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Research article

Using an ankle robotic device for motor performance and motor learning evaluation

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ABSTRACT

In this paper we performed the evaluation of ankle motor performance and motor learning during a goal-directed task, executed using the pediAnklebot robot. The protocol consisted of 3 phases (Familiarization, Adaptation, and Wash Out) repeated one time for each movement direction (plantarflexion, dorsiflexion, inversion, and eversion). During Familiarization and Wash out subjects performed goal-directed movements in unperturbed environment, whereas during Adaptation phase, a curl viscous force field was applied and it was randomly removed 10 times out of 200. Ankle motor performance was evaluated by means of a set of indices grouped into: accuracy, smoothness, temporal, and stopping indices. Learning Index was calculated to study the motor learning during the adaptation phase, which was subdivided into 5 temporal intervals (target sets). The outcomes related to the ankle motor performance highlighted that the best performance in terms of accuracy and smoothness of the trajectories was obtained in dorsiflexion movements in the sagittal plane, and in inversion rotations in the frontal plane. Differences between movement directions revealed an anisotropic behavior of the ankle joint. Results of the Learning index showed a capability of the subjects to rapidly adapt to a perturbed force field depending on the magnitude of the perceived field.

1. Introduction

Motor learning is defined as a process related to practice or experience that leads to improvement in the capacity of movement [1]. It implies two different capabilities: The first one is the acquisition of new motor skills, whereas the second one is the ability to adapt existing motor skills to changes in environmental condition [2]. Several studies have been conducted to better investigate the second feature [3, 4, 5], especially referred to the adaptation of reaching movements to dynamic perturbations induced by robotic devices [6, 7, 8, 9, 10, 11, 12, 13]. In these studies, subjects commonly interact with robots that apply perturbing forces to the hand in order to change the dynamic conditions of reaching movements. The presence of the perturbing force field leads to an inconsistency between the planning and the execution of the movement, resulting in a motor error in the first trial of the reaching movement [2]. These trials are characterized by trajectories deviating from a straight line, exhibiting a typical hooking pattern. When the subjects keep performing the perturbed reaching task, the trial by trial motor error rapidly decreases, tending towards a plateau close to the baseline values achieved in unperturbed reaching tasks [14]. This indicates that the performance in the perturbed task improves with practice becoming comparable with the performance in the unperturbed task. The decrease of motor errors is typically referred to as motor adaptation [15]. When the dynamic perturbation is unexpectedly removed after adaptation, subjects move the hand as the force field was still present, showing a hooking-pattern trajectory in the opposite direction. This implies that the sensorimotor system learned an internal model to counteract the dynamic perturbation [14] and to anticipate the forces in order to eliminate the motor error.

In more recent years, thanks to the development of active ankle-foot orthoses, motor learning and adaptation has also been analyzed in motor tasks involving lower limbs. These kind of devices are designed with actuators and controllers and can generate controllable torque and forces to the individuals using it [16]. Emken *et al.* [17] analyzed the motor learning process during gait applying viscous force fields with different amplitude to the leg. The authors found that motor learning can be accelerated by transiently amplifying the environmental dynamics. In addition, the same authors provided evidence that, during gait, motor adaptation of lower limbs can be modeled as a process in which the motor system minimizes a cost function that is the weighted sum of kinematic

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error and effort [18]. Noel et al [19] developed an electrohydraulic actuated ankle-foot orthosis sable to generate force fields during walking. They used this device to evaluate if the neural control of locomotion can be modified by applying force fields toward dorsi-flexion during two specific phases of the walking, i.e. push-off and mid-stance [20]. The authors have found that the motor adaptation is present only when the force field is applied during mid-stance phase, whereas kinematics of the push-off phase remained similar to control when the force field was unexpectedly removed. Using the same experimental setup, Blanchette *et al.* [21] evaluated the effect of a force field applied to the ankle in plantarflexion direction during the swing phase, focusing on the analysis of the electromyographic activity of five lower limb muscle groups. They have discovered an increase in tibialis-anterioris muscle activation that carried over after perturbation removal, leading to an increased ankle dorsiflexion. Therefore, a training based on this approach may have the potential of improving the gait of persons with drop foot.

All the studies regarding the analysis of motor adaptation in lower limb have always been focused on the whole gait cycle, or on a single walking phase. Since walking is a complex task, it is difficult to establish if the motor adaptation is strictly related to the behavior of the single ankle joint, or if it depends on the entire kinematic chain of the lower limbs. From a literature review, it emerges that there is a lack of studies on the motor adaptation involving the ankle joint exclusively, while all other lower limb joints stay at rest. For this reason, it is clearly necessary to develop new protocols for the ankle joint in order to study the motor learning during goal-directed tasks, as already proposed in the studies focused on the upper limbs.

The evaluation of motor learning abilities related to the ankle is strictly dependent on the kinematic performance of the joint during goal-directed tasks. In literature, only two studies [22, 23] were focused on deeper understanding the ankle kinematic behavior during goal-directed tasks. The authors examined two different aspects of ankle movements and compared them to what was already known for the upper limb. In particular [22], investigated the trade-off between speed and accuracy during goal-directed movements performed with the ankle in both dorsi-plantarflexion and inversion-eversion direction. It was the discovered that, as for the upper limb goal-directed movements, the speed accuracy relation can be described using the Fitts' law. The second study [23], instead, was focused on the analysis and modeling of the ankle speed profile during the execution of goal-directed movements in both sagittal and frontal plane. The results suggested a remarkable similarity between the top-performing models that described the speed profiles of ankle movements and the ones previously found for the upper limb. However, an objective analysis of the ankle trajectories related to a goal-directed task is a facet of the ankle movements that, to the best of authors' knowledge, has not yet been examined in the literature. Such an analysis can provide relevant insights into ankle movements, quantifying kinematic parameters, like the accuracy and the smoothness of the performed trajectories, in different movement directions.

Consequently, this study has got a twofold aim. On one side we aim to characterize trajectories related to dorsi-plantarflexion and inversion-eversion of the ankle performing goal-directed movements analyzing a set of kinematic indices; on the other side, we want to evaluate motor learning in goal-directed tasks using a viscous force field able to perturb the ankle during movements.

2. Materials and methods

2.1. Subjects

Twenty-two healthy subjects aged from 22 to 30 years old were enrolled in the study. The inclusion criteria were: (i) absence of neurological and visual deficits, (ii) physiological range of motion (ROM) for ankle, (iii) adequate anthropometric measures in order to freely move the ankle in the robot workspace, and (iv) right-footedness. The dominant leg was established by asking them to kick a ball [24].

Written informed consent was obtained from all subjects. The protocol was compliant with the ethical standards outlined in the Declaration of Helsinki. Research ethics approval for this study was obtained from the Ethical Committee of Sapienza University (approval date: 18-02-2016).

2.2. Experimental setup and procedure

All measurements were conducted by using pediAnklebot [25]. Subjects were seated in front of the PC monitor with knee flexed at 45°. They wore the knee brace fixed to the lower-limb by means of velcro straps, and a shoe of proper size, firmly tightened to the foot with shoelaces to prevent foot slippage. The main body of the robot was attached to the knee brace and the end-effectors were connected by means of two universal joints to a bracket fixed at the bottom of the shoe. The calf was leaned against an aluminum support covered with the foam rubber. The robot was laterally attached to the chair to make collected data free from the weight of the robot itself, and to improve data repeatability. The robot is equipped with two linear encoders and two load cells to acquire displacements and forces of each end-effector at 200 Hz. From the acquired data, rotations and moments of the ankle were obtained as reported in [26, 27]. The robot is equipped with proper shoe and knee brace and has been mainly designed for children, but it can be used by adults in the sitting configuration, by changing the dimension of shoe and knee brace, and using a new ad-hoc developed linkage between brace and chair. The experimental setup is illustrated in Figure 1.

In order to evaluate motor learning, a new game has been designed and developed. The game scenario is reported in Figure 2. Ankle rotations in inversion-eversion (IE) and dorsi-plantarflexion (DP) directions moved a pointer (yellow dot) in the game scenario along the x and y axis, respectively. Therefore, each point of the game was described by two coordinates that corresponded to inversion and plantarflexion angles in the IE-DP space. Before starting the experimental procedure, the foot at 90° with respect to the shank was selected as neutral position [28] and it corresponded to the yellow dot placed at the center of the game scene. This position represented the foot starting position at the beginning of the game. Once the game started, a picture appeared on the monitor. The subject was asked to move the pointer to reach the object (target), performing a movement as fast as possible, trying to maintain a straight trajectory, without stopping during the movement. Moreover, the subject was instructed to stop the ankle movement as soon as the target was reached, and to wait until a new target appeared on the scene.

The game was divided into two subgames characterized by different movement directions. In the first, named DP, ankle dorsi- and plantarflexion rotations moved the pointer to reach the targets alternatively positioned up (Figure 2(a)) and down (Figure 2(b)) on the monitor. In the second, named IE, targets were positioned in the right (Figure 2(c)) and the left sides of the scene (Figure 2(d)). Right and left movements corresponded to eversion (inversion) and inversion (eversion) rotations if they were performed with the right (left) ankle. In both DP and IE games, targets were positioned 10° from the center in each movement direction and the pointer was free to move in every part of the scene.

In order to evaluate the motor learning, the robot was used to perturb the ankle movements. The perturbation consisted in a curl viscous field, i.e. a pattern of force vectors with amplitude proportional to the instantaneous speed of the foot and direction perpendicular to the corresponding direction of the velocity vector [13]:

$$\begin{bmatrix} \tau_{IE} \\ \tau_{DP} \end{bmatrix} = \begin{bmatrix} 0 & \lambda \\ -\lambda & 0 \end{bmatrix} \begin{bmatrix} \dot{\theta}_{IE} \\ \dot{\theta}_{DP} \end{bmatrix} \tag{1}$$

where τ and $\dot{\theta}$ are the commanded torque applied by the robot and the angular velocity of the ankle, respectively; subscripts IE and DP refer to inversion-eversion and dorsi-plantarflexion directions, respectively; the coefficient λ represents the damping of the virtual environment in which the pointer is moving and its value was set to 2 $\frac{\text{Nm}}{\text{rad}}s$, constant for the

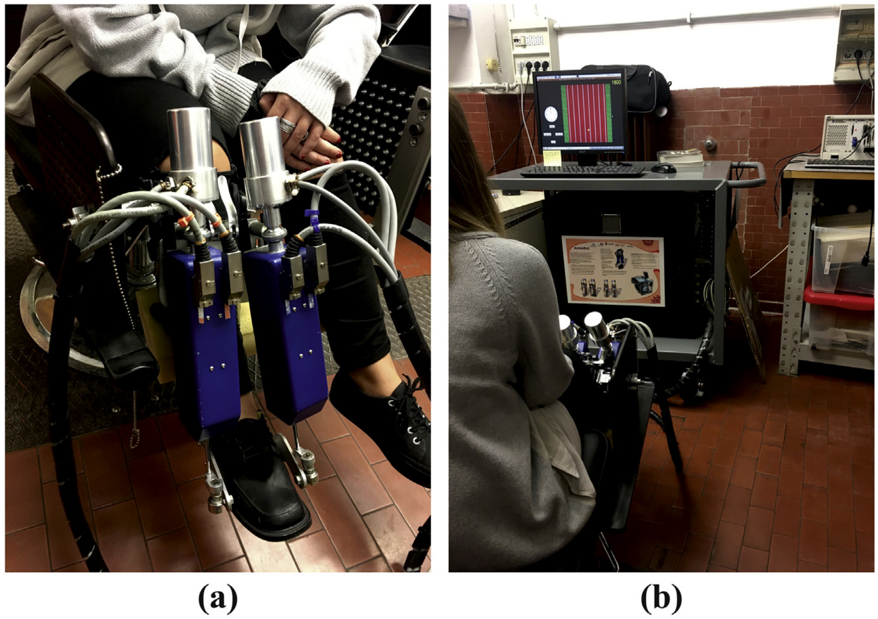

Figure 1. Experimental setup: subject wearing the robot (a) and subject performing the experimental task (b).

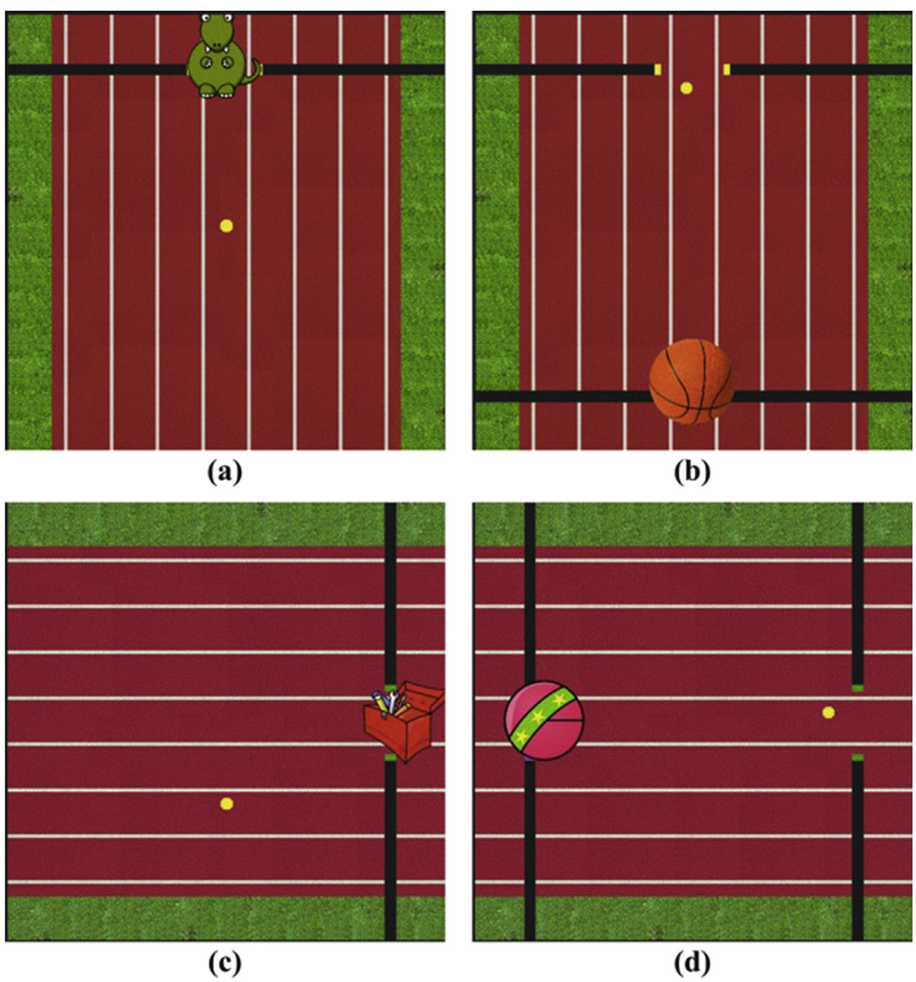

Figure 2. Game scenario with up- (a), down- (b), right- (c), and left-target (d).

entire session. The selected value of λ was chosen taking into account a pilot trial session in order to obtain a tradeoff between a sufficient deviating torque and an acceptable level of fatigue felt by subjects.

The protocol (Figure 3) consisted of four trial sessions (TS1, TS2, TS3, TS4) for each of which the curl viscous force field was applied only during one movement direction (upward, downward, rightward, leftward). Specifically, each trial session was composed by the following experimental phases:

- Familiarization phase (FA): 80 goal-directed movements in the unperturbed environment;
- Adaptation phase (AD): 200 goal-directed movements of which only the ones performed in the selected direction were perturbed with the curl viscous force field; the field was randomly switched off 10 times (catch trial);
- Wash out phase (WO): 20 goal-directed movements in the unperturbed environment.

The familiarization phase aimed both to make the subject more familiar with the experimental setup and to evaluate ankle behavior during movements in an unperturbed environment. The adaptation phase allowed the evaluation of motor learning. The Wash out phase was needed to avoid subjects' accommodation to the force field.

Subjects performed the protocol with both dominant (right) and non-dominant (left) limb.

2.3. Data analysis

Data acquired by the sensors of the robot were processed offline. The recorded position of the pointer was filtered with a 6th order, zero phase shift low-pass Butterworth filter, with a cut-off frequency of 10 Hz, and then differentiated, with a two point differentiation, to obtain speed, acceleration, and jerk. The global data obtained from each subject was divided into single goal-directed movements. In particular, each movement was assumed to start when the speed magnitude became greater than the 10% of the peak speed; the movement was assumed to end when the speed dropped and remained below the 10% of the peak speed [6].

Goal-directed movements were accurately screened and trajectories were discarded when following cases occurred: (i) the movement began before the new target appearance, i.e. the initial velocity was not equal to zero, and (ii) when the subject stopped the ankle movement before reaching the target. The number of discarded trials was lower than 5% for each subject. Finally, data was divided into four groups, according to movement directions. For each trial session, only movements toward a single direction were analyzed, i.e. upward, downward, rightward, and leftward for TS1, TS2, TS3, and TS4, respectively.

To characterize the kinematics of the movement a set of indices was evaluated and grouped into: (i) accuracy indices, (ii) smoothness indices, (iii) temporal indices, and (iv) stopping indices.

The accuracy indices are the *Length Ratio* (*LR*) and the *Lateral Deviation* (*LD*). In particular, *LR* is the ratio between the path actually travelled and the ideal one, i.e. the minimum distance between the beginning of the trajectory and the center of the target to be reached. Higher values of *LR* represent a goal-directed task performed with a lower accuracy. *LD* is defined as the highest deviation from the straight line connecting the starting and the target position. *LD* value increases when the movement accuracy decreases.

The smoothness indices are *Speed Metric* (*SM*) and *Normalized Jerk* (*NJ*). *SM* is measured as the ratio between the mean and the peak speed. The *SM* value increases when movement smoothness increases. *NJ* is the Normalized Jerk as proposed by Teulings et al. [29]. Lower values of *NJ* indicate smoother movements.

The set of temporal indices is constituted by the *Duration of Movement* (*T*), and by two novel indices herewith proposed: *Time Position Symmetry* (*TPS*) and *Time Velocity Symmetry* (*TVS*). *T* is the time between the movement onset and the movement termination, evaluated according to the speed threshold. The remaining two indices quantify the temporal symmetry of kinematic parameters of the trajectory. In particular, *TPS* is defined as follow:

$$TPS = \frac{\Delta t_i}{T} \qquad i = E, P \tag{2}$$

For the DP movements, i is equal to E and *TPS* represents the temporal duration (Δt_E) of the eversion rotations with respect to the total duration of the trajectory. Analogously, for IE movements, i is equal to P and *TPS* represents the temporal duration (Δt_P) of the plantarflexion rotations with respect to the total duration of the trajectory. In fact, when the foot is moved upward or downward the corresponding ankle rotation is not a pure dorsiflexion or plantarflexion, because a component of inversion or eversion is always present. The same occurs for rightward or leftward movements, for which the presence of dorsi-plantarflexion rotation is not negligible. *TPS* ranges from 0 to 1. Considering dorsi-plantarflexion rotations (inversion-eversion rotations), a *TPS* value equal to 0 reveals the absence of eversion (plantarflexion) rotations during the movement, whereas *TPS* equal to 1 denotes the absence of inversion (dorsiflexion) rotations. If *TPS* value is equal to 0.5, it means an equal temporal duration of the inversion and eversion rotations (dorsiflexion and plantarflexion rotations) during dorsi-plantar flexion movements (inversion-eversion movements). Instead, the value of *TPS* higher than 0.5 in the plantarflexion and dorsiflexion movements means a prevalence of eversion with respect to inversion rotations, whereas in eversion and inversion rotations indicates a prevalence of plantarflexion movements with respect to dorsiflexion ones. Thus, *TPS* can be considered as a measure of the contribution of the ankle rotation around the secondary movement axis, i.e. inversion/eversion in dorsiflexion/plantarflexion movements and vice versa. *TVS* is defined as the time t_v in which the velocity peak occurred normalized to the duration the trajectory.

$$TVS = \frac{t_v}{T} \tag{3}$$

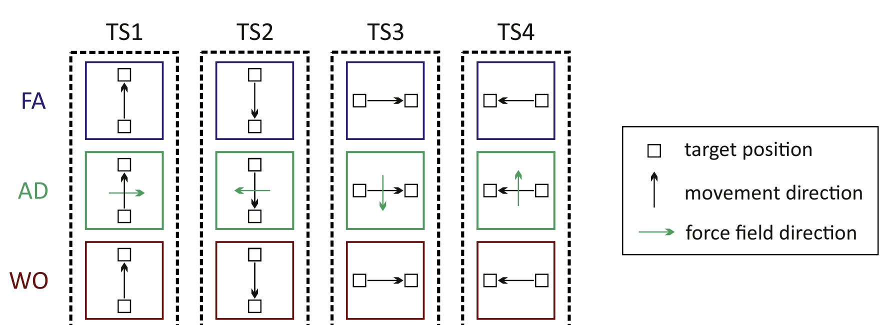

Figure 3. Experimental protocol: the four trial sessions (TS1, TS2, TS3, and TS4) composed of three phases (FA, AD, and WO). The squares represent target positions, the black arrows the movement direction analyzed for the correspondent target set, and the green arrow the direction of the force field.

TVS ranges from 0 to 1. The more *TVS* value is close to 1, the more the velocity peak occurs close to the end of the trajectory. Thus, a *TVS* value close to 0.5 represents a perfect bell shaped trajectory that is characterized by a velocity profile with the peak close to the middle of the trajectory.

The stopping indices are *Delay* (ΔT) and *Dispersion* (σ_{trj}). They evaluate the ability of the subject to stop the movement once the target is reached [30]. ΔT represents the temporal delay, normalized with respect to T, between the time in which the subject hits the target and the end of the movement evaluated accordingly to the velocity threshold. ΔT value close to zero indicates that the subject rapidly stops the ankle movement after hitting the target. σ_{trj} is defined as:

$$\sigma_{trj} = std\left(\sqrt{(d_{IE} - C_{IE})^2 + (d_{DP} - C_{DP})^2}\right) \tag{4}$$

where d_{IE} and d_{DP} are the coordinates of trajectory performed after hitting the target, and C_{IE} and C_{DP} are the coordinates of the target center position. Thus, σ_{trj} is a measure of the dispersion of the trajectory travelled after hitting the target. Low values of σ_{trj} imply a trajectory confined in a small area.

All the aforementioned indices were evaluated in both the familiarization and adaptation phases (excluding catch trials).

Finally, to evaluate motor adaptation to the force field, the *Learning Index* (*LI*) was calculated as follow:

$$LI = \frac{|LD_{catch}|}{|LD_{catch}| + |LD_{fielded}|} \tag{5}$$

where LD_{catch} was the *LD* related to a catch trial and $LD_{fielded}$ was the *LD* related to a movement performed in the perturbed environment. In particular, for the AD phase of each trial session, 5 *LIs* were calculated considering only the movement direction which the force field is applied in (100 over 200 movements). Then, *LIs* were obtained averaging *LD* values of a target set composed by 20 trials (i.e. 2 catch trials and 18 force-fielded trials). The more the subject's motor adaptation increases the more *LI* value is close to 1. *LI* was evaluated for each movement direction.

2.4. Statistical analysis

Two-way repeated measures ANOVA were performed to compare the kinematic indices considering, as independent variables, the movement directions (dorsiflexion vs. plantarflexion and inversion vs. eversion) and the experimental phases (familiarization and adaptation). If the interaction effects were significant, the interactions were broken down by comparing each phase at each movement direction and vice versa with a paired t-test. As regards the LI, one-way repeated measure ANOVA was performed to find differences among the 5 target sets. A Bonferroni's test for multiple comparisons was performed when statistical differences were found. Moreover, as regard indices which are bounded between 0 and 1, i.e. SM, TPS, TVS and LI, the logistic regression [31]was applied before performing the ANOVA. All data was test for normality with Shapiro Wilk test and sphericity was checked. If sphericity was violated, Greenhouse-Gasser correction was assumed.

3. Results

Figure 4 shows the trajectories during the entire experimental protocol performed with the dominant (right) limb for a representative subject. The first row is related to the familiarization phase of the four movement directions, whereas the second row represents the same movements performed in the perturbed environment. Considering the FA phase, it seems that dorsiflexion and inversion movements are more accurate, showing a lower dispersion of the trajectories. Considering trajectories related to the AD phase, effects of the force field are evident (catch trials), especially in dorsi-plantarflexion directions, whereas both

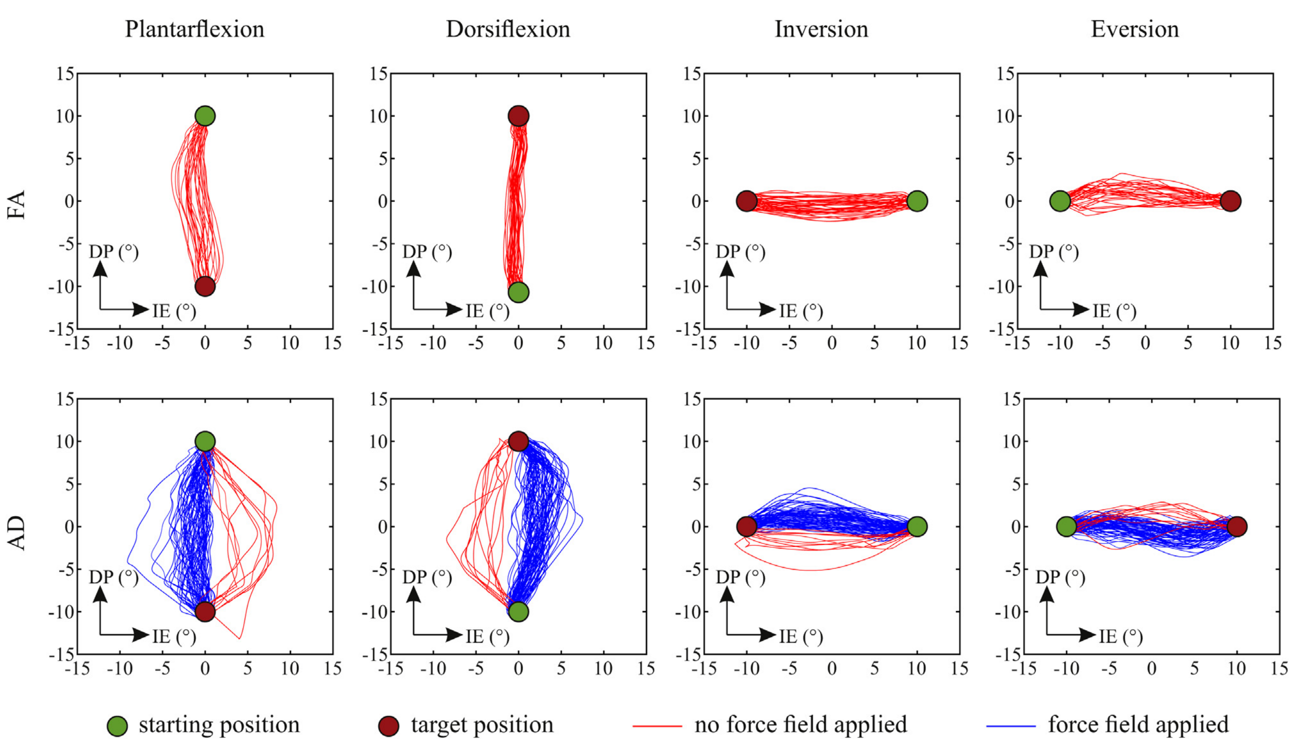

Figure 4. Trajectories of the dominant limb for a representative subject during familiarization (FA) in the first row and adaptation (AD) phase in the second row, divided in the four movement directions (plantarflexion, dorsiflexion, inversion, and eversion in the first, second, third, and fourth column, respectively). The red traces correspond to movements with no disturbing field, i.e. all trials in FA and only catch trials in AD. The blue traces correspond to movements affected by the disturbing force during AD.

fielded and catch-trial trajectories presented a less hooked shape in inversion-eversion movements.

Mean values among subjects for both dominant and non-dominant limb of accuracy, smoothness, temporal and stopping indices are reported in Figures 5, 6, 7, and 8, respectively.

3.1. Dorsi-plantar flexion movements

Mean and standard deviation values of all the kinematic indices averaged across subjects for plantarflexion and dorsiflexion movements, related to both dominant and non-dominant limb, are reported in Table 1.

Looking at the accuracy indices, *LR* presented statistical differences between both phases (dominant: $p < 0.01$, $F = 126.79$; non-dominant: $p < 0.01$, $t = 20.76$ plantarflexion FA vs plantarflexion AD and $p < 0.01$, $t = 14.14$ dorsiflexion FA vs dorsiflexion AD) and directions (dominant: $p < 0.01$, $F = 12.34$; non-dominant: $p < 0.01$, $t = 5.00$ plantarflexion FA vs dorsiflexion FA and $p < 0.01$, $t = 7.60$ plantarflexion AD vs dorsiflexion AD) in both limbs, showing the highest value in AD plantarflexion. *LD* resulted higher in AD phase than FA in both directions for both limbs (dominant: $p < 0.01$, $t = 7.94$ plantarflexion FA vs plantarflexion AD and $p < 0.01$, $t = 9.63$ dorsiflexion FA vs dorsiflexion AD; non-dominant: $p < 0.01$, $F = 162.83$). Comparing movement directions LD resulted higher in plantarflexion movements for both phases in the non-dominant limb ($p = 0.03$, $F = 5.26$) whereas, for the dominant limb, it resulted higher in plantarflexion movement in FA ($p = 0.03$, $t = 2.26$) and dorsiflexion movement of AD ($p < 0.01$, $t = 3.79$).

Considering the smoothness indices, *SM* resulted higher in dorsiflexion movement than in plantarflexion one (dominant: $p < 0.01$, $F = 75.29$; non-dominant: $p < 0.01$, $F = 89.03$) and in FA than in AD (dominant: $p < 0.01$, $F = 98.15$; non-dominant: $p < 0.01$, $F = 156.21$), for both limbs. *NJ* resulted lower in FA than in AD for both plantarflexion and dorsiflexion in both limbs (dominant: $p < 0.01$, $F = 20.48$; non-dominant: $p < 0.01$, $t = 6.08$ plantarflexion FA vs plantarflexion AD and $p < 0.01$, $t = 3.34$ dorsiflexion FA vs dorsiflexion AD); moreover, *NJ* was lower in dorsiflexion than plantarflexion of AD ($p < 0.01$, $t = 4.38$) only for the non-dominant limb.

In regards to the temporal indices, *T* presented statistical differences between phases (dominant: $p < 0.01$, $F = 35.17$; non-dominant: $p < 0.01$, $t = 5.95$ plantarflexion FA vs plantarflexion AD and $p < 0.01$, $t = 4.54$ dorsiflexion FA vs dorsiflexion AD) and movement directions (dominant: $p = 0.01$, $F = 7.58$; non-dominant: $p < 0.01$, $t = 3.56$ plantarflexion FA vs dorsiflexion FA and $p < 0.01$, $t = 6.93$ plantarflexion AD vs dorsiflexion AD) for both limbs. Considering the dominant limb, *TPS* showed a higher value in dorsiflexion than in plantarflexion only for the AD phase (dominant: $p < 0.01$, $t = 10.89$; non-dominant: $p < 0.01$, $t = 16.38$). Moreover, *TPS* was lower in AD than in FA for the plantarflexion movements (dominant: $p < 0.01$, $t = 6.35$; non-dominant: $p < 0.01$, $t = 5.24$) and the opposite occurred for the dorsiflexion one (dominant: $p < 0.01$, $t = 7.45$; non-dominant: $p < 0.01$, $t = 5.24$). Considering the non-dominant limb, *TPS* showed statistical differences in phases and directions. In particular, comparing plantarflexion and dorsiflexion movements, *TPS* was lower in the latter for AD ($p < 0.01$, $t = 16.38$). Considering differences between phases, *TPS* was higher in FA than in AD for dorsiflexion movements ($p < 0.01$, $t = 5.24$), whereas the opposite occurred for plantarflexion movements ($p < 0.01$, $t = 10.82$). As regard *TVS* for both limbs, statistical differences emerged in both movement directions and phases: it resulted higher in plantarflexion than in dorsiflexion (dominant: $p < 0.01$, $F = 17.07$; non-dominant: $p < 0.01$, 4.48) and higher in FA than AD (dominant: $p < 0.01$, $F = 20.39$; non-dominant: $p < 0.01$, 18.49).

Finally, as regards the stopping indices, ΔT showed higher values in plantarflexion than in dorsiflexion for both phases and both limbs (dominant: $p < 0.01$, $F = 46.58$; non-dominant: $p < 0.01$, 16.70). Considering σ_{trj}, it resulted higher in plantarflexion movements in both phases for both limbs (dominant: $p < 0.01$, $t = 9.75$ plantarflexion FA vs dorsiflexion FA and $p < 0.01$, $t = 9.24$ plantarflexion AD vs dorsiflexion AD; non-dominant: $p < 0.01$, $t = 10.41$ plantarflexion FA vs dorsiflexion

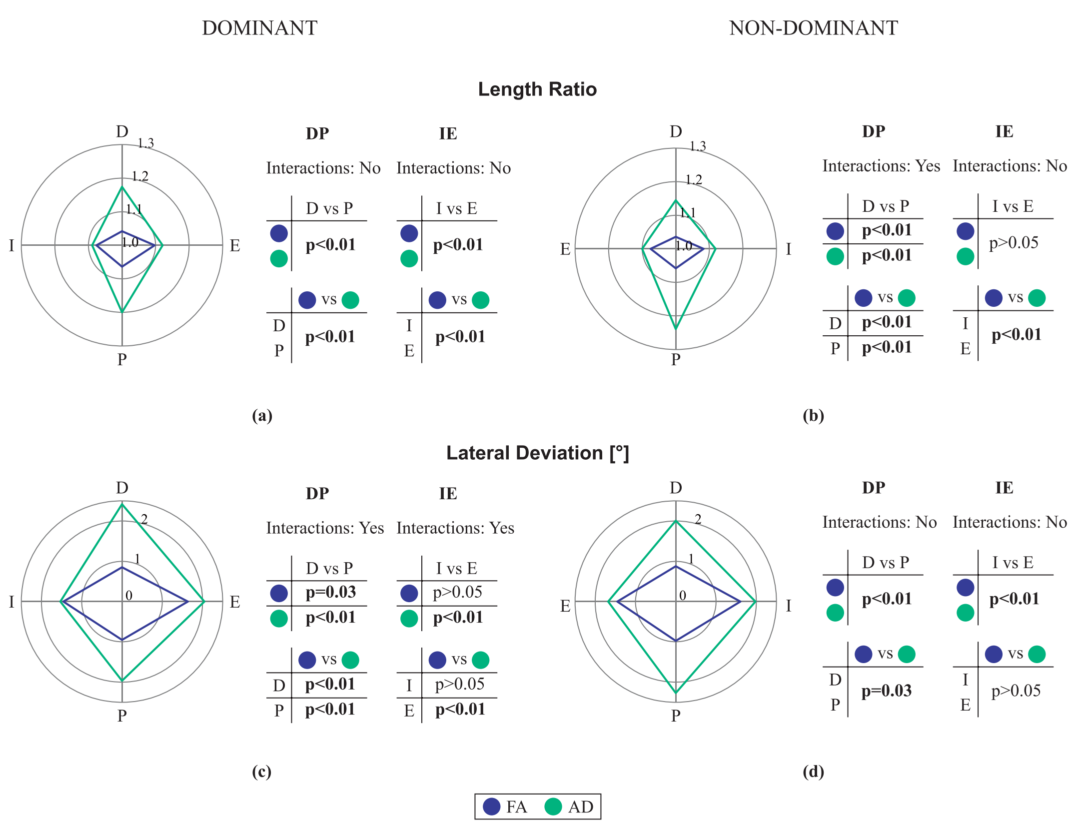

Figure 5. Accuracy indices: mean values among subjects and statistical results of LR (a, b) and LD (c, d) related to dominant (a, c) and non-dominant (b, d) limb.

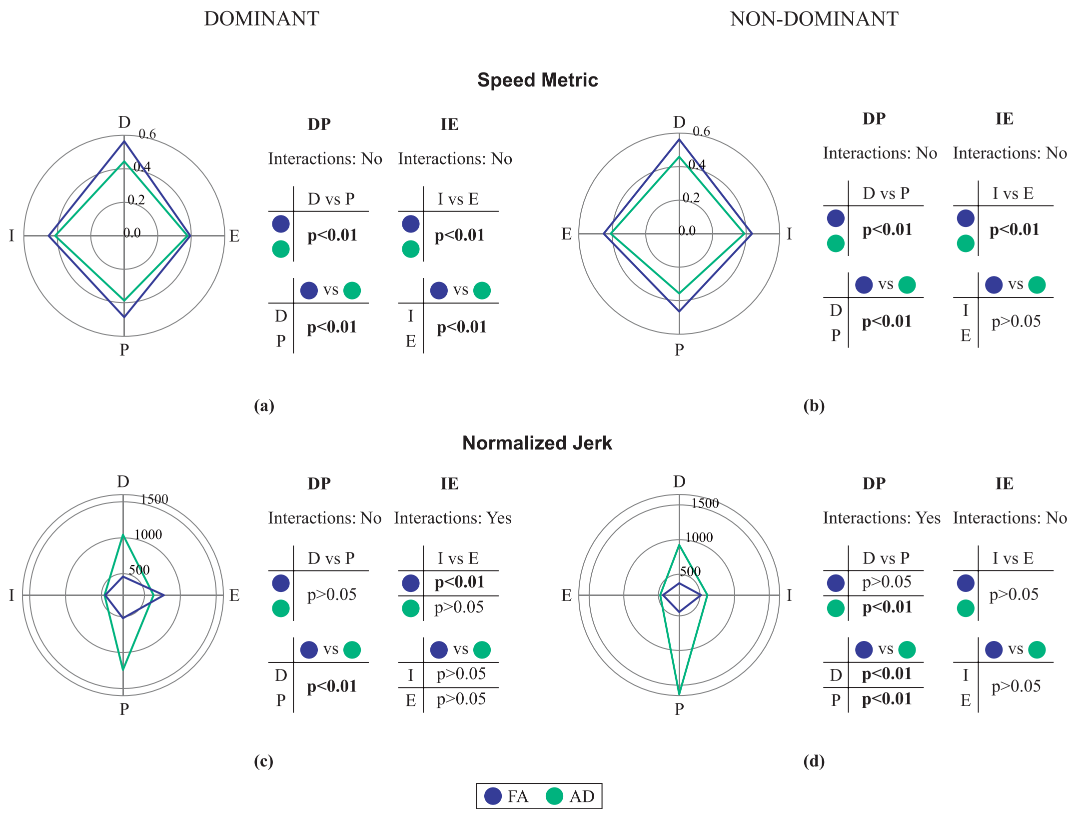

Figure 6. Smoothness indices: mean values among subjects and statistical results of SM (a, b) and NJ (c, d) related to dominant (a, c) and non-dominant (b, d) limb.

FA and $p < 0.01$, $t = 6.21$ plantarflexion AD vs dorsiflexion AD). Moreover, σ_{trj} resulted higher in FA than in AD for plantarflexion movements for both limbs (dominant: $p = 0.03$, $t = 2.25$; non-dominant: $p < 0.01$, $t = 3.60$).

3.2. Inversion-eversion movements

Mean and standard deviation values of all the kinematic indices averaged across subjects for plantarflexion and dorsiflexion movements, related to both dominant and non-dominant limb, are reported in Table 2.

Looking at the accuracy indices, *LR* resulted lower in inversion than eversion for both phases in the dominant limb ($p < 0.01$, $F = 12.26$); moreover, *LR* was higher in AD than in FA for both movement directions and limbs (dominant: $p < 0.01$, $F = 14.84$, non-dominant: $p < 0.01$, $F = 60.53$). *LD* was higher in AD than in FA ($p < 0.01$, $t = 4.35$) for the eversion of dominant limb and for both movements of the non-dominant one ($p < 0.01$, $F = 78.70$). In terms of the differences between movement directions, *LD* was higher in eversion than inversion for AD and for the dominant limb ($p < 0.01$, $t = 4.36$).

Considering the smoothness indices, *SM* resulted higher in FA than AD, for both directions and for both limbs (dominant: $p < 0.01$, $F = 11.85$, non-dominant: $p < 0.01$, $F = 35.52$); moreover, *SM* resulted higher in inversion than in eversion for both phases ($p < 0.01$, $F = 18.17$) of the dominant limb. In relation to*NJ*, a statistical difference occurred only between inversion and eversion in FA related to the dominant limb ($p < 0.01$, $F = 12.77$), showing the lowest value in inversion movements.

Considering the temporal indices, *T* resulted higher in eversion than inversion in both phases for the dominant limb ($p < 0.01$, $F = 10.47$) and vice versa for the non-dominant one ($p < 0.01$, $F = 8.86$), whereas no differences emerged between phases. *TPS* showed statistical differences between movement directions; in particular, it was higher in inversion than in eversion for FA ($p < 0.01$, $t = 4.14$) and vice versa for AD ($p < 0.01$, $t = 4.03$) in the dominant limb, and for FA ($p < 0.01$, $t = 4.77$) and AD ($p < 0.01$, $t = 7.47$) in the non-dominant one. Moreover, *TPS* was higher in FA than in AD for the inversion of the dominant limb ($p < 0.01$, $t = 7.61$) and for the eversion of the non-dominant one ($p = 0.01$, $t = 2.70$); instead, *TPS* was lower in FA than in AD for the eversion of dominant limb ($p < 0.01$, $t = 3.41$). No differences of *TVS* were found between phases and directions for both limbs.

Finally, considering the stopping indices, ΔT resulted higher in inversion than eversion of the dominant limb for both phases ($p = 0.01$, $F = 9.52$). σ_{trj} was higher in FA than in AD for the inversion related to the dominant limb ($p < 0.01$, $t = 3.30$) and for both movement directions performed with the non-dominant limb ($p = 0.01$, $F = 8.16$).

3.3. Learning index

LI values related to the dominant limb, grouped in the four movement directions are reported in Figure 9.

With respect to plantarflexion movements, statistical differences were found between T1 and T3 ($p < 0.01$, $t = 3.00$), T1 and T4 ($p = 0.05$, 3.13), and T1 and T5 ($p = 0.03$, $t = 3.42$). Looking at dorsiflexion, Bonferroni's tests demonstrated differences between T1 and all target sets (T1 vs T2 $p = 0.02$, $t = 3.60$, T1 vs T3 $p = 0.04$, $t = 3.31$, T1 vs T4 $p < 0.01$, $t = 5.05$, and T1 vs T5: $p = 0.01$, $t = 3.88$) and between T3 and T4 ($p = 0.04$, $t = 3.30$). As regard eversion movements, statistical differences were found between T1and all target sets (T1 vs T2 $p = 0.04$, $t = 3.22$, T1 vs T3 $p = 0.03$, $t = 3.55$, T1 vs T4 $p = 0.01$, $t = 4.00$, and T1 vs T5 $p = 0.01$, $t = 3.85$). No differences were found among target sets for the inversion direction.

LI values related to the non-dominant limb grouped in the four movement directions are reported in Figure 10.

In relation to plantarflexion movements, statistical differences were found between T1 and T2 ($p = 0.03$, $t = 3.37$), T1 and T3 ($p < 0.01$, $t = 4.60$), T1 and T4 ($p = 0.03$, $t = 3.36$), and T1 and T5 ($p < 0.01$, $t = 4.20$). Looking at dorsiflexion movements, statistical differences were found between T1 and T3, T4 and T5 (T1 vs T3 $p < 0.01$, $t = 3.41$, T1 vs T4 $p =$

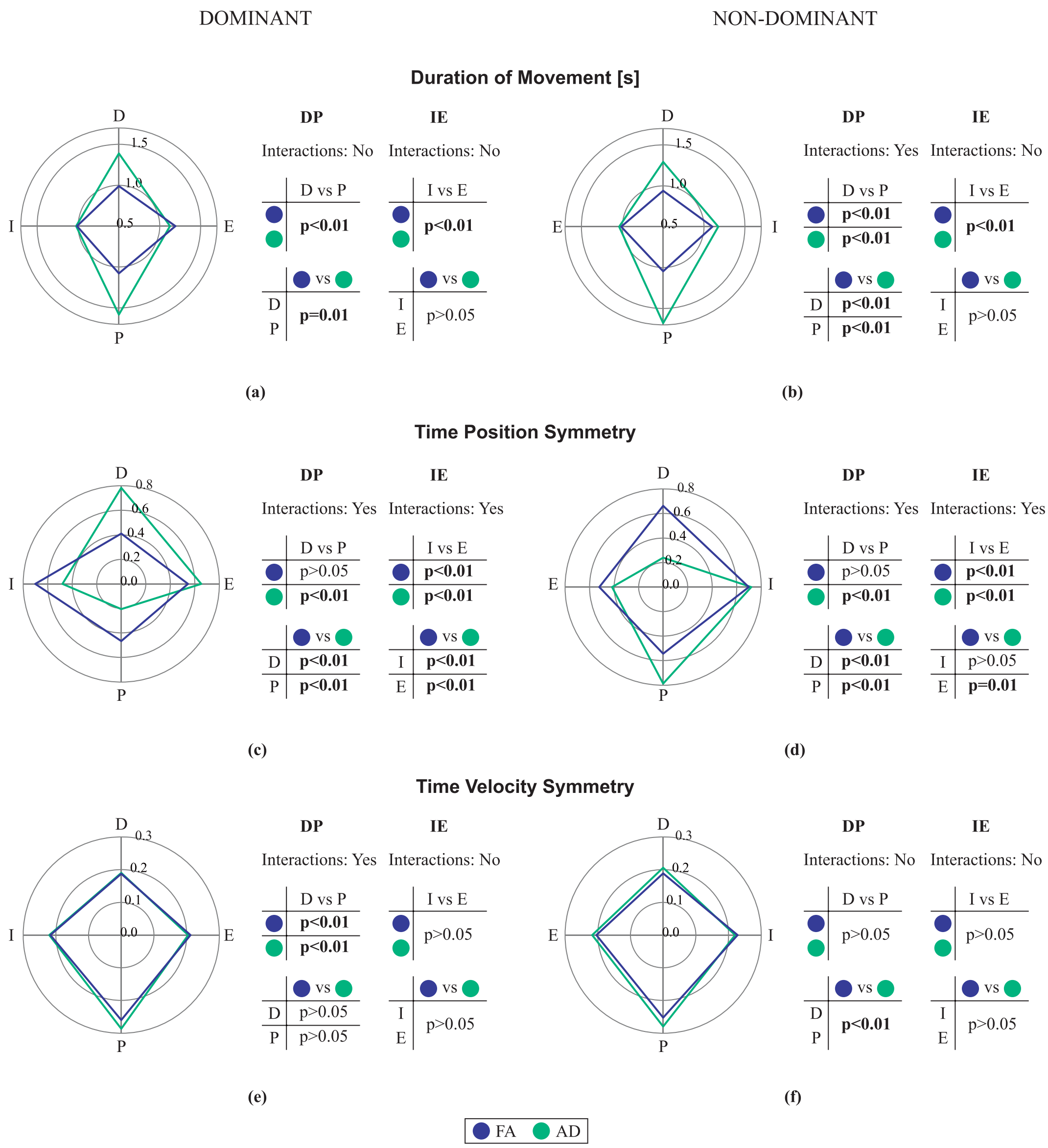

Figure 7. Temporal indices: mean values among subjects and statistical results of T (a, b), TPS (c, d) and TVS (e, f) related to dominant (a, c, e) and non-dominant (b, d, f) limb.

0.05, t = 3.81, and T1 vs T5 $p < 0.01$, t = 3.76). LIs evaluated in inversion direction showed differences between T1 and T4 ($p < 0.01$, t = 4.86) and T1 and T5 (p = 0.04, t = 3.24). No differences were found among target sets for the eversion direction.

4. Discussion

In this study, a set of indices was used to assess ankle motor performance during rotations in the sagittal and frontal planes, under two different dynamic conditions due to an unperturbed and a perturbed environment by means of a curl viscous force field. Motor tasks performed within the unperturbed environment were analyzed in order to evaluate if the ankle motor performance is related to the movement direction. Successively, a comparison was made between the kinematic indices gathered during motor tasks performed with and without perturbations to analyze if the presence of a curl force field affects ankle movements. Finally, LI was analyzed in the different movement directions in order to study the presence of motor learning.

4.1. Is the ankle kinematic behavior strictly related to the movement direction in the unperturbed environment?

The comparison between movement directions related to both limbs in the unperturbed environment highlighted a different kinematic behavior of the ankle in plantarflexion and dorsiflexion rotations. In particular, dorsiflexion movements resulted more accurate and smoother than the plantarflexion, in the unperturbed environment. However, focusing on the smoothness indices, statistical differences found for SM were not always confirmed by NJ. It could be due to the higher variability of *NJ* than *SM*, as showed by the standard deviations reported in Table 1. In fact, even if the mean value of *NJ* in dorsiflexion was lower than in plantarflexion, its low repeatability did not let differences to emerge. Indeed, unlike the *SM*, the jerk indices, even those including a normalization factor, as the one used in this study, depends on other kinematic parameters such as the movement amplitude and time [32]. Their variability could cause a not negligible inter-subject variability of the *NJ*. Moreover, a different behavior between the two smoothness indices has already been reported by Merlo *et al.* [33], which found a weak

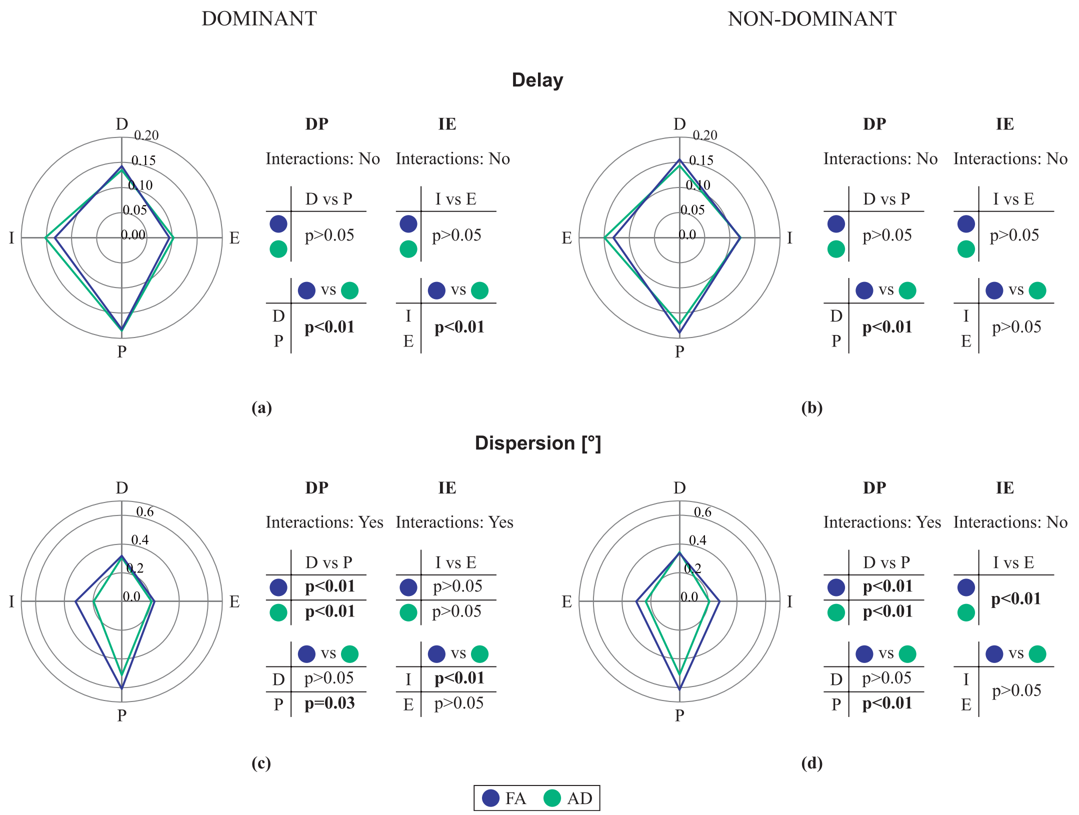

Figure 8. Stopping indices: mean values among subjects and statistical results of ΔT (a, b) and σ_{trj} (c, d) related to dominant (a, c) and non-dominant (b, d) limb.

Table 1. Means and standard deviations (in parenthesis) of all the indices averaged across subjects for plantarflexion (P) and dorsiflexion (D) movements of familiarization (FA) and adaptation (AD) phase, related to both dominant and non-dominant limb.

	Dominant				Non-dominant			
	FA		AD		FA		AD	
	P	D	P	D	P	D	P	D
LR	1.06 (0.05)	1.04 (0.03)	1.20 (0.04)	1.17 (0.04)	1.06 (0.03)	1.04 (0.02)	1.24 (0.04)	1.14 (0.05)
LD (°)	0.95 (0.33)	0.85 (0.30)	1.97 (0.65)	2.42 (0.85)	0.98 (0.33)	0.88 (0.37)	2.27 (0.63)	2.01 (0.58)
SM	0.48 (0.07)	0.56 (0.06)	0.39 (0.04)	0.44 (0.04)	0.46 (0.05)	0.56 (0.05)	0.36 (0.04)	0.46 (0.06)
NJ	521 (541)	459 (438)	1241 (682)	1042 (769)	445 (305)	371 (332)	1630 (908)	923 (919)
T (s)	1.08 (0.45)	0.99 (0.35)	1.58 (0.45)	1.39 (0.37)	1.05 (0.35)	0.94 (0.33)	1.68 (0.45)	1.29 (0.39)
TPS	0.47 (0.19)	0.41 (0.20)	0.21 (0.09)	0.78 (0.10)	0.54 (0.19)	0.66 (0.16)	0.79 (0.09)	0.24 (0.09)
TVS	0.51 (0.08)	0.44 (0.07)	0.44 (0.10)	0.37 (0.10)	0.49 (0.06)	0.45 (0.09)	0.42 (0.07)	0.40 (0.09)
ΔT	0.18 (0.05)	0.14 (0.05)	0.18 (0.07)	0.13 (0.05)	0.19 (0.05)	0.16 (0.04)	0.17 (0.05)	0.14 (0.04)
σ_{trj}(°)	0.61 (0.16)	0.32 (0.12)	0.51 (0.16)	0.30 (0.09)	0.62 (0.13)	0.34 (0.12)	0.51 (0.14)	0.34 (0.09)

correlation between the speed metric and the normalized jerk in robotic tasks performed with the upper limb. As regards the duration of movements, dorsiflexion presented a shorter time duration. It is interesting to notice how a faster execution of the movement reflected into an increase of performance in terms of accuracy and smoothness. Thus, considering these two parameters, it emerged that dorsiflexion movements showed the best performance. Our findings suggest that gravitational force influences the processes of controlling movement execution [34]. For this reason we can hypothesize that finer movements are probably needed to complete the motor task when it is performed against gravity.

Rotations in the frontal plane highlighted differences in inversion and eversion movements related to the dominant limb. Focusing on *LR*, *SM*, *NJ* and *T*, results proved that inversion movements were more accurate, smoother and faster than the eversion in the unperturbed environment. Thus, as reported for the rotations in sagittal plane, movements performed with the highest accuracy and smoothness were the ones executed more quickly. The better motor performance found for the inversion rotation can be explained by considering the different physiological ranges of motion of the ankle in inversion and eversion that are approximatively 23° and 12°, respectively [35]. In fact, subjects were asked to reach the targets positioned 10° away from the center of the screen, regardless to the direction of rotations. While this value is inside the physiological range of motion of both rotations, it is very close to the ROM limit for the eversion rotations, implying an increase of difficulty for the subject to complete the motor tasks maintaining the same level of accuracy and smoothness.

This paper presented a novel index, i.e. *TPS*, which allowed quantifying the coupling between the two degrees of freedom (DOF) of the ankle. Although ankle motions are often simply described as rotations around mutually perpendicular axes, the actual anatomic axes are not intersecting and not reciprocally orthogonal, and their position changes with the ankle rotation [36]. This complexity could introduce a not

Table 2. Means and standard deviations (in parenthesis) of all the indices averaged across subjects for inversion (I) and eversion (E) movements of familiarization (FA) and adaptation (AD) phase, related to both dominant and non-dominant limb.

	Dominant				Non-dominant			
	FA		AD		FA		AD	
	I	E	I	E	I	E	I	E
LR	1.08 (0.03)	1.10 (0.04)	1.09 (0.03)	1.12 (0.05)	1.08 (0.03)	1.08 (0.04)	1.12 (0.03)	1.10 (0.06)
LD (°)	1.46 (0.48)	1.64 (0.56)	1.53 (0.41)	2.04 (0.78)	1.60 (0.37)	1.45 (0.59)	1.98 (0.35)	1.68 (0.62)
SM	0.45 (0.06)	0.40 (0.06)	0.41 (0.05)	0.38 (0.06)	0.43 (0.06)	0.45 (0.06)	0.39 (0.05)	0.41 (0.05)
NJ	447 (310)	770 (503)	456 (360)	627 (410)	515 (542)	434 (368)	607 (464)	474 (373)
T (s)	1.01 (0.31)	1.19 (0.32)	1.02 (0.35)	1.12 (0.33)	1.10 (0.37)	1.01 (0.34)	1.17 (0.34)	1.04 (0.36)
TPS	0.70 (0.14)	0.54 (0.18)	0.48 (0.16)	0.65 (0.15)	0.70 (0.16)	0.52 (0.18)	0.72 (0.13)	0.42 (0.14)
TVS	0.45 (0.09)	0.45 (0.13)	0.46 (0.07)	0.45 (0.13)	0.46 (0.12)	0.45 (0.07)	0.43 (0.08)	0.45 (0.09)
ΔT	0.13 (0.06)	0.09 (0.05)	0.15 (0.05)	0.10 (0.05)	0.12 (0.06)	0.13 (0.06)	0.12 (0.05)	0.15 (0.06)
σ_{trj}(°)	0.32 (0.19)	0.23 (0.13)	0.20 (0.14)	0.21 (0.12)	0.28 (0.20)	0.30 (0.18)	0.21 (0.11)	0.24 (0.13)

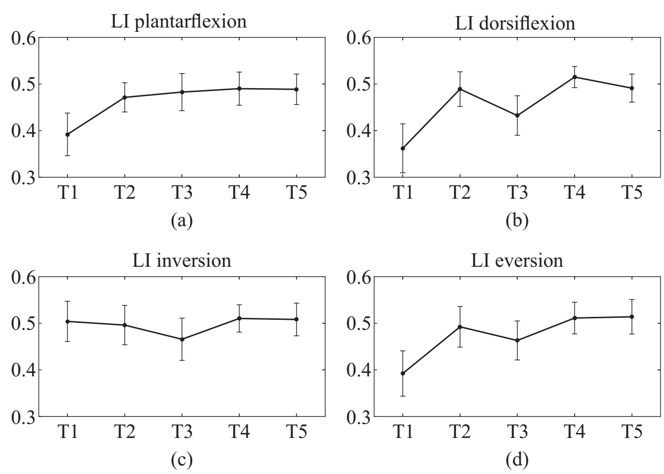

Figure 9. Mean and standard deviation of Learning Index for the 5 target sets relative to the dominant limb, averaged across subjects and grouped in the four movement directions: plantarflexion (a), dorsiflexion (b), inversion (c) and eversion (e).

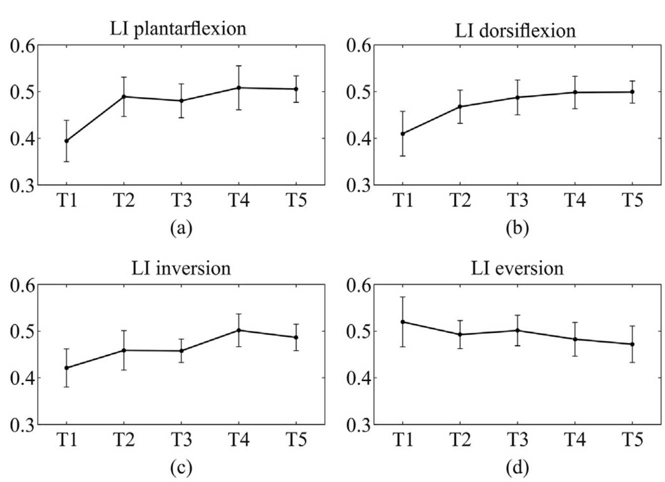

Figure 10. Mean and standard deviation of Learning Index for the 5 target sets relative to the non-dominant limb, averaged across subjects and grouped in the four movement directions: plantarflexion (a), dorsiflexion (b), inversion (c) and eversion (e).

negligible biomechanical coupling between the two degrees-of-freedom (DOF) when the required movements has to be performed in only one direction. In the present work, *TPS* highlighted no differences in the contribution of the secondary rotations between plantarflexion and dorsiflexion movements for the dominant limb. The mean values of *TPS* were close to 0.5, indicating the symmetry of the secondary rotations in both plantar and dorsal directions. Thus, it seemed that in the unperturbed environment, subjects were able to balance the contribution of rotations around the secondary axis with the dominant limb. On the contrary, the same ability in balancing the contribution of the secondary rotation did not characterize the dorsiflexion movements of the non-dominant limb, for which *TPS* showed a higher contribution of eversion rotations. This finding could be ascribed to the built-in highest dexterity in the execution of movements with the dominant limb. As regards the coupling between the two DOFs in the frontal plane, it emerges that subjects were able to balance the secondary rotations only in eversion movements, whereas an unneglectable plantarflexion component was always present during inversion movements.

Focusing on the bell-shape profile of the velocity, it emerged that the peak occurred earlier in dorsiflexion movements than in plantarflexion for both limbs. These results could be explained considering that subjects needed to contrast the gravity force while the foot is moved upwards, exerting a greater force at the beginning of the movement, which allowed to reach the maximum speed earlier in dorsiflexion than in plantarflexion. This hypothesis is confirmed considering that no differences in *TVS* were found in the frontal plane. In fact, the gravity force affects inversion and eversion movement in the same way. This finding appears to be in contrast with the one found by Michmizos *et al.* [23]. Specifically, the authors found differences in the skewness of the speed profile between movements equally affected by the gravity force, thus concluding that the gravity force did not play the main role in the ankle movements. This inconsistency could be ascribed to two factors related to the different experimental setup. The former is the different ROM covered during the movements, that was 12° in [23] and 20° in this study. The latter is related to the different visual feedback; in fact, even if the ankle was free to move in all DOFs in both studies, the pointer on the monitor was constrained to move itself along a vertical or horizontal axis during the ankle rotations in [23]. On the contrary, in our work the pointer was free to move on the screen following the ankle rotations irrespective to the target position. Therefore, in our study, since the visual feedback reflected exactly the rotations performed by the ankle, we hypothesize that subjects tried to continuously correct the ankle trajectory if the pointer moved away from the ideal trajectory, thus modifying the speed profile.

The stopping indices confirmed that the gravity force is a dominant factor in DP movements, as the indices were higher in plantarflexion than in dorsiflexion. In particular, when the foot was moved downwards, subjects encountered higher difficulty to stop the movement because they needed to counter gravity force in order to maintain the ankle in a fixed position after hitting the target.

As a general consideration, we can observe that, whereas for the sagittal plane the differences between plantarflexion and dorsiflexion movements resulted the same in both limbs, the differences emerged in the dominant limb for inversion and eversion rotations are not confirmed in the non-dominant limb. Moreover, rotations performed in same anatomical plane resulted in different kinematic performance depending on movement direction, thus unveiling an anisotropic behavior of the ankle joint.

4.2. Which are the effects of the force field on ankle movements?

The presence of the force field affected the ankle motor performance. As regards the accuracy of the trajectory, the presence of the force field led to a decrease of the performance for all movement directions in both limbs. The same occurred for the smoothness, which decreased when the movements were performed in the perturbed environment. However, whereas *SM* was higher in FA than in AD for all movement directions, *NJ* showed statistical differences only in dorsi-plantarflexion movements. The difference between the two smoothness indices could be ascribed to the higher variability of NJ that was not able to elucidate the different smoothness levels due to the force field in eversion and inversion directions. Considering the temporal indices measured in dorsi-plantarflexion movements, the force field determined an increase of the time needed to end the movement. In addition, a relevant variation of the velocity profile was observed, moving it away from a perfect bell shape since TVS values were lower than 0.5, bringing out an anticipation of the peak velocity. On the contrary, *T* and *TVS* did not change in inversion-eversion movements, suggesting a lower perception of the force field in frontal plane. Finally, the presence of the force field dramatically affected *TPS* in both dominant and non-dominant limb, leading to a prevalence of inversion rotations in plantarflexion movements and eversion rotations in dorsiflexion ones, for the dominant limb, and vice versa for the non-dominant one. This result could be easily explained considering the direction of the applied force field that pushed toward right in plantarflexion movements and toward left in dorsiflexion ones.

As regards the stopping indices, the presence of the force field affected only the dispersion parameter, showing that the trajectory travelled after hitting the target was more spread in the unperturbed environment, even though the time needed to stop the foot did not change between the two phases. As already discussed before, in the unperturbed environment, subjects needed to perform a finer movement while moving upward in order to contrast the effect of the gravity force, whereas in dorsiflexion movement subjects let the foot go down to reach the target, with a consequent increase of the dispersion index. On the contrary, in the perturbed environment, subjects cannot drop the foot during the plantarflexion as in the familiarization phase. In fact, they needed to tune the movement to compensate the effect of the force field that pushed the foot rightward or leftward, thus resulting in a reduction of the dispersion.

Motor learning was evaluated in each movement direction by means of Learning Index. From our findings, it emerged the presence of learning in all movement directions except the inversion for the dominant limb and the eversion for the non-dominant one. In all movement directions for which the motor learning occurred, *LI* showed statistical differences between the first target set and the other ones, meaning that the motor learning arose in the initial part of the task, whereas its level remained unaltered during the remain part of the task. These findings are in line with those found by Stockinger*et al.* [2] in planar reaching movements performed by healthy subjects with the upper limb. The authors found that, at the beginning of the adaptation phase, the *LI* curve was characterized by a rapid increase that decayed with ongoing practice and finally reached a plateau. However, even if the trend of *LI* curves was similar, the values found in [2] were higher than those obtained in this study. Considering *LI* as a good measure of force field prediction [8], our findings showed a different capacity of the CNS in generating a feed forward control strategy when the movements are performed with the lower limbs. It suggests a lower ability of subjects in the accurate movement generation using a feed forward control strategy [2, 8]. A further relevant finding is the similar behavior of dominant and non-dominant limbs in motor learning as highlighted by the *LI* curve trends. This outcome could suggest that the side of the limb involved in the task does not affect the ability of the CNS to form a feed forward control strategy, which is responsible of motor learning [2].

Movement directions for which *LI* values showed no statistical differences among the target sets corresponded to leftward movements for both libs. It is worthy to notice that for these rotations the applied force field was directed opposite to gravity force. In particular, in the eversion (inversion) rotation performed with the dominant (non-dominant) limb, the gravity force acted in the same direction of the applied force field, increasing its magnitude. An opposite occurred for the inversion (eversion) rotation, with a consequent decrease of the force field intensity. It could suggest that the motor learning ability is not related to the movement direction, but it depends on the magnitude of the applied force field actually perceived by subjects. We should hypothesize that to elicit the CNS ability to form a feed forward control strategy, independently by the movement direction, the magnitude of the applied force field should be increased.

As a general consideration, it is possible to observe that the force field was perceived by subjects leading to a deterioration of the motor performance, especially in the rotations in the sagittal plane. This confirmed the anisotropic behavior of the ankle joint, since the presence of the force field does not affect all movement directions in the same way. In particular subjects were forced to directionally tune the control strategy of the ankle in order to perform straight trajectories. Actually, the anisotropic behavior of the ankle joint related to kinematic parameters do not involve in the learning index. In fact, the lack of motor learning in two rotations (inversion for the dominant limb and eversion for the non-dominant one) is due to reduced perception of the force field caused by the gravity force.

5. Conclusions

In this paper, a protocol aiming at the evaluation of ankle motor performance and motor learning during goal-directed tasks is proposed. Tasks were performed with both limbs in four different movement directions with and without the presence of a perturbing force field. Motor performance was quantified by means of a set of kinematic indices, whereas motor learning was evaluated through the Learning Index. The outcomes of the kinematic indices related to the task performed in the unperturbed environment showed that the best performance in term of accuracy and smoothness of the trajectories were obtained for the dorsiflexion movements in the sagittal pane and for the inversion rotations in the frontal plane. Moreover, numerous differences were found between movement directions, thus unveiling an anisotropy behavior of the ankle joint. As regard the motor learning, it emerged that, even though subjects can rapidly adapt to the perturbed environment, they showed a lower ability in the accurate movement generation using a feed forward control strategy, if compared to the upper limb. Furthermore, the motor learning ability resulted affected by the magnitude of the force field.

Future study will investigate the effect of different magnitude of perturbing force field to deeper investigate the motor learning process related to ankle joint movements.

Declarations

Author contribution statement

F. Martelli: conceived and designed the experiments; performed the experiments; analyzed and interpreted the data; wrote the paper.

E. Palermo: performed the experiments; analyzed and interpreted the data; wrote the paper.

Z. Del Prete: analyzed and interpreted the data; wrote the paper.

S. Rossi: conceived and designed the experiments; analyzed and interpreted the data; wrote the paper.

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Competing interest statement

The authors declare no conflict of interest.

Additional information

No additional information is available for this paper.

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