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The impact of walking speed on the kinetic behaviour of different foot joints

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ABSTRACT

Background: The foot and ankle complex consists of multiple joints which have been hypothesized to fulfill a significant role in the lower limb kinetic chain during human locomotion. Walking speed is known to affect the lower limb kinetic chain function. Yet, this effect still has to be investigated throughout multiple joints of the foot and ankle complex.

Research question: What is the effect of walking speed on the kinetic behaviour of multiple joints of the foot and ankle complex?

Methods: This observational cross-sectional study investigated 15 asymptomatic male subjects. A three-and four-segment kinetic foot model was used to calculate power output and mechanical work during normal and high walking speed. One-dimensional Statistical Parametric Mapping (1D-SPM) linear regression was performed to examine the relationship between walking speed and kinetic data. Effect size calculations (Cohen's D) were included to quantify the amount of effect that walking speed has on power output and mechanical work in multiple foot joints.

Results: Three-segment kinetic measurements showed a significant positive correlation between walking speed and power output in the ankle ($p = 0.003$) and first metatarsophalangeal joint ($p = 0.0007$). Peak power generation increased in the ankle ($d = 1.59$), chopart ($d = 1.51$) and first metatarsophalangeal ($d = 1.25$) joints during high-speed walking. The three joints combined produced net +0.097 J/kg in normal and +0.201 J/kg in high-speed walking. Four-segment kinetic measurements showed a significant positive correlation between walking speed and power output at the ankle ($p = 0.036$), chopart ($p = 0.0001$), lisfranc ($p < 0.0001$) and first metatarsophalangeal ($p = 0.0063$) joints. Peak power generation increased in the ankle ($d = 1.32$), chopart ($d = 1.27$), lisfranc ($d = 1.22$) and first metatarsophalangeal ($d = 1.47$) joints during high-speed walking. Four joints combined produced net +0.162 J/kg in normal and +0.261 J/kg in high-speed walking.

Significance: These results add additional insight into foot function during increased walking speed.

1. Introduction

Improving our insight into the complex behaviour of the lower limb kinetic chain during human locomotion requires the analysis of power output and mechanical work. Multi-segment kinetic foot models were designed to investigate such measures at the level of the foot [1–4]. Kinetic models typically use inverse dynamics approaches that partition the total ground reaction forces across foot joints. Two distinct methods have been used: 1) a proportionality scheme applied on the combination of pressure and force plate data [1,5] and 2) a targeted foot placement on two adjacent force platforms [2].

Both approaches within multi-segment kinetic foot models are still explorative in nature, though the importance should not be underestimated as they can further improve our knowledge on both asymptomatic and symptomatic human locomotion [1,3,6]. For example, the different foot joints seem to provide an energy-neutral system, meaning that the ratio between positive and negative work produced over one stance phase is close to zero (−0.012 ± 0.054 J/kg) during normal walking speed [7]. Recently, the clinical relevance of multi-segment kinetic foot modelling was demonstrated in research on planovalgus feet in children, where ankle and midfoot power generation was found to be reduced with respectively 38% and 37% in children with

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planovalgus feet compared to control subjects [3]. Furthermore, a plethora of studies have identified potential parameters that can alter the foot and lower limb kinematics and kinetics, among which musculoskeletal pathologies [8], neuromuscular fatigue [9] and spatio-temporal parameters, such as step width [10], strike pattern [11] and walking speed [12–14].

With respect to the latter, some research has been published on how walking speed may affect lower limb kinetics. Farris and Sawicki found no difference in the proportion of hip, knee and ankle joint power to total power across different walking speeds [15]. Ebrahimi et al. suggested that positive relative ankle work and negative relative distal foot work during stance phase do increase significantly with increased walking speed [16]. Literature furthermore states that increased walking speeds demand increased power output and mechanical work in lower limb joints [12,15,16]. Yet to our knowledge, no study so far has actually investigated this influence by making use of three- and four segmented kinetic foot models.

These insights would, however, substantially aid in improving our knowledge on foot biomechanics and could form a foundation in diagnostics, prevention and management of pathologies affecting foot function. For example, ankle osteoarthritis is often associated with reduced walking speeds [17–19] and it is therefore of clinical importance to determine the influence of this phenomenon on the kinetics of multiple foot joints. Furthermore, clinical and research domains focusing on foot orthotics, prostheses and exoskeletons may benefit greatly from these new insights.

The aim of this study was to establish the effect of walking speed on the kinetic behaviour in multiple joints of the foot. A three-segment kinetic foot model was used because of its psychometric characteristics [3] and opportunity for comparison with similar studies. Even though the four-segment kinetic foot model [1] is to date less investigated, it was implemented in the current study as it may give additional information on forefoot function by measuring chopart and lisfranc joint kinetics. We hypothesize that increasing walking speeds demand higher mechanical work and power output in all joints of the foot. Furthermore, in contrast to the earlier described so-called energy-neutral foot system during walking at comfortable speed [7], we do not expect to encounter this phenomenon during increased walking speeds [15,16].

2. Methods

2.1. Participants

Data from 15 healthy asymptomatic male subjects without history of musculoskeletal lower limb injuries (ages 35.8 ± 6.06 years, height 1.78 ± 0.05 m, body mass 73.7 ± 9.9 kg; mean ± standard deviation) were included. Participants signed an informed consent and the ethical committee of the University Hospitals Leuven (S57147) approved the study.

2.2. Multi-segment foot modelling

The marker placement protocol of the Instituto Ortopedico Rizzoli Foot Model [20] was used for the integration of two specific multi-segment kinetic foot models [1]. The first model encompassed the three-segment kinetic foot model described by Bruening et al. [2] and later modified by Deschamps et al. [1] and includes the ankle, chopart and first metatarsophalangeal (MTP 1) joints [2,6]. The inter-segmental angle calculations in this model were as follows: ankle (Shank-Calcaneus angle), chopart (Calcaneus-metatarsus angle) and MTP 1 (First-metatarsophalangeal angle). The following joint center definitions were applied: ankle (midpoint between medial and lateral malleoli), chopart (midpoint between navicular and cuboid bone, with cuboid bone being projected at 2/3 distal distance between the peroneal tubercle and the base of fifth metatarsal) and MTP 1 (projection of MTP 1 marker ½ distance to the floor). The four-segment kinetic foot model, described by Deschamps et al., encompasses a shank segment as well as four foot joints: ankle, chopart, lisfranc and MTP 1 [1]. To facilitate readability, inter-segmental angle calculations were defined as follows: ankle (Shank-Calcaneus angle), chopart (Calcaneus-Midfoot angle), lisfranc (Midfoot-Metatarsus angle) and MTP 1 (First-metatarsophalangeal angle). Joint center definitions for this four-segment foot model were: ankle (midpoint between medial and lateral malleoli), chopart (midpoint between the cuboid and the navicular bone), lisfranc (base of the second metatarsal) and MTP 1 (projection of MTP 1 marker ½ distance to the floor).

2.3. Data collection

After instrumentation, participants carried out five walking trials at their self-selected walking speed over an instrumented walkway of 10 m. This walkway integrates, in the middle, a plantar pressure platform (Footscan, dimension 0.5 m x 0.4 m, 4096 sensors, 2.8 sensors per cm^2, Rsscan International, Olen, Belgium), placed on top of a force plate (Advanced Mechanical Technology Inc, Watertown, MA, USA). Data of the pressure and force plate were synchronized using an Rsscan 3D box. A passive optoelectronic motion analysis system (Vicon Motion System Ltd., Oxford Metrics, UK) consisting of 10 T-10 cameras (sampled at 100 Hz) surrounded the walkway to track kinematic data. After this, five trials were collected where participants walked at their highest possible walking speed without running.

2.4. Data analysis

Data analysis was identical with that described by Deschamps et al. [1]. The specific data analysis used in this study is succinctly described in supplementary material (word-file: data analysis). All one-dimensional data were normalized to 100% of the stance phase and the mean was calculated based on the five recorded trials.

One-dimensional Statistical Parametric Mapping (SPM) linear regression [21] was performed using open-source code (www.spm1d.org) in Matlab (R2014a, 8.3.0.532, The Mathworks Inc, Natick, MA) to examine the relationship between walking speeds and kinetic data. Cohen D's effect size of increased walking speed was determined by calculating the mean difference between normal and increased walking speed and dividing this outcome by the pooled standard deviation for following variables: peak power generation and absorption and positive and negative work. Effect size was interpreted according to Ellis' thresholds, in which an effect size of d = 0.20 is seen as a small effect, d = 0.50 a medium effect, d = 0.80 a large effect and d = 1.30 a very large effect [22].

3. Results

The average normal walking speed (NWS) was equivalent to 1.28 (± 5%) m/s and the average high walking speed (HWS) to 1.76 (± 4%) m/s, resulting in a 37.5% difference between both walking speeds.

3.1. Power calculations associated to the three-segment kinetic foot model

A significant positive correlation existed between walking speed and ankle joint power during the mid-stance phase ($p = 0.003$) and terminal stance phase ($p < 0.0001$) as well as in the MTP 1 joint during pre-swing phase ($p = 0.0007$). Critical t-values for the SPM tests were respectively $t = 3.169$ and $t = 3.293$ (Fig. 1).

Ankle peak power generation increased from 2.79 (± 0.54) Watt/kg in NWS walking to 3.77 (± 0.68) Watt/kg in HWS walking, which is a very large effect (d = 1.59). Also, peak power generation in both the chopart and MTP 1 joints increased from respectively 0.9 (± 0.24) Watt/kg and 0.29 (± 0.24) Watt/kg in NWS to respectively 1.38 (± 0.38) Watt/kg and 0.65 (± 0.33) Watt/kg in HWS, which is

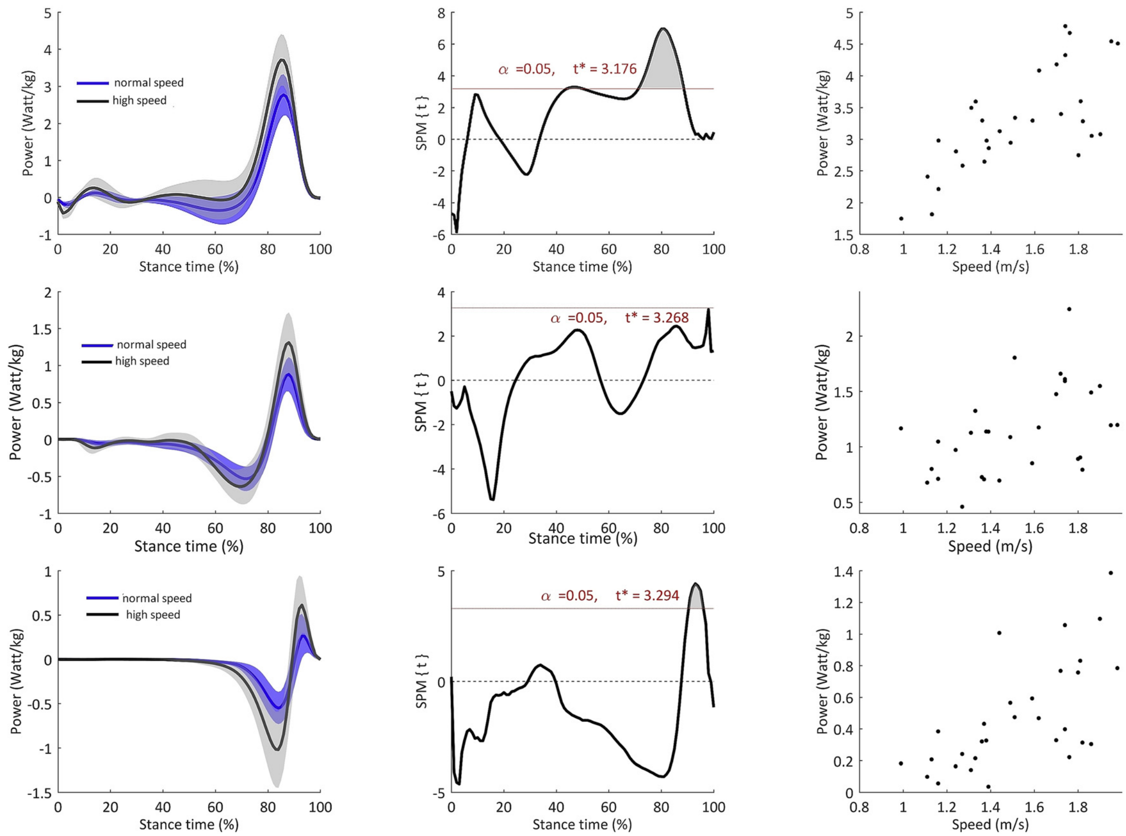

Fig. 1. Left: kinetic waveforms of the ankle (upper), chopart (middle) and MTP 1 (lower) joints during stance phase of normal (blue line) and high (grey line) pace walking. Standard deviations are visualized as bands. Middle: graphical visualization of outcomes of the one-dimensional statistical parametric mapping linear regression statistics. Grey bands visualize a significant difference between waveforms. Right: relationships between normal and high walking speeds (x-axis) and peak power of the ankle (upper), chopart (middle) and MTP 1 (lower) joints during stance phase of walking. MTP 1: First Metatarsophalangeal joint. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

respectively a very large (d = 1.51) and large effect (d = 1.25).

Peak power absorption increased to a small degree (d = 0.4) in the ankle joint and to a medium degree (d = 0.53) in the chopart joints in the HWS condition. Peak power absorption at MTP 1 increased with a very large effect (d = 1.59) (Table 1).

The ankle, chopart and MTP 1 joints respectively contributed 38%, 40% and 22% of the total power absorption during NWS, whereas during HWS, this participation changed respectively to 28%, 37% and 35% (Table 3).

3.2. Power calculations associated to the four-segment kinetic foot model

A significant positive correlation was found between walking speed and power output of the ankle during loading response (p = 0.036) and terminal stance phase ($p < 0.0001$). The same correlation was found for the chopart joints during terminal stance (p = 0.0001), the lisfranc joints during mid-stance ($p < 0.0001$) and in the MTP 1 joint during pre-swing (p = 0.0063). Critical t-values for the SPM tests were respectively t = 3.157, t = 3.227, t = 3.324 and t = 3.305 (Fig. 2).

In the ankle, peak power generation increased from 2.60 (± 0.49) Watt/kg during NWS to 3.35 (± 0.64) Watt/kg during HWS. For chopart, we found a rise from 1.19 (± 0.31) Watt/kg to 1.71 (± 0.49) Watt/kg, whereas peak power generation in the lisfranc joints increased from 0.4 (± 0.11) Watt/ kg to 0.59 (± 0.19) Watt/kg. Peak power generation in MTP 1 rose with an average of 0.35 W/kg. These observed changes in peak power generation of the ankle, chopart, lisfranc and MTP 1 joints had large and very large effect sizes of respectively d = 1.32, d = 1.27, d = 1.22 and d = 1.47 (Table 1).

Table 1
Effect of walking speed on power output (Watt/kg).

Variable	Mean (± SD)		*Mean change [95% C.I.]*		Variable	Mean (± SD)		*Mean change [95% C.I.]*	
	Normal speed	*High speed*		*Effect size*		*Normal speed*	*High speed*		*Effect size*
Peak generation (Watt/kg)					Peak generation (Watt/kg)				
Ankle	2.79 (± 0.54)	3.77 (± 0.68)	0.98 [0.68, 1.28]	1.59	*Ankle*	2.60 (± 0.49)	3.35 (± 0.64)	0.75 [0.42, 1.08]	1.32
Chopart	0.9 (± 0.24)	1.38 (± 0.38)	0.48 [0.34, 0.62]	1.51	*Chopart*	1.19 (± 0.31)	1.71 (± 0.49)	0.53 [0.35, 0.7]	1.27
MTP 1	0.29 (± 0.24)	0.65 (± 0.33)	0.36 [0.26, 0.46]	1.25	*Lisfranc*	0.4 (± 0.11)	0.59 (± 0.19)	0.18 [0.09, 0.27]	1.22
					MTP 1	0.2 (± 0.13)	0.54 (± 0.3)	0.35 [0.24, 0.46]	1.47
Peak absorption (Watt/kg)					Peak absorption (Watt/kg)				
Ankle	0.51 (± 0.22)	0.58 (± 0.11)	0.06 [-0.05, 0.17]	0.4	*Ankle*	0.57 (± 0.27)	0.64 (± 0.18)	0.07 [-0.06,0.20]	0.31
Chopart	0.55 (± 0.17)	0.66 (± 0.24)	0.11 [0.05, 0.17]	0.53	*Chopart*	0.36 (± 0.19)	0.45 (± 0.27)	0.09 [0.02, 0.16]	0.39
MTP 1	0.57 (± 0.19)	1.1 (± 0.43)	0.52 [0.34, 0.7]	1.59	*Lisfranc*	0.04 (± 0.05)	0.12 (± 0.14)	0.08 [0.03, 0.13]	0.76
					MTP 1	0.75 (± 0.27)	1.31 (± 0.56)	0.56 [0.36, 0.76]	1.27

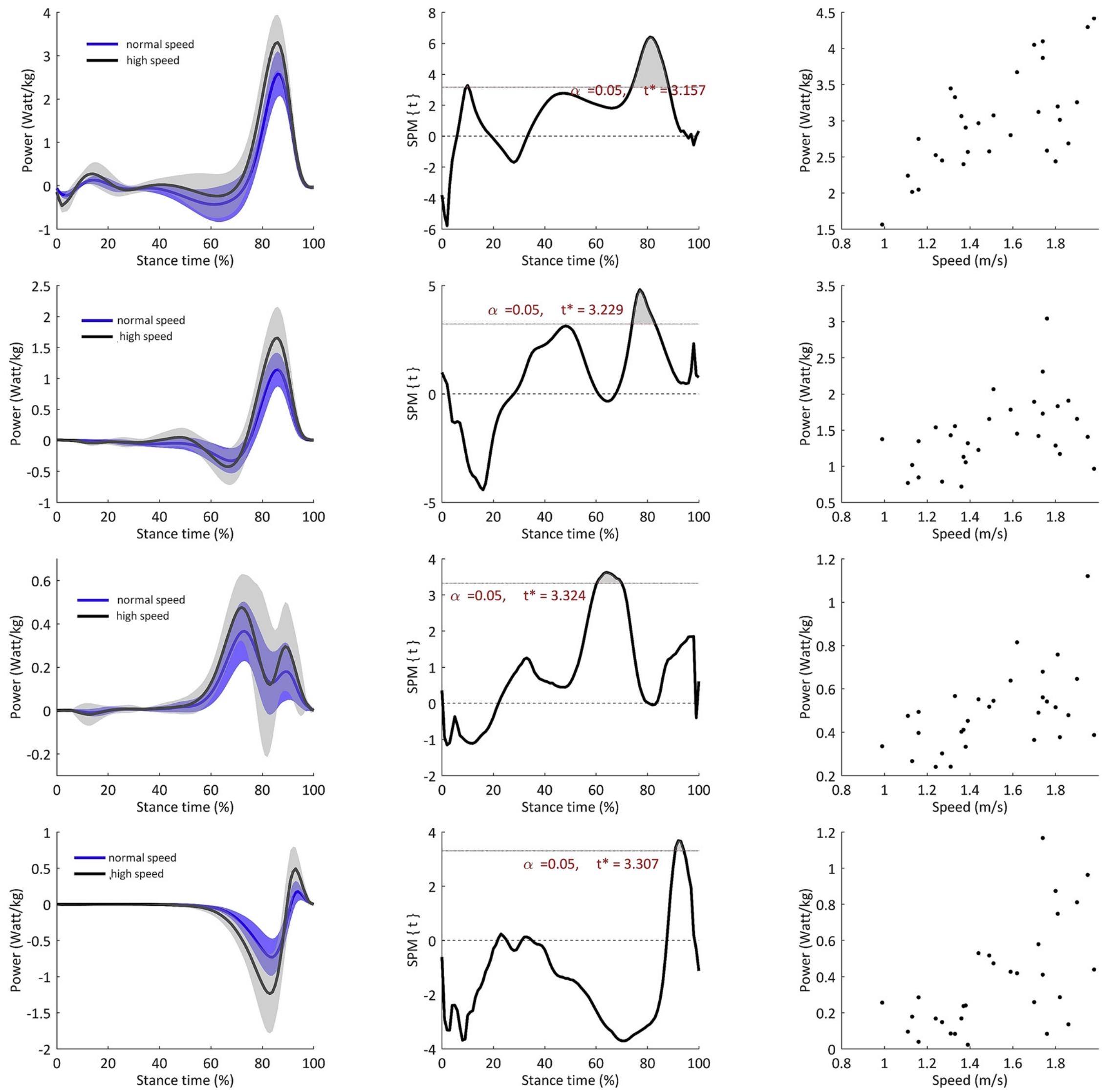

Fig. 2. Left: kinetic waveforms of the ankle (first row), Chopart (second row), Lisfranc (third row) and MTP 1 (fourth row) joints during stance phase of normal (blue line) and high (grey line) pace walking. Standard deviations are visualized as bands. Middle: graphical visualization of outcomes of the one-dimensional statistical parametric mapping statistics. Grey bands visualize a significant difference between kinetic waveforms. Right: relationships between normal and high walking speeds (x-axis) and peak power of the ankle (first row), Chopart (second row), Lisfranc (third row) and MTP 1 (fourth row) joints during stance phase of walking. MTP 1: First Metatarsophalangeal joint. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Regarding peak power absorption during the stance phase of walking with HWS, the ankle, chopart, lisfranc and MTP 1 segments demonstrated increases of respectively 0.07 W/kg, 0.09 W/kg, 0.08 W/kg and 0.56 W/kg, with corresponding small, medium and large effect sizes of respectively d = 0.31, d = 0.39, d = 0.76 and d = 1.26 (Table 1).

The relative influence of joints to the total power generation remained close to equal for both NWS and HWS, with joint contributions of 58% in the ankle, 25% in the chopart joints, 15% in the lisfranc joints and 2% in MTP 1. The relative joint influence of power absorption changed from NWS to HWS, with a decrease of 12% in the ankle and an increase of 11% in MTP 1 (Table 3).

3.3. Work calculations associated to the three-segment kinetic foot model

During NWS walking, the ankle, chopart and MTP 1 joints produced a total positive work of 0.32 J/kg (ankle = 0.25 J/kg, chopart = 0.06 J/kg and MTP 1 = 0.01 J/kg) and negative work of -0.22 J/kg (ankle = -0.09 J/kg, chopart = -0.09 J/kg and MTP 1 = -0.05 J/kg) over the complete stance phase. Consequently, mechanical work of these three segments combined to a net amount of + 0.1 J/kg (Table 2).

In HWS walking, a net mechanical work of + 0.2 J/kg was produced in all three foot joints during stance phase, due to a positive work total of 0.43 J/kg (ankle = 0.33 J/kg, chopart = 0.08 J/kg and MTP 1 = 0.02 J/kg) and a negative work total of -0.23 J/kg (ankle = -0.06 J/kg, chopart = -0.09 J/kg and MTP 1 = -0.08 J/kg). The mean net change of 0.10 J/kg between NWS and HWS walking came with a very large effect size (d = 3.45) (Table 2).

3.4. Work calculations associated to the four-segment kinetic foot model

The ankle (0.23 J/kg), chopart (0.1 J/kg), lisfranc (0.06 J/kg) and MTP 1 (0.01 J/kg) joints combined produced a total positive work of 0.39 J/kg during NWS. During HWS, the total mechanical work produced in the foot increased to 0.51 J/kg (ankle = 0.29 J/kg, chopart = 0.13 J/kg, lisfranc = 0.07 J/kg and MTP 1 = 0.02 J/kg). As for the negative mechanical work, 0.23 J/kg was absorbed in the foot

Table 2
Effect of increased walking speed on work (J/kg).

Variable	Mean (± SD) Normal pace	Mean (± SD) High Pace	Mean change and [95% C.I.]	Effect size
Positive work (J/kg)				
Ankle	0.25 (± 0.05)	0.33 (± 0.06)	0.08 [0.04, 0.12]	1.39
Chopart	0.06 (± 0.02)	0.08 (± 0.02)	0.02 [0.00,0.04]	0.95
MTP 1	0.01 (± 0.01)	0.02 (± 0.01)	0.01 [0.00, 0.02]	1.39
Negative work (J/kg)				
Ankle	-0.09 (± 0.04)	-0.06 (± 0.03)	0.02 [-0.00, 0.5]	0.62
Chopart	-0.09 (± 0.03)	-0.09 (± 0.03)	0.00 [-0.02, 0.03]	0.11
MTP 1	-0.05 (± 0.02)	-0.08 (± 0.04)	0.03 [0.01, 0.06]	0.98
Net result	+ 0.1 J/kg	+ 0.20 J/kg	0.10 [0.08, 0.12]	3.45

Variable	Mean (± SD) Normal pace	Mean (± SD) High Pace	Mean change and [95% C.I.]	Effect size
Positive work (J/kg)				
Ankle	0.23 (± 0.05)	0.29 (± 0.06)	0.07 [0.03, 0.11]	1.21
Chopart	0.10 (± 0.03)	0.13 (± 0.03)	0.03 [0.01, 0.05]	1.07
Lisfranc	0.06 (± 0.02)	0.07 (± 0.03)	0.01 [-0.01, 0.03]	0.35
MTP 1	0.01 (± 0.01)	0.02 (± 0.01)	0.01 [0.00, 0.02]	1.16
Negative work (J/kg)				
Ankle	-0.1 (± 0.05)	-0.08 (± 0.04)	0.02 [-0.01, 0.05]	0.44
Chopart	-0.05 (± 0.03)	-0.05 (± 0.03)	0.00 [-0.02, 0.02]	0.03
Lisfranc	-0.00 (± 0.00)	-0.01 (± 0.01)	0.01 [0.00, 0.01]	0.75
MTP 1	-0.07 (± 0.03)	-0.11 (± 0.05)	0.03 [0.00, 0.06]	0.8
Net result	+ 0.16 J/kg	+ 0.26 J/kg	0.1 [0.08,0.12]	3.29

Table 3
Relative influence of foot joints on power output (%) during normal and high walking speed.

	Power generation		Power absorption	
	Normal speed	*High speed*	*Normal speed*	*High speed*
Ankle	79	77	38	28
Chopart	18	18	40	37
MTP 1	3	5	22	35
Total Foot	100	100	100	100
Ankle	58	58	43	31
Chopart	25	25	22	21
Lisfranc	15	14	2	4
MTP 1	2	3	33	44
Total Foot	100	100	100	100

during NWS (ankle = -0.1 J/kg, chopart = -0.05 J/kg, lisfranc = -0.01 J/kg and MTP 1 = -0.08 J/kg). The latter increased to 0.25 J/kg during HWS (ankle = -0.08 J/kg, chopart = -0.05 J/kg, lisfranc = -0.01 J/kg and MTP 1 = -0.11 J/kg). This resulted in a net mechanical work of + 0.16 J/kg for NWS and + 0.26 J/kg for HWS, embodying a very large effect (d = 3.29) (Table 2).

4. Discussion

This study investigated the impact of walking speed on kinetics in multiple joints of the foot. Using an inverse dynamics approach, we partitioned the mechanical power and work throughout three and four-segment modelling. Computations associated with the three segment kinetic foot model were incorporated to allow for better comparisons with similar research [1]. The four-segment kinetic foot model granted exploratory insight into the kinetic behaviour of the chopart (transverse tarsal) and lisfranc (tarsometatarsal) joints.

Both peak power generation and positive work increase with increasing walking speed. As already proven by forward dynamic simulations, positive muscle work during the stance phase is primarily provided by ankle plantar-flexors [14]. This may explain the significant increase of joint power generation as well. Based on the relative joint influence calculations one can conclude that the absorption at the ankle declines with increasing walking speed, whereas the influence of the MTP 1 on relative power absorption increases. Temporal visualization of this phenomenon showed that during NWS, a distinct amount of power absorption in the ankle takes place during 35–75% of the stance phase, whereas this is not the case during HWS. A possible explanation for this phenomenon is that due to earlier forefoot contact, the MTP 1 joint tends to rapidly take over the absorbing role of the ankle [23] towards the second half of stance phase. Chopart and MTP 1 joints combined produce only up to 21% and 23% of the total power generation during respectively NWS and HWS. We believe that there might be two possible explanations for the smaller amount of propulsive behaviour in the more distal located anatomical entities compared to the more proximal entities. First, the insertions of the major ankle plantar flexors (m. gastrocnemius, m. soleus and m. plantaris) at the calcaneal bone have more influence at the ankle joints. Second, the intrinsic musculature in chopart and MTP 1 joints are local stabilisers with small cross-sectional areas, therefore producing small rotational moments [24]. Joints distal to the ankle tend to absorb relatively more power than the ankle joint. This is largely due to the anatomy of the passive structures in the foot, where the four foot arches (medial and lateral longitudinal arches and posterior and transverse metatarsal arches) coalesce into a functional half dome [25] responsible for flexibly adapting to load changes during dynamic activities [24]. This power absorption could also be explained by the extension mobility of the metatarsal-phalangeal joints [2,4,26] or the compressibility of the fat pad underneath the metatarsal heads during push-off [27].

Calculations with the four-segment kinetic foot model suggested that changing from NWS to HWS does not alter the relative power output contribution of the chopart and lisfranc joints, despite significant increases in peak power generations and absorptions. Since the increase in both segments is found during the propulsion phase (> 50% of stance phase cycle), it is plausible to say that the Windlass mechanism is even more predominant during HWS. This mechanism is activated when the metatarsophalangeal joints extend during propulsion and lengthens the plantar fascia (and possibly other muscle-tendon structures that comprise the longitudinal arch), which induces tension and facilitates positive work generation at the chopart and lisfranc joints during push-off [28]. Influence of the lisfranc joints on foot energetics during higher walking speeds tends to be rather small in comparison to the chopart joints in the four-segment model, which is predominantly caused by the stiffness of the tarsometatarsal joints [1]. The latter phenomena demonstrate the importance of measuring joint kinetics using a four-segment kinetic foot model, since the impact on foot function induced by the chopart and lisfranc joints can be better understood. We further believe that implementing a four-segment model in future studies on foot joint kinetics will aid in further documenting the complex foot function during walking.

Interestingly, we did not observe a so-called ‘energy-neutral system’ of the foot and ankle subsections during steady-gait, in contrast to the study by Takahashi et al. [7]. In fact, our results showed a positive net mechanical work outcome of +0.10 J/kg when combining all subsections of the foot (three-segment kinetic foot model). A possible explanation might be the different methodologies used in both studies. Where we used a rigid-body mechanics approach, Takahashi et al. opted for a non-rigid body mechanics approach [29]. The latter model incorporates, but does not distinguish, the absorbing function of the fat pads located underneath the heel and metatarsal heads, which negatively impacts the net result of mechanical work outcome in the foot. Despite the observed difference in net mechanical work during NWS, it is of interest to note the effect of increased walking speed on the net mechanical work, which rises towards 0.20 J/kg. This may have significant repercussions in the clinical decision-making of foot pathologies and design of foot orthotics as discussed in the study by Takahashi et al [7].

A limitation to the study is that we only collected data at two walking speeds, namely 1.28 m/s and 1.76 m/s. This way, no assumptions can be made on slow walking as observed in pathological situations, where for example, patients with ankle osteoarthritis produce an average walking speed of 1.10 m/s [30]. Another limitation is that the used multi-segment kinetic foot model is based on a proportionality scheme that carries an estimation-error of maximum 3% for all shear forces [3].

This study provides a first attempt towards gaining additional insight into the kinetic characteristics in multiple joints of the foot during normal and high-speed walking. Increased walking speeds demand higher power output and mechanical work in different foot joints and contrary to our hypothesis, the foot joints must produce net positive mechanical work to walk at normal speed. Since walking speed does result in different foot kinetics, future studies ought to be cautious when comparing asymptomatic control subjects with symptomatic subjects if their walking speed differs.

Conflict of interest

The authors have no conflict of interest to report.

The authors declare that no financial and personal relationship exist which could have influence (bias) their work.

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Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.gaitpost.2018.12.022.

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