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# Aging effects on electrical and hemodynamic responses in the sensorimotor network during unilateral proximal upper limb functional tasks

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## A B S T R A C T

Healthy aging leads to poorer performance in upper limb (UL) daily living movements. Understanding the neural correlates linked with UL functional movements may help to better understand how healthy aging affects motor control. Two non-invasive neuroimaging methods allow for monitoring the movement-related brain activity: functional near-infrared spectroscopy (fNIRS) and electroencephalography (EEG), respectively based on the hemodynamic response and electrical activity of brain regions. Coupled, they provide a better spatiotemporal mapping. The aim of this study was to evaluate the effect of healthy aging on the bilateral sensorimotor (SM1) activation patterns of functional proximal UL movements. Twenty-one young and 21 old healthy participants realized two unilateral proximal UL movements during: i) a paced reaching target task and ii) a circular steering task to capture the speed-accuracy trade-off. Combined fNIRS-EEG system was synchronised with movement capture system to record SM1 activation while moving. The circular steering task performance was significantly lower for the older group. The rate of increase in hemodynamic response was longer in the older group with no difference on the amplitude of fNIRS signal for the two tasks. The EEG results showed aging related reduction of the alpha-beta rhythms synchronisation but no desynchronisation modification. In conclusion, this study un-covers the age-related changes in brain electrical and hemodynamic response patterns in the bilateral sensori-motor network during two functional proximal UL movements using two complementary neuroimaging methods. This opens up the possibility to utilise combined fNIRS-EEG for monitoring the movement-related neuroplasticity in clinical practice.

### Keywords:

Motor-control

Neuroplasticity

fNIRS

EEG

Reaching and tracing task

Brain asymmetry

## 1. Introduction

Decrease in motor performance with aging [1] alters more specifically the daily life tasks involving upper limb (UL) movements (e.g. hand control for reaching tasks; [2]). As the aging population is increasing, understanding the neural correlates linked with UL functional movements may help to better understand how healthy aging affects the level of brain activation and lateralization of the sensorimotor network related to the speed and accuracy of movement [3,4].

In healthy young adults, an asymmetry in SM1 activation patterns have been generally reported for unilateral distal UL movements, where the contralateral hemisphere is more activated than the ipsilateral one [5–7]. This brain asymmetry for UL movement control is understood in

the context of the greater proportion of the corticospinal tract emanating from the contralateral SM1 [8]. Nevertheless, both high handgrip strength, manual dexterity or task complexity can increase substantially the ipsilateral SM1 activation [6,9] and excitability [10], thus reducing the hemispheric asymmetry of the brain.

In healthy people, age-related physiological changes alter the neural circuitry at the structural level, especially in the SM1 [11], the cortico-striatal and the cortico-cerebellar networks [12], and at the functional level with regard to neuronal plasticity [13]. Several models have emerged in the literature to explain these modified brain reorganisation patterns regularly observed in healthy aging during both cognitive tasks and cognitive-locomotor dual tasks. Two major models (compensation-related utilization of neural circuits hypothesis

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-CRUNCH, [14]; and hemispheric asymmetry reduction in older adults -HAROLD, [15]) suggest in elderly additional neural resources to compensate when facing elevated task requirements due to reduced neural efficiency, and a higher recruitment in the opposite (ipsilateral) hemisphere for the same task, respectively. Hence, a more bilateral brain activation would occur. More recently, brain reorganisation was revealed with hyperactivation of the task-specific motor networks during distal hand-movements in elderly compared to young to maintain the grip performance [4]. In addition, producing unilateral fine hand motor tasks, such as finger tapping tasks, recruits more bi-hemispheric areas for older adults with a greater activation of SM1 [16–18]. With aging, functional movements involving proximal UL joints (shoulder, elbow) are more affected and could present modification in the speed and accuracy components of the movement [2]. Nevertheless, there is no information about the effects of healthy aging on changes in brain activation patterns of the sensorimotor network during unilateral multi-joint proximal UL movement tasks seen in daily life.

In order to evaluate sensorimotor network activation during functional proximal UL movements in real-world settings, two non-invasive neuroimaging methods can be used. First, functional near infrared spectroscopy (fNIRS) relying on the concentration changes in oxygenated (HbO<sub>2</sub>) and deoxygenated (HbR) haemoglobin at the cortical regions of interest, has been used to measure sensorimotor network activation during distal UL movements in healthy young adults [6, 19–21] and in older healthy adults [22,23]. A limited number of studies have investigated proximal UL movements, such as reaching movements. Although in healthy adults it has been shown by using fNIRS that the contralateral brain was mainly activated [20], a more bilateral activation pattern is observed in proximal compared to distal UL movements [24]. Secondly, electroencephalography (EEG) measures functional brain activity directly by detecting the variations of electrical currents at the scalp from local electric fields produced by neuronal activity [25]. Event-related power changes in the neural oscillatory activities contained in EEG signals have been used to decode the movement [26]. These power modifications are contained within specific frequency bands of the SM1 (alpha-mu – 8–13 Hz and beta – 14–29 Hz). These oscillations over the SM1 are characterized by a power decrease during movement execution (event-related desynchronization, ERD) while the resting period is characterized by a power increase (event-related synchronization, ERS), which respectively indicate the level of excitation and inhibition in the sensorimotor network [27]. Some studies have shown a bilateral alpha-mu and beta ERD during unilateral hand movements in young and older populations [26–28]. However, a limited number of EEG studies have investigated age related differences during UL movements and found controverted results for ERD [28,29] and for ERS [30,31].

Both fNIRS and EEG methods provide independent non-invasive and portable recordings of brain activation related to movement without any strong movement constraint [32]. Combining the two neuroimaging systems together on a single headset could provide better spatio-temporal information of the brain activation patterns of SM1 over the two hemispheres [21,33], with high temporal resolution of the neuronal and hemodynamic components [34]. The proposed fNIRS-EEG association could highlight the inter-hemispheric reorganisation of both SM1s and the evolution of event-related brain oscillations with aging during miscellaneous unilateral proximal UL movements.

Therefore, the main objective of the present study was to evaluate the effects of physiological aging on the electrical (EEG) and hemodynamic (fNIRS) responses over bilateral SM1 regions during unilateral proximal UL movements. More specifically, we aimed to unveil hemispheric differences in the level of brain hemodynamic and electrical oscillation patterns in young and older adults during unilateral multi-joint UL movements, one with no constraints on performance (paced reaching task) and one that has speed-accuracy constraints (circular steering task). We hypothesised that with aging, the bi-hemispheric electrical and hemodynamic responses during the two proximal UL

movement tasks will be greater. As a secondary objective we wanted to explore aging effects on the brain pattern related to the motor performance in the circular steering task. We expected a lower neural efficiency in the older group due to a greater bilateral brain activation pattern for maintaining reduced motor performance.

## 2. Material and methods

### 2.1. Participants

A total of 42 right-handed healthy adults were recruited and assigned to a young or older group of 21 participants (see Table 1). Participants were recruited through students mailing lists for the young group and via local associations for the older one. To be included, participants had to be aged either between 18 and 40 years old (Young group), either between 60 and 90 years old (Old group) and to be right-handed assessed by the Edinburgh Handedness Inventory [35]. Exclusion criteria were the existence of neurological (including a history of traumatic brain injury) or motor disorders at the level of the upper limb (history of tendinous disease, arthritis, surgery). All participants gave informed written consent before participating to the study. In accordance to the Declaration of Helsinki, the study was approved by the local Ethics Committee (EuroMov IRB, permit number 1912B).

### 2.2. Experimental design

All participants came to the laboratory for a single 1 h session. The experiment took place in a quiet isolated room. The participants were equipped with the fNIRS-EEG neuroimaging systems and then had to perform two functional UL tasks while seated, a paced reaching arm task and a circular steering task. The setup allowed for combined and synchronous recordings of UL kinematics and brain activity (fNIRS and EEG) using lab streaming layer (LSL, <https://github.com/labstreaminglayer/App-LabRecorder>; see *Functional motor tasks* section).

### 2.3. Functional motor tasks

More details about the methodology of the functional motor tasks are presented in a recent methodological paper (see figure 5 in [36]). The description of the two UL motor tasks in sitting position are as follows:

#### 2.3.1. Paced reaching task

Participants were seated on a chair with armrests and had to reach a target (a ball) in front of them, at 80 cm of height and at a distance allowing a complete extension of the right arm. A Kinect sensor (V2, Microsoft, United-States) was placed in front of them at 1.70 m height and at 1.60 m from the target. Five free movements per block of 20 s were paced by the voice (“go”; “stop”). After one block of practice with each hand, they had to perform the paced reaching task for 3 blocks with their non-dominant hand and then for 3 blocks with their dominant hand. Each block was interspaced by ~20 s of rest. Then, participants performed the same task for 3 blocks with each hand in a movement-constrained condition, where their shoulders were held fixed to limit trunk tilting movements. Participants were requested to minimize head movement throughout the protocol. This protocol with a constrained reaching task condition was applied to fit with our clinical trial in stroke

**Table 1**  
Characteristics of the participants for each group (n = 21).

Characteristics	Young group	Old group
Age (years) (SD)	25.1 ( ± 4.7)	73.1 ( ± 6.7)
Sex (female/male)	11/10	11/10
Handedness score (SD)	0.94 ( ± 0.14)	0.96 ( ± 0.08)

Groups handedness score was compared using unpaired Student t-tests; there was no significant difference.

patients (<https://clinicaltrials.gov/ct2/show/NCT04291573>) to determine the effects of trunk compensation. In the current study, since there were no significant differences between reaching task conditions for either healthy group, only the results of the free movement condition are reported.

### 2.3.2. Circular steering task

Participants were seated on a chair in front of a graphic tablet (A3 size; Wacom, Kazo, Japan) with a stylus moulded onto a mouse pad. The task consisted in a circular steering task based on the speed-accuracy trade-off [37]. The task was delivered on the computer using a lab-made software, the LSL-Mouse (<https://github.com/KarimaBak/LSL-Mouse>). The software displayed on a 24-inch screen a circular target (33-inch circumference) with a 2 cm tunnel in which the participants had to move a cursor as fast as possible in a clock-wise direction during blocks of 20 s. Blocks of training were done in order to familiarize with the task requirement providing verbal encouragement to focus on the speed of the task. Participants had to speed up if errors (any time outside the 2 cm circle tunnel boundaries) were less than 15% (based on pilot testing) during the training. During the experimental task, no more instructions were given. The task then consisted on 3 blocks for each arm (20 s of task with 20 s of rest). Participants started first with their non-dominant hand.

### 2.4. Brain activity

During the whole session, participants were equipped by a customized EEG-fNIRS headcap with an inter-optode distance of ~3 cm allowing to record the left and right sensorimotor cortical regions during the 2 functional motor tasks (see Fig. 1). EEG and fNIRS signals were simultaneously measured using a wireless Starstim fNIRS integration system (Starstim8, Neuroelectrics, Barcelona, Spain; Octamon+, Artinis Medical Systems, the Netherlands). EEG and fNIRS signals were recorded for approximately 10 min for each task. The Brodmann's areas covered by the different fNIRS channels were extracted via the NFRI function [38] from the MNI (Montreal Neurological Institute) coordinates (Table 2). Cartesian coordinates were obtained with a 3-dimensionnal digitizer (Patriot®, Polhemus Inc., USA) allowing the surface locations of the fNIRS optodes and EEG electrodes. The

**Table 2**

Locations of fNIRS probes and EEG electrodes over the sensorimotor cortices. On the left, combination receiver (Rx) and transmitter (T) label for the probes and selected electrodes of the international 10–10 EEG system. On the middle, MNI coordinates for each fNIRS and EEG channel (n = 16) with x, y, and z coordinates. On the right, Brodmann area (BA) correspondence (number, name and %) extracted from the NFRI function. R and L are for right and left hemispheres, respectively.

fNIRS EEG Characteristics	MNI coordinates			BA
	X	Y	Z	
fNIRS – Rx1-Tx1 R	26	-5	90	6, 4
fNIRS – Rx1-Tx2 R	45	-9	81	6, 4
fNIRS – Rx1-Tx3 R	44	-27	85	4, 6, 3, 1
fNIRS – Rx1-Tx4 R	24	-23	94	4, 6, 3
fNIRS – Rx2-Tx5 L	-24	-3	90	6, 4
fNIRS – Rx2-Tx6 L	-48	-7	79	6, 4
fNIRS – Rx2-Tx7 L	-43	-24	87	4, 6, 3, 1
fNIRS – Rx2-Tx8 L	-24	-20	94	4, 6, 3
EEG – C4 R	58	-18	72	6, 3, 2, 1
EEG – FC2 R	24	27	77	6, 8
EEG – FC6 R	69	14	38	6, 9
EEG – CP2 R	26	-48	93	7, 5, 3, 4
EEG – C3 L	-62	-25	71	4, 2, 1
EEG – FC1 L	-38	-1	83	6, 4
EEG – FC5 L	-76	-13	46	6, 2, 1
EEG – CP1 L	-31	-65	86	7, 5

BA: 1,2,3 (primary somatosensory cortex); 4 (primary motor cortex); 5 (somatosensory association cortex); 6 (premotor and supplementary motor cortex); 7 (parietal cortex); 8 (frontal eye-field); 9 (frontal cortex).

positioning of the 16 channels constituted 4 NIRS and 4 EEG channels per hemisphere. The arrangement is presented on Fig. 1. After being equipped, participants were asked to perform a wrist extension task to check if the movement produced was inducing a hemodynamic response.

#### 2.4.1. Electroencephalography – EEG

The EEG used was an 8-channels design. Electrodes were located on and around SM1 cortices: right hemisphere (C4, FC2, FC6, CP2) and left hemisphere (C3, FC1, FC3, CP1) according to the international 10–10 EEG system. The electrodes (NG Geltrode, Neuroelectrics) were filled up with electro-gel (Signa Gel®). References electrodes (CMS, DRL) were placed over the right earlobe using an earclip. The EEG spectrum was sampled at 500 Hz. The EEG device was connected using WIFI and controlled by the computer through a software interface (Neuroelectrics Instrument Controller, NIC v 2.0).

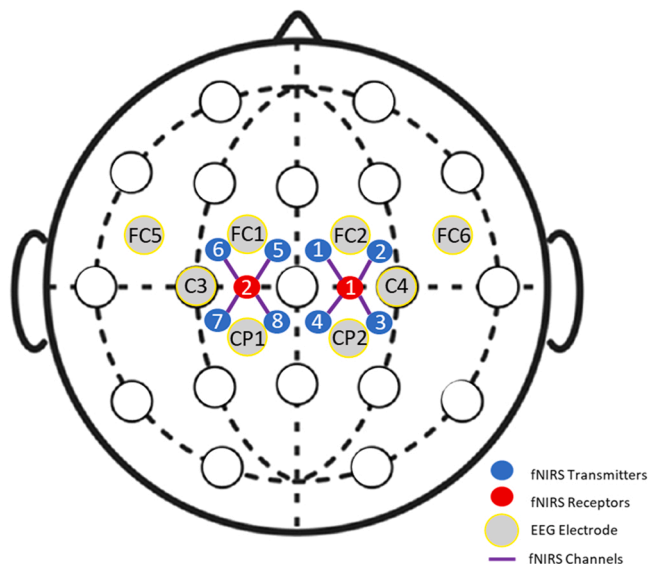
#### 2.4.2. Functional near infrared spectroscopy – fNIRS

A continuous-wave fNIRS system using two wavelengths was used for measuring the changes in HbO<sub>2</sub> and HbR overlying the left and right SM1 at a sampling frequency of 10 Hz. The optodes (2 receivers -light-detection probe- and 8 transmitters -light-emission probe- resulting in a total of 4 channels per hemisphere, Fig. 1) were placed securely on the participant's head using a fit neoprene cap to get a stable optical contact with the scalp by generating an equal amount of pressure. The two receivers were placed on C1 and C2 locations of the 10–10 EEG system. The respective four transmitters were located at a distance of 3 cm from the receivers thanks to plastic holders. The fNIRS device was connected using Bluetooth and controlled through a software interface (Oxysoft, v 3.2.51.4).

### 2.5. Data analysis

#### 2.5.1. Task performance

For the reaching task, the analysis of UL reaching kinematics was allowed by an LSL-Kinect software (LSL-KinectV2: <https://github.com/KarimaBak/LSL-KinectV2>; [39,40]). All the participants correctly reached the target and the mean peak of velocity of the hand to reach the



**Fig. 1.** Placement of the 8 fNIRS and 8 EEG channels covering the bilateral sensorimotor cortices represented on the international 10–10 EEG system layout (see Table 1 for MNI locations). In grey, the 8 EEG electrodes. For the fNIRS layout, the blue circles are the 8 transmitters and the red circle are the 2 receivers (1: right hemisphere; 2: left hemisphere).

ball (mm/s) was computed. For the kinematics analysis, 5 participants during Kinect recording of the paced reaching task, were unusable.

For the circular steering task, in order to quantify the speed-accuracy trade-off, an index of effective performance (IPE, [41]) was calculated. IPE (bits/s) was defined as function of an index of effective task difficulty (IDE):  $IDE = \frac{2\pi Re}{We}$ , with IDE in bits, Re the mean radius of the circular

drawing and We the effective target width path  $We = 4.133 * \sigma$ , [42] where  $\sigma$  is the standard-deviation of the radius);  $IPE = \frac{IDE}{MT}$ , with IPE in

bits/s, and MT, movement time in seconds. The speed (number of laps per seconds) and accuracy (percentage spent inside the 2 m tunnel) of the movement were also computed. For each participant and each arm, the median of the 3 blocks for each parameter was assessed.

### 2.5.2. Brain activity (fNIRS and EEG)

All fNIRS raw data analysis was undertaken using HOMER toolbox on MATLAB (Homer2 NIRS processing package, [43]) using the generated files of the Lab Recorder (xdf files). Following previously used pipeline [44], optical density time series were first converted into relative concentrations of HbO<sub>2</sub> and HbR using the modified Beer-Lambert law corrected by a differential pathlength factor depending on the age of the participant ([45], function hmrOD2Conc of Homer2). All data were band-pass filtered (Low cutoff: 0.08- high cutoff: 0.009 Hz; function hmrBandpassFilt of Homer2) to eliminate physiological confounds such as low-frequency drift, Mayer-wave systemic oscillations, breathing rate and heart rate components [46]. Onset of each movement and rest periods were then identified on the fNIRS signals. Motion artifacts were visually checked and removed manually when a sudden change of HbO<sub>2</sub> and HbR traces were observed. The corresponding epoch was then excluded. With visual rejection and outliers' detection, 14% of the fNIRS signals were rejected for the Young group, and 8% for the Older group. Each block of 5 movements was then corrected with a zero-baseline value based of the onset of the first movement. The relative concentration changes of HbO<sub>2</sub> and HbR were computed within the motor task intervals (0–20 s) to reflect the immediate hemodynamic response [47]. After having verified the neurovascular coupling occurrence during functional movement (i.e., rapid increase in HbO<sub>2</sub> with a lower-amplitude reduction or flat signals in HbR), the value and latency of the larger variation (peak), the average and the area under the curve values were computed for each block of 20 s, each participant and for each haemoglobin species. Given the findings were similar among the three measures of magnitude (peak, average and area under the curve), and that HbO<sub>2</sub> is most sensitive for motor task-related hemodynamic changes, the relative changes in peak HbO<sub>2</sub> concentration were employed as the indicator of brain activity; the time to peak HbO<sub>2</sub> concentration is also presented. These parameters were averaged by tasks (paced reaching; circular steering) and hand condition (dominant; non-dominant).

All EEG data analysis was undertaken using EEGLAB toolbox on MATLAB ([48], version 2021.1) using the generated files of the Lab Recorder (xdf files). First, data were converted from nanovolts to microvolts and re-referencing with the channel-average method and the signal was resampled at 250 Hz. It was then filtered using a band-pass filter between 0.3 and 50 Hz. Due to EEG-device troubles during acquisition, 4 participants of the Older group on the reaching task and 2 on the circular steering task were excluded. All signals were manually reviewed, those affected by noisy channels, noisy trials, and trials with abnormal movement were rejected. With visual inspection and outliers' detection, for the Young group, 2% of the signals were rejected and 6% for the Older group. Onset of each movement and rest periods were then detected on the signal. The event-related spectral perturbations (ERSP) in the alpha (8–13 Hz) and beta (14–29 Hz) rhythms were calculated as a visualisation of the average power changes in these specific time-frequencies. They provide details about the event-related desynchronization (ERD; percentage of power decrease in a specific frequency band relative to the baseline – i.e., rest) and synchronisation

(ERS; percentage of power increase in a specific frequency band relative to the task). A baseline correction was applied by subtracting the mean power changes in a 1 s baseline period from each spectral estimation, to normalize the ERSF.

To quantitatively evaluate the ERSF patterns for the different UL motor tasks, the relative power using the instantaneous power spectrum (Pn), the mean power spectra (Prest) during the rest period (Trest) and the mean power spectra (Ptask) during the task period (Ttask) were computed as follows [26]:

$$P_{rest} = \frac{1}{|T_{rest}|} \sum_{n \in T_{rest}} P_n,$$

$$P_{task} = \frac{1}{|T_{task}|} \sum_{n \in T_{task}} P_n,$$

$$RP(n) = \frac{P_n - P_{rest}}{P_{rest}} \times 100$$

$$RP = \frac{P_{task} - P_{rest}}{P_{rest}} \times 100$$

Mean ERSFs were calculated and averaged by tasks (paced reaching and circular steering) and hand conditions (dominant and non-dominant) during rest and movement periods.

$$RP = \frac{P_{rest} - P_{task}}{P_{task}} \times 100$$

## 2.6. Statistical analyses

Statistical analyses were performed using R version 3.6.1 and RStudio version 2022.07.06. Parametric tests were used as the data were normally distributed after using the Shapiro-Wilk test and by creating Q-Q plots. Effect sizes were reported with the partial eta square ( $\eta^2_p$ ) for the effects of the ANOVAs [49], considering a small (0.02), medium (0.13) and large (0.26) effect [50]. Statistical significance was set at  $p < .05$ . Significant observations were interpreted when  $\eta^2_p > .02$ . All values are reported mean (SD) unless specified otherwise. No three-level interaction effects were found, so only two-level interaction effects were reported with each factor combinations. For significant interactions, post-hoc Tukey comparisons were performed. Note that degrees of freedom of the analysis are different as there were not always the same number of subjects excluded for the different variables.

A mixed ANOVA with age (young and old) as a between-subject factor, and hand (dominant and non-dominant hand) as within-subject factors was performed on the movement parameters (circular steering task: IPE, speed and accuracy; paced reaching task: velocity peak).

A mixed ANOVA with age (young and old) as a between-subject factor, and hand (dominant and non-dominant hand) and hemisphere (ipsilateral and contralateral to the movement) as within-subject factors was performed on the fNIRS  $\Delta$ HbO<sub>2</sub> parameters (peak, time to peak) and on the EEG Alpha and Beta ESRFs.

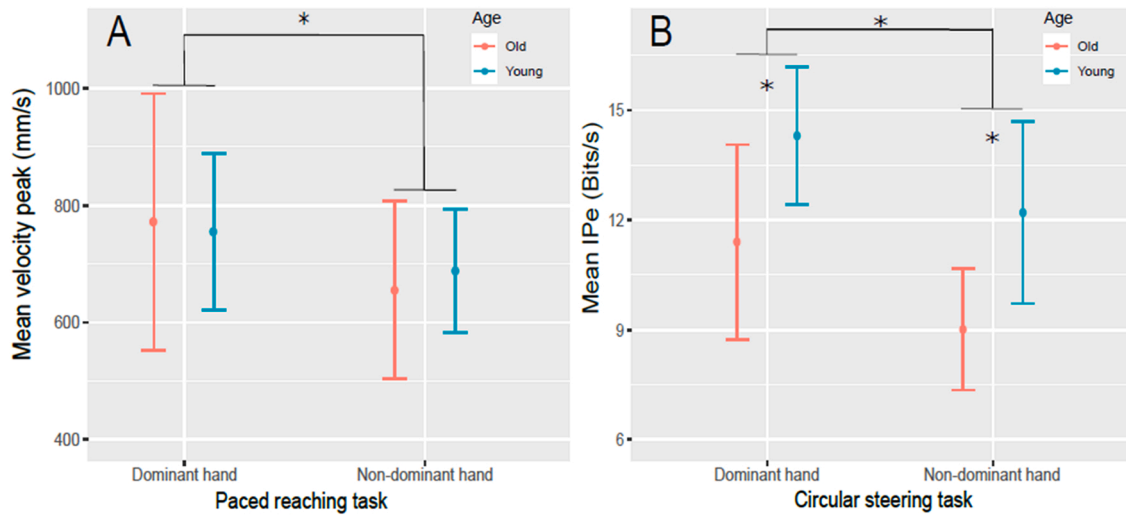
In order to explore the relationship between performance of circular steering task and brain activation (fNIRS peak  $\Delta$ HbO<sub>2</sub>) according to the age groups, an efficiency ratio (eR = (Performance - brain activation) / (Performance + brain activation) \* 100) was calculated. An independent t-test with Age (young and old) as factor was performed on eR. Finally, a spearman rank correlation analysis was applied for both age groups independently between motor performance indices of the circular steering task and brain activation parameters.

## 3. Results

### 3.1. Tasks performance (Fig. 2)

On the paced reaching task, there was no Age effect ( $F_{(1,35)} = 0.027$ ,  $p = .871$ ,  $\eta^2_p = .00$ ) or Age x Hand interaction ( $F_{(1,35)} = 1.71$ ,  $p = .199$ ,





**Fig. 2.** Upper limb motor task performances (mean  $\pm$  SD) for the two age groups and according to the hand dominance. A) Velocity peak differences for hand dominance on the paced reaching task, B) Index of performance (IPe) differences on the circular steering task for the young and older groups and hand dominance (\* for statistically significant differences at  $p < .05$ ).

$\eta^2_p = .05$ ). Nevertheless, we found that for both groups the mean velocity peak was higher with the dominant hand than with the non-dominant one (Hand:  $F_{(1,35)} = 23.59$ ,  $p = .000$ ,  $\eta^2_p = .40$ ).

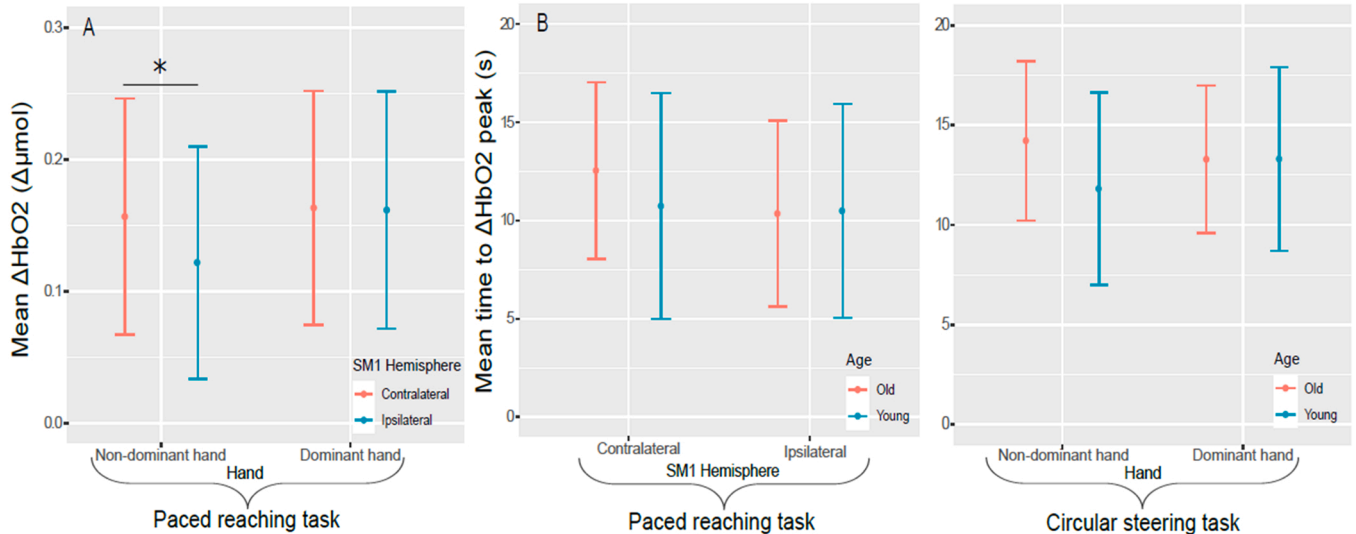
We found a higher performance (IPe) in the young group and with the dominant hand (Age:  $F_{(1,40)} = 26.79$ ,  $p = .000$ ,  $\eta^2_p = .40$ ; Hand:  $F_{(1,40)} = 45.87$ ,  $p = .000$ ,  $\eta^2_p = .53$ ) with no Age  $\times$  Hand interaction ( $F_{(1,40)} = 0.19$ ,  $p = .664$ ,  $\eta^2_p = .00$ ). For the speed component, we found higher values in the young group and with the dominant hand (Age:  $F_{(1,37)} = 6.71$ ,  $p = .014$ ,  $\eta^2_p = .15$ ; Hand:  $F_{(1,37)} = 25.55$ ,  $p = .000$ ,  $\eta^2_p = .41$ ), but no Age  $\times$  Hand interaction ( $F_{(1,37)} = 0.41$ ,  $p = .241$ ,  $\eta^2_p = .04$ ). For the accuracy component, we did not find any significant main or interaction effects (Age:  $F_{(1,40)} = 2.23$ ,  $p = .136$ ,  $\eta^2_p = .06$ ; Hand:  $F_{(1,40)} = 1.17$ ,  $p = .29$ ,  $\eta^2_p = .03$ ; Age  $\times$  Hand interaction:  $F_{(1,40)} = 0.06$ ,  $p = .813$ ,  $\eta^2_p = .00$ ).

### 3.2. Brain activity

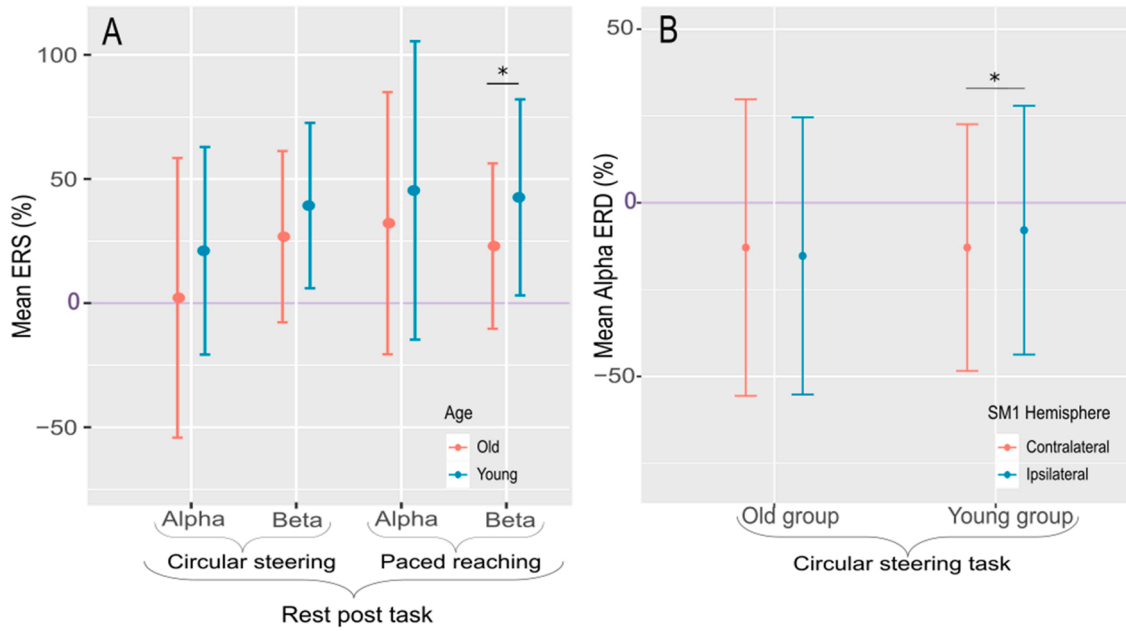
Brain activity (fNIRS: peak of  $\Delta\text{HbO}_2$  and time to peak; EEG: ERD and ERS) during paced reaching and circular steering tasks are presented in Fig. 3 (fNIRS) and Fig. 4 (EEG) and the statistical results are detailed in Table 3.

The analysis of the mean  $\Delta\text{HbO}_2$  peak on the paced reaching task showed there was no main effect of Age. The Hand  $\times$  Hemisphere interaction (complemented with Tukey's post-hoc mean comparisons) revealed a higher activation in the contralateral side compared to the ipsilateral one, but only when the task was performed with the non-dominant hand. On the same task, the analysis of the time to peak  $\Delta\text{HbO}_2$  showed a significant Age  $\times$  Hemisphere interaction. Tukey's post-hoc mean comparisons revealed that for the older group, there was a Hemisphere effect with a longer time to peak  $\Delta\text{HbO}_2$  in the contralateral than ipsilateral side.

The analysis of the mean  $\Delta\text{HbO}_2$  peak on the circular steering task



**Fig. 3.** fNIRS main results. A) fNIRS-derived mean ( $\pm$  SD)  $\Delta\text{HbO}_2$  peak in the contralateral (red) and ipsilateral (blue) sensorimotor (SM1) hemisphere during paced reaching task as a function of hand used (no significant Age effects, only a Hand  $\times$  hemisphere interaction, \*  $p < .05$ , see Table 3). B) fNIRS-derived mean ( $\pm$  SD) time to peak  $\Delta\text{HbO}_2$  for old (red) and young (blue) age group as a function of SM1 hemisphere (paced reaching task Age  $\times$  Hemisphere interaction) or hand used (circular steering task Age  $\times$  Hand interaction).



**Fig. 4.** EEG main results. A) Mean ( $\pm$  SD) event related synchronization (ERS) in Alpha (8–13 Hz) and Beta (14–29 Hz) power during the post task resting periods. B) Mean ( $\pm$  SD) event related desynchronization (ERD) in Alpha power for the contralateral (in red) and ipsilateral (in blue) sensorimotor (SM1) hemispheres during the circular steering task. (\* for statistically significant difference at  $p < .05$ , see Table 3).

showed there were no main effects of Age, Hemisphere, or Hand nor any Interactions. The analysis of the time to peak  $\Delta$ HbO<sub>2</sub> showed an Age x Hand interaction. Post-hoc comparison revealed that the time to peak for the dominant hand, in comparison to the non-dominant one, was

shorter for the young group and longer for the older group.

The analysis of the mean ERSP on the paced reaching task showed a main effect of Age on the Beta ERS with a higher post movement synchronisation of the younger group. No Age Hand, Hemisphere or

**Table 3**

Statistical results of the ANOVA on the EEG and fNIRS brain parameters for the paced reaching and circular steering tasks.

	A.	AGE	HEMISPHERE	HAND
	<i>EEG</i>			
<i>Paced reaching task</i>	Alpha ERD	$F_{(1,34)} = 2.09, p = .158, \eta^2_p = .06$	$F_{(1,34)} = 0.03, p = .871, \eta^2_p = .00$	$F_{(1,34)} = 0.82, p = .371, \eta^2_p = .02$
	Beta ERD	$F_{(1,34)} = 0.52, p = .476, \eta^2_p = .02$	$F_{(1,34)} = 0.30, p = .586, \eta^2_p = .01$	$F_{(1,34)} = 0.77, p = .767, \eta^2_p = .00$
	Alpha ERS	$F_{(1,33)} = 2.50, p = .123, \eta^2_p = .07$	$F_{(1,33)} = 0.39, p = .538, \eta^2_p = .01$	$F_{(1,33)} = 1.69, p = .203, \eta^2_p = .05$
	Beta ERS	<b><math>F_{(1,29)} = 4.60, p = .041, \eta^2_p = .14</math></b>	$F_{(1,29)} = 1.62, p = .213, \eta^2_p = .05$	$F_{(1,29)} = 0.37, p = .549, \eta^2_p = .01$
	<i>fNIRS</i>			
	$\Delta$ HbO <sub>2</sub> peak	$F_{(1,35)} = 0.94, p = .338, \eta^2_p = .03$	<b><math>F_{(1,35)} = 20.88, p = .000, \eta^2_p = .37</math></b>	$F_{(1,35)} = 1.05, p = .313, \eta^2_p = .03$
	Time to peak	$F_{(1,38)} = 2.99, p = .092, \eta^2_p = .07$	<b><math>F_{(1,38)} = 18.56, p = .000, \eta^2_p = .33</math></b>	$F_{(1,38)} = 2.01, p = .164, \eta^2_p = .05$
	<i>EEG</i>			
<i>Circular steering task</i>	Alpha ERD	$F_{(1,31)} = 0.18, p = .672, \eta^2_p = .01$	$F_{(1,31)} = 0.41, p = .526, \eta^2_p = .01$	$F_{(1,31)} = 0.94, p = .341, \eta^2_p = .03$
	Beta ERD	$F_{(1,31)} = 0.43, p = .517, \eta^2_p = .01$	$F_{(1,31)} = 0.04, p = .838, \eta^2_p = .00$	$F_{(1,31)} = 0.24, p = .625, \eta^2_p = .01$
	Alpha ERS	$F_{(1,33)} = 4.10, p = .051, \eta^2_p = .11$	$F_{(1,33)} = 0.02, p = .882, \eta^2_p = .00$	<b><math>F_{(1,33)} = 7.00, p = .012, \eta^2_p = .18</math></b>
	Beta ERS	$F_{(1,33)} = 2.44, p = .128, \eta^2_p = .07$	$F_{(1,33)} = 1.98, p = .169, \eta^2_p = .06$	<b><math>F_{(1,33)} = 4.88, p = .034, \eta^2_p = .13</math></b>
	<i>fNIRS</i>			
	$\Delta$ HbO <sub>2</sub> peak	$F_{(1,34)} = 0.00, p = .995, \eta^2_p = .00$	$F_{(1,34)} = 0.08, p = .774, \eta^2_p = .00$	$F_{(1,34)} = 0.36, p = .552, \eta^2_p = .01$
	Time to peak	$F_{(1,37)} = 1.42, p = .242, \eta^2_p = .03$	$F_{(1,37)} = 1.47, p = .233, \eta^2_p = .03$	$F_{(1,37)} = 0.17, p = .683, \eta^2_p = .00$
	<b>B.</b>			
	AGE x HEMISPHERE		AGE x HAND	
	<i>EEG</i>			
<i>Paced reaching task</i>	Alpha ERD	$F_{(1,34)} = 0.76, p = .391, \eta^2_p = .02$	$F_{(1,34)} = 0.11, p = .742, \eta^2_p = .00$	$F_{(1,34)} = 0.91, p = .346, \eta^2_p = .03$
	Beta ERD	$F_{(1,34)} = 3.99, p = .054, \eta^2_p = .11$	$F_{(1,34)} = 0.00, p = .963, \eta^2_p = .00$	$F_{(1,34)} = 0.00, p = .982, \eta^2_p = .00$
	Alpha ERS	$F_{(1,33)} = 0.81, p = .376, \eta^2_p = .02$	$F_{(1,33)} = 1.11, p = .299, \eta^2_p = .03$	$F_{(1,33)} = 0.44, p = .510, \eta^2_p = .01$
	Beta ERS	$F_{(1,29)} = 0.92, p = .346, \eta^2_p = .03$	$F_{(1,29)} = 0.12, p = .727, \eta^2_p = .00$	$F_{(1,29)} = 3.57, p = .069, \eta^2_p = .11$
	<i>fNIRS</i>			
	$\Delta$ HbO <sub>2</sub> peak	$F_{(1,35)} = 0.04, p = .849, \eta^2_p = .00$	$F_{(1,35)} = 0.01, p = .919, \eta^2_p = .00$	<b><math>F_{(1,35)} = 9.78, p = .004, \eta^2_p = .22</math></b>
	Time to peak	<b><math>F_{(1,38)} = 8.95, p = .005, \eta^2_p = .19</math></b>	$F_{(1,38)} = 0.00, p = .948, \eta^2_p = .00$	$F_{(1,38)} = 0.04, p = .847, \eta^2_p = .00$
	<i>EEG</i>			
<i>Circular steering task</i>	Alpha ERD	<b><math>F_{(1,31)} = 7.25, p = .011, \eta^2_p = .18</math></b>	$F_{(1,31)} = 0.26, p = .617, \eta^2_p = .01$	$F_{(1,31)} = 1.21, p = .279, \eta^2_p = .04$
	Beta ERD	$F_{(1,31)} = 0.33, p = .570, \eta^2_p = .01$	$F_{(1,31)} = 0.73, p = .399, \eta^2_p = .02$	$F_{(1,31)} = 0.02, p = .879, \eta^2_p = .00$
	Alpha ERS	$F_{(1,33)} = 0.01, p = .945, \eta^2_p = .00$	$F_{(1,33)} = 0.00, p = .983, \eta^2_p = .00$	$F_{(1,33)} = 0.00, p = .978, \eta^2_p = .00$
	Beta ERS	$F_{(1,33)} = 0.85, p = .364, \eta^2_p = .03$	$F_{(1,33)} = 0.06, p = .814, \eta^2_p = .00$	$F_{(1,33)} = 0.69, p = .412, \eta^2_p = .02$
	<i>fNIRS</i>			
	$\Delta$ HbO <sub>2</sub> peak	$F_{(1,34)} = 0.00, p = .984, \eta^2_p = .00$	$F_{(1,34)} = 0.00, p = .997, \eta^2_p = .00$	$F_{(1,34)} = 3.13, p = .086, \eta^2_p = .08$
	Time to peak	$F_{(1,37)} = 0.01, p = .928, \eta^2_p = .00$	<b><math>F_{(1,37)} = 5.66, p = .023, \eta^2_p = .13</math></b>	$F_{(1,37)} = 1.29, p = .263, \eta^2_p = .03$

A. Effects of AGE (Young vs Old), Hemisphere (Contralateral vs Ipsilateral) and Hand (Dominant vs Non-dominant) on the Alpha and Beta ERSP. B. Two-levels interaction of the model. In bold, results with  $p < .05$  and  $\eta^2_p > .02$ .

Interaction effect were found on the other EEG dependant variables (Alpha ERS, Alpha ERD, Beta ERD, see Table 3).

The analysis of the mean ERSP on the circular steering task showed no main effect of Age on the Beta and Alpha power modifications. However, there was an Age x Hemisphere interaction on the Alpha ERD. Post-hoc comparisons showed that Alpha ERD was significantly greater in the contralateral than ipsilateral sensorimotor area for the young group ( $p = .012$ ), with no significant differences between hemispheres for the older group ( $p = .197$ ). The results also showed a main effect of the Hand on the Alpha and Beta ERS, with a higher synchronisation after movements with the dominant than non-dominant hand.

Finally, looking at the difference between the changes in brain activity (fNIRS peak O2Hb) and circular task performance (Ipe), we found that the eR was greater for the young group in comparison to the older group ( $F_{(1,37)} = 5.48$ ,  $p < .05$ ,  $\eta^2_p = .034$ ). Moreover, we found a positive correlation between the accuracy component of the circular steering task and the level of activation in fNIRS for both groups (ipsilateral:  $p < .05$ ,  $\rho = .29$ ; contralateral:  $p < .05$ ,  $\rho = .26$ ). Nevertheless, we did not find any correlation between the movement parameters and the ERD-ERS.

#### 4. Discussion

This study is the first to investigate the age-related changes in the hemodynamic and electrical responses of the ipsilateral and contralateral sensorimotor areas in relation to unilateral proximal UL motor performance. For that purpose, we recorded the brain and the UL movements synchronously using combined fNIRS/EEG neuroimaging and kinematics methods. Two functional proximal UL tasks encountered in daily life were explored: a paced reaching target task and a circular steering task to capture the speed-accuracy trade-off. The primary aim was to determine potential hemispheric and hand dominance differences of the brain hemodynamic (fNIRS) and electrical (EEG) activity during proximal UL motor tasks with physiological aging. The fNIRS time course analysis (time to peak  $\Delta\text{HbO}_2$ ) showed aging effects on the rate of the neurovascular coupling response in the bilateral SM1 during both proximal UL tasks depending on the hemisphere and on the hand used. The Alpha ERD showed a greater contralateral to ipsilateral Alpha ERD for the young group in the circular steering task. Moreover, aging led to smaller power in the Beta ERS of the paced-reaching task; a similar trend was observed for Alpha ERS for circular steering task. The secondary aim was to explore the effect of aging on the relationship between motor performance and brain related activity in order to characterise the neural efficiency of movement. Looking to the relationship between brain pattern and motor performance, even though the fNIRS peak  $\text{HbO}_2$  and EEG Alpha and Beta ERD were similar between age groups during the circular steering task, performance (Ipe and speed) and efficiency ratio ( $eR = \text{Ipe}/\text{Peak } \Delta\text{HbO}_2$ ) were lower in the older group.

Altogether, our study showed some age-related differences in fNIRS and EEG parameters. On the paced reaching task, with similar fNIRS amplitude and performance, we found that with physiological aging there was a delayed time to peak  $\Delta\text{HbO}_2$  amplitude in the contralateral than ipsilateral hemisphere. On the circular steering task, we found a delayed time to peak  $\Delta\text{HbO}_2$  amplitude in the bilateral sensorimotor network when the task was performed with the non-dominant hand but not with the dominant hand. These results can be explained by a reduction of the speed of increase in cerebral blood flow following a neural activation causing an impaired neurovascular coupling response [51]. In addition, the results provided by EEG Alpha ERD suggest an aging related modification of excitatory-inhibitory capacity in the bilateral sensorimotor network, such that while the young group showed a greater Alpha ERD in the contralateral than ipsilateral hemisphere, the older subjects required similar levels of Alpha ERD in the bilateral sensorimotor network. This suggests that a greater excitatory capacity (Alpha ERD) was necessary from the ipsilateral sensorimotor network

for the older group to achieve this circular steering task performance. This could be explained by the reduction of interhemispheric inhibition present in older adults that is leading to an increase bilateral activity [52]. Moreover, older subjects presented a lower Beta (paced reaching) and Alpha ERS (trend for the circular steering,  $p = 0.051$  and  $\eta^2_p = .11$ ) traducing a reduced ability to disengage the activated neural circuits at the end of the tasks. Looking to the literature around the effect of aging, the ERS analysis is presenting contrasting results. Some teams showed a reduced ERS with aging [30] when others did not observe effects of aging on the magnitude of ERS [31]. Our results on EEG are thus, giving complementary results to the already existing contrasting literature of aging. Finally, this potential age-related decrease in ERS could be a key EEG parameter to understand the performance reduction on the circular steering task.

Overall, these novel findings suggest that physiological aging might reduce the neural efficiency to perform functional proximal UL movements, particularly when performance (speed-accuracy) needs to be taken into consideration. Herein, older participants showed a reduced motor performance and efficiency ratio. In addition, the older group had a slower neurovascular coupling response (i.e., a longer fNIRS time to peak  $\Delta\text{HbO}_2$ ) and reduced ability to disengage (inhibit) movement related neural circuits (i.e., EEG ERS) in the bilateral sensorimotor areas. These age-related modifications in bilateral sensorimotor network activation coupled with lower performance (circular steering task: Ipe, Speed) and efficiency ratio (eR), could suggest that the ability to efficiently recruit the bilateral sensorimotor network to perform the proximal UL tasks was lower for the old group compared to the young group.

Regarding our main fNIRS outcome, we did not find any Age x Hemisphere effect on both tasks and difference in the level of activation. Our results differ from previous studies that have highlighted aging-related differences in asymmetry of brain activity in distal UL tasks and dual tasks, such as dual-task walking and hand motor tasks with greater dexterity [4,16], and most notably during purely cognitive tasks requiring executive function [15]. Nevertheless, we did find an age x hemisphere effect with the EEG. Knowing the ability of the EEG to differentiate the excitatory state to the inhibitory one, we can hypothesize that in this task, to reach the higher performance required, the older subjects were over-activating their bilateral SM1. It is nevertheless really complicated to compare our results to previous studies, because none has looked to the Age x Hemisphere interaction during proximal UL movements. An explanation to this difference of age effect could also be in the area covered. Possibly, as the motor network involved in motor action is including more brain areas with aging, the controversial results in the literature could be explained by the brain areas explored [53]. For instance, as the circular steering task is involving a cognitive control, it could have been interesting to look to the parietal and frontal areas [54]. However, since our fNIRS-EEG measurement zone did not extend to more upstream motor control centres in the prefrontal cortex, any increases in the amplitude and/or area of activation to compensate for the reduced inhibitory capacity of the sensorimotor network during resting periods in older adults could not be verified. Although previous studies have confirmed such an extended prefrontal activation pattern to compensate for inefficient neural processing in sensorimotor circuits in older population [17,55], future studies with a more extended prefrontal region measurement should be undertaken to confirm such a mechanism in our experimental protocol.

Looking to the specific brain pattern of the tasks, we found that peak  $\Delta\text{HbO}_2$  and EEG ERD/ERS parameters on the two unilateral proximal UL movements evaluated showed a relative bilateral SM1 activation for most of the task conditions. This could be an explanation for the absence of aging effect on the asymmetry of SM1 activation as the younger subjects already yield, on these tasks, a bilateral SM1 activation for most of the conditions. The more proximal nature of the UL movement (mainly involving the elbow and shoulder joints) to perform the paced reaching and circular steering tasks already reduces any asymmetry of lateralised sensorimotor network activation because the elbow and



shoulder joints classically involve more ipsilateral activations than distal movements [24].

For the paced reaching task there were no overt difficulty constraints, and in the literature the modification of brain activation for motor control in aging is linked to the evolution of performances [14, 56]. Moreover, studies that have looked to the brain asymmetry evolution with aging on reaching tasks used a speed component for the movement [4,20]. However, our reaching task was paced with a 2 s period to reach the target and 2 s of rest back to the sitting position. Regarding the similar strategy used by all participants (i.e., faster with the dominant hand), a similar more bilateral activation pattern was required for both groups, which can explain the absence of further asymmetry reduction of brain activation in the older population.

The second movement explored, the circular steering task has a higher speed-accuracy movement control, which can explain the bi-hemispheric sensorimotor network activation of both populations. This absence of brain asymmetry could be due to the type of strategy used to perform the task. It has been shown on more distal UL tasks that complex movements are organized more bilaterally in the sensorimotor areas, particularly the primary and premotor areas [9]. Furthermore, the speed during this circular steering task was not controlled but biased to focus on the speed while maintaining constant a level of accuracy. It is indicated in the literature that with an increased speed, there is an increased ipsilateral activation to support the contralateral hemisphere [57]. The poorer performance on a speed-accurate proximal movement is in accordance with literature saying that there was a decline in manual dexterity in old age and that the time to accomplish a task was increasing [58].

Looking to the association of fNIRS and EEG for investigation of functional task motor control seems promising. These two methods are considered as standard in the field of functional neuroimaging. Measurements obtained from each of these two modalities provide complementary information related to functional activity of the brain. With a high temporal resolution, EEG can detect electrophysiological that span different frequency band and can occur simultaneously of in succession. For the fNIRS, the temporal indecision is not allowing such differentiation of electrophysiological signatures with for instance no differentiation between inhibitory and excitatory synaptic activity [59, 60]. Reversely, the EEG is recording responses of more widespread areas that could lead to spatial differentiation difficulties [24]. With our result, we reinforce the interest for the integration of these two methods for monitoring of brain hemodynamic and electrical activity in several healthy and pathological population for a variety of tasks (see [61] for a review).

This study presents a few methodological limitations. First, due to an EEG device issue, fewer old subjects were included in the final EEG data set. This could have led to false effects, or reversely to effects that could not be identified with fewer data, which could be the case with the close to our significance cut off found for the circular steering task Alpha ERS ( $p = 0.051$ ). Regarding the inclusion criteria, the older subjects recruited in this study were physically active, thus, it could have an influence on the brain responses and physical performances that were close to the young group. In further research it would be relevant to look if there is specific brain reorganization when the motor performances are more or less affected by aging [15,62]. We did not apply systemic physiology artifact correction for fNIRS signals by means of a short distance channel regression procedure. We adopted in the present study the following approaches when defining the experimental setup to minimize false positives. The study protocol was designed for inducing important contrast between the functional motor task and the resting period, employed UL movements on a horizontal plane below the level of the shoulders and heart for avoiding large blood pressure changes affecting HbO<sub>2</sub> and HbR concentrations, and used adequate pressure provided by the headcap to induce a partial transient blockage of the skin circulation during the fNIRS study [63] as witnessed by the presence of the well-defined circles over the skin that disappears about

15 min after the test session. Further investigations are needed to confirm our first findings by using simultaneous recording from deep- and shallow-separation optodes.

## 5. Conclusion

This study using combined fNIRS-EEG results showed that during real-world functional UL tasks involving proximal arm joint movements (shoulder and elbow), the bi-hemispheric sensorimotor network activation pattern seen in the young group was not affected by aging. For the circular steering task that had a speed-accuracy constraint, although it appears that healthy aging brain activity was at similar levels as their younger counterparts, the global performance was lower, suggesting a reduced neural efficiency. Indeed, EEG results showed that aging affected synchronization of the post task-related bilateral sensorimotor brain rhythms (Beta ERS) and increased ipsilateral desynchronization (Alpha ERD) for the circular steering task. We demonstrated the interest of using two types of UL tasks when monitoring brain activity, a task that is discrete with loose time-accuracy constraint (paced reaching) and another task which is continuous with high speed-accuracy constraints (circular steering). These proximal UL tasks can have clinical translation, such that if we look at post-stroke neuro-rehabilitation, the proposed tasks could be used to evaluate the evolution of the level of brain activity and performance during functional proximal UL movements that can be performed by most post-stroke patients as compared to more distal UL movements. Since the majority of stroke survivors are also affected by aging, knowing which activation we can expect in a healthy brain of an older person could help to better observe and understand how the brain is evolving through a neuro-rehabilitation program.

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## CRediT authorship contribution statement

**Camille O. Muller:** Writing – original draft, Investigation, Analysis, Visualization, Data curation, Formal analysis. **Stéphane Perrey:** Supervision, Writing – review & editing, Conceptualization, Project administration. **Karima Bakhti:** Writing – review & editing, Conceptualization, Methodology. **Makii Muthalib:** Writing – review & editing, Conceptualization, Methodology, Data curation. **Gérard Dray:** Writing – review & editing, Analysis, Visualization, Methodology. **Binbin Xu:** Writing – review & editing, Analysis, Visualization, Methodology, Data curation. **Denis Mottet:** Writing – review & editing, Conceptualization, Methodology, Data curation. **Isabelle Laffont:** Supervision, Writing – review & editing, Conceptualization.

## Declaration of Competing Interest

The authors have no conflict of interest to declare.

## Data availability

Data will be made available on request.

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