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Intersegmental coordination in severe hip osteoarthritis: is gait control influenced?

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Abstract

Background: The gait mechanism requires an efficient intersegmental coordination in order to ensure the displacement of the body while simultaneously maintaining the postural stability. However, intersegmental coordination may be disrupted by neurological or orthopaedic involvement, this increasing the metabolic cost associated with excessive or prolonged muscle co-contraction.

Research question: Our aim was to evaluate and to understand how hip OA affects lower limbs coordination during gait by using the kinematic segmental covariation law method and predict the energy expenditure.

Methods: In order to evaluate the influence of unilateral alteration of the lower limbs on the gait mechanism, three groups namely 63 hip osteoarthritis patients, 65 chronic hemiparetic stroke patients and 72 healthy subjects performed an instrumented gait analysis. The subjects had to walk barefoot for at least 3 min at a self-selected speed on a force measuring motor-driven treadmill. The biomechanical variables (kinematic, kinetic and energetical cost) were simultaneously recorded.

Results: The comparison between the three groups was tested using a repeated measure ANOVA. All biomechanical parameters show significant differences between the 3 groups highlighting the gait alteration for the patients groups. However, the energetic cost remains normal in the hip osteoarthritis group despite of the alteration of the other variables. A multivariate analysis allowed to identify the independent variables affecting more specifically their gait mechanisms.

Significance: This study showed the importance of quantitative functional evaluation in order to better understand the impact of hip osteoarthritis on the gait mechanism. The biomechanical analysis provides objective evidence of the altered gait mechanism and more

particularly of the intersegmental coordination in these patients. This gait analysis is therefore an interesting tool in the functional evaluation of the patient to better guide the diagnosis.

Keywords: Hip osteoarthritis; Gait analysis; Biomechanics; Intersegmental coordination; Gait control.

1. Introduction

Gait disorders can be classified in terms of the hierarchy of lowest, middle, and highest sensorimotor levels. Lowest-level gait disorders include musculoskeletal or primary muscle diseases, peripheral neuropathies or radiculopathies. Middle-level gait disorders include spastic gaits due to hemiparesis or paraparesis, cerebellar syndromes, and parkinsonian gaits. Finally, highest-level gait disorders include gait difficulties due to damage to cerebral hemispheres or psychogenic problems such as cautious gait, subcortical disequilibrium, and frontal gait. The quantitative assessment of this latter level is quite complex given the presence of extrinsic (attention, understanding, behavior) and intrinsic (impaired balance, gait ignition failure, freezing, shuffling) gait factors. The two firsts levels also show various disturbances of the human gait mechanism and are usually assessed in clinical gait laboratories [1]. The alteration of the gait mechanism results from interaction changes between the neurological system and the mechanical demands of the locomotor task.

The gait mechanism requires a correct intersegmental coordination in order to ensure the efficient displacement of the whole body while simultaneously maintaining the postural stability and limiting energetic expenditure [2-4]. Therefore, the intersegmental coordination involves synergetic activation of a large number of muscles to control simultaneously the postural stability and the dynamic equilibrium [5]. However, intersegmental coordination may be disrupted by neurological or orthopaedic involvement, this increasing the metabolic cost associated with excessive or prolonged muscle co-contraction [6-7]. The analyse of intersegmental coordination presents an useful clinical tool to better understand how pathology can affect the muscle strategies during gait [8]. Otherwise, to resolve this complex activation, some authors such as Grillner [9] have shown that the central nervous system uses a hierarchical and synchronized organization of the motor control during gait. The cyclic, rhythmic and alternating movements of the gait are thus generated and controlled by central

pattern generators (CPG) that are located to a large extent within the spinal cord, but are under the continuous influence of central signals. This approach hypothesizes that synergetic muscle activation of each lower limbs are controlled by these CPG and that the locomotion would result from the coupled activity of these CPG reflecting intersegmental lower limbs kinematic coordination. From this point of view, other authors [10-13] proposed an original approach of human gait, studying the kinematic segmental coordination of lower limbs by the calculation of angles of these segments (thigh, shank and foot) relative to the vertical and to the forward gait direction. This approach based on kinematic segmental covariation (KSC) law, seems to reflect in part the gait control (lower limb coordination) by CPG [10]. Lacquaniti et al. [12] postulated that planar covariation of limb segments might simplify the control of posture and locomotion by reducing the effective degrees of freedom of muscle activation. Moreover, Bianchi et al. [14] concluded that this planar covariation is a reliable predictor of the mechanical energy expenditure and could be used by the nervous system for limiting the overall energy expenditure. The KSC law's application has been mainly studied in middle-level of gait disorder more specifically in central nervous system diseases such as in chronic hemiplegic stroke patients, [*e.g.* 15-17] but much more rare in lowest-level gait disorders as in patients with joint disorders [18-19]. These studies allowed to show that subjects modify their intersegmental coordination in order to compensate weakness of the affected limb, using different muscular synergies compared to asymptomatic subjects. However, does unilateral peripheral (articular) involvement has the same gait control characteristics as unilateral central (neurological) involvement?

The aim of the present study was to evaluate how severe hip OA affects lower limbs coordination during gait by using the KSC law method and predict the energy expenditure. Due to the severe hip OA, we expected that patients would demonstrate different coordination

patterns with changes in the intersegmental coordination observed both in the affected limb but also in the non-affected limb in order to optimize energetic consumption.

2. Methods

2.1. Participants

In order to evaluate the influence of the severe hip OA on the gait mechanisms and in particular on the coordination of the lower limbs during gait by using the KSC law method, we decided to compare the data of these patients (OA group) with the data of patients with also unilateral alteration of the lower limbs namely chronic hemiplegic stroke patients (hemiplegic group).

The recruitment for OA group was performed by using the list of patients (n=213) consulting for severe OA at the department of orthopaedic surgery and traumatology of the Brussels Cliniques Universitaires Saint-Luc and Brussels Cliniques de l'Europe - Saint Elisabeth in Belgium between January 2016 and January 2017. The patients older than 45 years were invited by phone. After receiving a detailed explanation of the content and objectives of the study, they were given the possibility to participate on a voluntary basis and an appointment was scheduled to obtain written consent and perform the experimentation. All OA patients, included in this study (n=63, aged 59-75 years old), were diagnosed by 4 board-certified orthopedic surgeons (OC, JED, DP, MVC) with severe hip OA in the grade IV - end-stage as defined by clinical examination and standard radiographic [20], namely classified as severe hip OA. Patients were excluded of hip OA group if unable to ambulate without the use of an assistive device, had pain in more than one lower extremity joint on either limb, had neuro-musculoskeletal diseases, had prior lower extremity joint replacement surgery, had cardio-pulmonary problems or had comprehension problems.

For stroke patients group, we used the gait analysis recordings previously from our outpatient rehabilitation unit between 2000 and 2016. Of our database, we selected 65 chronic hemiplegic stroke patients presenting with spastic hemiparesis which performed an instrumented gait analysis. The inclusion criteria were spastic hemiparesis secondary to stroke, ≥ 6 months since stroke, ability to walk independently without an assistive device and older than 45 years. The exclusion criteria were inability to walk on a treadmill for sufficient time to complete a metabolic analysis (>2 minutes) and troubles of comprehension.

We used the lab norm to compute reference values in 72 healthy subjects, older than 45 years who were asked to perform an instrumented gait analysis. The neurological patients and healthy subjects were selected in order to perform homogenous groups in terms of age, height, weight and gait speed compared to OA patients. The anthropometrics data of each group are summarized in Table 1.

The study was approved by the local ethics committee (B403201523492), and all orthopaedic patients gave written informed consent prior to participation. For neurological and healthy subjects, we used retrospective data not requiring written consent in this case.

2.2. Instrumented gait analysis

In order to study gait, a three-dimensional gait analysis (3DGA) assessment was performed. Each subject was equipped with 19 reflective markers located on specific anatomical landmarks [21]. The subjects had to walk barefoot for at least 3 min at a self-selected speed on a force measuring motor-driven treadmill (Mercury LTmed, HP Cosmos, Germany). A motion capture system with eight infrared cameras (Elite, BTSbioengineering, Italy) measured, at a sampling rate of 200 Hz, the three-dimensional coordinates of reflective markers. Kinematic and kinetic data were simultaneously recorded for 40s and averaged on 10 successive strides. The gait analysis methods were similar for the three groups.

From kinematics data, we computed cadence and step length parameters. On each angular displacement curve in sagittal plane (pelvis, hip, knee and ankle), we measured the range of motion, defined as peak-to-peak amplitude. The total muscular mechanical work (Wtot) was also assessed. It corresponds to the sum of the external work (Wext), *i.e*. the work performed by the muscles to move the center of body mass relative to the surroundings, and the internal work (Wint), *i.e*. the work performed by the muscles to move the body segments relatively to the center of body mass [22]. The external work was computed from strain gauges measuring 3D-ground reaction forces according to Cavagna [23]. The internal work was computed from kinematics data measured with the motion capture system.

The assessment of metabolic energy cost of gait was performed with an ergospirometer (Medisoft, Belgium). Moreover, the metabolic cost of walking was determined by the subject's oxygen consumption (VO_2) and carbon dioxide production (VCO_2). The net energy cost was calculated as 'the metabolic cost of walking minus the metabolic cost of standing' divided by speed.

In order to quantify the elevation angles, a kinematic analysis of the lower limb segments in the sagittal plane with respect to the vertical was performed. The elevation angle of the thigh ([αt] trochanter - lateral epicondyle of the femur), shank ([αs] fibula head - lateral malleolus) and foot ([αf] lateral malleolus - fifth metatarsal) were computed (Figure 1). We analyzed the planar covariation of those elevation angles during the gait cycle, following the method described by Borghese et al. [10]. When plotting the elevation angles of each thigh, shank and foot against the other, we obtained 3D-position-space gait path, as shown in Figure 1 (right panel). The plane orientation and the shape of the loop reflected the phase relationship between the different segments and therefore the timing of the intersegmental coordination. The path progressed over time in the counterclockwise direction, with heel contact and toe-off phases corresponding, respectively, to the top and to the bottom of the loop. The statistical

structure underlying the distribution of the phase diagram associated with the observed changes of the elevation angles was described by a principal component (PC) analysis which enabled us to highlight the relationships between the variables analyzed. The PCs were computed by pooling together the sample of time-varying angles in a covariance 3x3 matrix. The first (PC1) and the second (PC2) components lie on the plane of angular covariation and describe the global form of the gait loop, whereas the third component (PC3) is orthogonal to the plane. The percentage of variance (PV) of each component illustrated by PV3 is an index of planarity of the loop (0% corresponds to an ideal plane). The percentage of variance demonstrated by PC1 and PC2 reflects the shape of the gait loop. The greater the value of PV1 relative to PV2, the more eccentric (closer to a line segment) the elliptic loop becomes. In normal subjects at spontaneous walking speed, PV1 and PV2 explain ≥ 99% of the percentage of variance.

2.3. Statistical analysis

We performed a one way analysis of variance to compare the anthropometric variables of each group in order to quantify the homogeneity between groups on parametric data (mean ± SD). A Kruskal-Wallis One Way Analysis of Variance on Ranks was used to compare the nonparametric data (median [range]).

Subsequently, a comparison between the three groups was performed in order to quantify significant differences in mechanical (Wint, Wext and Wtot), kinematics (range of motion) and energetical (cost) variables. A one way analysis of variance was used on parametric data (mean ± SD). A Kruskal-Wallis One Way Analysis of Variance on Ranks was used for nonparametric data (median [range]).

Following these analyses, a multivariate analysis (backward stepwise regression) was used to assess the influence of biomechanical parameters (work, range of motion and coordination)

on cost. The p-value of significance was <0.05. All statistical analyses were performed using SigmaStat software.

3. Results

3.1. Patients anthropometrics parameters

No significant differences between the three groups were observed for the anthropometric characteristics (Table 1).

3.2. Biomechanical variables

The biomechanical parameters gather the kinematics data (range of motion - ROM), lower limb coordination (elevation angles of pathological side and healthy side), mechanical work (Wtot, Wext and Wint) and energetical cost.

More particularly, the kinematics data (Figure 2) show a significant difference both for OA group and for hemiplegic group compared to healthy group: cadence (healthy: 100.3 step min^{-1}, OA: 108.8 step min^{-1}, hemiplegic group: 78.1 step min^{-1}), step length (healthy: 0.54m, OA: 0.44m, hemiplegic group: 0.43m) and range of motion (pelvis for healthy: 12.3°, OA: 3.7°, hemiplegic group: 8.5°; hip for healthy: 37.0°, OA: 27.0°, hemiplegic group: 26.4° and knee for healthy: 49.9°, OA: 38.6°, hemiplegic group: 24.5°). Only ankle does not show significant differences between both groups. Concerning the comparison between the OA group and the hemiplegic group, we observe significant differences for cadence, pelvis and knee ROM.

For the elevation angles data (Figures 3 and 4), significant differences are observed between OA and hemiplegic group for PV1 (pathological side: 84.1-OA *vs* 78.3% hemiplegic; healthy side: 81.7 *vs* 73.1%) and for PV3 in healthy side only (0.57 *vs* 0.85%), but also between the hemiplegic group and the healthy group for PV1 in healthy side (73.1 *vs* 80.7%) and for PV3 in healthy side (0.85 *vs* 0.62%). PV1 in pathological side showed also significant differences between the OA group and the healthy group (84.1 OA *vs* 81.5%). However,

when healthy subjects walk at their spontaneous speed (± 4.5 km h^{-1}), no significant difference is observed between OA and healthy groups (84.1 OA *vs* 85.8 healthy).

The mechanical work data (Figure 3) of the OA and hemiplegic groups show an increase of the total muscular mechanical work (Wtot) due to a significantly increase of external work (Wext) as compared to healthy subjects (Wext: healthy: 0.28 $J.kg^{-1}.m^{-1}$/ OA: 0.45/ hemiplegic: 0.5). However, the energy cost is only greater for hemiplegic group compared to OA and healthy group (healthy: 3.0 $J.kg^{-1}.m^{-1}$/ OA: 2.69/ hemiplegic: 5.32). For OA group, the internal work (Wint) is significantly different compared to the healthy and hemiplegic groups (healthy: 0.23 $J.kg^{-1}.m^{-1}$/ OA: 0.2/ hemiplegic: 0.25).

These results show that there are multiple relationships explaining the alteration of the gait for the OA patients. However, the energetic cost which is a pertinent indicator of the efficiency of gait activity remains normal in the OA group despite of the alteration of the other variables. That is why we conducted a multivariate analysis to understand the influence of the whole of biomechanical variables on the energetic cost. Results show that the cost (coefficient 7.329 ± 1.089) could be predicted from a linear combination of 3 independent variables: walking speed (coefficient -1.13 ± -0.68; p-value <0.001), Wint (coefficient 7.841 ± 0.336; p-value <0.001) and PV1 (coefficient -0.034 ± -0.146; p-value 0.013).

4. Discussion

Our aim was to understand how hip OA affects lower limbs coordination during gait by using the KSC law method and predict the energy expenditure. We showed that energy expenditure was normal despite excessive muscular mechanical due to impaired kinematics. We showed that lower limb coordination predicted partly the energy expenditure.

Our kinematic data and muscular mechanical work confirm the previous studies in both for OA group [24-31] and for hemiplegic group [*e.g.* 32]. Effectively, we showed that OA and

hemiplegic patients have an increase of total muscular mechanical work (Wtot) due to an important increase of external work (Wext) as compared to healthy subjects (Figure 2). This increase of external work is associated with altered locomotor pendulum mechanism. This alteration of gait mechanism is mainly due to a decrease of ROM and of step length. One possibility is that orthopedic or neurological disorder may disrupt mechanical energy exchange because of the reduced ROM, especially in the hip flexion/extension but also in the knee flexion/extension. This reduction may limit the rise and fall of the center of mass as a result reducing the ability to store gravitational potential energy allowing the vertical rise in the center of body mass [31]. The reduction of hip and knee ROM can also be associated with higher levels of muscular work necessary to move the center of mass owing to a decrease in the efficiency of exchanging potential energy and kinetic energy in these individuals [29-31]. These compensation strategies cause alteration in normal kinematic patterns (particularly in the sagittal and frontal planes) resulting in significant reduction of joint loading in the affected side and causing increased joint loads, particularly in the contralateral limbs [24-28]. According to some authors [25-26, 33], these adaptive gait strategies developed by the patients with hip OA become usual which could explain why the energy cost is not affected and therefore not significantly different from healthy subjects. These patients seem to adapt their gait strategy to limit the energy expenditure during walking and thus to have the most efficient possible gait in comparison to pain and/or stiffness due to OA. This could correspond to a law of optimization of the functioning of the system which implies an adapted regulation of the muscular forces in order to move [5, 34]. This adapted regulation results in a cost function considered as an indicator of the process of optimizing the resources of the system [34]. Conversely, the hemiplegic patients expend more energy to walk than hip OA and healthy groups. This higher expenditure of the energical cost is related to neurological impairments in these patients [35]. The increased of the energical cost indeed relate to an

increase of the mechanical work that could be explained by altered pattern of movement (stiff-knee gait). The increased mechanical work is mainly produced by the healthy limb to lift the center of body mass, but poorly related to the efficiency of work production, confirming previous results [36].

These first results confirm and support the observed results for elevation angles. As a reminder, human gait is an activity that requires an intersegmental coordination involving synergetic activation of a large number of muscles that is coordinated in part by the central nervous system. Some authors [10-14] postulated that kinematic segmental covariation (KSC) law highlights and reflects the gait control and more particularly the lower limb coordination and is a reliable predictor of the mechanical energy expenditure which could be used by the nervous system for limiting the overall energy expenditure. Our results allowed to show that patients modify their intersegmental coordination in order to compensate weakness of the affected limb, using different muscular synergies compared to the healthy group (Figure 4). Our results nevertheless suggest that the intersegmental coordination of the hip OA group corresponds more to the data observed in our healthy group. When we compare OA patients with healthy subjects walking at their spontaneous speed, coordination becomes normal. This seems to mean that OA patients are able to adapt effective coordination to optimize their gait and reduce their energetic cost. Only the PV1 in pathological side show significant differences that could be explained by the effect of reduction of the ROM in affected side and more specifically on the hip of the affected side. The reduction of these ROMs thus directly alters the coordination of the affected limb which is illustrated by a phase shift compared to the healthy group. The elevation angles data show for the healthy side only an insignificant discrepancy of phase relationship between the different segments and therefore the timing of the intersegmental coordination. These results imply that these patients were adopting a coordination pattern that is closer to that of the healthy group which could explain why the

energy cost is not affected. This last result confirms our observed results for kinematic data and muscular mechanical work and shows that our hip OA patients are in a process of the gait optimization with a cost/benefit ratio as optimal as possible compared to their hip OA disorders. This can be illustrated by the implementation of synergies and muscular couplings controlling the segmental chain of the lower limbs and allowing control of the degrees of freedom of the ankle - knee - hip during gait while minimizing the energy cost [34]. Finally, the system seeks to operate in the least costly manner possible [34].

These results show that there are multiple linked factors explaining the alteration of the gait for the OA patients.

Our study however has some limitations. It focuses on a gait mechanism analysis by using in part the KSC law method. However, as stated by Lacquaniti et al. [12], an analysis of intersegmental covariation cannot be simply attributed to biomechanical changes, because the degrees of freedom of muscle activity do not match those of the center of body mass and of the lower limb kinematic. Electromyographic activity patterns during walking are indeed much more complicated than a simple alternative activity between flexors and extensors. It would be then interesting to complete this study with data on electromyographic activity patterns in order to analyse the influence of hip OA on muscular activity during gait. We focused only on coordination in sagittal planes. The study of the coordination in the frontal and transverse planes should be included in future.

To conclude, our study shows the importance of this analyzing in order to better understand the impact of hip osteoarthritis on the gait mechanism. This biomechanical analysis of gait provides objective evidence of the altered mechanism of locomotion in patients with severe hip osteoarthritis. These measurements may be relevant to assess impairment in ambulatory patients and may be also useful in clinical research to improve

knowledge of the effect of treatment. The gait analysis is therefore an interesting tool in the functional evaluation of the patient to better guide the diagnosis.

Declarations

Author's contributions: All authors worked collectively to develop the protocols and method described in this paper. LW, CD and PM were principal investigators responsible for the fieldwork, the statistical analysis and the writing of the manuscript. All authors contributed to the interpretation of results. All authors read and approved the final manuscript.

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Conflict of interest statement: The authors declare that they have no proprietary, financial, professional, or other personal competing interests of any nature or kind.

Ethics Approval: This study was conducted according to the declaration of Helsinki and had ethical approval from the Comité d'Ethique Hospitalo-Facultaire de l'Université catholique de Louvain (B403201523492). Participants gave written informed consent prior to data collection.

Consent for publication: Participants gave written informed consent to data treatment in this research study and permission to publish anonymous data and results.

Availability of data and supporting materials: Data and software are the intellectual properties of the Université catholique de Louvain as well as the Fonds de Recherche Clinique (FRC). In addition, further studies on these data are ongoing which explain that data will not be shared at this stage.

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Table 1. Anthropometric characteristics of the three groups.

	Hip OA patients N=63	Hemiplegic patients N=65	Healthy Subjects N=72	*p-value*
Age (years)	65 [59-75]	59 [56-68.5]	61.6 [50-78.4]	0.071
Weight (kg)	75 [66-85]	76 [64.75-87]	73.8 [63-84]	0.70
Height (m)	1.68 ± 0.08	1.7 ± 0.08	1.69 ± 0.09	0.29
Speed (km h^{-1})	2.5 [1.8-3.5]	2 [1.7-2.5]	2.5 [1.25-3.75]	0.085

Figure 1.

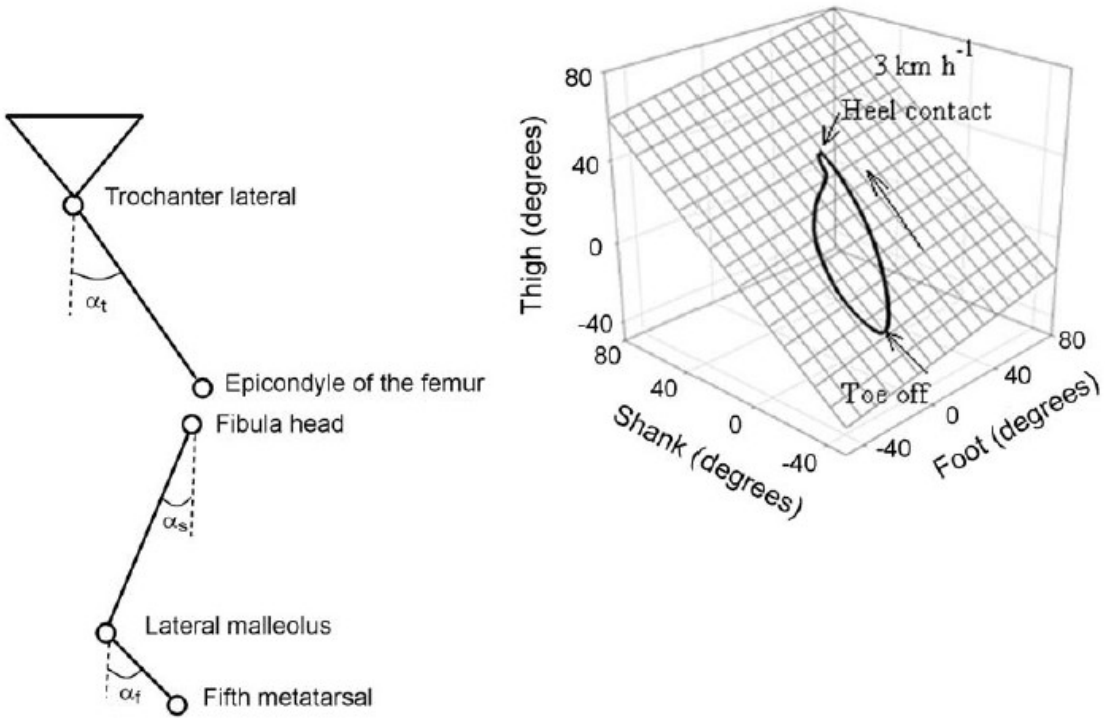

Trochanter lateral
α_t
Epicondyle of the femur
Fibula head
α_s
Lateral malleolus
α_f
Fifth metatarsal
3 km h^{-1}
Heel contact
Toe off
Thigh (degrees)
80
40
0
-40
Shank (degrees)
Foot (degrees)

Figure 2.

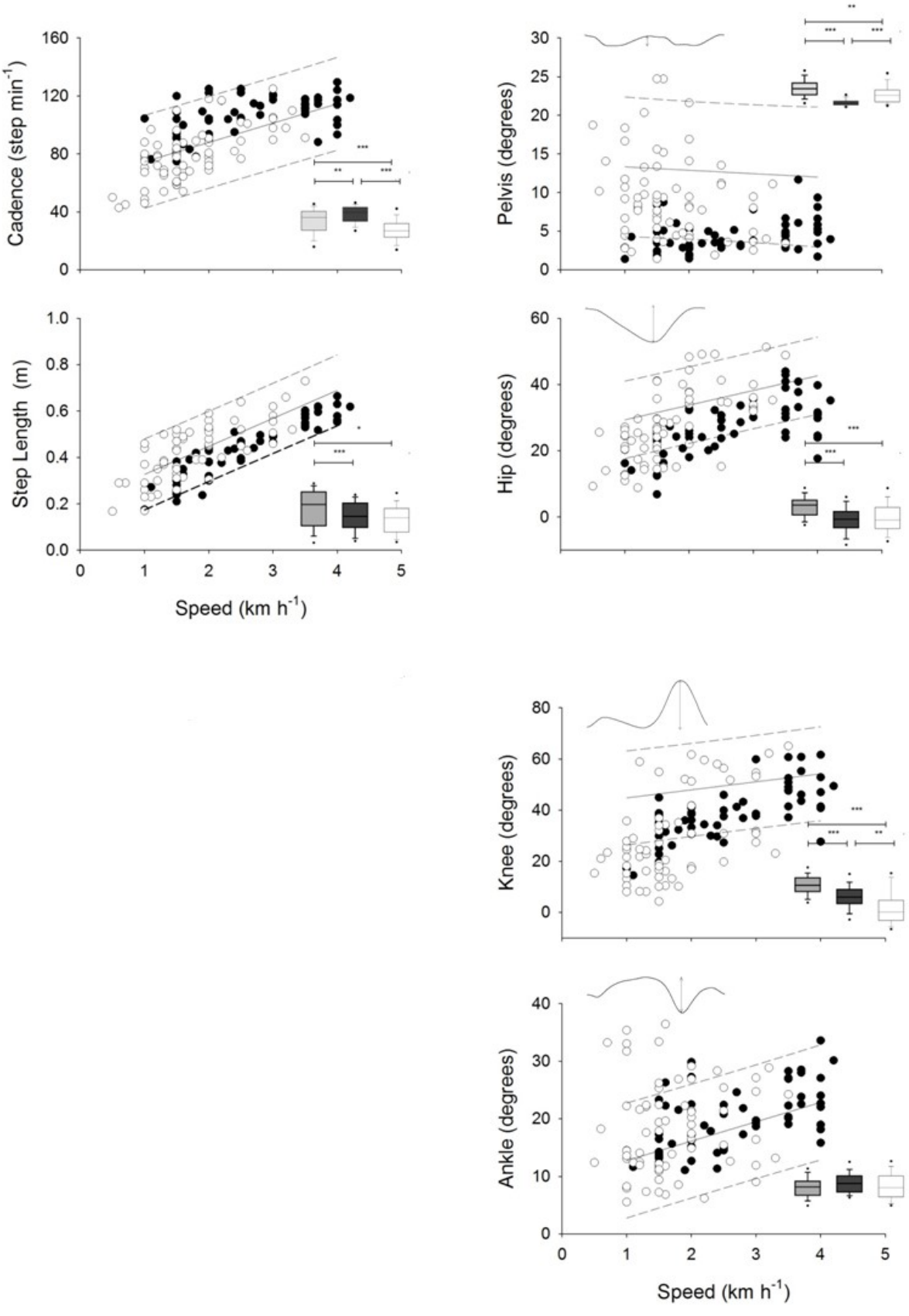

Cadence (step min-1)
Pelvis (degrees)
Step Length (m)
Hip (degrees)
Knee (degrees)
Ankle (degrees)
Speed (km h-1)
Speed (km h-1)

Figure 3.

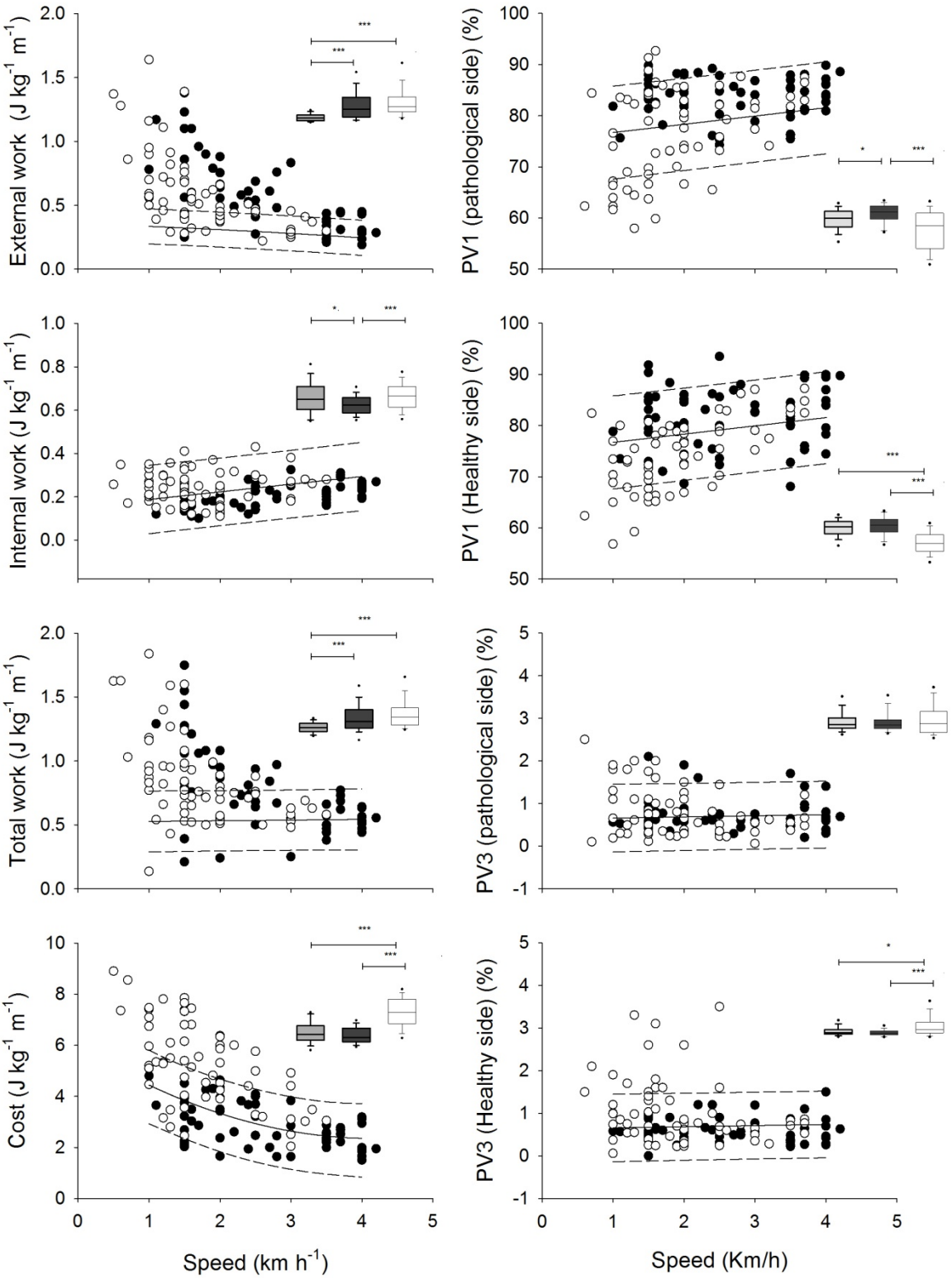
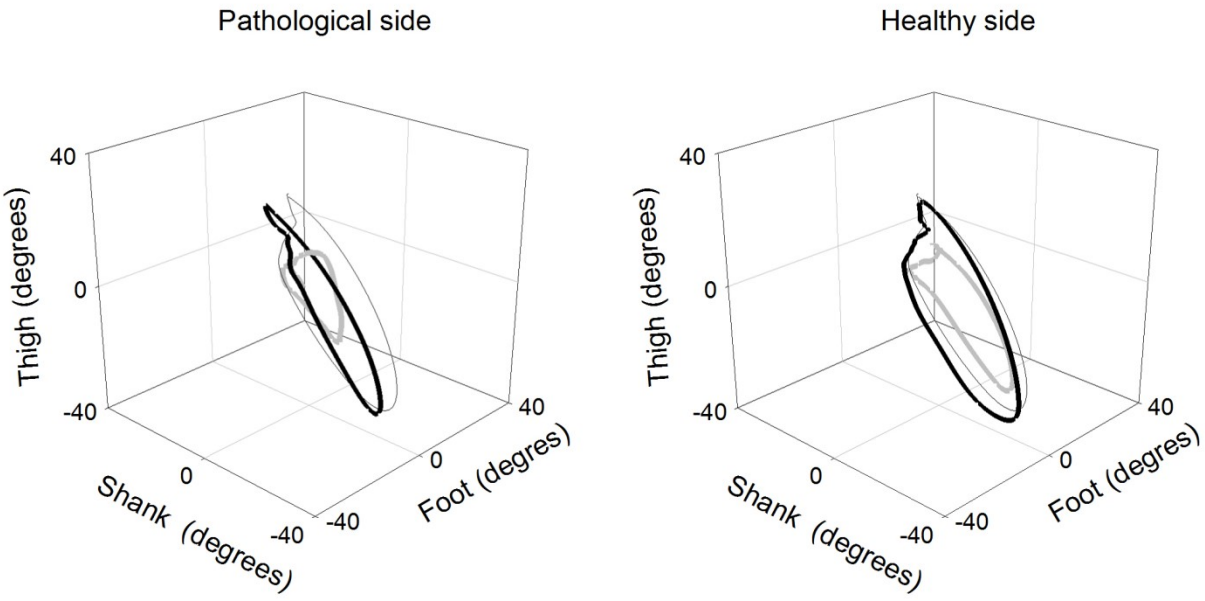

External work (J kg-1 m-1)
PV1 (pathological side) (%)
Internal work (J kg-1 m-1)
PV1 (Healthy side) (%)
Total work (J kg-1 m-1)
PV3 (pathological side) (%)
Cost (J kg-1 m-1)
PV3 (Healthy side) (%)
Speed (km h-1)
Speed (Km/h)

Figure 4.


Pathological side
Healthy side
Thigh (degrees)
Shank (degrees)
Foot (degres)
40
0
-40

Captions

Figure 1.

The elevation angle of each limb segment in the sagittal plane corresponds to the angle between the segment and the vertical. *at* is the angle between the thigh and the vertical, *as* the angle between the shank and the vertical, and *af* the angle between the foot and the vertical. The right part of the figure shows the 3D-position-space gait path of a normal subject (elevation angles of the thigh, shank, and foot plotted one versus the others). The path is progressing in time in the counter-clockwise direction with heel contact and toe-off phases corresponding to the top and bottom of the loop.

Figure 2.

Kinematics variables are expressed as function of walking speed.

Each group is represented by a color: the boxplots and the black symbols correspond to the group OA, white to the hemiplegic group and gray to the healthy group.

Black and dotted lines represent the quadratic extrapolation which then applied to the mean and 95% confidence intervals obtained by healthy subjects walking from 1 to 5 km h-1.

Figure 3.

All biomechanical variables are expressed as function of walking speed.

The left column contains data relating to the mechanical work and cost. The elevation angles of pathological and healthy sides are shown in the right column.

Each group is represented by a color: the boxplots and the black symbols correspond to the group OA, white to the hemiplegic group and gray to the healthy group.

Black and dotted lines represent the quadratic extrapolation which then applied to the mean and 95% confidence intervals obtained by healthy subjects walking from 1 to 5 km h-1.

Figure 4.

This figure show the 3D-position-space gait path (elevation angles of the thigh, shank, and foot plotted one versus the others). The path is progressing in time in the counter-clockwise direction with heel contact and toe-off phases corresponding to the top and bottom of the loop.

The left row contains data for the pathological side; the data for the healthy side are shown in the right row.

The thick black line represents the curve of OA group. The thick grey line represents the curve of hemiplegic group and the thin dark line represents the curve of healthy group.