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An easily applicable method to analyse the ankle-foot power absorption and production during walking

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ABSTRACT

Background: Power and work at the ankle joint during gait are usually computed considering the foot as a rigid body [1–6] (Ankle Joint method, AJ). The foot is instead a deformable structure and can absorb and produce work by pronation/supination, foot arch deformation and other intrinsic movements. A different approach, named "the Distal Shank method (DS)" [7–12] considers all these aspects without increasing the complexity of the protocol, and thus it seems promising for clinical applications [12].

Research questions: a) To characterize the differences in power and work computed using the two mentioned methods for a relatively large number of subjects walking at different velocities, barefoot and with different shoes; b) To assess the practical feasibility of the DS method for clinical applications.

Materials and methods: Eighteen healthy subjects were evaluated while walking barefoot at slow, natural and fast velocity. Shod walking was analysed at natural velocity. Four subjects were also analysed while walking in high-heel shoes. The power at the ankle joint was computed with both the AJ and the DS methods. We then compared the obtained results.

Results: The DS method showed a consistent negative peak of power absorption during the load acceptance phase, barely visible with the AJ method. The maximum power production calculated with the DS method was significantly lower. The work at the end of the stride cycle was lower with the DS method, and in most conditions even negative, thus indicating higher energy dissipation.

Significance: We confirmed on a large cohort of healthy subjects and in different walking conditions that neglecting foot deformations during gait leads to underestimate power absorption and overestimate power production. The DS method does not require a complex gait analysis protocol, nor additional time for the analysis, and can provide information of clinical interest, related to foot mechanical alterations.

1. Introduction

The foot is not a rigid body. It is a complex structure composed by 23 bones connected by ligaments and subjected to the action of several muscles. When loaded by body weight during walking, the structure undergoes deformations consisting of rearfoot pronation/supination, lowering of the medial arch, changing of the widening between first and fifth metatarsi, flexion of the phalanges in the late stance phase, and many other intrinsic movements occurring both in the stance and swing phases. Multi-segmental foot models were proposed to capture and describe all these aspects [13,14]. In particular, the so-called Heidelberg foot model [13] aims at the analysis of those intrinsic movements that have a clinical relevance and can be measured without any arbitrary assumption about internal rotational axes. Other models, like the Oxford foot model [14] consider the foot as composed of three segments: hindfoot, forefoot and hallux, and aims at providing the relative rotations among them. Although the number of proposed foot models is relatively high (comments about this can be found in [15]) seldom they are implemented in gait analysis. The reason can be related to the complex and delicate procedure of markers placement required by these models, a critical aspect that can strongly affect the results.

In most gait analysis studies, a simplified representation of the foot is adopted, consisting in a single rigid body connected to the shank by a rotational hinge. Its use has been consolidated in several papers and textbooks [1–6]. This method, that henceforth will be named "the Ankle Joint method (AJ)", can provide only a general description of dorsi-plantar flexion of the forefoot and, eventually, of its eversion-inversion, but it is unsuitable to capture the rearfoot movements and

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establish how much power is absorbed and produced by the effect of foot deformations [1314]. The limits of this model become particularly evident in the evaluation of deformable prosthetic feet, designed to store and release elastic energy during the walking cycle. Exactly for this reason, Prince and colleagues [7] developed a method for foot power calculation which intrinsically considers the effects of foot deformations. The ankle-foot power is calculated as the sum of the translational and rotational power associated to the movement of the shank rigid segment. Several authors provided evidence of the better accuracy and reliability of this method, named “the Distal Shank method (DS)”, with respect to the AJ method even for the natural foot power estimation [8–12]. In particular, these studies highlighted that the AJ method overestimates the amount of power produced in the push-off phase. These considerations have surprisingly received poor attention from the scientific community, and the AJ method is currently adopted in most applications. Only over the last few years the limitations of the AJ method were again underlined in some recent publications, particularly because of the possible implications of an incorrect evaluation of power in clinical practice [10–12].

In the wake of these previous studies, we investigated the opportunity to adopt the DS method in a clinical setting, by comparing the results obtained with the two methods and considering the compatibility of the DS method with the commonly used gait analysis protocols. To this purpose we evaluated the power and the work profiles of the ankle-foot in different walking conditions (different speeds and shoes) for a relatively large number of subjects. Our work will also provide reference data for power calculation in healthy young subjects.

2. Methods

2.1. Power and mechanical work calculation

In order to compute the power due to ankle and foot during the stance phase of walking Zelik and colleagues [12] proposed the following formula:

$$P_{distal\_shank} = F_{GRF} \cdot (V_{shank} + \omega_{shank} \times r_{COP/shank}) + M_{free} \times \omega_{shank} \tag{1}$$

Where P_{distal_shank} is the power transmitted to the shank; F_{GRF} is the ground reaction force; V_{shank} is the velocity of the shank's centre of mass (COM); ω_{shank} is the rotational velocity of the shank segment; $r_{COP/shank}$ is the position of the centre of pressure (COP) relative to the shank's COM; M_{free} is the free moment of ground reaction (only the vertical component of this moment is different from zero).

The formula was rearranged by the same authors [12] as:

$$P_{distal\_shank} = F_{GRF} \cdot V_{shank} + F_{GRF} \cdot (\omega_{shank} \times r_{COP/shank}) + M_{free} \cdot \omega_{shank} \tag{2}$$

Which is equivalent to:

$$P_{distal\_shank} = F_{GRF} \cdot V_{shank} + (r_{COP/shank} \times F_{GRF} \cdot \omega_{shank}) + M_{free} \cdot \omega_{shank} \tag{3}$$

And thus, finally:

$$P_{distal\_shank} = F_{GRF} \cdot V_{shank} + (M_{GRF/shank} + M_{free}) \cdot \omega_{shank} \tag{4}$$

By considering the shank as a rigid body, the formula (4) is valid for any point belonging to the shank, in particular for the point at the base of the shank: the ankle joint center (AJC). Substituting the shank's COM with the AJC, it becomes:

$$P_{distal\_shank} = F_{GRF} \cdot V_{AJC} + (M_{GRF/AJC} + M_{free}) \cdot \omega_{shank} \tag{5}$$

Where $M_{GRF/AJC} = r_{COP/AJC} \times F_{GRF}$ is the moment due to the ground reaction force in relation to the ankle joint centre, and the sum $M_{GRF/AJC} + M_{free}$ is the total external moment M_{Ext} transferred to the shank (Fig. 1, on the right). It clearly appears that the power transmitted to the shank is the sum of a translational power ($P_{transl} = F_{GRF} \cdot V_{AJC}$) and a rotational power ($P_{rot} = M_{Ext} \cdot \omega_{shank}$). Although the foot is not analysed, the foot deformations are intrinsically affecting the power computed in this way.

In the traditional method of ankle joint power computation (the AJ method, see Fig. 1, on the left), the power is computed as:

$$P_{AJ} = M_{AJC} \cdot \omega_{AJC} \tag{6}$$

where P_{AJ} is the power at the ankle joint; M_{AJC} is the ankle joint internal moment; ω_{AJC} is the angular velocity of the relative rotation between foot and shank.

The mechanical work, in both cases, was computed by integrating the power over time:

$$W = \int P(t)dt \tag{7}$$

Positive values mean that the foot is transmitting work to the shank, while negative values mean that the work is absorbed by the foot.

2.2. Experimental protocol implementation

Retroreflective markers (spherical, 15 mm diameter) were positioned on the medial and lateral malleoli and on medial and lateral femoral epicondyles of both limbs of our experimental subjects. These were intended for the identification of the longitudinal axis of the shank and to define a local reference system of Cartesian axes. Two additional markers were put on the first and fifth metatarsal heads in order to identify the foot longitudinal axis and its local reference frame. These last two markers were only used for the computation of the relative angular velocity of the ankle joint, as required by the more traditional AJ method and were not required by the DS method.

Kinematic and dynamic data were collected by means of a motion analysis system (Smart-E, BTS, Italy) with 6 cameras (sampling frequency of 120 Hz), and a force platform (sampling frequency of 960 Hz, Kistler, 9286 A A, Switzerland). The centre of the ankle joint (AJC) was defined as the midpoint of the markers placed on the medial and lateral malleoli; the longitudinal axis of the shank (Z_{sh}) was identified as the line connecting the ankle joint centre to the midpoint between the medial and lateral femoral epicondyles (point P); the longitudinal axis of the foot (Z_{foot}) was defined as the line which connects the midpoint of the markers on the first and fifth metatarsal heads to the ankle joint centre. The posterior-anterior axis of the shank (Y_{sh}) was obtained as the perpendicular to the plane defined by the markers on the medial and lateral malleoli and the point P. The medial-lateral axis of the shank (X_{sh}, exiting to the right) was then obtained by the cross product of Y_{sh} and Z_{sh}. The angular velocity of the shank (ω_{shank}) was obtained by first defining the three Euler angles of the shank reference frame in relation to the absolute reference frame (the laboratory frame). Named θ, χ, ϕ the nutation, precession, and rotation angles respectively, the three components of the angular velocity in the shank reference frame are defined by:

$$\begin{aligned} \omega_{x\_sh} &= \dot{\chi} \sin\theta \sin\phi + \dot{\theta} \cos\phi \\ \omega_{y\_sh} &= \dot{\chi} \sin\theta \cos\phi - \dot{\theta} \sin\phi \\ \omega_{z\_sh} &= \dot{\chi} \cos\theta + \dot{\phi} \end{aligned} \tag{8}$$

And the absolute components of the angular velocity are obtained by a rotation transformation:

$$\omega_{shank} = {}^{A}_{A'}[R_{sh}]\omega_{\_sh} \tag{9}$$

Where $\omega_{\_sh}$ is the angular velocity vector in the shank reference system A' (whose components are ω_{x_sh}, ω_{y_sh}, ω_{z_sh}); ${}^{A}_{A'}[R_{sh}]$ is the rotation matrix of the shank reference system A' in relation to the laboratory reference system A.

Concerning the AJ method, the angular velocity of the foot in relation to the shank was computed assuming that the flexion/extension axis was perpendicular to the plane defined by the longitudinal axes of foot and shank. The orientation of the vector ω_{AJC} was thus obtained by the cross product of the two axes, and the relative angle was computed as:

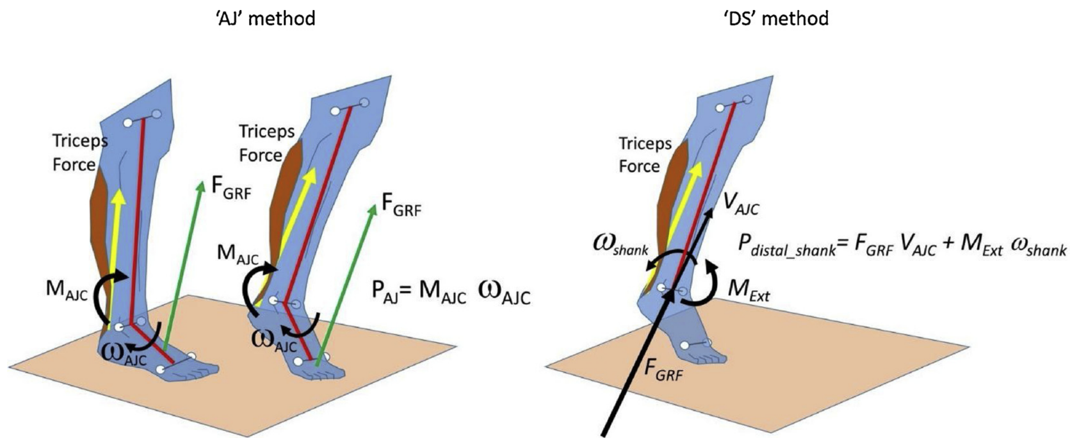

Fig. 1. Models of power computation. Left: the ankle joint (AJ) method; right: the distal shank (DS) method (see the *Methods* section for details). F_{GRF} = ground reaction force; M_{AJC} = moment of the external forces applied at the ankle joint; ω_{AJC} = angular velocity of the ankle joint; P_{AJ} = power at the ankle joint; M_{EXT} = external moment applied to the shank; ω_{Shank} = angular velocity of the shank; V_{AJC} = linear velocity of the center of the ankle joint; P_{distal_shank} = power due to ankle and foot.

$$\alpha_{AJC} = \arccos\left(\frac{Z_{foot} \bullet Z_{sh}}{|Z_{foot} \bullet Z_{sh}|}\right) \tag{10}$$

The modulus of the angular velocity (ω_{AJC}) was obtained by computing the time derivative of the relative angle:

$$\omega_{AJC} = \dot{\alpha}_{AJC} \tag{11}$$

2.3. Participants and walking conditions

We enrolled 18 healthy subjects: 9 males and 9 females (mean age: 26 ± 2 years). We excluded from the analysis subjects with orthopedic pathologies, major orthopedic surgery or motor disorders. The Ethical Committee of our Institution authorized the investigation and the participants signed an informed consent which could be revoked at any moment by the subjects.

All participants were asked to walk in two different conditions: barefoot and shod. While walking barefoot they performed the task at three different velocities: slow, normal and fast (the speed was self-selected by the subject itself); in shod conditions they walked at natural speed. The worn shoes were common sport shoes that each subject used currently in his/her daily life. In addition to these walking trials, 4 female subjects (mean age: 27 ± 1 years) were asked to walk also in high heel shoes (average height 8 ± 2 cm SD) at their preferred speed. For the analysis, we considered only the trials in which the subject stepped on the force platform with a single foot, without artificially lengthening or shortening the step. Trials with truncated or double steps on the force plate were excluded from further analysis. For each subject and walking condition, we collected five trials for each limb.

2.4. Parameters of interest and Statistical Analysis

We calculated the mean power and mechanical work along the stride cycle for each subject in each different walking condition with the DS and AJ methods. Power and work were normalized to the mass of the subject. Then we extracted the following parameters from the curves: maximum value of the power, between 80% and 90% of the stance phase duration; first minimum, that occurred within the first 20% of the stance phase; second minimum, at about the 70% of the stance phase. As to the mechanical work, we measured the work absorbed, produced and the final value at the end of the stride cycle. For each parameter and condition, we calculated the Mahalanobis distance between the observations and removed the outlier values [16]. We tested the distribution of each parameter for normality by the Shapiro-Wilk test. To those parameters that were non-normal distributed we applied a logarithmic transformation and tested again the transformed parameter for normality. The statistical comparison across the two methods was conducted by means of a matched pairs analysis performed with the JMP statistical package (JMP 13, SAS Institute Inc., Cary, NC, USA) with p-value set at 0.05.

3. Results

3.1. Barefoot condition

Fig. 2 reports the average curves (+/- one standard deviation) of power (left) and work (right) computed by the two methods for the barefoot walking at natural velocity. Interesting qualitative differences appeared at glance in all walking conditions. The most striking one was a consistent negative peak of power absorption that appeared with the DS method at load acceptance and was barely visible with the AJ method (Fig. 2, panel A). The curve representing the work was affected by this power absorption and exhibited a sort of downward shift in this phase. The peak of power production appeared considerably smaller when calculated with the DS method in relation to the one obtained with the AJ method (Fig. 2, panel A). This difference resulted in a lower final work calculated with the DS method with respect to the AJ method. Also the work absorbed during the first part of the stance phase (from 0% to approximately 70%) and produced during the terminal stance (from 70% to 100%) appeared different across the two methods (Fig. 2, panel B). The statistical analysis confirmed that all the observed differences were statistically significant for all gait velocities (Table 1).

The final work calculated with the DS method was on average negative for all gait velocities except for fast walking. Conversely, the final work resulted from the AJ method was always positive. All the parameters computed by both methods increased at the higher velocities.

3.2. Shod condition

The power and work curves obtained during shod walking and in the barefoot condition exhibited similar morphology and values (Fig. 3, panels A and B). The statistical significance of the differences between the two methods was confirmed for all parameters in the shod condition as well (Table 2). During walking with high-heel shoes, the power and work curves exhibited peculiarities that were not present in the other walking conditions (Fig. 3, panels C and D). In particular, during the load acceptance phase the DS method revealed two peaks of absorption instead of the one obtained with the AJ method. Furthermore, the maximum power and the produced work were lower with respect to the other walking conditions, and this appeared in both methods. Walking

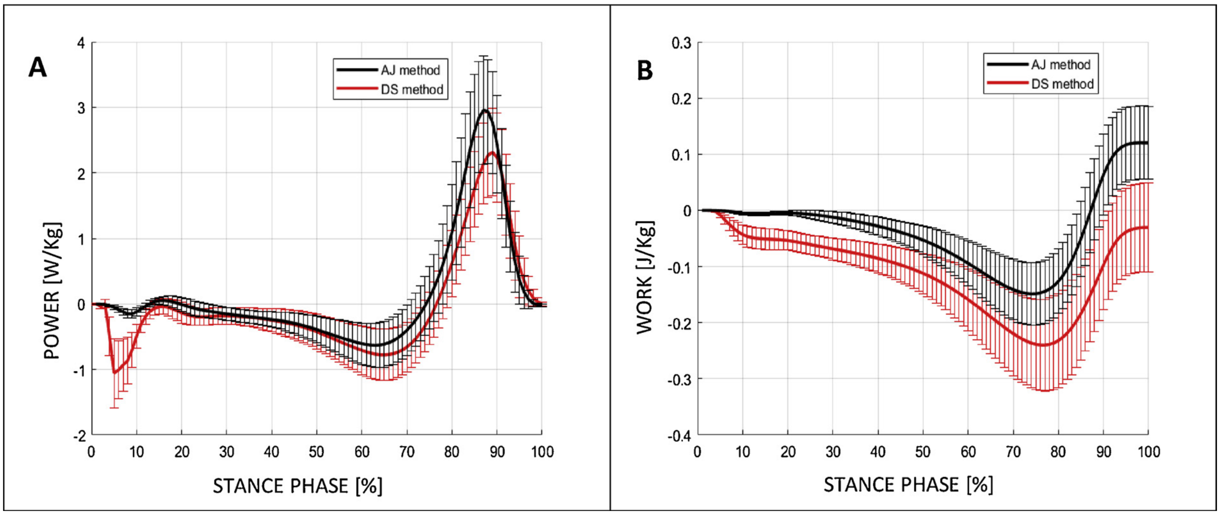

Fig. 2. Average curves (+/- 1 standard deviation) of Power (A) and Work (B) obtained at normal velocity from all subjects in barefoot condition (black line: AJ method; red line: DS method) (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

in high-heel shoes was the only condition in which the average final work was negative with both methods. All the evaluated parameters were different across the two methods also in high-heels condition, except for the second minimum of the power and for the work produced (Table 2).

4. Discussion

In this study, we compared the DS method and the AJ method for power and work calculation during different walking conditions. The results of previous studies [7–12] were confirmed and were corroborated by the analysis of a larger cohort of healthy young subjects, walking at different velocities, either barefoot and with shoes. These data can provide a reference for future studies. Of note, the DS method does not require a complex marker protocol, and thus it can be proposed for a wider clinical use. The application of this method is also feasible to data collected with the most consolidated protocols. Indeed, all of them can provide the 3D kinematics of the shank, the ground reactions forces and the ankle joint moments that are needed for the ankle-foot power computation by the DS method. Thus, the application in clinical practice would not require any substantial change in the commonly adopted procedure. The advantage is that the power and the work absorbed and produced by foot structure are assessed more comprehensively [12], with both qualitative and quantitative difference with respect to the AJ method. Previous studies [12,17–19] have shown that the AJ method overestimates the power produced at late stance phase and underestimates the absorbed one. Our results showed that during walking barefoot at natural speed the peak of power production was 29% higher with the AJ method than with the DS method. The absorbed power in the load acceptance phase was instead only 20% of that obtained with the DS method, and at the midstance phase it was 16% lower. Similar results were obtained for all walking conditions and confirmed the above observations. Our findings, in terms of the generated power, are also consistent with data obtained with more complex multi-segmental foot models [1718]. Indeed, the value of the power peak we obtained with the DS method (2.4 W/Kg) was very similar to the one reported by Dixon and colleagues [17] obtained with the Oxford Foot Model (2.3 W/Kg).

The differences in power estimation between the DS and AJ methods became striking in the high heels walking condition during the load acceptance phase. With the DS method, we found a double-peaked curve, likely due to the gap between the heel and the forefoot of the shoe, which split the load acceptance into two parts (first the contact of the heel and then the contact of the forefoot). The assumptions underlying the foot model of the AJ method prevented it from capturing these dynamics.

An analysis that takes into account the foot deformability and the movements out of the flexion/extension plane can be relevant for clinical applications. Indeed, it can reveal interesting features of the pathological gait related to alterations of the mechanical structure of

Table 1
Values of the main parameters of power and work curves during walking at three different velocities; all comparisons across the two methods were statistically significant ($p < 0.05$). Values are shown as *mean (standard deviation)*.

	SLOW		NORMAL		FAST	
	AJ	DS	AJ	DS	AJ	DS
Power peak [W/Kg]	2.19 (0.64)	1.70 (0.51)	3.15 (0.68)	2.44 (0.56)	4.21 (0.84)	3.64 (0.78)
First Min Power [W/Kg]	−0.18 (0.07)	−0.65 (0.29)	−0.23 (0.09)	−1.16 (0.47)	−0.47 (0.22)	−2.60 (0.88)
Second Min Power [W/Kg]	−0.70 (0.24)	−0.82 (0.29)	−0.79 (0.30)	−0.94 (0.36)	−0.70 (0.38)	−0.93 (0.48)
Final Work value [J/Kg]	0.06 (0.05)	−0.05 (0.07)	0.10 (0.06)	−0.03 (0.07)	0.21 (0.09)	0.045 (0.10)
Work Produced [J/Kg]	0.22 (0.06)	0.18 (0.05)	0.24 (0.05)	0.19 (0.04)	0.30 (0.07)	0.24 (0.06)
Work Absorbed [J/Kg]	−0.16 (0.05)	−0.23 (0.06)	−0.14 (0.05)	−0.22 (0.07)	−0.08 (0.05)	−0.20 (0.07)

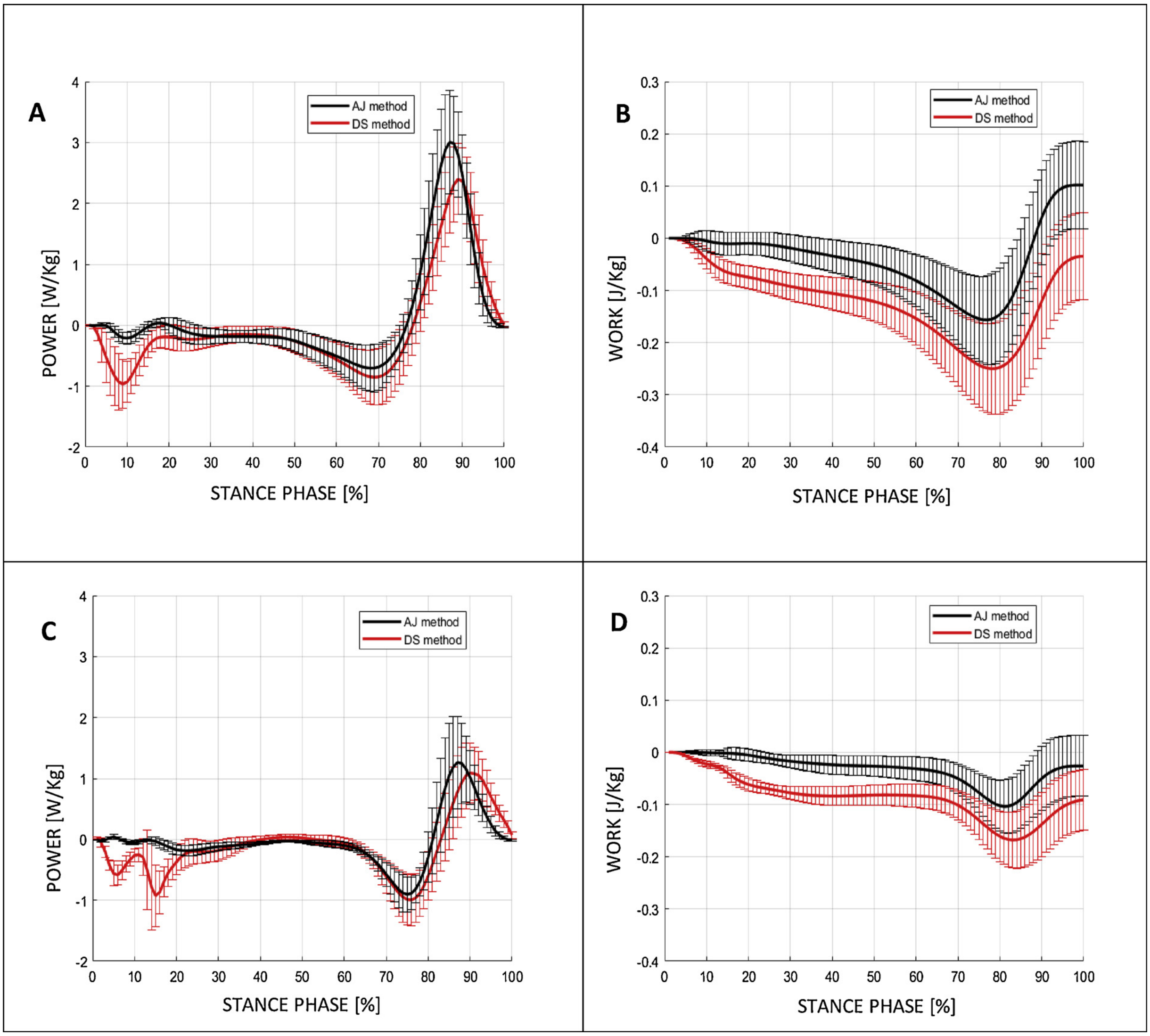

Fig. 3. Average curves (+/- 1 standard deviation) of Power (A, C) and Work (B, D) obtained at normal velocity from all subjects walking in sports shoes (A, B) and high heels (C, D) (black line: AJ method; red line: DS method) (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

Table 2
Values of the main parameters of power and work curves during walking in sport shoes and high heels; all comparisons across the two methods were statistically significant ($p < 0.05$) except for the Second Min Power and the Work Produced during walking with high heels. Values are shown as *mean (standard deviation)*.

	SPORTS SHOES		HIGH HEELS	
	AJ	DS	AJ	DS
Power peak [W/Kg]	3.09 (0.69)	2.49 (0.57)	1.37 (0.60)	1.18 (0.43)
First Min Power [W/Kg]	−0.30 (0.09)	−1.08 (0.44)	−0.22 (0.09)	−1.31 (0.61)
Second Min Power [W/Kg]	−0.78 (0.33)	−0.95 (0.39)	−0.98 (0.25)	−1.08 (0.40)
Final Work value [J/Kg]	0.09 (0.06)	−0.03 (0.75)	−0.02 (0.05)	−0.09 (0.06)
Work Produced [J/Kg]	0.24 (0.05)	0.19 (0.04)	0.08 (0.04)	0.08 (0.03)
Work Absorbed [J/Kg]	−0.15 (0.05)	−0.23 (0.07)	−0.11 (0.03)	−0.17 (0.05)

the foot and of the neural-motor control.

In the case of patients with cerebral palsy, for example, a common feature is the equine foot. This anomaly is characterized by enhanced plantarflexion which leads to toe-walking and can be associated to structural deformities, spasticity of the triceps surae muscles, contractures. The quantification of the ankle-foot power during the stance phase, computed as with the DS method, is fundamental to understand the mechanical conditions of the foot and to plan for proper interventions. There are also many orthopaedic pathologies, such as flatfoot, arthritis, tendonitis which affect the feet and would benefit from a reliable quantitative measure of power for treatment planning and assessment. In the field of prosthetics and orthotics, the power parameters are fundamental to assess the effectiveness of these devices. For example, prosthetic feet for lower limb amputations often incorporate the concept of energy store and return (ESR devices). As the name suggests, these prosthetic feet absorb energy in the loading phase and return it, partially, during the push off phase. As already demonstrated by previous studies [7,12], a correct power calculation is fundamental to assess the efficiency of the system and to analyse the effects of the aesthetic cover and shoes. This can have a relevant effect on the criteria of prosthetic design.

We have shown that similar values of power computed by the DS method can also be obtained by increasing the complexity of the multi-segmental model, as in the nine-segments model proposed by MacWilliams and colleagues [18]. However, the advantage of the DS method with respect to the multi-segmental models is that it can be easily introduced in the clinical practice, without changing the common acquisition procedure. In addition, if the pathologic conditions present bone deformities and make the positioning of markers on the foot particularly difficult, the DS method can provide ankle-foot power measurement without markers placed on the foot. This can be particularly appreciated when evaluating children or analysing shod walking and walking with ankle-foot orthoses.

Conflicts of interest statement

On behalf of all authors, the corresponding author states that there is no conflict of interest.

Author contributions

VF and CAF designed the experiment. CAF recruited the participants. VF and LH acquired the data. VF, LH and CP analyzed the data. VF, CP and CAF wrote the article. CP and CAF reviewed the manuscript.

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