



Age-Related Adaptations of Lower Limb Intersegmental Coordination During Walking

Mathieu Gueugnon^{1,2}, Paul J. Stapley³, Anais Gouteron^{2,4,5}, Cécile Lecland⁶,
Claire Morisset^{1,2}, Jean-Marie Casillas^{1,2,4,5}, Paul Ornetti^{1,2,4,7} and Davy Laroche^{1,2,4*}

¹ INSERM, CIC 1432, Module Plurithématique, Plateforme d'Investigation Technologique, Dijon, France, ² CHU Dijon-Bourgogne, Centre d'Investigation Clinique, Module Plurithématique, Plateforme d'Investigation Technologique, Dijon, France, ³ Neural Control of Movement Laboratory, Faculty of Science, Medicine and Health, School of Medicine, University of Wollongong, Wollongong, NSW, Australia, ⁴ INSERM, UMR 1093-CAPS, Université de Bourgogne Franche Comté, UFR des Sciences du sport, Dijon, France, ⁵ Department of Physical Medicine and Rehabilitation, Dijon-Bourgogne University Hospital, Dijon, France, ⁶ Damartex Group, Roubaix, France, ⁷ Department of Rheumatology, Dijon University Hospital, Dijon, France

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*Correspondence:

Davy Laroche
davy.laroche@chu-dijon.fr

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Lower-limb intersegmental coordination is a complex component of human walking. Aging may result in impairments of motor control and coordination contributing to the decline in mobility inducing loss of autonomy. Investigating intersegmental coordination could therefore provide insights into age-related changes in neuromuscular control of gait. However, it is unknown whether the age-related declines in gait performance relates to intersegmental coordination. The aim of this study was to evaluate the impact of aging on the coordination of lower limb kinematics and kinetics during walking at a conformable speed. We then assessed the body kinematics and kinetics from gait analyses of 84 volunteers from 25 to 85 years old when walking was performed at their self-selected speeds. Principal Component Analysis (PCA) was used to assess lower-limb intersegmental coordination and to evaluate the planar covariation of the Shank-Thigh and Foot-Shank segments. Ankle and knee stiffness were also estimated. Age-related effects on planar covariation parameters was evaluated using multiple linear regressions (i.e., without a priori age group determination) adjusted to normalized self-selected gait velocity. Colinearity between parameters was assessed using a variation inflation factor (VIF) and those with a VIF < 5 were entered in the analysis. Normalized gait velocity significantly decreased with aging ($r = -0.24$; $P = 0.028$). Planar covariation of inter-segmental coordination was consistent across age ($99.3 \pm 0.24\%$ of explained variance of PCA). Significant relationships were found between age and intersegmental foot-shank coordination, range of motion of the ankle, maximal power of the knee, and the ankle. Lower-limb coordination was modified with age, particularly the coordination between foot, and shank. Such modifications may influence the ankle motion and thus, ankle power. This observation may explain the decrease in the ankle plantar flexor strength mainly reported in the literature. We therefore hypothesize that this modification of coordination constitutes a neuromuscular adaptation of gait control accompanying a loss of ankle strength and amplitude by increasing the knee power in order to maintain gait efficiency.

We propose that foot-shank coordination might represent a valid outcome measure to estimate the efficacy of rehabilitative strategies and to evaluate their efficiency in restoring lower-limb synergies during walking.

Keywords: gait analysis, aging, planar covariation, biomechanics, locomotor control

INTRODUCTION

Human walking is a common task with efficient motor control. Synergic muscle activation for the control of limb movements requires the integration of inputs from the central nervous system and feedback from proprioceptive sensors in the muscles, tendons, and limbs. In healthy persons, the neural command ensures a rhythmic, stable gait with a highly consistent intersegmental coordination, and overall walking patterns. This coordination, corresponding to the process of mastering redundant degrees of freedom of the body into a controllable system, allows the efficiency of gait by maintaining dynamic equilibrium, and the lowest energetic cost during gait (Bernstein, 1967; Lacquaniti et al., 1999). The movement coordination during gait might therefore reflect neuro-muscular synergies. While an inability to modulate the intersegmental coordination may induce gait deviations, it might also provide insights into the organization and adaptation of gait patterns with pathology or aging (Winter et al., 1990).

Declining mobility and gait performance is one of the major functional hallmarks of aging (Boyer et al., 2017). Age-related differences in gait performance include a decrease in gait speed, a reduction in step length, and/or an increased cadence (McGibbon and Krebs, 2001; Lewis and Ferris, 2008). These changes are associated with impaired balance control, a reduction of muscle strength, and mass as well as an increase of the energy cost of walking (Sepic et al., 1986; Winter et al., 1990; Judge et al., 1996; Kerrigan et al., 2000, 2001; Pavol et al., 2002; Cofré et al., 2011; Frimenko et al., 2015). As a result, the coordination was impaired with aging and linked to a history of falls in the past year (Hutin et al., 2011; Chiu and Chou, 2012; Ghanavati et al., 2014; James et al., 2017; Hafer and Boyer, 2018; Hutin et al., 2011; Ghanavati et al., 2014). However, these studies used standard frequency-decomposition methods to evaluate the intersegmental coordination during walking (i.e., continuous relative phase and coding vector). While these methods are well documented, they do not provide a complete overview of the gait processing of the lower limb intersegmental coordination due to their analysis of singular parameters. Indeed, it is known that during human walking, the lower-limb coordination is controlled through a coupling of all the segments (thigh, shank, and foot) in order to simplify the spatiotemporal control of locomotion and equilibrium (Borghese et al., 1996; Lacquaniti et al., 2002). The elevation angles of these segments are consequently related. When lower-limb segment rotations (temporal changes in the elevation angles) are plotted one vs. each others, they covary along a plane and constitute a loop [i.e., *covariation plane*, (Ivanenko et al., 2008; Lacquaniti et al., 2012a)]. Principal component analysis (PCA) was used to analyse that plane and when applied produced three components. In normal walking,

the first and the second components define the robustness of the planarity of the loop whereas the third component defines its orientation. In this context, the properties of the covariation plane provide insights about how the central nervous system controls the limbs during walking and therefore might reflect the adaptation of the neural and neuromuscular systems with aging. In particular, the work of Lacquaniti and others (for a details see Ivanenko et al., 2006; Lacquaniti et al., 2012b) postulated that planar covariation may provide a link between neuromuscular control and mechanics of gait by matching the control of lower limb muscle patterns to those of the body's center of mass (Bleyenheuft and Detrembleur, 2012). Consequently, this study aimed to evaluate the impact of aging on the coordination of lower limb kinematics and kinetics during walking at comfortable speed using the planar covariation of elevation angles. We hypothesized that the planar covariation of elevation angles should be modified throughout the lifespan in order to adapt the locomotor pattern to the constraints of aging. To this end, we assessed effects of walking speed and age on the pattern and variability of lower limb intersegmental coordination in a cohort of healthy subjects from 25 to 85 years old.

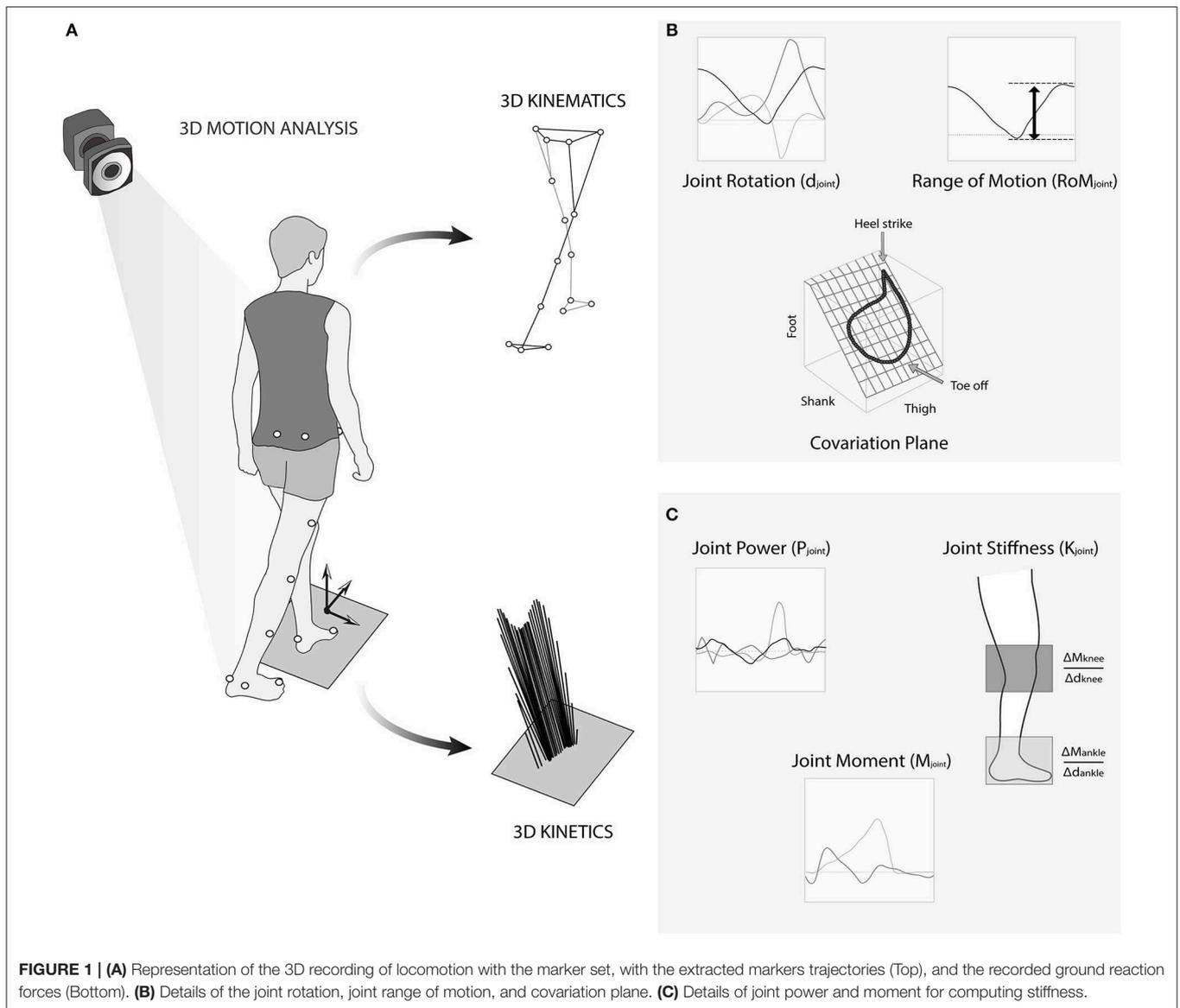
MATERIALS AND METHODS

Participants

Eighty-four volunteers (51 women and 33 men) from 28 to 85 years old were recruited from a previous asymptomatic cohort (clinical trial registration: NCT02042586) to participate in this prospective study. All showed no symptomatic musculoskeletal, neurological, or cardiovascular disease. Exclusion criteria were significant pain, ankle, hip or foot disorders, chronic back pain, Alzheimer's disease, Parkinson's disease, motor neuron disorders, non-stabilized diabetes mellitus, cardiac or respiratory insufficiency, and any inability to understand the procedures. The study protocol was approved by the local ethics committee (CPP Est I, Dijon, France). The study was conducted in compliance with the principles of Good Clinical Practice and the Declaration of Helsinki, and all patients gave their informed consent.

Task and Procedure

Participants were asked to walk 10 times barefoot while following a straight-line path, 10 meters in length traced on the floor. After each walking trial, they were asked to return to the starting point. They were instructed to adopt a natural and comfortable gait speed, as if they walked "along the street." Lower body kinematics (i.e., movements of pelvis, hips, knees, and ankles sagittal, frontal, and transverse plane) during walking were measured using an 8 optoelectronic camera motion capture system (Vicon MX, Vicon[®], Oxford, UK) sampling at 100 Hz. The marker set used, the Plug-in-Gait marker set (Davis et al.,



1991), was composed of 16 reflective markers positioned on specific anatomical landmarks on the lower limb (see Laroche et al., 2014 for placement on a representative participant).

Data Analysis

Marker trajectories were recorded by the optoelectronic camera allowing to reconstruct embedded coordinate systems associated to each rigid body segment (pelvis, femur, tibia, and foot) defining then a complete 3-dimensional model of the lower limb. To access kinetics data (i.e., joint moment and power), ground reaction forces were also recorded with two force platforms (AMTI[®], USA) sampled at 1,000 Hz (**Figure 1A**).

Marker trajectories were interpolated with Woltring polynomial and then filtered with a low pass zero phase shift Butterworth filter with a respective cut off frequency of 10 Hz. Similarly, ground reaction forces were filtered with a low

pass zero phase shift Butterworth filter with a respective cut off frequency of 50 Hz (van den Bogert and de Koning, 1996). Displacements of the center of mass (CoM), joint kinematics and kinetics were calculated with the Nexus software (Vicon[®], Oxford, UK) using inverse dynamic on the Plug-in-Gait model. The gait events were detected using a method proposed by Zeni et al. (2008) and expressed by gait cycle. Briefly, this method defines the heel-strike and the toe-off as the instant where the foot, respectively, begins to move backward and forward in the pelvis frame.

We chose the most representative variables of the gait kinematics, kinetics, and stiffness that could be associated with neuromuscular adaptation during gait (Lacquaniti et al., 2012b; Herssens et al., 2018). We first computed amplitude of displacements of CoM in the vertical plane (Amp_{CoM}). We then extracted the gait speed (v), step width, step length, and

computed the Froude number $Fr = v^2/g.L$ with g as the acceleration due to gravity and L , the subject's leg length. This parameter allows normalizing the velocity across participants (Saibene and Minetti, 2003). We also computed variability of the step length and step width (Herrensens et al., 2018). From the joint kinematics, we computed the range of motion (ROM_{Hip} , ROM_{Knee} , and ROM_{Ankle}) during walking, defined as the sum of the peak flexion and extension, or peak dorsal and plantar flexion (Figure 1B). From the joint kinetics, we extracted the maximal positive power during gait for each joint (P_{Hip} , P_{Knee} , and P_{Ankle}) and computed the associated joint moments normalized by the subject's body weight (M_{joint}) (Figure 1C). We then estimated the stiffness of the knee and ankle joints (K_{Knee} and K_{Ankle}) using the torsional spring model (Farley and Morgenroth, 1999; Kuitunen et al., 2002). The stiffness ($Nm.kg^{-1}.deg^{-1}$) was calculated as a change in the joint moment divided by the change in joint angular displacement in the middle of the ground contact phase (Hobara et al., 2013; Figure 1C).

The spatio-temporal structure of the lower limb intersegmental coordination was evaluated using a principal component analysis. Three segments per lower limb were taken into account: the feet (defined as the virtual lines joining the marker located in the second metatarsal head and the marker

located in the lateral malleolus), the shanks (defined as the virtual lines joining the marker located in the lateral malleolus and the marker located in the lateral femoral condyle), and the thighs (defined as the virtual lines joining the marker located in the lateral femoral condyle and the marker located in antero-superior iliac spine). Such analyses were computed randomly for one lower limb independently by means of the covariance matrix of the angular variation of foot, shank, and thigh segments as described previously (Borghese et al., 1996; Bianchi et al., 1998; Lacquaniti et al., 2002; Ornetti et al., 2011). The first two principal eigenvectors, accounting for almost 99% of data variance, correspond to the "covariation plane" ($Var_{CovPlane}$). The temporal coupling between the elevation angles of the shank and the thigh segments (μ_1) is illustrated with the first eigenvector and its projection on the thigh axis. The temporal coupling between the elevation angles of foot and shank segments were given by the third eigenvector (μ_3) normal to the plane. All these parameters were obtained for each gait cycle allowing to obtain two values per subject (mean and standard deviation).

Statistical Analysis

Data analysis was performed with Stata statistical software (version 15.1, Statacorp, College station TX, USA). We first applied univariate correlation between age and either normalized gait speed or planar covariation indices (μ_1 , μ_3). We applied stepwise regression analysis to identify the most relevant variables associated with age. Entry criterion of the three stepwise procedures was set at 0.20 and stay criterion at 0.10. The procedure stopped when no more variables satisfied the previous criteria. In order to validate the model, the colinearity between variables and the residuals homogeneity were checked, respectively, by the calculation of the Variance Inflation Factor (VIF) and the read of residuals vs. predicted values graphic. A VIF value higher than 5 enabled us to admit colinearity between variables (Kutner et al., 2004), those variables were then removed from the model, if necessary. Data from the gait analysis were entered as follows into the multivariate stepwise linear regression model:

- Kinematics variables (Amp_{CoM} , ROM_{Hip} , ROM_{Knee} , ROM_{Ankle} , $Var_{CovPlane}$, μ_1 , μ_3 , step width, Froude Number)
- Kinetics variables (P_{Hip} , P_{Knee} , and P_{Ankle})

TABLE 2 | Multivariate linear regression model between kinematics and kinetics variable and age.

Variables	β	z	p	Partial R^2	95% CI	VIF
ROM_{Ankle}	-1.55	-3.98	<0.001	0.22	-2.3; -0.8	2.18
P_{Knee}	16.0	2.10	0.04	0.05	1.1; 31	1.37
P_{Ankle}	-11.7	-4.13	<0.001	0.14	-17.3; -6.2	1.56
$Var_{CovPlane}$	-23.7	-3.09	0.002	0.09	-38.8; -8.7	1.84
μ_3	-135.5	-2.65	0.008	0.05	-235.7; -35.4	1.11

ROM_{Ankle} , range of motion of the ankle in the sagittal plane; P_{Knee} , maximal positive knee power; P_{Ankle} , maximal positive ankle power; μ_3 , foot-shank coordination; 95%CI, 95% confidence interval; VIF, variance inflation factor.

TABLE 1 | Population characteristics.

	Mean \pm SD
Age (years)	60.7 \pm 15.2
Weight (kg)	68.5 \pm 13.3
Height (cm)	164.6 \pm 9
Gait speed ($m.s^{-1}$)	1.09 \pm 0.16
Fr	0.14 \pm 0.02
Step Width (m)	0.075 \pm 0.02
SD Step Width (m)	0.018 \pm 0.01
Step length (m)	0.59 \pm 0.06
SD Step length (m)	0.021 \pm 0.01
Amp_{CoM} (cm)	2.82 \pm 0.52
ROM_{Hip} ($^\circ$)	40.2 \pm 4.56
ROM_{Knee} ($^\circ$)	55.6 \pm 4.52
ROM_{Ankle} ($^\circ$)	26.1 \pm 4.93
P_{Hip} (W)	1.13 \pm 0.37
P_{Knee} (W)	0.49 \pm 0.21
P_{Ankle} (W)	2.95 \pm 0.67
K_{Knee} ($Nm.kg^{-1}.deg^{-1}$)	0.07 \pm 0.07
K_{Ankle} ($Nm.kg^{-1}.deg^{-1}$)	0.06 \pm 0.03
$Var_{CovPlane}$ (%)	99.3 \pm 0.24
μ_1	0.44 \pm 0.04
μ_3	-0.16 \pm 0.03

Fr, Froude number corresponding to a normalized self-selected gait speed; Amp_{CoM} , amplitude of displacements of the center of mass in the vertical plane; ROM_{Hip} , range of motion of the hip in the sagittal plane; ROM_{Knee} , range of motion of the knee in the sagittal plane; ROM_{Ankle} , range of motion of the ankle in the sagittal plane; P_{Hip} , maximal positive hip power; P_{Knee} , maximal positive knee power; P_{Ankle} , maximal positive ankle power; K_{Knee} , knee stiffness; K_{Ankle} , ankle stiffness; $Var_{CovPlane}$, explained variance of the covariation plane; μ_1 , shank-thigh coordination; μ_3 , foot-shank coordination; SD, Standard Deviation.

- Stiffness variables (K_{Knee} and K_{Ankle})
- Gait instability parameters (Standard deviation of step length and step width).

Statistical significance was defined as $P < 0.05$. The parameter estimates, 95% confident interval and partial R-square are given and compared to Cohen's suggestions (Cohen, 1992).

RESULTS

The characteristics of participants are summarized in **Table 1**.

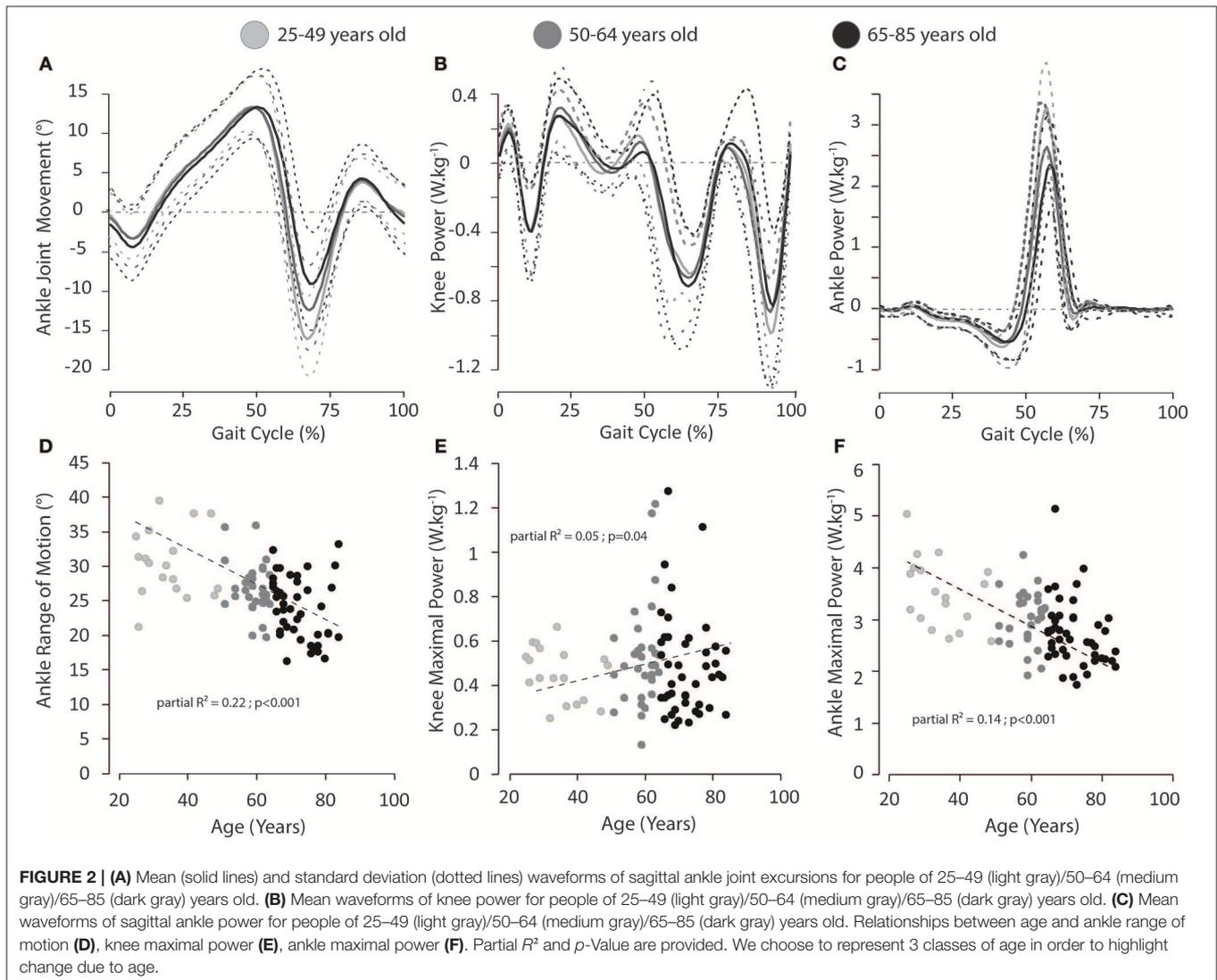
We performed univariate correlations between normalized self-selected gait speed (Fr) and age and planar covariation ($Var_{CovPlane}$) and age. A negative significant weak correlation ($r = -0.24$; $P = 0.028$) was found between normalized gait speed and age of the participants. Furthermore, we performed a multiple stepwise linear regression analysis between age and parameters computed from gait analysis (see methods for details). The regression model provided a moderate explanation of the

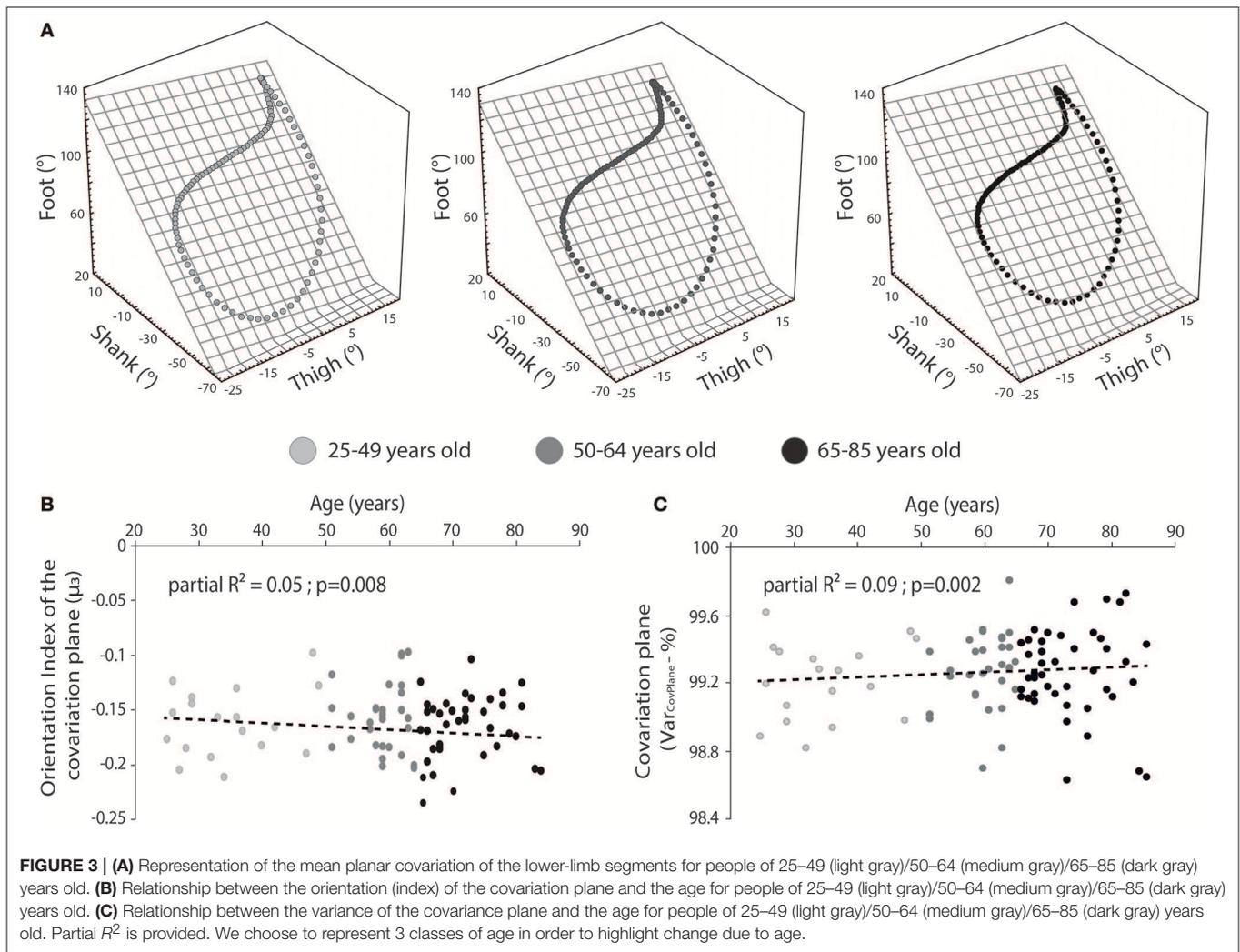
variance ($F = 6.20$; adjusted $R^2 = 0.49$; $p < 0.001$) and revealed no significant relationship between age and normalized gait speed ($p = 0.11$). However, significant relationships were found between age and range of motion of the ankle, maximal power of the knee, and the ankle (**Table 2**; **Figure 2**), percentage of planar covariation and the intersegmental foot-shank coordination (**Table 2**; **Figure 3**).

DISCUSSION

The present study aimed to assess the impact of non-pathological aging on the coordination of lower limb kinematics and kinetics during walking at conformable speed using the planar covariation of elevation angles. We showed the adaptation of planar covariation of lower-limb segments throughout the lifespan and the related kinematics and kinetics during walking.

Our results are first consistent with previous studies that showed a significant effect of aging on gait performance (Boyer





et al., 2017), especially walking speed. a significant relationship between aging and normalized gait speed was found which would attest to a decline in speed with age. Interestingly, this relationship was not evident in the multivariate model indicating that confounding variables may have been present. Indeed, while aging is associated with a reduction in gait speed, it has been previously detailed that it also produces a broad range of physiological and biomechanical changes on the walking apparatus, from the loss of muscular strength and mass, to a reduction in joint range of motion (Pavol et al., 2002; Delmonico et al., 2009; Billot et al., 2010; Cattagni et al., 2014). In our study, we corroborate and extend these results by showing that these changes occur specifically at the level of the knee/shank and ankle/foot during walking. Moreover, stiffness at both ankles and knees seems have no influence on joint motion in our results and are in line to those reported by others with no evolution of joint stiffness with age (Ochala et al., 2004; Collins et al., 2018). In the same vein, variability of step length, and step width previously reported as gait instability surrogates did not reach significance in our model. One possible explanation is that

the comfortable walking velocity might have optimized balance during gait. Further study implementing more complex balance constraints need to explore the contribution of these parameters in the aging process.

Outcomes extracted from lower-limb coordination, ankle motion, and plantar-flexors muscles seem to be a key target for both scientist and therapist. More precisely, ankle power, ankle sagittal kinematic, and shank-foot coordination seem to be reduced with aging whereas knee power seems to increase. Such modifications may reveal a potential adaptive mechanism occurring throughout the lifespan. Consequently, we believe that the planar covariation method provides basic insights into how the central nervous system controls limbs during walking by taking into account the global coordination of the thigh, shank, and foot segments. One can expect that lower-limb coordination has been modified with aging in order to compensate for the weakness progressively shown with aging and especially after the 6th decade of life. Such modifications of lower-limb coordination have been previously reported when the locomotor apparatus is impaired (Laroche et al., 2007; Ornetti

et al., 2011; Leurs et al., 2012). However, a previous study (Bleyenheuft and Detrembleur, 2012) failed to observe lower-limb coordination difference with aging. It could be explained by the weak statistical power and the absence of the shank-foot coordination, that seems to be modulated with aging. Thus, the planar covariation method seems to highlight the adaptation of the decline in the neuromuscular system with aging (Lacquaniti et al., 2012b). It could be argued therefore, that lower-limb coordination may act as a compensatory mechanism for physiological, and biomechanical changes in order to optimize the locomotor control and the dynamical balance (Ivanenko et al., 2006). Recently, Song and Geyer (2018) proposed a computer simulation to investigate the physiological causes of altered gait with aging. They found potential evidence that muscle-activation changes dominantly contribute to the reduced walking speed. In other words, the alteration of ankle power with aging could be one of the primary symptoms of the physiological decline due to aging. Further work should investigate muscular activation along lifespan in order to corroborate this hypothesis. A particular attention has to be done on prevention programs specifically designed to enhance the strength and coordination of lower-limb muscles and determine its potential effect of ankle power, lower-limb synergies, and gait speed.

This study does however, have several limitations. First, the power of the multiple regression was limited by the number of volunteers. However, the advantage inherent in this limitation is that only very strong relationships could be demonstrated. Despite the linear relationship between age and walking parameters, this study did not provide longitudinal data of volunteers. However, in the majority of studies, only groups are compared. We provide in this study data from young adults to aging people that may highlight changes during the whole lifetime. Second, the absence of the maximal strength of the volunteers to quantify the functional capacity and possibly the related gait performance should be noted.

In conclusion, this study showed age-related effects on gait performance. In particular, the modification of shank-foot

coordination could constitute a neuromuscular adaptation of the changes (biomechanical, physiological, etc.) occurred with aging. Furthermore, our results might have implications for clinical research and practice. Indeed, these four specific parameters could be relevant outcomes to measure efficacy of rehabilitative strategies and to evaluate their efficiency for restoring lower-limb synergies during walking. Consequently, it may be interesting to focus gait rehabilitation on the improvement of ankle amplitude and power as well as foot-shank coordination with healthy and pathological elderly people.

DATA AVAILABILITY

The datasets generated for this study are available on request to the corresponding author.

ETHICS STATEMENT

The study protocol was approved by the local ethics committee (CPP Est I, Dijon, France). The study was conducted in compliance with the principles of Good Clinical Practice and the Declaration of Helsinki and all patients gave their informed consent.

AUTHOR CONTRIBUTIONS

DL, PO, CL, and CM designed the experiment. DL and PO performed the experiments. MG, DL, AG, J-MC, PS, PO, and CL analyzed the data. MG, DL, and PS drafted the manuscript. MG, DL, PS, AG, J-MC, and PO critical revision of the article for important intellectual content. All authors give final approval of the version to be submitted.

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Conflict of Interest Statement: The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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