Effect of walking speed during gait in water of healthy elderly



*Published Version:*
Fantozzi S., Cortesi M., Giovanardi A., Borra D., Di Michele R., Gatta G. (2020). Effect of walking speed during gait in water of healthy elderly. GAIT & POSTURE, 82, 6-13 [10.1016/j.gaitpost.2020.08.112].

*Availability:*
This version is available at: https://hdl.handle.net/11585/771015 since: 2020-09-09

*Published:*
DOI: http://doi.org/10.1016/j.gaitpost.2020.08.112

 

# Introduction

Gait adaptations in the elderly population were widely investigated with the purpose to early detect gait impairments, and thus to identify potential fall risks [1] and plan dedicated rehabilitative programs aiming to improve the quality of life in this population [2,3]. Observed gait alterations were associated to an overall decrease in muscle strength due to loss of motor neurons, and to reduced muscle fibers, muscle sensitivity, and aerobic capacity [4]. In comparison with young adults, healthy elderly show reduced range of motion of the hip and ankle joints, as well as reduced peak torques of the hip and knee joints, and reduced power at the knee and ankle joints [5–11]. Since decreased walking speed is typical of the elderly, the gait is normally analyzed at comfortable and increased speeds, to identify age-related vs speed-related changes [11]. This type of assessment is necessary, as walking speed affects spatio-temporal, kinematic and dynamic parameters of gait [12]. Kerrigan and coworkers showed that a reduction of peak hip extension, ankle plantar-flexion, and ankle power generation was evident at different walking speeds, due to both a subtle hip flexion contracture and the concentric weakness of ankle plantar flexors [13].

Aquatic exercises are frequently included in rehabilitation programs for the elderly, with the aim of improving or at least maintaining the gait function [14–16]. Thanks to the specific physical properties of water (density, hydrostatic pressure, buoyancy, viscosity, thermodynamics), aquatic exercise involves reduced weight bearing, augmented drag and increased central venous pressure [17]. A buoying environment, such as water, allows the elderly to explore a variety of movements that are important to enhance motor skills [18], postural control [14] and the ability to cross obstacles during gait [19]. Moreover, aquatic exercise is particularly beneficial for individuals suffering from musculoskeletal or/and neurological disorders [17]. Among other types of movement, walking is of primary interest in aquatic exercise [17]. Thus, understanding relevant biomechanical characteristics of gait in healthy individuals and, furthermore, comparing gait patterns in elderly to those of young adults, is of primary importance to effectively design a specific water-based rehabilitation program.

Previous authors investigated kinematics of gait in water by means of waterproofed video cameras [20–25]. Two studies showed different individual gait adaptations in elderly compared to young adults [20,22]. However, those studies were: i) limited to only one or two consecutive steps, ii) performed at self-selected comfortable walking speed, that is, without assessing the possible effects of walking speeds on gait kinematics, and iii) focused on the lower limbs joints without any assessment of the trunk and/or pelvis segments. Moreover, the set-up and post-processing procedures related to the analysis of video recordings are time-consuming, thus not allowing an immediate assessment of the execution of the task during aquatic exercise sessions. To overcome these limitations, wearable inertial magnetic sensors (IMMUs) have been extensively used in movement analysis. Technology based on IMMUs was recently exploited for gait [26–28], balance [29] and squat [30] assessment, demonstrating good feasibility for kinematic analysis also in water.

The present study investigated the 3D kinematics of healthy elderly during walking in water using IMMUs. The primary objective was to investigate the effect of walking speed on the gait spatio-temporal and joint kinematics characteristics. The secondary objectives were to compare the gait descriptors with respect to: i) land walking, and ii) young adults.

# Materials and Methods

## Participants

Nine healthy elderly participants were involved (4 males and 5 females, mean and standard deviation age: 73.5±5.8 years old, height: 167.0±8.2 cm, body mass 68.0±9.7 kg). All participants were free from known musculoskeletal, neurological, cardiac or pulmonary diseases. The participants freely gave written informed consent to participate. The study was approved by the Bioethics Committee of the University of Bologna. Data from 11 healthy young adults (6 males and 5 females, 27.0 ± 3.4 years old, 174.2 ± 8.2 cm height, 70.2 ± 11.8 kg body mass), available from a previous study, was used for comparison [28].

## Procedures/Experimental Protocol

Five wearable inertial magnetic sensors (Opal, APDM USA, 128Hz, gyroscope: ±2000°/s; accelerometer: ±6g, magnetometer: ±6Gauss) were calibrated, inserted into round waterproofed plastic boxes, and attached by means of elastic modular bands and bi-adhesive tape [28]. Only the right lower limb was instrumented, one sensor for each body segment was utilized: trunk (flat portion of the sternum), pelvis (aligned with the posterior superior iliac spines), thigh (laterally in the central third), shank (laterally above the malleolus), and foot (over the flat portion of the lateral part of the metatarsal area) [28].

The elderly performed 3 walking barefoot trials (10m) at a self-selected comfortable speed in a water condition (WW) (1.2-m depth water at a temperature of 28°C) and in a land condition (LW). Before WW data acquisition, the participants executed a walking trial for acclimatization purposes. Furthermore, with the goal to assess the influence of walking speed on gait parameters using a wider range of speeds, the participants were asked to perform two additional 10-m walking trials at a deliberately slow walking speed in LW and WW. In the WW trials, participants received no instructions regarding upper-limb movements similarly of protocols adopted during aquatic rehabilitation programs verifying that they were not used for propulsive purposes. The self-selected walking speed in slow trials resulted 18% and 23% lower than in the comfortable speed trials, for LW and WW, respectively.

A previously validated protocol - Outwalk [31,32], already used with a sample of young adults for the analysis of walking in water [28], was used. Following the Outwalk protocol, the body was modelled as an open kinematic chain constituted by 5 rigid segments (trunk, pelvis, right thigh, right shank, right foot) connected by ball-and-socket (trunk-pelvis, hip, ankle) and ‘loose’ double-hinge (knee) joints. The sagittal and frontal planes of the right side were considered and analyzed. The static posture was considered as the zero-offset-condition so joint angles in the dynamic trials were estimated with respect to that alignment condition of the body segments. Additional details regarding the protocol and the fixing procedure are reported in previous studies [28,31].

# Data Analysis

Joint angles were estimated decomposing joint matrices computed between adjacent segments, filtered (3Hz 2nd order Butterworth low-pass filter), segmented using an automatic algorithm based on wavelet analysis [33], and time-normalized into gait cycles. In order to remove outliers in the segmentation algorithm, only gait cycles with Intraclass Correlation Coefficient (ICC) values of the medio-lateral angular velocity of the shank with respect to the mean pattern of young adults greater than 0.6 were selected. Subsequently, in order to have consistent value for each participant, only gait cycles with ICC values of hip, knee, and ankle flexion-extension angles simultaneously greater than 0.75 [28] were selected for analysis. Finally, to have the same number of gait cycles for each participant and environment, a deletion procedure was applied selecting the gait cycles belonging to the central part of the 10-m walking trial, and excluding the gait cycles before and after the turns. Thirty gait cycles per participant were selected, resulting in a total number of 270 gait cycles of each joint. For the comparison between environments, 15 gait cycles at self-selected walking speed and 15 gait cycles at slow speed were used, while for the comparison with young adults, 30 gait cycles at self-selected speed were used.

For each gait cycle, the following spatio-temporal and joint kinematics variables were estimated: stride, stance and swing times [s], stance and swing percentages [%], stride length [cm], normalized stride distance (stride distance/leg length) [34], number of steps, and walking speed [cm/s] [33]. Furthermore, for each joint angle were calculated the Range Of Motion (ROM) [deg], minimum [deg], maximum [deg], percentages of gait cycle at minimum and maximum [%], and values of the angle at foot strike and at foot off [deg] points. Reference bands for joint angles were represented with mean and standard deviations.

To identify the effects of environment (LW/WW) and of age (elderly/young), by taking into account the effect of average speed of each assessed gait cycle, two linear mixed models were used. We used this statistical approach because linear mixed models allows to optimally model the within- (cycle-to-cycle) and between-subject variability of examined variables within the present data set.

The first model, shown in equation 1, was applied using LW and WW data from the elderly sample, whereas the second model, shown in equation 2, was applied using WW data at the self-selected comfortable speed from elderly and young adults:

$$y_{ij} = u_{oj} + \beta_0 + \beta_1 \cdot x_{1ij} + \beta_2 \cdot x_{2ij} + \beta_3 \cdot (x_{1ij} \cdot x_{2ij}) + \varepsilon_{0ij} \quad (1)$$

$$y_{ij} = u_{oj} + \beta_0 + \beta_1 \cdot x_{1ij} + \beta_2 \cdot x_{2ij} + \beta_3 \cdot (x_{1ij} \cdot x_{2ij}) + \varepsilon_{0ij} \quad (2)$$

where $\beta_0$ represent the fixed intercepts, $u_{oj}$ represent the random effects associated to the $j^{th}$ subject, the subscripts ij indicate values of variables measured in the $i^{th}$ cycle clustered in the $j^{th}$ subject, $y$ represents the outcome variable, $x_1$ represents the environment in equation 1 (coded with a dummy equal to 1 or 0 for the WW and LW, respectively) or the age in equation 2 (coded with a dummy equal to 1 or 0 for the elderly and young adults conditions, respectively), $x_2$ represents the dimensionless walking speed, and $\varepsilon_0$ is the random error component.

In the models we used, the coefficients ($\beta_0$, $\beta_1$, $\beta_2$, $\beta_3$) are interpreted in the following way:

$\beta_0$ (fixed intercept): the intercept of the baseline category;

$\beta_1$: the difference in the intercept between the baseline category (coded as 0) and the other category (coded as 1);

$\beta_2$: the slope for the continuous predictor (walking speed) of the baseline category;

$\beta_3$ (interaction): the difference in the slope for the continuous predictor (walking speed) between the baseline category (coded as 0) and the other category (coded as 1).

In both linear mixed models, the dimensionless speed (walking speed/sqrt(g*leg_length)) was used as a predictor variable attempting to account for the influence of anthropometric differences [34]. Scatterplots showing the effects of dimensionless walking speed and environment or age group on the examined variables were visually inspected for data consistency

Data collection was performed using the Motion Studio Software (APDM, USA), while data processing was performed using Matlab® language (The Mathworks™, USA, version R2018a). Statistical analyses were performed using the R statistical software (version 3.5.2). When not otherwise indicated, data are presented as mean ± standard deviation.

# Results

Walking speed and normalized walking speed for elderly in both conditions and for young adults in WW are reported in Table 1.

*Table 1 here*

## *Water vs land conditions*

Kinematic parameters showing a significant effect of the WW vs LW in elderly are shown in Table 2, together with the corresponding estimated coefficients of the linear mixed models. A reduced walking speed was observed in WW (27.9±8.3 cm/s, 0.09±0.03 dimensionless value) with respect to LW (69.3±10.7 cm/s, 0.24±0.04 dimensionless value). Considering the spatio-temporal parameters, a smaller stride distance with larger stride duration was observed in WW (Table 2).

*Table 2 and Figure 1 here*

Similar joint angles patterns, although with some distinctive differences, were found in the sagittal (Figure 1, left) and frontal (Figure 1, right) planes. Analyzing the hip joint in the sagittal plane, higher flexion values were observed at the foot off (approximately 15°), at the minimum value (approximately 11°) and at the maximum value (approximately 11°) in WW vs LW (Table 2, Figure 1). Different effects of walking speed were observed in the two environments (Table 2). Analyzing the knee joint, a larger flexion at foot strike (approximately 28°) was observed in WW vs LW. This difference can be also explained by the different effects of walking speed in the two conditions (Table 2, Figure 2). Indeed, one-unit increase of dimensionless walking speed in LW, induced a slight (non-significant) decrease of knee flexion-extension at foot strike, and an increase of 111° of knee flexion-extension in WW.

*Figure 2 here*

In the frontal plane, the trunk-pelvis and the hip joints showed negligible differences between conditions. From the pattern of reference bands (Figure1), an indistinct pattern of mean values for WW in the trunk-pelvis joint can be observed. Analyzing the mean value over the stance phase of the ankle inversion-eversion, an approximately 9° larger value was observed in WW. For each unit increase of dimensionless walking speed, a decrease of 23° and increase of 6° of mean value during stance of ankle inversion-eversion were estimated in LW and WW, respectively. The different effect of walking speed in the two environments can thus partially explain the above results (Table 2).

### *Elderly vs young adults in WW*

Lower walking speed (29.5±7.9 cm/s and 57.5±14.1 cm/s, for elderly and young adults, respectively) and stride distance were observed in elderly. Among all analyzed parameters, those showing a significant effect of age are shown in Table 3. In the sagittal plane, a slightly more flexed hip joint in elderly was observed, whereas the knee joint showed no differences between ages (Figure 3). In the pre-swing phase, a less plantar-flexed ankle joint was estimated in the elderly (approximately 9°). The different effect of walking speed in the two different ages can partially explain the differences between mean values in young adults vs elderly (Table 3, Figure 4). No differences were observed in the frontal plane, with the exception of a greater variation of trunk-pelvis angle in elderly.

*Table 3, Figure 3 and Figure 4 here*

## Discussion

In the present study, we assessed the gait kinematics in healthy elderly during walking in water as compared to walking on land and to the walking characteristics of young adults in water only. It is well known that the use of IMMUs allows a fast set-up and data analysis, making this technology potentially suitable for the routine practice during a session of aquatic exercise. It must be said that, this exploitation requires the use of algorithms and devices able to provide robust and reliable orientation of the sensor with respect to the global frame. Nevertheless, the most important advantage of using IMMUs instead of video analysis is the possibility of using a high number of gait cycles. Thanks to the non-restricted field of view, a closer analysis of the gait of a person can be performed. A novel approach of this study was to assess the influence of walking speed on gait parameters during walking in water. Linear mixed models showed how, for most of the gait parameters, not only the single predictors (speed, environment, and age), but also their interactions, significantly affect the gait kinematics.

Analyzing the gait spatio-temporal parameters of elderly in WW with respect to LW, the shorter stride distance and longer stride duration are compatible with the reduction of walking speed (Table 2) and congruent with what was found in previous studies using video analysis [20,22]. The percentage of duration of the stance phase was slightly increased when compared to LW, about 3%, contrary to what was shown in a previous study [20]. Comparing these parameters with those shown by young adults walking in water, the main difference was a 50% reduction of stride distance (Table 3), while only a 20% reduction was found by Barela et al. [20]. The differences with the study of Barela et al. [20] could be explained by the lower walking speed in LW and in WW shown by the present elderly. Nevertheless, this speed was consistent with previous studies assessing walking on land in 75-years old elderly [4]. Furthermore, although the gait was performed at the same water level used in the study of Barela (approximately at the xyphoid process) [20], the difference in the mean height between the two groups was slightly higher than in that study (7cm vs 5cm). The differences in the spatio-temporal parameters of walking on LW with respect to WW might be explained by the adaptations to the water environment and its specific physical properties such as drag, viscosity, and buoyancy. When the level of water immersion is increased, a person experiences modified sensorial information also, such as a lower plantar feed-back and a reduced vision of his/her steps. This causes a perception of instability and, together with the higher drag, could contribute to the reduction of walking speed, stride distance and to the increasing of stride duration in the elderly.

Analyzing the joint kinematics of elderly in WW with respect to LW, a more flexed hip in the pre-swing and swing phases, a more flexed knee during the foot strike and the terminal swing phase, and a less-plantar flexed ankle, were observed on the sagittal plane. Furthermore, a higher inversion of the ankle joint during stance was observed on the frontal plane (Figure 1, Table 2). These findings confirm observations of studies using video analysis [20,22]. Comparing the same parameters with respect to young adults walking in water, we observed a more flexed hip and a less plantar-flexed ankle (Figure 3, Table 3). These results confirm previous observations related to sagittal plane parameters [22] in an elderly population on average 10 years younger than that of the present study. Jabbar et al. [22] hypothesized a direct relation between the effect of the hydrodynamic force (drag) and the pattern observed in elderly and young adults on the sagittal plane at the hip joint. As no significant differences were found in the trunk-pelvis joint in the sagittal plane, it seems that to overcome the higher resistance in gait progress due to the higher density of water with respect to air, the participants would tend to incline forward the trunk-pelvis segment as if it was one rigid segment, probably resulting in a higher flexion of the hip. The modifications shown here for the hip and the ankle joints angles more enhanced in the elderly, are also consistent with previous results concerning gait analysis of elderly in LW, where the influence of walking speed was investigated [13], and with the hypothesized possible subtle hip flexion contracture and ankle plantar-flexor concentric weakness of the older population [13]. From this point of view, aquatic exercises could be used for strengthening the hip flexors in a safe environment and in a less loaded condition. However, a more comprehensive investigation regarding the muscle activation of the lower limbs and the inclination of the trunk against the water should be performed.

Linear mixed models underlined how gait parameters were affected by the environment, age and walking speed, and also by the interaction between these variables (Tables 2 and 3, Figure 2 and 4). An interaction between speed and environment indicates that speed changes have a different effect on kinematic variables in WW as compared to LW, while an interaction between speed and age indicates that speed changes have a different impact in elderly compared to young adults. The significant speed x environment interactions observed for the majority of examined variables (Table 2) can be explained considering the different properties of water as compared to air in terms of density/viscosity and thus in the hydrodynamic force. Instead, the significant age x speed interactions observed for most variables (Table 3), suggest that the different musculoskeletal conditions of elderly vs young adults can affect the way gait kinematics changes with varying speed in the two populations. The linear mixed modelling analysis allows a better interpretation of the effect of the examined variables. Indeed, when the interaction coefficient ($\beta_3$) is significant, there is a different effect of walking speed in the two conditions (water vs land or elderly vs adults) and the respective lines have different slopes (Figure 2 a,b,d and Figure 4 b,c,d). In the opposite case (that is, no significant interaction), the two lines are parallel (Figure 2c and Figure 4a). In a few cases, a large vertical dispersion was observed around the fitted lines (Figure 4d), highlighting the importance of visually assessing the examined effects by means of plots with raw data of single participants, besides examining the model coefficients. These findings highlight the importance of assessing walking speed during walking in water, as differences in gait parameters between environments or age groups can, at least partially, be a consequence of differences in walking speeds and on how speed interacts with the environment or age.

The fear of falling, connected with a reduced balance capacity, undermines seriously the quality of life in elderly. The control of body stability and postural position is governed not only by the interaction and the integration of sensorial inputs (visual, vestibular and proprioceptive) but also by the experience, neuromuscular strategies, specific demands of the motor task, environment and age [35]. In this context, performing aquatic exercise is particularly important for elderly because it allows the possibility of exploring a variety of movements in a safe environment. This condition can be achieved thanks to the physical characteristics of water and especially to the hydrodynamic and hydrostatic forces that sustain an exercising person. In this respect, the higher variability of the trunk-pelvis kinematics in the frontal plane shown by elderly with respect to young adults (Figure 3) highlights an environment where the elderly are forced to explore a variety of movements in a safer situation for the fear of falling with respect to dry-land, and thus can be seen as a guidance for motor skill acquisition. However, after having performed a dedicated validation of the measurement, a specific investigation of the orientation of the trunk should be accomplished to better understand the role of this body segment during progression in the water environment.

The difference in height between the two groups lead to a variability in the immersion level, going from the xiphoid process to the shoulder, and reasonably determining a difference in the offloading body weight of about 6%, although without variations in the drag (68.1±4.8N vs 68.4±5.8N, elderly vs young adults, respectively [36]). Nevertheless, as the water level can't be adjusted in most swimming pools, the issue of the effect of subject's height on in-water walking kinematics is important and warrants to be further investigated.

Despite the great advantages of the IMMU technology, on the other side, video analysis allows a higher accuracy when estimating joint kinematics and provides the position of anatomical landmarks. For this reason, if significant differences between conditions were lower than about 10 degrees, they were not taken into account and commented. For similar reasons, connected with the accuracy of kinematics reconstruction, knee ab-adduction was not considered reliable in the present analysis and thus no comparison with previous results [22] for this specific degree of freedom was performed.

In conclusion, the present study has analyzed the 3D gait kinematics parameters of lower limbs and trunk segment during walking in water of healthy elderly, taking into account for the first time the effects of the dimensionless walking speed. The results showed specific patterns for elderly population during aquatic exercise, highlighting the importance of future investigations employing age as a continuous variable to understand more in detail its interaction with walking speed and environment. The differences shown by elderly with respect to young adults about hip and ankle joint on the sagittal plane were influenced not only by speed and age, but also by the interaction between the two variables. For this reason, in particular during WW, assessing the walking speed is fundamental as gait parameters can be different due to speed not only as a consequence of the different environment.

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# Figures

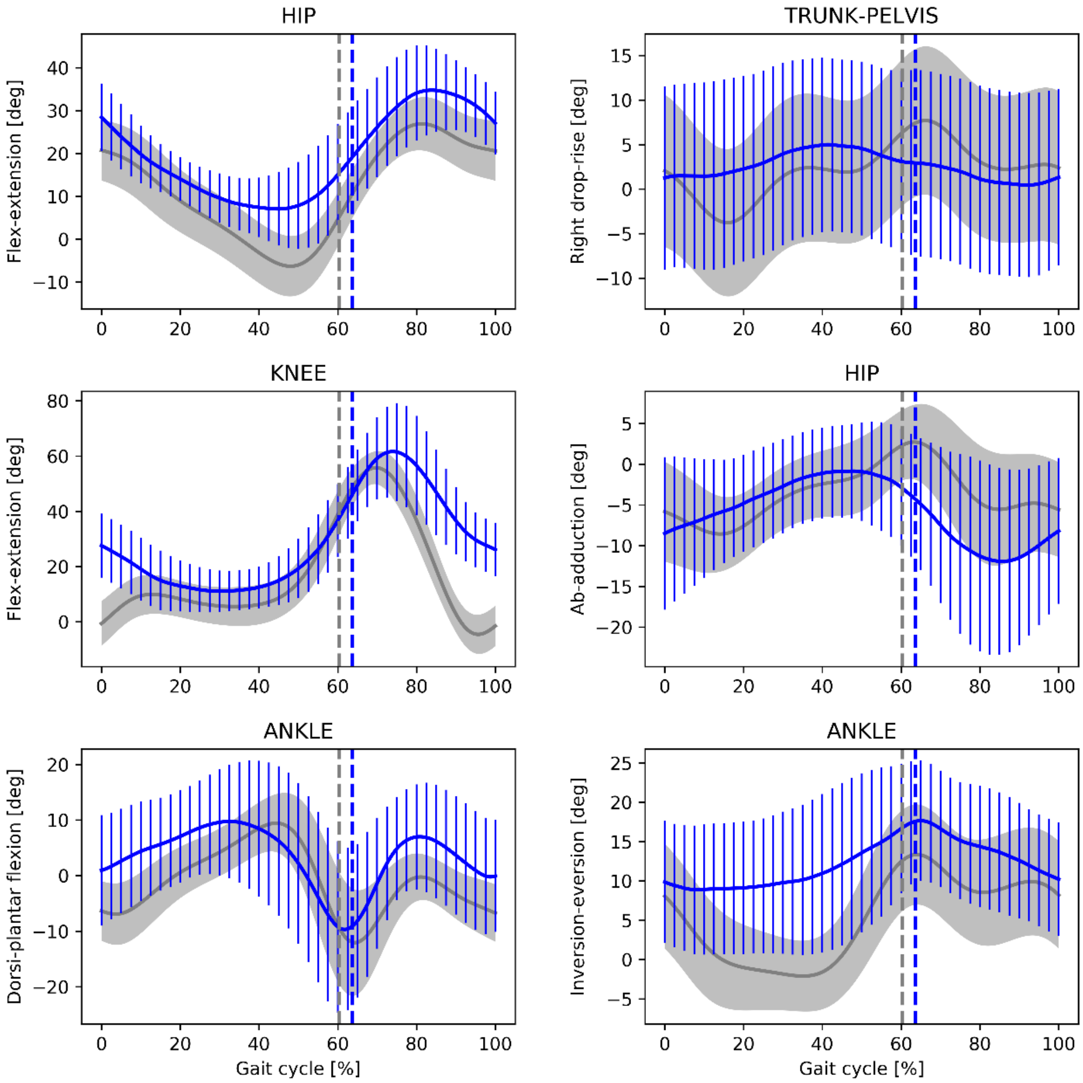


*Figure 1: Normative bands of joint kinematics (mean and standard deviations) in elderly in land (grey shaded) and water (blue striped) conditions on the sagittal (left) and frontal (right) planes. The vertical dashed lines correspond to the mean values of the "foot off" in land (black) and water (blue) conditions expressed in percentage of the gait cycle (0 and 100% correspond to "foot strike"). The reference angle (0°) for all joints was obtained in static posture, see main text.*

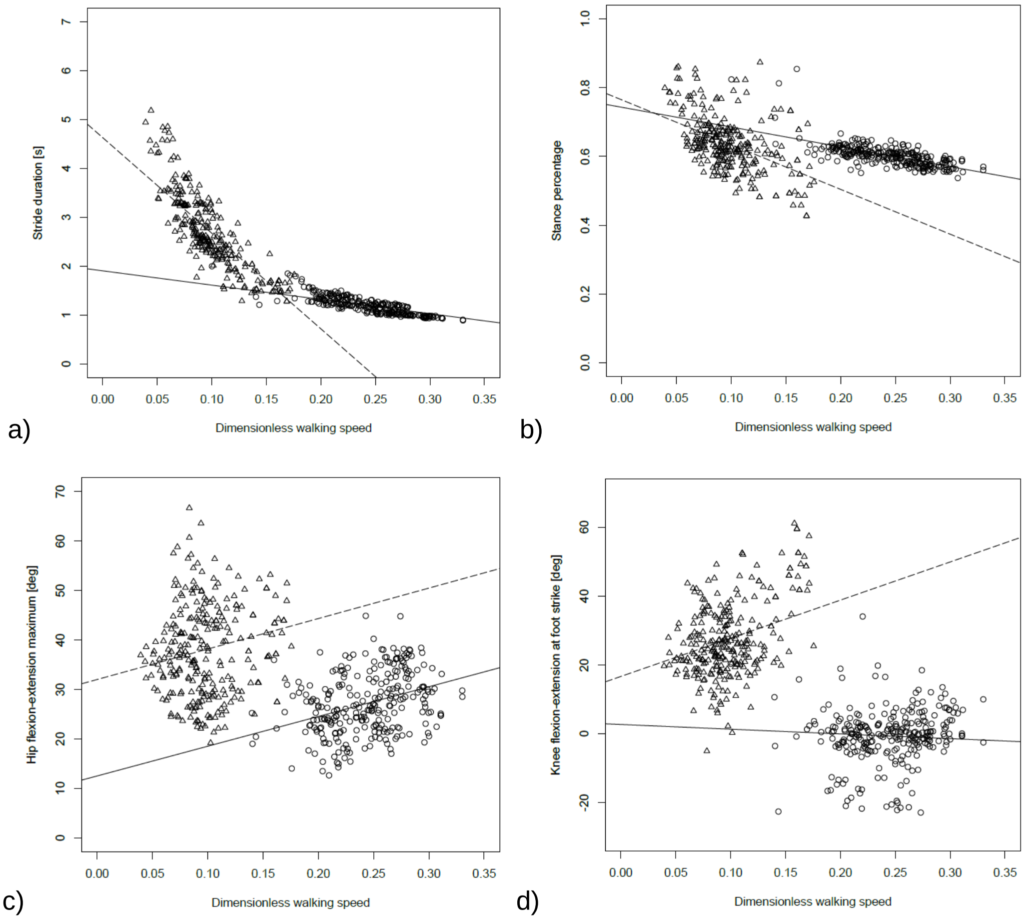


*Figure 2: Scatterplot showing the values of stride duration (a), of stance percentage (b), of maximum hip flexion-extension (c), and of knee flexion-extension at foot strike (d) as a function of dimensionless walking speed in the LW (circles) and WW (triangles) conditions. The lines represent the trends of the four parameters vs. dimensionless walking speed, as estimated with linear mixed models for the two conditions LW: solid line, WW: dashed line). The reference angle (0°) for all joints was obtained in static posture, see main text.*

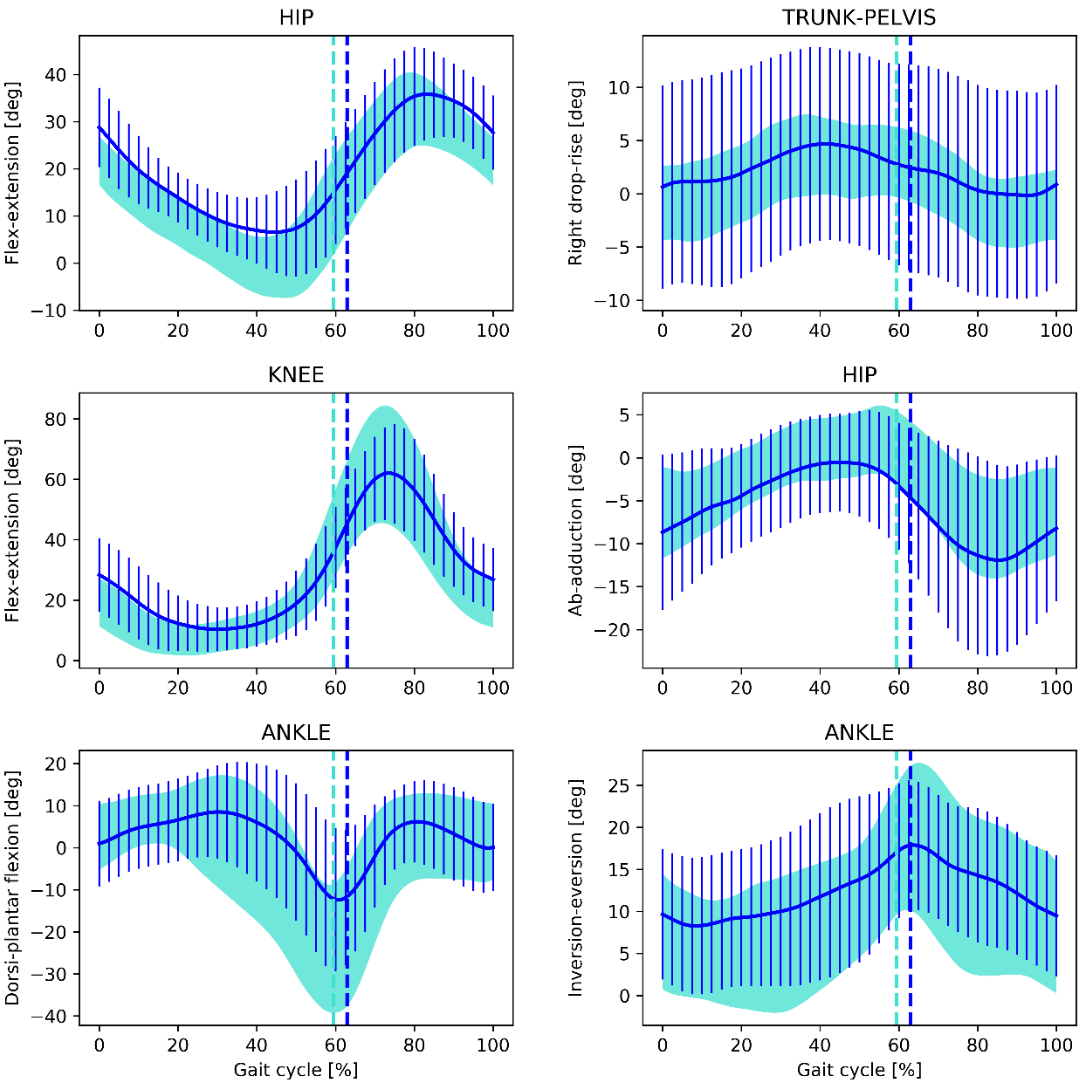


*Figure 3: Normative bands of joint angles (mean and standard deviation) of elderly (blue striped) and of young (light blue shaded) adults in the water condition. The vertical dashed lines correspond to the mean values of the "foot off" for elderly (blue) and young adults (light blue) conditions expressed in percentage of the gait cycle (0 and 100% correspond to "foot strike"). The reference angle (0º) for all joints was obtained in static posture, see main text.*

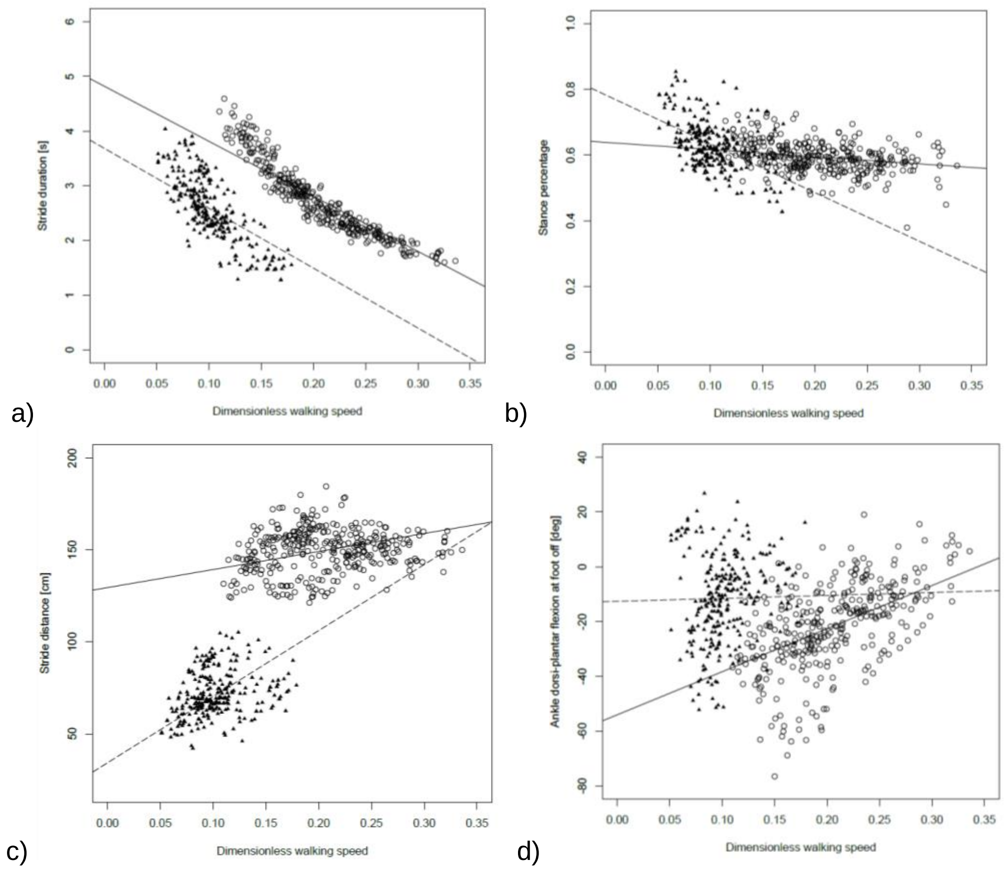


*Figure 4: Scatterplot showing the values of stride duration (a), of stance percentage (b), of stride distance (c), and of ankle dorsi-plantar flexion at foot off (d) as a function of dimensionless walking speed for young adults (circles) and elderly (full-triangles) conditions. The lines represent the trends of the four parameters vs. dimensionless walking speed, as estimated with linear mixed models for the two populations, young adults: solid line, elderly: dashed line). The reference angle (0°) for all joints was obtained in static posture, see main text.*

# Tables

**Table 1.** Mean and standard deviation values of the walking speed for elderly (in the two environments) and for young adults (in the in-water environment) (n=30 gait cycles for each participant and for each environment). For elderly, data are presented for all trials together (at self-selected and slow speeds) and for only trials at self-selected speeds.

| | | **Mean ± std.dev** | |
|---|---|---|---|
| | | Walking Speed [cm/s] | Normalized Walking Speed |
| **Elderly (all trials)** | Dry-land | 69.3±10.7 | 0.24±0.04 |
| | In-water | 27.9±8.3 | 0.09±0.03 |
| **Elderly (trials at self-selected speeds)** | Dry-land | 74.9±7.9 | 0.26±0.03 |
| | In-water | 29.5±7.9 | 0.10±0.03 |
| **Young Adults (trials at self-selected speeds)** | In-water | 57.5±14.1 | 0.21±0.05 |

**Table 2.** Mean and standard deviation values (n=30 gait cycles for each participant and for each environment, n=540 gait cycles overall) of the parameters that showed a significant effect of environment and/or of walking speed for the elderly sample. The estimated coefficients of the linear mixed models are reported ($\beta_0$ is the fixed intercept of the equation; $\beta_1$ multiplies the variable representing the environment; $\beta_2$ multiplies the normalized walking speed; $\beta_3$ multiplies the environment and the normalized walking speed). * $p<0.05$.

| | | **Mean ± std.dev** | | **Coefficients of Linear mixed models** | | | |
|---|---|---|---|---|---|---|---|
| | | **Dry-land** | **Under-water** | **$\beta_0$ Intercept** | **$\beta_1$ Environment** | **$\beta_2$ Normalized speed** | **$\beta_3$ Environment * Normalized speed** |
| **Spatio-temporal** | Stride duration [s] | 1.2±0.2 | 2.7±0.8 | 1.9* | 2.7* | -2.9* | -16.6* |
| | Stance percentage [%] | 60.3±3.5 | 63.7±8.2 | 0.7* | 0.0 | -0.6* | -0.7* |
| | Stride distance [cm] | 81.2±8.3 | 70.4±11.5 | 51.2* | -0.38 | 123.2* | 75.6* |
| | Normalized stride distance | 1.0±0.1 | 0.9±0.1 | 0.6* | 0.0 | 1.5* | 0.9* |
| **Hip** | Flexion-Extension at foot off [deg] | 5.3±5.8 | 19.9±9.4 | 19.7* | -6.4* | -59.0 | 127.1* |
| | Flexion-Extension maximum [deg] | 27.1±6.2 | 38.1±9.5 | 12.5* | 19.5* | 60.1* | 1.5 |
| | Flexion-Extension minimum [deg] | -6.6±7.1 | 4.3±6.9 | -3.6 | 0.2 | -12.2 | 90.5* |
| | Flexion-Extension mean value (swing)[deg] | 21.1±5.6 | 30.8±7.7 | 13.2* | 12.9* | 32.5* | 15.7 |
| **Knee** | Flexion-Extension at foot strike [deg] | -0.6±8.1 | 27.6±11.3 | 2.7 | 14.0* | -13.7 | 124.5* |
| | Flexion-Extension maximum [deg] | 56.7±5.9 | 64.1±16.9 | 42.6* | 26.4* | 58.2* | -107.9* |
| | Flexion-Extension minimum [deg] | -5.3±6.6 | 9.3±7.0 | -5.1 | 17.5* | -0.85 | -31.3* |

| | | | | | | | |
|---|---|---|---|---|---|---|---|
| | Flexion-Extension mean value [deg] | 17.7±5.5 | 29.0±7.4 | 10.9* | 15.4* | 27.9* | 0.0 |
| | Flexion-Extension mean value (swing)[deg] | 28.7±6.2 | 46.3±11.9 | 11.6* | 29.9* | 70.5* | -21.6 |
| **Ankle** | Inversion-Eversion mean value during stance [deg] | 1.9±4.8 | 11.2±7.4 | 7.6* | 3.0 | -23.4* | 29.8* |

**Table 3.** Mean and standard deviation values of the parameters that showed a significant effect of age and/or of dimensionless walking speed in water condition (n=30 gait cycles for each participant, considering 9 elderly and 11 young adults, n=600 gait cycles overall). The estimated coefficients of linear mixed models are reported ($\beta_0$ is the fixed intercept of the equation; $\beta_1$ multiplies the variable representing the age; $\beta_2$ multiplies the normalized walking speed; $\beta_3$ multiplies the age and the normalized walking speed). * $p<0.05$.

| | | **Mean ± std.dev** | | **Coefficients of Linear mixed models** | | | |
|---|---|---|---|---|---|---|---|
| | | **Young Adults** | **Elderly** | $\beta_0$ Intercept | $\beta_1$ Age | $\beta_2$ Normalized speed | $\beta_3$ Age * Normalized speed |
| **Spatio-temporal** | Stride duration [s] | 2.8±0.7 | 2.5±0.6 | 4.8* | -1.1* | -10.1* | -0.9 |
| | Stance percentage [%] | 59.4±4.8 | 62.9±7.7 | 0.6* | 0.1* | -0.2* | -1.3* |
| | Stride distance [cm] | 150.7±12.2 | 71.6±12.6 | 129.6* | -95.3* | 98.1* | 261.4* |
| | Normalized stride distance | 1.88±0.12 | 0.9±0.1 | 1.6* | -1.2* | 1.2* | 3.1* |
| **Hip** | Flexion-Extension at foot off [deg] | 12.6±8.9 | 20.2±8.9 | -9.2* | 37.3* | 106.1* | -183.1* |
| | Ab-Adduction minimum [deg] | -10.1±6.4 | -15.0±9.8 | -12.3* | -5.2 | 7.8 | 16.5 |
| **Ankle** | Dorsi-Plantar flexion at foot off [deg] | -21.3±15.5 | -11.5±15.7 | -54.0* | 41.4* | 157.3* | -146.5* |
| | Dorsi-Plantar flexion minimum [deg] | -27.0±15.5 | -18.9±14.8 | -65.5* | 32.2* | 186.6* | -47.6 |

# Introduction

Gait adaptations in the elderly population were widely investigated with the purpose to early detect gait impairments, and thus to identify potential fall risks [1] and plan dedicated rehabilitative programs aiming to improve the quality of life in this population [2,3]. Observed gait alterations were associated to an overall decrease in muscle strength due to loss of motor neurons, and to reduced muscle fibers, muscle sensitivity, and aerobic capacity [4]. In comparison with young adults, healthy elderly show reduced range of motion of the hip and ankle joints, as well as reduced peak torques of the hip and knee joints, and reduced power at the knee and ankle joints [5–11]. Since decreased walking speed is typical of the elderly, the gait is normally analyzed at comfortable and increased speeds, to identify age-related vs speed-related changes [11]. This type of assessment is necessary, as walking speed affects spatio-temporal, kinematic and dynamic parameters of gait [12]. Kerrigan and coworkers showed that a reduction of peak hip extension, ankle plantar-flexion, and ankle power generation was evident at different walking speeds, due to both a subtle hip flexion contracture and the concentric weakness of ankle plantar flexors [13].

Aquatic exercises are frequently included in rehabilitation programs for the elderly, with the aim of improving or at least maintaining the gait function [14–16]. Thanks to the specific physical properties of water (density, hydrostatic pressure, buoyancy, viscosity, thermodynamics), aquatic exercise involves  reduced weight bearing, augmented drag and increased central venous pressure [17]. A buoying environment, such as water, allows the elderly to explore a variety of movements that are important to enhance motor skills [18], postural control [14] and the ability to cross obstacles during gait [19]. Moreover, aquatic exercise is particularly beneficial for individuals suffering from musculoskeletal or/and neurological disorders [17]. Among other types of movement, walking is of primary interest in aquatic exercise [17]. Thus, understanding relevant biomechanical characteristics of gait in healthy individuals and, furthermore, comparing gait patterns in elderly to those of young adults, is of primary importance to effectively design a specific water-based rehabilitation program.

Previous authors investigated kinematics of gait in water by means of waterproofed video cameras [20–25]. Two studies showed different individual gait adaptations in elderly compared to young adults [20,22]. However, those studies were: i) limited to only one or two consecutive steps, ii) performed at self-selected comfortable walking speed, that is, without assessing the possible effects of walking speeds on gait kinematics, and iii) focused on the lower limbs joints without any assessment of the trunk and/or pelvis segments. Moreover, the set-up and post-processing procedures related to the analysis of video recordings are time-consuming, thus not allowing an immediate assessment of the execution of the task during aquatic exercise sessions. To overcome these limitations, wearable inertial magnetic sensors (IMMUs) have been extensively used in movement analysis. Technology based on IMMUs was recently exploited for gait [26–28], balance [29] and squat [30] assessment, demonstrating good feasibility for kinematic analysis also in water.

The present study investigated the 3D kinematics of healthy elderly during walking in water using IMMUs. The primary objective was to investigate the effect of walking speed on the gait spatio-temporal and joint kinematics characteristics. The secondary objectives were to compare the gait descriptors with respect to: i) land walking, and ii) young adults.

# Materials and Methods

## Participants

Nine healthy elderly participants were involved (4 males and 5 females, mean and standard deviation age: 73.5±5.8 years old, height: 167.0±8.2 cm, body mass 68.0±9.7 kg). All participants were free from known musculoskeletal, neurological, cardiac or pulmonary diseases. The participants freely gave written informed consent to participate. The study was approved by the Bioethics Committee of the University of Bologna. Data from 11 healthy young adults (6 males and 5 females, 27.0 ± 3.4 years old, 174.2 ± 8.2 cm height, 70.2 ± 11.8 kg body mass), available from a previous study, was used for comparison [28].

## Procedures/Experimental Protocol

Five wearable inertial magnetic sensors (Opal, APDM USA, 128Hz, gyroscope: ±2000°/s; accelerometer: ±6g, magnetometer: ±6Gauss) were calibrated, inserted into round waterproofed plastic boxes, and attached by means of elastic modular bands and bi-adhesive tape [28]. Only the right lower limb was instrumented, one sensor for each body segment was utilized: trunk (flat portion of the sternum), pelvis (aligned with the posterior superior iliac spines), thigh (laterally in the central third), shank (laterally above the malleolus), and foot (over the flat portion of the lateral part of the metatarsal area) [28].

The elderly performed 3 walking barefoot trials (10m) at a self-selected comfortable speed in a water condition (WW) (1.2-m depth water at a temperature of 28°C) and in a land condition (LW). Before WW data acquisition, the participants executed a walking trial for acclimatization purposes. Furthermore, with the goal to assess the influence of walking speed on gait parameters using a wider range of speeds, the participants were asked to perform two additional 10-m walking trials at a deliberately slow walking speed in LW and WW. In the WW trials, participants received no instructions regarding upper-limb movements similarly of protocols adopted during aquatic rehabilitation programs verifying that they were not used for propulsive purposes. The self-selected walking speed in slow trials resulted 18% and 23% lower than in the comfortable speed trials, for LW and WW, respectively.

A previously validated protocol - Outwalk [31,32], already used with a sample of young adults for the analysis of walking in water [28], was used. Following the Outwalk protocol, the body was modelled as an open kinematic chain constituted by 5 rigid segments (trunk, pelvis, right thigh, right shank, right foot) connected by ball-and-socket (trunk-pelvis, hip, ankle) and ‘loose’ double-hinge (knee) joints. The sagittal and frontal planes of the right side were considered and analyzed. The static posture was considered as the zero-offset-condition so joint angles in the dynamic trials were estimated with respect to that alignment condition of the body segments. Additional details regarding the protocol and the fixing procedure are reported in previous studies [28,31].

## Data Analysis

Joint angles were estimated decomposing joint matrices computed between adjacent segments, filtered (3Hz 2nd order Butterworth low-pass filter), segmented using an automatic algorithm based on wavelet analysis [33], and time-normalized into gait cycles. In order to remove outliers in the segmentation algorithm, only gait cycles with Intraclass Correlation Coefficient (ICC) values of the medio-lateral angular velocity of the shank with respect to the mean pattern of young adults greater than 0.6 were selected. Subsequently, in order to have consistent value for each participant, only gait cycles with ICC values of hip, knee, and ankle flexion-extension angles simultaneously greater than 0.75 [28] were selected for analysis. Finally, to have the same number of gait cycles for each participant and environment, a deletion procedure was applied selecting the gait cycles belonging to the central part of the 10-m walking trial, and excluding the gait cycles before and after the turns. Thirty gait cycles per participant were selected, resulting in a total number of 270 gait cycles of each joint. For the comparison between environments, 15 gait cycles at self-selected walking speed and 15 gait cycles at slow speed were used, while for the comparison with young adults, 30 gait cycles at self-selected speed were used.

For each gait cycle, the following spatio-temporal and joint kinematics variables were estimated: stride, stance and swing times [s], stance and swing percentages [%], stride length [cm], normalized stride distance (stride distance/leg length) [34], number of steps, and walking speed [cm/s] [33]. Furthermore, for each joint angle were calculated the Range Of Motion (ROM) [deg], minimum [deg], maximum [deg], percentages of gait cycle at minimum and maximum [%], and values of the angle at foot strike and at foot off [deg] points. Reference bands for joint angles were represented with mean and standard deviations.

To identify the effects of environment (LW/WW) and of age (elderly/young), by taking into account the effect of average speed of each assessed gait cycle, two linear mixed models were used. We used this statistical approach because linear mixed models allows to optimally model the within- (cycle-to-cycle) and between-subject variability of examined variables within the present data set.

The first model, shown in equation 1, was applied using LW and WW data from the elderly sample, whereas the second model, shown in equation 2, was applied using WW data at the self-selected comfortable speed from elderly and young adults:

$$y_{ij} = u_{oj} + \beta_0 + \beta_1 \cdot x_{1ij} + \beta_2 \cdot x_{2ij} + \beta_3 \cdot (x_{1ij} \cdot x_{2ij}) + \varepsilon_{0ij} \quad (1)$$

$$y_{ij} = u_{oj} + \beta_0 + \beta_1 \cdot x_{1ij} + \beta_2 \cdot x_{2ij} + \beta_3 \cdot (x_{1ij} \cdot x_{2ij}) + \varepsilon_{0ij} \quad (2)$$

where $\beta_0$ represent the fixed intercepts, $u_{oj}$ represent the random effects associated to the j$^{th}$ subject, the subscripts ij indicate values of variables measured in the i$^{th}$ cycle clustered in the j$^{th}$ subject, $y$ represents the outcome variable, $x_1$ represents the environment in equation 1 (coded with a dummy equal to 1 or 0 for the WW and LW, respectively) or the age in equation 2 (coded with a dummy equal to 1 or 0 for the elderly and young adults conditions, respectively), $x_2$ represents the dimensionless walking speed, and $\varepsilon_0$ is the random error component.

In the models we used, the coefficients ($\beta_0$, $\beta_1$, $\beta_2$, $\beta_3$) are interpreted in the following way:

$\beta_0$ (fixed intercept): the intercept of the baseline category ~~(coded as 0)~~;

$\beta_1$: the difference in the intercept between the baseline category (coded as 0) and the other category (coded as 1);

$\beta_2$: the slope for the continuous predictor (walking speed) of the baseline category ~~(coded as 0)~~;

$\beta_3$ (interaction): the difference in the slope for the continuous predictor (walking speed) between the baseline category (coded as 0) and the other category (coded as 1).

In both linear mixed models, the dimensionless speed (walking speed/sqrt(g*leg_length)) was used as a predictor variable attempting to account for the influence of anthropometric differences [34]. Scatterplots showing the effects of dimensionless walking speed and environment or age group on the examined variables were visually inspected for data consistency

Data collection was performed using the Motion Studio Software (APDM, USA), while data processing was performed using Matlab® language (The Mathworks™, USA, version R2018a). Statistical analyses were performed using the R statistical software (version 3.5.2). When not otherwise indicated, data are presented as mean ± standard deviation.

# Results

Walking speed and normalized walking speed for elderly in both conditions and for young adults in WW are reported in Table 1.

*Table 1 here*

### *Water vs land conditions*

Kinematic parameters showing a significant effect of the WW vs LW in elderly are shown in Table 2, together with the corresponding estimated coefficients of the linear mixed models. A reduced walking speed was observed in WW (27.9±8.3 cm/s, 0.09±0.03 dimensionless value) with respect to LW (69.3±10.7 cm/s, 0.24±0.04 dimensionless value). Considering the spatio-temporal parameters, a smaller stride distance with larger stride duration was observed in WW (Table 2).

*Table 2 and Figure 1 here*

Similar joint angles patterns, although with some distinctive differences, were found in the sagittal (Figure 1, left) and frontal (Figure 1, right) planes. Analyzing the hip joint in the sagittal plane, higher flexion values were observed at the foot off (approximately 15°), at the minimum value (approximately 11°) and at the maximum value (approximately 11°) in WW vs LW (Table 2, Figure 1). Different effects of walking speed were observed in the two environments (Table 2). Analyzing the knee joint, a larger flexion at foot strike (approximately 28°) was observed in WW vs LW. This difference can be also explained by the different effects of walking speed in the two conditions (Table 2, Figure 2). Indeed, one-unit increase of dimensionless walking speed in LW, induced a slight (non-significant) decrease of knee flexion-extension at foot strike, and an increase of 111° of knee flexion-extension in WW.

*Figure 2 here*

In the frontal plane, the trunk-pelvis and the hip joints showed negligible differences between conditions. From the pattern of reference bands (Figure1), an indistinct pattern of mean values for WW in the trunk-pelvis joint can be observed. Analyzing the mean value over the stance phase of the ankle inversion-eversion, an approximately 9° larger value was observed in WW. For each unit increase of dimensionless walking speed, a decrease of 23° and increase of 6° of mean value during stance of ankle inversion-eversion were estimated in LW and WW, respectively. The different effect of walking speed in the two environments can thus partially explain the above results (Table 2).

### *Elderly vs young adults in WW*

Lower walking speed (29.5±7.9 cm/s and 57.5±14.1 cm/s, for elderly and young adults, respectively) and stride distance were observed in elderly. Among all analyzed parameters, those showing a significant effect of age are shown in Table 3. In the sagittal plane, a slightly more flexed hip joint in elderly was observed, whereas the knee joint showed no differences between ages (Figure 3). In the pre-swing phase, a less plantar-flexed ankle joint was estimated in the elderly (approximately 9°). The different effect of walking speed in the two different ages can partially explain the differences between mean values in young adults vs elderly (Table 3, Figure 4). No differences were observed in the frontal plane, with the exception of a greater variation of trunk-pelvis angle in elderly.

*Table 3, Figure 3 and Figure 4 here*

## Discussion

In the present study, we assessed the gait kinematics in healthy elderly during walking in water as compared to walking on land and to the walking characteristics of young adults in water only. It is well known that the use of IMMUs allows a fast set-up and data analysis, making this technology potentially suitable for the routine practice during a session of aquatic exercise. It must be said that, this exploitation requires the use of algorithms and devices able to provide robust and reliable orientation of the sensor with respect to the global frame. Nevertheless, the most important advantage of using IMMUs instead of video analysis is the possibility of using a high number of gait cycles. Thanks to the non-restricted field of view, a closer analysis of the gait of a person can be performed. A novel approach of this study was to assess the influence of walking speed on gait parameters during walking in water. Linear mixed models showed how, for most of the gait parameters, not only the single predictors (speed, environment, and age), but also their interactions, significantly affect the gait kinematics.

Analyzing the gait spatio-temporal parameters of elderly in WW with respect to LW, the shorter stride distance and longer stride duration are compatible with the reduction of walking speed (Table 2) and congruent with what was found in previous studies using video analysis [20,22]. The percentage of duration of the stance phase was slightly increased when compared to LW, about 3%, contrary to what was shown in a previous study [20]. Comparing these parameters with those shown by young adults walking in water, the main difference was a 50% reduction of stride distance (Table 3), while only a 20% reduction was found by Barela et al. [20]. The differences with the study of Barela et al. [20] could be explained by the lower walking speed in LW and in WW shown by the present elderly. Nevertheless, this speed was consistent with previous studies assessing walking on land in 75-years old elderly [4]. Furthermore, although the gait was performed at the same water level used in the study of Barela (approximately at the xyphoid process) [20], the difference in the mean height between the two groups was slightly higher than in that study (7cm vs 5cm). The differences in the spatio-temporal parameters of walking on LW with respect to WW might be explained by the adaptations to the water environment and its specific physical properties such as drag, viscosity, and buoyancy. When the level of water immersion is increased, a person experiences modified sensorial information also, such as a lower plantar feed-back and a reduced vision of his/her steps. This causes a perception of instability and, together with the higher drag, could contribute to the reduction of walking speed, stride distance and to the increasing of stride duration in the elderly.

Analyzing the joint kinematics of elderly in WW with respect to LW, a more flexed hip in the pre-swing and swing phases, a more flexed knee during the foot strike and the terminal swing phase, and a less-plantar flexed ankle, were observed on the sagittal plane. Furthermore, a higher inversion of the ankle joint during stance was observed on the frontal plane (Figure 1, Table 2). These findings confirm observations of studies using video analysis [20,22]. Comparing the same parameters with respect to young adults walking in water, we observed a more flexed hip and a less plantar-flexed ankle (Figure 3, Table 3). These results confirm previous observations related to sagittal plane parameters [22] in an elderly population on average 10 years younger than that of the present study. Jabbar et al. [22] hypothesized a direct relation between the effect of the hydrodynamic force (drag) and the pattern observed in elderly and young adults on the sagittal plane at the hip joint. As no significant differences were found in the trunk-pelvis joint in the sagittal plane, it seems that to overcome the higher resistance in gait progress due to the higher density of water with respect to air, the participants would tend to incline forward the trunk-pelvis segment as if it was one rigid segment, probably resulting in a higher flexion of the hip. The modifications shown here for the hip and the ankle joints angles more enhanced in the elderly, are also consistent with previous results concerning gait analysis of elderly in LW, where the influence of walking speed was investigated [13], and with the hypothesized possible subtle hip flexion contracture and ankle plantar-flexor concentric weakness of the older population [13]. From this point of view, aquatic exercises could be used for strengthening the hip flexors in a safe environment and in a less loaded condition. However, a more comprehensive investigation regarding the muscle activation of the lower limbs and the inclination of the trunk against the water should be performed.

Linear mixed models underlined how gait parameters were affected by the environment, age and walking speed, and also by the interaction between these variables (Tables 2 and 3, Figure 2 and 4). An interaction between speed and environment indicates that speed changes have a different effect on kinematic variables in WW as compared to LW, while an interaction between speed and age indicates that speed changes have a different impact in elderly compared to young adults. The significant speed x environment interactions observed for the majority of examined variables (Table 2) can be explained considering the different properties of water as compared to air in terms of density/viscosity and thus in the hydrodynamic force. Instead, the significant age x speed interactions observed for most variables (Table 3), suggest that the different musculoskeletal conditions of elderly vs young adults can affect the way gait kinematics changes with varying speed in the two populations. The linear mixed modelling analysis allows a better interpretation of the effect of the examined variables. Indeed, when the interaction coefficient ($\beta_3$) is significant, there is a different effect of walking speed in the two conditions (water vs land or elderly vs adults) and the respective lines have different slopes (Figure 2 a,b,d and Figure 4 b,c,d). In the opposite case (that is, no significant interaction), the two lines are parallel (Figure 2c and Figure 4a). In a few cases, a large vertical dispersion was observed around the fitted lines (Figure 4d), highlighting the importance of visually assessing the examined effects by means of plots with raw data of single participants, besides examining the model coefficients. These findings highlight the importance of assessing walking speed during walking in water, as differences in gait parameters between environments or age groups can, at least partially, be a consequence of differences in walking speeds and on how speed interacts with the environment or age.

The fear of falling, connected with a reduced balance capacity, undermines seriously the quality of life in elderly. The control of body stability and postural position is governed not only by the interaction and the integration of sensorial inputs (visual, vestibular and proprioceptive) but also by the experience, neuromuscular strategies, specific demands of the motor task, environment and age [35]. In this context, performing aquatic exercise is particularly important for elderly because it allows the possibility of exploring a variety of movements in a safe environment. This condition can be achieved thanks to the physical characteristics of water and especially to the hydrodynamic and hydrostatic forces that sustain an exercising person. In this respect, the higher variability of the trunk-pelvis kinematics in the frontal plane shown by elderly with respect to young adults (Figure 3) highlights an environment where the elderly are forced to explore a variety of movements in a safer situation for the fear of falling with respect to dry-land, and thus can be seen as a guidance for motor skill acquisition. However, after having performed a dedicated validation of the measurement, a specific investigation of the orientation of the trunk should be accomplished to better understand the role of this body segment during progression in the water environment.

The difference in height between the two groups lead to a variability in the immersion level, going from the xiphoid process to the shoulder, and reasonably determining a difference in the offloading body weight of about 6%, although without variations in the drag (68.1±4.8N vs 68.4±5.8N, elderly vs young adults, respectively [36]). Nevertheless, as the water level can't be adjusted in most swimming pools, the issue of the effect of subject's height on in-water walking kinematics is important and warrants to be further investigated.

Despite the great advantages of the IMMU technology, on the other side, video analysis allows a higher accuracy when estimating joint kinematics and provides the position of anatomical landmarks. For this reason, if significant differences between conditions were lower than about 10 degrees, they were not taken into account and commented. For similar reasons, connected with the accuracy of kinematics reconstruction, knee ab-adduction was not considered reliable in the present analysis and thus no comparison with previous results [22] for this specific degree of freedom was performed.

In conclusion, the present study has analyzed the 3D gait kinematics parameters of lower limbs and trunk segment during walking in water of healthy elderly, taking into account for the first time the effects of the dimensionless walking speed. The results showed specific patterns for elderly population during aquatic exercise, highlighting the importance of future investigations employing age as a continuous variable to understand more in detail its interaction with walking speed and environment. The differences shown by elderly with respect to young adults about hip and ankle joint on the sagittal plane were influenced not only by speed and age, but also by the interaction between the two variables. For this reason, in particular during WW, assessing the walking speed is fundamental as gait parameters can be different due to speed not only as a consequence of the different environment.

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