A Robotic Device for Measuring Human Ankle Motion Sense

Qiyin Huang, *Student Member, IEEE*, Bin Zhong, *Student Member, IEEE*, Naveen Elangovan, Mingming Zhang, *Senior Member, IEEE*, and Jürgen Konczak, *Senior Member, IEEE*

***Abstract*—Proprioceptive signals about ankle motion are essential for the control of balance and gait. However, objective, accurate methods for testing ankle motion sense in clinical settings are not established. This study presents a fast and accurate method to assess human ankle motion sense acuity. A one degree-of-freedom (DOF) robotic device was used to passively rotate the ankle under controlled conditions and applied a psychophysical forced-choice paradigm. Twenty healthy participants were recruited for study participation. Within a trial, participants experienced one of three *reference* velocities (10°/s, 15°/s, and 20°/s), and a smaller *comparison* velocity. Subsequently, they verbally indicated which of the two movements was faster. As outcome measures, a just-noticeable-difference (JND) threshold and interval of uncertainty (IU) were derived from the psychometric stimulus-response difference function for each participant. Our data show that mean JND threshold increased almost linearly from 0.53°/s at the 10°/s *reference* to 1.6°/s at 20°/s ($p < 0.0001$). Perceptual uncertainty increased similarly (median IU = 0.33°/s at 10°/s and 0.97°/s at 20°/s; $p < 0.0001$). Both measures were strongly correlated ($r_s = 0.70$). This implies that the bias of the human ankle motion sense is approximately 5 - 8% of the experienced movement velocity. We demonstrate that this robot-aided test produces quantitative data on human ankle motion sense acuity. It provides a useful addition to the current measures of ankle proprioceptive function.**

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Qiyin Huang, Naveen Elangovan, and Jürgen Konczak are with the Human Sensorimotor Control Laboratory, Center for Clinical Movement Science, and the School of Kinesiology, University of Minnesota, Minneapolis, MN 55455 USA (e-mail: huan1480@umn.edu; naveen@umn.edu; jkonczak@umn.edu).

Bin Zhong and Mingming Zhang are with the Shenzhen Key Laboratory of Smart Healthcare Engineering, Department of Biomedical Engineering, Southern University of Science and Technology, Shenzhen, Guangdong 518055, China (e-mail: 11955001@mail.sustech.edu.cn; zhangmm@sustech.edu.cn).

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NOMENCLATURE

ANOVA	Analysis of Variance
CoP	Center of pressure
EMG	Electromyography
IU	Interval of uncertainty
IQR	Interquartile range
ICC	Intraclass correlation coefficient
JND	Just-noticeable-difference
PROM	Passive range of motion
PF/DF	Plantarflexion/Dorsiflexion
RMS	Root mean square
SEM	Standard error of measurement
SD	Standard deviation

I. INTRODUCTION

JOINT motion sense refers the ability to detect limb movement or to discriminate between different movement velocities [1]. It is one of the sensory modalities of proprioception, the sense of body/limb awareness. Proprioceptive information about movement velocity is mainly encoded in signals derived from dynamic nuclear bag fibers of muscle spindles, mechanoreceptors embedded in skeletal muscles [2]. Proprioceptive information about ankle position and motion is important for the neural control of gait and balance [3]. Numerous neurological conditions, such as stroke [4] and cerebral palsy (CP) [5], are associated with impaired ankle proprioception, which leads to deficits in postural control during stance and locomotion [6], [7]. The available clinical scales (e.g., modified Nottingham Sensory Assessment) commonly used in clinical exams provide basic data on function, such as an examiner rating a patient's ability to perceive movement and/or its direction on an ordinal scale as "absent", "impaired" to "normal" [8], [9], [10]. That is, the applied test scales are only sensitive to detect large deficits of proprioceptive function. However, there are psychophysical methods available that can provide a more comprehensive evaluation of motion sense while yielding data at a higher resolution [11]. These methods are designed to measure either motion sense *sensitivity* or *acuity*. Detection thresholds representing the smallest detectable movement velocity would represent a measure of *sensitivity*. Other typical measures of motion sense *sensitivity* are the time/displacement until movement detection or the identification of motion direction [12], [13].

Intact ankle proprioception is crucial for balance control [14]. A decline in ankle proprioceptive acuity is known to negatively affect balance and gait. Respective proprioceptive deficits have been documented in aging [15] and populations affected by stroke [6], cerebral palsy [7], and Parkinson's disease [16]. In this context, capturing motion and position sense acuity is clinically meaningful because impairments in these proprioceptive submodalities are closely linked to observable motor dysfunction [17]. Moreover, motion and position sense can be differentially affected within and between patients by the phase of the disease. For example, people with chronic stroke may present with abnormalities in arm position or motion sense or both [18], and they may improve in measures of proprioceptive function as part of therapy [19]. That is, the assessment of both senses provides a more comprehensive profile of proprioceptive status. This also implies that the available tests of motion sense acuity should be sensitive enough to detect subtle differences in proprioceptive function to provide clinical utility [17], such as providing quantitative data representing a continuum from "absent" to "normal" to quantify therapeutic success over time.

An assessment of motion sense *acuity* involves the presentation of two distinct movement velocities, where a joint or limb is rotated passively either by an investigator or a mechatronic device [11]. The perceiver identifies which of the two presented velocities is faster. Subsequently, a psychometric stimulus-response function is derived [20], which provides two outcome measures of motion sense acuity: 1) a discrimination or *just-noticeable-difference* (JND) threshold [21] as a measure of bias or systematic error, and 2) the *interval of uncertainty* (IU) as a measure of precision or random error [22]. The human ability to just notice a difference between two sensory stimuli is proportional to the stimulus intensity - a relationship expressed in the *Weber-Fechner* laws [23], [24], [25], [26]. For motion sense, this implies that a JND threshold should increase as movement velocity increases with the ratio between the JND threshold and a given movement velocity representing *Weber* fraction (K) [25].

Unlike position sense, human motion sense has been understudied [27]. To date, available reports focused mostly on upper limb joints (e.g., finger, elbow, shoulder) [27], [28], [29], [30]. However, the procedures applied in these reports incompletely addressed confounding factors that can affect measures of motion sense acuity. For example, matching tasks that require a user to actively replicate a given joint velocity [28], are unable to differentiate the sensory from the motor contribution. The resulting perceptual judgment is a composite of both proprioceptive and motor function. This distinction becomes important when testing the proprioceptive function of clinical populations who have compromised motor control, as it becomes difficult to discern if the failure to replicate a joint velocity is due to a motor problem. In movement velocity discrimination tasks, participants can utilize movement time [27] or displacement [29], [30] as motion cues to judge the differences between movement velocities. For instance, if two velocities are provided over the same displacement, then the faster velocity is associated with a shorter time period, allowing the observer to simply count. If two velocities are provided at the same time interval, the faster motion will result in a longer displacement, which then allows the observer to judge between positions to infer differences in motion. Thus, it is imperative that the availability of such unwanted motion cues is tightly controlled when assessing motion sense.

Limb motion sense of the lower extremity has rarely been studied systematically, and focused solely on motion sense sensitivity [4], [31]. Possible reasons for the paucity of available data on motion sense acuity concern the necessity of a device that 1) can deliver highly controlled and precise movement velocities, and 2) avoids providing extra displacement or time sensory cues.

This study aims to address the above shortcomings. First, it presents a system to assess ankle motion sense acuity that controls for confounding position and time cues. Second, it shows that the developed robotic assessment system can generate quantitative data on ankle motion sense bias and precision. Finally, the obtained data can be compared to motion sense acuity of other human joints [32], [33]. This helps to examine diseases such as diabetes that tend to affect distal joints first in the disease process and then progress to other joints and limb segments [34].

II. Methods

A. Participants

Twenty healthy adults (mean ± SD age 23.2 ± 3.5 years, F: 11) participated in this study. All participants self-reported with no neurological impairments, musculoskeletal or orthopedic injuries in lower extremities within 12 months prior to testing. All study procedures were reviewed and approved by the University of Minnesota Institutional Review Board. Before testing, all participants provided written informed consent and completed the footedness questionnaire [35] to determine the dominant foot.

B. The Robotic Ankle Proprioception Assessment System

The robotic ankle proprioception assessment system (APASr) used in this study can deliver dorsi/plantarflexion movements (see **Fig. 1A**). Its technical details are described elsewhere [36]. In short, the actuator consists of a DC motor (305013, Maxon, 200 W) with a gearbox (326664, Maxon, gear ratio: 51:1) and a built-in 14-bit optical encoder (575827, Maxon). The RMS velocity error for repeated rotations at a 20°/s target velocity was of 0.091°/s. The device was programmed to generate precise speeds and trajectories with well-tuned internal PID gains to move the ankle passively. The maximum torque provided by the actuator is 12 Nm. To determine the sensitivity of ankle robot, we evaluated the positioning accuracy of the robotic ankle by measuring the error between the actual trajectory and a sinusoidal reference trajectory signal with an amplitude of 20 degrees for a time period of 12 seconds. The repetitive positioning accuracy was 0.04° ± 0.11° (mean ± SD), and the maximum error was

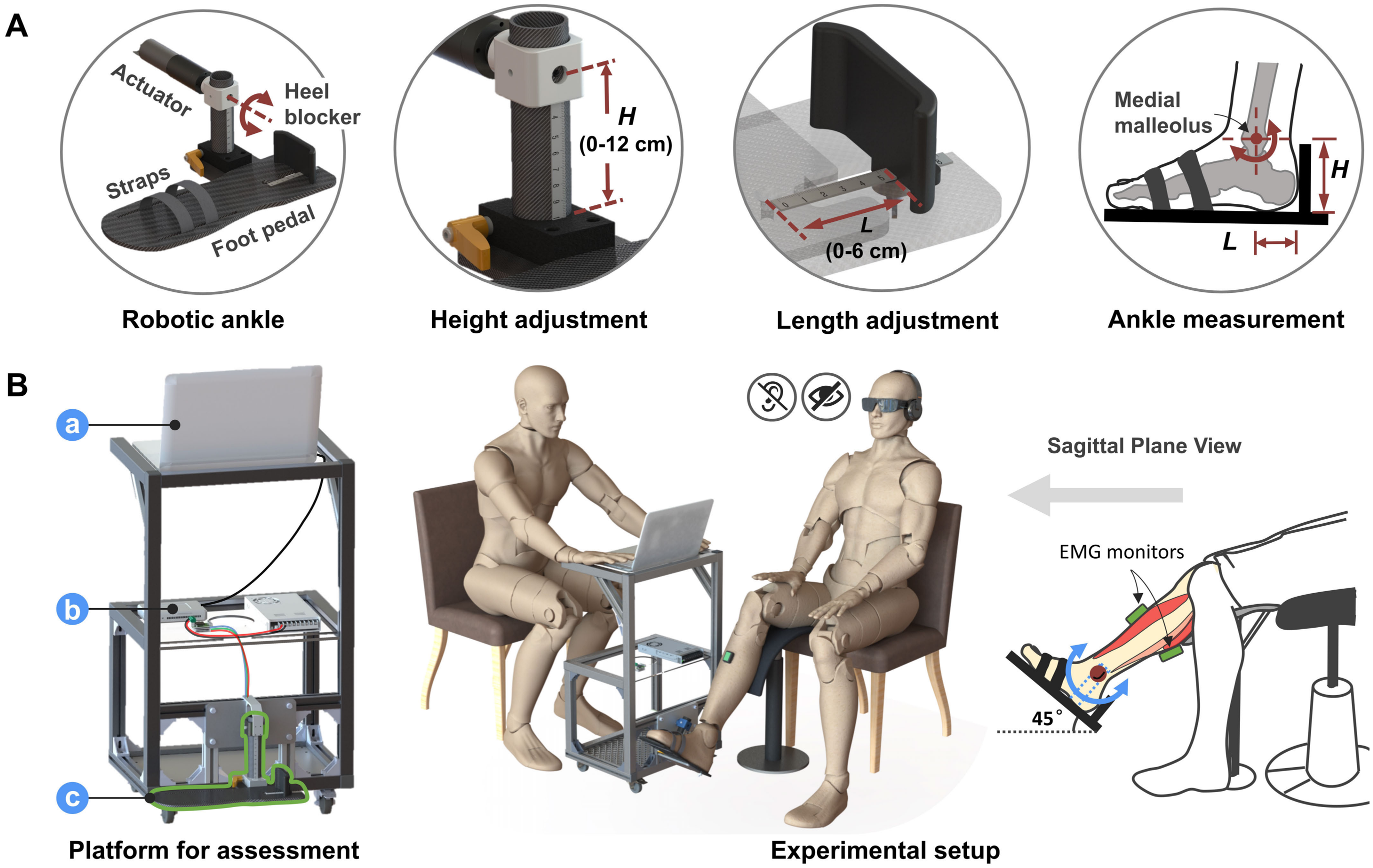

Fig. 1. **A**. Robotic ankle proprioception system. The system includes an actuator (motor, gearbox, and encoder), and the foot pedal with foot straps to secure the forefoot position. The maximum range of ankle height (carbon fiber shaft rod) and foot length (heel blocker) are adjustable to human anthropometrics. The position of the lateral malleolus (the center point of the cross-hairs) was determined relative to the base of the heel (height (H) and length (L)) to align the motor axis with the human ankle joint axis. **B**. Overview of the complete system consisting of a) the computer to control the robot, b) the actuation system, and c) the robotic ankle. The participant was blindfolded to block vision and received auditory background noise via headphones to mask auditory cues. The height of the support pedestal was adjusted to unload the leg and allow the foot to rest on the foot pedal. Surface electromyography (EMG) was placed on two major muscles (i.e., tibialis anterior and gastrocnemius) to monitor muscular activity in real-time and ensure passive movements during the testing. When active muscular activity was detected during a trial, the trial was repeated.

always below 0.2°. Foot straps stabilize the testing foot on a carbon fiber foot pedal (L: 33 cm; W: 13 cm). A carbon fiber shaft rod (maximum H: 12 cm) was attached to the actuator's output shaft to adjust the ankle to align with the motor axis. The use of carbon fiber components provides high durability while achieving a lower mass of the device, which reduces the necessary torque output during ankle rotation. A heel blocker screwed in the foot pedal allows foot position to be adjustable forward/backward within a 6 cm range. Calibrated scales were included in the foot pedal to guarantee accurate foot positions for various foot sizes. The electronic control apparatus was mounted on a mobile platform with lockable K-92-50F type wheels (see Fig. 1B). The test protocol software uses a custom-written MATLAB code to determine the stimulus presentation order and stimulus size difference required for position and motion sense testing. It is integrated with a customized control software routine developed in LabVIEW to control the output of the DC motor.

C. Procedure

During the assessment, participants sat on a chair with their dominant foot resting on the foot pedal (Fig. 1B). The distance and height of the lateral malleolus from the heel were measured to align the axis of rotation of the ankle joint with the center of rotation of the robot's actuator. In addition, ankle *passive range of motion* (PROM) in plantarflexion (PF) and dorsiflexion (DF) were measured by an occupational therapist who used a handheld goniometer. The tested leg was allowed to rest on a custom leg support to maintain the foot in an open kinematic chain with the ankle joint at 90° relative to the shank (see Fig. 1B). The participants were instructed to keep the tested ankle relaxed. Surface EMG was used to monitor the muscular activity of the tibialis anterior and gastrocnemius in real-time to ensure passive movements during the testing. Those trials where active muscular activity was detected were repeated and participants were reminded to relax their foot. Participants wore opaque glasses and headphones playing pink noise to exclude extraneous visual and auditory cues during testing (Fig. 1B).

The testing procedure assessed motion sense acuity at three *reference* velocities (V_R): 10°/s, 15°/s, and 20°/s. The *reference* velocities were chosen considering the PROM of the ankle joint in plantarflexion (49.4 - 62.1°) [37] to ensure that the participants have sufficient time (>2.5s) to experience the

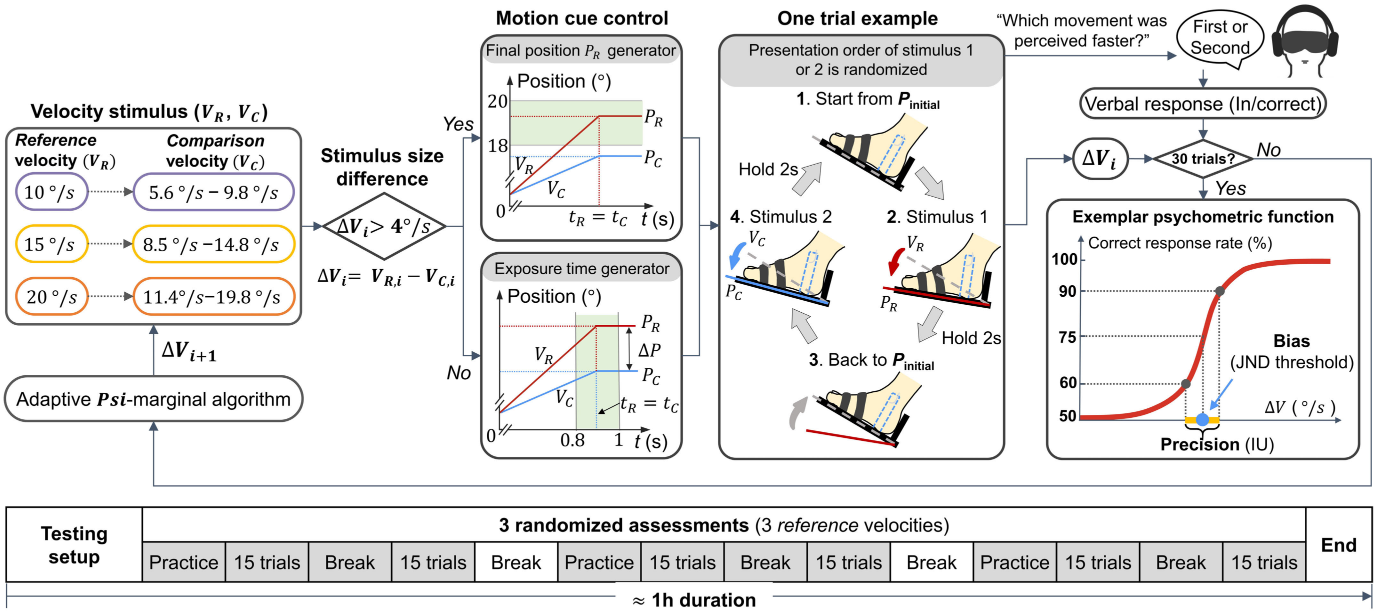

Fig. 2. Test protocol and timeline of the experimental procedure. Based on the stimulus size difference ($\triangle V_i$) between the *reference* (V_R) and *comparison* stimulus (V_C) determined by the adaptive psi-marginal algorithm, a subsequent motion cue control algorithm determined the corresponding final positions (P_R and P_C) for each velocity stimulus. The amplitude of the final position P_R or the exposure time of experienced velocities (V_R, V_C) was then randomized within a range of allowable positions and exposure times (see green zones). After the robot rotated the foot at two different velocities (V_R, V_C), the participant verbally indicated which movement was perceived as faster. After 30 trials, a stimulus-response psychometric function was fitted on the stimulus size difference and correct response rate data.

given velocities. A person who has insufficient time to perceive an imposed velocity, will tend to guess, which ultimately alienates the test results [12], [21]. Additionally, the maximum exposure time (<6.21s) avoids potential memory issues (i.e., the first movement is no longer in working memory) [21].

In each trial (see Fig. 2), the ankle robot plantarflexed the participant's ankle at the *reference* velocity V_R and the *comparison* velocity V_C ($<V_R$) from the initial neutral position ($P_{initial}$). After two seconds, the robot moved the ankle back to the initial position. During testing, the order of the *reference* and *comparison* velocities (first or second) between trials was randomized. To minimize the confound due to time, the time duration (t_R, t_C) for the pair of velocities in each trial was the same but varied between trials. This means the final position P_R for *reference* velocity was always larger than the final position P_C for *comparison* velocity. When the stimulus size difference ($\triangle V$) between the *reference* and *comparison* stimulus was larger than 4°/s, a randomized final position ($P_R \in [18°, 20°]$) was generated by the final position generator (see Fig. 2), which provided larger exposure times (>1.8s for 10°/s *reference*, >1.2s for 15°/s *reference*, and >0.9s for 20°/s *reference*) of the two velocity stimuli. When $\triangle V$ decreased below 4°/s, then the exposure times for V_R and V_C were randomly selected to be between 0.8 - 1.0s ($t_R = t_C$). This caused the difference between the two angular displacements to start below 4° and assured fast convergence towards the known average ankle position sense discrimination threshold of healthy young adults (∼2.4°) [38]. Thus, the differences between displacements were under the position sense thresholds, thereby controlling for the velocity cues based on judging differences in the final positions of P_R and P_C (see Fig. 3). At the end of each trial, the participant verbally indicated which of the two movements was faster (first or second). Based on the participant's responses, the subsequent stimulus size difference between the *reference* and the *comparison* velocity stimuli was selected by an adaptive Bayesian (psi-marginal) algorithm [20].

The assessment of each of the three *reference* velocities consisted of 30 trials (<15 min.) for a total of 90 trials. The order of three *reference* velocities was randomized between participants. Before each assessment, participants performed three practice trials to become familiar with the procedure. Breaks were offered after 15 completed trials or when the participant requested a rest. The experimental duration was around one hour, including setup, practice, and breaks (see Fig. 2).

D. Proprioceptive Measures

After the completion of the testing, a logistic Weibull function was fitted for the stimulus size difference data (i.e., the difference between *reference* and *comparison* ankle velocity stimuli) and the correct response rate data for each participant. Based on the fitted function, the stimulus size difference corresponding to the 75% correct response rate was identified as the *just-noticeable difference* (JND) threshold as a measure of perceptual bias [21]. The corresponding *interval of uncertainty* (IU) represents the range of the stimulus size difference between 60% and 90% probability of correct response. It is a measure of perceptual precision (see Fig. 2).

For this test, a JND motion sense threshold represents the minimum difference in velocity required to discriminate a *comparison* from the *reference* velocity (i.e., 10°/s, 15°/s, and 20°/s).

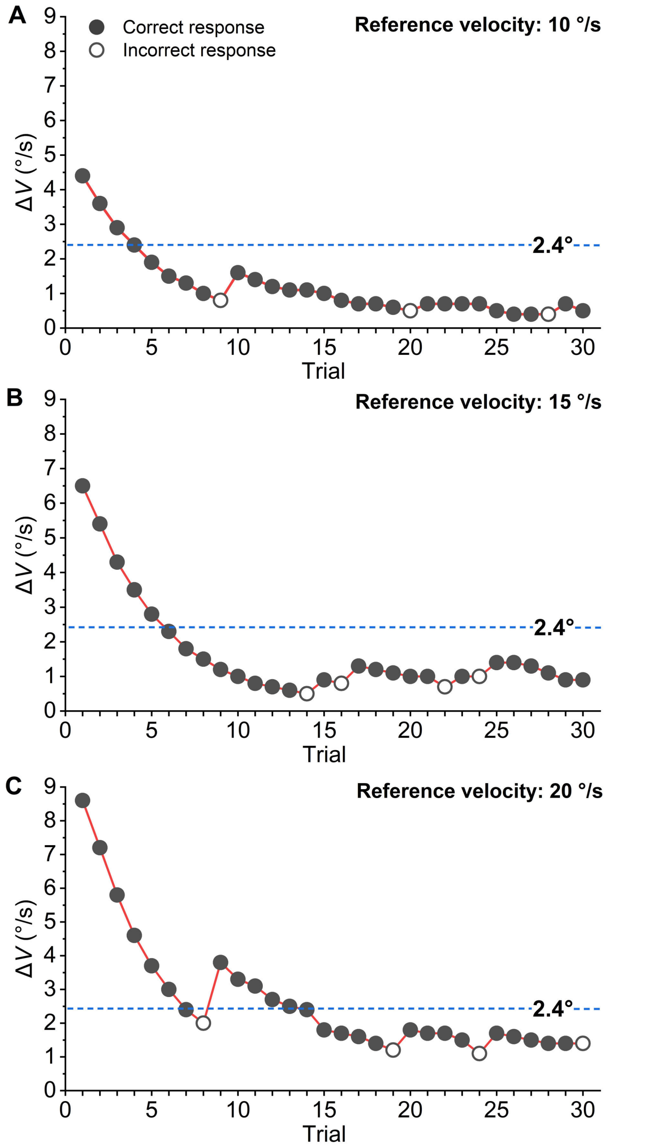

Fig. 3. Exemplar trial data at the three *reference* velocities for one participant (**A**. 10°/s, **B**. 15°/s, **C**. 20°/s). The y-axis shows the experienced velocity difference ($\triangle V$) between *reference* and *comparison* stimulus. Each datapoint represents ($\triangle V$) at a given trial. The blue dotted lines represent the known average value of the ankle position sense discrimination threshold [38]. Below this threshold, differences in final joint position are not perceivable, and thus, could not have been used as cues to judge differences in movement velocity.

III. Statistical Analysis

To obtain sufficient statistical power to detect statistical differences between the three tested *reference* velocities, we performed a priori power analysis, which yielded an estimated total sample size of N = 20. The normality and sphericity of the outcome measures (i.e., JND threshold and IU) were evaluated using Shapiro-Wilk and Mauchly's tests individually. Only JND thresholds were normally distributed and met the sphericity assumption. Subsequently, a one-way repeated measures ANOVA was performed to determine differences between JND thresholds at three *reference* velocities (10°/s, 15°/s, and 20°/s). Effect size was reported using Eta-squared (η^2) where $\eta^2 = 0.01$ corresponds to a small, $\eta^2 = 0.06$ to a medium, and $\eta^2 = 0.14$ to large effect size [39]. Post-hoc analyses were conducted with a Bonferroni adjustment for the paired *reference* velocity comparisons.

A non-parametric analysis was conducted for IU using the Friedman test since it did not meet normality. Effect size was reported with Kendall's W, which was 0.1 (small), 0.3 (moderate), and above 0.5 (strong) effect [39]. The Conover's test was used for the post-hoc analyses accordingly. Spearman's correlation analyses were conducted to determine the relationship between two proprioceptive measures (i.e., bias and precision). The correlation r_s between 0.1 - 0.3 was considered weak, 0.4 - 0.6 moderate, and 0.7 - 0.9 strong [40].

Finally, *Weber* fraction (K) was computed as the ratio between the mean threshold of the sample and a respective *reference* velocity (i.e., JND/V_R) [25]. The software R 4.1.2 was used for the statistical analysis.

TABLE I
Descriptive Statistics for Outcome Measures

Proprioceptive measures	V_R (°/s)	Mean (°/s)	Range (°/s)	K
JND threshold	10	0.53	0.30 – 0.93	0.05
	15	1.03	0.54 – 1.53	0.07
	20	1.60	0.66 – 2.82	0.08
	V_R (°/s)	**Median (°/s)**	**Range (°/s)**	
IU	10	0.33	0.07 – 1.00	
	15	0.66	0.15 – 1.60	
	20	0.97	0.41 – 4.67	

***Note**: V_R denotes the *reference* velocities. JND threshold data were normally distributed and are presented as mean. IU data are presented as median. *Weber* fraction (K) was computed as the ratio between the mean JND threshold and a respective *reference* velocity (i.e., JND/V_R). A higher value of K indicates that a larger difference between the *reference* and *comparison* velocity was necessary to perceive the two experienced velocities as different.

IV. Results

Across all participants, median ankle PROM in plantarflexion was 50° (range: 25° - 60°) and 15° in dorsiflexion (range: 5° - 20°). There was a single participant who presented with a restricted PROM (25°) in PF and two participants with a PROM of 40°. The respective mean JND thresholds and median IU values and *Weber* fractions for the 10°/s, 15°/s, and 20°/s *reference* velocities are shown in **Table I**. The corresponding linear regression equation of the JND means at the three *reference* velocities was JND $= -0.55 + 0.11 \times V_R$ yielding a coefficient of determination $R^2 = 0.99$. The slope of the fitted line was $K = 0.11$, representing the overall *Weber* fraction. In addition, the JND threshold was significantly different at the three different *reference* velocities with a large effect size ($F_{(2,38)} = 56.01$, $p < 0.0001$, $\eta^2 = 0.60$). Post-hoc analyses with a Bonferroni adjustment revealed that the pairwise differences between three *reference* velocities were statistically significantly different (see **Fig. 4** and **Fig. 5**). In summary, these results indicate that the JND motion sense threshold increased proportionally with increases in the velocity stimulus. In other words, ankle motion sense acuity follows *Weber's* law.

IU was significantly different at the three *reference* velocities with a large effect size ($X_F^2(2) = 23.7$, $p < 0.0001$,

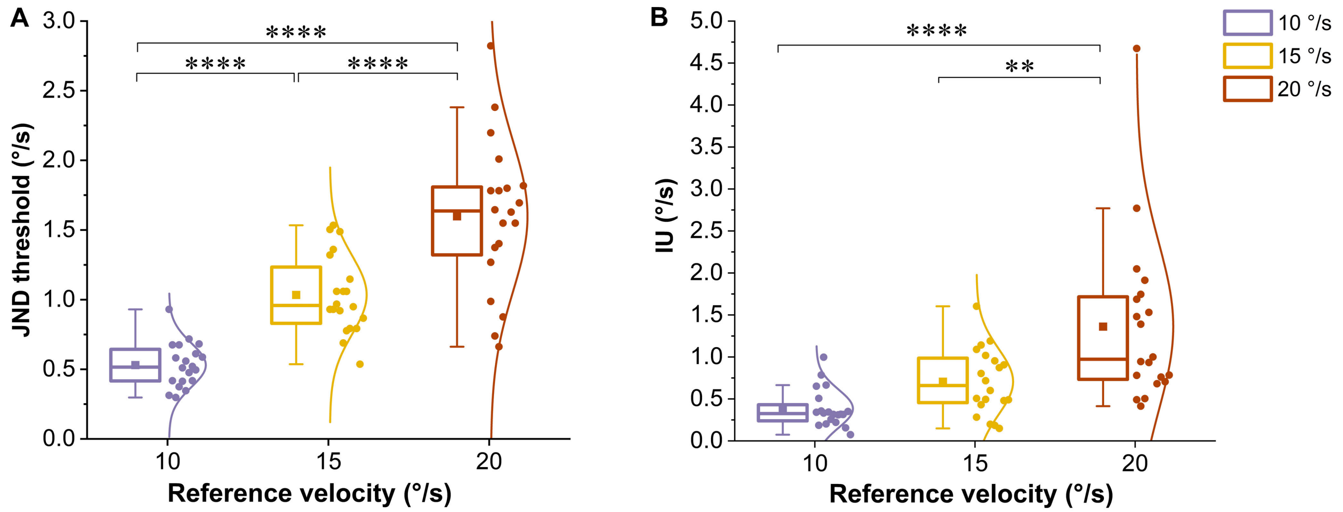

Fig. 4. Motion sense bias and precision at three *reference* velocities. Each box represents the 25-75th percentile. The middle line within a box represents the median. The solid square represents the mean, and the whiskers represent the 1st and 99th percentile. **A**. Bias (JND threshold) and **B**. precision (IU) in all participants. Adjacent circles represent individual subject data and the corresponding distribution. Significant differences are marked based on post-hoc pairwise comparisons (**: $p < 0.01$, ****: $p < 0.0001$).

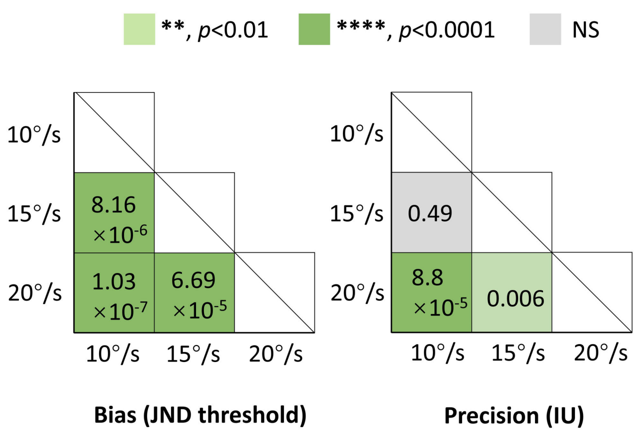

Fig. 5. Post-hoc comparisons between paired *reference* velocities. NS = not significant.

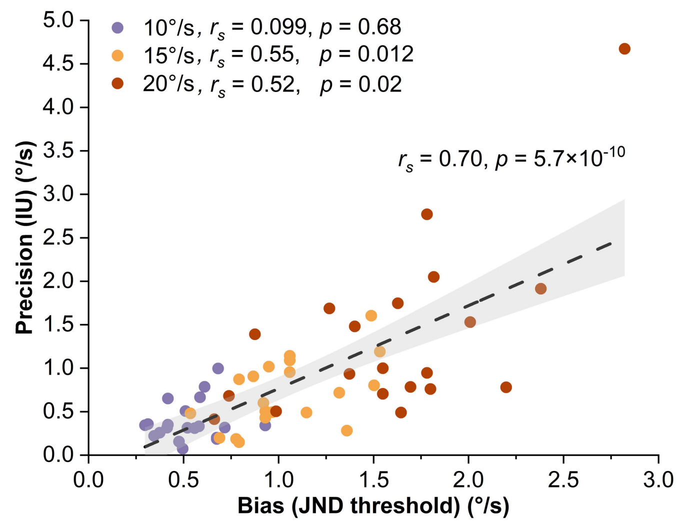

Fig. 6. Relationship between motion sense bias (JND threshold) and precision (IU). Each data point represents the coordinates of a respective JND threshold and IU of an individual participant. Shown are the data for three *reference* velocities. The dashed line represents the linear fit of the JND threshold and IU data. The grey area represents the 95% confidence interval.

$W = 0.59$). Post-hoc analyses using Conover's test revealed IU at 20°/s *reference* velocity was significantly higher than that at 10°/s and 15°/s (see **Fig. 5**).

A significant strong positive correlation was found between motion sense JND threshold and IU ($r_s = 0.70$, $p < 0.0001$) with the increase of *reference* velocity. Moreover, significant moderate correlations were found between two proprioceptive outcome measures at 15°/s ($r_s = 0.55$, $p = 0.012$) and 20°/s ($r_s = 0.52$, $p = 0.02$) *reference* velocities (see **Fig. 6**).

V. Discussion

Motion sense is a submodality of proprioception next to active and passive position sense, and the sense of force [1]. There is a large body of empirical evidence indicating that proprioceptive signals about ankle motion are critical for balance control [3], [4], [6], [14], [41]. Yet, at present, no comprehensive data on ankle motion sense in humans are available. This study applied a robotic system to obtain psychophysical measures of ankle motion sense in healthy human adults. The main outcomes and the scientific yield of our study are summarized as follows: First, we present a robot-aided assessment system that produces quantitative data on ankle motion sense acuity. The system is portable, and the testing time is less than 15 minutes. These features are important when considering the system's potential use in clinical settings. Second, we implement a method that minimizes the potential confounds of position and time cues. The method provides two outcome measures characterizing the accuracy of motion sense: A JND threshold as a measure of bias, and an interval of uncertainty as a measure of motion sense precision. Third, we document that the accuracy of the human ankle motion sense follows *Weber's* law. That is, accuracy diminishes at the higher joint angular velocities. We will discuss these findings in more detail below.

A. Sensitivity of the Device and Test-Retest Reliability

The functional range of motion of the human ankle is between 25 - 60° for plantarflexion (see data in **Table I**). This implies that in order to test an individual's ability to discern between two ankle joint angular velocities, one needs to select *reference* velocities that assure that the perceiver has sufficient time to experience a given joint rotation. Therefore, our assessment protocol selected *reference* velocities between 10 - 20°/s that translated into maximum joint movement times between 1.25 - 6 seconds. We found that the resulting motion sense thresholds of the human participants ranged between 0.3 - 2.82°/s. In contrast, the robot's actuator had an RMS velocity tracking error of 0.091°/s for rotation at a 20°/s target velocity. This implies that the resulting psychophysical threshold data were at least three times above the RMS velocity error, assuring that the device has enough sensitivity to assess human ankle motion sense acuity.

Another important aspect of the assessment procedure relates to its test-retest reliability. To assure that the system can reliably track therapeutic success in clinical populations requires that observable changes over time are not just the results of within-subject variability. Moderate-to-excellent test-retest reliability of the system has been demonstrated in an earlier study [42] yielding an average ICC = 0.88 and a standard error of measurement of 0.0197°/s for the JND motion sense threshold at a 5°/s *reference* velocity on repeated testing over a three-week period.

One caveat concerns the assumption that the human ankle has a fixed center of rotation during dorsiflexion/plantarflexion that is perfectly aligned with the motor's axis of rotation. To ensure consistency in testing and to reduce possible misalignment between the human ankle joint axis and the motor's axis as best as possible, the height and the length of the foot pedal were adjusted to align both axes at the beginning of each experimental session. The reported moderate-to-excellent test-retest reliability observed during repeated testing sessions suggests that the influence of human-robot axis misalignment on the obtained measures of ankle proprioceptive function is minimal.

B. Human Ankle Motion Sense Bias Follows Weber's Law

To our knowledge, this is the first study to systematically evaluate the acuity of the motion sense at the ankle joint. In general, acuity refers to one's ability to discriminate between two sensory stimuli. Here we document that ankle motion sense acuity follows *Weber's* law. This fundamental law of human perception states that the ability to perceive a change in a given stimulus declines proportionally to its magnitude or intensity [26]. Our findings indicate a diminished ability to perceive differences at larger velocities. The respective JND thresholds, as marker of bias, increased linearly as the magnitude of *reference* velocity increased. If expressed in relative terms, the bias of the human ankle motion sense is approximately 5% to 8% of the *reference* velocity (10 - 20°/s).

This finding is consistent with the proprioceptive acuity measures in upper limb joints. For example, the bias of the shoulder motion sense acuity ranged between 7 - 20% for *reference* velocities between 30 - 50°/s [27]. For the elbow joint, a difference of 4.5°/s was discriminated at 15°/s, which increased to 16°/s at a 75°/s *reference* velocity [29]. These studies used the same psychophysical discrimination threshold method as our study to measure joint motion sense acuity.

The corresponding *Weber* fraction that expresses the relationship between threshold and *reference* velocity (i.e., JND/V_R) in our study was $K = 0.11$. This value is comparable to *Weber* fractions for other proprioceptive submodalities such as limb position sense, as well as for other exteroceptive or somatosensory sensory modalities such as vision and touch. For example, the *Weber* fraction for position sense discrimination thresholds at the wrist is 0.09 [23], and for force discrimination at the wrist, elbow and shoulder joints values between 0.08 to 0.13 were reported [43]. Similar values have been found for vision (i.e., brightness discrimination) [44], audition (loudness) [45], [46], and touch [24], [26].

C. Challenges and Pitfalls of Motion Sense Testing

Motion sense testing is more complicated than position sense testing because it requires a device that rotates a limb or limb segment at precise velocities. Moreover, it can be difficult to disentangle position and motion sense, as the testing of joint motion sense inherently means a change in joint position. It has been argued that such motion-position sense coupling challenges quantifying motion sense without accounting for a possible position sense effect [29]. Finally, it is imperative not to provide the perceiver with other motion cues to judge velocity differences. For example, when comparing two joint velocities during a constant displacement, people can simply count the time to infer the higher velocity [27]. Similarly, when keeping the exposure time of a given velocity constant, the perceiver can infer that a larger displacement is associated with a higher velocity [29], [30]. To control for these undesired motion cues, we implemented an adaptive testing procedure that controls for these two motion cues (see **Fig. 2**).

However, the tactile sense from the foot pressure as an extra motion cue cannot entirely be eliminated when the foot pedal begins to rotate. The testing set the participant at a sitting position with the shank (lower limb) unloaded to minimize cutaneous feedback from the foot pedal. The weight of the leg is supported by the pedestal at the knee so that the shank is non-weight bearing, and subsequently, no force due to the weight of the leg is applied (see **Fig. 1B**). Yet somatosensory information about the light touch is likely present and encoded by the signals derived from Merkel disk receptors [47], mechanoreceptors embedded in the basal epidermis [2].

Motion sense testing at the ankle is constrained by the range of motion of the ankle joint. The total range of plantarflexion/dorsiflexion of the tibiotarsal joint is approximately 50 - 70°. Yet, even at slow gait velocities (<0.6 m/s), the average angular velocity of the ankle during plantarflexion is approximately 62°/s [48]. However, psychophysical testing at such angular velocities is difficult. The evaluated person has very little time ($<$1.1s) to perceive an imposed velocity, will tend to guess, and ultimately alienates the test results (e.g., high variability of the perceptual response) [12], [21]. We here tested velocities that were considerably lower than those experienced during normal gait to obtain reliable responses.

D. Implications for Clinical Testing

Present clinical practice to determine ankle proprioceptive status uses the clinical rating scales [9], [10] or biomechanical measures such as the center of pressure (CoP) derived variables (e.g., sway area, sway path) [49], [50]. Rating scales are coarse and are suitable only to detect severe forms of dysfunction, while biomechanical measures of balance dysfunction can only indirectly reflect on proprioceptive status. Moreover, motion sense is typically not assessed at all because of the difficulties in imposing precise angular velocities. Robotic technology has been developed to fill this void. For example, recent studies using a wrist/arm exoskeleton robotic system documented its usefulness in the evaluation of the upper limb position sense in healthy individuals and stroke survivors [18], [19]. The device used in the current study extended such application to the ankle joint. The robotic system couples a simple, yet precise mechatronic device with an accurate and reliable psychophysical discrimination threshold testing method to obtain quantitative data on ankle motion or position sense acuity. An intact motion sense at the ankle is essential for the control of dynamic balance during gait and other forms of locomotion [14]. In contrast, research on stroke survivors showed that ankle motion sense deficits can impair a person's ability to position and load the foot repeatedly during walking [4].

Consequently, a system that can quickly and reliably provide data on motion sense status with sufficient sensitivity constitutes a useful addition to the diagnostic arsenal of clinics assessing and treating ankle dysfunction. This is important given that the currently available clinical scales (e.g., modified Nottingham Sensory Assessment and Rivermead Assessment of Somatosensory Perception) obtaining the observer-based ordinal scales lack such sensitivity [9], [10]. Importantly, the system obtains two distinct measures of proprioceptive acuity, that is, bias (systematic error) and precision (random error), which allows for the comprehensive assessment of proprioceptive function in a clinical setting or for clinical research [51], [52], because people with proprioceptive deficits may show abnormal bias and/or precision [52].

The data on ankle motion sense acuity provided in this study complement the available data on ankle proprioceptive position sense. This is important because either aspect of ankle proprioception can change with age, disease, or therapy. For example, arm position and motion sense deficits two weeks post-stroke show a similar prevalence (56% and 61%). However, after six months, 40% of stroke survivors still exhibited a position sense deficit, while 25% showed a motion sense deficit [18].

Finally, the system is suitable for evaluating therapeutic success with its capability to differentiate and monitor subtle proprioceptive changes. Documenting changes in motion sense acuity over the time course of therapy provides the clinician with a set of somatosensory measures that reflect on the effectiveness of a rehabilitation training program and allows for making informed decisions to adapt the therapy.

VI. Conclusion

To the best of our knowledge, this study is the first to systematically evaluate the acuity of the motion sense at the ankle joint. Our results document that the accuracy of this sense follows *Weber's* law. That is, accuracy diminishes at the higher ankle joint angular velocities. The testing method utilized a robotic device - coupled with an adaptive psychophysical paradigm to objectively measure human ankle motion sense minimizing the potential confounds of position and time cues. It yields two distinct measures of proprioceptive acuity (i.e., bias and precision), allowing for the comprehensive assessment of proprioceptive function. Importantly, the robotic system can easily be moved between care environments, and assessment times are less than 15 minutes - both attributes make the device suitable for use in clinical settings.

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