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A Soft Exosuit Assisting Hip Abduction for Knee Adduction Moment Reduction During Walking

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***Abstract*—The knee joint experiences significant torques in the frontal plane to keep the body upright during walking. Excessive loading over time can lead to knee osteoarthritis (OA), the progression of which is correlated with external knee adduction moment (KAM). In this paper, we present a wearable soft robotic exosuit that applies a hip abduction torque and evaluate its ability to reduce KAM. The exosuit uses a portable cable actuation system to generate torque when desired while remaining unrestrictive when unpowered. We explored five different force profiles on healthy participants (N=8) walking on an instrumented treadmill at 1.25 m/s. For each force profile, we tested two peak force levels: 15% and 20% of bodyweight. We observed KAM reductions with two of the five profiles. With Force Profile 2 (FP2), peak KAM was reduced by 9.61% and impulse KAM by 12.76%. With Force Profile 5 (FP5), we saw reductions of peak KAM by 6.14% and impulse KAM by 21.09%. These initial findings show that the device has the ability to change walking biomechanics in a consistent and potentially beneficial way.**

***Index Terms*—Wearable Robotics, Soft Robot Applications, Physically Assistive Devices, Prosthetics and Exoskeletons.**

I. Introduction

THE human knee joint handles a great deal of stress during walking. Exposure to abnormally high joint loading can damage the articular cartilage and is thought to be a cause of osteoarthritis (OA) [1], which is the most common type of arthritis [2]. Knee Osteoarthritis is a disease that affects an estimated 16% of people 15 years of age or older, and almost 23% of people 40 years and older globally [3]. The medial compartment of the knee, which takes 2.5 times greater joint loads than the lateral compartment [4], [5], is the most common site for knee OA [6]. Symptoms include pain and restriction of movement as the disease progresses [7]. These limitations in turn contribute to a decrease in quality of life [8].

The loading on the medial compartment that is thought to cause OA can be estimated by Knee Adduction Moment (KAM) [9], which is a popular surrogate measure in the field that is correlated with the progression of knee OA [10], [11], [12]. KAM is the torque about the knee in the frontal plane of the body due to the Ground Reaction Force (GRF) and its lever arm (the perpendicular distance from the knee joint center to GRF vector) in the frontal plane. It is reacted by the bone and cartilage of the medial compartment of the tibiofemoral joint, hence the connection between KAM and medial compartment loading [9]. Because OA is complex in both cause and presentation, no study that we know of has been able to prove that KAM causes knee OA. However, the correlation of KAM with knee OA and the ability to use KAM to predict future disease progression provides strong evidence that high KAM contributes to knee OA. Peak KAM is an often-used predictor of knee OA progression, but impulse KAM, the integral of KAM through the gait cycle, provides additional information and is potentially a more sensitive metric for distinguishing between knee OA severities [13], [14], [15].

Between the two elements of KAM (lever arm and GRF), knees with OA have been shown to have higher KAM caused by a larger lever arm, more so than by a larger GRF magnitude. This is because lever arm is affected by lower limb alignment, while the GRF magnitude is related to body weight. This implies that the lever arm is an important metric for predicting the progression of knee OA [16].

Many studies have investigated using gait modifications to reduce KAM, with the intention to decrease pain and further progression of OA. These include foot progression angle [17], wider step width [18], [19], and medio-lateral trunk sway [20]. Greater toe-out or toe-in foot angle during walking shifts the GRF closer to the knee joint center due to the shift in the center of pressure (COP), thereby reducing the peak KAM by 13% [17]. However, inconsistent changes were found in the peak KAM and the impulse KAM; toe-in walking reduced the peak KAM but increased the impulse KAM while toe-out walking reduced the impulse KAM but increased the peak KAM [21]. Similarly, a 10 - 20 cm increase in step width was suggested to reduce peak KAM by 19 - 28% [18], [19], but narrower step width was also reported to lead to having a smaller KAM [22]. Lateral trunk lean (4 - 12 degrees) has also been shown to reduce peak KAM by 7 - 25% and impulse KAM by 8 - 18%, but at the expense of some joint discomfort during retraining [20].

Another KAM reduction strategy in the literature is to target increasing hip abduction moment, the torque in the frontal plane about the hip joint. The hip abductor muscles exert a torque to prevent the pelvis from rotating about the femur in the frontal plane during the single-support stance phase of walking. If the hip abduction torque is low, a person will

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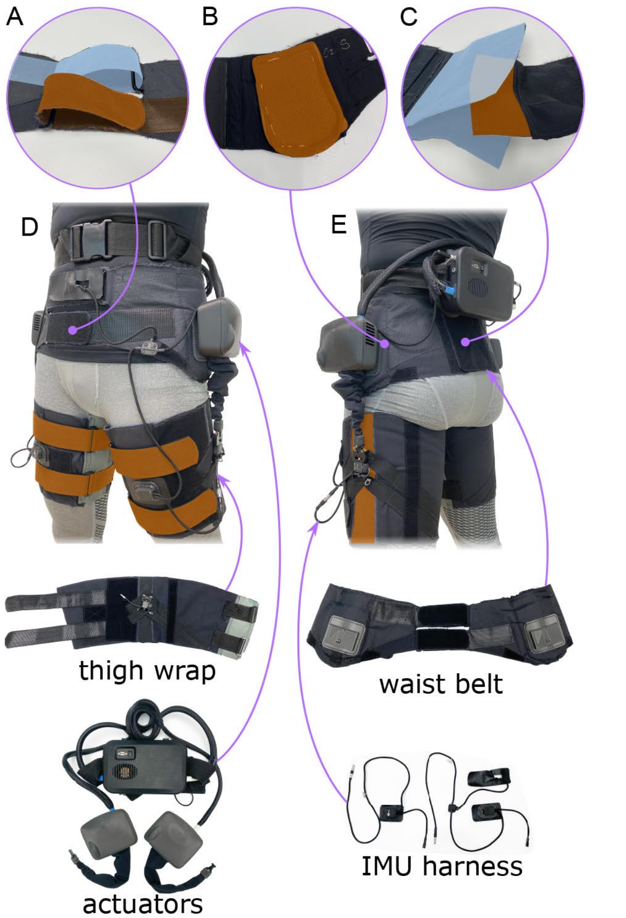

Fig. 1. The textile aspects of the suit design are designed for modularity and comfort. A) Front hook-and-loop straps allow different tightness higher or lower on the body. B) Foam padding distributed pressure to the bony aspects of the hip. C) A sandwiched hook-and-loop back closure allows high adjustability. D) Thigh wrap closures let the user set proximal and distal tightness independently (highlighted in orange). E) A lateral elastic strip accommodates changing thigh shape as muscles flex (highlighted in orange).

exhibit more contralateral pelvic drop, which in turn shifts the body's COM contralaterally [23], increasing KAM [24]. A study showed that greater hip abduction moment during gait can protect against the structural progression of knee OA [25], and suggested that this was likely due to lessened pelvic drop and subsequently lesser KAM and knee loading. This has motivated interventions targeting hip abductor strengthening. Surprisingly, such interventions have not shown reductions in KAM [26], [27]. While the reason for this discrepancy remains unknown, it inspired us to ask if we could increase hip abduction moment during walking regardless of biological muscle strength. In doing so, we thought to explore how an externally-applied hip abduction moment might affect KAM during walking.

One possible method to add this hip abduction moment is with soft wearable assistive robots. This field focuses on applying forces to the body using primarily compliant actuation methods like pneumatics or cables, anchored to textile components rather than rigid joints. Cable-driven soft wearable robots, often called exosuits, have emerged as a popular way to apply assistive force to the lower body due to their high power density, high bandwidth, and low profile [28]. Exosuits that target hip joint assistance have shown promise for reducing the metabolic cost of walking [29]. However, most hip exosuits to date focus on assisting the sagittal plane of the body. Very little research has been done on assisting the frontal plane, despite its pivotal importance in gait.

We propose a soft wearable robotic exosuit that provides torque about the hip joint in the frontal plane, with the goal of reducing KAM during walking gait. We explore five different assistance force profiles with different relative peak magnitudes, and tested each at two different overall force levels. In this paper, we introduce the design of the textile, sensing, and actuation system of the device in detail in Section II. Section III presents the controller and force profiles used to time the assistance magnitude of the device throughout walking gait. Sections IV and V walk through our experimental protocol on human subjects (N=8) to validate the device's ability to reduce KAM during level walking. Finally, we discuss the results and implications for future research in Section VI.

II. Device and Components

A. Textile-Based Suit Design

The suit is composed of a waist belt and two thigh wraps that are donned separately. The waist belt has a sandwiched hook-and-loop closure in the back, allowing for about 10cm of adjustment between the two actuator mounts to accommodate different waist sizes as shown in Figure 1C. Another velcro closure in the front is designed to allow the user to easily don the suit and differentially tighten the top and bottom of the waist belt, pictured in Figure 1A. 11 mm of foam padding was sewn to the inside of the waist belt underneath the actuator mounts to avoid pressure concentrations on bony aspects of the lateral hip, as shown in Figure 1B.

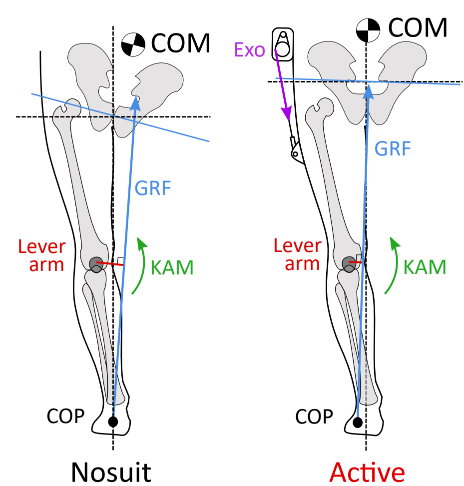

Fig. 2. Left: Hips with weak abductor muscles may experience greater contralateral pelvic drop (exaggerated in this figure for clarity) during single stance. Right: The exosuit proposed in this paper adds a hip abduction torque (purple) to help stabilize the angle of the pelvis, keeping the body Center of Mass (COM) less laterally offset from the knee, reducing the GRF lever arm about the knee and thus lowering KAM.

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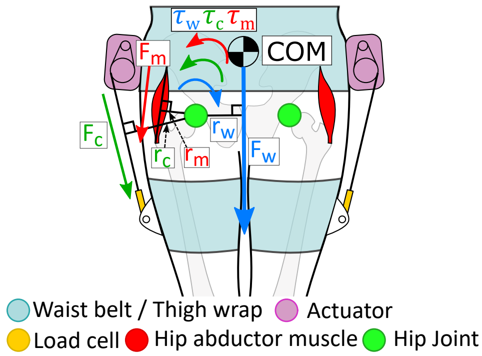

Fig. 3. Free body diagram of the functional principle of the device. During the stance phase of walking, the force from the weight of the body F_w causes a torque about the hip joint $r_w \times F_w$ that causes the pelvis to lean contralaterally. The hip abductor muscles resist this lean by generating an opposing torque $r_m \times F_m$. This torque is assisted by the robot with a parallel torque $r_c \times F_c$ generated by spooling in the cable on the device.

The thigh wraps have similar velcro closures allowing the user to tighten the top and bottom of the wrap separately to a comfortable degree, highlighted in Figure 1D. They also have an elastic strip that sits along the lateral side of the thigh that allows the wrap to expand circumferentially as the thigh changes shape throughout the gait cycle, shown in Figure 1E. Crossing over the elastic strip are angled loops of inextensible fabric where the distal end of the actuator cable is attached, with a higher and lower mounting position to accommodate different bodies.

B. Actuation System

To minimize the burden of wearing the suit, the actuation system was designed to be small and light, with torque and speed capabilities geared to provide the desired assistance levels on the human body during walking. We used two custom mobile actuators, which were designed using a rope winch architecture. Each actuator has a motor (U5; T-Motor, Jiangxi, China), a timing belt reduction (2.2:1 gearing ratio), and a metal cylindrical drum (8.8 mm diameter), which are connected in series. The actuator spools and unspools a high tension rope (1.8 mm diameter; SK78; Marlow Ropes, Hailsham, UK). A magnetic encoder (AS5145B; AMS A.G., Premstaetten, Austria) is placed behind the motor to monitor the rotational position of the motor in order to measure the length of the spooled rope. The textile components of the suit weigh 0.59 kg, and the electromechanical components weigh 2.28 kg, for a total system weight of 2.87 kg.

Both actuators are controlled by a motor controller board that uses two motor controllers (Gold Twitter; Elmo Motion Control Ltd., Petach-Tikva, Israel) running force and position controllers (described in Section III). In addition, one 32-bit ARM microprocessor (Cortex-M7; Atmel, San Jose, CA, USA) reads in all sensor data via CAN communication, computes high-level control algorithms including gait segmentation, force trajectory generator, and closed-loop PI controller, and sends the required current commands to the motor controllers through CAN communication. Two 3450-mAh Li-Ion smart batteries (RRC2054; RRC Power Solutions, Homburg, Germany) power the actuators and the electronics.

The device applies assistance torque to the hip joint in the frontal plane by pulling on the rope to the desired load, which distributes forces through the textile components and soft tissue and onto the underlying bone. Thus the torque being applied to the hip joint can be approximated as the cross product of the vector from the hip joint to the cable (perpendicular to the cable) with the force vector from the actuator along the cable, as illustrated in Figure 3. We expect this torque to assist the hip abductor muscles in stabilizing the pelvis in the frontal plane, reducing the lever arm that contributes to high KAM (pictured in Figure 2).

C. Sensor System

The exosuit contains two Inertial Measurement Units (IMUs) (one on the front of each thigh) and two load cells (one at the distal end of each cable, shown in Figure 3) to enable a closed-loop force controller shown in Section III.

On each leg, an IMU (MTi-3 AHRS; Xsens Technologies B.V., Enschede, Netherlands) is placed on the frontal region of the thigh wrap to measure the sagittal-plane angular displacement and angular velocity of the thigh. The IMU measurements are used in the gait detection algorithm in Figure 4. In addition, one load cell (LSB200; Futek Advanced Sensor Technology Inc., Irvine, CA, USA) was integrated in the thigh wrap of each leg to measure hip abduction force generated by the actuator. The force measurements are used in the closed loop force controller of the low-level controller described in Section III.

III. Controller Design

A. High and Low-Level Controller

We designed the controller to apply force to the wearer during the stance phase of walking without restricting body motion during the swing phase of walking. The controller consists of a high-level and low-level controller as shown in Figure 4. The high-level controller highlighted in green estimates the percentage through the gait cycle at a given time based on the thigh IMUs via a gait event detection algorithm described in subsection B. It then generates the desired force or position command based on the gait cycle (GC) percentage using a force profile trajectory generator. During the stance phase (heel strike (HS) to toe-off (TO)), the force profile trajectory generator commands a desired force based on an assistance profile described in subsection C. Right after toe-off, the position profile trajectory controller commands the motor to fully push out the cable and maintain this cable position until the next gait event.

The low-level controller runs closed-loop current control for the actuation system to track the desired force or position command trajectories generated by the high level controller. It uses measured force ($F_{measured}$, from the load cells) or position ($P_{measured}$, from the encoder) to calculate the error from the desired force or position (F_{err}, P_{err}), which feeds into a PI filter that outputs a desired current (C_{des}) to the actuator unit torque controller. The system is controlled by

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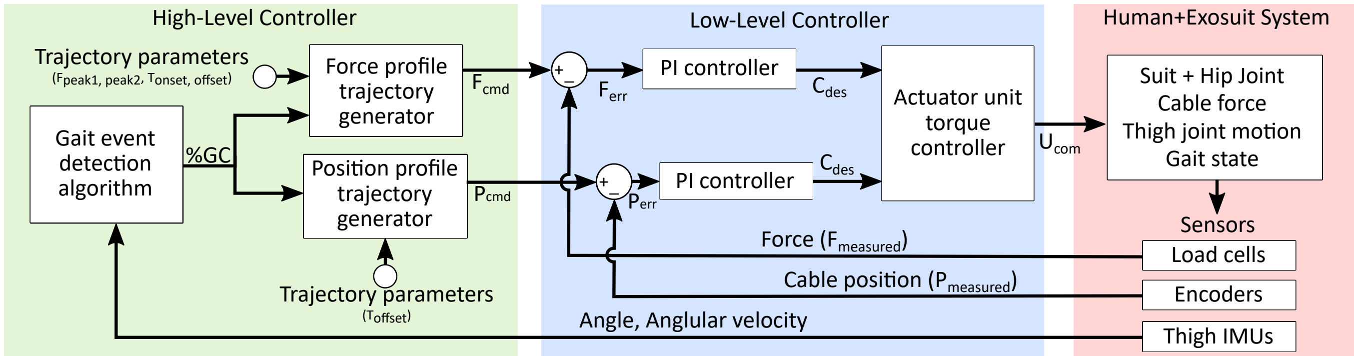

Fig. 4. Diagram of overall controller architecture. From the high level controller, thigh IMUs are used to estimate gait cycle through the gait event detection algorithm. Based on this estimate, a desired force or position profile trajectory is generated. The force or position command is sent to the low-level controller, which closes the loop on force or position using sensor data from measuring the actuator's cable tension or length. Finally, the low-level controller regulates control effort (U_{com}) for the actuator system to track the desired force or position generated by the high-level controller.

the PI controller with F_{err} during the stance phase of walking or the PI controller with P_{err} pushing out the cable during the swing phase of walking. Finally, the actuator unit torque controller regulates control effort (U_{com}) for the actuator system to track the desired force or position.

B. Gait Event Detection

In order to track gait cycle percentage while walking, the system uses the thigh-mounted IMU on each leg to detect maximum hip flexion angle (MHF), then measures stride time as the time between two consecutive MHF events. This allows the gait cycle to be expressed as a percentage from MHF to subsequent MHF. Hip flexion angle can be accurately approximated by measuring thigh angle relative to the inertial coordinate system (as shown in Figure 5A) with each thigh IMU. The algorithm identifies MHF as the first positive thigh angle peak after a negative thigh angle peak.

To predict the onset and offset timing of the stance phase within the gait cycle, we collected baseline motion capture data from all participants and measured the relationship between MHF, HS, and TO, as shown in Figure 5A. Across all participants, HS occurred 12.18 ± 0.74% after MHF, and TO occurred 60.36 ± 1.46% after HS. From this, we set HS = 12%, TO = 72% after MHF for all participants as the default setting.

C. Assistance Profiles

The assistance profiles used in this study were based on the shape of biological hip abduction torque during walking, shown in Figure 5A. During the stance phase of walking, there are two large positive moment regions: one in early stance corresponding to pelvic angle stabilization during weight acceptance, and another in late stance as the body's center of mass is lowered and the limb braces for push-off.

Since this is the first study to our knowledge to test different hip abduction assistance profile shapes, we had to explore a variety of shapes. The shape of hip abductor muscle activity is a two-peak shape, but the relative height of the two peaks is different for each muscle [30]. In a previous paper that simulated hip abduction assistance targeting metabolic

TABLE I
FORCE PROFILES

Profile #	F_{peak1}	F_{mid}	F_{peak2}
1	100%	80%	100%
2	65%	50%	100%
3	100%	50%	65%
4	N/A	N/A	100%
5	100%	N/A	N/A

reduction, the authors found that the most efficient assistance profile exhibited relative peak heights that were different from the biological hip abduction moment [31]. From this understanding, we decided to test a range of relative peak heights, including two single-peak profiles (FP4 and FP5) to represent the extreme cases where one peak was insignificant in height relative to the other. This resulted in five profiles with key values listed in Table I and the resulting shapes shown in Figure 5C.

We also varied the percent gait cycle after MHF at which assistance was started and stopped, T_{onset} = 12 ± 1% and T_{offset} = 72 ± 2%, based on the timing described in the previous section and subjective feedback from the wearer. For FP1, T_{mid} was set as the middle timing between T_{onset} and T_{offset}. For FP2 and FP3, T_{mid} was set to be closer to the lower peak, which allowed for a more gradual onset and offset of assistance force and improved subjective comfort. T_{peak1} was defined as the middle timing between T_{onset} and T_{mid}, and T_{peak2} was defined as the middle timing between T_{mid} and T_{offset}.

IV. Human Subject Testing

A. Experimental Protocol

To evaluate the performance and the biomechanical efficacy of the hip abduction device during walking, we conducted an experimental protocol on eight healthy subjects with no musculoskeletal injuries (age 28.8 ± 3.7 y.o., weight 77.4 ± 8.8 kg, height 172.4 ± 7.1 cm). The study consisted of two experimental visits per participant: suit familiarization and testing, which were held on separate days 2-8 days apart.

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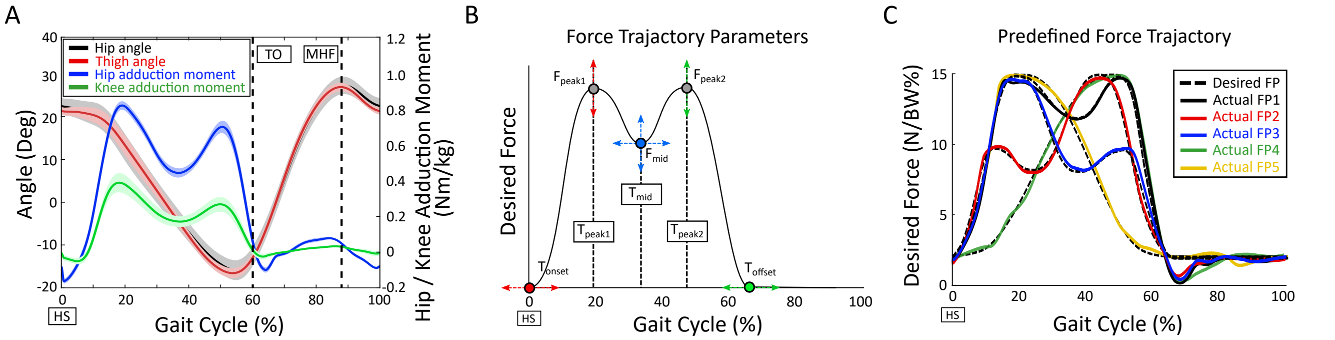

Fig. 5. Overview of gait segmentation and assistance profiles, A) Average sagittal plane hip angle, thigh angle, hip abduction moment, and knee adduction moment in the frontal plane of participants (N=8) during walking. Shaded regions indicate standard error; B) Force profile trajectory generator. The generator contains trajectory parameters (F_{peak1}, F_{peak2}, T_{onset}, T_{offset}); C) Predefined desired force trajectory and actual experimental force output of a representative participant. Five different force profiles (force profile 1-5 (FP1-FP5)) were designed.

On both days, each participant walked on an instrumented treadmill (Bertec, Columbus, Ohio, US) at a speed of 1.25 m/s.

The first testing visit was a baseline and suit familiarization session where the participant experienced the exosuit and the predefined force profiles described in Section III were applied. During the testing, each force profile was tuned based on user's feedback. First, applying each profile to the subject with a low-level force magnitude (15% of bodyweight (BW), which is $\sim$18% of biological hip abduction moment and is the lowest level of assistance that was effective in previous exosuit systems [32]), timing parameters including T_{onset}, T_{offset}, and T_{mid} were tuned to be earlier or later (± 2% of GC) based on subjective feedback. Next, the high-level force magnitude was set to be 20% of BW, which is $\sim$23% of biological hip abduction moment and the highest magnitude that all participants found comfortable. At the end, the participants walked for five randomized four-minute bouts, experiencing each of the five different force profiles with low-level, then high-level force, in order to get familiar with the functioning of the system.

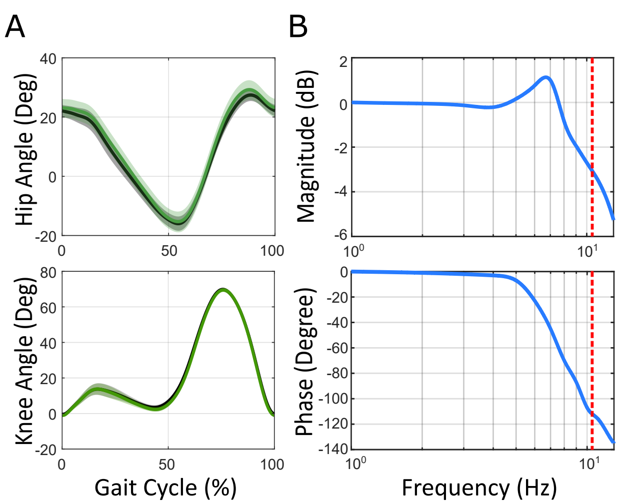

Fig. 6. Mechanical transparency and system performance; A) Graphs show average joint angles of all participants in the sagittal plane during walking, nosuit condition vs. slack condition. Shaded regions indicate standard error; B) Bandwidth test results of the actuator on one subject. The vertical red dashed line marks the -3dB frequency.

During the testing day, subjects had a five-minute warm-up followed by twelve different two-minute data collection bouts, with rest breaks between bouts: one "no suit" condition, one slack condition, five active conditions with the low assistance, and five active conditions with the high assistance. In each two-minute trial for the active conditions, the force level ramped up gradually after thirty seconds of slack walking in the first minute. The active conditions were randomized to minimize the effects of adaptation or fatigue on the statistical analysis. All study procedures were approved by the Harvard Medical School Institutional Review Board.

B. Data Collection and Analysis

To collect kinematic data, optical motion capture (Qualisys, Gothenburg, Sweden) was used, and reflective markers were placed on anatomical bony landmarks on both legs (iliac crest, trochanter, lateral/medial knee, lateral/medial ankle, lateral/medial toe, lateral/medial heel, tip of toe, and heel) and on cluster plates on the thigh and shank segments for tracking. Raw marker data were collected from each participant at 120Hz and ground reaction force (GRF) were collected at 2 kHz. Kinematic and kinetic data were processed using Qualisys and Visual 3D (C-motion Inc., Rockville, MD, USA). Joint kinematics and kinetics were calculated using inverse models of each individual.

C. Statistical Analysis

For each participant, KAM measurements were normalized with respect to body weight. The peak KAM and impulse KAM were evaluated during the stance phase. Repeated measures analysis of variance (ANOVA) including 7 conditions (nosuit, slack, active conditions (FP1, FP2, FP3, FP4, FP5)) were performed to verify the effect of assistance on the peak KAM and impulse KAM for all participants. Additional ANOVA was used to verify the effect of different force level assistance (low assistance, high assistance) on peak KAM and impulse KAM (significance level α = 0.05; MATLAB, MathWorks Inc., Natick, MA, USA).

The assistance profiles that resulted in the largest group-level KAM reduction were selected for in-depth analysis. Force profile 2 (FP2) yielded the highest reduction in the peak KAM, while force profile 5 (FP5) gave the most reduction in impulse KAM. Two sided paired t-tests were used to identify statistically significant differences in lever arm and center of pressure (COP) of the foot between these two active conditions and the no suit condition (significance level α = 0.05; MATLAB, MathWorks Inc., Natick, MA, USA).

V. Results

A. Mechanical Transparency and System Performance

The transparency of the exosuit design was evaluated by comparing kinematics between the nosuit condition and the slack condition (the exosuit donned but unpowered) for all participants during walking. The mean joint angles on the left side across all participants were comparable in the sagittal plane as shown in Figure 6A. The RMS deviation (MEAN ± SE (Deg)) in mean joint angles between two conditions are as follows: hip: 4.59 ± 1.25, and knee: 0.83 ± 1.97. In addition, the RMS deviation in a range of joint angles are as follows: hip: 1.79 ± 0.36, and knee: 1.02 ± 0.45. The hip joint angle had the large deviation, but it had a similar range of joint angle. This may be due to the fact that pelvis markers had to be attached on the waist belt component, which can cause angle offsets if the belt shifted position over the course of testing. There were also no statistically significant differences ($p > 0.050$) in mean and range of joint angles.

To evaluate the force tracking performance of the controller, the bandwidth of the actuator was tested experimentally while the actuator was worn on a subject. The actuator force bandwidth was measured by commanding swept sinusoidal force signals (from 1 to 13Hz) with a force magnitude between 15% and 25% of biological hip abduction moment, which was representative of the force magnitudes used for the experimental protocol. The bandwidth was calculated as the frequency at which the gain reached -3dB. The results in Figure 6B show a bandwidth of 10.26Hz.

B. Knee Adduction Moment (KAM)

Participants showed statistically significant reductions in peak and impulse knee adduction moment (KAM) for FP2 and FP5, as shown in Figure 7C. FP2 achieved higher peak KAM reduction, and FP5 caused a higher impulse KAM reduction. Peak KAM was reduced by 9.61 ± 2.27% with FP2 (p = 0.002) and by 6.14 ± 2.73% with FP5 (p = 0.031) compared with the nosuit condition (Figure 7A). The impulse KAM was reduced by 12.76 ± 3.71% with FP2 (p = 0.008) and by 21.09 ± 4.10% with FP5 (p = 0.002) compared with the nosuit condition (Figure 7B). Nosuit and slack conditions were not significantly different for peak KAM and impulse KAM ($p \geq 0.741$), validating that the suit does not increase or decrease KAM when unpowered. We saw a decreasing trend in impulse KAM related to the nosuit condition for all force profiles, but no significant changes were observed in peak KAM or impulse KAM for FP1, FP3, and FP4 compared with the nosuit condition ($p \geq 0.421$) or the slack condition ($p \geq 0.396$).

In addition, no significant trend was found with respect to the different assistance force levels (low and high) for reducing the peak KAM and impulse KAM ($p > 0.996$), shown in

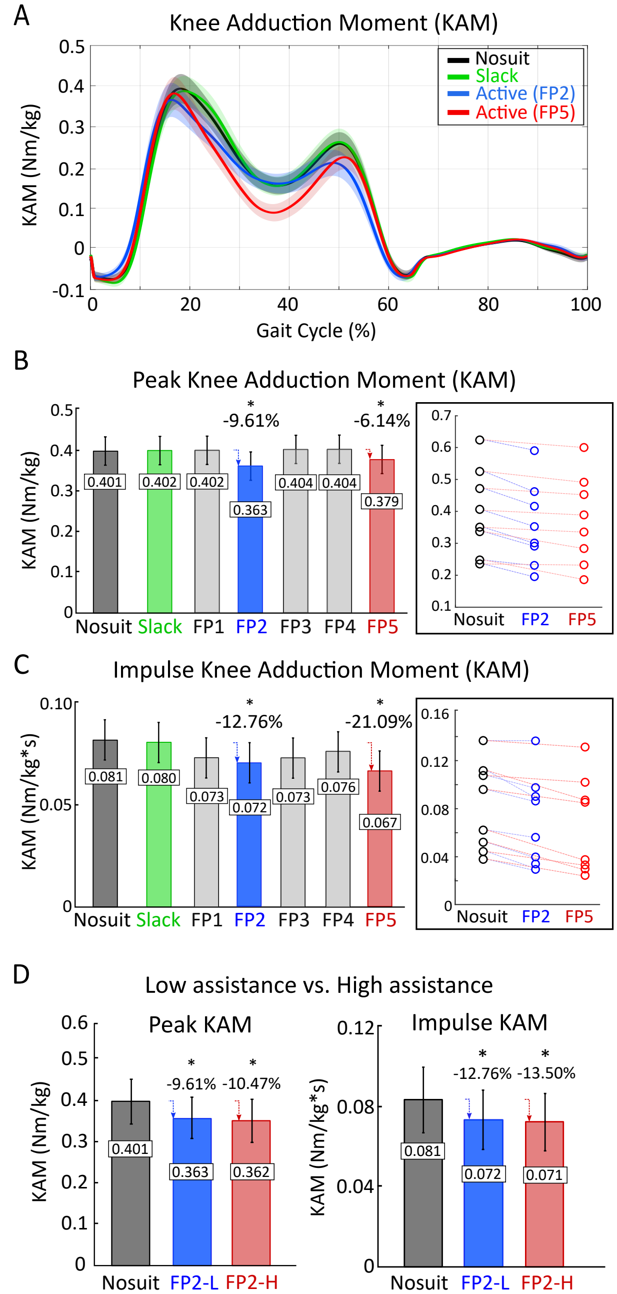

Fig. 7. Changes in knee adduction moment for both legs during treadmill walking with different force profiles for all participants (N=8); A) Average knee adduction moment (KAM) in different conditions (nosuit, slack, active with force profiles (FP2, FP5) plotted versus gait cycle percentage. Shaded regions indicate standard error; B) Average and individual peak KAM during walking (MEAN ± SE); C) Average and individual impulse KAM during the stance phase of walking (MEAN ± SE); D) Comparison between low assistance (15% of body weight) and high assistance (20% of body weight) for force profile 2 (FP2): average peak and impulse knee adduction moment (KAM) (MEAN ± SE) with nosuit, low assistance (FP2-L), and high assistance (FP2-H). High assistance did not yield significantly different results compared with low assistance ($p > 0.05$). Error bar indicates standard error. The bars with * represent statistically significant differences with the nosuit condition ($p < 0.050$).

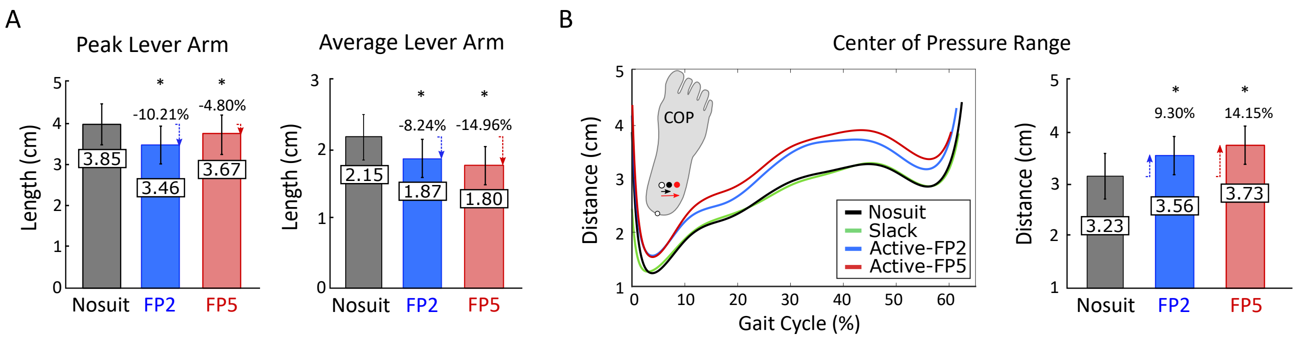

Fig. 8. A) Changes in lever arm (the perpendicular distance from the knee joint center to the GRF vector) during treadmill walking with different force profiles; average peak lever arm (MEAN ± SE) during the stance phase of walking, and average mean lever arm (MEAN ± SE) during the stance phase of walking; B) Center of Pressure (COP) distance from heel marker during the stance phase of walking; Average COP distance (MEAN ± SE) was increased by FP2 and FP5. The bars with * represent statistically significant differences with the nosuit condition ($p < 0.050$).

Figure 7D. This does not imply that the level of assistance force to the hip joint has no effect on the KAM reduction.

C. Lever Arm

Lever arm was measured by the perpendicular distance from the knee joint center (middle point between medial anatomical knee marker and lateral anatomical knee marker) to the GRF vector in the frontal plane. Hip abduction assistance resulted in reduced peak lever arm of 10.21 ± 1.65% ($p < 0.001$) with FP2, and 4.80 ± 1.72% with FP5 ($p = 0.021$). The assistance also achieved an average lever arm reduction of 8.25 ± 3.12% ($p = 0.033$) with FP2 and 14.96 ± 2.43% ($p < 0.001$) with FP5 (Figure 8A).

The lever arm seemed to be reduced by the increased range of medial-lateral center of pressure (COP), as shown in Figure 8B. The COP distance was calculated as the length from heel marker to COP position during the stance phase of walking. Average COP distance was increased by 9.30 ± 2.60% ($p = 0.002$) with FP2 and by 14.15 ± 2.24% ($p < 0.001$) with FP5.

VI. Conclusion and Discussion

In this study, we designed and validated the initial performance of a hip abduction assist soft exosuit to reduce knee adduction moment (KAM) during walking.

Hip abduction, the movement of the pelvis relative to the femur in the frontal plane, is critically important to walking, despite its small range of motion, because it keeps the pelvis (and thus the whole upper body) aligned above the stance limb which bears the body's weight. The exosuit proposed in this paper applies assistive force to the wearer in parallel with the hip abductor muscles, generating a hip abduction torque during the stance phase of walking, and allows free motion during the swing phase. The exosuit is capable of delivering relatively high-frequency (10.26 Hz) force profiles, which was validated with a -3dB bandwidth test. While unpowered, the exosuit is highly mechanically transparent to the wearer, which was verified by comparing the no suit and slack conditions, which showed no significant differences in the mean and range of lower limb joint angles, or in KAM.

We developed a highly flexible force trajectory generator which allowed us to quickly iterate through a variety of one-peak and two-peak assistance force profiles. We investigated five different force profiles based on variations of a typical biological hip abduction moment shape. We found that, for two of the five profiles, assistance from the suit resulted in significant reductions in KAM. These same two profiles showed a reduction in the lever arm during walking and increased mediolateral COP of the foot. It may be that the assistive force from the suit reduces KAM by increasing the hip abduction angle during stance, thereby shifting the body's COM laterally, which aligns it more directly above the knee joint, reducing the lever arm about the knee.

We expected to see a trend between overall assistance level and KAM reduction. However, we observed no difference between 15% bodyweight and 20% bodyweight assistance. One possible explanation could be that the exosuit was partially "cueing" the wearer to perform more hip abduction, acting as a stimulus more so than an added force, which might be less sensitive to force level. It could also be that soft tissue deformation absorbs power between the suit and the skeleton, evening out the force experienced by the skeleton and rendering the 5% bodyweight difference between conditions insignificant. More research would be necessary to separate the effects of added work and cueing.

This preliminary study showed lower peak KAM reductions compared to the reductions found with other clinical methods presented in the introduction, but also showed higher impulse KAM reductions. However, our study was limited to testing on a healthy population with lower baseline KAM than these other studies, so direct comparisons to those interventions are tenuous. Since the subjects in this study had low baseline KAM, they may have been limited in how much assistance they could receive from the device. For patients with knee OA who may benefit more from the device, we could expect greater KAM reductions. In addition, this device does not require gait retraining or major gait modifications that can lead to inconsistent outcomes [21], [22] or discomfort [20]. We believe this method of applying assistance torque directly with a soft robotic wearable device may offer a new approach for gait retraining, perhaps in conjunction with other therapies, by providing consistent KAM reduction without consciously changing a person's normal gait pattern.

This study had several limitations that could be addressed in future work. First and foremost, our experiment did not reveal a correlation between force profile shape and KAM reduction. As such, in future studies, more research will be conducted to understand how each assistance force profile changes walking biomechanics and how they perform at various walking speeds. In addition, future research will be done with populations with knee OA to better investigate the KAM reductions this device could achieve in comparison to other clinical interventions, and what other kinematic changes it might cause. Finally, a longitudinal study must be done to investigate how lowering KAM using the device could affect the progression of knee OA. These studies could provide insights on the development of robotic wearable assistive devices, enabling a new methodology for addressing knee OA.

REFERENCES

[1] B. Jackson, A. Wluka, A. Teichtahl, M. Morris, and F. Cicuttini, "Reviewing knee osteoarthritis — a biomechanical perspective," *Journal of Science and Medicine in Sport*, vol. 7, no. 3, pp. 347–357, 2004.

[2] R. C. Lawrence, C. G. Helmick, F. C. Arnett, R. A. Deyo, D. T. Felson, E. H. Giannini, S. P. Heyse, R. Hirsch, M. C. Hochberg, G. G. Hunder, M. H. Liang, S. R. Pillemer, V. D. Steen, and F. Wolfe, "Estimates of the prevalence of arthritis and selected musculoskeletal disorders in the United States," *Arthritis & Rheumatism*, vol. 41, no. 5, pp. 778–799, 1998.

[3] A. Cui, H. Li, D. Wang, J. Zhong, Y. Chen, and H. Lu, "Global, regional prevalence, incidence and risk factors of knee osteoarthritis in population-based studies," *EClinicalMedicine*, vol. 29, p. 100587, 2020.

[4] O. D. Schipplein and T. P. Andriacchi, "Interaction between active and passive knee stabilizers during level walking," *Journal of Orthopaedic Research*, vol. 9, no. 1, pp. 113–119, 1991.

[5] L. Sharma, D. E. Hurwitz, E. J. A. Thonar, J. A. Sum, M. E. Lenz, D. D. Dunlop, T. J. Schnitzer, G. Kirwan-Mellis, and T. P. Andriacchi, "Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis," *Arthritis & Rheumatism*, vol. 41, no. 7, pp. 1233–1240, 1998.

[6] B. L. Wise, J. Niu, M. Yang, N. E. Lane, W. Harvey, D. T. Felson, J. Hietpas, M. Nevitt, L. Sharma, J. Torner, C. E. Lewis, Y. Zhang, and M. O. M. Group, "Patterns of compartment involvement in tibiofemoral osteoarthritis in men and women and in whites and African Americans," *Arthritis Care & Research*, vol. 64, no. 6, pp. 847–852, 2012.

[7] D. J. Hunter and S. Bierma-Zeinstra, "Osteoarthritis," *The Lancet*, vol. 393, no. 10182, pp. 1745–1759, 2019.

[8] K. L. Dominick, F. M. Ahern, C. H. Gold, and D. A. Heller, "Health-related quality of life among older adults with arthritis," *Health and Quality of Life Outcomes*, vol. 2, no. 1, p. 5, 2004.

[9] D. Zhao, S. A. Banks, K. H. Mitchell, D. D. D'Lima, C. W. Colwell, and B. J. Fregly, "Correlation between the knee adduction torque and medial contact force for a variety of gait patterns," *Journal of Orthopaedic Research*, vol. 25, no. 6, pp. 789–797, 2007.

[10] T. Miyazaki, M. Wada, H. Kawahara, M. Sato, H. Baba, and S. Shimada, "Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis," *Annals of the Rheumatic Diseases*, vol. 61, no. 7, p. 617, 2002.

[11] N. Foroughi, R. Smith, and B. Vanwanseele, "The association of external knee adduction moment with biomechanical variables in osteoarthritis: A systematic review," *The Knee*, vol. 16, no. 5, pp. 303–309, 2009.

[12] A. Baliunas, D. Hurwitz, A. Ryals, A. Karrar, J. Case, J. Block, and T. Andriacchi, "Increased knee joint loads during walking are present in subjects with knee osteoarthritis," *Osteoarthritis and Cartilage*, vol. 10, no. 7, pp. 573–579, 2002.

[13] C. O. Kean, R. S. Hinman, K. A. Bowles, F. Cicuttini, M. Davies-Tuck, and K. L. Bennell, "Comparison of peak knee adduction moment and knee adduction moment impulse in distinguishing between severities of knee osteoarthritis," *Clinical Biomechanics*, vol. 27, no. 5, pp. 520–523, 2012.

[14] L. E. Thorp, D. R. Sumner, J. A. Block, K. C. Moisio, S. Shott, and M. A. Wimmer, "Knee joint loading differs in individuals with mild compared with moderate medial knee osteoarthritis," *Arthritis & Rheumatism: Official Journal of the American College of Rheumatology*, vol. 54, no. 12, pp. 3842–3849, 2006.

[15] M. Creaby, Y. Wang, K. L. Bennell, R. Hinman, B. Metcalf, K.-A. Bowles, and F. M. Cicuttini, "Dynamic knee loading is related to cartilage defects and tibial plateau bone area in medial knee osteoarthritis," *Osteoarthritis and Cartilage*, vol. 18, no. 11, pp. 1380–1385, 2010.

[16] M. A. Hunt, T. B. Birmingham, J. R. Giffin, and T. R. Jenkyn, "Associations among knee adduction moment, frontal plane ground reaction force, and lever arm during walking in patients with knee osteoarthritis," *Journal of Biomechanics*, vol. 39, no. 12, pp. 2213–2220, 2006.

[17] P. B. Shull, R. Shultz, A. Silder, J. L. Dragoo, T. F. Besier, M. R. Cutkosky, and S. L. Delp, "Toe-in gait reduces the first peak knee adduction moment in patients with medial compartment knee osteoarthritis," *Journal of Biomechanics*, vol. 46, no. 1, pp. 122–128, 2013.

[18] B. J. Fregly, "Computational Assessment of Combinations of Gait Modifications for Knee Osteoarthritis Rehabilitation," *IEEE Transactions on Biomedical Engineering*, vol. 55, no. 8, pp. 2104–2106, 2008.

[19] F. Stief, J. Holder, Z. Feja, A. Lotfolahpour, A. Meurer, and J. Wilke, "Impact of subject-specific step width modification on the knee and hip adduction moments during gait," *Gait & Posture*, vol. 89, pp. 161–168, 2021.

[20] M. A. Hunt, M. Simic, R. S. Hinman, K. L. Bennell, and T. V. Wrigley, "Feasibility of a gait retraining strategy for reducing knee joint loading: Increased trunk lean guided by real-time biofeedback," *Journal of Biomechanics*, vol. 44, no. 5, pp. 943–947, 2011.

[21] M. Simic, T. Wrigley, R. Hinman, M. Hunt, and K. Bennell, "Altering foot progression angle in people with medial knee osteoarthritis: the effects of varying toe-in and toe-out angles are mediated by pain and malalignment," *Osteoarthritis and Cartilage*, vol. 21, no. 9, pp. 1272–1280, 2013.

[22] B. D. Street and W. Gage, "The effects of an adopted narrow gait on the external adduction moment at the knee joint during level walking: Evidence of asymmetry," *Human Movement Science*, vol. 32, no. 2, pp. 301–313, 2013.

[23] C. D. MacKinnon and D. A. Winter, "Control of whole body balance in the frontal plane during human walking," *Journal of Biomechanics*, vol. 26, no. 6, pp. 633–644, 1993.

[24] J. Takacs and M. A. Hunt, "The effect of contralateral pelvic drop and trunk lean on frontal plane knee biomechanics during single limb standing," *Journal of Biomechanics*, vol. 45, no. 16, pp. 2791–2796, 2012.

[25] A. Chang, K. Hayes, D. Dunlop, J. Song, D. Hurwitz, S. Cahue, and L. Sharma, "Hip abduction moment and protection against medial tibiofemoral osteoarthritis progression," *Arthritis & Rheumatism*, vol. 52, no. 11, pp. 3515–3519, 2005.

[26] E. A. Sled, L. Khoja, K. J. Deluzio, S. J. Olney, and E. G. Culham, "Effect of a Home Program of Hip Abductor Exercises on Knee Joint Loading, Strength, Function, and Pain in People With Knee Osteoarthritis: A Clinical Trial," *Physical Therapy*, vol. 90, no. 6, pp. 895–904, 2010.

[27] K. Bennell, M. Hunt, T. Wrigley, D. Hunter, F. McManus, P. Hodges, L. Li, and R. Hinman, "Hip strengthening reduces symptoms but not knee load in people with medial knee osteoarthritis and varus malalignment: a randomised controlled trial," *Osteoarthritis and Cartilage*, vol. 18, no. 5, pp. 621–628, 2010.

[28] A. T. Asbeck, S. M. D. Rossi, K. G. Holt, and C. J. Walsh, "A biologically inspired soft exosuit for walking assistance," *The International Journal of Robotics Research*, vol. 34, no. 6, pp. 744–762, 2015.

[29] M. Xiloyannis, R. Alicea, A.-M. Georgarakis, F. L. Haufe, P. Wolf, L. Masia, and R. Riener, "Soft Robotic Suits: State of the Art, Core Technologies, and Open Challenges," *IEEE Transactions on Robotics*, no. 99, 2021.

[30] A. Zacharias, T. Pizzari, A. I. Semciw, D. J. English, T. Kapakoulakis, and R. A. Green, "Comparison of gluteus medius and minimus activity during gait in people with hip osteoarthritis and matched controls," *Scandinavian Journal of Medicine & Science in Sports*, vol. 29, no. 5, pp. 696–705, 2019.

[31] C. L. Dembia, A. Silder, T. K. Uchida, J. L. Hicks, and S. L. Delp, "Simulating ideal assistive devices to reduce the metabolic cost of walking with heavy loads," *PLOS ONE*, vol. 12, no. 7, p. e0180320, 2017.

[32] B. T. Quinlivan, S. Lee, P. Malcolm, D. M. Rossi, M. Grimmer, C. Siviy, N. Karavas, D. Wagner, A. Asbeck, I. Galiana *et al.*, "Assistance magnitude versus metabolic cost reductions for a tethered multiarticular soft exosuit," *Science robotics*, vol. 2, no. 2, p. eaah4416, 2017.