Biomechanical Comparison of Assistance Strategies Using a Bilateral Robotic Knee Exoskeleton

Dawit Lee, Bailey J. McLain, Inseung Kang, and Aaron J. Young Ph.D.

***Abstract*— Despite there being studies that have investigated the effects of human augmentation using a knee exoskeleton, comparing different assistance schemes on a single knee exoskeleton has not been studied. Using a light-weight, low-profile bilateral knee exoskeleton system, this study examined and compared the biomechanical effects of three common assistance strategies (biological torque, impedance, and proportional myoelectric controllers) exhibiting different levels of flexibility for the user to control the assistance. Nine subjects walked on a 15% gradient incline surface at 1.1 m/s in the three powered conditions and with the exoskeleton unpowered. All the assistance strategies significantly reduced the metabolic cost of the users compared to the unpowered condition by 3.0% on average across strategies (p < 0.05), led by the significant reduction in the biological knee kinetic effort and knee extensor muscle activation (p < 0.05). Between assistance strategies, the metabolic cost and biomechanics displayed no statistically significant differences. The metabolic and biomechanical results indicate that powered extension assistance during early stance can improve performance compared to the unpowered condition. However, the user's ability to control the assistance may not be significant for human augmentation when walking on an inclined surface with a knee exoskeleton.**

***Index Terms*— Wearable Robotics, Sloped Walking, Robotic Exoskeleton, Knee Exoskeleton, Biomechanics**

I. Introduction

The field of powered exoskeleton technology has been actively developed over the past years. Exoskeleton technology holds a large potential to help improve human mobility and physical capability. Therefore, the effectiveness of using exoskeletons has been widely investigated in various applications including augmenting human performance [1, 2], assisting impaired populations [3-5], and therapeutic purposes [6, 7]. The current gold standard measurement for human performance augmentation in the field is metabolic cost [8]. This is achieved by replacing the user's biological joint effort required by muscles to perform a certain locomotor task [6]. For walking, many exoskeletons target the hip and/or ankle joints because these joints contribute to the majority of the total positive mechanical work during level-ground walking [6]. Despite there being fewer studies to date that have targeted the knee joint for human performance augmentation, previous studies have investigated the efficacy of utilizing knee exoskeletons in assisting the user with different locomotor tasks [9-16]. The majority of these studies have focused on one of the following tasks: developing high torque, light-weight and low profile devices [9, 15-17], control strategies for level-ground walking [10], or investigating the effect of the exoskeleton system on dynamic locomotion including level ground and sloped walking [11-14].

During level-ground walking, the knee joint produces a very small amount of positive mechanical work compared to the other lower-limb joints, and this led the majority of the previous exoskeleton research to focus on targeting the ankle or the hip joint. However, the contribution of the knee joint becomes greater during inclined walking, up to approximately 25% of the total positive power [18]. The positive power generation of the knee joint during incline walking is primarily present during the early stance phase through an extension moment [19]. Importantly, the efficiency of muscle is much lower for power generation (positive power) than power absorption (negative power), meaning that humans consume a larger amount internal energy to achieve the same absolute amount of energy for positive power than negative power [20]. Thus, one potentially promising strategy that knee exoskeletons have exploited is to provide early stance phase knee extension support. For instance, our previous work examined providing early stance assistance with a biological torque controller using a unilateral knee exoskeleton during incline walking [12]. Our results indicated that only half of the individuals achieved metabolic reduction with the assistance while the other half did not. The primary biomechanical difference between individuals with (responders) and without (nonresponders) metabolic reduction was that nonresponders exhibited increased muscle activation of the knee extensor group on the unassisted leg when assistance was provided whereas the responders did not. This suggests that assisting the user's knee joints bilaterally could potentially be a better strategy for improving the energetics of the user than assisting unilaterally.

An important question in the field of exoskeletons is what is the most optimal exoskeleton assistance to reduce the user's metabolic cost? Previous works, investigating solutions for exoskeleton control, have introduced a variety of controllers [4]. One of the most common control schemes seen in the field is a biological torque controller in which the assistance profile is based on human biomechanics data [12, 21-23]. With this scheme, the assistance profile stays consistent during every gait cycle. Another common type of controller is an impedance controller which models a joint as

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a spring-damper system [24-26]. An impedance controller regulates the torque output based on the input kinematics from the user, allowing the user to step up or down the torque by controlling their kinematics. Frequently, the mechanical dynamics of the knee joint during the stance phase of walking is modeled as a quasi-stiffness model [27, 28], suggesting that an impedance controller could be employed for an assistance strategy. Lastly, a controller frequently utilized is a myoelectric controller [21, 29, 30]. The myoelectric controller utilizes surface electromyography (EMG) signals from the user as an input. With the use of the biological muscle activation to control the assistance timing and magnitude, the user can adjust the assistance to their gait for step-to-step variability in real-time. Although a myoelectric controller may pose a few notable limitations such as change of EMG signals from muscle fatigue, signal noise, and electrodes losing contacts, it still is one of the most common controllers in the exoskeleton field. Previous studies demonstrated that myoelectric controllers allowed the users to walk more naturally, which led to a larger reduction in muscle activation or metabolic cost compared to the biological torque controllers using hip and ankle exoskeletons [21, 31]. Results indicate that allowing the user more control of the assistance may be more beneficial than allowing less because the powered assistance becomes more synchronized to the user's intention.

Despite there being a variety of controllers, it is not well known how the users' biomechanics compare with different types of controllers while utilizing a knee exoskeleton. Frequently, in a given study, only a single controller is investigated [12, 14]. Thus, to better compare different assistance strategies for a robotic knee exoskeleton, we studied the differences in the biomechanical effectiveness of three common assistance strategies for assisting able-bodied adults during inclined walking using a bilateral knee exoskeleton: a biological torque controller (BT), an impedance controller (IM), and a proportional myoelectric controller (PM). The overall aim of each controller design was to assist the biological effort that is normally exerted around the knee joint by the knee extensor muscle group during the early stance phase.

In this paper, we present the design and the control of a lightweight, low-profile, autonomous bilateral knee exoskeleton that was used to test the biomechanical effects of the three assistance strategies in this study. The actuator of this exoskeleton is half the weight of our previous version, while the torque capability was only reduced by a quarter [12]. Then, using this exoskeleton, we present a biomechanical and metabolic comparison of three control strategies on the knee exoskeleton during incline walking. The first hypothesis of the study was that the bilateral assistance (all 3 controllers) would reduce the metabolic cost of the user compared to the unassisted condition, which is the unpowered condition for this study. The second hypothesis of the study was that among the three assistance strategies, the metabolic reduction compared to the unpowered condition would be the largest under PM, the second largest under IM, and the smallest under BT. This hypothesis is based on the ability of the user to adjust the assistance in real-time. PM provides the user the most control over the assistance, and BT provides the least since the assistance profile is fixed for the controller. IM falls between BT and PM as the assistance profile is adjusted based on the knee joint angle, so the user can adjust the magnitude of assistance but only based on a strict torque-angle relationship.

II. Powered Knee Exoskeleton Design

A. Mechatronic Design

The one degree of freedom robotic knee exoskeleton was designed to assist the flexion and extension of the knee joint (Fig. 1A and 1B). We prioritized the design criteria to minimize the overall weight of the device, which is directly correlated to the metabolic cost penalty due to the added mass to the user's body [32]. Our exoskeleton system employs a quasi-direct drive similar to previous knee exoskeleton systems [15-17]. While our design has comparable performance compared to the previous knee exoskeletons with the quasi-direct drive mechanism, this design is the first bilateral, autonomous, robotic knee exoskeleton with the quasi-direct drive mechanism that had its performance evaluated during human locomotion.

The actuator utilizes a high torque density Brushless DC motor (U8 LITE, T-motor, China) incorporated with a custom made single-stage 6:1 planetary gear. By using a low ratio gear reduction, the overall actuator exhibits actuator dynamics with low friction where the residual interaction torque (e.g., actuator backdriven by the user) between the user and the exoskeleton is minimized. This feature allows the actuator to maintain a high torque bandwidth capability [33]. Additionally, this high efficiency in the gear transmission allows the actuator output torque to be directly correlated to the motor output torque and does not require an additional torque sensing at the output, allowing open-loop torque control. The peak torque of the actuator is 17.4 Nm (Table I), corresponding to 31% of the peak biological moment of a male with 50th percentile body weight in the United States during incline walking [34]. The actuator's maximum angular velocity (while providing peak torque) is 49.9 rad s^{-1} which is greater than the maximum angular velocity of the knee joint during walking [35]. Lastly, the actuator has a range of motion from -20° to 90° in flexion covering a full range of motion of the biological knee joint during walking. Wearing the exoskeleton leads to approximately a 7% increase in the moment of inertia about the knee joint in the sagittal plane during swing phase of a male with 50th percentile body weight and height in the United States [34, 36].

The motor was powered by a 22.2V 3600mAh LiPo battery (Venom Power, USA). A 19-bit resolution absolute magnetic encoder (Orbis, Renishaw, UK) tracked the position of the motor, and the measured angle of the motor was divided by the gear ratio to represent the output angle of the actuator

TABLE I. KNEE EXOSKELETON SPECIFICATION

Max. Cont. Torque (Nm)	7.2
Peak Torque (Nm)	17.4
Max. Speed (rad·s^{-1})	49.9
Actuator Mass (kg)	0.56
Unilateral Exoskeleton Mass (kg)	1.5
Range of Motion (°)	-20° to 90° in flexion

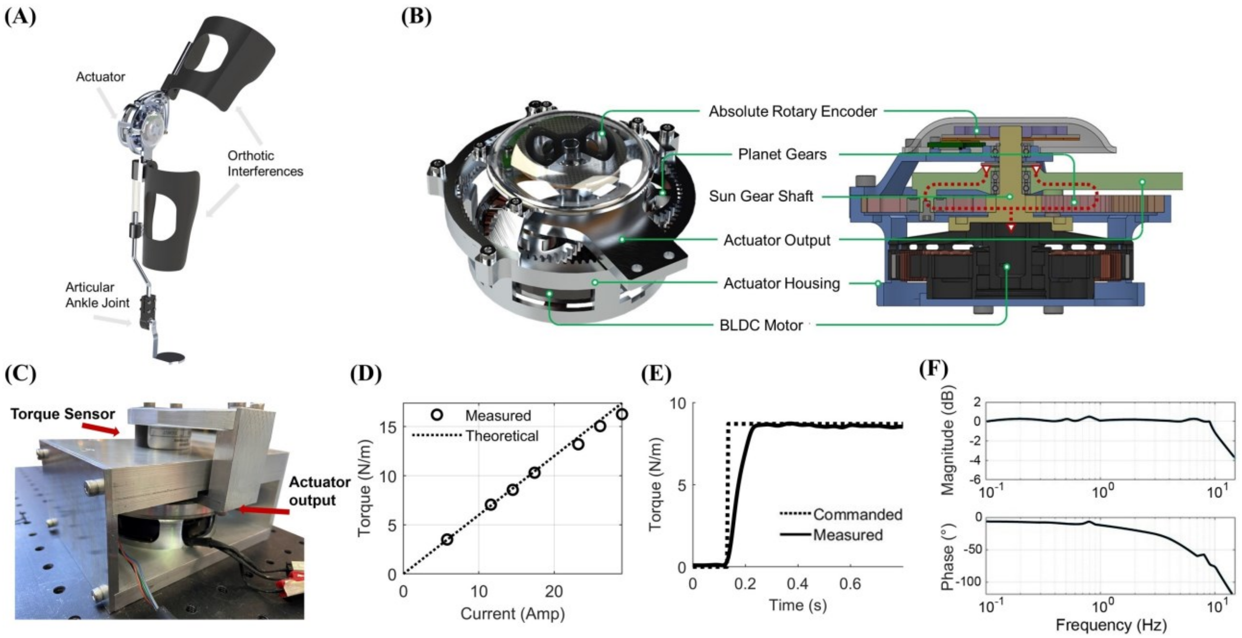

Fig. 1. (A) The full exoskeleton assembly for a unilateral system. The assembly includes the knee and ankle joints where the knee joint is powered by the actuator and the ankle joint is an articular joint. (B) The isometric view of the actuator assembly with the main components labeled. The red dotted lines indicate the load path of the power transmission between the actuator and the user. (C) Configuration of the actuator and a torque sensor for the actuator benchtop testing. (D) Steady-state measured torque values with various current inputs (circles). The dotted line shows the theoretical trend calculated based on the torque constant of the motor. (E) Step response of the actuator (50% of the maximum torque) under benchtop setting (commanded torque – dotted line, measured torque – solid line). (F) The bode plot of the actuator performance under benchtop setting in response to sinusoidal commanded torque profiles.

assembly. The actuator assembly was connected to the thigh and shank orthotic interfaces to allow power transmission between the actuator and the user's leg. The electrical current was controlled (PI closed-loop control) by an open-source VESC motor controller using an analog signal input. A force-sensitive resistor was placed at the user's heel for each leg to detect the heel-contact. All inputs and outputs of sensors and the motor controller were integrated by a custom-made printed circuit board (PCB), and a microprocessor (myRIO 1900, National Instruments, USA) was used to control the device with a control-loop rate updating at 200 Hz.

The characterization of the actuator performance was performed on a benchtop setting (Fig. 1C). During testing, the actuator housing and the end effector were statically mounted on a frame with an external torque sensor (Transducer Techniques, CA) coupled in series. We performed step response testing by commanding various electrical current inputs to validate the linearity of the actuator response. The steady-state result indicated a linear relationship ($R^2 = 0.99$) between the commanded current and the measured torque (Fig. 1D). As shown with an exemplary step response result (Fig. 1E), illustrating a 50% of the actuator torque output (8.7 Nm), the steady-state response of the measured torque yielded approximately 8.56 Nm, indicating about 1.6% deviation from the theoretical value. This slight deviation of steady-state measured torque compared to the theoretical value (while it only occurred in the high torque command region) may have resulted from the motor hysteresis due to thermal loss. To validate this heat loss more systematically, we conducted thermal testing by commanding the maximum torque. Our actuator outputted the peak torque for 34 seconds while the motor winding temperature raised to approximately 145 °C. As the conventional exoskeleton assistance does not require a peak torque to be applied for more than a couple of hundred milliseconds during the gait cycle, our result assured that our actuator output can reliably provide assistance without overheating. Lastly, we conducted torque bandwidth testing using a sinusoidal torque input in various frequencies (ranging from 0.1 ~ 20 Hz). For each sinusoidal input, we set the peak torque with 90% of the actuator's maximum capability. Our test result showed that our actuator's torque bandwidth was 17 Hz which was computed by taking a -3dB magnitude decay point from a generated bode plot (Fig. 1F).

B. Controller Design

1) Biological Torque Controller (BT)

Similar to previous biological torque controllers designed for the hip and the ankle joints, the BT for this study was designed to closely follow the profile of the knee joint moment during the early stance phase of incline walking [23, 31]. The parabolic-shaped assistance profile provided active knee extension assistance for the first 30% of the estimated gait cycle with its peak reaching 30% of the peak biological knee moment occurring during incline walking, about 0.65 Nm/kg (see Fig. 2) [19]. After the assistance phase, the commanded torque was set to 0 Nm for the rest of the gait cycle. The user's gait phase percentage was estimated by dividing the current time since the most recent heel contact by the user's average stride duration. The average stride duration was calculated by averaging the duration of the past five strides and was updated at every heel-contact detected by an FSR attached to the bottom of the user's shoe.

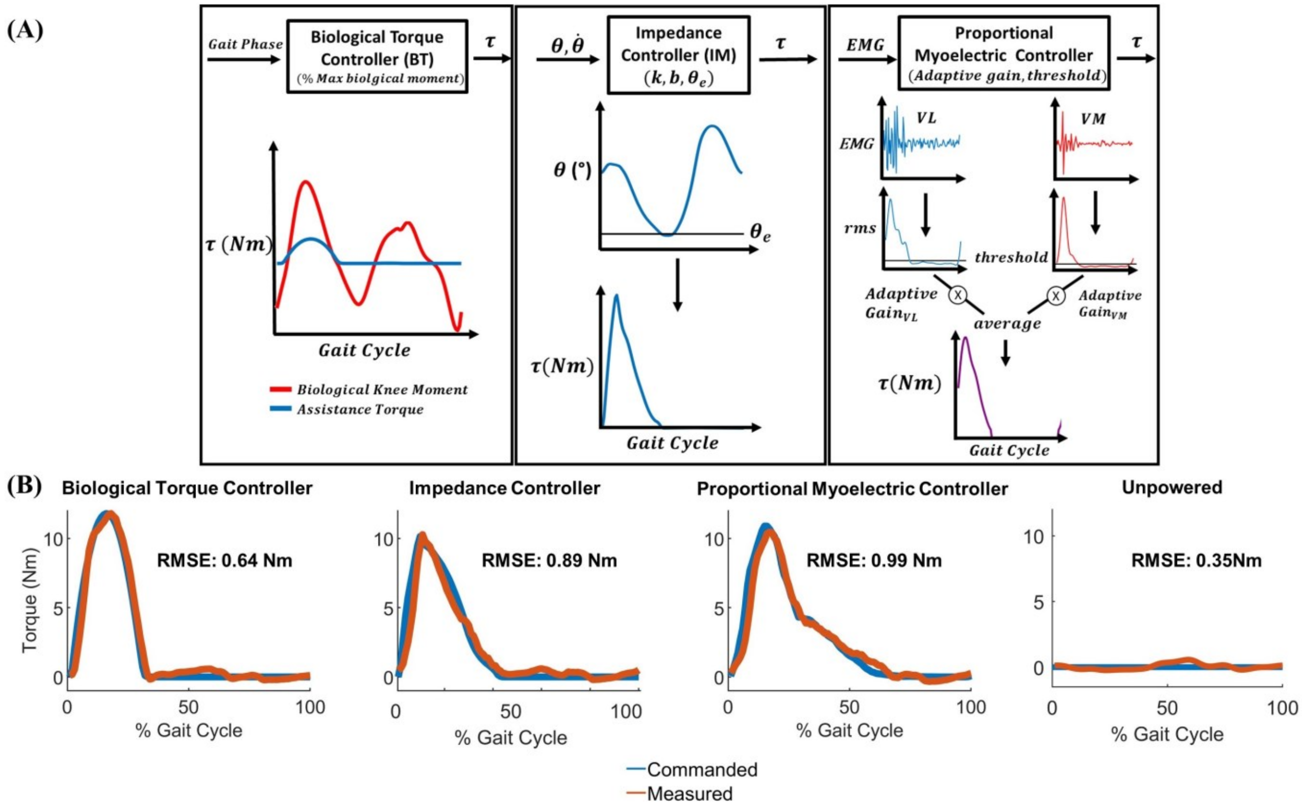

Fig. 2. (A) General diagram of the controllers: BT, IM, and PM. (B) The torque tracking of the actuator for each controller during incline walking for validation. The torque was measured using a torque sensor installed between the actuator output and the orthotic interface. The gains multiplied to the commanded torque were 1.1 for the biological torque controller and 1.15 for the impedance controller and the proportional myoelectric controller. The RMSE between the commanded and measured torques was below 1 Nm for all walking conditions including the unpowered condition for this representative test subject.

2) Impedance Controller (IM)

The impedance controller was designed to provide assistance based on the generic IM method using (1), where τ is the assistance torque, k is the stiffness constant, θ is the measured angle, θ_e is the equilibrium angle, b is the damping coefficient, and $\dot{\theta}$ is the measured velocity [37].

$$\tau = k(\theta - \theta_e) - b\dot{\theta} \quad (1)$$

For the experiment, the assistance was intended to assist positive power generation of the knee joint, therefore, the damping coefficient was set to zero. Assisting the knee joint with a very high damping coefficient should be more suitable for decline walking where the knee joint performs significant negative power during the early stance phase. θ_e remained zero degrees, fully extended knee, for all subjects, and k was modulated for each subject. The assistance was active for the first 40% of the estimated gait cycle, and the way in which the estimated gait cycle was determined for IM was the same as the way used for BT. In order to avoid a sudden spike of the assistance between the assistance and non-assistance phases, the θ_e was ramped from the θ at heel-contact to zero degrees for the first 100ms of the assistance phase.

3) Proportional Myoelectric Controller (PM)

Many studies have developed proportional myoelectric controllers for lower-limb joints [21, 38, 39]. The purpose of the controller is to provide assistance in response to the activation of the muscle. With this, the user is allowed to freely control the timing, duration, and magnitude of the assistance. The design of the controller in this study was based on the adaptive PM designed for the ankle exoskeleton by *Koller et. al.*, which allows a long-term adaptation of the controller to the user's muscle activation [40]. This method allows the controller to achieve a consistent maximum assistance torque over time even if there is a reduction in the muscle activation level due to the user adapting to the exoskeleton assistance. We used a combination of two knee extensor muscles (Vastus Lateralis and Vastus Medialis) to proportionally control the assistance. In real-time, the root-mean-squares (RMS) of raw EMG signals sampled at 1000Hz were calculated with a 100ms moving window to create the linear envelope of each Vastii muscle, and this moving window introduced a slight time delay (~50 ms) between maximum activation of the muscle to the maximum commanded assistance torque. The threshold of muscle activation for each muscle was determined while the subject was standing still in order to avoid triggering assistance when the muscle was minimally active. The commanded assistance torque was determined using (2) and (3).

$$G_i = \frac{1}{N}\left(\sum_{j=i-N}^{i-1} g_j\right) \quad (2)$$

$$\tau = G_i \times (RMS\ EMG - threshold) \quad (3)$$

Firstly, the threshold subtracted maximum point of the linear envelope for every stride, g_i, was stored in a buffer keeping the maximum points of the most recent 25 strides from the current stride ($N = 25$). The maximum points in the buffer were averaged using (2) to estimate the adaptive-gain necessary, G_i, to map the maximum point of the linear envelope to the maximum targeted assistance magnitude. This adaptive gain was updated at every heel-contact and remained consistent throughout each gait cycle. The commanded assistance torques, τ, were computed for the two Vastii muscles separately and averaged to represent the final commanded torque (Fig. 2A).

III. METHODS

A. Experimental protocol

Nine able-bodied subjects [5 female/4 males, 21.6 (mean) ± 3.2 (standard deviation) years, 173.1 ± 7.4 cm, 67.0 ± 5.5 kg] provided written informed consent to participate in the following experiment approved by the Georgia Institute of Technology Institutional Review Board (Protocol #/ approval date: H19167/ May 15th, 2019). The biomechanical effectiveness of different powered knee exoskeleton assistance schemes was tested with the subject walking on a 15% gradient at 1.1 m/s on an instrumented split-belt treadmill (Bertec Corporation, Ohio). The user walked wearing a control box weighing 1.3 kg containing batteries, an EMG measurement unit, a PCB, and a microprocessor.

Each experiment involved two visits. The first visit was for the fitting of the device and training on the three exoskeleton controllers. During this visit, the subject walked with each controller for 15 minutes to adapt to walking with assistance. For BT, the maximum torque was set to assist 30% of the maximum biological knee extension moment during incline walking. For each subject, we collected the average integrated commanded torque per gait cycle during BT and tuned the appropriate stiffness value for IM and the maximum targeted assistance magnitude for PM to match the integrated torque per gait cycle between controllers. The goal for tuning assistance parameters was to have the controllers provide as similar an amount of assistance as possible, so that it does not affect the comparison in biomechanical outcomes. On the second visit, data were collected. Metabolic cost, muscle activity, motion capture data, ground reaction force, and user preference data were collected. Prior to collecting data for each condition, the subject walked for five additional minutes on the controller in order to re-acclimate to the assistance. During data collection, the subjects walked for six minutes in four conditions in the randomized order for each subject: unpowered (UN), BT, IM, and PM. The device was powered-off under UN. To investigate and compare the biomechanical effectiveness of the assistance strategies using the knee exoskeleton, we chose UN as the baseline.

Fig. 3. Experimental setup during incline walking. The orthotic interfaces and the location of the actuator were adjusted to each subject's legs.

B. Data Acquisition and Analysis

1) Metabolic Cost

The metabolic cost of walking was collected using indirect calorimetry (Parvo Medics, UT). The metabolic cost of walking (W/kg) was calculated using the modified Brockway equation based on oxygen consumption and carbon dioxide production [41]. The resting metabolic rate was subtracted from the gross metabolic rate from the walking trials to calculate the net metabolic cost of walking in each walking condition. The metabolic rate of the last two minutes of each walking trial was included in the analysis.

2) Muscle Activity

The muscle activity of six major muscles acting on the knee joint (3 flexors and 3 extensors) of the right leg were collected at 1000 Hz for analysis: vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), biceps femoris (BF), semitendinosus (ST), and lateral gastrocnemius (GA), utilizing surface electromyography (EMG) electrodes (Biometrics Ltd, VA). The raw signal was band-pass filtered between 20 and 400 Hz, full-wave rectified, and low-pass filtered at 6 Hz to create the linear envelope of each muscle channel. Then, the envelope was normalized to the maximum activation of the average linear envelope during UN for each muscle. Finally, the RMS of the linear envelope was calculated for each walking condition and was compared across conditions to assess the differences in muscle activation. We removed one subject's ST and another subject's RF EMG from the analysis because EMG signals had significant motion artifact that the digital filtering technique was unable to completely remove.

3) Biomechanics

Motion capture data (VICON, UK) and the ground reaction force from an instrumented split-belt treadmill were collected. The motion capture data were collected at 200 Hz. We utilized the same lower-limb marker set detailed in our previous study [12], which includes 28 markers. All markers were placed on the subject's body. Motion capture data and ground reaction force data were low pass filtered at 6 Hz before joint kinematics and kinetics were processed in OpenSim. The commanded torque of the device was subtracted from the knee joint moment calculated with Opensim to represent the biological knee joint moment. The last two minutes of biomechanics data were analyzed for the right leg to match the EMG data collected only on the right leg. The maximum biological extension moment and positive power of the hip and the ankle joints during the whole gait cycle were compared between conditions. For the knee joint, since the powered assistance was mostly active during 0-40% of the gait cycle, we performed a specific kinetic analysis during this phase of the gait cycle.

4) Controller

We calculated the average commanded torque for each subject and each controller. Using this result, we evaluated if the average torque was consistent between powered conditions. Additionally, the average peak assistance magnitude across subjects for each controller was calculated and compared between controllers.

5) User Preference

At the end of the study, the user was asked to rank the walking conditions that required the most physical effort (Rank 1 being requiring the least effort and Rank 4 being requiring the most effort).

6) Statistical Analysis

One-way repeated measures ANOVA test with post hoc Bonferroni correction test (alpha = 0.05) was used to determine if quantitative differences exist in walking conditions. The outcome measurement was the dependent variable, and subjects (random) and controller conditions (fixed) were the independent variables. All quantitative data in the result section are presented as mean ± standard error of mean (SEM).

IV. RESULTS

A. Energetics

All assistance conditions had a significantly lower ($p < 0.05$) metabolic cost than the UN condition (Fig. 4). All assistance conditions yielded very similar metabolic reductions from UN (averaged reduction: 3.1% for BT, 2.9% for IM, and 3.1% for PM) with no statistically significant difference between them.

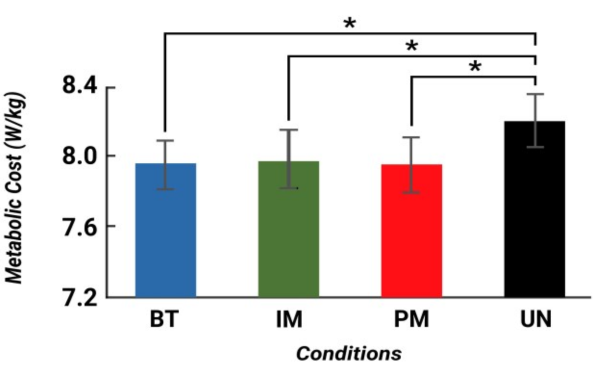

Fig. 4. Average metabolic cost for each assistance condition across subjects (N=9). Average metabolic cost for all assistance conditions are lower compared to the one during the unpowered condition. The error bars show mean ± 1 SEM. The asterisk (*) indicates statistically significant difference ($p < 0.05$).

B. EMG

The primary intention of the knee extension assistance is to replace a portion of the biological effort of the knee extensor muscle group. The result shows that all assistance modes consistently yielded significant reductions in the activation of the major knee extensor muscle group (VL, VM, RF) compared to UN, except for RF during PM (Fig. 5). The

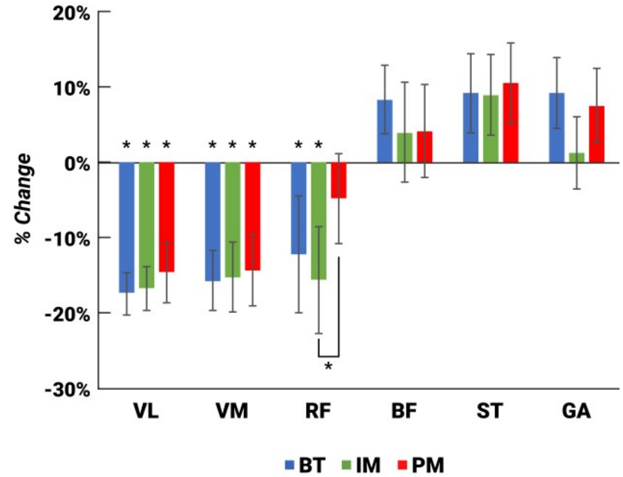

Fig. 5. The percent change of RMS EMG during assistance conditions from the UN condition. Positive/negative values indicate increase/decrease in EMG. The knee extensor muscle group generally exhibited a large reduction in activation with assistance compared to the UN. The asterisk (*) below the bars indicate statistically significant differences between the assistance conditions connected by lines and above the bars indicates statistically significant difference compared to UN ($p < 0.05$). The error bars show mean ± 1 SEM.

averaged percent reduction in muscle activity across assistance conditions was -16.0% for VL and -15.1% for VM. The reduction in the knee extensor was largely present during 0-40% of the gait cycle where the muscle group was mostly active. On average, the primary knee flexors, BF and ST, showed increases with all assistance modes compared to UN with no statistically significant differences. These increases were mostly attributed to increased activation during 0-50% of the gait cycle for some subjects. There were no significant differences in muscle activation level for each muscle between assistance conditions except for that the RF activated less during IM compared to walking with PM.

C. Biomechanics

During all assistance conditions, joint kinematic, moment, and power profiles exhibited very similar trends (Fig. 6). The most notable differences between UN and

Table II. Average peak biological moment (Nm/kg) and power (W/kg) for each joint across subjects. The outcomes with asterisks (*) showed statistically significant changes compared to the outcomes during UN ($p < 0.05$). There was no significant difference exists in the kinetic outcomes between assistance conditions. The outcomes are shown as mean (SEM).

Joint	Variable	UN	BT	IM	PM
Hip	Peak extension moment	1.12 (0.07)	1.08 (0.07)	1.03 (0.07)	1.08 (0.07)
	Peak positive power	1.97 (0.17)	2.11 (0.25)	2.07 (0.20)	2.06 (0.20)
Knee	Peak extension moment	0.96 (0.14)	0.63* (0.12)	0.65* (0.13)	0.65* (0.12)
	Peak positive power	1.54 (0.26)	1.06* (0.22)	1.08* (0.21)	1.07* (0.18)
Ankle	Peak plantarflexion moment	1.34 (0.07)	1.35 (0.07)	1.35 (0.07)	1.37 (0.07)
	Peak positive power	3.30 (0.17)	3.37 (0.18)	3.52 (0.22)	3.59 (0.17)

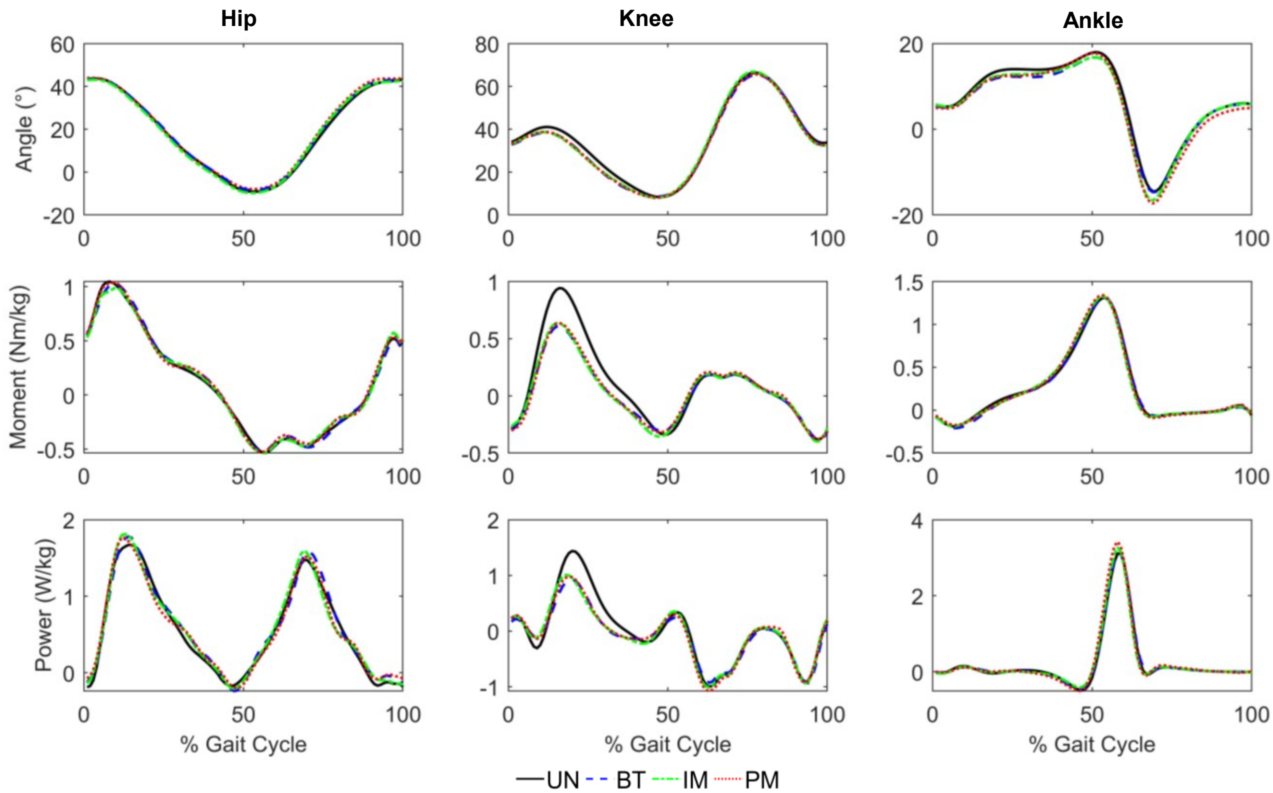

Fig. 6. The average profiles of joint kinematics (top row), biological moment (middle row), and biological power (bottom row). The first column is for the hip joint, second column for the knee joint, and the third column for the ankle joint. Large differences between UN and assistance conditions are present at the knee joint kinetics.

assistance conditions presented at the knee joint. The knee extension assistance during all assistance conditions significantly reduced the peak biological knee extension moment during the early stance phase compared to UN ($p < 0.001$, Table II). The reduction in peak biological extension moment led to a significant reduction in the peak biological positive power of the knee joint during the early stance phase with all assistance conditions compared to UN ($p < 0.01$). The biological kinetics of the hip and the ankle joints were not significantly altered by the assistance compared to UN (see Table II). The knee extension torque assistance caused a slight reduction in the peak knee flexion angle during the early stance phase during the assistance conditions compared to UN by around 2.5°: 41.3 ± 1.6° during UN, 38.7 ± 1.3° during BT ($p = 0.049$), 38.7 ± 1.3° during IM ($p = 0.058$), and 39.0° ± 1.3 during PM ($p = 0.109$). However, the outcome was very consistent between assistance conditions.

D. Controller

All assistance conditions provided active powered assistance primarily during the early stance phase (Fig. 7). The average assistance torque was very consistent and was not significantly different across assistance conditions: 0.0388 ± 0.0001 Nm/kg during BT, 0.0386 ± 0.0008 Nm/kg during IM, and 0.0383 ± 0.0009 Nm/kg during PM. If we look at the assistance profiles within a subject (Fig. 7), the variability of the peak commanded assistance torque was the highest under PM, second highest under IM, and the least under BT. Across subjects, the peak assistance torque during PM was significantly smaller than both BT ($p < 0.02$) and IM ($p < 0.03$): 0.192 ± 0.001 Nm/kg during BT, 0.190 ± 0.008 Nm/kg during IM, and 0.169 ± 0.006 Nm/kg during PM. The assistance during PM and IM remained active longer than the assistance during BT.

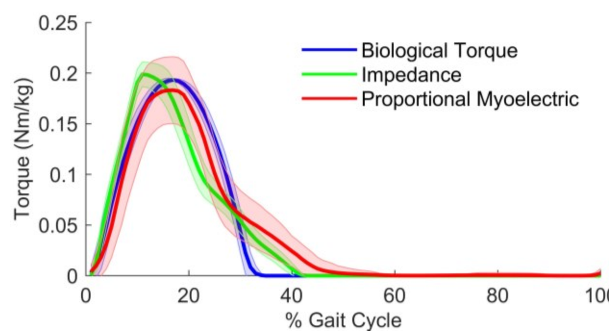

Fig. 7. Average commanded assistance torque profiles of a representative subject for each assistance condition. The shaded region represents ± 1 SD.

E. User Preference

All except one subject ranked UN as the condition that required the most physical effort among the four walking conditions (Fig. 8). All assistance conditions were significantly preferred more than UN in their effort perspective ($p < 0.02$), however, there was no statistically significant difference in the averaged ranks between assistance conditions: 3.9 ± 0.1 for UN, 2.0 ± 0.4 for BT, 1.7 ± 0.3 for IM, and 2.4 ± 0.2 for PM. Even though there was no significant difference in the averaged ranks between

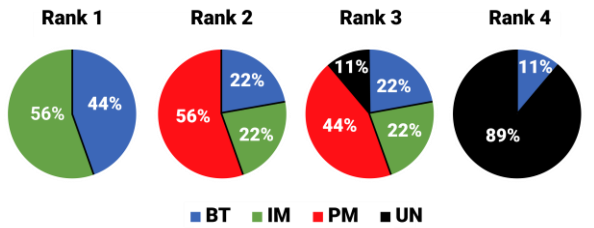

Fig. 8. The distribution of the user preference rank (rank 1 - 4: requiring the least effort – most effort). All subjects preferred BT or IM the most whereas most of the subjects preferred UN the least. All subjects preferred PM as their second or third most preferred condition.

assistance conditions, IM was preferred the most based on the averaged rank data whereas the PM was the least amongst the assistance conditions. All subjects chose BT or IM as their most preferred walking condition, and the PM was chosen the most as the second most preferred walking condition (see Fig. 8).

V. Discussion

All assistance strategies tested in this study significantly reduced the metabolic cost of the users compared to UN (p < 0.05) by approximately 3%. This confirms the first hypothesis, which is that bilateral assistance would reduce the metabolic cost compared to UN. Comparing assistance strategies, the metabolic cost displayed no statistically significant differences between them. This leads to rejecting the second hypothesis, that the metabolic reduction would be the largest under PM, the second largest under IM, and the smallest under BT based on the user's ability to control the assistance in a step-by-step manner. The secondary outcome measures of muscle activity and joint level biomechanics provided insight into the physiological mechanisms for the metabolic changes and corresponded to the metabolic cost changes.

The knee extension torque assistance was effective in significantly reducing the activation of the knee extensor muscle group for all assistance conditions compared to UN. This is similar to how powered assistance at the ankle or the hip joint reduced the activation of the agonist muscle group and metabolic cost of the user [21, 42]. The assistance did not significantly affect the activation of the knee flexor muscle group. However, for a subset of the subject group, we observed increased activation of the knee flexors during the early stance phase of the assistance conditions. This seems to be a result of an antagonistic contraction in response to the knee extension assistance that some subjects exhibited to stabilize and control the knee joint during the exoskeleton assistance, similar to increased activation in the Tibialis muscle with plantarflexion assistance using an ankle exoskeleton [43]. Between assistance conditions, the activation levels of the muscles collected for this experiment showed almost no significant differences. This indicates that the assistance strategy did not significantly affect the neuromechanical behavior of the muscles acting around the assisted joint. In the kinetic perspective, the assistance only significantly affected the knee joint during all assistance conditions. The assistance significantly reduced the peak biological knee extension moment (33.0% reduction on average) and positive power (30.5% reduction on average) during the early stance phase of all assistance conditions compared to UN, offloading a large amount of the biological kinetic effort of the knee joint. This led to a large reduction in the muscle activation of the knee extensor group and a small reduction in overall metabolic cost. Additionally, the subjects preferred the assistance conditions over UN for their subjective measurement of the effort required to complete the walking task. This was primarily because the assistance alleviated heavy activation of the knee extensor group. Corresponding to other outcome measurements, the preference ranking was not significantly different between assistance conditions. Therefore, the secondary outcome metrics of EMG, biomechanics, and user preference all lined up with the metabolic results to support the first hypothesis (assistance improves performance) and reject the second hypothesis (control strategy affects performance). The subjective preference did show that there was no subject who chose PM as the most preferred assistance condition. In particular, this data supports that there is a stronger preference towards consistent controllers during steady- state locomotion such as BT and IM compared to PM. This is likely due to the user's ability to easily learn the controller and that the controller provides consistent torque step-to-step. In cases with significantly more task variability, it may be that PM would be more advantageous by taking into account variability associated with the task which the other control strategies inherently lack.

While many previous studies have investigated the energetic effect of powered assistance targeting the hip or the ankle joint during walking, there are only a few studies that have explored the topic using a knee exoskeleton. During incline walking, assisting the hip or the ankle joint has achieved larger metabolic reductions compared to assisting the knee joint while the peak assistance was not always higher than this study (9.8% reduction compared to the no-exo condition using a hip exoskeleton with 7.5 Nm as the peak torque by *Seo et al.* and 12.2% reduction compared to the unpowered condition using an ankle exoskeleton with 0.35 Nm/kg Nm as the peak torque by *Galle et al.*) [22, 44]. *MacLean et al.*'s study showed a 4.2% metabolic cost reduction with assistance compared to walking without the exoskeleton using a knee exoskeleton, providing up to 60 Nm, during incline walking [14]. The largest metabolic cost reduction compared to UN in this study was with BT, 3.1%. The smaller metabolic reduction seen when assisting the knee joint may indicate that further research should investigate the best way to use a knee exoskeleton to drive down metabolic cost. At this time, fewer studies have investigated assistance strategies at the knee joint compared to the hip and ankle. Also, a separate study can be conducted with an exoskeleton capable of providing assistance at all three joints to fully answer if the knee joint is not an efficient joint to assist for metabolic reduction. With the exoskeleton, the same level of assistance can be separately provided to each joint while the other two joints are not assisted. By doing this, the efficiency of the powered assistance delivered to each joint using a single exoskeleton transfer in reducing the user's metabolic cost can be more transparently compared without other variables such as the hardware, subject, and walking speed.

The second hypothesis (control strategy affects performance) was because the controller, such as PM, can adapt to the user's step-to-step variability and provide more natural assistance, which was the case in the controller comparison study investigating PM and BT using a hip exoskeleton [21]. However, the result of this study is similar to the controller comparison study investigating PM and a mechanically intrinsic controller, equivalent to our BT, using an ankle exoskeleton by *Koller et al*. where the user's metabolic cost was not different between the two controllers [30]. A possible explanation is that the shape of the assistance profiles was generally similar across controllers both in this study and the ankle exoskeleton study, whereas the hip exoskeleton study had a substantial shift in the timing of the peak hip flexion assistance between PM and BT. If the timing of the peak assistance was noticeably different between controllers, the metabolic outcome may have yielded different results as shown with the hip exoskeleton [45].

The EMG results of the study show that there were no major differences in the activation levels of the knee extensors across assistance conditions. Therefore, the level of flexibility in controlling the assistance led to no major differences in how the user's muscle activation adapts. However, using an ankle exoskeleton, the mechanically intrinsic controller, which has a similar behavior with our BT, yielded a larger reduction in muscle activity of the agonistic muscles compared to the PM in *Koller et al*.'s study [30]. The study suggested that this could be since PM requires muscle activation to drive the assistance whereas the mechanically intrinsic controller guarantees consistent assistance regardless of how much the muscle activates, leading the user to ride along with the assistance with the mechanically intrinsic controller. In this study, this was not the case since VL and VM, the muscles used to control PM, exhibited very similar levels of activation across assistance conditions. Also, the peak biological positive power of the knee joint was very consistent across all assistance conditions. Therefore, the adaptation strategy of the user's muscles to the adaptive PM could differ between joints, as results differ between the use of an ankle exoskeleton and a knee exoskeleton.

Additionally, previous exoskeleton studies targeting the hip or the ankle joint showed that the powered assistance could also affect the muscle activation and biological kinetic behaviours of the unassisted joints, the redistribution of energy. However, the EMG and biological kinetic results of this study did not show the redistribution of energy amongst the lower-limb joints by assisting the knee joint [40, 46]. This could be why studies assisting the knee joint have not achieved the levels of metabolic reduction that ankle and hip exoskeletons have where assisting one joint can produce a benefit at another joint.

One limitation of the study is that the hypotheses were tested in a controlled environment at a single slope and speed. The outcome may be different when the exoskeleton is deployed in outdoor or indoor settings which cover a larger array of environmental conditions including speed, slope, and surfaces where the adaptation capability of an exoskeleton is emphasized. In this setting, a controller with greater flexibility for the user to control the assistance, such as PM, could be more suitable for yielding a larger biomechanical benefit. Another limitation is that the biomechanical effects of the assistance strategies were compared to the unpowered condition, not to walking without the exoskeleton.

VI. Conclusion

In conclusion, this paper presented the design of a light-weight, low-profile, bilateral knee exoskeleton system, and provided a detailed biomechanical comparison using the bilateral knee exoskeleton of three common exoskeleton control strategies including biological torque control, impedance control, and proportional myoelectric control. All three controllers yielded a similar level of significant metabolic reduction from the unpowered condition, where the largest reduction was 3.1%. with BT. The metabolic reduction was biomechanically caused by the significant reduction in the biological knee kinetic effort and knee extensor muscle activation. Between assistance conditions, there were no significant differences in trends in EMG, biomechanics, or energetics. This suggests that the user's controllability of the assistance profile at the knee joint is not significant for the human augmentation purpose during steady state walking on an inclined surface.

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