# Actuation Timing Perception of a Powered Ankle Exoskeleton and Its Associated Ankle Angle Changes During Walking

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***Abstract*— Robotic ankle exoskeletons have the potential to extend human ability, and actuation timing serves as one of the critical parameters in its controller design. While many experiments have investigated the optimal actuation timing values to achieve different objective functions (*e.g.* minimizing metabolic cost), studies on users' perception of control parameters are gaining interest as it gives information on people's comfort, coordination, and trust in using devices, as well as providing foundations on how the sensorimotor system detects the exoskeleton behavior changes. The purpose of this study was to evaluate people's sensitivity to changes in exoskeleton actuation timing and its associated exoskeleton ankle angle changes during walking. Participants (n =15) with little or no prior experience with ankle exoskeletons were recruited and performed a psychophysical experiment to characterize their just-noticeable difference (JND) thresholds for actuation timing. Participants wore a bilateral active ankle exoskeleton and compared pairs of torque profiles with different actuation timings and low peak torque (0.225 Nm/kg) while walking on the treadmill. The mean timing JND across participants was 2.8±0.6% stride period. Individuals exhibited different sensitivity towards actuation timing, and their associated exoskeleton ankle angle changes also varied. The variance in ankle angle changes might be explained by their differences in ankle stiffness and different ankle torques provided during walking. The results provide insights into how people perceive the changes in exoskeleton control parameters and show individual differences in exoskeleton usage. The actuation timing JND found in this study can also help determine the necessary controller precision.**

***Index Terms*— Ankle exoskeleton, actuation timing, human perception, biomechanics.**

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## I. Introduction

ANKLE exoskeletons have gone through years of developments and have shown the potential to augment human ability [1]. Controllers have been designed to help with different individual tasks, and the tuning of their associated parameters is needed to achieve desired human-system performance design goals, such as minimizing a specific objective function (*e.g.* metabolic cost [2]–[4]). As one of the major controller parameters, actuation timing has a substantial effect on the device's performance. Optimizing exoskeleton actuation timing can significantly reduce metabolic cost, though the optimum might vary dependent on several factors (*e.g.* the torque profile, the use of exoskeleton or prosthesis, the unilateral or bilateral assistance, etc.) [4]–[7]. Ingraham *et al.* [8] found that individuals were consistent in identifying their preferred actuation timing, suggesting that the selection of timing can strongly affect user comfort. Muscle activation and walking mechanics can also be affected by the changes in actuation timing [9], [10]. However, there is still a lack of understanding in the underlying perception of relevant exoskeleton control parameters, which could affect users' comfort, coordination, and even trust in using wearable devices. An important consideration in exoskeleton control architectures is to minimize the perceived interaction forces between humans and exoskeletons [11]. By quantifying people's ability to perceive the changes in exoskeleton actuation timing, we can gain more insights on how to improve the controller design.

Understanding human perception towards exoskeleton parameters can be beneficial in several aspects. First, it provides a foundation on how people perceive changes in external devices, which can instruct the improvement of the system to be more transparent to users during human-robot interaction. Second, investigating people's sensitivity can help determine the precision requirements of exoskeleton controllers. A controller that has variability within the perception of the user may improve user experience. By setting precision parameters to meet the lowest threshold, we can accommodate users across the perceptual spectrum. Third, setting appropriate step values inspired by the perception threshold has the potential to improve the implementation of optimization algorithms in how the solution space is investigated. Currently, the state of the art human-in-the-loop optimization algorithm takes about 2 minutes to estimate the metabolic cost for each control law [12], and the total training time can be significant for multi-gait assistant patterns as optimal values might vary under different conditions [4], [13]. Knowledge of perception thresholds can be used to define step values for discretizing the parameters in the design space and limit indistinguishable control law evaluations.

Human perception can be evaluated by conducting psychophysical experiments. The two-alternative forced choice (2AFC) task is commonly used in assessing user perception [14]–[18]. In this method, participants are asked to compare a series of stimuli pairs and select the one with the larger magnitude. Then the responses are used to evaluate their sensitivity. However, 2AFC is not always applicable, such as the situation where participants cannot detect relative magnitudes, but can only tell if the two choices are the same or not. The yes-no task simplifies the requested information from the participant by asking them to compare if two stimuli are the same or not. For example, in our pilot experiment [19], users could have difficulty specifying relative timing even when they could perceive a difference. This difficulty may arise from the inability to distinguish the exoskeleton behavior from the gait cycle features, and people's uncertainty of how 'early timing' and 'late timing' of the exoskeleton relate to the goal of the system and their expectation of timing as they walk. However, compared to 2AFC, the yes-no task will be less accurate if participants exhibit a strong bias toward any response alternatives. To address this bias, Green [20] proposed a psychometric function that incorporated false-alarm rate to gain a more accurate threshold estimation. Catch trials were introduced to calculate the false-alarm rate [21], and have been applied previously in acoustic perception [22].

In this study, we measured participants' ability to differentiate actuation timings of an ankle exoskeleton that supports gait. We determined the just-noticeable difference (JND) threshold for all participants as the minimum timing changes that can be reliably perceived. We further investigated the ankle angle changes associated with the different actuation timings. Knowing the actuation timing perception of ankle exoskeletons and its influence on walking mechanics can provide a foundation for future controller development and implementation.

## II. Methods

### A. Participants

Fifteen participants (13 males and 2 females; age: 23.9±4.2 years; mass: 72.3±10.2 kg; height: 1.76±0.08 m, mean±standard deviation) were recruited to participate in the study. Participants were all healthy, did not have mobility limitations, and did not have leg or foot injuries within three years. All participants had little or no prior experience with active ankle exoskeletons: 13 out of 15 had never worn an ankle exoskeleton before, one had 1-2 hours, and one had about 12 hours experience due to their previous participation in a separate exoskeleton study with the same system. All participants provided written informed consent, approved by the University of Michigan Institutional Review Board. (HUM00181678, approval date: 09/10/2020)

### B. Exoskeleton

Participants wore a bilateral ExoBoot [Dephy, Inc., Maynard, MA] (Fig. 1) for walking augmentation [23]. The

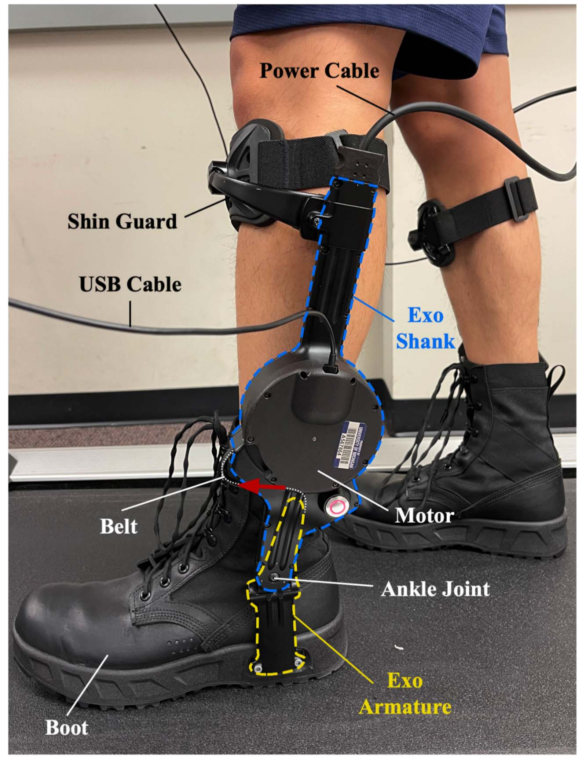
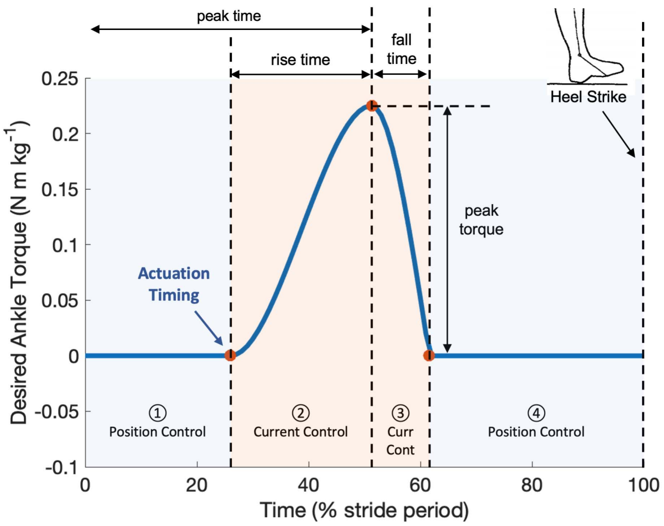


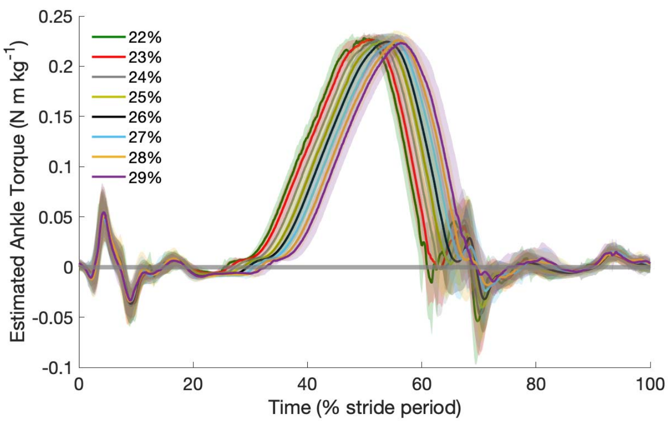

Fig. 1. The bilateral Dephy ExoBoot used in this study. The Exo shank was attached to the lower leg with the shin guard just below the knee. The motor rotated the Exo armature (along with the boot) around the ankle joint counterclockwise (left ExoBoot from the picture view) by pulling the belt (red arrow direction), providing plantarflexion torque to the users during the push-off portion in each gait cycle. Only plantarflexion torque can be generated as the force transmission only happens when the belt is pulled.

Exo shank contains a motor and is attached to the participants' lower leg with the shin guard affixed right below the knee. The motor connects the Exo armature via a belt. When the belt is pulled by the motor (indicated by the red arrow), the Exo armature and the boot rotate counterclockwise around the ankle joint as a whole, and a plantarflexion torque is provided to users. Only plantarflexion torque can be generated as the force transmission only happens when the belt is pulled. Participants wore a battery pack (not shown in the figure) to power the ExoBoot. The ExoBoot was connected to a Raspberry Pi (Model: 4B, Cambridge, UK) through the USB cables for the control and data recording. The data sampling rate was 1000 Hz. The ExoBoot contains several onboard sensors that measure the acceleration and angular velocity of the Exo shank, motor current and voltage, ankle angle, and velocity.

The ExoBoot helps people walk by providing plantarflexion torque to the ankle during push-off. The torque profile used in this study was originated from Zhang *et al.* [4]. In their work, the torque profile had four parameters: peak torque, peak time, rise time, and fall time (Fig. 2(a)). In our experiment, only the peak time was a variable, and the other three parameters were held constant to all participants throughout the experiments. The rise time and the fall time were set to be 25.3% and 10.3% stride period respectively, according to the average optimal values in minimizing metabolic cost [4]. The peak torque




(a) Desired Torque Pattern




(b) Estimated Torque Patterns

Fig. 2. The ankle torque profile provided by the ExoBoot in this study. (a) The desired torque pattern. The torque was determined as a function of time, normalized to stride period, and was divided into four regions. No torque was applied in region ① and ④. The torque profile in region ② and ③ was a cubic spline parameterized by four parameters: peak torque, peak time, rise time, and fall time. The actuation timing was the start point of the torque profile, as well as the transition between the region ① and ②. The heel strike was defined as 0% stride period. (b) The estimated ankle torque patterns under different actuation timings of a representative participant. The ankle torque was estimated from the recorded motor current (calculation details are in Section II-C.2). The estimation only held in region ② and ③ (current control), as the current in region ① and ④ (position control) was generated to maintain the 'no slack' status instead of providing torque. The shaded regions represent the associated standard deviation.

was set to be 0.225 Nm/kg due to the ExoBoot mechanical constraints to incorporate participants up to 100 kg. To be consistent with other literature [6], [7], we used actuation timing instead of peak time in this paper. The actuation timing, which was the timing of exoskeleton actuation onset (Fig. 2(a)), was calculated as:

$$\text{Actuation Timing} = \text{Peak Time} - \text{Rise Time} \tag{1}$$

Time was normalized to stride period. A stride period started from the heel strike of one foot and ended at the subsequent heel strike. We detected the heel strike using ankle angular velocity measured by the ExoBoot. The current stride period was estimated as the average of the previous three stride periods.

## C. Control Scheme

Each gait cycle was divided into four regions, as shown in Fig. 2(a). Region ① was from the heel strike to the onset of plantarflexion torque. Region ② was as the torque ascended. Region ③ was as the torque descended. Region ④ contained the remainder of the gait cycle to the next heel strike. Region ① and ④ used position control, and region ② and ③ implemented current control.

*1) Position Control:* In the position control scheme, the ExoBoot was commanded by providing target motor encoder values. This control scheme aimed to ensure the belt remained in a 'no slack' status without providing any torque. The 'no slack' status permitted a faster response when force generation was desired, thus supporting a more accurate actuation timing. Under the 'no slack' status, each exoskeleton ankle angle ($\theta_a$) corresponded to a motor angle ($\theta_m$) value. The desired motor encoder value was then determined by monitoring the exoskeleton ankle angle state as measured by the ExoBoot. This mapping was empirically determined by fitting a 4th order polynomial ($\theta_m = P(\theta_a)$).

*2) Current Control:* In the current control scheme, the desired motor current ($I_m$) was calculated using the desired motor torque ($T_m$): $I_m = T_m/k_t$, where $k_t = 0.14$ Nm/A is the q-axis torque constant [24]. Assuming no power loss in the belt transmission, the relation between ankle and motor torque is expressed as:

$$T_m = T_a \times \frac{\omega_a}{\omega_m} = T_a \times \frac{\partial \theta_a}{\partial \theta_m} = T_a \times \frac{1}{P'} \tag{2}$$

where $P$ is the empirically determined 4th order polynomial, $T_a$ is the desired ankle torque, $\omega_a$ is the ankle angular velocity, and $\omega_m$ is the motor angular velocity. The minimum current in current control scheme was set to 1200 mA to overcome the cogging torque and stiction.

## D. Experimental Protocol

*1) Familiarization:* Participants were given 5-10 minutes walking on the treadmill to get familiar with the ExoBoot. The treadmill speed was 1.25 m/s, and the ExoBoot's actuation timing was a constant 26% stride period during this familiarization process. In our pilot test, 26% stride period was the average preferred actuation timing selected by the participants [19]. One participant did not adapt to walking with the ExoBoot during the familiarization time and did not continue with the experiment. Though we investigated changing the treadmill speed and the actuation timing to accommodate this participant, he still felt uncomfortable walking, as his preferred timings for left and right legs were not the same, which was beyond our hardware setting.

*2) Perception Test:* After a two-minute rest, participants started the actuation timing perception test. Participants walked on the treadmill with the ExoBoot. The treadmill speed was 1.25 m/s.

We implemented a yes–no procedure and the method of constant stimuli to measure participants' perception towards actuation timing. During each trial, a pair of timings were presented to participants sequentially: a comparison timing and a reference timing. The order of these two timings within each trial was random. Participants walked in each timing for 5 strides — 10 strides in a single trial — and were asked 'Are the two timings equal or different?' at the end of the trial. Participants were notified when each timing started. Participants' responses could be either 'equal' or 'different'. After the response, participants were given an 8-second interval until the next trial started. During the interval, the ExoBoot was still providing the augmentation from the reference timing, but the participants were told to ignore this period for their comparison.

Catch trials — trials whose comparison timing equaled the reference timing — were introduced to better estimate the false-alarm rate [21], which was the probability of a 'different' response when the comparison timing and the reference timing were the same. This value will affect the calculation of the perception threshold, as will be discussed in the next section. Each trial had a 25% probability to be a catch trial.

The reference timing was constant throughout the experiment (26% stride period), while the comparison timing varied across trials. The simple up-down method was used to determine the following non-catch-trial comparison timing: if the answer was 'different', the comparison timing in the next non-catch-trial would be a $\Delta$ ($\Delta = 1\%$ stride period) closer to the reference timing; if the answer was 'equal', the next non-catch-trial comparison timing would be 1% further from the reference timing. A $\Delta$ was set to be 1% stride period due to the resolution in stride duration prediction. The initial comparison timing was $3\Delta$ away from the reference timing. Participants were asked to complete 9 sweeps: a sweep contains comparison timing changes in two directions: increasing and decreasing, as shown in Fig. 3. The test was conducted twice to approach the reference timing from both above and below. Therefore, two timing JNDs were determined for each participant, named early timing JND and late timing JND, respectively. Early and late timing JNDs were analyzed separately as there could be differences in the effect based on the phase of gait.

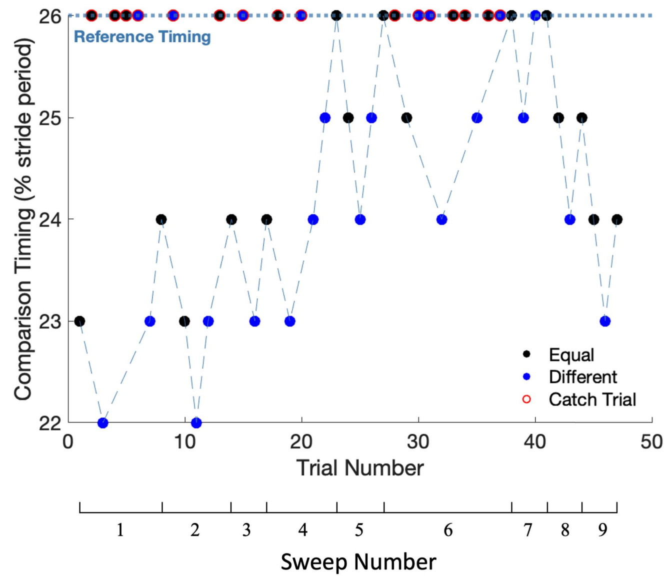



Fig. 3. An example of a participant's perception test where the comparison timing approached the reference timing from a lower value, which represents the early timing condition. Each trial contained a reference timing and a comparison timing. The figure presents the comparison timing of each trial (dots) as the reference timing was always the same (26% stride period, horizontal dashed line). Each trial had a 25% probability to be a catch trial (red circle), in which the comparison timing equaled the reference timing. The comparison timing in the subsequent non-catch-trial was determined based on the participant's previous response: 'Equal' (black dots) - a $\Delta$ (1 % stride period) further from the reference timing; 'Different' (blue dots) - a $\Delta$ closer to the reference timing. Each participant completed 9 sweeps (a sweep contains comparison timing changes in two directions: increasing and decreasing) in a single test.

### E. Data Analysis

*1) Actuation Timing JND:* The responses of each trial were used to fit the psychometric curve. The psychometric function introduced by Green [20] for the yes-no task was used in this study to incorporate the effect of false-alarm rate:

$$P(\text{different}) = \alpha + (1-\alpha)\frac{1}{1+e^{-k(x-m)}} \quad (3)$$

where $\alpha$ is the false-alarm rate, $k$ determines the slope of the psychometric function, $m$ is the mean of the logistic and is set to be the JND, corresponding to the 'different' probability of $(1+\alpha)/2$, which is the halfway point of the best performance (100%) and the chance performance ($\alpha$). The logistic function was fitted using the maximum likelihood criterion [25]. The fitting was performed in MATLAB (MathWorks, Natick, MA, USA).

As the actuation timing is represented in the percentage of the current stride duration (% stride period), the error in the stride duration prediction will lead to an error in the desired actuation timing. Therefore, in the data analysis, we used the measured actuation timing for the curve fitting instead of the desired actuation timing. The measured actuation timing was calculated using the following equation:

$$AT_{mea} = AT_{des} \times \frac{t_{est}}{t_{mea}} \quad (4)$$

where $AT_{mea}$ is the measured actuation timing, $AT_{des}$ is the desired actuation timing, $t_{est}$ is the estimated stride duration, and $t_{mea}$ is the measured stride duration. The measured actuation timing for each trial was then the average of the timings in its corresponding 10 strides (5 strides for the left foot and 5 strides for the right foot) and rounded to the nearest integer to eliminate the influence of a lower number of samples at higher timing differences. During the fitting, We removed trials with standard deviations larger than 1.5% stride period, as a large variation within a trial would affect participants' responses.

A sample psychometric curve fitting is shown in Fig. 4. Trials with the same comparison timing were summarized to a single point, and the corresponding 'different' response probability was calculated. In this study, the JND represented the smallest change in actuation timing that can be reliably perceived by participants. A lower JND value corresponds to

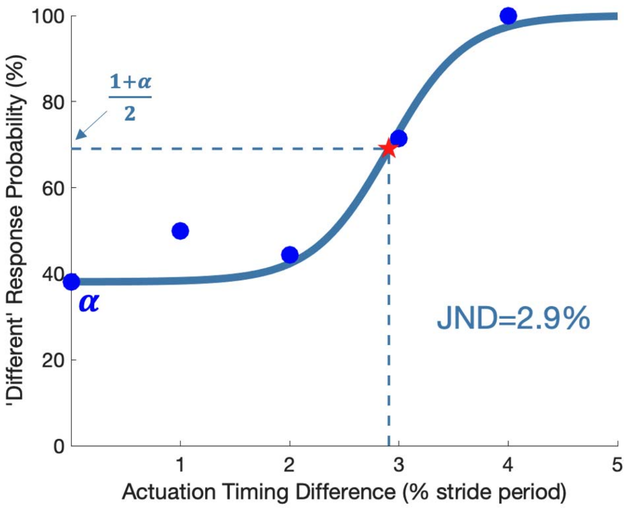



Fig. 4. A sample psychometric curve fitting. The x-axis is the difference between the comparison timing and the reference timing. The y-axis represents the portion of trials that participants responded 'different'. The JND was the timing corresponding to the 'different' response probability of $(1+\alpha)/2$ ($\alpha$: false-alarm rate). The curve was fitted using the maximum likelihood criterion with all the trial data points.

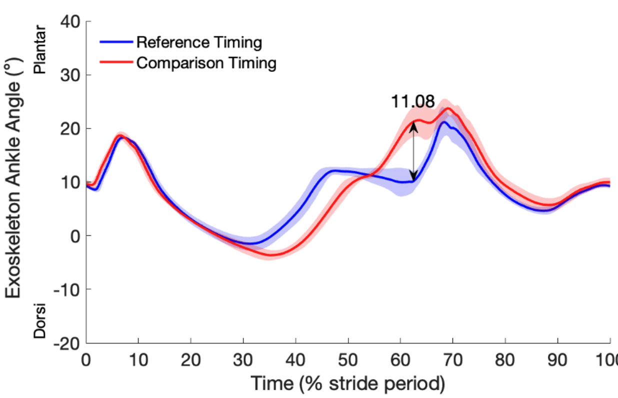



Fig. 5. An example of calculating the maximal exoskeleton ankle angle difference for a comparison timing, which was the largest deviation between two mean stride profiles. The shaded regions represent the associated standard deviation. (plantarflexion[+], dorsiflexion[−]).

a better perception. We performed paired t-test to compare the early and late timing JND.

In addition, we estimated the motor execution delay by calculating the actual time (% stride period) when the estimated ankle torque crossed 0 in region ② in Fig. 2(b).

*2) Kinematics:* Kinematic data were recorded to study the ankle angle changes under different actuation timings. The mean exoskeleton ankle angle profiles for each actuation timing were generated by averaging all strides of the same timing. The mean exoskeleton ankle angle difference profiles between each comparison timing and the reference timing were the average across trials with the same comparison timing. The maximal exoskeleton ankle angle difference was calculated for each actuation timing by finding the maximal difference between the two mean stride profiles, as shown in Fig. 5. We then calculated the maximal exoskeleton ankle angle difference as the function of actuation timing to examine the magnitude people altered their gait patterns as exoskeleton

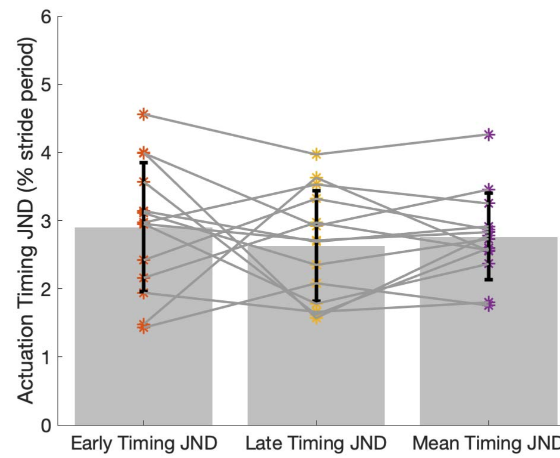



Fig. 6. The mean and standard deviation of early and late timing JND. The mean timing JND is the average of early and late timing JND. The grey lines were the JNDs for each participant. There was no significant difference between the early and late timing JND values.

behavior changed. We hypothesized that this magnitude varied across the participants as some users might be more easily influenced by the exoskeleton changing behavior. We also examined if the participant weight played a role in the estimated ankle angle magnitude as the torque was scaled by mass.

## III. Results

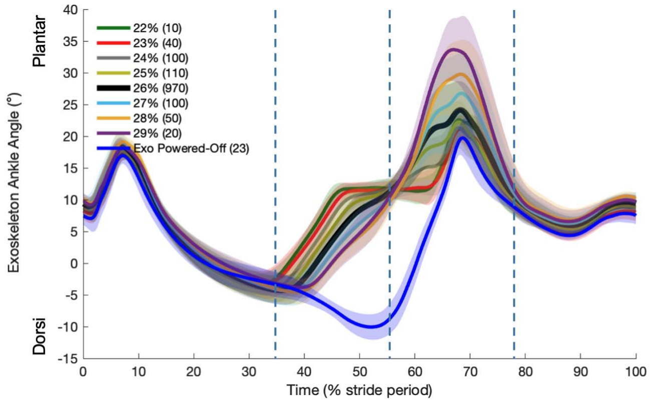
### *A. Actuation Timing JND*

The mean early and late timing JND was 2.9±0.9% and 2.6±0.8% stride period, respectively. The mean timing JND was 2.8±0.6% stride period. There was no significant difference between the early and late timing JND values (paired t-test: $t_{13} = 0.85$, $p = 0.41$, Fig. 6).

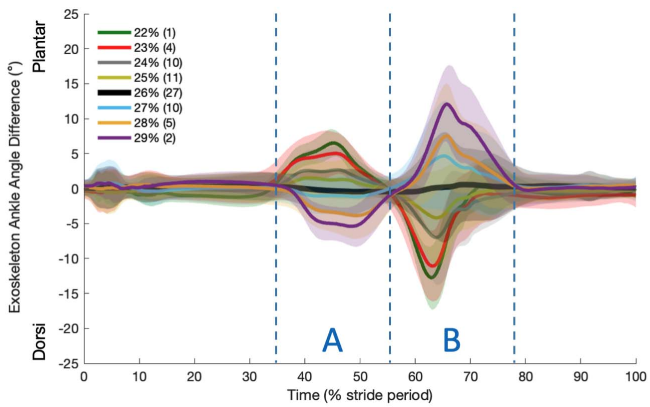
The mean error between the estimated stride duration and the measured stride duration was 0.0±2.0% stride period for all strides across participants. The mean absolute error was 1.4% stride period. Assuming the error accumulated linearly within each stride, the mean absolute error for the actuation timing would be about 0.4% stride period. Across all participants, 26 out of all 1124 trials were removed as the timing standard deviations were larger than 1.5% stride period. In addition, the motor execution delay was around 3.7% stride period on average.

### *B. Kinematics*

An example of a participant's stride profiles is shown in Fig. 7(a). The mean stride profiles under each actuation timing were plotted, as well as the profile of normal walking with the ExoBoot powered-off. The mean exoskeleton ankle angle difference between each comparison timing and the reference timing (26% stride period) are shown in Fig. 7(b). This representative participant highlights that for early actuation timing, the ankle was more plantarflexed during the terminal stance phase (region A, Fig. 7), and was more plantarflexed during the pre-swing and initial swing phase (region B, Fig. 7) with late actuation timing.




(a) Exoskeleton ankle angle profile under different actuation timings




(b) Exoskeleton ankle angle difference with respect to the reference timing

Fig. 7. Kinematics. (a) An example of a participant's exoskeleton ankle angle profiles under different actuation timings. The number in parenthesis indicated the number of strides taken in each timing. (b) The exoskeleton ankle angle difference profiles between each comparison timing and the reference timing. The averages were taken for all the trials that had the same comparison timing. The number in parenthesis was the number of trials for each timing. Region A is the terminal stance between around 35-55% stride period, and region B is the pre-swing and initial swing between around 55-78% stride period. (plantarflexion[+], dorsiflexion[−]).

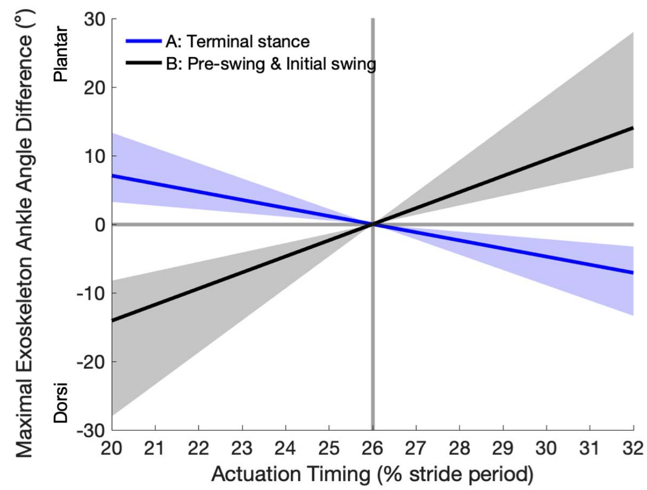



Fig. 8. The linear correlation between actuation timing and maximal exoskeleton ankle angle difference in both stance and swing phases. The mean (solid line) and range (shaded region) of the slopes were across all participants. The slope represents how easily the participant was influenced by the changes in exoskeleton actuation timing, reflected by their maximal exoskeleton ankle angle difference with respect to the reference timing. (plantarflexion[+], dorsiflexion[−]).

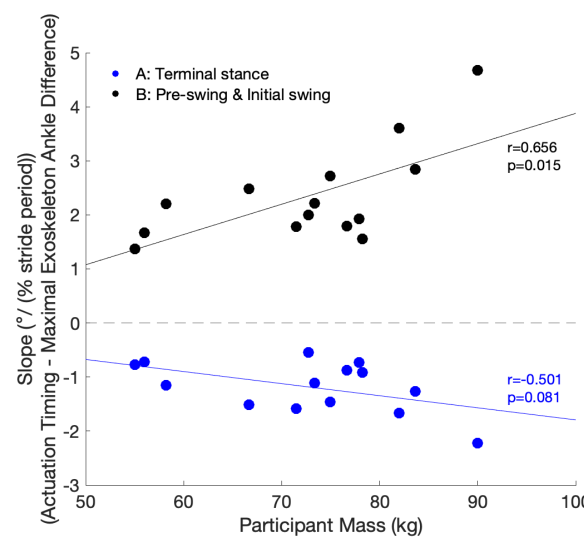



Fig. 9. The linear correlation between the slope (actuation timing - maximal exoskeleton ankle angle difference) and the participant mass. (plantarflexion[+], dorsiflexion[−]).

The maximal exoskeleton ankle angle difference was plotted as the function of the actuation timing (Fig. 8). A strong linear correlation was found between these two parameters. During the terminal stance, the maximal exoskeleton ankle angle difference decreased as the actuation timing increased ($r < -0.922$, $p < 0.002$ for all participants). When participants were in the pre-swing and initial swing phase, the maximal exoskeleton ankle angle difference increased as the actuation timing increased ($r > 0.952$, $p < 0.001$ for all participants). The slope of this linear relation had a strong significant correlation with participant mass during region B, and moderate but not significant correlation in region A, as shown in Fig. 9 (A: terminal stance: $r = -0.501$, $p = 0.081$; B: pre-swing & initial swing: $r = 0.656$, $p = 0.015$).

## IV. Discussion

Actuation timing has been an important parameter for exoskeleton application. While previous studies investigated the actuation timing in an optimization to minimize specific cost functions, such as metabolic cost, this experiment studied the user's perception of this control parameter. By measuring participants' sensitivity towards the change in exoskeleton actuation timing, we determined the just-noticeable difference (JND) thresholds across participants. We found that people's ankle angle profiles varied as the actuation timing changed, and the variances were also different between participants.

*1) Actuation Timing JND:* While we observed differences in the kinematics caused by the early and late actuation timings, the timing JNDs were similar. The mean timing JND of 2.8±0.6% stride period determined the minimum detectable change in the exoskeleton actuation timing. Individual difference in exoskeleton usage was observed. The lowest boundary (1.8% stride period) can be used to guide the design and control of wearable robotics, such as setting up the precision requirements for control parameters and setting appropriate step values to support parameter discretization for optimization algorithms.

Participants reported that in some trials they were unsure if they should respond 'equal' or 'different' because they felt like the two timings were similar. It was unclear what each participant's strategy was in this situation, which may lead to a biased response (participants' tendency to report 'different' or 'equal' when they were unsure). The mean false-alarm rate was 34±11%, which indicated the existence of response bias towards saying 'different'. The high false-alarm rate can also be influenced by the error in the exoskeleton actuation timing (0.4% stride period), which may have led to people perceiving that the timing had changed when it was supposed to remain the same. The natural variance in a participant's gait pattern could also affect perception as there would be a difference in the relative exoskeleton behavior as the participant's gait cycle features change. The high false-alarm rate will cause a biased estimation of perception threshold (larger than it actually is) [21], and results in a lower number of samples at higher timing differences in the simple up-down method, which also can increase estimation error. Future studies that require a more accurate timing may adopt other adaptive methods, such as weighted up-down [26] to increase the precision of the estimation.

Due to the method we used, the heel strike estimate was about 2.0% stride period earlier than the ground truth as assessed in preliminary studies. However, there was also a motor execution delay, which was about 3.7% stride period, resulting in the torque provided slightly later than the value indicated. This error will not affect the conclusions of the study, which are based on relative timing, but should be noted when comparing to other methods in the literature.

*2) Kinematics:* As seen in Fig. 7, participants tended to start the plantarflexion early with the early actuation timing because of the applied external torque. The late actuation timing would result in a more plantarflexed ankle position during the swing phase, as the exoskeleton torque was still on after toe-off.

The ankle angles in this paper are reported as exoskeleton ankle angles as this parameter was measured by the ExoBoot directly. There can be an offset between the exoskeleton ankle joint and the biological ankle joint [27]. Therefore, the underlying biological ankle angle pattern might have small differences from the measurements in Fig. 7(a). However, the observed trend should be similar as the movement of the exoskeleton relative to the leg can be assumed to be a function of the torque and gait phase. As torque was also determined by the gait phase, the offset could be regarded as a constant value for each gait phase. Therefore, the biological ankle angle difference (Fig. 7(b)) would be similar as it was the relative value. Direct measurement of the biological ankle joint could be included in future experiments for a more direct evaluation.

Individual participants had different ankle responses to the same actuation timing. As shown in Fig. 8, the change in actuation timing results in different ankle angle changes, with the larger slope corresponding to the larger ankle angle differences. From the figure, we can see that the slope varies across participants, with the largest slope about 3 times the smallest slope. This difference may be due to participants' ankle stiffness. With a more compliant ankle, a small change in actuation timing results in a larger magnitude biomechanical change. We also note that the average slope of the stance phase (1.18°/(% stride period), absolute value) was smaller than that of the swing phase (2.35°/(% stride period)), which might be explained by the ankle stiffness change within a gait cycle. As reported in Lee *et al.* [28], ankle stiffness is much larger in the stance phase compared to the swing phase and is largest during the terminal stance. The peak torque is another factor that affects the slope. Fig. 9 shows that participants with larger mass were more easily influenced by the actuation timing change. A potential explanation is that though the normalized torque was the same, the actual torque applied differed. People's resistance to external perturbations might not be strictly linear to their weight.

*3) Limitations and Future Work:* In this study, the selection of reference timing was based on the preferred timings of two pilot participants [19] and was set as a constant value throughout the experiment, as we found that people's criteria for 'preferred timing' varied. Some participants might regard it as being the most transparent, while others thought they should feel the applied torque to ensure the exoskeleton was actually 'helping' them. Using the participant's preferred value as the reference might result in an alternate estimation of JND. Future studies can investigate estimating JND for different reference actuation timings to further understand people's perception of exoskeleton behavior. In addition, future studies on training to use exoskeletons should also consider these different perspectives on preference and evaluate how designed usage aligns with different expectations.

A more accurate stride duration prediction method is needed in the future for a more precise estimation of the threshold, as it can decrease the variation between strides and also reduce the false-alarm rate. A smaller $\Delta$ can be selected to achieve a more precise threshold estimation. While the current study examined a young healthy sample, additional data that considers the effects of gender, age, and pathology is warranted. People with expertise wearing exoskeletons may also have different perception of changes in actuation timing, and these changes with experience should be further studied. The parameters selected defining the torque profile, including the magnitude may affect the JND and should be examined. The peak torque used in this study was around 0.225 Nm/kg, which was lower than values from the optimization study [4], which ranged between 0.6-0.8 Nm/kg. In addition, all timing elements were changed simultaneously in this study, but some may play a more important role (*e.g.* peak time might be more critical than rise time and fall time). Future studies should examine the perception of a single timing parameter at a time (*e.g.*, changing peak time while fixing start time and end time).

As participants in this study were given 5 strides after each actuation timing change to perceive the difference, we regarded the results as people's perception within transition periods as opposed to perception within steady-state. Understanding people's perception during transitions is important for consecutive strides to be perceived as similar. As an example, errors in heel strike detection and stride duration prediction will both lead to the shift of the whole torque profile and cause the change in actuation timing in the following stride. A desirable design would be such that small changes in stride parameter estimation would not be perceived, while a larger change indicative of a mode change would be perceived. As seen in this study, the mean absolute error in stride duration prediction was 1.4% stride period when walking on the treadmill. The error in stride timing will be dependent on the method for calculation and environment of usage, which can be even larger. Controller development should define specifications with precision such that people do not feel like the system is changing within a given mode to support people's trust in the device. In addition, For applications like multi-controller usage in multi-tasks [29], smooth transitions between task modes could be possible as long as the changes across steps are below the perceived threshold. For situations where transitions require a large change in parameter values and the design goals are for the transition not to be perceived, there is a potential to investigate the JND in a continuous manner. For example, the step value could be decreased and each step modulated until the participant notes a difference. The overall threshold of perceiving small continuous changes might be larger than observed in the current study that examined discrete changes. People's perception in steady-state may also be different as individuals take different strategies during early adaptation [30]. However, studies that examine steady-state perception will take a much longer time as the adaptation can take tens of minutes for each timing [31] and care should be taken in comparisons with two timings across intervals of tens of minutes as human memory may not be reliable for this perception task.

Individual differences in exoskeleton usage are also important as the strategy can impact the efficacy of the system [32]. Understanding how people perceive, coordinate with, and respond to exoskeletons can support the design of adaptive controllers in the future to accommodate users' changing behavior in the long term [33]–[35], or develop training methods to help people adapt to the usage of exoskeletons [31], [32].

## V. Conclusion

This study investigated the human perception towards exoskeleton actuation timing during walking and characterized the exoskeleton ankle angle changes as the torque actuation timing varied. We estimated the timing perception by presenting a series of paired actuation timings while participants walked on the treadmill with the exoskeleton. The mean timing JND across participants was 2.8±0.6% stride period. The actuation timings resulted in different ankle angle changes, which could be influenced by the difference in ankle stiffness and peak torque. The results from this study offer insight into how people perceive the changes in exoskeleton behavior. By understanding how people interact with and perceive wearable devices, a more precise and adaptive controller can be developed to maximize exoskeleton efficacy.

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