The Effect of Hip Exoskeleton Weight on Kinematics, Kinetics, and Electromyography During Human Walking*

Michael A. Normand†, Jeonghwan Lee, Hao Su‡, James S. Sulzer§¶

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†Michael A. Normand and Jeonghwan Lee have contributed equally to this work. Both are from the Mechanical Engineering at the University of Texas at Austin, Austin, TX

‡Hao Su is with Department of Mechanical and Aerospace Engineering, North Carolina State University and Joint NCSU/UNC Department of Biomedical Engineering, North Carolina State University, Raleigh, NC, 27695; University of North Carolina at Chapel Hill, Chapel Hill, NC 27599, USA.

§James Sulzer is the Department of Physical Medicine and Rehabilitation at MetroHealth Hospital and Case Western Reserve University, Cleveland, OH

¶Correspondence to: James Sulzer jss280@case.edu

1 Abstract

In exoskeleton research, transparency is the degree to which a device hinders the movement of the user, a critical component of performance and usability. Transparency is most often evaluated individually, thus lacking generalization. Our goal was to systematically evaluate transparency due to inertial effects on gait of a hypothetical hip exoskeleton. We predicted that the weight distribution around the pelvis and the amount of weight applied would change gait characteristics. We instructed 21 healthy individuals to walk on a treadmill while bearing weights on the pelvis between 4 and 8 kg in three different configurations, bilaterally, unilaterally (left side) and on the lumbar portion of the back (L4). We measured kinematics, kinetics, and muscle activity during randomly ordered trials of 1.5 minutes at typical walking speed. We also calculated the margin of stability to measure medial-lateral stability. We observed that loading the hips bilaterally with 4 kg had no changes in kinematics, kinetics, dynamic stability, or muscle activity, but above 6 kg, sagittal joint power was increased. Loading the lumbar area increased posterior pelvic tilt at 6 kg and decreased dynamic stability at 4 kg, with many individuals reporting some discomfort. For the unilateral placement, above 4 kg dynamic stability was decreased and hip joint power was increased, and above 6 kg the pelvis begins to dip towards the loaded side. These results show the different effects of weight distribution around the pelvis. This study represents a novel, systematic approach to characterizing

transparency in exoskeleton design (clinicaltrials.gov: NCT05120115).

2 Introduction

In recent decades, lower-body robot exoskeletons have assisted with heavy military loads (Zoss *et al.*, 2006), industrial worker fatigue (Abdoli-E *et al.*, 2006), hospital patient care (Suzuki *et al.*, 2007), and gait rehabilitation for the neurologically impaired (Lerner *et al.*, 2018; Zhang *et al.*, 2017). Their designs vary from kinematic chains to the ground (Esquenazi *et al.*, 2012; Farris *et al.*, 2013; Zeilig *et al.*, 2012) to body held devices (Lerner *et al.*, 2018; Lee *et al.*, 2017), particularly with hip exoskeletons (Zhang *et al.*, 2017; Lee *et al.*, 2016; Di Natali *et al.*, 2019; Yu *et al.*, 2020). Hip exoskeletons weigh between 2.4 kg to 11.6 kg (Chen *et al.*, 2020), with the bulk of this weight coming from their actuators and batteries, whose positions can be arranged to improve user experience. For example, the commercial gait trainer, Samsung Gems (Lee *et al.*, 2016), distributes its 2.8 kg of weight bilaterally with its actuators located on both hips. An exoskeleton designed for above-knee amputees only requires one side of actuation (Ishmael *et al.*, 2019), loading only one hip with 2 kg of weight. The S-Assist L-type exoskeleton elects to load the actuators on the lower back and utilizes cable driven transmission to power both legs (Lee *et al.*, 2017), with the total weight of 14.5 kg. Some designs can reduce their actuators' inertia by driving the exoskeleton remotely, such as a tendon driven knee exoskeleton, which enabled reduced the weight on the knee of 1.2 kg (Sulzer *et al.*, 2009). While low weight is desirable, it is unclear how much weight affects how people walk in an exoskeleton.

An important metric in user experience with robot exoskeletons is transparency, the degree by which a device hinders the movement of the user by gravity, inertia, friction or other resistance (Jarrassé & Morel, 2011). Transparency is altered according to device characteristics such as weight distribution and magnitude (Lerner *et al.*, 2018; Jin *et al.*, 2017; Browning *et al.*, 2007; Rossi *et al.*, 2013). Even though it was found that weight compensation alone was not enough to negate the weight effects of an exoskeleton (Jin *et al.*, 2017), studies systematically investigating how exoskeleton weight affects transparency are few. Browning et al. found that ankle loads increased muscle activation during late stance phase with healthy adults (Browning *et al.*, 2007). Other weight effects studies evaluated the effects of heavy backpack loads from military aged adults (Harman *et al.*, 2000) to school children (Ahmad & Barbosa, 2019). The military study found that backpacks of 20 kg increased the range of motion in the hips, knees, and ankles, while the children study found that backpacks of 15% body weight increased the stance phase duration. Another study focusing on weight placement and symmetry about the torso has shown the placement of 10% body weight can cause destabilization of the wearer during gait initiation (Caderby *et al.*, 2017). However, these studies have not systematically investigated the distribution of the load. A recent study investigated a comparison of weight distribution across the pelvis and thighs in middle-aged adults compared to younger individuals (Vijayan *et al.*, 2022). They found that the amount and distribution of the bilateral load between the pelvis and thighs affects joint loading during walking. However, as exoskeletons vary in laterality, we still lack an understanding of how such laterality of weight distribution affects gait biomechanics.

We tested for the effects of weight magnitude and distribution across the pelvis of healthy individuals while walking on a treadmill. We measured the gait deviations in kinematics, kinetics, stability, and muscle activation on 21 healthy individuals. We varied magnitude within a range commonly found, 4 kg, 6 kg and 8 kg and compared to no additional weight bearing. We varied placement of the weights on the pelvis to be supported bilaterally, unilaterally on the left side, and on the lumbar area of the back. Based on the previous weight studies, we predicted that, 1) increased weight will increase the range of motion in the hip and knees during initial stance phase, 2) unilateral placement will affect stability at all weight conditions, 3) unilateral and bilateral placements will require increased demand from the hips and knees in initial stance phase and the ankle in late stance phase. These findings will lead towards a more principled approach to transparency in exoskeleton design.

3 Methods

3.1 Participants

Twenty-one healthy participants (12 males, 9 females, age 26.8 $\pm$5.57 years, body height 172.8 $\pm$7.58 cm, body weight 65.9 $\pm$8.53 kg) were recruited for this study. Exclusion criteria included relevant musculoskeletal injuries, abnormal gait deviations, and weight bearing restrictions. Prior to the experiment, participants had their footedness evaluated to ensure that they were right footed. The University of Texas Institutional Review Board approved the experimental protocol and subjects were provided informed written consent.

3.2 Experimental Setup and Protocol

Participants were tasked with traversing a treadmill while bearing scuba weights secured with a diving belt (Scuba Choice, Los Angeles CA). Weights of 4 kg, 6 kg, and 8 kg were suspended on the pelvis in three configurations: weight evenly distributed between both anterior iliac crests (Bilateral, BI), weight on the 4th lumbar vertebrae (Lumbar, L), and weight on the non-dominant left anterior iliac crest (Unilateral, UNI). Figure 1 illustrates these placements.

We collected Motion capture marker data with a 13-camera motion capture system and 36 active markers attached to the lower body and torso segment (Phase Space, San Leandro, CA). Ground reaction forces (GRF) were measured through force plates in an instrumented split-belt treadmill (Bertec, Columbus, OH). Surface electromyography data (EMG) (Bortec, Calgary, AL) were collected from the rectus femoris (RF), medial hamstring (MH), tibialis anterior (TA), and lateral gastrocnemius (GAS) of each leg.

Participants walked for 1m30s at a speed of 1.1 m/s for every combination of weight (4, 6 and 8 kg), placement (BI, UNI, L), and no weight (NW) in pseudo-randomized order. The participants were exposed to each condition two times, for 20 total trials. After every five trials subjects were given a break of 2 minutes.

3.3 Data Processing

Using GRFs, we defined heel strike and toe off events, which were then used to identify gait phases. Starting from heel strike, the gait phases were defined as initial stance, mid stance, late stance, and swing, ending the gait cycle with the proceeding heel strike. For each trial, we ignored the first 30 seconds to account for familiarization to the condition. For outlier detection, we removed an individual stride if the waveform exceeded 2 inter-quartile-ranges from the median waveform for more than 40% of the gait cycle. The mean waveform from the last 30 strides was evaluated.

We recorded all biometric data at 960 Hz. First, all these signals were downsampled to 480 Hz. Force plate data were low-pass filtered with 4th order Butterworth filter at 20 Hz. Motion capture data were low-pass filtered with 4th order Butterworth filter at 6 Hz. Surface EMG signals were processed with a high-pass filter of 40 Hz, demeaned, rectified, and low-pass filtered at 4 Hz.

We used motion capture data and GRFs with an open-source musculoskeletal simulation software, OpenSim 4.3 (Delp *et al.*, 2007). We scaled a musculoskeletal model to match the anthropometry of each subject, and then performed inverse kinematics and dynamics for joint angles and moments, respectively.

3.4 Outcome Measures

We contextualised the joint motion for the hips and knees as Range-of-Motion (ROM), the difference between the maximum and minimum joint angles within a given period. We analyzed the effect of weight on the hip and knee sagittal plane motion during initial stance phase. We quantified pelvic tilt and obliquity as the average position within a gait phase.

We used the Margin of Stability (MoS) (Hof *et al.*, 2005; Hof, 2008) as a measure of medial-lateral stability during walking. With body kinematics data, we identified the medial-lateral center of mass (COM) location. We calculated the extrapolated center of mass (xCOM) in Eq. 1, where $\vec{r}$ and $\vec{v}$ is the COM position and velocity projected onto the ground plane and ω_o is the angular eigenfrequency of the physical body as an inverted pendulum. Lastly, MoS was solved for in Eq. 2 by calculating the distance between COP and xCOM at the time of toe off. We analyzed the effect of placement on MoS for both sides.

$$xC\vec{O}M = \vec{r} + \vec{v}/\omega_o \tag{1}$$

$$MoS = C\vec{O}P - xC\vec{O}M \tag{2}$$

Joint power was calculated using joint angular velocity and joint moment data from the OpenSim model and then normalised by the total weight of the participant, including any added weights. We contextualised joint power using the peak values during each gait phase. We analyzed the effect of placement on the peak joint power both hip and knee flexion/extension during initial stance and ankle plantarflexion/dorsiflexion during late stance.

EMG signals were normalized via the mean-dynamic method (Burden & Bartlett, 1999), centering the EMG signal around 1. For data analysis, the EMG signal was integrated along each gait phase to calculate the integrated EMG (iEMG) values. We analyzed the RF muscle activation during initial stance and the GAS muscle activation during late stance.

At the end of the session, participants were asked to identify their least favorite. These results were tabulated according to subject sex as the pelvis kinematics differ between the two sexes (Nguyen & Shultz, 2007).

3.5 Statistical Analysis

R 4.1.1 (2021 The R Foundation for Statistical Computing) was used for statistical analysis. One subject was removed from the dataset due to technical errors with recording the data.

We used a linear mixed regression model (*lme4* 1.1.27.1 (Bates *et al.*, 2015) and *lmerTest* 3.1.3 (Kuznetsova *et al.*, 2017)) with two fixed effects (weight and placement) along with a no weight condition and one random effect (subject) and $\alpha < 0.05$. A Tukey Honestly Significant Difference posthoc test was performed to determine pairwise differences between weights, placements, and their interactive effects.

Based on previous studies (Browning *et al.*, 2007; Harman *et al.*, 2000; Ahmad & Barbosa, 2019; Caderby *et al.*, 2017), we predicted the following gait deviations on the left side. Increased weight would increase sagittal plane range of motion during initial stance phase. The assymetrical distributions, L and UNI, would dip the pelvis orientation towards the weight. UNI placement will especially cause a decrease in the MoS. BI and UNI placements will cause an increase in effort required from the sagittal plane joints.

4 Results

4.1 Lower Limb Kinematics

We found an effect of weight on ROM for hip flexion/extension ($F_{2,371} = 5.28$, $p = .006$), where 8 kg decreased ROM from 4 kg (mean difference = $-0.36°$, $z = -3.16$, $p = .010$) and NW (mean difference = $-0.47°$, $z = -3.07$, $p = .011$): an average drop of 7% ROM. We observed an effect of placement ($F_{3,371} = 8.18$, $p < .001$), with BI placement decreasing hip ROM compared to L placement by 13% (mean difference = $-0.83°$, $z = -3.83$, $p < .001$). L placement had 7% higher ROM than UNI placement (mean difference = $0.47°$, $z = 3.93$, $p < .001$). Additionally, we found an interaction between weight and placement on the effect on hip ROM ($F_{4,371} = 3.27$, $p = .012$). At 6 kg, BI placement had decreased hip ROM by 18% compared to L placement (mean difference = $-1.18°$, $z = -4.358$, $p < .001$), and at 8 kg, L placement had 16% higher ROM than UNI placement (mean difference = $1.07°$, $z = 3.931$, $p = .004$). Tables S1 and S2 provide summary results for left and right hip ROM, respectively.

For knee ROM, we observed an effect of placement ($F_{3,371} = 6.88$, $p < .001$) but not weight ($F_{2,371} = 1.66$, $p = .19$). BI placement increase knee ROM over the both L placement by 5% (mean difference = $1.10°$, $z = 2.63$, $p = .034$) and UNI placement by 11% (mean difference = $2.16°$, $z = 4.16$, $p < .001$). Figure 2 illustrates joint kinematics throughout the gait cycle. Tables S3 and S4 provide summary results for left and right knee ROM, respectively.

4.2 Pelvis Kinematics

Placement had effect on the average position of pelvic tilt ($F_{2,371} = 10.85$, $p < .001$). L placement resulted in lower pelvic tilt than BI (mean difference $= -4.36°$, $z = -4.18$, $p < .001$), UNI (mean difference = $-3.84°$, $z = -3.88$, $p < .001$), and NW (mean difference = $-2.87°$, $z = -3.46$, $p = .002$) placements. Compared to other placements, L placement tilted the pelvis backwards on average $4.10°$. Table S5 provides summary results on pelvic tilt.

We observed an effect of weight placement on pelvic obliquity ($F_{2,371} = 6.04$, $p = .002$). UNI placement caused a $1.34°$ leftward dip in the pelvis from BI placement (mean difference = $-1.34°$, $z = -3.93$, $p < .001$) and a 0.93° dip from L placement (mean difference = $-0.93°$, $z = -3.93$, $p < .001$). Figure 3 illustrates the pairwise differences between placements for hip kinematics. Table S6 provides summary results on pelvic obliquity.

4.3 Stability

We observed an effect of placement on both the left side ($F_{2,370} = 21.10$, $p < .001$). BI placement had 7% increase in MoS to both L placement (mean difference = 0.91 cm, $z = 5.13$, $p < .001$) and UNI placement (mean difference = 1.03 cm, $z = 6.02$, $p < .001$). Table S7 provides summary results on MoS.

Table 1: Margin of Stability [cm]

Side		Placement		
	NW	BI	L	UNI
Left	13.9 ± 1.3	13.7 ± 1.7*	12.8 ± 1.3***	12.7 ± 1.3***
Right	15.0 ± 1.6	14.7 ± 1.7**	13.9 ± 1.5***	15.7 ± 1.4***

Side		Weight [kg]		
	NW	4	6	8
Left	13.9 ± 1.3	13.3 ± 1.8***	13.0 ± 1.3***	12.8 ± 1.3***
Right	15.0 ± 1.6	14.8 ± 1.8**	14.8 ± 1.6***	14.6 ± 1.7***

*denotes a significant difference between factor and NW condition
'*' for $p < .05$, '**' for $p < .01$, '***' for $p < .001$

We found an effect of placement on the right side ($F_{2,370} = 131.3$, $p < .001$). BI placement had 7% increase in MoS from the L placement (mean difference = 0.79 cm, $z = 6.80$, $p < .001$). UNI placement had the highest MoS, 7% higher than BI placement (mean difference = 1.09 cm, $z = 9.34$, $p < .001$) and 12% higher than L placement (mean difference = 1.88 cm, $z = 16.14$, $p < .001$), with an average increase of 10%.

We observed an interaction effect between weight magnitude and placement effects with the right side MoS ($F_{4,370} = 7.59$, $p < .001$). At 4 kg, BI placement had 3% higher MoS than L placement (mean difference = 0.48 cm, $z = 3.198$, $p = .018$), and at 4 kg BI placement had 5% lower MoS than UNI placement (mean difference = −0.80 cm, $z = -5.12$, $p < .001$). At 4 kg, L placement had 8% lower MoS than UNI placement (mean difference = −1.28 cm, $z = -8.38$, $p < .001$). Compared to NW, BI placement at 8 kg decreased MoS by 3% (mean difference = −0.50 cm, $z = -3.25$, $p = .016$), at 4 kg L placement had decrease of 5% (mean difference = −0.74 cm, $z = -4.86$, $p < .001$), but UNI placement had increase of 3% MoS at 4 kg (mean difference = 0.54 cm, $z = 3.52$, $p = .007$). L decreased the MoS on average of 0.21 cm per kg of added weight. Table 1 summarizes the MoS data for both legs.

4.4 Joint Power

We observed an effect on weight placement on peak power for hip flexion/extension during initial stance ($F_{3,371} = 22.7$, $p < .001$). BI placement had 25% higher average hip peak power than L placement (mean difference = 0.22 W/kg, $z = 7.28$, $p < .001$). L placement had 26% lower hip peak power than UNI placement (mean difference = -0.23 W/kg, $z = -6.08$, $p < .001$). Additionally we observed an interaction effect for the weight and placement effects on hip peak power ($F_{4,371} = 3.24$, $p = .012$). At 4 kg, BI placement had 14% increase in hip peak power than L placement (mean difference = 0.12 W/kg, $z = 3.42$, $p = .014$). At 4 kg, L placement was observed to have a 24% lower hip peak power than UNI placement (mean difference = -0.21 W/kg, -5.71, $p < .001$). Tables S8 and S9 provide summary results for left and right hip power, respectively.

Weight placement affected initial stance peak power in the knee ($F_{2,371} = 5.70$, $p < .001$). BI placement had 12% higher knee peak power than L placement (mean difference = 0.12 W/kg, $z = 2.67$, $p = .023$). Tables S10 and S11 provide summary results for left and right knee power, respectively.

We found late stance ankle power was affected by weight placement ($F_{3,371} = 11.9$, $p < .001$). BI placement had 10% higher ankle peak power than L placement (mean difference = 0.41 W/kg, $z = 3.98$, $p < .001$), an average increase of 11%. Figure 4 illustrates the effect of weight placement on joint power over the gait cycle. Tables S12 and S13 provide summary results for left and right ankle power, respectively.

4.5 EMG

We observed an effect of placement on the RF during early phase ($F_{3,371} = 7.54$, $p < .001$). We did not find a difference in muscle activity from L condition to both BI placement (mean difference = 0.20 AU, $z = 1.43$, $p =$

.46), and UNI placement (mean difference = 0.08 AU, $z = 0.34$, $p = .74$). Tables S14 and S15 provide results for left and right RF EMG, respectively.

We also found a significant effect of placement on the GAS during late stance ($F_{2,371} = 5.93$, $p < .003$). BI placement had 8% lower GAS activation than UNI placement (mean difference = −0.12 AU, $z = -3.29$, $p = .003$). L placement similarly had 5% lower GAS activation than UNI placement (mean difference = −0.08 AU, $z = -2.53$, $p = .023$). Tables S16 and S17 provide summary results for left and right GAS EMG, respectively.

We did not find an interaction effect on muscle activation for neither the left RF during initial stance phase ($F_{4,371} = 1.38$, $p = .24$) nor the left GAS during late stance phase ($F_{4,371} = 1.30$, $p = .26$). The surface EMG signals over the gait cycle are illustrated in Figure 5.

4.6 Participant Preference

We tabulated the participants' least preferred configurations in 2 organized by sex. We found that UNI configuration was the least favorite configuration of 80% of the participants (75% of males, 89% of females) and L configuration was the least preferred of 20% of participants (25% of males, 11% of females). BI condition was not mentioned as a least favorite configuration (see Table S18).

Table 2: Least Favorite Placements

Sex	BI	L	UNI
M	0	3	9
F	0	1	8

5 Discussion

The goal of this study was to determine the biomechanical effects of external pelvic loads on healthy adult gait. BI placement exhibited the greatest MoS, but increased power and altered kinematics compared to the NW condition; however, these effects were mitigated at lower weight (4 kg). We found that L placement minimally deviated sagittal kinetics and kinematics. Compared to the NW condition, L placement above 6 kg resulted in excessive posterior pelvic tilt, and above 4 kg had notable decrease in MoS. Also above 4 kg, UNI placement altered sagittal kinematics, increased kinetics, and reduced MoS and pelvic obliquity. Lastly, UNI placement was the only placement to distinctly alter muscle activation, increasing GAS activation. These results provide a novel guide to the inertial effects of weight placement and magnitude within a common range of exoskeleton weights.

BI placement is the most common weight distribution in exoskeleton designs (Chen *et al.*, 2020). Compared to NW in the initial stance phase, hip ROM decreased (0.71°) and knee ROM increased (1.74°), but the decrease in hip ROM was only present after 6 kg (0.99°). The decrease in hip ROM was unexpected since previous backpack studies found that increased weight increases ROM (Harman *et al.*, 2000; Ahmad & Barbosa, 2019). It is possible that backpacks increase posterior pelvic tilt which then requires greater hip ROM (Harman *et al.*, 2000). Indeed, with L placement hip ROM increased proportionally with weight. Given the relatively small changes in overall ROM, there may be little impact in user experience, and further, human gait may prioritize maintenance of kinetics over kinematics (Winter, 1984; Shemmell *et al.*, 2007; Lewis & Ferris, 2011). We expected the increase in joint power compared to the NW and L condition as the weight placement would be directly above the loaded limb. Hip power was increased after 6 kg for NW (6% of peak) and 4 kg for L (4% of peak). This suggests that BI placement at 4 kg may not alter gait significantly from the baseline. In summary, BI placement above 6 kg causes small changes in kinematics and increased joint power, but maintains stability below 8 kg.

L placement represents a newer iteration of hip exoskeletons with cable transmissions (Lee *et al.*, 2017; Chiu *et al.*, 2021). This condition was most similar to previous backpack studies (Harman *et al.*, 2000; Ahmad & Barbosa, 2019), which found increased ROM with the sagittal plane and posterior pelvic tilt. There were no observable changes in ROM across the sagittal plane, however our weights (4-8 kg) are lower than previous studies (6-44 kg). Despite the lack of change with sagittal kinematics and kinetics, 4 participants found L placement the least comfortable. L placement caused near vertical pelvic tilt (-0.43°) as predicted since L placement is an asymmetrical load. We can compare the change in average position to walking on a 10° downhill slope (Leroux *et al.*, 2002). L placement decreases MoS from NW at 4 kg (0.74 cm). To put this in perspective, Peebles et al. found that walking on an oscillating platform reduced MoS_{AP} by 0.6 cm (Peebles *et al.*, 2017). In summary, L placement above 4 kg reduces the MoS of the user, and was observed to have a general effect on posterior pelvic tilt.

UNI placement represents asymmetrical exoskeletons created for amputees (Ishmael *et al.*, 2019) or patients with hemiplegia (Kawamoto *et al.*, 2009). UNI placement caused a dip in pelvic obliquity (1.44° towards the left), as expected. UNI placement was also found to reduce MoS from the NW condition (1.3 cm), with significant effects occurring at 4 kg (1.0 cm). This suggests MoS in UNI placement, even at low weight, is comparable to a person with impaired walking ability (Peebles *et al.*, 2016). At 4 kg, UNI placement increased hip, knee and ankle joint power. Given the changes with BI placement at 8 kg which bears equivalent weight on the single limb, this effect is expected. UNI placement at 4 kg also resulted in higher GAS activation than both BI (8% increase) and L (5% increase) placements. We expected UNI placement to put more emphasis on the muscle activation of the loaded left side for the same reasons as BI condition. UNI placement was the least favorite placement by 17 of the 21 participants. In summary, even at our lowest weight setting, UNI placement reduces stability, alters kinematics, increases joint power and muscle effort.

This study had several limiting factors. The duration of each condition was short (1m30s). We have found that biometric data tends to stabilize at around 45 seconds which was consistent with previous literature (Noble & Prentice, 2006). Initial pilot testing used trial duration as long as 5 minutes, but there were no observable differences with shorter trial time. Using shorter trials allowed us to explore more experimental factors prior to fatigue. Further work is needed to examine the long-term adaptation to weight bearing. We modeled weight distribution as concentrated masses, which may not accurately reflect all hip exoskeleton weight distribution. Often, the human interface of a hip exoskeleton extends more distally along the thigh, which is shown to increase the weight effects (Browning *et al.*, 2007; Jin *et al.*, 2017). In this study we tested the analog of actuator placement along the hips, since actuators are the largest source of mass on a hip exoskeleton. With our oversimplified model, we still observed weight effects on gait, meaning that in a more distally distributed exoskeleton will carry these weight effects and more. The population observed was primarily healthy young adults. Previous work has shown that middle-aged adults react differently than younger ones under load (Vijayan *et al.*, 2022). Additional work is needed to examine the effects of weights on those most likely to use exoskeleton assistance, such as older individuals. We recruited right dominant individuals and loaded their non-dominant sides. Since an assistive exoskeleton would likely be used primarily to address an impaired limb, we chose this combination because it was more relevant than loading the dominant side. Thus with the current data, we are unable to make conclusions on how dominant side loading would affect gait biomechanics.

6 Conclusion

In this investigation, we observed the weight effects of hip exoskeleton configurations to determine best practices for transparent design. We found that the placement of the heavy components of the exoskeleton often causes more gait deviation than the weight of these components. We observed that lateral placement on the pelvis, such as in bilateral and unilateral placements, changed sagittal plane kinematics. Lumbar and unilateral placements changed pelvis orientation and decreased stability. Our findings indicate that bilateral placement, especially below 6 kg is the most comfortable and has the least effect on gait, whereas unilateral placement has measurable effects even at our lowest level (4 kg). These findings outline the weight effects of common load placements and provide insight for both mechanical and controller designs of transparent hip exoskeletons.

7 Conflict of Interest Declaration

None of the authors of this submission have any conflicts of interest to declare.

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8 Figure Legends

Figure 1. Weight placements during an experiment. Also shown are the motion capture LEDs, EMG electrodes, and split belt treadmill.

Figure 2. Kinematics data of the Left Hip and Knee, grouped by weight condition. On the y-axis, the positive and negative directions represent extension and flexion respectively. For this measure, the solid colored lines and the shading represent the mean measure and its standard deviation respectively. The dotted grey lines illustrate a change in gait phase, indicating the gait phases of early stance, mid stance, late stance, and swing, The colored bars on the bottom represent pairwise significance with the left bar having a larger measure than the right bar. Pairwise results for this figure reflect a significant change in RoM for a specific gait phase.

Figure 3. Kinematics data of the Pelvis based on the left gait cycle, grouped by placement condition. On the y-axis for pelvic tilt, the positive and negative directions represent posterior and anterior tilt respectively. On the y-axis for pelvic obliquity, the positive and negative directions represent a shift downwards towards the right and left sides respectively. Pairwise results reflect a change in average position for a specific gait phase.

Figure 4. Joint Power Data of sagittal plane motion, grouped by placement. On the y-axis for the hip and knee, the positive and negative directions represent extension and flexion respectively. On the y-axis for the ankle, positive and negative directions represent the plantarflexion and dorsiflexion respectively. Pairwise results reflect a significant change in peak power for a specific

gait phase.

Figure 5. Surface EMG Data of the left rectus femoris (RF) and lateral gastrocnemius (GAS). Pairwise results reflect a significant change in iEMG values for a specific gait phase.