



Standing postural control during an upper limb reaching task in a mixed reality condition: analysis of center of pressure and lower limb muscle activities

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Abstract

Mixed reality headset monitor display (MR-HMD) is an emerging technology and has been considered for use in health-care and rehabilitation. We aimed to examine the effects of MR-HMD on balance and postural control during a dynamic movement of the upper limb in a standing posture. Twenty participants performed a unilateral repetitive reaching task with or without MR-HMD in the present study. With MR-HMD, the effects of presence or absence of visual feedback on successful reaching were examined. The velocity and 95% confidence ellipse area of the center of pressure (COP) and the surface electromyography (EMG) of four lower limb muscles were recorded during the task. The assessments of head and eye movements with MR-HMD were used to examine the effects of those on COP changes. As a result, velocity (medial-lateral direction) and ellipse area of COP, and EMG activity of the soleus muscle were significantly smaller with visual feedback in the MR-HMD condition than those without MR or visual feedback. The head and eye movements with MR-HMD were significantly smaller than eye tracking conditions, while the ellipse area of COP was oppositely larger with MR due to the upper limb movements. Our findings suggest that wearing MR-HMD causes better stability compared to physical environment that induce specific postural control depending on the visual feedback of the interactions between individual's body and the virtual object.

Keywords Mixed reality · Center of pressure · Electromyography · Body stability · Postural control · Head and eye movement tracking

1 Introduction

Postural control is a crucial ability in daily life and involves the integration of visual, vestibular, proprioceptive, and tactile senses that are incorporated into the various neuromuscular functions (Chiba et al. 2016). Aging, physical degeneration, and pathological factors are vulnerable to an increased risk of falls due to increased instability in balance (Silsupadol et al. 2006; Toraman and Yildirim 2010; Chang and Do 2015). The center of pressure (COP) is an

effective assessment to examine postural control (Quijoux et al. 2021), and previous studies have shown greater COP tends to reflect weaker body control in dynamic and static equilibrium states (Bonnet and Szaffarczyk 2017; Fujikake et al. 2009). Maintaining a standing position requires the synergistic action of the hip and peri-ankle muscles (Henry et al. 1998) and, when disturbed, healthy populations preferentially use the peri-ankle strategy to adjust their posture back to a stable state (Maylor et al. 2001). The muscles around the ankle joint are thought to play an important role in adjusting posture, and it is known that instability always accompanies the increased lower limb muscle activity (Gebel et al. 2019).

Rehabilitation strategies regarding balance include improving muscle strength, proprioception, and vestibular function (Herdman et al. 2001). Although these approaches are considered effective in improving balance, traditional therapies can be a burden on the patient's time, finances,

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and energy (Lei et al. 2019). Most recently, extended reality involving virtual reality (VR), mixed reality (MR), and augmented reality (AR) has also been used in training for the prevention of falls and rehabilitation of muscle injuries in the lower extremities as it can be designed to be more relevant to everyday or specific environments (Schuermans et al. 2022). For example, the effectiveness of immersive VR technology in dynamic and static balance through providing an environment that has no interaction with the real environment is still controversial (Chen et al. 2016). There is evidence proposing that virtual reality as a tool for balance training produces comparable effects to traditional therapy (Prasertsakul et al. 2018), while other studies reported that VR produces physical instability due to conflicting visual and vestibular systems (Saldana et al. 2020; Robert et al. 2016). One previous study has also suggested that postural sway during VR interventions was consistent with swaying with eyes closed during quiet standing and that this instability may persist until the end of immersion in VR (Shahnawaz et al. 2021).

On the other hand, MR is a technology that combines VR and AR, which can interact with virtual and real objects (Fu et al. 2022), and it exists on a spectrum between entirely virtual and real environments (Rauschnabel et al. 2022). With regard to motor performance, MR interventions have demonstrated the capacity to enhance motivation and facilitate learning and exercise (Anderson et al. 2013). However, MR technology available at the time differs from the current devices. Presently, MR devices include projection devices and computer screens at first (Krukowski et al. 2014; Gotsis et al. 2012), and once the headset monitor display (HMD) is developed the MR-HMD is thought to be a potentially effective and motivating strategy for medical care, e.g., clinical interface and surgery (Martin et al. 2020; Chen et al. 2017). MR-HMD has advantages primarily in representing the depth of motion in 3-dimensional space (Butz et al. 2022). In contrast to VR where the user is removed from the real world and fully immersed in the virtual environment (Eswaran et al. 2022), MR does not completely filter the real environment, and the MR-HMD can alleviate the anxiety associated with using immersive devices (Shahnawaz et al. 2021). On the other hand, in comparison with the conventional AR which is constrained to simulating physical objects in real-time using digital technology within the physical environment (Wang et al. 2021), MR facilitates real-time interaction between the physical world and the virtual environment and objects (Rokhsaritalemi et al. 2020). Koop et al. verified the agreement of gait performance when using MR-HMD and compared the performance with conventional biomechanical measurements, confirming that it can be used as a tool for measuring balance (Koop et al. 2020). Furthermore, Lee et al. (2019) found no sole effect

of MR-HMD on body sway when comparing the COP with or without visual feedback. To date, however, there has been no clear evidence that postural control during dynamic upper limb movement varies depending on the MR-HMD condition. Reaching movements of the upper limb are consciously and unconsciously engaged in numerous daily activities, which require the coordination of both upper and lower limbs with the trunk, especially in standing situations (Huang and Brown 2013). Previous study has also demonstrated that the reaching movement involved all body directions, in which constant adjustment of the COP is needed, causing interference with posture (Stamenkovic et al. 2021).

By utilizing a reaching task in a standing position, this study was conducted to verify whether postural control with MR-HMD differs from that in the physical environment by comparing differences in body sway and lower extremity electromyography (EMG). Moreover, the state of balance was also compared when completing MR tasks with and without visual feedback to explore the factors that affect postural control in MR. Further, the influence of head and eye movements on the COP with MR was also examined, because it has been shown that the head instability increase the risk of postural instability (Paloski et al. 2006), and that the eye movements may also potentially affect postural control (Jahn et al. 2002). We used these control conditions to explore the impacts, if any, of head and eye movement on the COP changes in the MR condition. We hypothesized that the MR condition would lead to better postural control than the physical environment even during a dynamic movement, and that the visuomotor integration and interaction in the MR condition may play a crucial role in maintaining the body posture.

2 Methods

2.1 Participants

Twenty right-handed (Flanders Handedness survey questionnaire, 9.3 ± 1.7) (Nicholls et al. 2013; Okubo et al. 2014) healthy adults were involved in this study (13 females, mean age, 27.7 ± 6.2 years, height, 162.0 ± 6.3 cm, weight, 53.8 ± 6.4 kg, means $\pm$ SD). A sample size of 20 was calculated based on a priori power analysis for one-way analysis of variance (ANOVA) with repeated measures (G*power ver. 3.1.9.6). We applied an effect size of 0.3 and an alpha of 0.05 to achieve a power of 0.8. The effect size was determined according to the Cohen's statement that the values of 0.0 to 0.5 indicate a small effect (Cohen 1992). All participants provided written informed consent prior to the experiments. The experimental procedures and protocols were approved by the Ethics Committee of Kyoto University

Graduate School and Faculty of Medicine (Permit No. C1492) and were performed according to the Declaration of Helsinki.

2.2 Apparatus

2.2.1 Mixed reality setup

An MR head-mounted stereoscopic display (HoloLens2, Microsoft Corporation, USA) was used to construct the mixed reality environment, by which users can interact with real and virtual environments (resolution, 2048×1080 pixels per eye; refresh rate, 60 Hz). The wearer can view not only the object setting in the device through the glasses but also the surroundings. In addition, a depth camera, and four additional cameras with built-in sensors were used for environmental sensing and detecting ambient light (Hanna et al. 2018). The platform is equipped with a Windows system that allows users to operate through gestures directly through the MR glasses. Two infrared cameras enable sight tracking in real-time at a frequency of 30 Hz and manipulate the objects in the scene by eye movement.

In addition to the MR glasses, a computer (Surface, Microsoft Corporation, USA) was used for manipulation of the task. In this experiment, the angle parallel and perpendicular to the line of sight was set to 30 degrees respectively by the software (Rehamaru software, Techlico Inc, Japan). The realistic environmental conditions were controlled by a separate computer (HP Inc. USA) and implemented by Microsoft PowerPoints.

2.2.2 Center of pressure

A force platform (TF-6090, Tech Gihan Co. Ltd., Japan) was used to collect COP data at 1000 Hz initially. The data were exported to data acquisition software (LabChart7, AD Instruments, Sydney, Australia) and low pass filtered with a cut-off frequency of 20 Hz. In this study, the length (mm), mean velocity (mm/s), mean velocity in medial-lateral direction (mean velocity ML), anterior-posterior direction (mean velocity AP), and 95% confidence ellipse area (mm^2) of COP were calculated, which was referred to the formula for COP as follows:

$$\text{SWAYLENGTH} = \sum_{n=1}^{N-1} \left[(\text{AP}[n+1] - \text{AP}[n])^2 + (\text{ML}[n+1] - \text{ML}[n])^2 \right]^{1/2}$$

$$\text{SWAYLENGTH}_{\text{AP}} = \sum_{n=1}^{N-1} |\text{AP}[n+1] - \text{AP}[n]|$$

$$\text{SWAYLENGTH}_{\text{ML}} = \sum_{n=1}^{N-1} |\text{ML}[n+1] - \text{ML}[n]|$$

$$\text{MEAN VELOCITY} = \frac{\text{SWAYLENGTH}}{\text{TIME}}$$

$$\text{AREA} = \pi ab = 2\pi F_{0.05[2, n-2]} [s_{\text{AP}}^2 s_{\text{ML}}^2 - s_{\text{APML}}^2]^{1/2}$$

where N stands for the number of points of the signal, n for the coordinates of the COP per unit time ($1 \leq n \leq N$), a and b for the major and minor radii of the 95% confidence ellipse, $F_{0.05[2, n-2]}$ for the F statistic at a 95% confidence level for a bivariate distribution with n data points, and S for the standard deviation (Prieto et al. 1996).

2.2.3 Surface EMG recordings

EMG was recorded from the peroneus longus (PL), tibialis anterior (TA), gastrocnemius medialis (GM), and soleus (SOL) muscles of the right lower limb. A pair of silver-bar electrodes (Bagnoli-4 EMG System, Delsys, USA) was used to collect the signal from these muscles. The placement of electrodes was followed by SENIAM guideline (SENIAM 2014). The EMG signals were amplified and band-pass filtered at 20 Hz–2 kHz. Data acquisition software (LabChart7, ADInstruments, Sydney, Australia) with a PowerLab analog-to-digital converter (PowerLab 8/30, AD Instruments, Sydney, Australia) was used to record the EMG activities with a sampling rate of 5 kHz. Before the task, EMG activity with maximum voluntary contraction (MVC) was obtained in each muscle. The integral EMG (iEMG) of each muscle during the task was calculated and normalized to the MVC. The signals from the EMG electrodes were stored on a lab computer for later offline analysis (LabChart7, ADInstruments, Sydney, Australia).

2.3 Experimental procedures

The experiment was conducted in a soundproof room with the participant standing upright on a force plate, with their feet together and looking straight ahead. With MR, the distance from the participant's eyes to the projected object was adjusted between 65 and 70 cm, and the angle parallel and perpendicular to the line of sight was adjusted to 30 degrees. This process was operated in MR, which was equal to participants viewing a 55 cm x 55 cm projection on a screen. The participants were then asked to perform the motor tasks with their right upper limb in MR condition. The target area was adjusted to 55 cm x 55 cm, the same range in the MR condition.

Regarding the motor task, ten red spheres labeled with serial numbers from 1 to 10 with a white background were presented with the MR device (with MR condition) or with the display (without MR condition) (Fig. 1). The participants were instructed to put their right hand near their chest as the initial position and then to ‘reach and touch the spheres in number order as fast and accurate as possible after the start cue (10 reaches in a session). For the condition without MR, participants were asked to touch the targets on display set in front of them. The display was adjusted to a position so that the participants could touch it by reaching naturally and receive tactile feedback similar to the MR condition. The start instruction is followed by a 5-s countdown. After the 5 seconds, the participants started the task of reaching and touching the target with their right index finger and then put the right hand back to their chest after each reach. After completing the last reach (number 10) in a session, the participant’s arm returned to its natural drooping position.

2.4 Tasks and instructions

Before the main protocols, participants completed a 30 s eyes-open stand with bipedal (two-legged standing with feet together) and unipedal stances (right or left one-legged standing). With two-legged standing, it has been suggested that the measurement of COP performance can reflect personal balance ability (Ageberg et al. 2003). With one-legged

standing, on the other hand, the interlimb postural comparison has been studied numerously, but the results are contradictory and do not lead to a consensus (Paillard and Noé 2020). Because we aimed to determine the differences in balance and postural control between with and without MR conditions (reaching tasks in the MR condition or in the real environment, see below for details), these control data would help us to exclude the effects of individual balance difference and/or interlimb difference if any.

Totally five motor tasks were included in the main protocols, i.e., three reaching tasks and two tracking tasks. In the reaching tasks, the participants were encouraged to perform reaching and touching the target with or without MR. In the tracking tasks, on the other hand, the participants were asked to eliminate the targets by moving the head (head tracking) or eyes (eye tracking) instead of reaching movements. Each task was repeated three times in different consecutive sessions (total of 15 sessions), and the order of tasks was randomized between the participants. Prior to each task, several trials were performed to familiarize the participants with the tasks.

2.4.1 Reaching tasks

Three reaching tasks are, (1) touching the targets without the MR device (noMR), participants wore an MR device on their heads, but the device’s display was lifted and

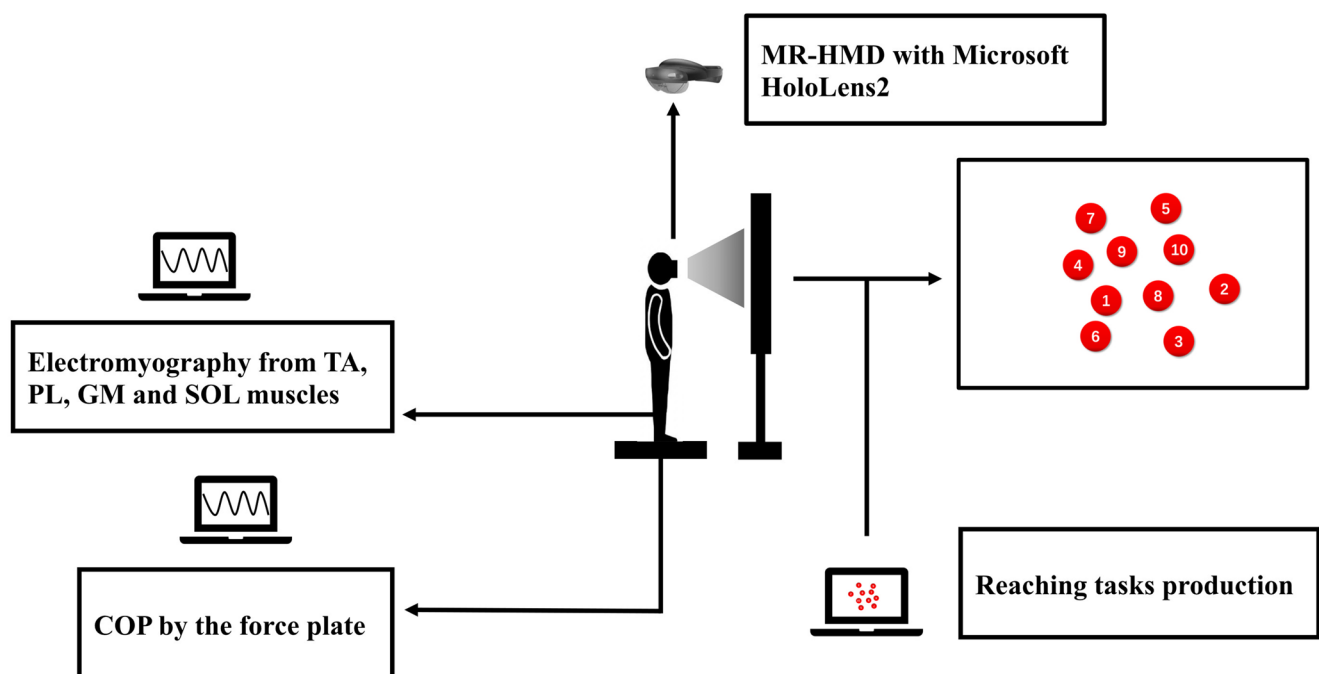


Fig. 1 The illustration of the experimental equipment. The participant performed reaching tasks to touch the red spheres one-by-one with or without the mixed reality headset monitor display (MR-HMD) of Microsoft HoloLens2. Center of pressure (COP) was recorded by a

force plate under the feet of the participant, and electromyography (EMG) signals were recorded from four lower limb muscles. PL, peroneus longus muscle; TA, tibialis anterior muscle; GM, gastrocnemius muscle; SOL, soleus muscle

participants saw the task being presented on the front display in the real environment, (2) eliminating the targets by touching them in the MR condition (MR (target−)), participant completes the motor task through MR, which is set up in the same way as the “noMR” condition, and received a visual feedback of touch (elimination of the target within 1 s), and (3) touching the targets in the MR condition but the target was not eliminated (MR (target+)). The most important characteristic of MR was the interaction of virtual (spheres) and real (their own hand) objects which was represented repeatedly as elimination of the target one by one. By comparing the results of tasks (1) and (2), we aimed to investigate whether changes in COP and lower limb muscle activity exhibited differences between with and without MR conditions. Because this task can be assumed to be a control condition for MR without visual feedback of successful touch, we aimed to further explore whether the interaction of virtual and real objects in the MR condition plays a role in mediating the COP as well as EMG activity if any, by comparing the results of tasks (2) and (3).

2.4.2 Tracking tasks

Since head and eye movements should also be taken into account as affecting balance (Larsson et al. 2016), we adopted two additional tasks in the MR condition to examine whether the effects of head or eye movement on the biomechanical parameters. Two tracking tasks are, (4) eliminating the targets by moving the head without eye and trunk movements (head tracking with MR, HT-MR), and (5) eliminating the targets by moving the eyes without head and trunk movements (eye tracking with MR, ET-MR). A cursor in the MR environment was presented and linked to head or eye movement, respectively, by which participants moved their head or eyes and received visual feedback of eliminating the target by moving the cursor on the target within 1 s.

2.5 Data analysis

The completion time, the mean velocity of COP, mean velocity in the medial-lateral direction and anterior-posterior direction, 95% confidence ellipse area of COP, normalized iEMG, rectangular area of head movement, and eye movement were calculated under all conditions. Data from one participant while standing on the left foot was not successfully recorded because of the malfunction in the force plate, and the data of normalized COP velocity and area by left stance was $N=19$. The completion time was calculated from

the moment after the countdown ended when the tasks were launched to the time point to finish the last reach of number 10 in conditions. The rectangular area of head movement and eye movement were analyzed through tasks (2), (4), and (5). The rectangular area was calculated by multiplying the maximum distance moved in the lateral direction by the maximum distance moved in the longitudinal direction for the head and eye movements recorded in the MR device, respectively.

2.6 Statistical analysis

The Shapiro-Wilk test was used to test the normality of the distribution of values. For parameters that were normally distributed and log-transformed to be normally distributed, one-way ANOVA with repeated measures was performed with mean velocity and 95% confidence ellipse area, velocity ML, velocity AP between “noMR”, “MR (target−)”, and “MR (target+)”. Except for the velocity ML, the rest of the parameters are log-transformed to conform to a normal distribution. For cases that did not follow Mauchly’s test of sphericity, Greenhouse-Geisser corrected significance values were reported for these results. For parameters that violated the normal distribution, which included completion time, normalized COP velocity, normalized COP area, normalized iEMG, and rectangular area of head and eye movement, the Friedman test was performed to check the difference among conditions. Apart from the main results, we also examined the difference in the COP, eye movement, and head movement among the conditions of MR (target−), Head tracking, and Eye tracking to explore the impacts of head and eye movement on each parameter in the MR condition. The P values for the post-hoc examination among multiple comparisons were adjusted using a Bonferroni correction. The level of statistical significance was defined as $P<0.05$. The data values are expressed as the means $\pm$ standard deviations (SD) or median with interquartile range (IQR).

3 Results

3.1 Completion time

The Friedman test showed the significant difference in completion time among the conditions (median value, noMR, 10.82 s (IQR: 2.95s); MR (target−), 14.91 s (IQR: 5.82s); MR (target+), 11.84 s (IQR: 4.13s), $P<0.001$).

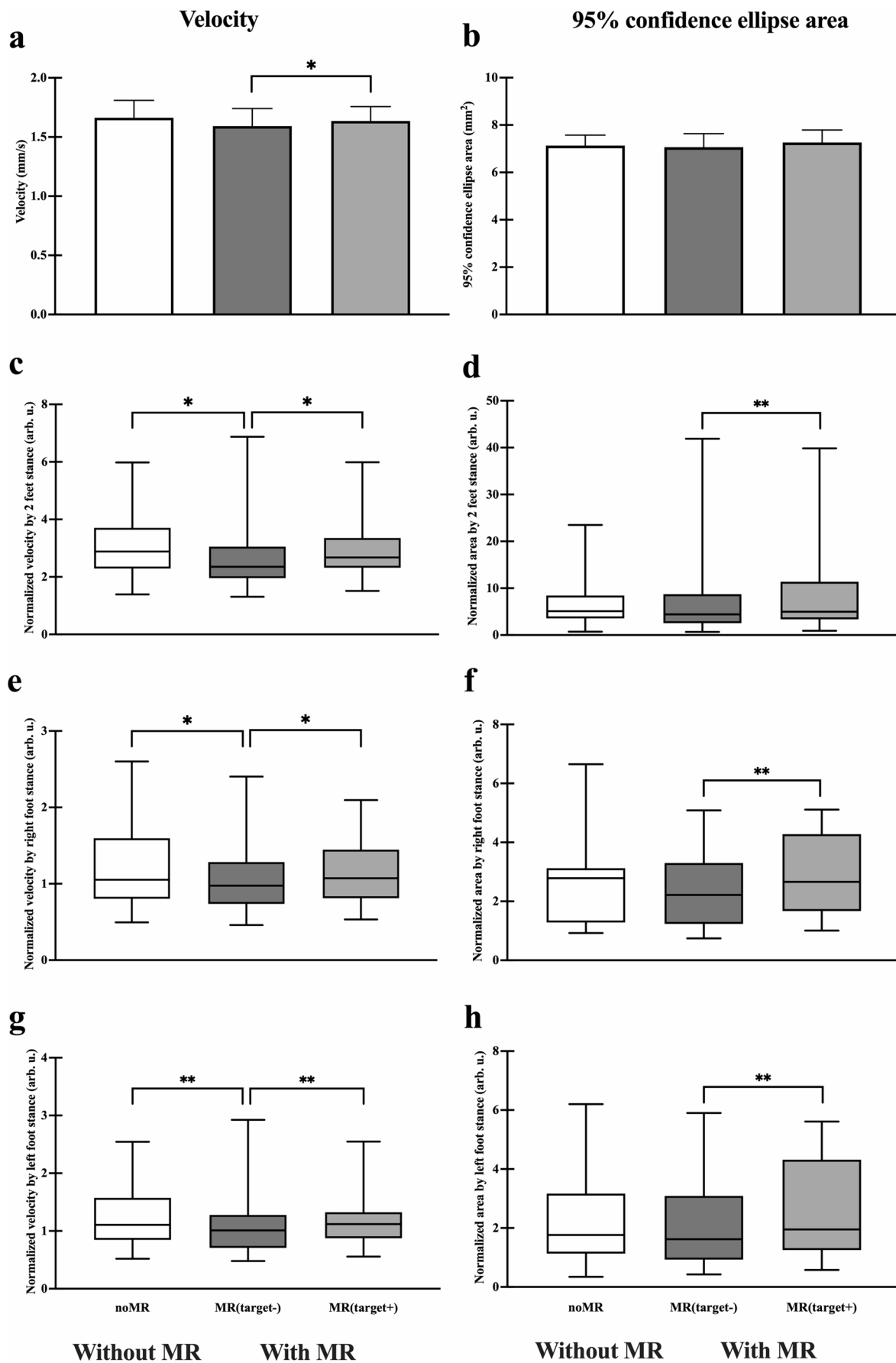


Fig. 2 Mean velocity (a) and 95% confidence ellipse area (b) after logarithmic transformation among conditions of “noMR”, “MR (target−)”, and “MR (target+)”, the former is a logarithmic conversion with a base of 10. c, e, and g show the normalized center of pressure (COP) velocity by static stance, right foot stance, and left foot stance. d, f, and h show the normalized COP area by static stance, right foot stance and left foot stance. Vertical bars represent standard deviations, asterisks represent statistical significance (*: $P < 0.05$, **: $P < 0.01$, ***: $P < 0.001$)

3.2 Center of pressure

Figure 2 summarizes the group mean velocity and 95% confidence ellipse area of COP among the “noMR”, “MR (target−)”, and “MR (target+)” conditions. Regarding the velocity (Fig. 2a), a significant difference was detected among the conditions ($F(1.4, 26.59) = 4.20$, $P < 0.05$). Post-hoc analysis revealed that the COP velocity was significantly lower with MR (target−) than with MR (target+) ($P < 0.05$). However, no significant difference was found when

comparing “noMR” condition with the “MR (target−)” condition ($P = 0.10$), and “MR (target+)” condition ($P = 0.87$). The COP profiles during two- and one-legged standing are shown in Table 1. The normalized COP velocity by two and one leg standing also showed similar results among the conditions (Fig. 2c, $\chi^2(2) = 7.50$, $P < 0.05$; Fig. 2e, $\chi^2(2) = 7.50$, $P < 0.05$; Fig. 2g, $\chi^2(2) = 9.58$, $P < 0.01$). The COP velocity was significantly lower with MR (target−) than noMR condition and MR (target+) condition ($P < 0.05$, respectively).

Regarding the 95% confidence ellipse area (Fig. 2b), one-way ANOVA showed no significant difference among these conditions ($F(2, 38) = 2.71$, $P = 0.08$). Normalized to two- and one-legged standing (Fig. 2d, f, and h), Friedman test revealed a significant difference in the normalized area among the conditions ($P < 0.05$). Post-hoc analyses performed by Bonferroni correction indicated that the normalized area of COP was significantly lower with MR (target−) condition than MR (target+) condition ($P < 0.01$, respectively).

To further examine the directional difference in COP among conditions, the COP velocity in the ML and AP directions is shown in Fig. 3. One-way ANOVA showed a significant difference in ML direction (Fig. 3a, $F(2, 38) = 9.77$, $P < 0.001$) but not in AP direction (Fig. 3b, $F(2, 38) = 2.47$, $P = 0.10$). Post-hoc analyses revealed that the velocity in ML direction was significantly lower with MR (target−) condition than that with noMR condition ($P < 0.05$) and MR (target+) condition ($P < 0.001$).

Table 1 Velocity and 95% confidence ellipse area of COP during standings

	Velocity (mm/s)	95% Confidence ellipse area (mm ²)
Two-legged standing	15.66 (14.76–16.95) ***	231.57 (197.36–409.78) ***
	IQR: 2.44	IQR: 201.33
One-legged standing (right)	38.71 (35.55–47.86) IQR: 14.18	651.82 (488.41–709.12) IQR: 296.07
One-legged standing (left)	36.52 (35.13–47.3) IQR: 9.29	606.12 (555.56–993.64) IQR: 611.36

Values are median with 95% CI (shown in brackets) and IQR, *** $P < 0.001$. 95% CI: 95% confidence interval; IQR: Interquartile range (IQR)

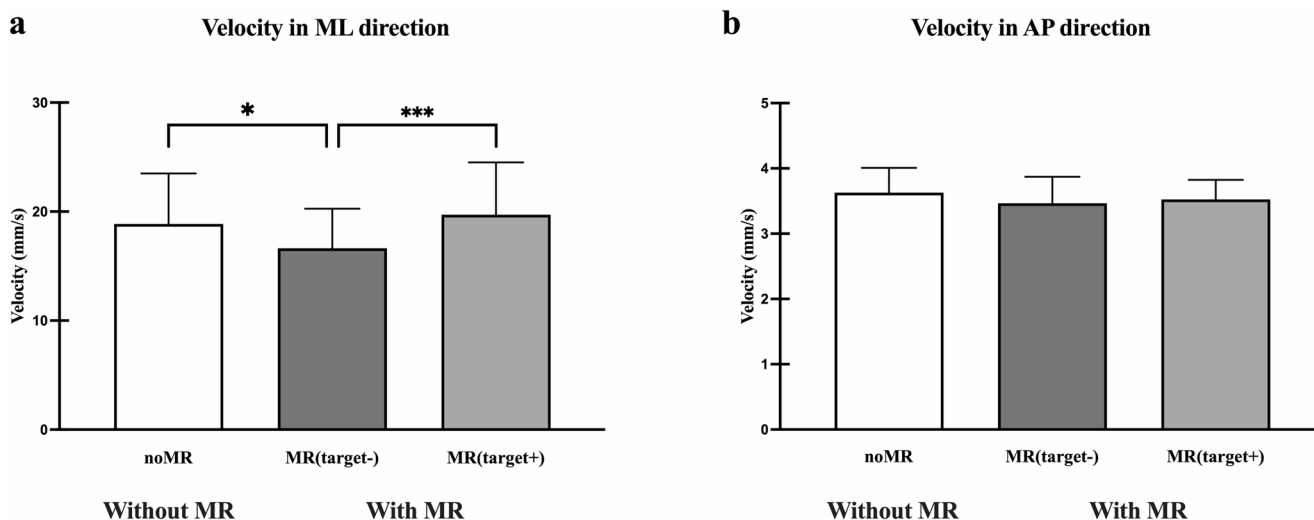


Fig. 3 Mean velocity in the medial-lateral (ML) and anterior-posterior (AP) directions among the conditions. Vertical bars represent standard deviations, asterisks represent statistical significance (*: $P < 0.05$, **: $P < 0.01$, ***: $P < 0.001$)

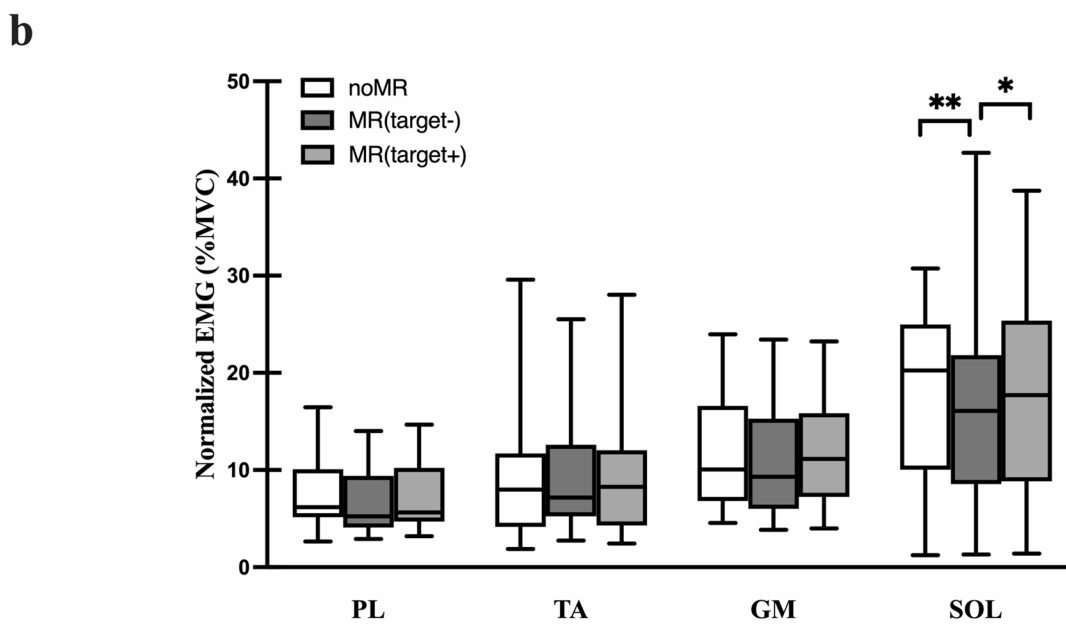
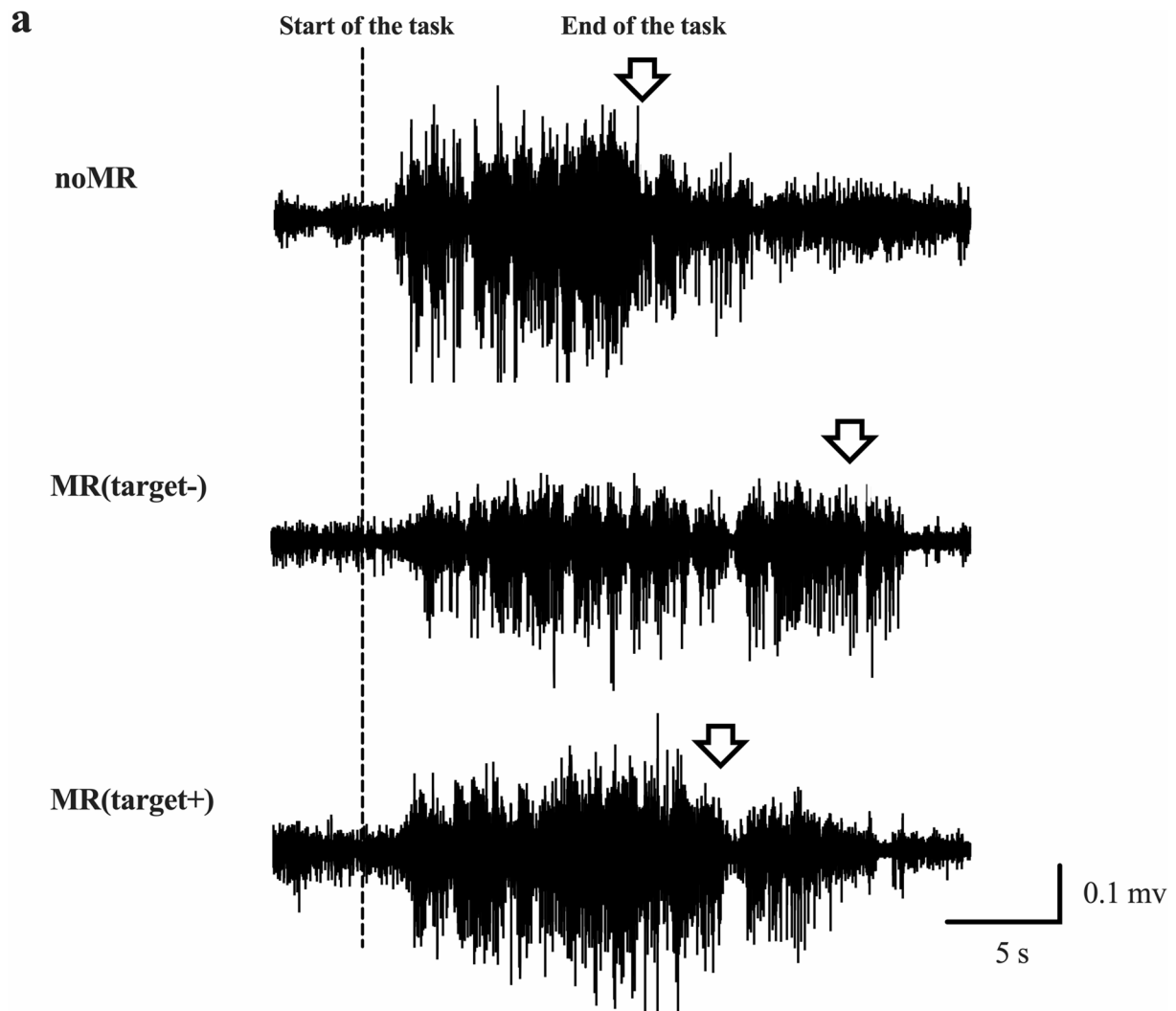


Fig. 4 a The representative recordings of surface electromyography (EMG) of soleus muscle in the conditions. The vertical dashed line indicates the start of the motor task, and the arrows indicate the end of the motor task. **b** The normalized EMG of four lower limb muscles among the conditions. The results are presented as median value. Vertical bars represent min to max; asterisks represent statistical significance (*: $P < 0.05$, **: $P < 0.01$, ***: $P < 0.001$)

3.3 iEMG

Original EMG tracings of the SOL muscle in a representative individual and the group mean EMG activity of all four muscles among the conditions are presented in Fig. 4. Although no significant difference was found in PL, TA, and GM muscles, a significant difference among the conditions was detected in the SOL muscle ($\chi^2(2) = 9.70$, $P < 0.01$). The iEMG activity of SOL muscle was significantly smaller with MR (target−) condition compared to noMR conditions ($Z = 3.00$, $P < 0.01$), and MR (target+) ($Z = 2.21$, $P = 0.03$), which is consistent with the results of COP.

3.4 Head and eye movements

With the MR (target−) condition, we further examined the contribution of head and eye movement to the COP changes. Figure 5 shows representative recordings of COP, eye movement, and head movement with MR (target−) condition as well as HT-MR and ET-MR conditions, and Fig. 6 summarizes the group mean data. The COP area was significantly larger with MR (target−) condition than that in HT-MR and ET-MR condition (Fig. 6a, $F(1.5, 28) = 72.48$, $P < 0.001$; post-hoc, $P < 0.05$, respectively). The rectangular area of the head movement was expectedly lower in the ET-MR condition than MR (target−) and HT-MR conditions (Fig. 6b, $P < 0.01$, respectively). Conversely, the rectangular area of eye movement was significantly smaller with the MR (target−) condition than the other two conditions (Fig. 6c, $P < 0.05$, respectively).

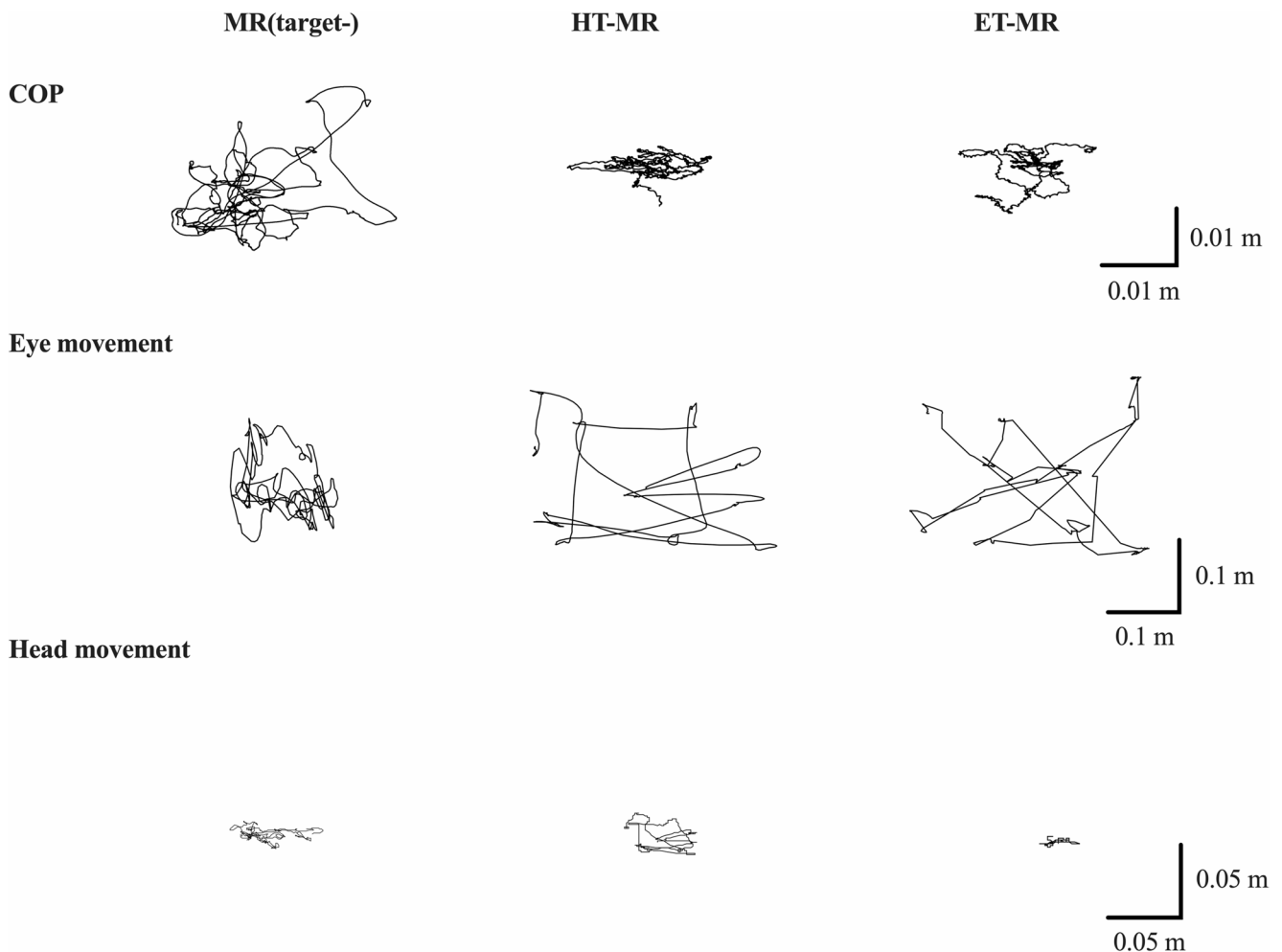
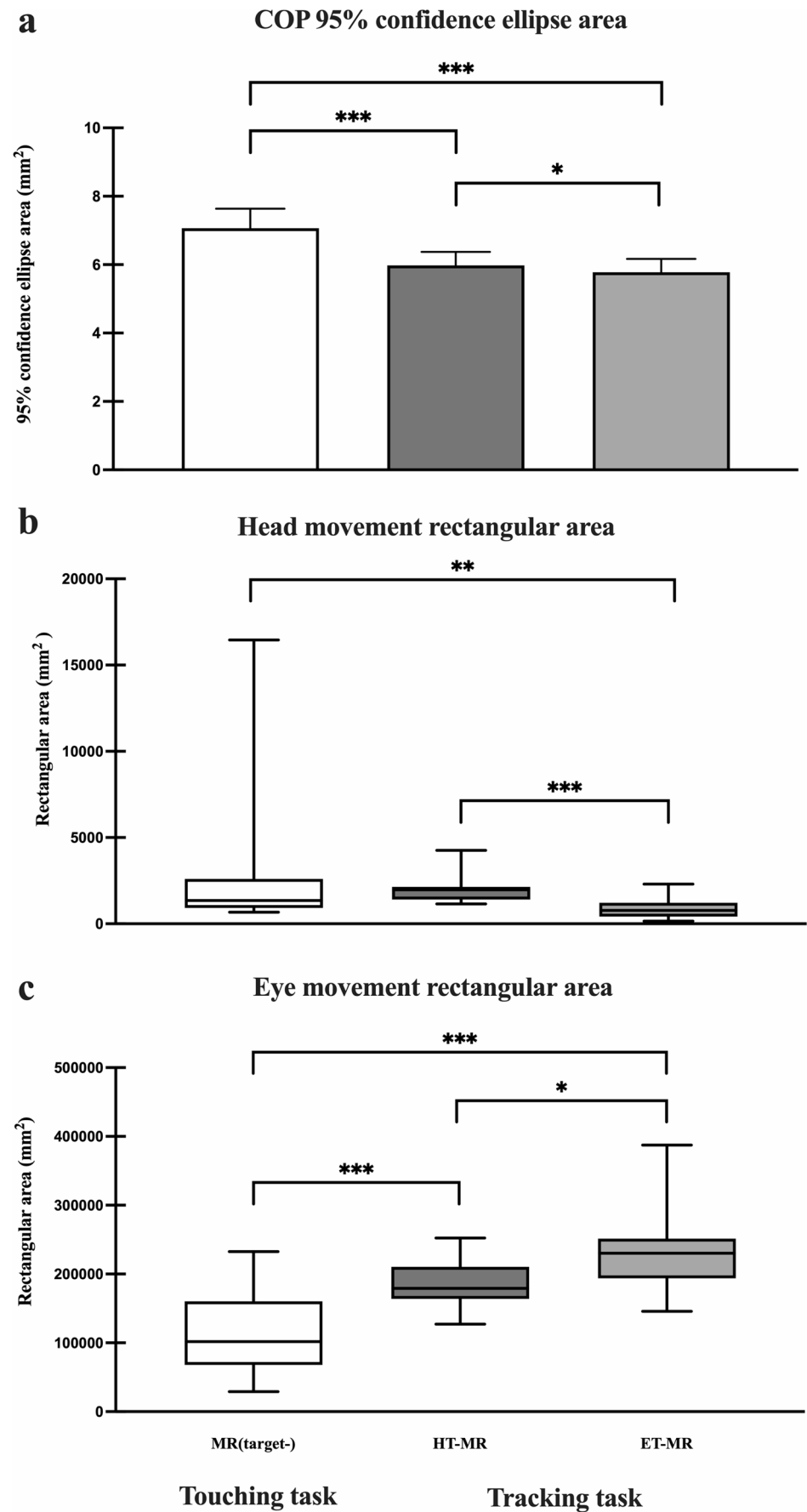


Fig. 5 The representative recordings of center of pressure (COP), eye movement, and head movement during reaching task with MR (target−), HT-MR, and ET-MR, respectively. Note that the smaller eye and head movements accompanied with larger COP changes with MR

Fig. 6 The 95% confidence ellipse area of center of pressure (COP) (a), rectangular area of head movement (b), and eye movement (c). The results are presented as median value. Vertical bars represent min to max; asterisks represent statistical significance (*: $P < 0.05$, **: $P < 0.01$, ***: $P < 0.001$)



4 Discussion

The major finding from this study was that the body sway was significantly smaller with MR than that without MR during an upper limb reaching task in an upright standing position. The smaller change in COP accompanied the lower EMG activity of the lower limb muscle, i.e., SOL muscle. Interestingly, in the condition of MR without visual feedback (MR (target+)), the change in COP became larger and that was comparable to that without MR, suggesting that the virtual tactile feedback with MR (visual feedback of successful reaching) played a crucial role in decreasing the body sway as well as lower limb muscle activity. To the best of our knowledge, this study provides the first evidence that MR with visual feedback may affect postural control and decrease body sway during reaching tasks.

4.1 MR

The HoloLens2 used in this study is a semi-immersive device. It allows the user to interact naturally with the object, as opposed to VR, which complements the real-world environment without replacing it, which may be one of the advantages of MR. Some researchers have suggested its motivational value for patients with shoulder pain (Conдино et al. 2019) and physical disabilities (Howard and Davis 2022). Since MR is still in an exploratory phase in medical applications, there is a great variation between current MR rehabilitation programs, and the main goal is to demonstrate the feasibility and safety of this application (Howard and Davis 2022). In the present study, we have challenged the application of MR in a reaching task during upright standing and tested the changes in the biomechanical parameters. Although we found a distinct difference in the change of COP and lower limb muscle activity between the conditions without and with MR, whether the visuomotor integration and interaction involved in MR play a role in maintaining the balance was unclear. To answer this question, we utilized a control condition for MR, namely MR without visual feedback (MR (target+)), and found that the changes in COP and muscle activity became larger than those with MR (target-) and were even comparable to those in the noMR condition. By tracking the head and eye movements with the MR device, we further examined the impacts of these on motor performance, i.e., COP.

4.2 Postural control and lower limb muscle activity with MR

Since the individual bipedal and unipedal balance ability might vary among individuals (Table 1) and then influence on the standing balance during a motor task, we have

normalized the COP velocity to the bipedal and unipedal stances to exclude the effects of individual balance difference and/or interlimb difference (Ageberg et al. 2003; Hertel et al. 2002; Paillard and Noé 2020). Although there was no significant difference between noMR and MR (target-) conditions with the velocity data (Fig. 2a), a significant difference could be detected when using the normalized velocity (Fig. 2c, e, and g), suggesting that normalizing the COP velocity to the bipedal and unipedal stances would be more sensitive to detect the difference between the conditions.

By comparing the changes in COP and iEMG in various conditions with and without MR, we found less body sway and iEMG when participants conducted the MR task compared to the physical condition. It should be noted that the completion time in MR (target-) condition was longer than that in the noMR or MR (target+) condition, even though the identical instruction of the motor task was given to the participants, i.e., to reach and touch the target 'spheres' in number order as fast and accurate as possible. It can be assumed that visual feedback of successful touch in the MR (target-) condition took more time than the other two conditions, in which only tactile feedback was provided by touching the actual display.

While a previous study has mentioned interference with balance due to conflicting visual and vestibular input sensations when using VR (Nishiike et al. 2013), the present results demonstrated that MR does not cause the disturbance as VR does. On the contrary, the MR (target-) condition even produced better postural control than that in the noMR and MR (target+) conditions. Maintaining standing posture involves anteroposterior as well as mediolateral displacement or rotations (Müller and Redfern 2004). Our results of COP velocity revealed differences in the ML direction but not in the AP direction. The movement in the AP and ML directions corresponded to different postural control strategies (Day et al. 1993; Winter et al. 1996). When disturbances to balance occur, control of the trunk and hips is required to move the body back to balance (Winter 1995). Sway in AP direction benefited more from the synergy of the plantar flexors and dorsiflexors around the ankle joint, while that in ML direction has been proposed with the activity of the abductor and adductor muscles around the hip (Gatev et al. 1999). The forward extension movement did not induce a greater amplitude of ankle movement (Fujiiwara et al. 2011) thereby not triggering a difference in the movement of the COP in AP direction. Lower velocities in the ML direction in the MR (target-) condition may indicate better control of the hips and trunk. Although the profile of iEMG of SOL activity is often used to explain ankle movements in the AP direction, Cimadoro et al. (2013) showed that high iEMG activity in SOL when the body was

restricted to ML movements. In the present study, therefore, muscle activity of SOL may contribute to the body sway in the ML direction.

4.3 Effects of visual information during standing

While a previous study has shown that biomechanical and cardiovascular variables were unaffected by wearing MR glasses during several cognitive tasks (Cometti et al. 2018), it is of interest to question why MR reduced the COP as well as lower limb muscle activity during a reaching task in the present study. Previous studies have discussed the importance of peripheral visual information on physical stability, and it is known that the peripheral field of view plays an important role in controlling posture (Diniz-Filho et al. 2015). Blocking of peripheral vision increases body sway, especially in standing conditions, which proves the importance of field of view on the impact of balance (Horiuchi et al. 2017; Duh et al. 2001; Streepey et al. 2007). Therefore, the better or comparable performance in the MR (target±) conditions compared to that without MR (noMR) could be explained by the fact that the MR device did not block the peripheral vision and has the same peripheral field of view as the real environment. The importance of leg and foot visualization on balance has also been emphasized in the previous study (Pastel et al. 2020). When performing a balance beam via VR, they showed that the number of foot impacts and completion times were significantly less in the case of visualized feet than that of nonvisualized feet (Pastel et al. 2020). In our study, the position of the body and extremities, especially the upper and lower extremities, were not completely obscured when wearing MR glasses and can be perceived by the wearer.

Maintaining standing has demands on attentional abilities (Basso et al. 2014), especially in daily life where different cognitive tasks need to be performed together, which requires a reallocation of attention from postural control to cognitive tasks (Borel and Alescio-Lautier 2014). It has been suggested that better balance performance may benefit from attention shifting from postural control to cognitive tasks (Berger and Bernard-Demanze 2011; Lakhani and Mansfield 2015). A memory-spatial task might induce lower acceleration of the center of gravity (Berger and Bernard-Demanze 2011), while a visual feedback task over two consecutive days could increase entropy of COP samples, with no changes in participants' postural stability (Lakhani and Mansfield 2015). These results led the authors to speculate that postural control may be delegated to a highly automated process, resulting in a better balance performance, when the attention was shifted from the postural task to the cognitive task. In the present study, the repetitive reaching and touching of the targets also involves a cognitive demand, which

might be emphasized in such a brand-new MR (target−) condition than that in the MR (target+) and noMR conditions. Thus, the more stable COP and EMG performance in the MR (target−) condition may have benefited from a different allocation of cognitive resources during the task compared to the other conditions, and perhaps triggering more automated postural control. The lower movement velocity and muscle activity possibly benefitted from more attention resource shifting from posture control to the reaching task when interacting with the virtual targets which have elimination feedback in the MR (target−) condition. Of interest, we found that even with the same MR condition (target− and target+), the body sway and lower limb muscle activity were significantly smaller when the numeric targets were eliminated after touching.

Apart from attention, it is suggested that the feedback attached to the target may also help improving stability performance. The participants in our study could receive tactile and visual feedback from the touch screen in the noMR condition, and even such a light touch has been shown to reduce postural sway (Chen and Tsai 2015). Notably, a MR (target−) condition that without the actual touch but a virtual touch induced better postural control than noMR condition in the present study, suggesting the overwhelming effect of MR compared to light touch.

The visual feedback involved in the MR condition might represents not only a sole feedback but also an interaction between the virtual and real objects in the MR (target−) condition, which is thought to be an important characteristic and advantage of MR. Of interest, the postural sway characterized by COP velocity and area in the MR (target+) condition was larger than MR (target−) condition and was comparable to the noMR condition. It appears that the way of interaction with the virtual targets but not the visual information per se plays a crucial role for the improvement in standing balance with MR. Speculatively, the decrease in the number of virtual targets as the task progressed might lead to attention to the motor strategies for the following reaches.

4.4 Effects of head and eye movements on COP

As in daily life, participants performing the reaching tasks in the present study could move their body (limbs, head, and eyes, etc.) in an unrestrained manner either with or without MR. The impacts of head and eye movement on the changes in COP are of interest to determine whether the decreased COP changes with MR (target−) condition are attributed to those movements. Despite the larger movements of the head and eyes in the head- and eye-tracking tasks, respectively, the change in COP was smaller than that during the reaching task with MR (Figs. 5 and 6). These results were congruent

with a previous study that showed a negative correlation between eye and postural control, i.e., the larger eye movements the more stable the participants (Bonnet and Szafarczyk 2017). In contrast, the inverse relationship observed between eye/head movements and COP suggests that the relatively minor head and eye movements accompanying larger COP changes may represent a coordinated and functional adaptation. Although we could not assess the head and eye movements in the conditions noMR and MR (target+) due to the technical difficulty, the results of the larger values of COP velocity and area led us to speculate that the head and eye movements may be smaller in these conditions than MR (target-) condition. Perhaps, the relatively larger movements of the eye or head in the MR (target-) condition than those in the noMR and MR (target+) conditions might be compensatory actions for the decreased COP, i.e., better postural control.

4.5 Limitations

Several potential limitations are involved in the present study. First, we have focused on the EMG activity from several muscles in the lower limb and found differences in the SOL muscle activity among conditions, while others such as the hip and trunk muscles were not assessed in the present study. Because it is known that the synergistic action of the hip and ankle muscles is required to maintain an upright standing position (Henry et al. 1998; Maylor et al. 2001), the contribution of other muscles apart from the muscle tested may be relevant for the present results. The assessment of kinematic performance by using three-dimensional motion analysis may also be useful to explore the whole-body movements during reaching tasks. Second, the participants wore the HoloLens during all tasks to satisfy the conditions, but the weight of the HMD was also a cause of physical instability (Ito et al. 2021; Imaizumi et al. 2020). Because the participants in our study were asked wear the HMD throughout the experiments, the load and the induced body sway during reaching here may be equivalent, but not equal to those in day-to-day life. Third, head and eye movements in conditions noMR and MR (target+) could not be assessed due to the technical difficulty with our experimental equipment as mentioned above. Considering that the stability of head and vision and how they are coupled with lower limb activity can explain the influence of proprioception on balance (Sozzi et al. 2019), future study is needed to explore the mechanism of the better functionality of postural control in MR (target-) condition. Finally, we utilized a relatively simple cognitive task combined with the reaching task in the present study. It has been demonstrated that different balance performance occurs with increasing cognitive task difficulty or balance difficulty (Barra et al. 2006), and hence

the postural control with MR can be further explored in the future by adding richer tasks with healthy and people who have disability on balance.

5 Conclusion

In summary, MR was found not to produce greater balance disturbances than in realistic environments. Instead, it seems that the upright standing posture is more stabilized in MR conditions through appropriate interactions between the individual's body and the virtual object. Our findings suggest that the use of MR can perform superimposed cognitive tasks with stable balance and could be applied as a desirable rehabilitation tool in the future.

Author contributions YZ and NL: Conception and design of research; YZ, AM, CO, and NL: Data acquisition; YZ and NL: Data analysis; YZ, AM, CO, KS, TS, and NL: Interpretation of results; YZ: Preparation of the first version of the manuscript; YZ, AM, CO, KS, TS, and NL: Editing of the manuscript; YZ, AM, CO, KS, TS, and NL: Approval of the final version of the manuscript.

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Data availability The datasets generated or analyzed during the current study are available from the corresponding author upon reasonable request.

Declarations

Competing interest The authors declare no competing interests.

Ethical approval The experimental procedures and protocols were approved by the Ethics Committee of Kyoto University Graduate School and Faculty of Medicine (Permit No. C1492) and were performed according to the Declaration of Helsinki.

Informed consent All participants provided written informed consent prior to the experiments.

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