Human Perception of Wrist Flexion and Extension Torque During Upper and Lower Extremity Movement

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Abstract—Real-world application of haptic feedback from kinesthetic devices is implemented while the user is in motion, but human wrist torque magnitude discrimination has previously only been characterized while users are stationary. In this study, we measured wrist torque discrimination in conditions relevant to activities of daily living, using a previously developed backdrivable wrist exoskeleton capable of applying wrist flexion and extension torque. We implemented a torque comparison test using a two-alternative forced-choice paradigm while participants were both seated and walking on a treadmill, with both a stationary and a moving wrist. Like most kinesthetic haptic devices, the wrist exoskeleton output torque is commanded in an open-loop manner. Thus, the study design was informed by Monte Carlo simulations to verify that the errors in the wrist exoskeleton output torque would not significantly affect the results. Results from ten participants show that although both walking and moving wrist conditions result in higher Weber Fractions (worse perception), participants were able to detect relatively small changes in torque of 12-19% on average in all grouped conditions. The results provide insight regarding the torque magnitudes necessary to make wrist-worn kinesthetic haptic devices noticeable and meaningful to the user in various conditions relevant to activities of daily living.

Index Terms—Kinesthetic devices, perception and psychophysics, biomechanics.

I. INTRODUCTION

TOUCH plays an important role in providing us with information needed to make physical actions, both subconsciously (e.g., withdrawal reflex after touching a hot stove) and consciously (e.g., squeezing a fruit to determine its ripeness). Mechanoreceptors in the skin allow humans to sense cutaneous interactions, such as vibration, force, and changes in temperature. In addition, position and force sensors within the muscles allow sensing of kinesthetic interactions that apply forces or torques to the joints. This information is integrated to inform motor responses.

Kinesthetic haptic devices can provide complex information to the user by displaying forces and/or torques of various magnitudes, and may also measure the user’s joint position for use as a control input. Effective design of these haptic devices requires knowledge of human proprioception and perception of applied forces and torques. Psychophysical and joint position matching tests have been used to quantify proprioceptive acuity in order to determine how well users are able to identify the position of their limbs in the absence of vision [1], [2]. Prior work has also quantified human proprioception of the upper extremity under various movement conditions such as active motions [3], and external forces at the elbow [4], wrist [5], and hand [6]. In addition, experiments have quantified perception of applied vibrotactile and force/torque feedback at the fingertips [7], [8], hand [9], [10], and elbow [3], [11].

Although most experiments investigate force/torque discrimination while the user is seated with minimal movement, wearable upper-limb exoskeletons have been developed with the intent of applying haptic feedback while upper [12] and/or lower extremity [13], [14] motion is occurring. Applications for these devices range from path-following guidance [13] to haptic feedback during teleoperation of robotic prosthetic devices [14]. To inform the design and control of devices intended to be used in conjunction with lower extremity movement, perception of applied forces and torques should be further characterized during walking. In addition, for systems where wrist movement is necessary for either teleoperation [14] or another movement objective [12], it is important to characterize perception of applied forces and torques during upper extremity movement as well.

Existing studies characterizing haptic perception of the upper extremity during movement are limited. Feyzabadi, et al. investigated force discrimination for individual upper extremity joints during movement, without comparison to stationary conditions [15]. Shull et al. investigated human classification of different haptic cues while stationary, walking, or jogging, and found that the classification accuracy decreased with increased movement [16]. The purpose of our work is to quantify torque discrimination at the wrist during both upper and lower extremity movement.

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The organization of this manuscript is as follows. Section II discusses metrics used to characterize haptic perception in prior work, focused on the wrist. In Section III, we describe benchtop testing to characterize the wrist torque output of a previously developed wearable wrist exoskeleton. We use these characterization results in Section IV to perform Monte Carlo simulations that inform the design of the subsequent user study. Section V presents the methods and results of a user study that investigates wrist torque discrimination during different types of motion. Finally, Section VI discusses the implications of our results and future work to be done in this area.

II. PRIOR WORK

Haptic perception can be characterized in multiple ways. One common metric used when conducting psychophysical testing is the Just Noticeable Difference (JND), which is the minimal difference in magnitude between two stimuli that allows the user to reliably perceive a difference. Because JND differs as a function of the magnitude of stimulus presented, it is typically normalized by the reference stimulus to calculate a Weber Fraction. Prior work shows the Weber Fraction to be relatively constant for haptic sensations [15], [17], [18], with some exceptions [19], [20]. In our study, the Weber Fraction is used to characterize perception of applied wrist torque.

Prior studies of human wrist torque discrimination are particularly relevant to the work discussed in this manuscript. Both Vicentini et al. and Laghi et al. investigated torque magnitude discrimination for three degrees of freedom of the wrist (flexion/extension, adduction/abduction, and pronation/supination) [10], [21]. During these experiments, participants were seated and instructed to minimize hand motion while torques were applied by a wrist-worn device [10], [21]. Vicentini et al. uses a preliminary three-alternative forced-choice method of constant stimuli test followed by a two-alternative forced-choice adaptive staircase procedure with three magnitudes of reference forces and torques. Their results show that the Weber Fractions associated with wrist force and torque discrimination in the range of magnitudes tested were not constant, with smaller reference torques resulting in higher Weber Fractions. Specifically for wrist torque flexion and extension torque discrimination, their results indicated that Weber Fractions varied between 0.1 and 0.4 when the applied torque was between 0.03 and 0.3 N-m [10]. Laghi et al. uses a method of constant stimuli with a reference stimulus of approximately 0.2 N-m, and found an average Weber Fraction of 0.1 [21].

Feyzabadi et al. investigated sensitivity of forces applied at the hand during upper extremity movement while seated [15]. A two-alternative forced-choice adaptive staircase procedure was used while users were asked to perform different movements during force application from a world-grounded manipulandum held in the hand. Although they did not perform a direct comparison to stationary conditions, these results can be compared to those of Vicentini et al. to obtain an estimate of how upper extremity movement affects perception of forces applied to the hand. Interestingly, in contrast to Vicentini et al. [10], Feyzabadi et al. [15] found relatively constant Weber Fractions of approximately 0.1 at the hand during wrist movement, even when testing similarly small reference magnitudes of 0.5 N. Although they did not report their data in terms of an equivalent torque perception, we use averaged anthropometric data of the length of the third metacarpal bone [22] to calculate that their reference force of 0.5 N would produce a reference torque of 0.038 N-m, a similar order of magnitude to that tested by Vicentini et al. [10].

In this study, we aim to determine the Weber Fractions of wrist flexion and extension torque magnitude perception during both upper and lower extremity movement, compared to stationary conditions. To our knowledge, this is the first characterization of how lower extremity movement affects upper extremity force/torque discrimination, as well as the first study to directly compare force/torque discrimination during stationary and movement conditions with the same methodology.

Because secondary tasks have previously been shown to negatively affect haptic perception [23], we hypothesize that the additional task and movement associated with walking will result in worse perception at the wrist. Although completing a desired wrist motion is also a secondary task, prior work shows that active compared to passive movement can improve perception in upper extremity tasks [24], [25]. In addition, Feyzabadi et al. [15] reports better overall force perception during wrist movement than that reported by Vicentini et al. [10], who asked participants to maintain a stationary wrist while testing. Therefore, we hypothesize that wrist motion will result in better perception compared to the stationary wrist condition.

III. WRIST EXOSKELETON BENCHTOP TESTING

We used a previously developed wrist exoskeleton [14] (Fig. 1) weighing 363 grams to apply torques to the wrist in this study. Benchtop characterization of the torque output was performed in order to develop an effective experimental protocol to ensure the results were not affected by potential inaccuracy of torque output. We observe that most psychophysical studies performed with commercial kinesthetic desktop devices do not perform such characterization, relying on the device to provide accurate open-loop force or torque output. Because we use a novel exoskeleton device that operates using open-loop torque output, we propose that such characterization is necessary to ensure accurate psychophysical results. Torque output could be affected by the presence and direction of wrist motion while torques are being applied. Benchtop tests that allowed motion of the palm plate of the wrist exoskeleton were previously performed [14], and here we develop additional tests that constrain both the forearm base and palm plate of the wrist exoskeleton during torque application. We include the details of both these tests in this manuscript for clarity.

Although the wrist exoskeleton is able to output torque magnitudes of up to 1 N-m, prior work only characterizes human perception at a maximum reference torque of 0.3 N-m [10]. Our preliminary testing investigating comfort and
user fatigue determined that a torque output of no more than 0.55 Nm was desirable for the repetitive step inputs necessary for perceptual testing. Thus, we focused our benchtop characterization and results between -0.55 Nm (wrist flexion torque) and +0.55 Nm (wrist extension torque), and the same range was used for the subsequent user study.

A. Hanging Mass Resistance Test

In the first test, previously reported in [14], the wrist exoskeleton was placed in a static testing rig and a range of known masses were hung from the palm plate while commanding a virtual spring, such that the further the palm plate traveled from its neutral position, the more resistance torque was applied. In this testing rig, the wrist exoskeleton applied torque to resist downward motion of the palm plate.

Both the wrist flexion and extension torque were tested by fixing the wrist exoskeleton right side up and upside down. Each of the five known masses tested were hung from the center, as well as both extremes of the palm plate, and tested in triplicate in each condition. In each trial, the mass was allowed to reach its steady state position, and the resulting commanded motor torque was averaged. Error was determined by comparing the averaged commanded torque to the known torque resulting from the mass hanging on the motor shaft with a known lever arm, compensating for the change in length as a result of angular displacement.

B. Integrated Force/Torque Sensor Test

Because static friction contributed to the resistance to motion of the wrist exoskeleton during the hanging mass testing described in Section III-A, we also directly measured the torque output. For the second benchtop test, we built a separate wrist exoskeleton identical to the original, except that it housed a 6-axis Nano17 force/torque sensor (ATI Industrial Automation, North Carolina, USA) in the joint opposite from the capstan drive. The forearm frame of the wrist exoskeleton was grounded, and input torques commanded as a step response were directly compared to the torque output measured with the force/torque sensor in the relevant direction. In this test, the maximum magnitude of the input torque was set to 0.4 Nm in each direction because of torque limits on the torque sensor. To account for hysteresis, the steady-state torque inputs were commanded to sequentially increase in increments of 0.1 Nm to the maximum, then decrease to zero in each direction. This testing was done in triplicate, starting both in flexion and extension, and the results were averaged.

C. Results

The results of both benchtop tests were grouped together for all analyses. Fig. 2(a) depicts the data collected during each benchtop test and the resulting relationship between input and output torque. This relationship was linear, with an $R^2$ value of 0.992, and root mean square error (RMSE) of 0.0258 Nm. The distribution of the resulting error is plotted in Fig. 2(b), with a mean of $5 \times 10^{-4}$ Nm and a standard deviation of
For subsequent simulations of the wrist exoskeleton error, the deviation of the mean from zero in this error distribution is treated as negligible.

IV. SIMULATION-INFORMED USER STUDY DESIGN

We sought to design a study that would allow us to use the wrist exoskeleton to accurately determine the Just Noticeable Difference and Weber Fraction of wrist torque magnitude in different movement conditions. Two common experimental methods used to determine Weber Fractions are the staircase method and the method of constant stimuli. In the staircase method, comparison of stimuli with large differences are initially presented, and these differences are adjusted for a predetermined number of trials based on the participant’s ability to correctly identify the larger stimulus [26]. In contrast, the method of constant stimuli compares the same magnitudes of reference and comparison stimuli many times in a random order to fit a psychometric curve [27]. Because of the random order of presentation in the method of constant stimuli, participants are less biased in their responses, but this method is typically more time-consuming than the staircase method. In this study we applied the method of constant stimuli, and used simulations based on pilot data to select a number of repetitions that would achieve rigorous results but not fatigue the participants.

Prior work with participants seated with a stationary wrist found that wrist torque perception varied with reference wrist torque magnitude within a range of reference torques from 0.03 N·m to 0.33 N·m [10]. To examine this relationship in our data, we chose three reference torque magnitudes of 0.1, 0.2, and 0.3 N·m to test in both wrist flexion and extension. Pilot testing with one individual determined that comparison torques equivalent to 30, 60, 90, 120, 150, and 180% of the reference captured the entire range of the psychometric curve.

No device can provide perfect torque output, and we wanted to ensure that our study design included a sufficient number of participants and repetitions to mitigate the effect of torque application error on the resulting perceptual results. To justify our selection of both number of participants and number of repetitions given our characterization of exoskeleton torque output accuracy described in Section III, we performed Monte Carlo simulations of the two-alternative forced-choice method of constant stimuli experiment both with and without consideration of wrist exoskeleton error.

Psychometric curves used as ground truth in simulation were determined using data collected during pilot testing. These psychometric curves were determined for one participant, who was presented with six comparison torques and ten repetitions each for small, medium, and large reference torques while seated with a stationary wrist. These psychometric curves assign a probability that the participant is able to correctly identify the larger torque given the change in magnitude between the reference and comparison. The experiment was performed in simulation taking only human error into account by repeating this comparison for a given number of repetitions, with the probability of the participant choosing the correct answer for a given repetition based on the ground truth psychometric curve data at the given change in magnitude between the two presented torque outputs. An additional experiment was simulated where noise was injected into both the reference and comparison torque values, based on a Gaussian distribution with a standard deviation equivalent to that characterized in the wrist exoskeleton static benchtop testing (Fig. 2(b)). This noise could effectively alter the change in magnitude between the presented torque outputs, affecting the probability of the participant choosing the correct answer.

Each experimental simulation (human error only and with the addition of wrist torque output error) was repeated 1000 times to determine the distribution of the results, in
which Weber Fractions from all simulated reference torques were grouped together. To enable simulation of multiple participants, a given number of simulations were randomly chosen from the distribution and averaged together to obtain resulting Weber Fractions. This was repeated 100 times in order to determine the distribution of the averaged results.

Based on the simulated individual participant outcomes, we found that repeating all possible comparisons 10 times each resulted in a high variance in the estimated Weber Fraction due to anticipated human perceptual error, with small effects of exoskeleton torque application error. For the 1000 simulations performed, the wrist exoskeleton error increased the average Weber Fraction by 14%, from 0.208 to 0.237. We also compared the outcome of each simulation condition with the averaged Weber Fractions from the psychometric curves. The simulations with the human error resulted in a difference of -4.4% from the averaged ground truth Weber Fraction (0.227), and the simulations with both human and wrist exoskeleton error resulted in a difference of +8.4%. The standard deviation of the Weber Fraction for each reference in the simulations was 0.098 without the wrist exoskeleton error and 0.114 with the wrist exoskeleton error. Fig. 3(a) demonstrates the range of simulated Weber Fractions for one reference torque (0.1 N·m of wrist extension torque), both with and without the wrist exoskeleton error.

Overall, with ten simulated participants experiencing all possible comparisons ten times each, the results were highly consistent; the standard deviation of estimated Weber Fraction with wrist exoskeleton error was less than 6% of the ground truth value. The Weber Fractions averaged across all references and participants resulted in a standard deviation of 0.011 without wrist exoskeleton error and 0.013 with wrist exoskeleton error. These standard deviations are much lower than those for the individual participants because they incorporate more data, which reduces the effects of noise. We also examined the change in Weber Fraction in the simulated group results and found an average increase in Weber Fraction of 13% when wrist exoskeleton error was added, from 0.2071 to 0.2351. Because the ground truth Weber Fraction was 0.2273, this resulted in a difference of -8.9% for the simulations with human error only and a difference of +3.4% for the simulations with both human and wrist exoskeleton error. Fig. 3(b) demonstrates the range of resulting simulated Weber Fractions when they were averaged across participants and references.

Based on these simulations, we conclude that the torque output error in the wrist exoskeleton would likely affect our results by artificially increasing the Weber Fractions by an average of 8.4%, with a fairly large standard deviation of 50% of the ground truth value. However, with ten participants repeating each reference and comparison pair ten times, we can still achieve a reliable estimate of wrist torque perception, with a mean error of only 3.4% and standard deviation less than 6% of the ground truth value. Our results also suggest that a future study characterizing the relationships between actual human perceptual errors, imprecision in experimental tools, study design, and calculated perceptual thresholds would result in a valuable tool for experiment design.

V. USER STUDY

A. Participants

Ten healthy participants (5 female and 5 male, age 26.9 ± 2.73 years, arm length 43.6 ± 3.88 cm) were recruited to participate in an experiment using the wrist exoskeleton to determine wrist flexion and extension torque magnitude perception in different movement conditions. Prior to beginning the study, the wrist exoskeleton was fit to each participant by determining an appropriate width of foam spacers placed between the participant’s forearm and the wrist exoskeleton, and the Velcro straps were tightened so that the fit was snug...
but not uncomfortable. This tight fit ensured that the wrist exoskeleton motion relative to the arm was minimal and that the applied torque was translated to forces on the arm instead of motion of the wrist exoskeleton relative to the arm. In addition, because the exoskeleton was driven by a backdrivable motor and capstan transmission, applied torque was accompanied by minimal vibratory or auditory stimuli that could have enhanced user perception. The study was approved by the Stanford University Institutional Review Board, and all participants provided written informed consent prior to testing.

B. Experimental Protocol

Participants performed the two-alternative forced-choice method of constant stimuli experiment with the study design determined in Section IV, where they were presented with a series of two different torque stimuli applied with the wrist exoskeleton and asked to verbally choose which was greater (stimulus one or stimulus two). The experimenter (who was seated out of view of the participant) recorded each response. Comparison testing performed for each reference torque block and the order of each reference and comparison pair were randomized.

A Speedgoat Target Machine interfacing with Simulink Real-Time and running at 1000 Hz was used to control the exoskeleton input torque. In each presentation, the stimuli were commanded for 1.5 seconds each with a 0.5 s pause in between, and an auditory cue with a dominant frequency of 520 Hz accompanied the start of each stimulus. Benchtop characterization of the exoskeleton determined that the rise time to reach 90% of steady-state torque was 13.4 ± 2.4 ms, while the settling time to reach less than 10% variation in the steady-state torque was 32.9 ± 9.6 ms. This was tested at multiple input torque commands up to 0.3 N-m in both flexion and extension.

Each participant completed the two-alternative forced-choice testing in four different conditions (Fig. 4(b)): (1) seated and asked to maintain a stationary neutral wrist angle, (2) seated and asked to oscillate their wrist at a frequency of 1 Hz, (3) walking at 1.25 m/s and asked to maintain a stationary neutral wrist angle, and (4) walking at 1.25 m/s and asked to oscillate their wrist at a frequency of 1 Hz. For the conditions where participants were asked to oscillate their wrist, they were able to hear a metronome at the target frequency alternating between 5.1 kHz and 2.5 kHz sound waves and see a visual representation of the desired frequency on a screen in
During seated conditions, they were asked to keep their arm by their side with their elbow fully extended so that they could not see the device during testing. In the walking trials, participants walked on a treadmill (Bertec, Ohio, US) at a speed of 1.25 m/s that was started and stopped by the experimenter. They were allowed to swing their arms as they normally would during walking.

At the start of each condition, participants practiced the task required for that condition for one minute. Then participants completed at least two practice torque comparison tests (more if desired) until they felt comfortable beginning the trial. The experimenter was able to visualize the wrist angle measurement from the wrist exoskeleton encoder as a function of time throughout the experiment to ensure that the participants were able to complete the task during each condition. The experimenter also monitored the participants during the experiment to ensure that their gaze was not consistently directed towards their wrist, as changes in wrist position could be correlated with torque magnitude. Wrist angle was collected at a sampling frequency of 500 Hz for later analysis. An example of the wrist angle and output torque as a function of time for both a stationary and oscillating wrist condition is shown in Fig. 5(a).

Because of the mental stamina required for the total of 1,440 comparisons per participant, testing was performed over two separate days, with two movement conditions performed per day. In addition, two-minute breaks were implemented between blocks, and participants took a ten-minute break between the two conditions during each day. The order of each block was semi-randomized to prevent fatigue, so that participants did not complete two walking conditions in the same day.

C. Data Analysis

1) Wrist Movement: Wrist angle was analyzed in order to verify that the participants were moving their wrist at the instructed frequency and to investigate potential correlations between wrist motion characteristics and Weber Fractions. To capture the transient wrist dynamics versus the low-frequency movement over time, wrist angle was analyzed in segments of five seconds each. The wrist motion in each segment was fit to a one-term Fourier model in MATLAB, defined as follows:

\[ f_x = a_0 + a_1 \cos(\omega t) + b_1 \sin(\omega t) \]  

This equation was used to determine the frequency (\(\omega\)), offset (\(a_0\)), and amplitude, \(\sqrt{a_1^2 + b_1^2}\). The majority of data were well-described by this Fourier fit, as the moving wrist trials were designed to be oscillatory, and the stationary wrist trials often oscillated a small amount as well due to wrist motion as a result of the output torques (example raw wrist motion data shown in Fig. 5(a)). However, there were segments of data in the stationary wrist trials where the optimal fitting coefficients had very low frequency and high amplitude, essentially resulting in a linear fit. For all segments where \(\omega\) was less then 0.05 Hz, amplitude was instead determined as half of the difference between the maximum and minimum angles for that segment, and the offset was determined by taking the mean value for that segment of data.

To determine the ultimate result for each trial, the median values of the frequency, amplitude, and offset from each 5-second segment within the trial were determined. The mean and standard deviation of the dominant frequency of oscillation was determined for all grouped moving wrist and stationary wrist conditions, and trials where this dominant frequency fell outside of three standard deviations from the mean were discarded from subsequent movement analyses. In total, this outlier detection resulted in 5 trials discarded of the 240 total collected in the study.
2) Weber Fraction: For each reference torque in each condition, psychometric curves were fit to a cumulative Gaussian sigmoid function with equal asymptotes using the psignifit toolbox [28]. JND was defined as the torque magnitude difference at which the participant could identify the larger stimulus with 75% accuracy. This occurred at two points of the psychometric curve (for torque magnitude differences both lower and higher than the reference), so these two points were averaged to calculate the final JND for each reference. Fig. 6 shows an example of JND curves collected in all movement conditions for one participant at a single reference torque. We also report two additional values determined from the psychometric curve: (1) the point of subjective equality, or the torque magnitude difference at which the participant had an equal likelihood of reporting that the stimulus was higher or lower, and (2) the lapse rate, or difference in the upper and lower asymptote from 100% accuracy.

After using the psychometric curves to determine JND values, these were normalized by the reference torque to calculate Weber Fractions. A three-way analysis of variance (ANOVA) followed by multiple comparison testing was performed to determine if the Weber Fractions were significantly affected by the reference magnitude, walking, or wrist motion. To obtain an overall estimate of the effect of upper and lower extremity movement on the resulting Weber Fractions, all reference torques were grouped together and percent differences were calculated for each condition, using a paired t-test for each group with Bonferroni corrections to account for multiple comparisons. We also performed post-hoc analyses to determine if there was a relationship between any quantitative wrist motion metrics (dominant frequency, amplitude, or offset) and the resulting Weber Fraction by checking for a significant correlation.

D. Experimental Results

1) Wrist Motion: The Fourier fit determined that participants were indeed able to follow the instructed 1 Hz wrist oscillation, with an overall dominant frequency of 1.00 ± 0.057 Hz (median ± standard deviation). In addition, the dominant amplitude for the wrist movement trials (16.86 ± 7.24°) was significantly greater than that of the stationary wrist trials (3.00 ± 5.39°), with a p-value of $6.3 \times 10^{-32}$. The small average amplitude during the stationary wrist trials demonstrates that participants followed instructions for this condition as well. Grouped results are shown in Fig. 5(b).

2) Weber Fraction: The Weber Fractions were significantly affected by walking ($F_{(5,101)} = 15.8$, $p = 0.001$) and wrist motion ($F_{(5,101)} = 31.2$, $p < 0.001$), which both resulted in higher Weber Fractions compared to the stationary conditions. The interaction terms were not significant. The reference torque also had a significant effect on the Weber Fractions ($F_{(5,101)} = 4.1$, $p = 0.001$), and multiple comparison testing showed that only conditions testing reference torques with the smallest magnitude of 0.1 N·m while walking with an oscillating wrist resulted in Weber Fractions significantly higher than a number of other conditions. Fig. 7(a) shows the mean and standard deviation of the Weber Fraction in each condition, as well as individual participant datapoints.

In comparing Weber Fractions independent of reference torque, all grouped conditions resulted in average Weber Fractions between 12 and 19% (Fig. 7(b)). In statistical testing, the significance level of 0.05 was adjusted to 0.0083 to account for multiple comparisons. Compared to the baseline seated and stationary wrist condition, wrist motion while seated reduced perception by 28.9% ($p = 5.0 \times 10^{-6}$), walking with a stationary wrist reduced perception by 20.1% ($p = 0.001$), and oscillation while seated reduced perception by 28.9% ($p = 5.0 \times 10^{-6}$), walking with an oscillating wrist reduced perception by 20.1% ($p = 0.001$), indicating a significant effect of wrist motion on perception.
walking with wrist motion reduced perception by 53.1% (p = 1.2 × 10⁻⁸) (Fig. 7(b)). In addition, walking with wrist motion reduced perception compared to walking with a stationary wrist by 26.8% (p = 0.002) (Fig. 7(b)). Despite the number of significant differences in Weber Fractions when comparing different grouped conditions, none of the quantitative wrist motion metrics (frequency, offset, or amplitude) were significantly correlated with Weber Fraction when treated continuously.

The median point of subjective equality for all fits was 0.0026 N·m with a standard deviation of 0.015. The point of subjective equality gives us additional confidence in our fits, as we would expect it to be zero if there was no bias towards wrist flexion or wrist extension, and it is less than 3% of the smallest reference torque value. The median lapse rate was 3.7 × 10⁻⁷ with a standard deviation of 0.0164, indicating that for the vast majority of the trials, participants were able to identify the comparison torques furthest from the reference with 100% accuracy.

VI. DISCUSSION

In this manuscript, we sought to quantify perception of applied wrist torque during upper and lower extremity motion compared to stationary. In order to do this, we utilized a previously built wearable wrist exoskeleton capable of accurately sensing wrist angle and providing up to 1 N·m of continuous wrist flexion and extension torque. Using two different methods of benchtop testing, we determined that the wrist exoskeleton was able to transmit torques with an RMS error of 0.0258 N·m. We used this characterization to design an experimental protocol to accurately determine Weber Fractions for wrist torque perception. We performed a two-alternative forced-choice experiment where we tested small, medium, and large reference torques in flexion and extension while each participant was both sitting and walking with a stationary and oscillating wrist to determine Weber Fractions. We found that both walking and wrist motion reduced participants’ abilities to perceive differences in wrist torque, and that perception in all conditions was within 10-20% of the reference stimulus.

The results supported our hypothesis that wrist torque magnitude would decrease while walking versus when seated. This decrease in sensitivity is an important consideration in the design of future wrist-worn haptic devices seeking to provide torque feedback while the user is moving their wrist or walking, especially given that previous work focuses primarily on characterizing torque sensitivity while seated [10], [15], [21]. Although we only investigated torque discrimination at one upper extremity joint in this study, it is likely that walking would also affect perception at other joints or areas of the body. Another study investigating the effect of walking on vibrotactile haptic cue classification found a 14% decrease in perceptual acuity [16], which is similar to the 21% decrease found in our experiment, suggesting that these results may generalize across multiple haptic modalities and stimulation locations. However, prior studies investigating the perception
of time delay [29] and direction detection [30] found the results to be affected by frequency and amplitude, suggesting that our results may differ with changing wrist oscillation frequency or amplitude as well. Additional studies investigating the effect of walking on different types of perception and movement input would provide further insight into how generalized these findings are for different haptic devices.

Our hypothesis that wrist motion would increase perceptual abilities compared to a stationary wrist was not supported by the data. Prior work suggests that active movement can increase proprioceptive abilities [24], [25], and it is possible that this would apply to torque discrimination as well. In addition, prior work characterizing perception of applied torque during wrist movement resulted in lower Weber Fractions [15] compared to experiments where the wrist was stationary [10], although these conditions were not directly compared with the same experimental setup. However, our results indicate that perception of applied torque declines with wrist movement.

We hypothesize that the task of oscillating the wrist at a certain frequency required mental effort that likely distracted the participants, as many participants qualitatively reported that they thought the task of oscillating their wrist at a given frequency was challenging. In addition, it is interesting to note that although Weber Fractions significantly increased in the movement condition compared to stationary, there was no significant correlation between the dominant wrist frequency and Weber Fraction, suggesting that it is possible that the effect on Weber Fraction may have more to do with the mental effort of performing the task than the motion itself. Because we only provided one minute of practice for a given task before starting the perceptual testing, it is possible that increased training time could decrease the mental effort required for this task, which may result in better perceptive capabilities.

Prior literature regarding the effect of reference magnitude on perception has been mixed, with one study reporting an almost four-fold increase in Weber Fraction with decreasing reference forces and torques at the wrist [10], and another reporting consistent Weber Fractions across similarly varying reference forces [15]. Our results indicate that reference torque did have a significant effect on perception, although the only two conditions that were significantly different occurred at the smallest reference torque magnitudes during walking with an oscillating wrist. Overall, although we did see a trend towards worse perception at smaller reference torques, the differences that we found were smaller than the changes found by Vicentini et al. [10].

There are several limitations of this study. Using a wearable wrist exoskeleton allowed users to move both their upper and lower extremity in more natural ways compared to a world-grounded device, allowing us to answer our research questions regarding the effect of this movement on haptic perception. Because users were allowed to swing their arms naturally while walking, this subtly changed the dynamics of the system and may have introduced additional variability, allowing us to measure haptic perception during motions representative of activities of daily living. In addition, although participants were not permitted to consistently gaze at the wrist during the experiment, there was no visual barrier preventing it. Therefore, it is possible that this could have a small effect on our results, since wrist position could be correlated with applied torque, even though participants were instructed to maintain a stationary or consistently oscillating wrist regardless of applied torque. There was also some error in the torque output of the device, likely affecting our results. Even so, our simulations suggest that the relatively small magnitudes of torque error in our device (RMSE <0.03 N-m), combined with the number of repetitions and participants tested, resulted in Weber Fractions that were artificially increased by less than 5% compared to ground truth. Because many studies report Weber Fractions without characterizing device error, our simulations suggest that these types of experimental design criteria are important to consider.

It is possible that the design of the wrist exoskeleton also caused some bias in perception of wrist flexion and extension torque. The highest Weber Fractions (worst perception) occurred at the smallest reference torque magnitudes while walking and oscillating the wrist. Of these two conditions, wrist flexion resulted in the largest Weber Fraction. However, the wrist exoskeleton was not uniform on the palmar versus dorsal side. The hand was secured on the palmar side with a palm plate consisting of rigid 3D-printed PLA, and a compliant Velcro strap on the dorsal side. For this reason, it may have been easier for participants to perceive wrist extension torque, because the resultant forces pushing against the hand were applied by a rigid versus compliant body, allowing participants to more effectively use tactile, as well as kinesthetic cues [31]. In addition, the palmar side of the hand consists of glabrous skin, which is known to have better perceptual capabilities than hairy skin [32], due to an increased density of myelinated afferent touch receptors [33]. Because force would be applied on the dorsal side of the hand with hairy skin during wrist flexion, this could contribute to differences in perception.

There are other aspects of wrist movement and perception that our study does not address. For example, performance in force matching tasks has been shown to deteriorate when bidirectional compared to unidirectional force is applied [34]. Because each direction (flexion and extension) was tested separately in our protocol, it is possible that the torque discrimination thresholds would be different if the two directions were tested together. In addition, although prior work has noted differences due to handedness in force matching tasks [35] and other proprioceptive tasks [36], we did not systematically explore the effects of handedness on the torque perception. Although one participant was left-handed, their results fell within the range of the others tested. Finally, our methods did not allow for repetition of comparisons within a single trial, which could increase user confidence and decrease JND values.

Like the complex mechanisms of human sensory feedback, the biological processes involved in human movement are multifaceted. Movement is the result of neural excitation, which results in activation of muscles that can cross multiple joints. Muscle activation then causes force application to move the joint to a new position, with a dependency on the complex musculoskeletal geometry of the human body. In this
study, we attempt to normalize wrist motion by instructing participants either to move their wrist at a given frequency or keep the wrist in a consistent, neutral position. Although participants were largely able to follow these instructions, some were more variable than others, with some smaller participants reporting difficulty resisting the larger torque magnitudes to keep their wrist in the desired position. In addition, the amplitude of wrist oscillation was allowed to vary, as providing participants with feedback of their wrist angle compared to the desired wrist angles to match would have provided participants with an alternate sensory pathway (visual) to detect differences based on the output torque. Future work should attempt to isolate different aspects of wrist movement, such as wrist position and velocity, in order to determine their individual effects on perception.

VII. CONCLUSION

In this manuscript, we quantify human wrist torque perception during different upper and lower extremity movement conditions. We find that both wrist movement and walking significantly decreased perception, and walking while moving the wrist resulted in 53% worse perception compared to the seated and stationary wrist condition. These increased perceptual thresholds should be taken into account when designing kinesthetic haptic devices with applications requiring real-world human movement.

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