Mean (solid) ± standard deviation (shaded) values for all individual gait cycles in the 30 seconds of data collection are shown for right hip F/E, A/A, and I/E joint angles, as measured by motion capture for (a) conditions 1-6 and (b) conditions 7-12. Each color represents a different emulated spring constant, as indicated in the legend.
Table 6.3: Average Energy Storage per Step: Right Knee

<table>
<thead>
<tr>
<th></th>
<th>$v$ (m s$^{-1}$)</th>
<th>$\kappa$ (N m rad$^{-1}$)</th>
<th>$\dot{U}$ (J/step)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.8</td>
<td>0</td>
<td>0.053</td>
</tr>
<tr>
<td>2</td>
<td>0.64</td>
<td>1.27</td>
<td>0.348</td>
</tr>
<tr>
<td>3</td>
<td>1.91</td>
<td>2.55</td>
<td>0.567</td>
</tr>
<tr>
<td>4</td>
<td>3.18</td>
<td>3.18</td>
<td>0.634</td>
</tr>
<tr>
<td>5</td>
<td>1.2</td>
<td>0</td>
<td>0.060</td>
</tr>
<tr>
<td>6</td>
<td>0.64</td>
<td>1.27</td>
<td>0.353</td>
</tr>
<tr>
<td>7</td>
<td>1.91</td>
<td>2.55</td>
<td>0.571</td>
</tr>
<tr>
<td>8</td>
<td>3.18</td>
<td>3.18</td>
<td>0.836</td>
</tr>
</tbody>
</table>

Due to the overall consistency of brace kinematics within conditions (as evidenced by the small angular standard deviations in Figure 6.4), energy stored per step during this experiment is, for the most part, directly correlated with the emulated spring constant regardless of gait velocity. Condition 12 presents an exception to this trend, with an average energy storage per step of 0.836 J compared to 0.634 J in condition 6. This dramatic shift can be explained by an examination of the knee kinematics for condition 12 in Figure 6.4. While the emulated spring is active (from $\sim$75% to $\sim$100% gait cycle) the angle measured by the brace encoder is noticeably lower than in condition 6, resulting in an increase of brace torque and therefore energy storage. A cause of this phenomenon is hypothesized at the end of Section 6.5.2.

6.5.2 Kinematics

A clear trend can be observed in Figure 6.4 for the knee flexion angles surrounding heel strike ($\sim$95% thru $\sim$10% gait cycle) for conditions 1-11, showing a decrease in peak knee extension angle with increasing spring stiffness. This result is not surprising in the presence of external flexion torques which experiences their peak values at $\sim$95% of the gait cycle for
all conditions. During weight acceptance (~0% thru ~25% gait cycle), knee flexion angles for conditions 1-6 and 7-11 converge to similar values of ~15° and ~18°, respectively.

Also observable in Figure 6.4, although not as clear, is a trend of increasing peak knee flexion angles (~75% gait cycle) with increasing spring stiffness for conditions 2-6 and 8-11. This may indicate an adaptation on the part of the subject, in anticipation of the resistive torques to occur during knee swing.

Knee flexion angle profiles were consistent between the two measurement methods (emulator encoder and motion capture) for conditions 1-6, indicating that the angles used by the control model were an accurate representation of the anatomical joint angles at 0.8 m s$^{-1}$ (Figure 6.5). Additionally, hip angles as measured by motion capture were generally consistent for these conditions, with no discernable trends (Figure 6.6).

However, the same level of consistency is not present in the kinematic data at 1.2 m s$^{-1}$: notably, conditions 7, 8 and 12. This is partially attributable to the poor quality of motion capture data for conditions 7-12, which experienced substantial intermittent marker obstruction behind the handrail supports on the treadmill, necessitating a considerable amount of manual “gap-filling” of marker trajectories during post-processing. In addition, trajectories for two of the four pelvis markers (Figure 6.1) merged for the duration of conditions 7-9, contributing to the observable inconsistencies observed in (Figure 6.6) for these conditions.5

A potential cause for the phenomenon described in Section 6.5.1 regarding condition 12 is a “drop” of counts by the brace encoder. Acknowledging the weakness of the motion capture data, this hypothesis is supported by a combination of the considerable discrepancy between encoder and motion capture data for the knee (Figure 6.5) and the lack of any notable shift in hip angles beyond the existing variability (Figure 6.6) during condition 12.

5The merged trajectories correspond to right and left posterior iliac spine markers. Because no static trial was collected with a marker in this merged location, the merged trajectory was assigned to the landmark defined for the right posterior iliac spine marker. This caused a substantial rotation of the pelvis segment toward the right leg in Visual3D, introducing a permanent positive bias to reported right hip internal rotation angles, and a negative bias to reported hip abduction angles during periods of hip flexion. A converse effect (i.e. permanent negative bias to internal rotation, and positive bias to adduction during periods of flexion) can be observed for left hip angles during these conditions (Appendix B).
6.5.3 Subject Feedback

Qualitative feedback provided by the subject during experimental conditions 1-11 was positive, and indicated the torques being applied by the emulator were noticeable but easily tolerable. During condition 3, the subject remarked that her gait felt “more stable” during the application of torques. During condition 12, the subject remarked that walking in the presence of the applied torques required noticeably more effort. This result, in combination with the observed alteration of brace kinematics for condition 12 as discussed above may indicate an altered motor control strategy to compensate for the torques being provided by the emulator. If this is the case, the absence of an increase in perceived exertion by the subject during condition 6 indicate that higher energy storage is tolerable per step at slower gait speeds.

6.6 Conclusions

The first goal of this study was to develop an experimental protocol to investigate mechanical energy storage using emulated clutched torsional springs during gait. In this regard, the experiment was successful: the control model consistently detected peak flexion/extension angles and heel strike events for every step in every condition, generated desired torque profiles representative of a clutched torsional spring, and applied torques to the braces closely following the desired trajectories.

The second goal of this study was to quantify the maximum amount of tolerable mechanical energy storage at the knee during gait. While an increase in perceived exertion was observed for the highest spring constant/velocity combination tested, the applied torques were nevertheless considered “tolerable” to the subject. Therefore, the second goal of the study was not met. However, the energy stored per step in condition 12, normalized to body mass, corresponds to approximately 7% of able-bodied ankle push-off work as reported in [82], indicating that knee extension during swing should be further explored as a target for energy storage.
6.7 Limitations

In addition to the lack of additional participants, relative movement between the emulator brace and subject was a significant limitation to this study, as it constrained the maximum emulated spring stiffness to a value substantially below the capabilities of the emulator. Future versions of this experiment would benefit from a second-generation leg attachment cuff to provide a more secure interface between emulator brace and wearer.

The obstruction of reflective markers by the handrail supports on the treadmill is another limitation - both to the quality of motion capture data and ease of data processing. Future experimentation should seek to address this issue via re-orientation of the treadmill in the motion capture space, rearrangement of motion capture cameras, and/or physical removal of the treadmill handrails.

Finally, EMG data was not processed for this study. Knowledge of muscle activity and how it changes in the presence of spring-like perturbations during gait would provide additional insight into the potential biomechanical effects of AMBLE and population/wearer-specific limitations to energy storage during gait.
7.1 Conclusions

A lower-limb exoskeleton emulator intended for study of a broad range of research questions (including control algorithms and mechanical parameters of gait exoskeletons, human/device interaction, human adaptation, and joint impedance during gait) has been developed. The emulator is capable of closed-loop control to apply joint torques in up to four simultaneous lower-limb degrees of freedom.

Five rigid braces for human interaction spanning the hip, knee, and ankle joints of the wearer provide a total of 10 degrees of freedom for the wearer. Adjustment points on the braces allow joint axis alignment for a large range of wearer sizes and body types. Composed primarily of aluminum, braces were designed to be as lightweight as possible to maximize device transparency.

The braces are actuated by four brushless DC motors, which were selected to maximize system response. Custom heat sinks were designed and implemented to optimize the torque-generation capacity of the motors, which are housed in an off-board cart along with other hardware for actuation and sensing. Torques generated by the motors are transmitted to the braces via cable transmission. Modular transmission attachment points on the braces allow specific emulator configurations to be easily customized.

Instrumentation built into the emulator allows torques at the braces to be controlled in real time at a frequency of 10 kHz. A custom control algorithm for the device was developed using Simulink software, featuring a novel gait event detection algorithm and position-feedback PID controller. The controller utilizes results from a characterization/calibration procedure in order to account for the nonlinear behavior of the cable transmission. Desired torque, brace angle and motor angle are the inputs to the controller; motor current is the
output. Performance evaluations of the controller demonstrated its ability to accurately and responsively generate desired torques at the emulator braces.

A preliminary experiment utilizing the emulator was conducted in which spring-like knee flexion torques were applied to a single subject during treadmill walking. The specific control software and experimental protocol developed will serve as a template for future experimentation. This study represents a first step toward quantifying the amount of external mechanical energy storage that is tolerable to a human during gait, an important metric that will inform the feasibility of AMBLE - a proposed new body-powered exoskeleton intended to provide gait assistance to individuals suffering from hemiparesis following stroke.

7.2 Future Work and Recommendations

Future extensions of this project will involve its application to experiments intended to evaluate new exoskeleton designs (such as AMBLE) and related control algorithms. The device will also be used to provide validation for results of predictive gait simulations with combined human and device mechanics.

7.2.1 Mechanical Design

Leg attachment cuff design should be revisited in order to improve the transmission of torques from braces to wearer. Mechanical strength analyses of structural components of the braces can facilitate a further reduction in brace mass.

7.2.2 Actuation and Transmission

The current limitation to the maximum actuation torque capability is the torque rating of the planetary gearhead. However, a simple reduction of motor pulley diameter would increase torque delivery to the braces for any given gearhead output.

7.2.3 Software and Controller

Currently, the PID controller relies on a manual characterization and calibration procedure to account for friction and compliance of the transmission cable. Future work should
seek to streamline these procedures and account for routing geometry of the transmission cable (a primary driver of variations in cable friction). In addition, experimentation should be conducted to identify additional dependencies of the transmission to further improve the accuracy of torque application.

7.2.4 Toward a Passive Exoskeleton for Gait Assistance

An expanded protocol for the preliminary energy storage study is currently awaiting Institutional Review Board approval. Future experimentation targeting energy storage during gait will form the basis for the design of AMBLE. Once numerical values for available energy per step are determined, they can be used to inform additional experiments to determine the most effective methods for returning the energy to improve hemiparetic gait.
REFERENCES CITED


**APPENDIX A - MANUFACTURER SPEC SHEETS**

**maxon EC motor**

**EC-4pole 30 Ø30 mm, brushless, 200 Watt**

**High Power**

![Motor Specifications Diagram](image)

---

### Motor Data

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nominal voltage</td>
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</tr>
<tr>
<td>No load speed</td>
<td>2400 rpm</td>
</tr>
<tr>
<td>No load current</td>
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</tr>
<tr>
<td>Stall torque</td>
<td>0.24 Nm</td>
</tr>
<tr>
<td>Stall current</td>
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</tr>
<tr>
<td>Max. efficiency</td>
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</tr>
<tr>
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</tr>
<tr>
<td>Terminal inductance</td>
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</tr>
<tr>
<td>Torque constant</td>
<td>7.4 Nm/A</td>
</tr>
<tr>
<td>Speed constant</td>
<td>15000 rpm/V</td>
</tr>
<tr>
<td>Speed/torque gradient</td>
<td>7.2 rpm/mNm</td>
</tr>
<tr>
<td>Mechanical time constant</td>
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<tr>
<td>Rotor inertia</td>
<td>3 g cm²</td>
</tr>
<tr>
<td>Thermal resistance (housing-ambient)</td>
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</tr>
<tr>
<td>Thermal resistance (winding-housing)</td>
<td>0 K/W</td>
</tr>
<tr>
<td>Thermal time constant (winding)</td>
<td>1100 s</td>
</tr>
<tr>
<td>Max. radial load, 5 mm from flange</td>
<td>8.0 N</td>
</tr>
<tr>
<td>Ambient temperature</td>
<td>-20°C to +100°C</td>
</tr>
<tr>
<td>Max. speed</td>
<td>25000 rpm</td>
</tr>
<tr>
<td>Axial play at axial load</td>
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</tr>
<tr>
<td>Radial play</td>
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<tr>
<td>Max. force for press fits (static)</td>
<td>0.1 N</td>
</tr>
<tr>
<td>Max. radial load, 5 mm from flange</td>
<td>0.21 N</td>
</tr>
<tr>
<td>Max. axial load (static)</td>
<td>0.21 N</td>
</tr>
<tr>
<td>Max. force for press fits (static)</td>
<td>0.21 N</td>
</tr>
</tbody>
</table>

---

### Specifications

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
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<tr>
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</tr>
<tr>
<td>Terminal resistance</td>
<td>1 Ohm</td>
</tr>
<tr>
<td>Terminal inductance</td>
<td>0.1 mH</td>
</tr>
<tr>
<td>Torque constant</td>
<td>7.4 Nm/A</td>
</tr>
<tr>
<td>Speed constant</td>
<td>15000 rpm/V</td>
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<tr>
<td>Speed/torque gradient</td>
<td>7.2 rpm/mNm</td>
</tr>
<tr>
<td>Mechanical time constant</td>
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<tr>
<td>Rotor inertia</td>
<td>3 g cm²</td>
</tr>
<tr>
<td>Thermal resistance (housing-ambient)</td>
<td>21 K/W</td>
</tr>
<tr>
<td>Thermal resistance (winding-housing)</td>
<td>0 K/W</td>
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<td>8.0 N</td>
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<tr>
<td>Ambient temperature</td>
<td>-20°C to +100°C</td>
</tr>
<tr>
<td>Max. speed</td>
<td>25000 rpm</td>
</tr>
<tr>
<td>Axial play at axial load</td>
<td>&lt; 0.01 mm</td>
</tr>
<tr>
<td>Radial play</td>
<td>&lt; 0.02 mm</td>
</tr>
<tr>
<td>Max. radial load (dynamic)</td>
<td>&lt; 0.002 Nm</td>
</tr>
<tr>
<td>Max. axial load (static)</td>
<td>&lt; 0.01 Nm</td>
</tr>
<tr>
<td>Max. force for press fits (static)</td>
<td>0.1 N</td>
</tr>
<tr>
<td>Max. radial load, 5 mm from flange</td>
<td>0.21 N</td>
</tr>
<tr>
<td>Max. axial load (static)</td>
<td>0.21 N</td>
</tr>
<tr>
<td>Max. force for press fits (static)</td>
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### Other Specifications

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<tr>
<td>Weight of motor</td>
<td>300 g</td>
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</tbody>
</table>

---

### Recommended Electronics:

- Encoder MR: 1024 - 5000 cpm, 3 channels
- Encoder HEDL 5540: 500 cpm, 3 channels
- Encoder HEDL 5530: 500 cpm, 3 channels
- Encoder EPOS HEDL 910: 500 cpm, 3 channels
- Encoder EPOS HEDL 900: 500 cpm, 3 channels
- Encoder EPOS HEDL 940: 500 cpm, 3 channels
- Encoder EPOS HEDL 950: 500 cpm, 3 channels
- Encoder EPOS HEDL 960: 500 cpm, 3 channels
- Encoder EPOS HEDL 970: 500 cpm, 3 channels
- Encoder EPOS HEDL 980: 500 cpm, 3 channels
- Encoder EPOS HEDL 990: 500 cpm, 3 channels

---

**Figure A.1: Motor Specifications**
### Ceramic Version

#### Planetary Gearhead GP 42 C

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<th>Diameter (mm)</th>
<th>Gearhead Length L1</th>
<th>Gearhead Length L2</th>
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</thead>
<tbody>
<tr>
<td>42 mm</td>
<td>3-15 Nm</td>
<td></td>
</tr>
</tbody>
</table>

#### Technical Data

- **Motor**: maxon Modular System
- **Sensor**: maxon gear
- **Brake**: maxon gear
- **Overall length (mm)**
  - M 12: 55.5
  - M 18: 55.5
  - M 24: 55.5

#### Part Numbers

<table>
<thead>
<tr>
<th>Model</th>
<th>Part Numbers</th>
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</thead>
<tbody>
<tr>
<td>RE 40, 170 W</td>
<td>260552* 203130 203134 203138 203142</td>
</tr>
<tr>
<td>RE 40, 150 W</td>
<td>203127 203132 203136 203140</td>
</tr>
<tr>
<td>RE 40, 150 W</td>
<td>203124</td>
</tr>
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<td>RE 40, 170 W</td>
<td>203135</td>
</tr>
<tr>
<td>RE 40, 150 W</td>
<td>203137</td>
</tr>
<tr>
<td>RE 40, 150 W</td>
<td>203139</td>
</tr>
</tbody>
</table>

#### Specifications

- **Max. continuous input speed**: 8000 rpm
- **Recommended temperature range**: -40...+100°C
- **Axial play at axial load**: < 0.06 mm
- **Radial play, 12 mm from flange**: max. 0.06 mm
- **Axial load (dynamic)**: 150 N
- **Axial force for press fit**: > 5 N
- **Direction of rotation, drive to output**: M 1:2
- **Absolute reduction**: from flange 120 N 240 N 360 N
- **Max. continuous torque**: 4.9 Nm
- **Max. intermittent torque**: 9.4 Nm
- **Max. motor shaft diameter**: 11.3 mm
- **Max. motor shaft diameter**: 22.5 mm
- **Mass inertia**: 0.6 g·cm²
- **Mass inertia**: 0.8 g·cm²
- **Mass inertia**: 1.0 g·cm²
- **Number of stages**: 3
- **Number of stages**: 4
- **Number of stages**: 4
- **Number of stages**: 1
- **Number of stages**: 2
- **Number of stages**: 3
- **Number of stages**: 4
- **Number of stages**: 4

#### Stock program

- **Motor**: maxon Modular System
- **Sensor**: maxon gear
- **Brake**: maxon gear
- **Overall length (mm)**
  - M 12: 55.5
  - M 18: 55.5
  - M 24: 55.5

#### Special program (on request)

- **Motor**: maxon Modular System
- **Sensor**: maxon gear
- **Brake**: maxon gear
- **Overall length (mm)**
  - M 12: 55.5
  - M 18: 55.5
  - M 24: 55.5

#### Overall length (mm)

- **M 12**: 55.5
- **M 18**: 55.5
- **M 24**: 55.5
FEATURES

• Up to 10 times the overload protection
• Overload is available in Tension and Compression
• Light weight
• Notable nonlinearity
• Loads up to 100 lb (445 N)
• Miniature size

SPECIFICATIONS

PERFORMANCE
Nonlinearity ±0.1% of RO
Hysteresis ±0.1% of RO
Nonrepeatability ±0.05% of RO

ELECTRICAL
Rated Output (RO) See chart on third page
Excitation (VDC or VAC) 10 max
Bridge Resistance See chart on third page
Insulation Resistance ≥500 MOhm @ 50 VDC
Connection #29 AWG, 4 conductor, spiral shielded silicone cable, 5 ft (1.5 m) long

MECHANICAL
Weight (approximate) 0.3 oz (9 g)
Safe Overload 1000% of RO
200% tension only (50–100 lb)
Material Aluminum (10 g–10 lb), stainless-steel (25–100 lb)

TEMPERATURE
Operating Temperature -60 to 200°F [-50 to 93°C]
Compensated Temperature 60 to 160°F [15 to 72°C]
Temperature Shift Zero ±0.01% of RO/°F [0.018% of RO/°C]
Temperature Shift Span ±0.02% of Load/°F [0.036% of Load/°C]

CALIBRATION
Calibration Test Excitation 5 VDC
Calibration (standard) 5-pt Tension
Calibration (available) Compression

Figure A.3: Load Cell Specifications (page 1/3)
Figure A.3: Load Cell Specifications (page 2/3)
<table>
<thead>
<tr>
<th>ITEM #</th>
<th>lb</th>
<th>N</th>
<th>Thread</th>
<th>RO (nom)</th>
<th>Bridge Resistance</th>
<th>Shunt Calibration Value</th>
<th>Deflection (in.)</th>
<th>Natural Frequency (Hz)</th>
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</thead>
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<tr>
<td>FSH02534</td>
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<td>0.1</td>
<td>#4-40</td>
<td>0.5 mV/V</td>
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<td>0.004</td>
<td>140</td>
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<td>0.2</td>
<td>#4-40</td>
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<td>0.008</td>
<td>140</td>
<td></td>
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<tr>
<td>FSH02535</td>
<td>50g</td>
<td>0.5</td>
<td>#4-40</td>
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<td>1000 Ohm nom</td>
<td>0.010</td>
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<td></td>
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<td></td>
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<td>#4-40</td>
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<td>0.007</td>
<td>530</td>
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<tr>
<td>FSH0091</td>
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<td>4.5</td>
<td>#4-40</td>
<td>2 mV/V</td>
<td></td>
<td>0.004</td>
<td>930</td>
<td></td>
</tr>
<tr>
<td>FSH0092</td>
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<td>0.004</td>
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<td></td>
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Figure A.3: Load Cell Specifications (page 3/3)
Figure B.1: Effects of Merged Marker Trajectories on Reported Hip Joint Angles (1/2)
For conditions 7-9 (black, red and orange lines), right hip internal rotation angles experience a permanent positive bias. Abduction angles experience a negative bias while the hip is in flexion.
For conditions 7-9 (black, red and orange lines), left hip internal rotation angles experience a permanent negative bias. Abduction angles experience a positive bias while the hip is in flexion.