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Normative Data of the External Work of Individual Limbs and of the Distribution of Joint Work During Stair Crossing

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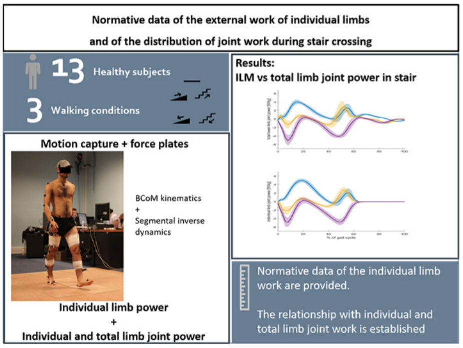
HIGHLIGHTS

- COM energy rate and summed joint work are consistent during stance in stair.
- Walking conditions significantly influence both COM and joint works production.
- Mechanical power patterns are different in stair versus slope and level walking.

GRAPHICAL ABSTRACT



Normative data of the external work of individual limbs and of the distribution of joint work during stair crossing

13 Healthy subjects

3 Walking conditions

Motion capture + force plates

BCoM kinematics
+
Segmental inverse dynamics

Individual limb power
+
Individual and total limb joint power

Results:
ILM vs total limb joint power in stair

Normative data of the individual limb work are provided.

The relationship with individual and total limb joint work is established

ABSTRACT

Background: Stair walking requires to elevate or lower the body center of mass and results in increased muscle contractions and consumed energy compared to level walking. Mechanical work produced by the body can be quantified through Individual Limb Method and the summed lower limb joint work but there does not exist normative data of these works in stair ascent and descent compared to slope ascent and descent of the same individuals.
Methods: Upstair and downstair walking were investigated at 0%, 5% and 12% inclinations and compared to upslope and downslope walking for thirteen able-bodied volunteers. Lower limb joint and individual limb powers and works were compared across walking conditions.
Findings: Work production and absorption required to elevate or lower the center of mass directly depend on the inclination to be crossed (about 0.35 J/kg for 5% slope, 0.9 J/kg for 12% slope and 1.6 J/kg for stair). However, the distribution among joints and between gait phases is different when considering stair versus slope walking. In particular, the role of the knee is exacerbated for work production in stair ascent (45% of total work) as well as for work absorption in stair descent (61% of total work). Also, more work production/absorption is performed during the swing phase for stair walking then for slope walking.

Keywords:
Lower limb
Slope
Stair
Locomotion
Biomechanics

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Interpretation: This study provides reference data of the Individual Limb mechanical work performed during stair walking and show that this method can substitute to summed lower limb joint one during the stance phase of stair walking.

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1. Introduction

The energy required for locomotion largely varies depending on the activity, in particular when elevating or lowering the Body Center of Mass (BCoM) during slope or stair crossing and depends of factors such as mass, height and inclination. From a mechanical point of view, the variation of kinetic energy of the system is equal to the sum of the powers of external and internal mechanical actions. In motion analysis, the classical modeling of the human body consists in representing the body segments as rigid bodies articulated by the joints. Thus, considering the body as a mechanical system, the sources of internal power originate from the mechanical actions of muscles on the anatomical joints. Therefore, it should be theoretically possible to link the mechanical work necessary to perform a given motion to the metabolic energy that is consumed for the functioning of muscles generating joint motion. Even if direct relationship between overall metabolic energy estimated from oxygen consumption and mechanical work has not been yet fully proven in the literature [1–3], this theoretical framework justifies the relevance of quantifying mechanical work in the attempt to understand the energy flow during human body motion.

Mechanical work quantification can be done by integrating mechanical power over a given period of time. However, the computation of mechanical power of a mechanical system consists in an algebraic sum of powers, which assumes that the mechanical energy produced in a joint can be absorbed by another [4]. From a physiological point of view, it means that the muscle should be able to retrieve energy when working in an eccentric mode and even to release it. Such types of transfer have been considered as unlikely to occur [4]. In an attempt to get closer to the estimation of the metabolic cost of locomotion, some authors have proposed to sum the joint work no more in an algebraic sense but by using absolute values either considering individual joints [4–6] or individual limbs [7]. In the individual limbs paradigm, the underlying modeling of gait is a succession of pendulum-like phases where power is mainly generated or absorbed during the step-to-step transition [8–10]. The ILM power can be estimated from the ground reaction forces and moments and the velocity of the BCoM, both accessible from force plate recording only, that could make it easier to implement than the computation of joint powers. ILM has been used to quantify mechanical work in the context of amputation [9], [11], [12], total ankle arthroplasty [3], obesity [13], cerebral palsy [14] or post-stroke [15], [16]. Most of these studies investigated level ground walking.

The comparison of this estimation of the internal work with the one made from the multibody modeling has also been performed from experimental data [12], [17], [18] and from a theoretical approach [10] for level and slope walking. In addition, the description of joint work distribution has already been reported during slope [19] and stair walking [20] but has never been made for the same cohort of persons nor compared to the ILM work during stair crossing.

In this context, the aim of the present study is to provide reference data for the mechanical work developed by healthy subjects during stair crossing compared to level ground and slope walking. Particularly, the computation of mechanical work from ILM method had never been made before during stair crossing. Indeed, stair crossing is recognized as a very constraining condition that distinguishes from slope walking not only considering the equivalent inclination that could not be easily achieved by a slope but also because of specific locomotion strategy to face the disruption induced by step-by-step ascent or descent [20]. It can be hypothesized that the specific profile of mechanical power during stair crossing particularly during the step-to-step transitions can be captured by the computation from the ILM method. A secondary objective is to show the influence of the walking conditions on the different estimation of mechanical works. The novelty of the study therefore comes from both the use of ILM method in stair crossing and the comparison for the same population of several non-standard conditions.

2. Methods

2.1. Subjects

Thirteen healthy subjects participated in the study (11 M/2 F, mean ± standard deviation, age: 39 ± 16, height: 175 ± 8 cm, mass: 69 ± 9 kg). The protocol was approved by the local ethics committee and all participants gave their informed consent.

2.2. Protocol

Data collection was conducted in two different motion analysis laboratories with the same equipment, protocol, and operators. Subjects were equipped with 54 markers to record segmental and articular kinematics of the whole body with a motion analysis optoelectronic system (Vicon V8i, UK) sampled at 100 Hz. Markers were positioned on specific anatomical landmarks in accordance with the protocol described by Pillet et al. [21].

A static trial was performed to define a reference position and two pictures were taken simultaneous. Then, subjects were asked to walk at a comfortable self-selected speed in 7 conditions including level walking on a 9 m pathway, ascent and descent of a gentle slope at 5% (2.8°), ascent and descent of a steep slope at 12% (6.8°) ascent and descent of a four step stairway (28.5 cm run for 17.5 cm rise which corresponds to an equivalent inclination of 32° i.e. 63%). The inclination values were chosen both to correspond to current values in the literature [22–25] and to be consistent with accessibility normative standards (normal slope inferior to 5% and maximal authorized slope of 12%). The same process led us to select the number of stairs to have a complete stair gait cycle (more than 3 steps) as in [26], [27] and in accordance with the normative recommendation for public space (28 cm of run for 16 cm of rise). All situations were instrumented with two force plates (AMTI, USA) sampled at 1000 Hz. Each stair was instrumented thank to the design of the specific staircase [28]. Five successful trials were recorded for each condition, a trial being considered successful if each lower limb hit the device on at least one of the force plates.

2.3. Data analysis

Force plate data were resampled at 100 Hz to be synchronized with motion capture data. Spatio-temporal parameters and anatomical frames, segmental and articular kinematics and kinetics of the lower limbs (ankle, knee, and hip), pelvis and trunk were computed according to Pillet et al. [29] following the recommendations of the International Society of Biomechanics. An

Table 1
Gait spatio-temporal parameters (mean, SD, min and max of the entire group) for the different conditions.

Mean (SD) Min – Max	Level	5% (3°) slope		12% (7°) slope		Stairs	
		Up	Down	Up	Down	Up	Down
Velocity (m/s)	1.26 (0.10) 1.22 – 1.55	1.26 (0.17) 0.85 – 1.56	1.27 (0.17) 0.87 – 1.57	1.26 (0.16) 0.86 – 1.51	1.27 (0.19) 0.81 – 1.61	0.58 (0.07) 0.49 – 0.74	0.61 (0.08) 0.50 – 0.75
Step Length(m)	0.70 (0.04) 0.58 – 0.75	0.70 (0.06) 0.53 – 0.78	0.67 (0.06) 0.51 – 0.75	0.71 (0.07) 0.55 – 0.85	0.66 (0.08) 0.51 – 0.79	0.34 (0.03) 0.30 – 0.41	0.36 (0.06) 0.29 – 0.47
Double support (%)	11 (2) 8 – 13	12(2) 7 – 15	11(2) 6 – 14	12 (2) 8 – 15	10 (2) 7 – 13	11 (2) 4 – 14	13 (2) 9 – 16

anthropometric model of the external shape of the segments was also built from the pictures taken during the static pose and personalized using the method described by [29]. This model allowed the estimation of personalized body segment inertial parameters from segmental densities. The densities were taken from Dempster [30] except for the thorax, where a modified density taken from Amabile et al. [31] was used. Body segment inertial parameters including segment centers of mass then serve for the computation of joint moments and of the Body Center of Mass (BCoM). Joint moments were obtained from classical inverse dynamics method and were used to compute individual joint power. Thus, joint power was computed as the dot product of joint moment and relative angular velocity of the distal segment relatively to the proximal segment of the joint. The sum of the ankle, knee and hip power was also assessed and referred as the summed lower limb joints power.

The position of the Body Center of Mass (BCoM) was computed as the barycenter of the same segmental centers of mass and BCoM kinematics inferred from segmental kinematics during the motion. Then, BCoM velocity could also be obtained in the global reference frame by fourth order finite differentiation after filtering with a Butterworth second order filter at 5 Hz. Then, the mechanical power of each individual limb was assessed by the dot product of the velocity of the body center of mass and the resulting ground reaction forces on the considered lower limb. All power data were analyzed by cycle defined as the period of time between a contact with the floor of one given lower limb to the following contact of the same limb in all the studied conditions. Individual limb power was compared to the above mentioned summed lower limb joints power.

Positive and negative works were calculated by numerical integration of powers over time for the entire cycle, which gives two estimations of the mechanical positive and negative works performed i.e. the ILM and the summed joint ones.

The effects of the walking conditions were investigated using a one-way ANOVA with repeated measures (and Kruskall Wallis nonparametric test for the non-normally distributed parameter). Investigated parameters were the peaks of positive and negative power and the values of positive and negative works for the 5 computed powers (ankle, knee, hip, sum of the joint and ILM). When the ANOVA showed a difference between groups, *post hoc* Tukey's tests were used for pairwise post-hoc comparisons. On total, 27 comparisons (7 walking conditions) were made for each of the 5 positive and 5 negative parameters (3 joint powers, summed joint power, ILM power). For the sake of clarity, in the result part, we exhaustively present only the values of mechanical work for the summed joint power and ILM work results. We also gave the result of statistical analysis in the text for power peaks and reported the joint parameters that revealed no significant differences between stair walking conditions and one of the other conditions with a level of significance of $p<0.01$.

3. Results

The walking velocities, step lengths and double support durations are given for the group of 13 subjects in Table 1. The walking velocities were similar across conditions except for stair ascent and descent. The step lengths were equivalent for level, and upslope walking, slightly smaller in downslope walking and even smaller in up stair and down stair walking. There was no significant difference in the double support duration across conditions.

Lower limb joint power curves are presented as corridors representing the average +/- one standard deviation for the entire group in Fig. 1. When walking on slopes, the global pattern of the joint power curves over a gait cycle was not affected by the slope. For each joint, a tendency of increase can be observed for the positive peaks while climbing the slope and the negative ones while descending the slope. However, only the hip positive peak at the beginning of stance revealed significantly different ($p<0.01$) between level (0.56 +/-0.36 W/kg), gentle (0.25 +/- 0.21 W/kg) and steep slopes (0.12 +/- 0.1 W/kg) in descent and between level and steep slope in ascent (1.66 +/-0.48 W/kg). When walking in stairs, differences were observed in the patterns of individual joint powers as compared to walking in slopes. In particular, peaks of power absorption at the ankle in descent and generation at the knee in ascent occurred in the early stance. In the same time, the hip power generated in ascent (1.23 +/- 0.57 W/kg) was smaller than while going down the steep slope even if the difference was not statistically significant.

Fig. 2 compares the summed lower limb joints powers and the external mechanical power performed by each individual leg on the COM according to the walking conditions. The average summed lower limb joints powers curve is very similar to the external mechanical power performed by each individual leg on the COM (individual limb power) for all walking conditions. As for the individual joint power, the values of the positive peaks increased during walking upslope and the values of negative peaks increased during walking downslope. Thus, the peak of positive power was significantly increased in a steep slope ascent (4.96 +/- 1.44 W/kg) compared to level walking (3.33 +/- 1.3 W/kg). The peak of negative power was significantly greater during steep slope descent (−4.15 +/- 1.65 W/kg) compared to level walking (−1.45 +/- 0.93 W/kg). Again, the pattern in stair is radically different and exhibits quasi exclusive positive values in ascent and conversely negative values in descent. Positive peaks of power appeared in early stance while going up and negative peaks in late stance while going down.

The positive and negative works of each joints were computed along the whole gait cycle and summarized in Fig. 3 for each walking conditions.

In ascent, even if all joint positive works increased, the proportion of the total work due to the individual joint evolved with the walking condition. In particular, in slope, the part of the power produced by the ankle and the knee decreased (from 44% and 21% during level walking to 38% and 18% during steep slope ascent) while the contribution of the hip increased (34% during level walk-

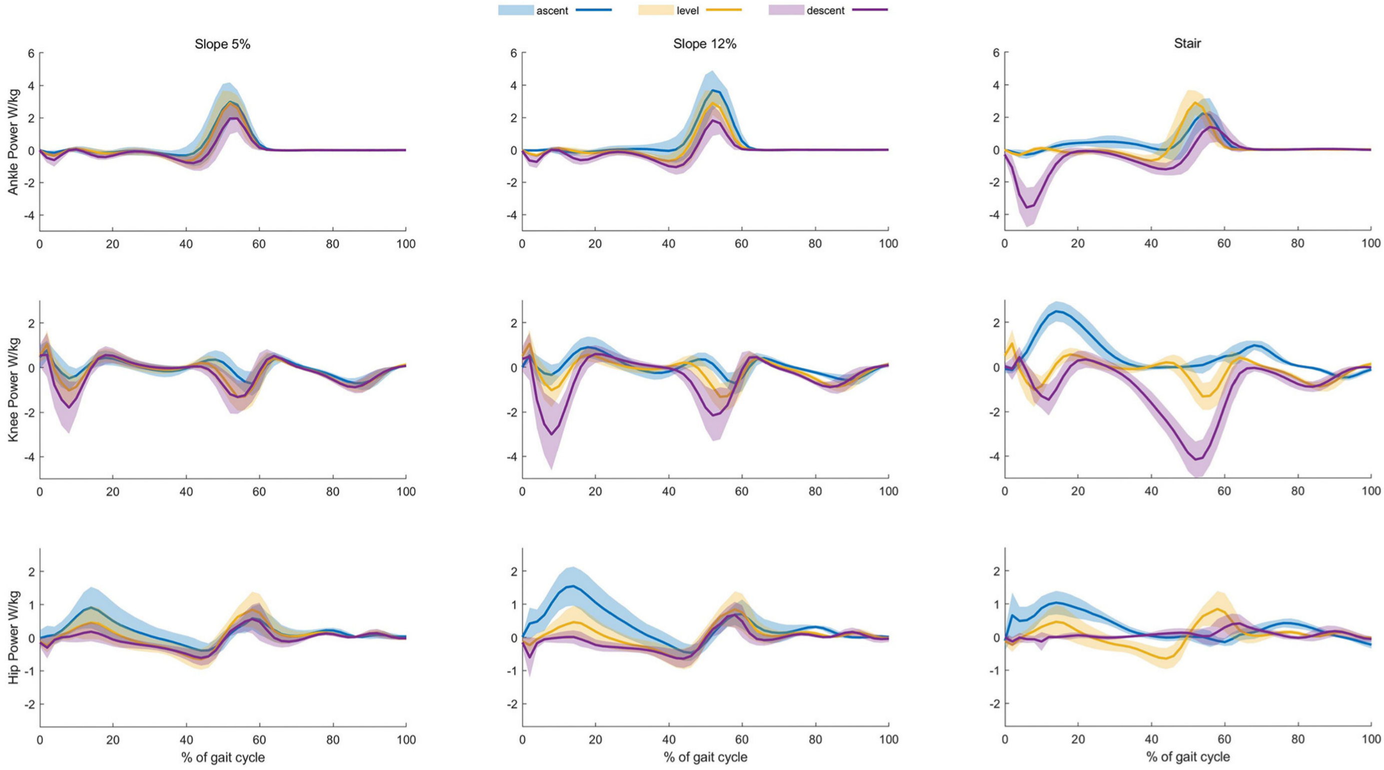

Fig. 1. Ankle, knee and hip joint power curves and corridors during ascent and descent of 5% slope, 12% slope and stair compared to level walking throughout gait cycle. In column: walking conditions (slope 5%, slope 12% and stair). In line: joint (ankle, knee and hip). On each graph are displayed the power curves for the condition in ascent and descent compared with level walking.

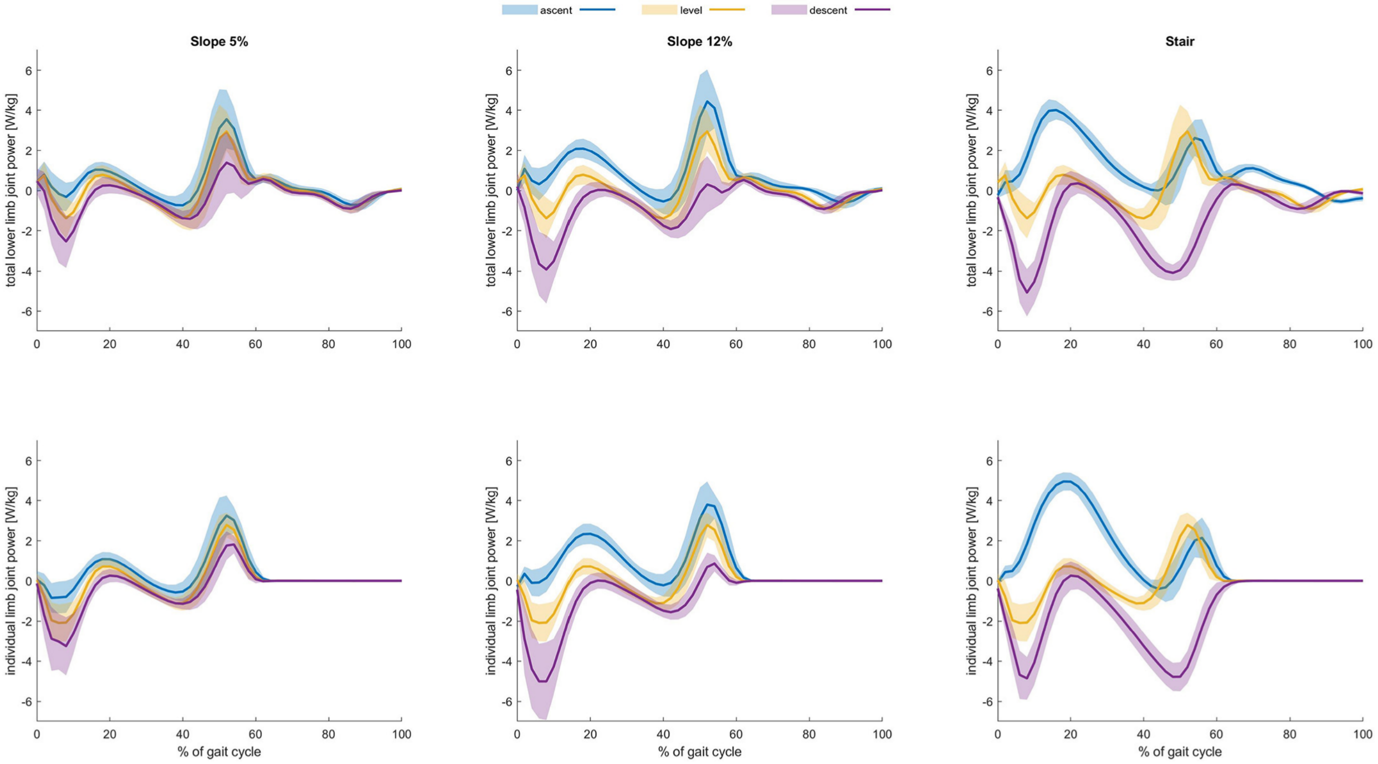

Fig. 2. Summed lower limb joints power (first line) and Individual limb power (second line) during ascent and descent of 5% slope, 12% slope and stair (first, second and third column respectively) compared to level walking throughout gait cycle.

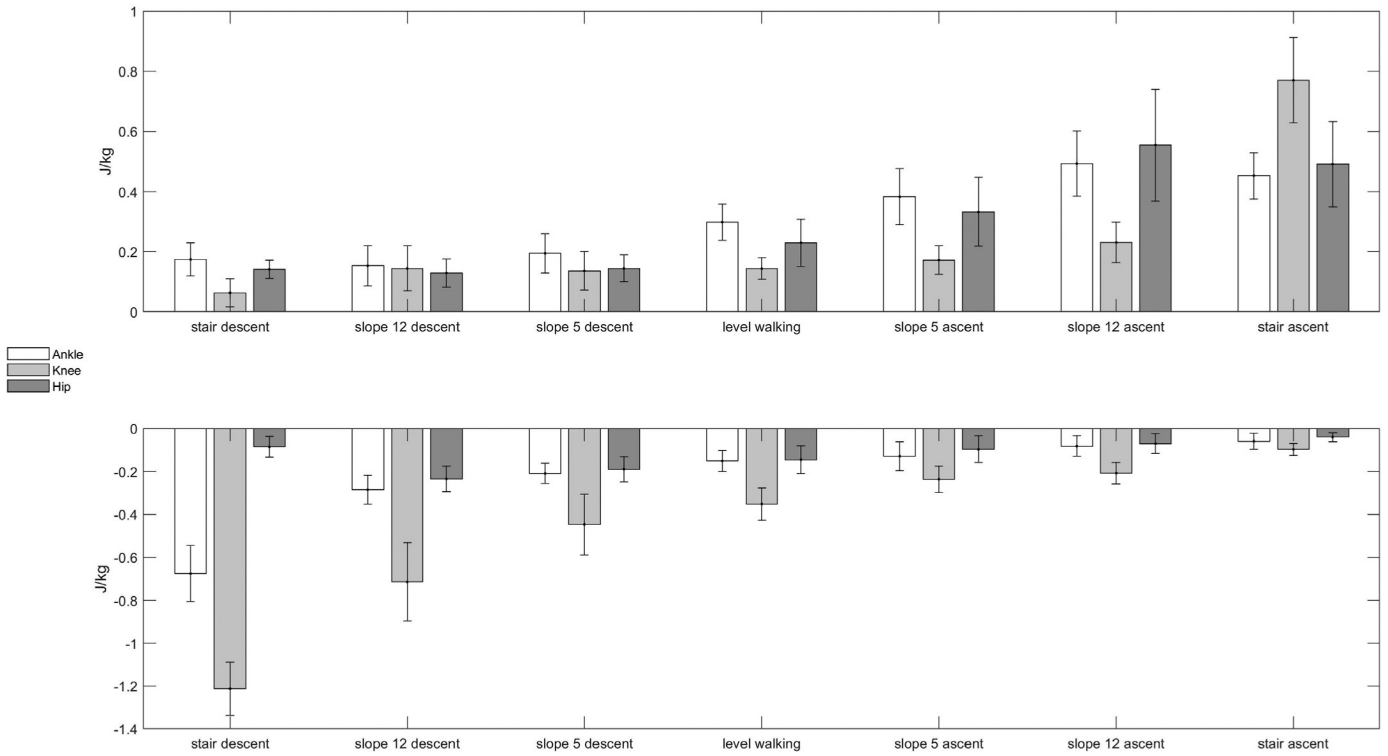

Fig. 3. Ankle (white), knee (light grey) and hip (dark grey) positive and negative work in all walking conditions (average and standard deviation over the entire group for the entire gait cycle).

ing to 43% during steep slope ascent) when the slope increased. In stair ascent, the distribution of work production was completely different and the contribution of the knee became predominant (45% of the work production compared to 26% and 28% for the ankle and the knee respectively). As concerns negative works, the total work logically decreased while the inclination increased but the proportion between joints remained slightly constant across conditions.

In descent, the negative work performed by the joint continuously increased with the slope except for the stair condition where a decrease of the hip negative work was observed. In terms of proportion, the distribution among joints remained relatively stable across downslope walking conditions (between 23% and 25% for the ankle, 53 and 58% for the knee and 19% and 22% for the hip) but this distribution was totally modified when walking downstairs with a very limited contribution of the hip (4%) while the absorption by the knee and the ankle represented 61% and 34% of the total respectively. Finally, the positive work performed by the joints decreased progressively with the inclination with no significant disruption due to the stair condition.

The positive and negative works computed along the stance phase were compared for the summed ankle–knee–hip power and the individual limb power and are presented in Fig. 4. The individual limb method estimated equal positive and negative works (0.34 +/- 0.07 J/kg) during level walking. On the contrary, using the summed ankle–knee–hip power, the positive work (0.43 +/- 0.09 J/kg) was overestimated and negative work (−0.39 +/- 0.07 J/kg) was underestimated compared to individual limb method evaluation. As expected, when inclination increased, the individual leg positive work increased, and the individual leg negative work decreased in ascent. Conversely, the individual leg negative work increased in ascent, and the individual leg positive work decreased in descent. The total work value is in accordance with the variation of potential energy of the body center of mass over a cycle (about 0.35 J/kg for 5% slope, 0.9 J/kg for 12% slope and 1.6 J/kg for stair). In particular, even if the equivalent inclination crossed when walking in the stair of the present study is about 5 times the one crossed over the steep slope, the step length decrease observed in stair (2 times smaller in stair vs slope) resulted in a moderate increase of the work. Finally, the total work computed over the stance phase of gait (provided by the ILM method that necessitates the force plates measurement) is underestimated due to the part of the power generated/absorbed during the swing phase (captured by the summed joint powers) particularly in stair locomotion.

The result of the statistical analysis revealed a significant influence of the walking condition on all the studied parameters. For the summed ankle-knee-hip work as for the external mechanical work computed by ILM method, pairwise comparisons confirmed significant differences except for the positive work produced in stair descent compared to steep slope descent ($p=0.97$ and $p=0.98$ respectively) and for negative work produced in stair ascent compared to steep slope ascent ($p=0.99$ and $p=0.14$ respectively). For joint works, the Table 2 summarized the difference that are not significant i.e. all others were significant at the level $p<0.01$.

4. Discussion

The present study aimed at comparing the change of energy rate of the center of mass (COM) and of the distribution of joint power contributing to this change in different walking conditions with a special focus on stair crossing. A secondary objective was to investigate the influence of the walking conditions of the mechanical works estimated from both methods. The normative reference data obtained in this study are useful to understand movement demands by determining the sources of the work production that can be modified by training or external factors [32].

Globally, the amount of work produced or absorbed is related to the increase or decrease of the equivalent inclination to be crossed. However, due to the physical architecture of the stair and the capabilities of human joints, the required work in stair is lower than in a slope of equivalent grade consistently with the decrease of the

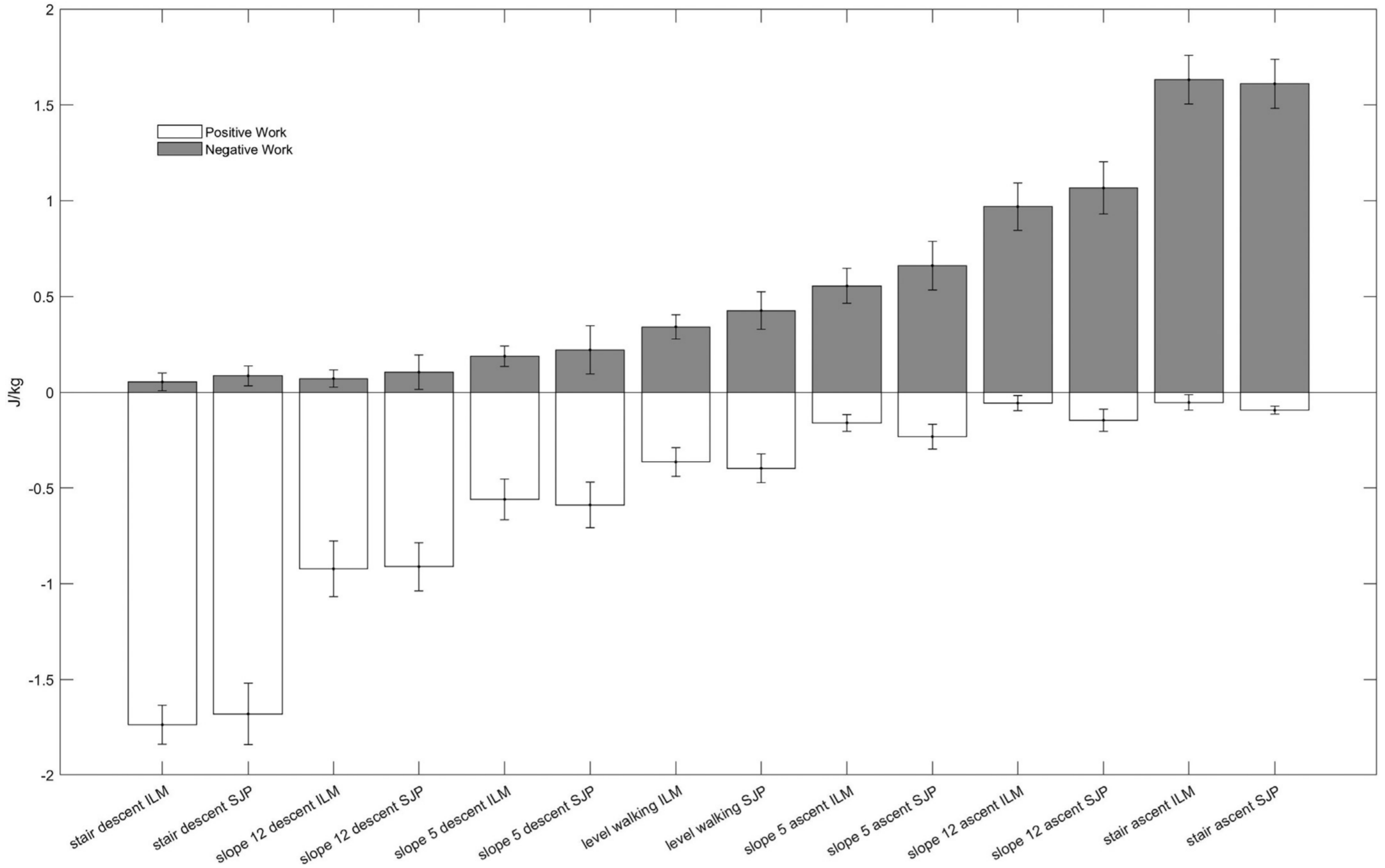

Fig. 4. Positive (white) and negative (dark grey) work obtained via the individual limb method (ILM) and using the summed lower limb joints power (SJP) in all walking conditions (average and standard deviations over the entire group for the stance phase only).

Table 2
Summary of the parameters that revealed no significant differences between stair ascent or descent and another condition identified from the post hoc pairwise comparisons between stair descent and slope descent (first line) and between stair ascent and slope ascent (second line). The table contained only the parameters for which p value was >0.01.

	12% slope descent	5% slope descent	Level
Stair descent	Ankle negative work Knee negative work Hip negative work	Ankle negative work Knee positive work Knee negative work Hip negative work	Ankle positive work Ankle negative work Knee positive work Knee negative work

	12% slope ascent	5% slope ascent	Level
Stair ascent	Knee positive work Knee negative work	Ankle negative work Knee positive work Knee negative work Hip negative work	Ankle positive work Ankle negative work Knee positive work Knee negative work Hip positive work Hip negative work

step length (2 times lower in stair than in steep slope). Indeed, if power patterns (joint and COM) when walking in slopes are similar to the ones when walking on level ground, they differ in stair ascent and descent. Particularly, energy generation in ascent and absorption in descent are almost equally distributed between the early and late stance phases. The power peaks are indeed exacerbated in early stance in ascent and in late stance in descent compared to slope walking even with a steep inclination (up to 5 W/kg).

The comparison between the external mechanical work at the COM and the summed joint works during stance shows consistent results for all walking conditions. The differences can be assumed to come from different sources related to the hypotheses surrounding each modeling [33], the contribution of soft tissues [17] or experimental uncertainties [18]. The study confirms that the use of ILM method is relevant to estimate the power produced by one single limb. It also exhibits the inherent limitations of the two methods, which justifies to consider them as complementary indicators of the mechanical work [18].

As concerns stair walking, the results corroborate previous findings available in the literature describing lower limb joint power [20], [34], [35]. The knee is the joint that contributes the most to both the production and absorption of work during stair ascent and descent respectively (45% in ascent and 61% in descent of the total work produced/absorbed respectively). The novelty of the study is to combine this analysis with the total work performed by the lower limb. The results demonstrate that walking in stair implies very different joint contributions than walking in slope. This is obviously due to the necessity to adapt to the geometric configuration of the stair and to the physiologic limits of the different joints. The overall work performed is however lower if related to the equivalent inclination of the stair. Thus, human joints seem to naturally adopt efficient strategy to this aim.

One limitation of the study is related to the walking velocities chosen by the volunteers that were different in stairs compared to the other walking conditions. These differences partially hindered the comparison of walking conditions, the external mechanical power being directly linked to this velocity. Other limitation lies in the experimental conditions as the stair is only three steps which does not allow to record steady-state stair walking but still remains a good compromise in the context of laboratory movement analysis. The size of the sample and the gender unbalance (11 males and 2 females) must also be considered as a limitation knowing that differences in muscle cocontractions has already been shown in the literature depending on gender [36], [37].

5. Conclusion

To conclude, the present study highlights the complementarity between external mechanical power of individual limbs and joint powers from inverse dynamics computation to fully understand the adaptations of the gait pattern in different walking conditions. It also provides normative data that could be used to compare pathologic population results to able-bodied persons in level, slope and stair walking.

Human and animal rights

The authors declare that the work described has been carried out in accordance with the Declaration of Helsinki of the World Medical Association revised in 2013 for experiments involving humans as well as in accordance with the EU Directive 2010/63/EU for animal experiments.

Informed consent and patient details

The authors declare that this report does not contain any personal information that could lead to the identification of the patient(s).

The authors declare that they obtained a written informed consent from the patients and/or volunteers included in the article. The authors also confirm that the personal details of the patients and/or volunteers have been removed.

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Author contributions

All authors attest that they meet the current International Committee of Medical Journal Editors (ICMJE) criteria for Authorship.

CRediT authorship contribution statement

Helene Pillet: Conceptualization, Funding acquisition, Methodology, Software, Writing – review & editing. **Boris Dauriac:** Data curation, Investigation, Writing – original draft. **Coralie Villa:** Investigation, Methodology. **Isabelle Loiret:** Funding acquisition, Supervision. **François Lavaste:** Funding acquisition, Supervision. **Xavier Bonnet:** Supervision, Validation, Writing – review & editing.

Declaration of competing interest

The authors declare that they have no known competing financial or personal relationships that could be viewed as influencing the work reported in this paper.

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References

[1] Griffin TM, Roberts TJ, Kram R. Metabolic cost of generating muscular force in human walking: insights from load-carrying and speed experiments. J Appl Physiol 2003;95(1):172–83. https://doi.org/10.1152/japplphysiol.00944.2002.

[2] Donelan J, Kram R, Kuo A. Simultaneous positive and negative external mechanical work in human walking. J Biomech 2002;35:117–24. https://doi.org/10.1016/S0021-9290(01)00169-5.

[3] Doets HC, Vergouw D, Veeger HEJ (Dirkjan), Houdijk H. Metabolic cost and mechanical work for the step-to-step transition in walking after successful total ankle arthroplasty. Hum Mov Sci 2009;28(6):786–97. https://doi.org/10.1016/j.humov.2009.05.001.

[4] a Willems P, a Cavagna G, Heglund NC. External, internal and total work in human locomotion. J Exp Biol 1995;198:379–93.

[5] DeVita P, Helseth J, Hortobagyi T. Muscles do more positive than negative work in human locomotion. J Exp Biol 2007;210(19):3361–73. https://doi.org/10.1242/jeb.003970.

[6] Montgomery JR, Grabowski AM. The contributions of ankle, knee and hip joint work to individual leg work change during uphill and downhill walking over a range of speeds. R Soc Open Sci 2018;5(8). https://doi.org/10.1098/rsos.180550.

[7] Zelik KE, Kuo AD. Mechanical work as an indirect measure of subjective costs influencing human movement. PLoS ONE 2012;7(2). https://doi.org/10.1371/journal.pone.0031143.

[8] Donelan JM, Kram R, Kuo AD. Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking. J Exp Biol 2002;205:3717–27. https://doi.org/10.1007/BF00430237.

[9] Houdijk H, Pollmann E, Groenewold M, Wiggerts H, Polomski W. The energy cost for the step-to-step transition in amputee walking. Gait Posture 2009;30(1):35–40. https://doi.org/10.1016/j.gaitpost.2009.02.009.

[10] Bonnet X, Villa C, Fodé P, Lavaste F, Pillet H. Mechanical work performed by individual limbs of transfemoral amputees during step-to-step transitions: effect of walking velocity. Proc Inst Mech Eng, H J Eng Med 2014;228(1):60–6. https://doi.org/10.1177/0954411913514036.

[11] Agrawal V, Gailey RS, Gaunaurd IA, O'Toole C, Finnieston AA. Comparison between microprocessor-controlled ankle/foot and conventional prosthetic feet during stair negotiation in people with unilateral transtibial amputation. J Rehabil Res Dev 2013;50(7):941–50. https://doi.org/10.1682/JRRD.2012.05.0093.

[12] Bonnet X, Villa C, Loiret I, Lavaste F, Pillet H. Distribution of joint work during walking on slopes among persons with transfemoral amputation. J Biomech 2021;129:110843. https://doi.org/10.1016/j.jbiomech.2021.110843.

[13] Browning RC, McGowan CP, Kram R. Obesity does not increase external mechanical work per kilogram body mass during walking. J Biomech 2009;42:2273–8. https://doi.org/10.1016/j.jbiomech.2009.06.046.

[14] Feng J, Pierce R, Do KP, Aiona M. Motion of the center of mass in children with spastic hemiplegia: balance, energy transfer, and work performed by the affected leg vs. the unaffected leg. Gait Posture 2014;39(1):570–6. https://doi.org/10.1016/j.gaitpost.2013.09.009.

[15] Mahon CE, Farris DJ, Sawicki GS, Lewek MD. Individual limb mechanical analysis of gait following stroke. J Biomech 2015;48(6):984–9. https://doi.org/10.1016/j.jbiomech.2015.02.006.

[16] Farris DJ, Hampton A, Lewek MD, Sawicki GS. Revisiting the mechanics and energetics of walking in individuals with chronic hemiparesis following stroke: from individual limbs to lower limb joints. J NeuroEng Rehabil 2015;12(1):24. https://doi.org/10.1186/s12984-015-0012-x.

[17] Zelik KE, Kuo AD. Human walking isn't all hard work: evidence of soft tissue contributions to energy dissipation and return. J Exp Biol 2010;213(24):4257–64. https://doi.org/10.1242/jeb.044297.

[18] Zelik KE, Takahashi KZ, Sawicki GS. Six degree-of-freedom analysis of hip, knee, ankle and foot provides updated understanding of biomechanical work during human walking. J Exp Biol 2015;218(6):876–86. https://doi.org/10.1242/jeb.115451.

[19] Nuckols RW, Takahashi KZ, Farris DJ, Mizrachi S, Riemer R, Sawicki GS. Mechanics of walking and running up and downhill: a joint-level perspective to guide design of lower-limb exoskeletons. PLoS ONE 2020;15(8):1–20. https://doi.org/10.1371/journal.pone.0231996.

[20] Riener R, Rabuffetti M, Frigo C. Stair ascent and descent at different inclinations. Gait Posture 2002;15:32–44. https://doi.org/10.1016/S0966-6362(01)00162-X.

[21] Pillet H, et al. APSIC: training and fitting amputees during situations of daily living. IRBM 2014;35(2):60–5. https://doi.org/10.1016/j.irbm.2014.02.005.

[22] Vrieling AH, et al. Uphill and downhill walking in unilateral lower limb amputees. Gait Posture 2008;28(2):235–42. https://doi.org/10.1016/j.gaitpost.2007.12.006.

[23] Vickers DR, Palk C, McIntosh AS, Beatty KT. Elderly unilateral transtibial amputee gait on an inclined walkway: a biomechanical analysis. Gait Posture 2008;27(3):518–29. https://doi.org/10.1016/j.gaitpost.2007.06.008.

[24] Fradet L, Alimusaj M, Braatz F, Wolf SI. Biomechanical analysis of ramp ambulation of transtibial amputees with an adaptive ankle foot system. Gait Posture 2010;32(2):191–8. https://doi.org/10.1016/j.gaitpost.2010.04.011.

[25] Wolf EJ, Everding VQ, Linberg AL, Schnall BL, Czerniecki JM, Gambel JM. Assessment of transfemoral amputees using C-Leg and Power Knee for ascending and descending inclines and steps. J Rehabil Res Dev 2012;49(6):831. https://doi.org/10.1682/JRRD.2010.12.0234.

[26] Riener R, Rabuffetti M, Frigo C. Stair ascent and descent at different inclinations. Gait Posture 2002;15(1):32–44. https://doi.org/10.1016/S0966-6362(01)00162-X.

[27] Bae TS, Choi K, Hong D, Mun M. Dynamic analysis of above-knee amputee gait. Clin Biomech 2007;22(5):557–66. https://doi.org/10.1016/j.clinbiomech.2006.12.009.

[28] Della Croce U, Bonato P. A novel design for an instrumented stairway. J Biomech 2007;40:702–4. https://doi.org/10.1016/j.jbiomech.2006.01.020.

[29] Pillet H, Bonnet X, Lavaste F, Skalli W. Evaluation of force plate-less estimation of the trajectory of the centre of pressure during gait. Comparison of two anthropometric models. Gait Posture 2010;31(2):147–52. https://doi.org/10.1016/j.gaitpost.2009.09.014.

[30] Dempster WT. Space requirements of the seated operator, geometrical, kinematic and mechanical aspects of the body with special reference to the limbs. Fort Belvoir, VA: Defense Technical Information Center; 1955.

[31] Amabile C, Choisne J, Nérot A, Pillet H, Skalli W. Determination of a new uniform thorax density representative of the living population from 3D external body shape modeling. J Biomech 2016;49(7):1162–9. https://doi.org/10.1016/j.jbiomech.2016.03.006.

[32] Alexander N, Strutzenberger G, Ameshofer LM, Schwameder H. Lower limb joint work and joint work contribution during downhill and uphill walking at different inclinations. J Biomech 2017;61:75–80. https://doi.org/10.1016/j.jbiomech.2017.07.001.

[33] Riddick RC, Kuo AD. Mechanical work accounts for most of the energetic cost in human running. Sci Rep 2022;12(1):1–11. https://doi.org/10.1038/s41598-021-04215-6.

[34] McFadyen BJ, Winter DA. An integrated biomechanical analysis of normal stair ascent and descent. J Biomech 1988;21(9):733–44. https://doi.org/10.1016/0021-9290(88)90282-5.

[35] Novak AC, Li Q, Yang S, Brouwer B. Mechanical energy transfers across lower limb segments during stair ascent and descent in young and healthy older adults. Gait Posture 2011;34(3):384–90. https://doi.org/10.1016/j.gaitpost.2011.06.007.

[36] Mengarelli A, Maranesi E, Burattini L, Fioretti S, Di Nardo F. Co-contraction activity of ankle muscles during walking: a gender comparison. Biomed Signal Process Control 2017;33:1–9. https://doi.org/10.1016/j.bspc.2016.11.010.

[37] Strazza A, et al. Increased co-contraction activity during push-off phase of walking in healthy women. IRBM 2021;42(1):48–54. https://doi.org/10.1016/j.irbm.2020.01.001.