Effects of Assistance During Early Stance Phase Using a Robotic Knee Orthosis on Energetics, Muscle Activity, and Joint Mechanics During Incline and Decline Walking

Dawit Lee, Eun Chan Kwak, Bailey J. McLain, Inseung Kang, and Aaron J. Young

***Abstract*— The knee joint performs a significant amount of positive or negative mechanical work during gradient walking, and targeted assistance during periods of high mechanical work could yield strong human augmentation benefits. This paper explores the biomechanical effects of providing knee extension assistance during the early stance phase of the gait cycle using a powered unilateral knee exoskeleton during gradient walking on able-bodied subjects. Twelve subjects walked on 15% gradient incline and decline surfaces with the exoskeleton providing knee extension assistance during the early stance phase of the gait cycle. For both incline and decline walking, the exoskeleton assistance reduced the muscle activation of the knee extensors on the assisted leg ($p < 0.05$). However, only approximately half the individuals responded to exoskeleton assistance positively by reducing their metabolic cost of walking for both incline and decline tasks. The results indicate that, unlike the individuals who did respond, the individuals who did not respond to the assistance may have penalized their metabolic cost by their biomechanical compensatory behaviors from the unassisted leg.**

***Index Terms*— Wearable robotics, sloped walking, robotic exoskeleton, biomechanics, knee orthosis.**

I. Introduction

OVER the past years, both research and industry fields have actively developed powered lower-limb exoskeletons. These devices can mainly be categorized into three major applications: human performance augmentation, mobility assistance, and rehabilitation [1]. Human performance augmentation utilizes exoskeleton devices to reduce the workload of users working in an industrial or factory setting as well as soldiers walking long distances carrying heavy loads [1]–[3]. Moreover, powered exoskeletons can also be employed to provide partial assistance to both healthy and disabled individuals to enhance their mobility. This assistance can be applied to different locomotion modes such as walking, sit-to-stand, stair ascent/descent, and ramp ascent/descent [4]–[6]. Lastly, exoskeletons are used in therapeutic and rehabilitative purposes for patients with gait deficiencies [7], [8].

One of the major goals in the field of exoskeleton research has been to reduce the human's energetic consumption by partially replacing the biological muscle work required to perform certain tasks [1]. In the case of walking (i.e. level-ground), which is often the primary focus, a majority of these exoskeletons are targeting either the ankle or the hip joint due to their core contribution in providing positive mechanical work [9]. The knee joint creates the smallest portion of the total positive mechanical work among the lower-limb joints in level-ground walking [9], making it the least common lower limb joint to augment for locomotion assistance. While these joint contributions may overshadow the knee joint in level-ground walking, its role becomes greater in the case of sloped walking. For instance, inclined walking is highlighted by a significant amount of positive power generation needed to carry the body's center of mass (COM) forward and upward while fighting against gravity. As the locomotion mode translates from level-ground to incline, the percent contribution of the knee joint to the total positive work of the leg over the stance phase significantly increases while that of the ankle joint decreases [10]. During decline walking, the knee joint mainly functions as a shock-absorber by creating a substantial amount of negative work during the weight-acceptance phase [11]. The trade-off trend between the ankle and the knee joint is also exhibited in terms of negative work between level-ground and decline walking [10]. This increase in the knee joint work is reflected by the increase of the knee extension moment in both incline and decline walking compared to level-ground walking [12]. During incline walking, the peak knee extension moment during the stance phase increases by more than five times compared to level-ground walking [11], and at least two times during decline walking [13].

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This increase in the knee extension moment generation is directly correlated to a substantial increase in the knee extensors muscle activation [12]. This activation eventually leads to an elevation of the patellofemoral compression force, which is more than four times during both walking incline and decline compared to level-ground walking [14]. Therefore, assisting the knee joint using an exoskeleton system offers the potential capability to assist in tasks that are often much more difficult and energetically demanding versus level-ground walking and also reduces the potential risks in damaging the knee joint from increased muscle activity.

A number of previous research works have been involved in developing and investigating the efficacy of using a knee exoskeleton for human augmentation. Many of these research works have also focused on developing robust mechanisms that are capable of adapting to the changes of the user's knee joint center of rotation position during flexion/extension movement [15]–[18]. Additionally, researchers have focused on developing a high torque, light-weight, and low-profile hardware or control strategy mainly for level-ground walking and sit-to-stand locomotion [4], [19], [20]. On the other hand, some researchers investigated and analyzed in terms of human biomechanics and energetics to understand how human interacts with the knee exoskeleton assistance during level-ground locomotion [21], [22]. In general, the list of previous knee exoskeleton research studies showed that, while there were decent progress in the device and controller designs, the field was still lacking in understanding better on the biomechanical effect of using the knee exoskeleton system. Especially, the energetics and biomechanics of using a knee exoskeleton during slope walking have not been explored.

Thus, we investigated the efficacy of using a knee exoskeleton for assisting able-bodied adults during incline and decline walking. In this paper, we present a study in the biomechanical effectiveness of providing knee extension assistance during the early stance phase of the gait cycle using a robotic unilateral knee exoskeleton during incline and decline walking on healthy adults. The first hypothesis of the study was that the extension torque assistance during the early stance phase would reduce knee extensor muscle activation of the assisted leg (AL). The second hypothesis was that the reduction in muscle activity would lead to a reduction in metabolic cost of the users. The hypotheses were based on previous literature that has demonstrated metabolic cost reduction when powered assistance was provided to partially replace the user's joint biological effort [5], [23], [24]. Our study is unique in that it is the first to investigate in depth of understanding the exoskeleton control strategy for human augmentation during slope walking at a muscle and joint biomechanical level to explain the metabolic effect of knee assistance. Our findings will provide valuable information for future exoskeleton designers to further optimize the control strategies to maximize human exoskeleton performance.

II. Powered Knee Exoskeleton Design

A. Mechatronic Design

The unilateral one degree-of-freedom knee exoskeleton (Fig. 1) was designed to partially replace the knee joint's biological moment during different types of walking (Table I). The maximum speed of the actuator matches the maximum speed of the knee joint during level-ground walking corresponding to approximately 5.8 rad$\cdot$ s^{-1} [25]. The peak torque of the device, 22 Nm, can provide about 35% of the peak biological knee moment of an average weight male in the United States during incline walking [26]. The range of motion of the device is -20 to 90° of knee flexion which covers the full ranges of walking. The actuator assembly comprises a brushless DC motor rated for 70W (EC 45 flat, Maxon Motor) coupled with a 113:1 planetary gearhead (GP42, Maxon Motor) and provides an active power assistance to the user. The direction of the actuator output is rotated by 90° with a 1:1 bevel gear (Misumi, Schaumburg) to align the axis of rotation of the exoskeleton output concentric to the axis of rotation of the user's biological knee joint. A 14-bit resolution absolute rotary encoder (Orbis, Renishaw) was attached to the rotational axis of the exoskeleton to measure the angular position of the knee joint. To allow for a closed-loop torque control, a reaction torque sensor (Transducer Techniques) was coupled in between the actuator and the exoskeleton output. A 22.2V, 3600 mAh LiPo battery (Venom Power) powered the exoskeleton and was tethered throughout the experiment. Force sensitive resistors (FSRs) were used to detect the contact between the foot and the ground. The control of the device was all executed through an on-board computer (myRIO 1900, National Instruments). All the external sensors were integrated by a custom printed circuit board.

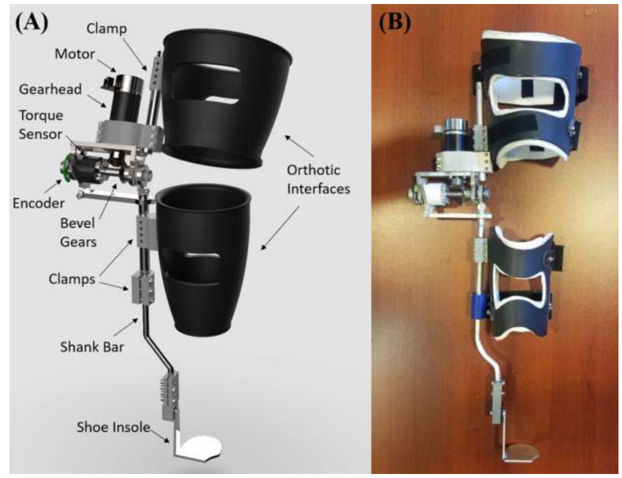

Fig. 1. Powered knee exoskeleton design. (A) CAD drawing of the device is shown with different actuator and sensor components on board. (B) Prototype of the knee exoskeleton. The orthotic shells and leg bar can be adjusted to accommodate different body sizes.

TABLE I
Knee Exoskeleton Specification

Power (W)	70
Max. Cont. Torque (Nm)	11
Peak Torque (Nm)	22
Max. Cont. Speed (rad$\cdot$s^{-1})	4.5
Max. Speed (rad$\cdot$s^{-1})	5.7
Actuator Mass (kg)	1.1
Exoskeleton Mass (kg)	2.7
Range of Motion (°)	110

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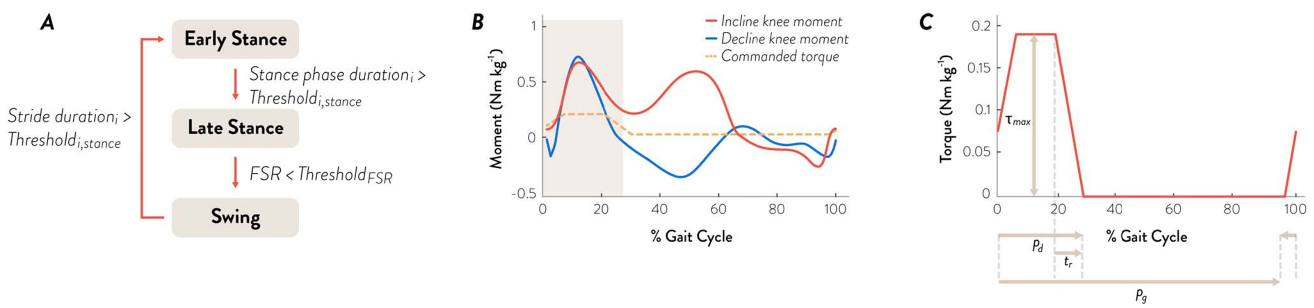

Fig. 2. Control strategy of the knee exoskeleton. (A) Finite-state machine used in the high-level layer for differentiating the gait cycle. (B) An example of knee exoskeleton assistance profile (dotted red). Human biological knee moments for incline (solid blue) and decline (solid red) walking are shown for comparison. (C) Control parameters to generate commanded assistance torque profile. Assistance timing, magnitude, and duration can be modulated.

B. Controller Design

1) High-Level Control: The high-level control uses a finite-state machine (FSM) to determine which phase of the gait cycle the user is walking in at a given time. The FSM divides a gait cycle into three different sub phases: early stance phase, late stance phase, and swing phase (Fig. 2A). Torque assistance is only active during the early stance phase. The transition from early stance phase to late stance phase occurs when the current stance phase duration, $Stancephaseduration_i$, exceeds the $Threshold_{i,stance}$. $Threshold_{i,stance}$, a value that determines the assistance duration, is defined as the product of average stance phase duration from previous five gait cycles and a constant ranging from 0 to 1. The transition between late stance to swing phase is determined by the toe off detection using the FSRs placed on the bottom of the user's shoe. The voltage readings from the FSR below the $Threshold_{FSR}$ indicates that the user's foot is no longer in contact with the ground. The transition from swing phase to early stance phase occurs when the current stride duration, $Strideduration_i$, exceeds the $Threshold_{i,stride}$. $Threshold_{i,stride}$ is the product of average stride duration from previous five gait cycles and a tunable constant, ranging from 0 to 1, which is determined experimentally based on the user's preference.

2) Mid-Level Control: Once the user's walking phase is determined, the mid-level controller determines the commanded assistance torque profile. During the gait cycle when the assistance is not provided, the exoskeleton is set to zero impedance mode, where the commanded torque is set to 0 Nm to cancel out any interaction torque (generated from the gear train) between the exoskeleton and user. This mode ensures that the device feels transparent to the user's intended movement. During the assistance mode (Fig. 2B), an extension torque assistance during early stance phase is provided. The commanded assistance torque profile during the assistance mode is determined by three different control parameters: p_g, τ_{max}, and p_d, which are all tunable (Fig. 2C). p_g is the assistance onset timing (percent gait cycle) and also corresponds to the start of early stance phase. For this parameter to be applied, the current gait phase percentage is estimated using the average stride duration from past five gait cycles. When the current gait phase passes p_g, the actuator starts commanding active torque assistance for the duration of p_d. The commanded torque assistance magnitude ranges from 0 Nm to the peak desired torque, τ_{max}, while linearly increases and decreases during the first and last part of assistance with a duration of t_r to smoothly provide the assistance to the user. For all trials, t_r remained constant with 110 ms.

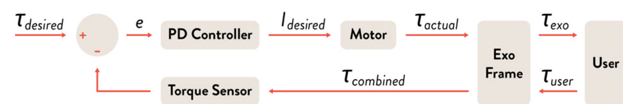

Fig. 3. Schematic of a closed-loop PD torque control. This control loop is necessary to match the $\tau_{desired}$ to the $\tau_{combined}$.

3) Low-Level Control: When the assistance torque profile is commanded to the low-level controller as the desired torque, $\tau_{desired}$, from the mid-level controller, the low-level controller aims to match $\tau_{desired}$ and the actual interactive torque, $\tau_{combined}$ (Fig. 3). $\tau_{combined}$ is a combination of the torque outputted by the actuator, τ_{exo}, and the user's biological torque input to the exoskeleton, τ_{user}, measured by the torque sensor. The error is computed by subtracting the $\tau_{combined}$ from $\tau_{desired}$. This error is inputted into the PD controller, which outputs the desired electrical current output, $I_{desired}$. Using the electrical current, the motor produces τ_{exo}. Then, $\tau_{combined}$ is measured and the error is calculated, then commanded into the PD controller for a closed-loop control. All control loops are updated at 200 Hz.

III. Methods

A. Experimental Protocol

The study was approved by the Georgia Institute of Technology Institutional Review Board, and informed written consent was obtained for all subjects. The biomechanical effectiveness of powered knee extension assistance during early stance phase in walking on a 15% gradient incline and decline at 0.7 m·s^{-1} conditions were tested. 15% gradient surface was often chosen for testing human walking in sloped condition [5], [10], [27]. Additionally, slopes greater than a 15% gradient surface is considered "very steep." [5] Therefore, we chose a 15% gradient surface as the slope level of walking for this experiment. Twelve healthy adult subjects (9 males/3 females, mean ± standard deviation, age: 22.6 ± 2.7, height: 1.73 ± 0.10 m, mass: 68.9 ± 9.9 kg) enrolled in this study. The experiment consisted of two visits. The first visit was

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designated for fitting and training with the exoskeleton. During this visit, the control parameters in the mid-level controller including p_g, τ_{max}, and p_d were tuned based on the subject's feedback with the goal of maximizing the effectiveness of the assistance for each subject. During the tuning process, the baseline values of p_g, τ_{max}, and p_d were 100%, 5 Nm, and 30%, respectively. Each step of change for p_g, τ_{max}, and p_d were ±2%, ±0.5 Nm, and ±2%, respectively. p_g was tuned first with τ_{max}, and p_d fixed at their baseline values. τ_{max} was tuned second, followed by p_d. p_g was based on the estimated percent gait cycle during the previous gait cycle and was tuned based on whether the onset timing is too early or too late. If p_g was earlier than 100% estimated percent gait, the onset timing of the assistance was initiated earlier than the estimated start of the following gait cycle. τ_{max} was tuned based on whether the magnitude of the torque assistance was too low or too high, and p_d was tuned based on whether the assistance duration was too short or too long. After all three control parameters were tuned, the subject tried ±2%, ±0.5 Nm, and ±2% of p_g, τ_{max}, and p_d, respectively, from the tuned control parameters and the final assistance parameters that assisted their walking the most were decided. Then, the subjects were allowed to walk with the device until they felt comfortable with the assistance (~20 minutes in each condition). Data collection occurred during the second visit (Fig. 4A). The subject's metabolic cost was collected using an indirect calorimetry (Parvo Medics, UT). The muscle activity was collected using surface electromyography (EMG) electrodes (Biometrics Ltd, VA). The motion capture data (VICON, UK) and ground-reaction force from an instrumented split-belt treadmill (Bertec Corporation, Ohio) were recorded. Before beginning the first walking trial, the resting metabolic cost of the subject wearing the exoskeleton was measured for six minutes. Prior to the start of data collection for walking trials, subjects practiced walking with both assistance mode and zero impedance to ensure that they felt comfortable walking with the device (~3 minutes in each condition). Subjects performed a six-minute walk test for each of four conditions: assistance mode and zero impedance mode for both incline and decline walking. Additionally, the subjects were asked whether the assistance provided during the assistance mode helped them walk with less effort. In order to investigate the biomechanical effectiveness of the knee extension assistance during the early stance phase, data analysis was done by comparing the data between the assistance mode and the zero impedance mode.

B. Data Acquisition and Analysis

1) Metabolic Cost and User Preference: The measured volume of oxygen uptake and carbon dioxide production were used to estimate the metabolic cost of the user using the modified Brockway equation [28]. The net metabolic cost of walking was calculated by subtracting the resting metabolic cost from the walking metabolic cost for each walking condition. The net metabolic cost for the last two minutes of the six-minute trial for each walking conditions was averaged to represent the subject's net metabolic cost for each walking mode. After finishing walking on each slope, the subjects were asked whether the assistance provided during the assistance mode helped them walk with less effort. The users provided their subjective answer to the question in a scale from 1 to 10 where 1 is strongly disagree and 10 is strongly agree.

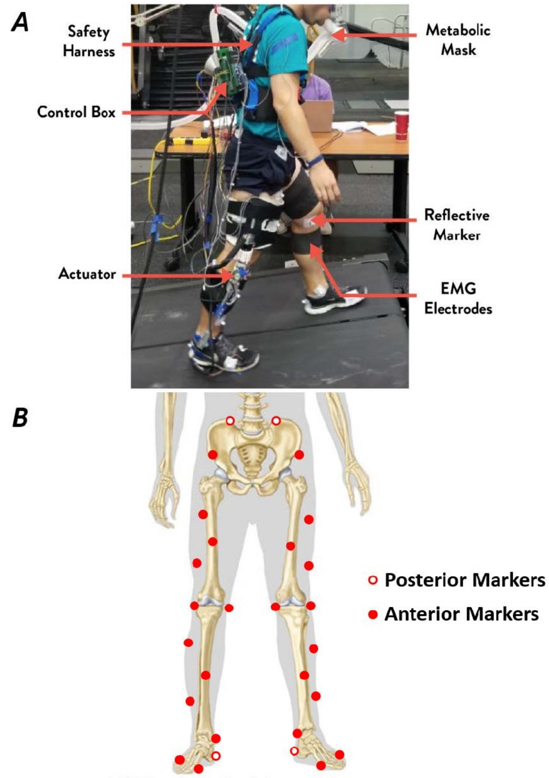

Fig. 4. (A) Experimental setup during incline walking. Metabolic mask, reflective markers, and EMG sensors were used for data collection. Subjects wore the control box on their back. (B) Demonstration of reflective marker placement on the user's body for motion capture. 28 markers were placed in total. Red-filled dots indicate the markers placed anteriorly and white-filled dots indicate the markers placed posteriorly.

2) Muscle Activity: The muscle activities were collected from six different major muscles acting on the knee joint of the assisted leg (AL) and unassisted leg (UAL): vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), biceps femoris (BF), semitendinosus (ST) and lateral gastrocnemius (GA). The EMG signal was sampled at 1000 Hz. The raw EMG signal was band-pass filtered between 20 to 400 Hz, full-wave rectified, and, low-pass filtered at 6 Hz to smooth the signal. Root mean square average (RMSA) of each of the muscle's last two minutes of the processed EMG was compared between the assistance mode and the zero impedance mode for data analysis. Three subjects' EMG on the UAL were not collected, and they were excluded for the EMG analysis for the UAL.

3) Biomechanics: The marker set includes 28 markers as shown in Fig. 4B. The markers are located in the following locations bilaterally: anterior superior iliac spine, posterior superior iliac spine, three markers on the thigh segment, medial knee joint, lateral knee joint, three markers on the shank segment, medial malleolus, lateral metatarsal phalangeal joint, medial metatarsal phalangeal joint, and the posterior

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calcaneus. The markers on the thigh and shank segments were placed around the knee exoskeleton orthotic on the right leg, then mirrored on the left leg. For two subjects with short leg lengths, one of the thigh markers on the AL was placed on the thigh orthotic, and the rest of the markers were placed on the user's body. This marker set was created as a marker template in Vicon Nexus software to allow for data capture, labeling, and the exporting of trajectory. Post-processing for kinematics and kinetics data was done via OpenSim. Starting with the Gait2392 model available in OpenSim [30], the model was modified to match the marker set created in Vicon. The marker trajectory data were used to calculate the joint angles through inverse kinematics, then was combined with the ground reaction force data from the instrumented treadmill to calculate the sagittal-plane inverse dynamics. The biomechanics data from the last two minutes of each of six-minute walking trials were analyzed. The heel-contact and toe-off were detected by comparing the ground reaction force in the vertical direction with setting 40 N as a threshold. The kinetic contribution of the exoskeleton to the knee joint on AL was subtracted from the output kinetic data of the knee joint on the AL from inverse dynamics analysis to represent the biological effort that the knee joint produced. The biomechanics data were collected on eight subjects. One subject's biomechanics for incline walking was excluded due to poor marker tracking performance during post-processing.

4) Multi-Linear Regression Muscle to Metabolic Cost Model: In both incline and decline walking, multi-linear regression models were created to estimate the changes in the user's metabolic cost based on multi-channel EMG data. A forward selection search was used to add up to three muscles that contributed to the estimation with statistical significance. The three subjects without the EMG on the UAL were not included for this analysis.

5) Statistical Analysis: A paired t-test was used to test the significance of the difference in quantities including muscle activity and power generation/absorption between the assistance mode and the zero impedance mode with a $p < 0.05$ criterion. A regression analysis was run between the percent change in metabolic cost and the user preference for incline walking and decline walking separately.

IV. Results

A. Commanded Assistance Torque Profile

1) Incline Walking: The mean τ_{max} during incline assistance mode was 10.0 $\pm$ 0.5 Nm, across all subjects (Table II). p_d averaged 43.2 $\pm$ 1.7 % of the estimated stance phase duration. Most subjects preferred p_g near heel-contact during inclined assistance mode. The RMS error of torque tracking of a representative torque tracking was smaller than 1.3 Nm for both assistance mode and zero impedance mode (Fig. 5).

2) Decline Walking: The average τ_{max} during decline assistance mode across all subjects was 8.5 $\pm$ 0.9 Nm, which was about 1.5 Nm lower than the average τ_{max} during the incline assistance mode. Both the average p_d and p_g during decline assistance mode were fairly consistent with the average p_d and p_g during incline assistance mode. The RMS error of a representative torque tracking was smaller than 1.5 Nm for both assistance mode and zero impedance mode as shown in Fig. 5.

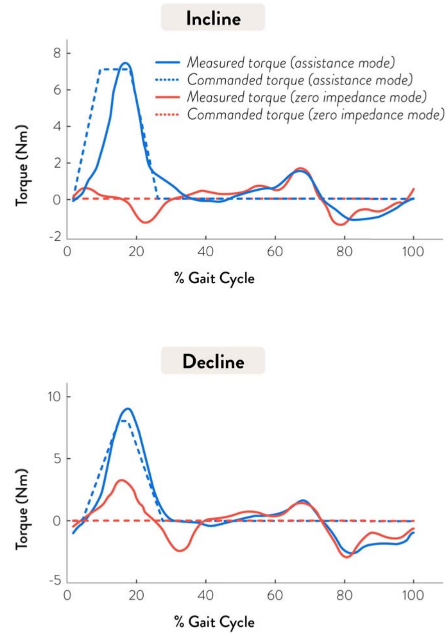

Fig. 5. A representative torque tracking during assistance mode and zero impedance mode for incline walking (top) and decline walking (bottom). The root mean square (RMS) error of the torque tracking was 1.21 Nm during incline assistance mode, 0.69 Nm during incline zero impedance mode, 1.27 Nm during decline assistance mode, and 1.43 Nm during decline zero impedance mode.

TABLE II
Assistance Parameters Across Different Modes

	τ_{max} (Nm)	p_d (%)	p_g (%)
Incline all subjects	10.0 (±0.5)	43.2 (±1.7)	99.8 (±0.9)
Decline all subjects	8.5 (±0.9)	42.3 (±1.2)	100.8 (±0.6)

Maximum torque, assistance duration, and assistance onset timing are presented as τ_{max}, p_d, and p_g respectively (mean $\pm$ SEM).

B. User Preference & Metabolic Cost

In the post-hoc analysis, the metabolic results have shown that approximately half of the subjects yielded net metabolic reduction greater than 1.0% with assistance in both incline and decline walking (Table III). For the purposes of further analyses, these subjects were considered as responders. Subjects not displaying a metabolic reduction greater than 1.0% were considered nonresponders. Consequently, we sought to identify the biomechanical cause(s) of the inconsistency in the presence of the metabolic reduction. For this purpose, each type of the following data was presented both as the overall trend across all subject as well as a comparison between the responders and nonresponders.

1) Incline Walking: During incline walking, subjects generally indicated that they were using less effort during the assistance mode compared to zero impedance mode based on their subjective answers to a question. For incline walking, none of the subjects answered below 5 (not agree or disagree). However, the user preference did not correspond to the percent

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TABLE III
CHANGE IN METABOLIC COST WITH ASSISTANCE AND USER PREFERENCE ACROSS SUBJECTS

	Incline walking		Decline walking	
Subject Identifier	Δ metabolic cost (%)	User preference	Δ metabolic cost (%)	User preference
1	**-6.4**	**8**	**-7.6**	**7**
2	**-4.5**	**8**	10.1	8
3	-0.1	10	**-1.1**	**10**
4	1.3	8	**-4.3**	**7**
5	4.4	6	-0.7	4
6	**-3.1**	**9**	3.5	9
7	**-3.5**	**9**	**-19.3**	**8**
8	**-2.6**	**10**	**-3.1**	**2**
9	0.4	8	18.9	7
10	**-1.6**	**8**	**-7.0**	**8**
11	3.7	10	**-10.3**	**10**
12	5.1	8	22.6	9
All subjects	-0.6	8.5	0.1	7.4
Responders	**-3.6**	**8.7**	**-7.5**	**7.4**
Nonresponders	2.5	8.3	10.9	7.4

The percent changes in metabolic cost in the assistance mode from zero impedance mode and user preferences in both incline and decline walking. Each subject was asked "Did the assistance help you walk with less effort? (1 – strongly disagree, 10 – strongly agree)", and the answer is presented in the bracket next to the percent change in metabolic cost. The responders' metabolic cost changes are bolded. Average across all subjects, responders, nonreponders are presented on the last three rows.

changes in metabolic cost with assistance during incline walking ($p = 0.51$). Six of the twelve subjects were responders during incline walking with an average metabolic reduction of 3.6%. The incline walking nonresponders exhibited on average of 2.5% increase in their metabolic cost. Both responders and nonresponders indicated that the assistance helped them walk with less effort (responders mean: 8.7, nonresponders mean: 8.3).

2) Decline Walking: During decline walking, two subjects, one responder and one nonresponder, disagreed with the statement that it required less effort to walk in the assistance mode compared to zero impedance mode. The responder commented that his/her hip joint on the AL had to "work more" with assistance. The nonresponder who disagreed with the statement simply stated that the assistance mode "took more effort." Similar to incline walking, the user preference did not correspond to the percent changes in metabolic cost with assistance during decline walking ($p = 0.85$). Seven of the twelve subjects were responders with an average metabolic cost reduction of 7.5% (Table II). Five subjects were nonresponders with an average metabolic cost increase of 10.9%. No difference was observed between the responders and nonresponders group's answer to the user preference question (responders mean: 7.4, nonresponders mean: 7.4) for decline walking.

C. Muscle Activity

1) Incline Walking: Across all subjects in incline walking, the muscles that exhibited statistically significant changes in activation with assistance were the knee extensors on the AL and the ankle planterflexor of the UAL (Fig. 6): VL 10.5 (mean) ± 3.8 (SEM) % reduction ($p < 0.05$), RF 9.6 ± 2.1% reduction ($p < 0.001$), VM 10.9 ± 2.1% reduction ($p < 0.001$), GA 5.6 ± 2.3% increase ($p < 0.05$).

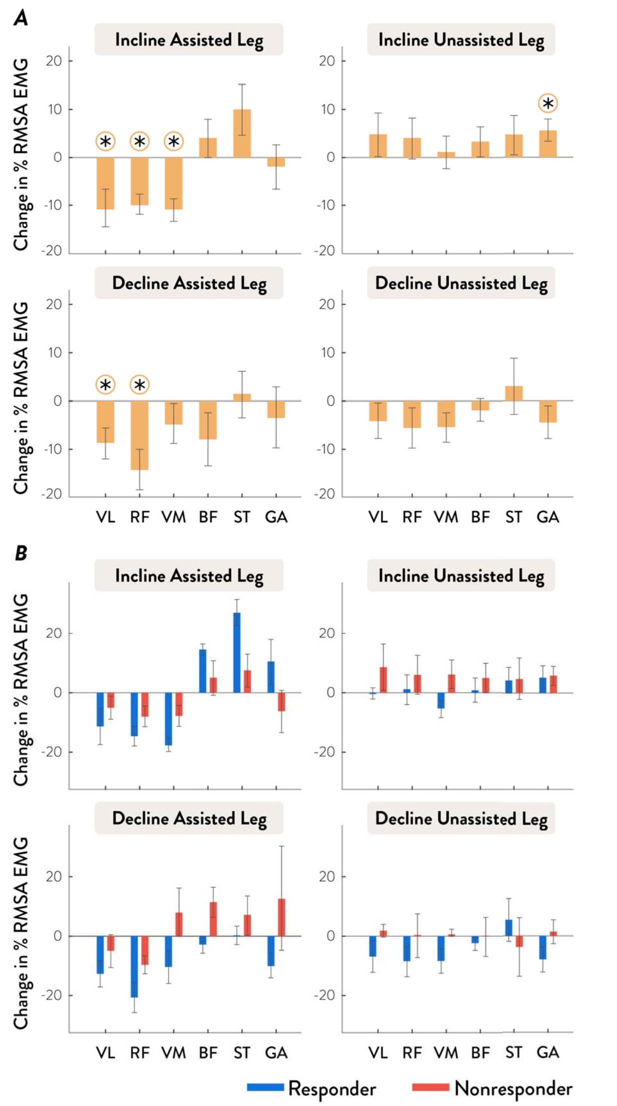

Fig. 6. Comparison in the average percent RMSA reduction RMS EMG in both AL (left column) and UAL (right column) during the assistance mode compared to the zero impedance mode in incline (top row) and decline walking (bottom row) for (A) across all subjects (B) responders (blue) versus nonresponders (red). The asterisks (*) indicate statistically significant changes across all subjects. The error bars show ±1 SEM.

A reduction in muscle activation of the knee extensor group on the AL was exhibited consistently in both the responders and nonresponders during incline walking. However, this reduction was much greater for the responders' than the nonresponders' by almost a factor of two for each of the knee extensor muscles. However, hamstring muscle group activation on the AL also increased with the assistance for both responders and nonresponders. Especially, the activation of the responders' ST on the AL, which increased by 27.1% with the assistance while the nonresponders averaged a 7.5% increase. The changes in muscle activation on the UAL showed a larger difference between the responders and nonresponders. Notably, the nonresponders muscle activation increased for all of the UAL's knee extensor muscles: VL 8.7%, RF 6.1%, and VM 6.2%. However, the responders did not show a consistent increase of muscle activation in the knee extensor group. Both

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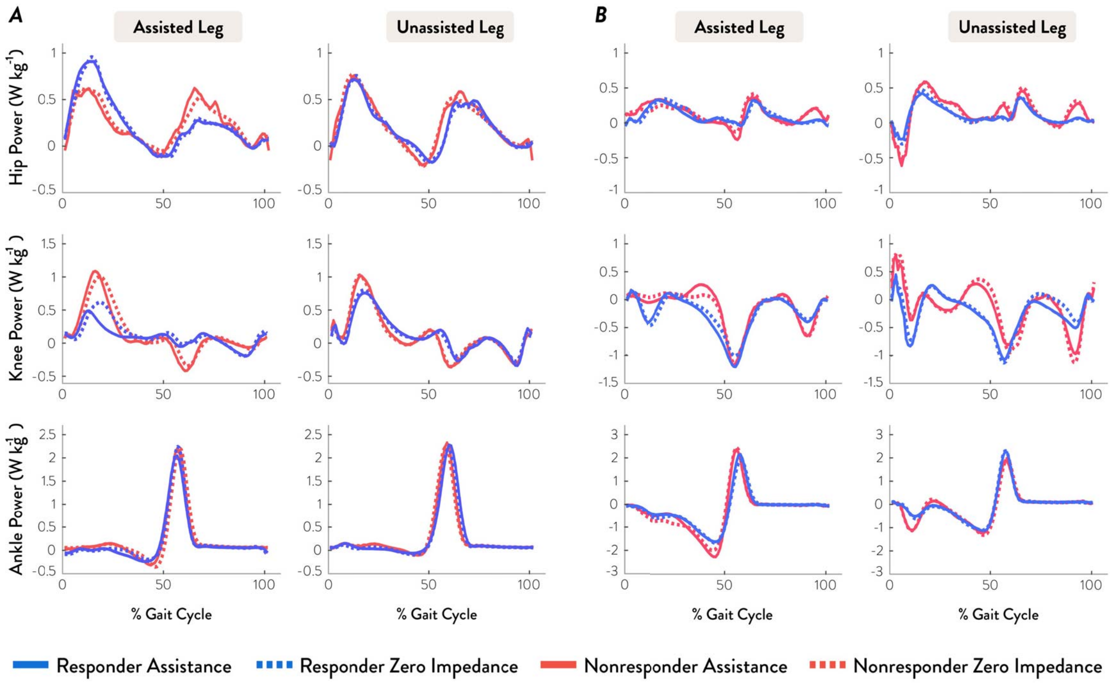

Fig. 7. The biological power profile of hip, knee, and ankle joints in the sagittal plane of the responders (blue) and the nonresponders (red) during (A) incline and (B) decline walking. The solid line represents joint power during the assistance mode and the dotted line represents the joint power during zero impedance mode.

responders and nonresponders increased the ST activation on the UAL.

2) Decline Walking: Across subjects, the assistance in decline walking significantly reduced activation in most of the knee extensors: VL 8.6 ± 3.8% ($p < 0.05$) and RF 14.1 ± 4.3% ($p < 0.01$). In contrast to the consistent increase of muscle activation on the UAL during incline walking, most of the muscles on the UAL during decline walking reduced their activation (no statistical significance).

Similar to the difference in the changes of knee extensor muscle activation of the AL between the responders and nonresponders during incline walking, both responders and nonresponders exhibited a reduction of the AL's VL and RF activation during decline walking. However, the reduction of responders' (VL 12.5% and RF 20.4%) was larger than that of the nonresponders' (VL 4.8% and RF 10.8%). The nonresponders' VM on the AL increased activation by 8.3% with assistance, whereas the responders' reduced by 10.0%. The nonresponders increased the activation of the hamstring muscle group on the UAL (BF 11.7% increase, ST 7.3% increase) whereas the responders reduced the activation of the BF by 2.4% and increased the activation of the ST by 0.5% on the UAL. During decline walking, the responders generally reduced muscle activation of the knee extensors of the UAL (VL 6.7%, RF 8.1%, and VM 8.0%). On the other hand, the nonresponders presented a slight increase in muscle activation in all of the knee extensors (VL 2.1%, RF 0.5%, and VM 0.8%).

D. Biomechanics

1) Incline Walking: Across all subjects, the positive power generation of the knee joint on the AL (Fig. 7), decreased by 18.4 ± 10.5% with the assistance compared to zero impedance mode. This reduction was not statistically significant ($p = 0.13$), because one subject exhibited a 34.9% increase, while the rest of the subjects consistently exhibited reduction. Additionally, the total biological positive power generation across all lower-limb joints (hip, knee, and ankle) of the AL was reduced by 4.6 ± 2.2% across the all subjects without significance ($p = 0.08$).

The responders exhibited a reduction of total biological positive power on both legs combined by 2.6% with assistance (Fig. 8). On the other hand, the nonresponders showed an increase in total positive power by 2.3%. The assistance substantially reduced the positive power generation of the knee joint on the AL by 0.038 W/kg per stride, which is equivalent to an average of 33.5% reduction compared to zero impedance mode for the responders. On the other hand, the nonresponders minimally reduced the positive power generation of the knee joint on the AL by 0.004 W/kg with the assistance. Interestingly, during the assistance mode, the responders decreased the biological positive power generation of the knee joint on the UAL by 3.2% compared to zero impedance mode. However, in parallel to the increase of the UAL's knee extensors EMG, the nonresponders increased the biological positive power of the UAL's knee joint by 14.0% on average.

2) Decline Walking: The extension torque assistance was primarily active during the first 25 percent of the gait cycle. Over the first 25 percent of gait cycle during zero impedance mode, both the responders and nonresponders were absorbing a similar level of total power using all joints on the AL (-0.55 W/kg for nonresponders and -0.52 W/kg for responders). However, the contribution of each joint to the total negative power was largely different between the responders and nonresponders. During the first 25 percent of the gait cycle in the zero impedance mode, the total negative power

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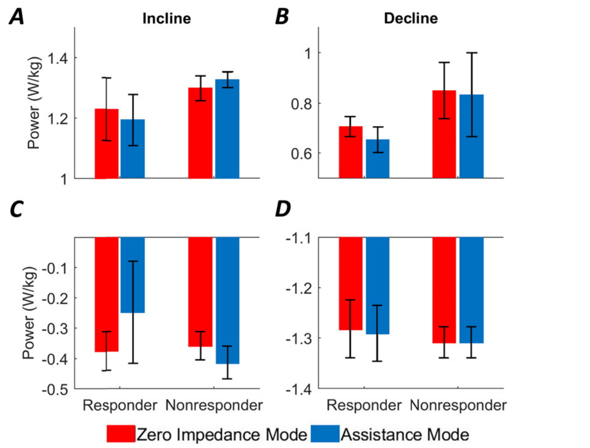

Fig. 8. The averaged total biological positive (top row) and negative (bottom row) power of both legs (ankle, knee, and hip combined) of the responders and nonresponders during the assistance mode (blue) and the zero impedance mode (red) for incline (A, C) and decline (B, D) walking. The error bars show ±1 SEM.

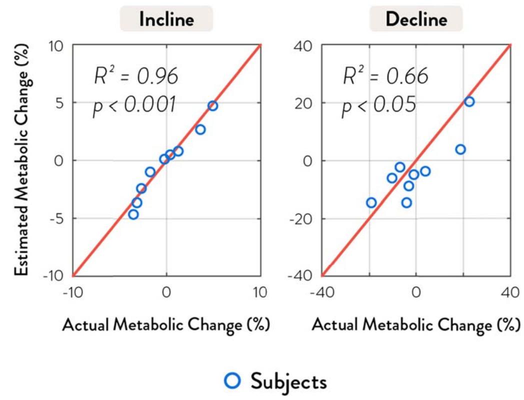

Fig. 9. Multi-linear regression models for estimating the metabolic change using the multi-channel RMSA EMG with assistance for incline (left) and decline (right) walking. For incline walking, ST on the AL and VM and ST on the UAL were used to estimate the metabolic change. For decline walking, VM and GA on the AL and VM on the UAL were used to estimate the metabolic change.

absorption from all lower-limb joints including the power input of the exoskeleton on the nonresponders' AL was heavily relying on the ankle joint. In this case, the ankle joint solely made 95.1% of the total negative power absorption of the AL including the power input of the exoskeleton. Consequently, the contribution of the nonresponders' knee joint to the total power absorption over the phase was minimal, resulting in only a 4.6% contribution to the total negative power absorption of the AL, including the power input of the exoskeleton. On the other hand, the responders utilized their knee joint more by absorbing 31.5% of the total negative power on the AL over the first 25 percent of the gait cycle. During the assistance mode in decline walking, both the responders and nonresponders increased the percent contribution of the knee joint to the total negative power of the AL compared to zero impedance mode.

E. Multi-Linear Regression

1) Incline Walking: The multi-linear regression model (Fig. 9) using the change in RMSA of VM and ST on the UAL and ST on the AL was able to explain approximately 96% of the variance in the metabolic change with assistance from the zero impedance mode ($R^2 = 0.96$, $p < 0.001$). The model indicated the change of metabolic cost was negatively related to the activation level of the ST on the AL and positively related to the activation level of the VM and ST on the UAL (weights: ST < VM).

2) Decline Walking: Three different muscle channels were used in the regression model for decline walking: VM and GA on the AL and VM on the UAL. The model has shown that all three muscle channels used were positively related to the metabolic change (weights: VM on the AL < GA on the AL < VM on the UAL). The model was able to explain 66% of the variance in the metabolic change with assistance from the zero impedance mode ($R^2 = 0.66$, $p < 0.05$).

V. Discussion

Across all subjects, the knee extension torque assistance during both incline and decline walking was effective in producing a significant reduction in the activation of the knee extensor muscles on the AL. Therefore, during both incline and decline walking, the first hypothesis that the knee extension assistance during the early stance phase would reduce the knee extensor muscle activation of the AL is accepted. This reduction is similar to how previous hip and ankle exoskeletons were able to reduce the activation of the muscles that normally create the type of motion by providing torque assistance at a given joint [5], [6], [23], [24]. Also, as previous research work has shown that the reduction of EMG activation is linked to the reduction of joint mechanical effort [23], the reduction of the knee extensor muscle activation is directly correlated to the reduction of the biological positive power generation of the knee joint on the AL during incline walking.

However, even with the reduction in the knee extensor muscle activation, the assistance did not yield metabolic reduction across subjects during both incline and decline walking. Therefore, the second hypothesis that the reduction in muscle activity would lead to a reduction in metabolic cost of the users is rejected for both incline and decline walking. We found that the multi-linear regression model using multi-channel EMG help to predict the changes in metabolic cost with assistance. Due to the assistance being provided unilaterally, the models did not contain the same muscles on both legs, as the activation of each muscle could respond differently between legs. The regression model for incline walking ($R^2 = 0.96$, $p < 0.001$) shows that the activation of VM and ST on the UAL in addition to the activation of ST on the AL were closely related to the changes in metabolic cost with assistance. During decline walking, the regression model ($R^2 = 0.66$, $p < 0.05$) shows that the activation of VM and GA on the AL and VM on the UAL were largely influencing the change in metabolic cost with assistance. These results indicate that the changes in muscle activation on the UAL in addition to the changes in muscle activation on the AL are critical for understanding the user's compensatory behavior and determining the potential metabolic reduction gain.

In both incline and decline walking, the subjects felt that they were using less effort with the assistance condition

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compared to the zero impedance condition, regardless of their changes in metabolic cost. This result indicates that the user's metabolic cost was not always parallel with the subjects' subjective feeling on their level of effort. This result is also similar to a previous powered hip exoskeleton study where the user's preference in the onset timing of assistance did not correspond well with metabolic cost [29].

Although many exoskeleton studies have found variations in how users respond [23], [24], none have gone into detailed biomechanical analysis of the differences between individuals who respond to exoskeleton assistance and those who do not. This study was unique in that approximately half the participants responded to assistance, allowing a comparison of trends between the two sub-groups within the experiment. The biomechanical behaviors of the users in terms of muscle activity and kinetics separated the responders and nonresponders in this experiment. First, although the reduction in muscle activation of the knee extensor group on the AL was commonly present in both groups in incline walking, the responders' muscle activation reduction was larger. However, the nonresponders' UAL penalized the metabolic cost by the neuromechanical strategy that adapted. Their UAL upregulated the activation of the knee extensors to increase the positive power generation of the joint. This is similar to the link between the increased effort from the knee joint on the UAL and the increase in metabolic cost using a unilateral ankle exoskeleton [23]. On the other hand, the responders did not demonstrate this neuromechanical strategy. Importantly, the increase in total biological positive power generation as well as the increase of the total biological negative power absorption of both legs combined were observed from the incline nonresponders when the responders have shown a reduction in both quantities. This neuromechanical behavior of the UAL the nonresponders presented during incline walking is a compensatory behavior as a result of the assistance on the AL. On the other hand, the responders were able to utilize the assistance more effectively for their locomotion and were able to achieve a reduction in metabolic cost.

During decline walking, the most distinctive difference in the muscle activation of the responders and nonresponders was that the responders presented a large reduction in the knee extensors muscle activation on both the AL and UAL, whereas the reduction of the nonresponders' knee extensor muscle activation was not as consistently present. Due to the limitation of the experiment, only two nonresponders' motion capture data was available for analysis, therefore, it may not be practical to conclusively explain the kinetic difference between the responders and nonresponders. However, the key factor to differentiate the kinetic behavior of the responders and nonresponders in decline walking was the first 25% of the gait cycle. During the phase in the zero impedance mode, the AL of the nonresponders was heavily relying on the ankle joint for the majority of the power absorption. However, the responders were utilizing the knee joint to cover about one-third of the total power absorption of the leg during the phase of the zero impedance mode. Therefore, promoting the users to utilize the assistance for power absorption may result in a reduction in metabolic cost during the decline walking.

Additionally, since the assistance was provided unilaterally, one hypothesis is that asymmetry contributed to differences in metabolic cost. We performed step time symmetry analysis because it may explain the increased metabolic cost for nonresponders [30]. However, this result did not show a relationship with the metabolic outcome for the users on both incline and decline walking. Therefore, we suggest that future works should investigate augmenting users with bilateral assistance to see if the assistance can alleviate the metabolic penalty from the activation pattern of the knee extensor group on the UAL that nonresponders presented with unilateral assistance in this study. By augmenting users bilaterally, users would not be able to compensate for the assistance on the UAL. Therefore, we expect both legs to display activation patterns similar to the AL in this study, eliminating the metabolic penalty from the neuromechanical behavior of the UAL leg. Using bilateral assistance, the hip and ankle exoskeletons achieved a metabolic reduction in incline walking [5], [31]. As shown in the previous ankle exoskeleton research work, bilateral assistance may be more effective in metabolic cost reduction compared to unilateral assistance [32]. Additionally, varying assistance levels and types of controller may yield better metabolic cost reduction [23]. Therefore, bilateral assistance with different control strategies is sought to be investigated in the future.

As a suggestion to the future researchers based on the observed biomechanical behavior of the nonresponders, we suggest a better training paradigm. Even though we provided an ample amount of time for training, a better training method may improve the metabolic outcome of the users. In addition to the instruction, "Adapt to the powered assistance with a goal of maximizing the effectiveness of the assistance," adding instructions on the way they are walking can be employed. One suggested method is providing feedback to the users during the training to prevent upregulating the activation of the knee extensors on the UAL for incline walking or promote downregulating the activation of the knee extensors on the UAL for decline walking. This can be in a form of real-time visual feedback used previously [33].

One limitation of the study is that the upper thigh marker was placed on the exoskeleton, instead of on the skin of the thigh segment for two users with short leg lengths. Even though the marker has a relatively minimal effect on the joint angle calculation compared to joint markers, the more ideal case would have been to attach it to the skin. Only two nonresponders' motion capture data were available for decline walking. Therefore, it was hard to draw a generalized conclusion on the kinetic behavior of the decline walking nonresponders. The magnitude of assistance solely depended on the user's subjective feeling. The result of the metabolic cost may also be different if the magnitude of the assistance was determined based on the subject's weight, one of the most common ways in the research field.

VI. Conclusion

Our extension torque assistance during the early stance phase during incline and decline walking significantly reduced the user's knee extensor activation of the AL. However, only approximately a half of the subject group exhibited

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metabolic cost reduction with assistance in both incline and decline walking. Based on the main biomechanical factors that differentiated the responders and nonresponders during incline and decline walking, the overall dynamic interaction between the AL and UAL have determined whether a user would get energetic benefit in this experiment. During incline walking, the nonresponders showed reduction in knee extensor muscle activation on the AL but the knee extensors of the UAL increased their activation and canceled out the potential energetic benefit. During decline walking, the presence of a reduction in muscle activation on both AL and UAL was more consistent in the responder group, which was not present in the nonresponders group. Additionally, the nonresponders did not exhibit a large negative power absorption using the knee joint whereas the responders did. Therefore, for future implementation of a unilateral knee exoskeleton for reducing metabolic cost, the user's muscle activation level and kinetic behavior of lower limb joints on both AL and UAL should be the most determinant factors.

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