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The influence of the aquatic environment on the control of postural sway

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ABSTRACT

Balance training in the aquatic environment is often used in rehabilitation practice to improve static and dynamic balance. Although aquatic therapy is widely used in clinical practice, we still lack evidence on how immersion in water actually impacts postural control. We examined how postural sway measured using centre of pressure and trunk acceleration parameters are influenced by the aquatic environment along with the effects of visual information. Our results suggest that the aquatic environment increases postural instability, measured by the centre of pressure parameters in the time-domain. The mean velocity and area were more significantly affected when individuals stood with eyes closed in the aquatic environment. In addition, a more forward posture was assumed in water with eyes closed in comparison to standing on land. In water, the low frequencies of sway were more dominant compared to standing on dry land. Trunk acceleration differed in water and dry land only for the larger upper trunk acceleration in mediolateral direction during standing in water. This finding shows that the study participants potentially resorted to using their upper trunk to compensate for postural instability in mediolateral direction. Only the lower trunk seemed to change acceleration pattern in anteroposterior and mediolateral directions when the eyes were closed, and it did so depending on the environment conditions. The increased postural instability and the change in postural control strategies that the aquatic environment offers may be a beneficial stimulus for improving balance control.

1. Introduction

Postural sway during quiet standing has been widely investigated to evaluate the postural stability in the young, elderly [1] and individuals with disabilities [2]. This is because the postural sway allows one to examine interplay of sensory information from visual, vestibular and somatosensory systems, and how they are integrated to generate corrective torques to maintain body equilibrium during quiet standing [3]. Various behavioral studies have been conducted to examine the influence of sensorial input on postural sway during quiet standing by modifying or perturbing one of the sensory modalities or mechanical constraints [4–6].

The aquatic environment has been widely used as a therapeutic modality to improve static and dynamic balance in various patient populations [7,8]. Immersion in water can be considered as a form of sensorial and mechanical perturbation that is applied to the person who is standing in water. In addition, closing eyes while standing in the aquatic environment could potentially lead to further instability as shown in previous studies in a different sensory perturbation scenario [9]. Understanding the underlining mechanisms of immersion in water on postural stability could enable us to develop targeted rehabilitative programs for aquatic environments. However, the influence of immersion on postural sway has been investigated only sporadically, even though it has been speculated for some time that training in the aquatic

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environment may improve postural stability during standing on dry land [8,10].

Two recent studies have reported that center of pressure (COP) parameters were larger in water in comparison to land [11,12]. However, these two studies performed single quiet standing trials for each condition of interest to calculate COP parameters. In Louder's study, the effect of vision was tested only in one trial with eyes open and one trial with eyes closed condition, and the difference observed was not significant. In addition, the order of experiments in two environments was not randomized among participants. To accurately calculate COP parameters and to investigate the effect of vision on postural sway, one needs to perform COP measurements during longer time periods (at least 90 s) and to carry out between 3 and 5 trials for every single experimental condition to obtain reliable COP parameters [13]. Therefore, the first purpose of this study was to investigate the influence of the aquatic environment on COP parameters during quiet standing, using a more robust experimental methodology, i.e. longer trial period (90 s) and with more repetitions for each experimental condition. In addition, the present study randomized the order of tests on land and in water among the participants.

In most therapeutic pools, the height of water is usually at the level of lumbar region. As a result, one can anticipate that the part of the body that is above the water line and the part of the body that is below the water line may have different dynamics. Therefore, the second objective of this study was to investigate the influence of the aquatic environment on acceleration of the trunk, exploring the contribution of the lower and upper trunk movements on postural sway in the aquatic environment. The lower trunk acceleration was previously used to evaluate postural sway in able-bodied subjects and people with Parkinson's disease [14,15]. However, the contribution of the upper trunk acceleration in relation to the lower trunk has been underexplored during postural sway and in particular during quiet standing in water.

Our hypotheses going into this study were the following. First, we hypothesized that the fluctuation of postural sway, measured by COP parameters and trunk acceleration, would be larger in water compared to standing on land. Second, we hypothesized that the ratio of upper trunk to lower trunk acceleration would be significantly higher when individuals were standing in water compared to standing on land, due to the influence of water resistance and body weight offloading on the lower part of the trunk. Third, we hypothesized that visual information (i.e., eyes open and closed conditions) could affect postural sway differently between standing in water and on dry land, since different sensory inputs on the immersed part of the body could increase the demand for visual input while standing in water.

2. Methods

2.1. Participants and location

Ten able-bodied volunteers (6 male and 4 female) without any known history of physical or mental impairments and any contraindication to immersion in thermal water were assessed (Table 1). Prior to enrolling into the study, participants reviewed and signed a written informed consent. Ethical approval was obtained by our institution.

Both tests in water and on dry land were performed at a therapy pool area in our clinical facility (Fig. 1). During tests in water, the level of immersion for all participants was around the umbilicus, and the water temperature was set at 34–35 °C.

2.2. Instrumentation

A waterproof force plate ORP-WP-1000 (AMTI, USA) was used to collect kinetic data, from which we obtained COP in anteroposterior (AP) and mediolateral (ML) directions. A 16-channels data acquisition system Powerlab16/35 (ADInstruments, USA) was used to collect the force plate signals with a sampling frequency of 1000 Hz. Two wireless body-worn inertial sensors (Physilog, GaitUp, Switzerland) sealed in waterproof bags were attached to the lower trunk region (L5/S1) and to the upper trunk region (head of sternum) using medical adhesives. Their 3D-acceleration signals were synchronously collected at a sampling frequency of 500 Hz.

A mechanical switch embedding a force sensitive resistor (FSR) sensor was used to synchronize inertial sensor signals with the force-plate signals. The exact same instrumentation was used in water and on dry land. We carefully controlled the aquatic environment by shutting down the water flow and monitoring the examiner and participant movement in water (Fig. 1).

2.3. Experimental procedure

Participants were requested to stand "as still as possible" with arms crossed in front of their chest and with a comfortable foot position. Feet contour and ankle line were marked with a water-resistant chalk (Fig. 1) on the force plate and kept exactly the same between the environments. A mild mechanical strike was applied on the FSR placed over the trunk inertial sensor to trigger the beginning of each trial. As such, the beginning of each trial was recorded by both FSR (connected to the data acquisition system) and inertial sensors (recording on an internal memory card). We waited approximately 10 s prior to start 100 s of data collection to avoid the potential influence of the mechanical strike on the sway

Table 1
Demographic and anthropometric measures of the participants.

Subject	Gender	Age (years)	Height (cm)	Body weight (N)	Apparent body weight (N)	%Offload
1	F	21	175	625.0	270.0	56.8
2	M	19	173	625.9	309.1	50.6
3	M	18	175	727.2	357.7	50.8
4	F	23	171	627.3	274.1	56.3
5	M	20	179	737.5	362.1	50.9
6	M	24	173	610.4	307.3	49.7
7	M	23	175	794.2	418.3	47.3
8	F	29	165	438.8	182.2	58.5
9	F	21	168	720.9	266.4	63.1
10	M	35	178	753.7	463.2	38.5
Mean	*6 M/4F*	*23*	*173*	*666.1*	*321.0*	*52.2*
SD		*5*	*4*	*103.0*	*81.7*	*6.8*

Note: %offload indicates the percentage of body weight offloading in water calculated as %offload = (BWland – BWwater/BWland) * 100, where BWwater and BWland indicate averages of the vertical force during quiet standing in water and on land, respectively.

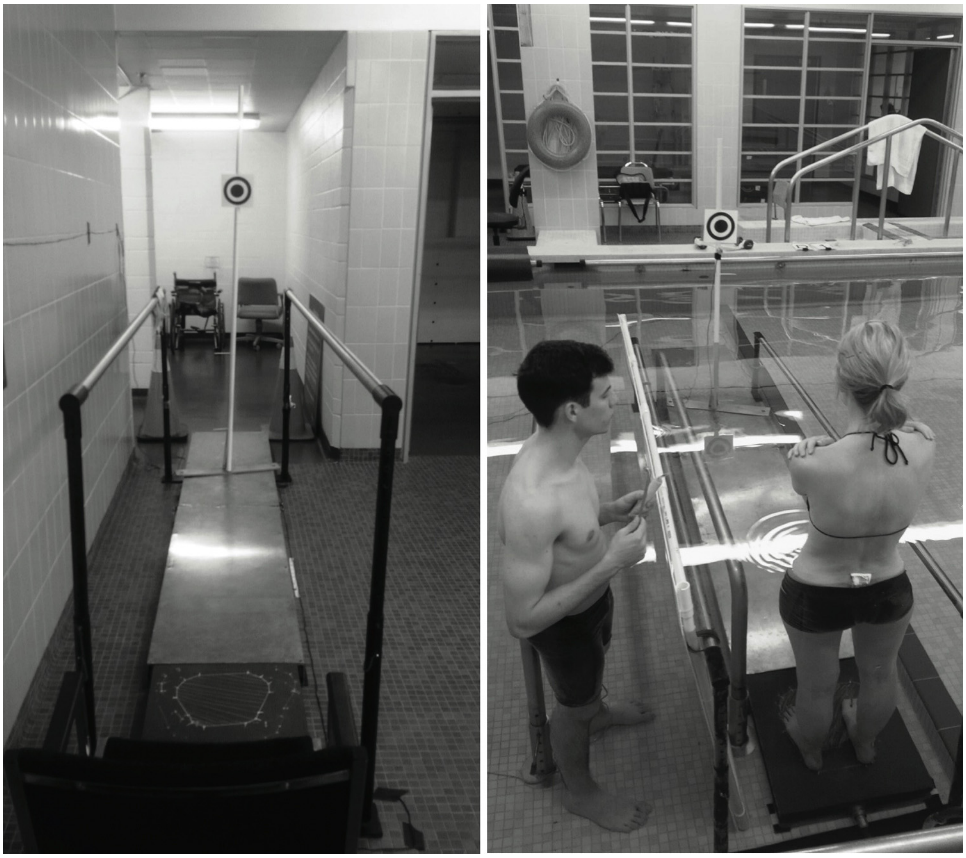

Fig. 1. Experimental set on land and in water. In water, participant stands on a waterproof force plate ORP-WP-1000 (AMTI, USA) measuring 0.50 × 0.46 × 0.08 m. When standing with eyes open, the participant looked at the target placed at 2-m distance. An aluminum custom-made pathway measuring 2.80 × 0.51 × 0.08 m was mounted in level with the waterproof force plate to allow participants to walk 4 steps forward during gait initiation protocol, the second part of our study.

of participants. Visual information was assigned randomly to 10 trials, five with eyes open (EO) and five with eyes closed (EC). In the case of EO condition, participants were asked to gaze at a fixed target positioned at the eye level at 2 m distance from the participant. Participants were allowed to rest between trials as needed to avoid fatigue. Tests on land and in water were performed in two consecutive days, whose order was randomized.

2.4. Data analysis

The percentage of body weight (BW) offloading (%offload) in water was calculated as %offload = 100 × (BW_{land} − BW_{water}/BW_{land}), where BW_{land} and BW_{water} indicated averages of the vertical force measured by force-plate in water and on land, respectively.

The COP signals were downsampled to 500 Hz and filtered using a 4th order zero-lag Butterworth digital filter, with low-pass cut-off frequency of 5 Hz [1,16]. From 100 s data, we eliminated the initial and final 5 s of the trials to avoid transient effects, and we analyzed 90 s data. The 3D-acceleration signals were used to calculate the acceleration vector. The acceleration signals were band-pass filtered with a 4th-order zero-lag Butterworth digital filter with band-pass frequencies between 0.15 and 5 Hz. The accelerometer data were corrected for horizontal tilt [17].

The mean COP position was measured as the distance between the overall anterior COP location in relation to the ankle line during 90 s quiet standing. Time- and frequency-domain parameters of the COP in AP and ML directions were analyzed. The parameters in time-domain were the following: root mean square (RMS_{COP}), mean velocity (MVELO), and hybrid-sway-area (AREA) [1]. In frequency-domain, we analyzed the frequency at 50% and 95% of the total (F50 and F95), centroidal frequency (CF), and frequency dispersion (FD). For the acceleration signal, RMS (RMS_{ACC}) and a measure of the first derivative of the acceleration (JERK) were calculated for upper and lower trunk accelerations in AP and ML directions [14]. In addition, we calculated the ratio of upper trunk to lower trunk for each acceleration parameter (i.e., RMS_{ACC} and JERK) in AP and ML directions. Within the 200 trials collected in the experiment, we eliminated three trials from the analyses due to technical issues.

2.5. Statistical analysis

Means and standard deviations were used to describe the dependent variables. A two-way repeated measure analysis of variance was conducted to detect the effect of two factors (i.e., environment and visual condition) for each parameter. In the presence of interaction between the two factors, a post-hoc Bonferroni test was performed. To explore the level of relationship of the %offloading with the anthropometric parameters, the Pearson's correlation coefficient was used. Analyses were conducted using the statistical package SPSS Statistics 23 (IBM, USA).

3. Results

3.1. Percentage of body weight offloading

Percentage of offloading in water varied from 38.5% to 63.1% and was found to inversely correlate with the height of the participants (r = −0.730, p = 0.017) (Table 1). There was no correlation between body weight and %offload (r = −0.470, p = 0.170).

3.2. COP parameters

The mean COP position was more anterior to the ankle line in water (F1,9 = 194.771, $\eta_p^2 = 0.956$, $p < 0.0001$) and with EC (F1,9 = 63.726, $\eta_p^2 = 0.876$, $p < 0.0001$) (Table 2). There was a significant interaction between environment and vision for COP position (F1,9 = 39.823, $\eta_p^2 = 0.816$, $p < 0.0001$). COP was significantly more anterior when individuals had EC in water than when they had it on land. The main effects of environment were significant for all COP parameters in time-domain ($p < 0.0001$ for all), indicating that all mean values were significantly larger in water than on land (Table 2). The main effects of visual condition were also significant for all parameters of COP in time-domain (from $p < 0.0001$ to $p = 0.012$), indicating that all mean values were significantly larger with EC than with EO. There were significant interactions between environment and visual condition for MVELO in AP direction (F1,9 = 6.121, $\eta_p^2 = 0.405$, $p = 0.035$) and for AREA (F1,9 = 8.830, $\eta_p^2 = 0.495$, $p = 0.016$), revealing that EC (compared to EO condition) increased the MVELO in the AP direction and AREA scores significantly more in water than it did on land.

Regarding the frequency-domain parameters, there was no significant interaction between environment and visual condition (Table 2). The main effect of environment was significant for F50 in ML direction (F1,9 = 6.421, $\eta_p^2 = 0.416$, $p = 0.032$), indicating F50 was larger in water than on land. The main effect of environment was significant for F95 (F1,9 = 19.818, $\eta_p^2 = 0.688$, $p = 0.002$), CF (F1,9 = 13.715, $\eta_p^2 = 0.604$, $p = 0.005$) and FD (F1,9 = 36.792, $\eta_p^2 = 0.803$, $p < 0.0001$) in AP direction, while they were smaller in water than on land. The main effect for visual condition was significant only for FD in ML direction (F1,9 = 9.613, $\eta_p^2 = 0.516$, $p = 0.013$), which showed that FD was smaller for EC condition compared to EO condition.

3.3. Acceleration parameters

The main effect of visual condition was significant for RMS_{ACC} and JERK indicating that they were significantly larger with EC than with EO, except for JERK of upper trunk in ML direction (F1,9 = 1.255, $\eta_p^2 = 0.122$, $p = 0.292$) (Table 3). The main effect of environment was significant for only RMS_{ACC} of the upper trunk in ML direction (F1,9 = 10.045, $\eta_p^2 = 0.527$, $p = 0.011$) and for the (upper trunk)/(lower trunk) ratio in ML direction (F1,9 = 16.298, $\eta_p^2 = 0.644$, $p = 0.003$), indicating that RMS_{ACC} was larger in water than on land. In the lower trunk, there was a significant interaction between environment and visual condition for RMS_{ACC} in AP direction (F1,9 = 7.003, $\eta_p^2 = 0.438$, $p = 0.027$) and RMS_{ACC} in ML direction (F1,9 = 8.265, $\eta_p^2 = 0.479$, $p = 0.018$), revealing that the effect of vision on these variables was influenced by the environment. RMS_{ACC} in AP direction in the lower trunk was significantly higher with EC compared to EO only in the case of land environment ($p = 0.006$). RMS_{ACC} in ML direction was significantly higher with EC compared to EO condition only in the aquatic environment ($p = 0.031$).

4. Discussion

We demonstrated that the aquatic environment increased COP fluctuation in the time-domain regardless of visual condition. Eye

Table 2
Center of pressure (COP) parameters during 90 s of quiet standing.

COP Parameters	Environment				*P* values		
	Land		Water		Two-way repeated ANOVA		
	Vision				Environment	Vision	Interaction
	EO	EC	EO	EC	Land vs. Water	EO vs. EC	
COP distance from ankle line [cm]	5.49 ± 1.25	5.99 ± 1.04	6.75 ± 1.01	7.68 ± 0.69	***0.000***	***0.000***	***0.000***
Time-domain measures							
RMS_{COP} AP [cm]	4.294 ± 1.159	5.409 ± 0.900	7.409 ± 2.385	9.337 ± 2.934	***0.000***	***0.003***	0.207
RMS_{COP} ML [cm]	2.432 ± 0.649	2.681 ± 0.879	3.668 ± 1.530	4.293 ± 1.311	***0.000***	***0.012***	0.334
MVELO AP [cm/s]	5.326 ± 1.172	7.462 ± 1.660	7.637 ± 1.577	11.003 ± 2.160	***0.000***	***0.000***	***0.035***
MVELO ML [cm/s]	2.997 ± 0.766	3.512 ± 0.869	4.856 ± 1.002	5.890 ± 1.577	***0.000***	***0.003***	0.075
AREA [cm^2]	8.330 ± 2.519	12.896 ± 4.239	21.626 ± 8.809	34.365 ± 15.533	***0.000***	***0.001***	***0.016***
Frequency-domain measures							
F50 AP [Hz]	0.111 ± 0.041	0.140 ± 0.063	0.137 ± 0.056	0.153 ± 0.070	0.070	0.159	0.419
F50 ML [Hz]	0.132 ± 0.042	0.166 ± 0.040	0.201 ± 0.096	0.214 ± 0.076	***0.032***	0.069	0.384
F95 AP [Hz]	0.724 ± 0.240	0.789 ± 0.210	0.518 ± 0.188	0.593 ± 0.218	***0.002***	0.055	0.844
F95 ML [Hz]	0.692 ± 0.191	0.756 ± 0.191	0.685 ± 0.147	0.666 ± 0.167	0.406	0.421	0.216
CF AP [Hz]	0.359 ± 0.086	0.385 ± 0.091	0.295 ± 0.069	0.327 ± 0.089	***0.005***	0.061	0.839
CF ML [Hz]	0.370 ± 0.081	0.401 ± 0.084	0.401 ± 0.092	0.386 ± 0.079	0.728	0.452	0.129
FD AP	0.794 ± 0.038	0.782 ± 0.046	0.748 ± 0.036	0.747 ± 0.036	***0.000***	0.638	0.240
FD ML	0.771 ± 0.042	0.748 ± 0.031	0.746 ± 0.049	0.722 ± 0.044	0.151	***0.013***	0.954

Note: Values are expressed as mean ± 1SD. *P*-values are presented for two-way repeated ANOVA. P-values are bolded in case of significance ($P < 0.05$). AP: anteroposterior. ML: mediolateral. EO: eyes open. EC: eyes closed. RMS_{COP}: root mean square of COP fluctuation. MVELO: mean velocity of COP. AREA: hybrid area of COP trajectory. F50 and F95: frequency below which contains 50% and 95% of the spectrum, respectively. CF: centroidal frequency. FD: frequency dispersion.

Table 3
Parameters of trunk acceleration during 90 s of quiet standing.

ACC Parameters	Environment				*P* values		
	Land		Water		Two-way repeated ANOVA		
	Vision				Environment	Vision	Interaction
	EO	EC	EO	EC	Land vs. Water	EO vs. EC	
Upper Trunk							
RMS_{ACC} AP [m/s^2]	0.049 ± 0.014	0.055 ± 0.011	0.049 ± 0.010	0.054 ± 0.010	0.844	***0.001***	0.973
RMS_{ACC} ML [m/s^2]	0.031 ± 0.007	0.033 ± 0.006	0.033 ± 0.006	0.036 ± 0.007	***0.011***	***0.016***	0.437
JERK AP [m/s^3]	4.624 ± 0.643	5.173 ± 0.791	4.751 ± 0.750	5.169 ± 0.669	0.729	***0.000***	0.485
JERK ML [m/s^3]	4.252 ± 1.352	4.282 ± 1.304	3.950 ± 1.031	4.120 ± 0.979	0.227	0.292	0.304
Lower Trunk							
RMS_{ACC} AP [m/s^2]	0.030 ± 0.007	0.037 ± 0.008	0.032 ± 0.009	0.036 ± 0.009	0.852	***0.002***	***0.027***
RMS_{ACC} ML [m/s^2]	0.021 ± 0.003	0.022 ± 0.004	0.020 ± 0.002	0.022 ± 0.003	0.948	***0.012***	***0.018***
JERK AP [m/s^3]	3.407 ± 0.661	3.998 ± 0.967	3.695 ± 0.829	4.022 ± 0.784	0.466	***0.000***	0.153
JERK ML [m/s^3]	5.195 ± 1.901	5.356 ± 1.992	4.048 ± 1.163	4.258 ± 1.145	0.153	***0.022***	0.532
Ratio of Upper Trunk to Lower Trunk							
RMS_{ACC} AP [m/s^2]	1.644 ± 0.371	1.477 ± 0.233	1.614 ± 0.482	1.580 ± 0.469	0.807	0.165	0.082
RMS_{ACC} ML [m/s^2]	1.492 ± 0.212	1.546 ± 0.166	1.632 ± 0.221	1.624 ± 0.175	***0.003***	0.440	0.249
JERK AP [m/s^3]	1.389 ± 0.221	1.351 ± 0.277	1.335 ± 0.295	1.331 ± 0.285	0.676	0.326	0.528
JERK ML [m/s^3]	0.847 ± 0.170	0.832 ± 0.172	1.087 ± 0.581	1.049 ± 0.464	0.223	0.367	0.638

Note: Values are expressed as mean ± 1SD. P-values are presented for two-way repeated ANOVA. P-values are bolded in case of significance ($P < 0.05$). AP: anteroposterior. ML: mediolateral. EO: eyes open. EC: eyes closed. RMS_{ACC}: root mean square of acceleration. JERK: a measure of time derivative of acceleration.

closure increased only MVELO in AP direction and AREA in water compared to on land. There was a predominance of low-frequency components in water environment compared to land. The trunk acceleration during postural sway increased predominantly with EC compared to EO, with exception for upper trunk acceleration in ML direction which increased in water compared to land.

4.1. COP fluctuation in water and on land

All time-domain measures of COP (RMS, MVELO and AREA) were larger in water than on land, which agrees with previous studies analyzing COP parameters in water [11,12] and in a micro-gravity environment [18]. COP amplitude measures have been related to the stability achieved by the postural control system, while COP velocity measures have been related to the amount of regulatory activity associated with this level of stability [19]. Therefore, our results indicate that the standing posture is less stable in water than on land, and that the central nervous system regulates the ankle joints more frequently to compensate for the instability.

One of the mechanisms to explain the instability in the aquatic environment may be the weightless condition caused by the buoyancy force. Gravity is always pulling the body forward during standing, as the COM is in front of the ankle joint requiring continuous exertion of plantarflexion torque to maintain stability. As the gravity toppling torque is reduced in water due to buoyancy, the required plantarflexion torque and the corresponding muscle activation of plantarflexors are smaller in water [20]. This automatic response may be via unloading on somatosensory system [21,22]. Nakazawa et al. suggested that unloading on somatosensory system may result in the enhance of H-reflex in water [23]. A combination of reduced muscle activity and enhanced stretch reflex may be some of the causes for postural instability in water.

Our findings indicated that F95, CF and FD in AP direction were significantly smaller in water than on land. These findings suggest a predominance of lower frequencies of COP variations in water in AP direction. An evaluation of postural sway in a micro-gravity environment showed that there was a predominance of low frequencies in unloading conditions and high frequencies in an overloading condition [18]. An inverted pendulum model may account for the phenomenon of predominance of low frequencies in a micro-gravity environment. The "eigen-frequency" of the body is $\sqrt{g/l}$, where g is the gravity and l is the distance from the COM to the ankle joint [24]. Since the effective gravity is smaller in water due to buoyancy, it leads to smaller eigen-frequency of the body resulting in the predominance of low frequency of postural sway.

4.2. Acceleration fluctuation in water and on land

In the present study, we chose to investigate the difference between the upper trunk acceleration signal (at the head of the sternum) and the lower trunk acceleration signal (at L5/S1) to evaluate if different patterns of upper and lower trunk accelerations would be present during partial immersion in water. We found no difference regarding trunk acceleration parameters between the environments, except for RMS_{ACC} in ML direction for upper trunk acceleration that was larger in water than on land. Because of this sole increment of RMS_{ACC} for upper trunk without any change for lower trunk acceleration, the (upper trunk)/(lower trunk) ratio was different between the environments in ML direction. These results suggest that, in water, upper trunk action is larger probably to compensate for postural instability in ML direction.

4.3. Interaction between environment and vision

We hypothesized that visual information could affect postural sway differently between water and land conditions, because sensory inputs following submersion in water may alter the perception of postural balance, and the demand for visual input could provide more reliable perception of balance condition. We observed that closure of the eyes had a more significant effect on the COP's MVELO in AP direction and AREA, as both variables exhibited increases with EC when the participants were standing in water as compared to standing on land. In addition, a simple strategy was assumed to compensate for postural instability in water with EC, which was to lean the body more forward. The increased COP's MVELO and AREA along with more anterior COP position with EC in water may suggest the decreased proprioceptive regulation exerted by the tibialis anterior when individuals are immersed in water [25]. In addition, the possible decreased sensory feedback from the foot sole in water may have jointly contributed to an increased instability [26].

Our findings demonstrated that upper and lower trunk RMS_{ACC} and JERK were affected predominantly by visual condition, being larger with EC condition compared to EO condition, although significant interactions was found between environment and visual condition in the lower trunk. For instance, RMS_{ACC} in ML direction was significantly higher with EC compared to EO only in the aquatic environment. This preliminary finding may suggest that with EC in water, a lower trunk strategy in ML direction may be required.

4.4. Limitations

Although controlling the aquatic environment to be as inert as possible, the unsteadiness of the water due to postural sway may have had some effect on the postural balance measurements and their repeatability. Yet, we did not have an instrument for measuring water oscillation while body swayed in water. Further studies analyzing the reliability of postural sway parameters in water are required.

5. Conclusions

We conclude that the aquatic environment increased postural instability: i.e., the center of pressure parameters increased in water regardless of the visual information; there was a predominance of low frequencies of sway in water compared to land; the larger upper trunk acceleration in mediolateral potentially suggests increased mediolateral trunk instability in water. The increased postural instability and the change in postural control strategies that the aquatic environment offers may be a beneficial stimulus for improving balance control. Further investigation is warranted.

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Conflict of interest

None.

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