Distal-to-proximal joint kinetics redistribution is a main contributor to reduced walking economy in older adults

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Complete List of Authors:	Delabastita, Tijs; KU Leuven, Human Movement Biomechanics Research Group, Department of Movement Sciences Hollville, Enzo; KU Leuven, Human Movement Biomechanics Research Group, Department of Movement Sciences Catteau, Andreas; KU Leuven, Human Movement Biomechanics Research Group, Department of Movement Sciences Cortvriendt, Philip; KU Leuven, Human Movement Biomechanics Research Group, Department of Movement Sciences De Groote, Friedl; KU Leuven, Human Movement Biomechanics Research Group, Department of Movement Sciences Vanwanseele, Benedicte; KU Leuven, Human Movement Biomechanics Research Group, Department of Movement Sciences
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23 8 Tijs Delabastita, Enzo Hollville, Andreas Catteau, Philip Cortvriendt, Friedl De Groote,
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25 9 Benedicte Vanwanseele
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28 10 Affiliation
29
30
31 11 Human Movement Biomechanics Research Group, Department of Movement Science, KU
32
33 12 Leuven, Belgium
34
35
36 13 Corresponding author
37
38
39 14 Benedicte Vanwanseele
40
41
42 15 benedicte.vanwanseele@kuleuven.be
43
44
45 16 Tervuursevest 101 – bus 1501, 3001 Leuven, Belgium
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21 Abstract

22 Background

23 Age-related neural and musculoskeletal declines affect mobility and the quality of life of old
24 people. To date, the mechanisms underlying the reduced walking economy in older adults still
25 remain elusive. In this study, we wanted to investigate which biomechanical factors were
26 associated with the higher energy cost of walking in older compared to young adults.

27 Methods

28 Fourteen young (24 ± 2 years) and older (74 ± 4 years) adults were tested. Plantarflexor
29 strength and Achilles tendon stiffness were evaluated during a dynamometer test. Medial
30 gastrocnemius fascicle length, ground reaction forces, joint kinematics and oxygen
31 consumption were measured during walking treadmill at $0.83 \text{ m}\cdot\text{s}^{-1}$ and $1.39 \text{ m}\cdot\text{s}^{-1}$. Energy cost
32 of walking, lower-limb joint kinetics, muscle-tendon unit and tendinous tissues length were
33 calculated.

34 Results

35 Energy cost of walking was higher at $0.83 \text{ m}\cdot\text{s}^{-1}$ (+16%; $p = 0.005$) and plantarflexor strength
36 lower (-31%; $p = 0.007$) in older adults. Achilles tendon stiffness ($p = 0.66$) and medial
37 gastrocnemius fascicle length changes did not differ between young and older adults. Ankle
38 kinetics reduction was compensated by hip kinetics increase in older adults during walking. The
39 hip extensor moment was the only significant predictor of the energy cost of walking.

40 Conclusion

41 The higher energy cost of walking in older adults is mainly associated with the distal-to-
42 proximal redistribution of joint kinetics in older adults during walking possibly due to

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plantarflexor weakness. In our study, medial gastrocnemius fascicle and tendinous tissue behavior did not explain the higher energy cost of walking between young and older adults.

Keywords

ultrasound, gait, energy cost, muscle-tendon, elderly, joint power

PROOF

INTRODUCTION

Slow preferred walking speed is a predictor of mobility limitations and hospitalization ¹ and is related to the high energy cost of walking in older adults. Specifically, a rise in energetic cost can predict the rate of preferred walking speed decline after the age of 65 years ². It has therefore been suggested that older adults compensated for their poor walking economy by adopting a preferred slow walking speed ². However, the biomechanical factors associating with such poor walking economy remain equivocal. A better understanding of these factors may help designing tailored training interventions to improve walking economy and performance in older adults.

Reduced walking economy and impaired mobility in older adults result from a broad range of age-related neural and musculoskeletal changes ³⁻⁵. As a result, older adults have marked deficits in push-off intensity or propulsive power during walking ⁶. In addition, ankle plantarflexor moments during push-off are lower and hip extensor moments during mid-stance are higher during walking in older than in young adults ^{7,8}. This compensatory mechanism is well-known as distal-to-proximal redistribution of positive joint work and power within the lower limb to generate propulsive power during walking ^{9,10}. Previously, using a double inverted pendulum model, it was found that during walking, lower push-off forces at the ground contact point of the trailing leg and higher compensatory hip extensor moments at the leading leg increased the energy dissipation when redirecting the center of mass velocity ¹¹. This might partly explain why the proximal shift in joint moments would be related to higher whole-body energy cost of walking in older adults ¹². Although lower plantarflexor moments functionally impair older adults' push-off intensity ¹³, it yet remains unclear if they are also related to the energy cost of walking.

The age-related deficits in propulsive power during walking is associated with hallmark age-related reductions in ankle plantarflexor muscle power and work (for review, see ⁶). In addition, Achilles tendon behavior and properties are altered with ageing ^{14,15} and may be another

72 explanation for the higher energy cost of walking observed in older compared to young adults.

73 Indeed during walking, the Achilles tendon acts as a spring enabling the triceps surae to

74 efficiently produce force near optimal length (i.e., quasi-isometrically) and at very low

75 contraction velocity ¹⁶. Any impairment in Achilles tendon stiffness may in turn affect power

76 generation by altering the operating lengths and contraction velocities of the triceps surae. Older

77 adults generally exhibit lower Achilles tendon stiffness (for review, see ¹⁷) which may increase

78 Achilles tendon elongation during walking compared to young adults ¹⁸. This might force the

79 triceps surae to operate at shorter length and high shortening velocity during the stance phase

80 of walking ^{18–20}, thus increasing the generating cost of doing work and likely the whole-body

81 metabolic cost of walking ²¹. Very recently, Conway and Franz (2020) observed that medial

82 gastrocnemius fascicles were shorter during walking at preferred and fast walking speed in

83 older adults, which was associated with worse capacity to enhance push-off intensity ¹⁹. These

84 authors also observed slower fascicle shortening contraction velocity in older adults whatever

85 the walking speed. Besides, Stenroth et al. (2017) have suggested that older adults adopted a

86 slow preferred walking speed which improves their triceps surae force generation ability,

87 mainly because of plantarflexor weakness and the favorable force-velocity potentials ²⁰.

88 Although simulation studies already showed that Achilles tendon stiffness influences the energy

89 cost of walking ²¹, direct experimental evidence is lacking regarding the relationship between

90 Achilles tendon stiffness, triceps surae behavior and the energy cost of walking in older adults.

91 The main aim of this study was to investigate which biomechanical factors were associated to

92 the higher energy cost of walking in older compared to young adults at slow and fast walking

93 speed. We first evaluated how young and older adults differ in terms of Achilles tendon

94 stiffness, maximal plantarflexor strength, energy cost of walking, walking capacity, medial

95 gastrocnemius fascicle, muscle-tendon unit and tendinous tissue behavior, and joint kinematics

96 and kinetics. We then used the same variables to predict the energy cost of walking in young

and older adults. We hypothesized that energy cost of walking, hip extensor and flexor moments and medial gastrocnemius fascicle shortening velocity during walking would be higher in older than in young adults. Moreover, we expected to find lower Achilles tendon stiffness, medial gastrocnemius operating length, maximal plantarflexor strength, ankle plantarflexor moments during walking, and lower maximal walking capacity in older than in young adults. We also expected to find an association between the higher energy cost of walking seen in older adults and lower Achilles tendon stiffness, unfavorable medial gastrocnemius operating length and velocity, lower maximal plantarflexor strength and larger hip kinetics during walking.

METHODS

Participants

We determined the sample size of this study using a priori statistical power estimations. The overall effect size of the difference in Achilles tendon stiffness between young and older adults, reported in a systematic review was 1.2¹⁷. The effect size of the difference in maximal ankle plantarflexor moment during walking and the energy cost of walking between young and older adults reported in previous studies was 1.2 (young $136 \pm 27 \text{ N.m}^{-1}$, older $102 \pm 29 \text{ N.m}^{-1}$)⁹ and 1.3 (young $0.179 \pm 0.02 \text{ ml O}_2.\text{kg.m}^{-1}$, older $0.205 \pm 0.02 \text{ ml O}_2.\text{kg.m}^{-1}$)²², respectively. With α -error probability of 0.05 and statistical power of 0.8, a sample size of 13 young and older subjects would be sufficient to detect age-related differences in Achilles tendon stiffness, maximal ankle plantarflexor moment during walking and in the energy cost of walking using two-tailed Mann-Whitney U tests.

When piloting the study, we noticed that the ultrasound image quality was worse in older than in young adults. To account for exclusion of older subjects due to poor image quality, we recruited 16 young adults (age: 23 ± 2 years; body mass: $63 \pm 12 \text{ kg}$; length: $165 \pm 7 \text{ cm}$; 8 males, 8 females) and 18 older adults (age: 75 ± 4 years; body mass: $73 \pm 12 \text{ kg}$; length: $170 \pm$

9 cm; 9 males, 9 females). We included healthy young (between 20 and 30 years) or older (between 68 and 85 years old) subjects that did not participate in competitive sports and did not perform structured exercise more than twice a week over the last three months. This study was approved by the Medical Ethics Committee UZ/KU Leuven. The subjects signed written informed consent before participating in the study in accordance with the declaration of Helsinki.

Experimental design

We tested the participants during a single session comprising three tests: a) an overground walking test, where we quantified maximal walking capacity, b) a treadmill walking test, where we collected data to quantify medial gastrocnemius fascicle length and contraction velocity, gait kinematics and kinetics and the energy cost of walking, and c) a dynamometer test, where we collected data to quantify Achilles tendon stiffness and maximal plantarflexor strength.

Overground walking

To quantify maximal walking capacity, we measured the distance covered during a six-minute walk test, i.e. the six-minute walk distance, with participants being barefoot ²³.

Treadmill walking

Actual testing only started after an extensive familiarization of at least 20 minutes to the instrumented treadmill (M-Gait, Motekforce Link, Amsterdam, The Netherlands) ²⁴. During testing, the subjects walked barefoot at 0.83 m.s⁻¹ (3 km.h⁻¹) and at 1.39 m.s⁻¹ (5 km.h⁻¹) for 5 to 8 minutes. Between the familiarization, and between the testing trials, the subjects could rest until they sufficiently recovered. The participants walked barefoot to enable precise foot marker placement and to avoid inter-subject variability due to shoe and insole type.

Energy cost of walking

To quantify the energy cost of walking, we continuously measured oxygen consumption and carbon dioxide production (Oxycon Mobile Pro, Carefusion, Houten, The Netherlands). We only proceeded to the next trial after at least 5 minutes and if the subjects attained a steady state oxygen uptake and carbon dioxide production for at least 2 minutes²². We asked the subjects to maintain their normal dietary pattern and to avoid intense physical activities during 24 hours before the start of the measurement. The subjects did not eat or drink for 2 hours before starting the static standing gas analysis that was performed before treadmill walking.

We averaged raw oxygen uptake during the steady state phase in the last two minutes of walking at 0.83 m.s⁻¹ and 1.39 m.s⁻¹ (Oxycon Mobile Pro software, Carefusion, Houten, The Netherlands). We subtracted the oxygen uptake during static standing from the oxygen uptake during walking to obtain net oxygen uptake. We normalized this value to body weight and to distance covered to obtain the normalized net oxygen uptake during walking²². Normalized net oxygen uptake is a good predictor for energy cost of walking in young and older adults²².

Lower-limb joint kinematics and kinetics

Motion capture data and ground reaction forces were collected during three full strides of walking after 5 to 8 minutes at 0.83 m.s⁻¹ and at 1.39 m.s⁻¹. We captured the trajectories of 68 reflective markers using thirteen infrared cameras (Vicon, Oxford, UK) recording at 100 Hz. We used an extended full body plug in gait marker set²⁵. We measured ground reaction forces using the two built-in force plates in the treadmill at 1000 Hz.

We scaled an OpenSim's gait 2392-model to the subject's anthropometry using the marker data collected during a static standing trial²⁶. Afterwards, we calculated joint kinematics with a Kalman smoothing algorithm²⁷ using the marker data collected during walking. We used the kinematics and ground reaction forces in OpenSim's inverse dynamics tool to calculate joint moments. Next, joint power and joint work were computed. Ankle and hip joint power were

calculated as the product of the joint moment and the joint angular velocity (derivative of joint angular displacement). Joint work was calculated by the integration of joint power using the trapezium method. Joint kinetics parameters (i.e., moments, power, work) were normalized to the subject's body weight. Afterwards, we identified the maximal ankle plantarflexor moment, maximal hip extensor moment and maximal hip flexor moment per stride before averaging over 3 strides per participant.

Medial gastrocnemius fascicle length and velocity

To quantify medial gastrocnemius fascicle length and contraction velocity, we collected two-dimensional ultrasound images of the medial gastrocnemius muscle fascicles of the right leg. We used a PC-based ultrasound system (Echoblaster 128, UAB Telemed, Vilnius, Lithuania) to record ultrasound images at 60 Hz using a flat shaped ultrasound probe (wave frequency 8 MHz, field of view 60 mm). We carefully placed the probe in line with the muscle fascicles and perpendicular to the deep aponeurosis of the medial gastrocnemius²⁸. The probe placement was similar to the placement used in a previous study²⁹. We synchronized motion capture and ultrasound data with an electrical pulse at the start and at the end of ultrasound imaging.

The ultrasound images collected during treadmill walking were processed by tracking three lines representing the superficial aponeurosis, the deep aponeurosis and the orientation of the medial gastrocnemius muscle fascicles with a semi-automated algorithm³⁰. We linearly extrapolated these lines and calculated medial gastrocnemius fascicle length as the distance between the two aponeuroses parallel to the muscle fascicles and the pennation angle as the angle between the muscle fascicles and the deep aponeurosis¹⁶. The first derivative of the medial gastrocnemius fascicle length with respect to time was calculated to determine medial gastrocnemius fascicle contraction velocity. We normalized medial gastrocnemius fascicle length to the average fascicle length measured during the static standing trial. Afterwards, we calculated medial gastrocnemius fascicle length operating range throughout the whole stance

193 phase and throughout terminal stance and pre-swing (50 to 85 % and 85 to 100 % of the stance
194 phase, respectively). We defined operating range as the difference between minimal and
195 maximal normalized fascicle length throughout the time range of interest²⁰. We also calculated
196 medial gastrocnemius fascicle contraction velocity at maximal ankle plantarflexor moment and
197 maximal fascicle contraction velocity throughout terminal stance and pre-swing²⁰.

198 Additionally, we calculated medial gastrocnemius tendinous tissue length by subtracting medial
199 gastrocnemius fascicle length multiplied by the cosine of its pennation angle from the medial
200 gastrocnemius muscle-tendon length¹⁶ calculated using OpenSim's muscle analysis tool. We
201 also normalized medial gastrocnemius tendinous tissue length to its length measured during the
202 static standing trial.

203 **Dynamometer testing session**

204 The participants lay prone on the dynamometer seat with the left foot on the footplate (Biodex
205 System 3, Biodex Medical Systems, Inc., New York, USA). The ankle was positioned in 90°
206 flexion (neutral position) with the knee and hip fully extended. The foot was tightly strapped to
207 the footplate. Before each trial, we carefully aligned the mid-point of the line between the
208 medial and the lateral ankle with the dynamometer axis. Participants performed four explosive
209 isometric plantarflexions. Peak ankle torque was averaged across the four trials and used as a
210 measure of plantarflexor strength (34).

211 We captured the trajectories of 10 reflective markers using three mobile infrared cameras
212 (Vicon, Oxford, UK) recording at 100 Hz. The marker set and marker placement were similar
213 as during treadmill walking, but we only used the markers on the lateral side of the left leg²⁵.

214 We also collected ultrasound images of the medial gastrocnemius muscle-tendon junction
215 during explosive contractions and passive rotations. A PC-based ultrasound system captured
216 images at 30 Hz using the same probe, wave frequency and field of view as during treadmill

walking. To visualize the medial gastrocnemius muscle-tendon junction, we placed the probe at the distal end of the medial gastrocnemius muscle belly ⁵. We slightly moved the probe to obtain images perpendicular to the aponeuroses. We synchronized motion capture and ultrasound data with an electrical pulse at the start and at the end of ultrasound imaging.

We quantified Achilles tendon force by dividing the ankle plantarflexor moment by the Achilles tendon moment arm calculated using a hybrid method combining three-dimensional motion analysis and ultrasound ³¹. Briefly, ultrasound was used to measure the perpendicular distance from the surface of the skin to the midline of the Achilles tendon, while motion capture was used to determine the perpendicular distance from the bottom of the probe to the ankle joint center. The difference between these two measures was defined as the Achilles tendon moment arm (31). We calculated Achilles tendon elongation during the explosive contractions by tracking the muscle-tendon junction displacement ³⁰. We accounted for the effect of ankle joint rotation on Achilles tendon elongation during isometric plantarflexions by subtracting the tendon elongation measured during a passive rotation of the foot ³². During these passive rotations, we set up the system to move the foot from end-range dorsiflexion to end-range plantarflexion within the comfort limits of the patient at 15°/s. We tracked the muscle-tendon junction displacement ³⁰ and calculated ankle joint rotation from the marker trajectories with a Kalman smoothing algorithm ²⁷. We defined Achilles tendon stiffness as the slope of the linear fit to the force-elongation data between 20 and 80 % of the maximal plantarflexor strength for all 4 explosive contractions. We used the median Achilles tendon stiffness across four trials ³³. For five subjects, we could only use 3 trials due to technical issues.

238 Statistical analysis

We used Mann-Whitney U tests to compare young and older adults with respect to the discrete parameters Achilles tendon stiffness, maximal plantarflexor strength, energy cost and six-minute walk distance. We used non-parametric statistical tests throughout the study to account

for non-normally distributed outcomes. We applied Sidak-corrections for multiple comparisons (critical alpha-level at 0.0253 for the comparisons of oxygen uptake and at 0.05 for other comparisons).

To compare young and older adults with respect to the time-varying parameters ankle and hip joint kinetics, medial gastrocnemius fascicle length and contraction velocity, tendinous tissue length and muscle-tendon unit length, we used repeated measures ANOVA's with statistical non-parametrical mapping (SNPM) ³⁴. We applied Sidak-corrections for multiple comparisons (critical alpha-level at 0.0253 for the comparisons of ankle and hip joint moments and at 0.0127 for other comparisons). We performed these analyses in Matlab 2015b (The Mathworks, Inc., Natick, USA).

Partial correlations using non-parametric Spearman's rank correlations adjusted for step frequency were performed to investigate the relationships between the energy cost of walking and, the Achilles tendon stiffness, maximal plantarflexor strength, joint kinetics, medial gastrocnemius fascicle length, and six-minute walk distance. Data for young and older adults were pooled as we did find similar correlation coefficients between both age groups. We applied Sidak-corrections for multiple correlations. The critical alpha level was set to 0.05 for correlations between Achilles tendon stiffness, maximal isometric plantarflexor force, and six-minute walk distance. The critical alpha level was set to 0.0253 for all other correlations.

Based on non-parametric partial correlations, we additionally performed a multiple regression analysis using the three significant independent variables (i.e., hip extensor moment, Achilles tendon stiffness and plantarflexor strength) as inputs to predict the energy cost of walking. Two independent models were tested corresponding to the two controlled walking speeds used in this study. We used bootstrapping to account for non-normally distributed variables. We checked the assumptions of multicollinearity using tolerance and variance inflation factor and

reported no violation of the assumptions. All statistical analyses were performed in SPSS Statistics 20 (IBM Corp., Armonk, USA).

Results

We excluded four older participants due to problems to perform isometric contractions without compensations (two participants) and insufficient ultrasound image quality (two participants). As gender influences Achilles tendon stiffness ⁵, we excluded the last two young male participants included in the study to have the same distribution of males and females in both groups. The results of 14 healthy young adults (age: 24 ± 2 years; height 170 ± 10 cm; body mass 63 ± 13 kg; 7 male and 7 female) and 14 healthy older adults (age 74 ± 4 years, height 165 ± 6 cm, weight 73 ± 12 kg, 7 male and 7 female) are reported.

Young vs. older adults

Older adults exhibited a higher net energy cost of walking than young adults when walking at 0.83 m.s⁻¹ (on average +16%) while we observed a trend for higher net energy cost of walking at 1.39 m.s⁻¹ (on average +12%; p = 0.044; alpha level ≤ 0.0253; Table 1). Moreover, six-minute walk distance was lower in older than in young adults (on average -17%; Table 1). Maximal plantarflexor strength was also lower in older than in young adults (on average -31%). However, we observed similar Achilles tendon stiffness between young and older adults (p = 0.663).

No statistical differences were observed in medial gastrocnemius muscle-tendon unit, fascicle and tendinous tissue length changes and velocity between young and older adults during the stance phase of walking at both speeds (all p-values > 0.05; Fig. 1). We also observed similar fascicle operating length at both slow and fast walking speeds in both populations (Fig. 1C).

There was a distal-to-proximal joint kinetics redistribution during walking at both slow and fast speeds in older adults (Fig. 2-4). Ankle plantarflexor moment during terminal stance propulsion was lower in older than in young adults (-9 and -14% at slow and fast speed, respectively; Fig.

2A) while hip extensor moment during loading response and mid-stance was higher in older than in young adults (+32% and +30% at slow and fast speed, respectively; Fig. 2B). The age-related differences in hip moments were more pronounced at higher walking speed. When increasing walking speed, hip extensor moment during loading response and mid-stance increased more in older than in young adults (Fig. 2B). Hip flexor moment during pre-swing was also higher in older than in young adults (Fig. 2B). Similarly, we observed larger hip extension power at mid-stance (+25% and +32% at slow and fast speed, respectively) and hip flexion power at terminal stance (+31% and +24% at slow and fast speed, respectively) in older adults (Fig. 3B). This compensates for reduced ankle power during propulsion at slow (-20%) and fast walking speeds (-30%; Fig. 3A). We also found higher positive hip work and lower positive ankle work in older compared to young adults (Fig. 4).

Partial correlation analysis

Partial non-parametric correlations between the energy cost of walking at 0.83 m.s⁻¹ and 1.39 m.s⁻¹ and the main variables, adjusted for step frequency, are reported in Table 2. Higher energy cost of walking was significantly associated with higher hip extensor moment (r: .38 to .48), lower Achilles tendon stiffness (r: -.57 to -.52), lower plantarflexor strength (r: -.47 to -.43), higher hip positive work (r: .45 to .51), and higher hip extension power at mid-stance (r: .45 to .56).

Multiple regression analysis

The multiple regression models predicted up to 38% of the variance in the energy cost of walking (Table 3). The independent variables (i.e., hip extensor moment, Achilles tendon stiffness and ankle plantarflexor strength) statistically predicted the energy cost of walking at both 0.83 m.s⁻¹ ($R^2 = 0.45$; adjusted $R^2 = 0.38$; $p = 0.002$) and 1.39 m.s⁻¹ ($R^2 = 0.42$; adjusted $R^2 = 0.35$; $p = 0.004$). However, only hip extensor moment was independent predictor of the

energy cost of walking in these models (p-values: 0.012 and 0.040; standardized coefficients β : 0.562 and 0.381, respectively at 0.83 m.s⁻¹ and 1.39 m.s⁻¹).

Discussion

In this study, we experimentally compared the walking economy of young and older adults and identified the biomechanical factors that were associated with the energy cost of walking at slow and fast speeds. We observed that older adults had a higher energy cost of walking compared to young adults at matched speed. This was accompanied by lower ankle kinetics (i.e., moment, power and work) during the propulsion phase of walking, mainly compensated by larger hip kinetics in early/mid-stance. In addition, we found that the hip extensor moment, Achilles tendon stiffness and plantarflexor strength were associated with the energy cost of walking. However, only the hip extensor moment was an independent predictor of the energy cost of walking. The current study highlights the major influence of the age-related distal-to-proximal redistribution of joint kinetics on the energy cost of walking, with increasing demand placed at the hip joint increasing the energy cost of walking in older adults.

The effect of ageing on walking economy has been studied extensively with consistent findings of higher gross and net cost of walking in healthy older adults compared to their younger counterparts (e.g., for review see ³⁵). In line with previous observations, we also reported a 16% to 12% higher net energy cost of walking in older than in young adults at both slow (0.83 m.s⁻¹) and fast speeds (1.39m.s⁻¹) ^{22,35,36}. This increase in the energy cost of walking in older adults is multifactorial and remains unclear. Previous hypotheses about the role of gait instability ²² and individual leg work ³⁷ have failed or partly explain (i.e., antagonist activity ³⁸) this age-related higher energy cost of walking. Based on our results, we posit that the higher mechanical output from muscles spanning the hip, i.e. extensors during early/mid-stance and flexors during

late stance/early swing, is one of the main contributors to the higher energy cost of walking observed in older adults.

Which driven mechanisms?

Prior studies highlighted the key role of neural control/strategies in the age-related increase in the energy cost of walking ^{36,38}. Specifically, increased antagonist leg muscle coactivation explained around one third of the oxygen consumption increase in older adults ^{36,38}. We now discuss about other potential driven mechanisms explaining the age-related decreased walking economy in older adults.

Age-related distal-to-proximal redistribution of joint kinetics

The higher mechanical output from the hip is mainly caused by the well-known age-related distal-to-proximal redistribution of joint kinetics during walking ^{9,10} with a shift strategy in mechanical output from the ankle to the hip in older adults. In addition, it has previously been shown in young adults that an experimental reduction of ankle plantarflexor power during walking was accompanied by a higher reliance to the proximal leg muscles and an increase in the energy cost of walking ³⁹. In our study, we observed that changes in ankle kinetics were inversely related to hip muscle kinetics with an age-related reduction of 20 to 30% in ankle power and increase of 25 to 32% in hip power. We also observed similar magnitude of positive ankle work reduction (-26 to -27%) and hip positive work increase (29 to 30%) in older adults during walking at both speeds. However, the age-associated reduction in peak ankle moment during push-off was relatively smaller in comparison to the higher hip extensor moment. Deficits in ankle moment emerged as the lone genuine functionally limiting impairment in older adults ¹³. Recently, Conway and Franz demonstrated that ankle moment was the main limiting factor of push-off intensity and thereby walking performance (i.e., walking speed). In our study, we observed lower peak ankle moment in older adults during walking at both slow and fast

speeds but this was not correlated to the energy cost of walking ($r: .067$, $p = 0.74$ at 0.83 m.s^{-1} ; $r: .266$, $p = 0.18$ at 1.39 m.s^{-1}). The overall decrease in ankle kinetics has consequences on propulsive power generation in older adults ⁶ and partly arise from lower ankle plantarflexor strength (-31% in our study), a decrease in plantarflexors force-generating capacity and/or their inability to elicit the use of propulsive reserve capacity during walking ⁶. Besides, Achilles subtendons dynamics ⁴⁰ and Achilles tendon moment arm ⁴¹ could also have affected peak ankle joint moment, work and power during walking in older adults and may indirectly play a role in the redistribution of joint kinetics towards proximal hip muscles.

Using multiple regression analyses, we identified hip extensor moment as the only significant predictive factor of the energy cost of walking, explaining between 35% and 38% of the variance in the energy cost of walking (Table 2). Together, these results agree with the majority of previous studies regarding the 'biomechanical plasticity/elasticity' of older adult's gait pattern ^{8,10,38,42} and highlights the key role of the hip to compensate decreased ability of older adults to use their ankle musculature during walking. Although experimental evidence is lacking, it has previously been suggested that the hip muscles (long fibers and short tendons with little elastic energy storing capacity) generate muscle forces during walking less efficiently than the triceps surae muscles (short fibers and long tendons with higher elastic energy storing capacity) ²⁰. In addition, a previous study based on computer simulations already suggested that using the hip extensor muscles during mid-stance instead of the ankle muscles during push-off to redirect the center of mass during walking increases the energy cost of walking ⁴³. Although, how much leg swing contributes to the metabolic cost of walking is still debated, it is clear that the hip flexors are active during swing and consume metabolic energy ⁴⁴. In our study, we found that older adults exhibited larger peak power produced by hip flexors during late stance/pre-swing compared to young adults which could also contribute to the overall increase in the energy cost of walking with age. More direct evidence should confirm this result in the future.

Influence of tendon properties and fascicle dynamics

We found a significant negative partial correlation between Achilles tendon stiffness and the energy cost of walking at both speeds (r : $-.52$ and $-.57$), with a trend for explaining part of the variance in the energy cost of walking at faster walking speed (i.e., $1.39 \text{ m}\cdot\text{s}^{-1}$). However, we did not find age-related differences in Achilles tendon stiffness, which is in contrast with most of previous literature ¹⁷. Therefore, in our study sample, Achilles tendon stiffness was not associated with age-related differences in the energy cost of walking. A selection bias might explain why Achilles tendon stiffness was not different between young and older participants in our study. Possibly, young adults were weaker and/or older adults were stronger in our study than in other studies. For instance, maximal isometric plantarflexor strength was on average 31% lower in older than in young adults in our study but 39% lower in a study that showed that Achilles tendon stiffness is lower in older than in young adults ⁵.

Furthermore, the similar Achilles tendon stiffness found between young and old adults is consistent with our observations of similar gastrocnemius medialis muscle-tendon interactions during walking in old and young adults. Medial gastrocnemius fascicle length and fascicle contraction velocity did not differ between young and older adults at both slow and fast walking speeds. This result is in agreement with a previous study ²⁰ at matched speed (i.e., $\sim 1.35 \text{ m}\cdot\text{s}^{-1}$) corresponding to the fast walking speed in our study. We further demonstrated that this similar behavior between age groups is also persistent at slower matched walking speed. Due to triceps surae working on the ascending limb of the force-length relationship during walking ^{20,21}, the shortening of the fibers tends to limit force-generation capacity and it is not surprising to find similar fascicle length changes between age groups. Unlike at matched speed, previous studies showed that older adults improved their triceps surae force-generation ability due to more favorable force-velocity potentials when walking at preferred speed ^{19,20}. Therefore, despite altered tendon properties or not (e.g., not found in our study), medial gastrocnemius in older

adults can operate relatively efficiently at both matched and preferred walking speeds ²⁰. While other findings on lateral gastrocnemius ¹⁸ are in line with these results, it seems that there is more room for debate on the role of soleus and the age-related differences in force-generation ability during walking ^{20,45}. It is possible that unfavorable shifts in soleus force-length and force-velocity potentials reduce the force-generating capacity in older adults, which may in turn drive older adults to use more hip extensors rather than ankle plantarflexors, and by extension increase the energy cost of walking.

Limitations

This study is not without limitations. Our results are dependent on the use of two-dimensional ultrasound to calculate three-dimensional muscle fascicle lengths and muscle-tendon junction displacements. Therefore, extreme care was taken to align the probe with the muscle fascicles and/or aponeuroses before treadmill and dynamometer testing. Our sample size was limited with a small number of young and older participants (N = 28). To calculate partial correlations, young and older adults were treated as one group based on the fact that changes in oxygen uptake ²², muscle-tendon function ⁴ and six-minute walk distance ⁴⁶ were gradual with age. To control for potential influence of age group on partial correlations, we visualized each relationship using scatter plots (see supplementary material). Causality cannot be inferred based on the results from the multiple regression analyses. However, we mainly based our interpretation using experimental findings for which the models confirm the key role of the hip joint in the energy cost of walking. Other study design may help to unravel which mechanisms drive the redistribution of joint kinetics during walking. Such design may be inspired by recent work with assisting/resisting treadmill setup where propulsive forces can be monitored/controlled to affect ankle push-off power ^{13,42} while the energy cost of walking could be simultaneously measured. The method used to evaluate Achilles tendon stiffness is

imperfect. Specifically, during isometric contractions on dynamometer, there is a lift of the heel from the footplate which decreases the accuracy of force measurements as well as ankle and muscle-tendon junction kinematics despite appropriate joint angular rotation corrections during processing³² (see Methods). Lastly, since our measurements only focused on medial gastrocnemius, our findings cannot be generalized to whole triceps surae. This is especially true considering the influence of ageing on Achilles tendon and sub-tendons dynamics (e.g., more uniform sub-tendons deformation in older adults during walking;¹⁴) and the differences in fascicle dynamics/function between triceps surae muscles^{18,20,45}.

Conclusion

In this study, we found a reciprocal interplay between hip and ankle function independent of walking speed with ageing which was associated with the energy cost of walking. We posit that the age-related reduction in older adults walking economy mainly comes from an increase in hip joint contribution to propulsive power during walking. This shift towards proximal leg muscles strategy with age likely comes from distal kinetics impairments (e.g., ankle power) during walking and other age-related neural and musculoskeletal changes (e.g., plantarflexor weakness) without altering medial gastrocnemius fascicle and tendinous tissue interactions. This study offers further evidence for the central role of hip extensors and flexors to the hallmark reduction in walking economy with ageing.

Perspectives

Training interventions targeting ankle plantarflexor do not always translate into improved ankle moment and power, and increased push-off intensity. In contrast, a training intervention that effectively target both ankle muscle strength and power, and increase propulsive forces during walking might reduce the energy cost of walking in older adults. Recent work using real-time biofeedback during treadmill walking^{13,42,47} combined with traditional resistance and

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461 plyometric training seems promising in order to restore or improve push-off capacity with
462 ageing (e.g., propulsive reserve capacity; ⁶).

463 **Conflicts of interest**

464 All authors declare that they have no conflict of interest

PROOF

REFERENCES

1. Cesari M, Kritchevsky SB, Penninx BWHJ, Nicklas BJ, Simonsick EM, Newman AB, Tylavsky FA, Brach JS, Satterfield S, Bauer DC, Visser M, Rubin SM, Harris TB, Pahor M. Prognostic value of usual gait speed in well-functioning older people - Results from the health, aging and body composition study. *J Am Geriatr Soc* 2005;53:1675–1680.
2. Schrack JA, Zipunnikov V, Simonsick EM, Studenski S, Ferrucci L. Rising energetic cost of walking predicts gait speed decline with aging. *Journals Gerontol - Ser A Biol Sci Med Sci* 2016;71:947–953.
3. Vandervoort AA. Aging of the human neuromuscular system. *Muscle and Nerve* 2002;25:17–25.
4. Narici M V, Maffulli N, Maganaris CN. Ageing of human muscles and tendons. *Disabil Rehabil* 2008;30:1548–1554.
5. Stenroth L, Peltonen J, Cronin NJ, Sipilä S, Finni T. Age-related differences in Achilles tendon properties and triceps surae muscle architecture in vivo. *J Appl Physiol* 2012;113:1537–1544.
6. Franz JR. The Age-Associated Reduction in Propulsive Power Generation in Walking. *Exerc Sport Sci Rev* 2016;
7. Boyer KA, Johnson RT, Banks JJ, Jewell C, Hafer JF. Systematic review and meta-analysis of gait mechanics in young and older adults. *Exp Gerontol* 2017;95:63–70.
8. Kulmala JP, Korhonen MT, Kuitunen S, Suominen H, Heinonen A, Mikkola A, Avela J. Which muscles compromise human locomotor performance with age? *J R Soc Interface* 2014;11:.

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51
52
53
54
55
56
57
58
59
60

9. DeVita P, Hortobagyi T. Age causes a redistribution of joint torques and powers during gait. *J Appl Physiol* 2000;88:1804–1811.

10. Waanders JB, Hortobágyi T, Murgia A, Devita P, Franz JR. Advanced Age Redistributes Positive but Not Negative Leg Joint Work during Walking. *Med Sci Sports Exerc* 2019;51:615–623.

11. 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;205:3717–3727.

12. Wert DM, Brach J, Perera S, VanSwearingen JM. Gait biomechanics, spatial and temporal characteristics, and the energy cost of walking in older adults with impaired mobility. *Phys Ther* 2010;90:977–985.

13. Conway KA, Franz JR. Increasing the propulsive demands of walking to their maximum elucidates functionally limiting impairments in older adult gait. *J Aging Phys Act* 2020;28:1–8.

14. Franz JR, Thelen DG. Imaging and simulation of Achilles tendon dynamics: Implications for walking performance in the elderly. *J Biomech* 2016;49:1403–1410.

15. Stenroth L, Sillanpää E, McPhee JS, Narici M V., Gapeyeva H, Pääsuke M, Barnouin Y, Hogrel JY, Butler-Browne G, Bijlsma A, Meskers CGM, Maier AB, Finni T, Sipilä S. Plantarflexor muscle-tendon properties are associated with mobility in healthy older adults. *Journals Gerontol - Ser A Biol Sci Med Sci* 2015;

16. Lichtwark GA, Bougoulas K, Wilson AM. Muscle fascicle and series elastic element length changes along the length of the human gastrocnemius during walking and running. *J Biomech* 2007;40:157–164.

17. Delabastita T, Bogaerts S, Vanwanseele B. Age-Related Changes in Achilles Tendon

- 511 Stiffness and Impact on Functional Activities: A Systematic Review and Meta-
512 Analysis. *J Aging Phys Act* 2020;27:116–127.
- 513 18. Mian OS, Thom JM, Ardigò LP, Minetti AE, Narici M V. Gastrocnemius muscle-
514 tendon behaviour during walking in young and older adults. *Acta Physiol* 2007;
515 19. Conway KA, Franz JR. Shorter gastrocnemius fascicle lengths in older adults associate
516 with worse capacity to enhance push-off intensity in walking. *Gait Posture*
517 2020;77:89–94.
- 518 20. Stenroth L, Sipilä S, Finni T, Cronin NJ. Slower Walking Speed in Older Men
519 Improves Triceps Surae Force Generation Ability. *Med Sci Sports Exerc* 2017;49:158–
520 166.
- 521 21. Lichtwark GA, Wilson AM. Optimal muscle fascicle length and tendon stiffness for
522 maximising gastrocnemius efficiency during human walking and running. *J Theor Biol*
523 2008;252:662–673.
- 524 22. Malatesta D, Simar D, Dauvilliers Y, Candau R, Borrani F, Prefaut C, Caillaud C.
525 Energy cost of walking and gait instability in healthy 65- and 80-yr-olds. *J Appl*
526 *Physiol* 2003;95:2248–2256.
- 527 23. Enright PL. The Six-Minute Walk Test. *Respir Care* 2003;48:783–785.
- 528 24. Meyer C, Killeen T, Easthope CS, Curt A, Bolliger M, Linnebank M, Zörner B, Filli L.
529 Familiarization with treadmill walking: How much is enough? *Sci Rep* 2019;9:1–10.
- 530 25. Van Rossom S, Smith CR, Zevenbergen L, Thelen DG, Vanwanseele B, Van Assche
531 D, Jonkers I. Knee cartilage thickness, T1ρ and T2 relaxation time are related to
532 articular cartilage loading in healthy adults. *PLoS One* 2017;12:1–16.
- 533 26. Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Guendelman E, Thelen

1
2
3 534 DG. OpenSim: Open-source software to create and analyze dynamic simulations of
4
5 535 movement. IEEE Trans Biomed Eng 2007;
6
7
8 536 27. De Groote F, De Laet T, Jonkers I, De Schutter J. Kalman smoothing improves the
9
10 537 estimation of joint kinematics and kinetics in marker-based human gait analysis. J
11
12 538 Biomech 2008;
13
14
15 539 28. Bolsterlee B, Gandevia SC, Herbert RD. Ultrasound imaging of the human medial
16
17 540 gastrocnemius muscle: How to orient the transducer so that muscle fascicles lie in the
18
19 541 image plane. J Biomech 2016;49:1002–1008.
20
21
22
23 542 29. Lichtwark GA, Wilson AM. Interactions between the human gastrocnemius muscle and
24
25 543 the Achilles tendon during incline, level and decline locomotion. J Exp Biol
26
27 544 2006;209:4379–4388.
28
29
30
31 545 30. Farris DJ, Lichtwark GA. UltraTrack: Software for semi-automated tracking of muscle
32
33 546 fascicles in sequences of B-mode ultrasound images. Comput Methods Programs
34
35 547 Biomed 2016;
36
37
38 548 31. Manal K, Cowder JD, Buchanan TS. Subject-specific measures of achilles tendon
39
40 549 moment arm using ultrasound and video-based motion capture. Physiol Rep 2013;1:1–
41
42 550 8.
43
44
45
46 551 32. Arampatzis A, Monte G De, Karamanidis K. Effect of joint rotation correction when
47
48 552 measuring elongation of the gastrocnemius medialis tendon and aponeurosis. J
49
50 553 Electromyogr Kinesiol 2008;
51
52
53
54 554 33. Werkhausen A, Cronin NJ, Albracht K, Paulsen G, Larsen A V., Bojsen-Møller J,
55
56 555 Seynnes OR. Training-induced increase in Achilles tendon stiffness affects tendon
57
58 556 strain pattern during running. PeerJ 2019;2019:.
59
60

- 1
2
3 557 34. Pataky TC, Vanrenterghem J, Robinson MA. Zero- vs. one-dimensional, parametric vs.
4
5 558 non-parametric, and confidence interval vs. hypothesis testing procedures in one-
6
7 559 dimensional biomechanical trajectory analysis. *J Biomech* 2015;48:1277–1285.
8
9
10 560 35. Das Gupta S, Bobbert MF, Kistemaker DA. The Metabolic Cost of Walking in healthy
11
12 561 young and older adults – A Systematic Review and Meta Analysis. *Sci Rep* 2019;9:1–
13
14 562 10.
15
16
17 563 36. Ortega JD, Farley CT. Effects of aging on mechanical efficiency and muscle activation
18
19 564 during level and uphill walking. *J Electromyogr Kinesiol* 2015;
20
21
22 565 37. Ortega JD, Farley CT. Individual limb work does not explain the greater metabolic cost
23
24 566 of walking in elderly adults. *J Appl Physiol* 2007;102:2266–2273.
25
26
27 567 38. Hortobágyi T, Finch A, Solnik S, Rider P, De Vita P. Association between muscle
28
29 568 activation and metabolic cost of walking in young and old adults. *Journals Gerontol -*
30
31 569 *Ser A Biol Sci Med Sci* 2011;
32
33
34 570 39. Huang T wei P, Shorter KA, Adamczyk PG, Kuo AD. Mechanical and energetic
35
36 571 consequences of reduced ankle plantar-flexion in human walking. 2015;3541–3550.
37
38
39 572 40. Clark WH, Franz JR. Triceps surae muscle–subtendon interaction differs between
40
41 573 young and older adults. *Connect Tissue Res* 2020;61:104–113.
42
43
44 574 41. Rasske K, Thelen DG, Franz JR. Variation in the human Achilles tendon moment arm
45
46 575 during walking. *Comput Methods Biomech Biomed Engin* 2017;20:201–205.
47
48
49 576 42. Browne MG, Franz JR. More push from your push-off: Joint-level modifications to
50
51 577 modulate propulsive forces in old age. *PLoS One* 2018;13:1–14.
52
53
54 578 43. Kuo AD. Energetics of actively powered locomotion using the simplest walking model.
55
56 579 *J Biomech Eng* 2002;124:113–120.
57
58
59
60

1
2
3
4
5
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7
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9
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42
43
44
45
46
47
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49
50
51
52
53
54
55
56
57
58
59
60

580 44. Umberger BR. Stance and swing phase costs in human walking. J R Soc Interface
581 2010;7:1329–1340.

582 45. Panizzolo FA, Green DJ, Lloyd DG, Maiorana AJ, Rubenson J. Soleus fascicle length
583 changes are conserved between young and old adults at their preferred walking speed.
584 Gait Posture 2013;38:764–769.

585 46. Enright PL, Sherrill DL. Reference Equations for the Six-Minute Walk.pdf. Am J
586 Respir Crit Care Med 1998;158:1384–1387.

587 47. Conway KA, Crudup KL, Lewek MD, Franz JR. Effects of Horizontal Impeding Force
588 Gait Training on Older Adult Push-Off Intensity. Med Sci Sport Exerc 2020;Publish
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591 Tables

Parameters	Young adults	Older adults	P-value	α_{SID}
Achilles tendon stiffness (N.mm ⁻¹)	233 ± 92	219 ± 108	0.663	0.05
Maximal plantarflexor strength (N)	2051 ± 531	1406 ± 299	0.007	0.05
Energy cost of walking at 0.83 m.s ⁻¹ (mlO ₂ .kg ⁻¹ .m ⁻¹)	0.108 ± 0.02	0.128 ± 0.02	0.005	0.023
Energy cost of walking at 1.39 m.s ⁻¹ (mlO ₂ .kg ⁻¹ .m ⁻¹)	0.110 ± 0.02	0.125 ± 0.02	0.044	0.023
Six-minute walk distance (m)	665 ± 59	551 ± 41	< 0.001	0.05

592 **Table 1.** Comparisons between Achilles tendon stiffness, maximal plantarflexor strength,
593 energy cost of walking and six-minute walk distance between young and older adults. Values
594 are presented as mean ± SD; α_{SID} represents critical alpha level corrected for multiple
595 comparisons with Sidak-method.

	0.83 m.s ⁻¹		1.39 m.s ⁻¹	
	Partial <i>r</i>	<i>p</i> value	Partial <i>r</i>	<i>p</i> value
Achilles tendon stiffness (N.mm ⁻¹)	-.516	.006	-.566	.002
Plantarflexor strength (N)	-.434	.024	-.473	.013
Hip extensor moment (N.m ⁻¹ .kg ⁻¹)	.481	.011	.384	.048
Hip power (W.kg ⁻¹)	.448	.019	.564	.002
Hip positive work (J.kg ⁻¹)	.505	.007	.453	.018
Ankle moment (N.m ⁻¹ .kg ⁻¹)	-.067	.739	-.266	.180
Ankle power (W.kg ⁻¹)	-.237	.234	-.234	.240
Ankle positive work (J.kg ⁻¹)	-.110	.585	-.017	.931
MG fascicle velocity (L/L _{standing})	-.027	.893	.075	.712
Walking distance (m)	-.339	.083	-.303	.124

597 **Table 2.** Partial non-parametric correlations between independent variables and the energy cost
598 of walking at 0.83 m.s⁻¹ and 1.39 m.s⁻¹. Adjusted for step frequency. MG means medial
599 gastrocnemius. This value corresponds to the fascicle velocity at peak ankle moment during
600 stance.

	R^2	Adjusted R^2	Standard error of the estimate	Unstandardized β coefficient	Standardized β coefficient	p value
<i>Model 1: 0.83 m.s⁻¹</i>	<i>0.448</i>	<i>0.379</i>	<i>0.015</i>			0.002
(Constant)				.093		
Hip extensor moment				.098	.562	0.012
Plantarflexor strength				-4.50E ⁻⁵	-.159	.426
Achilles tendon stiffness				-5.86E ⁻⁶	-.225	.252
<i>Model 2: 1.39 m.s⁻¹</i>	<i>0.418</i>	<i>0.345</i>	<i>0.018</i>			0.004
(Constant)				.027		
Hip extensor moment				.024	.381	0.040
Plantarflexor strength				.0001	-.142	.507
Achilles tendon stiffness				.0001	-.360	.074

Table 3. Adjusted multiple regression models at both slow and fast speeds with energy cost of walking as dependent variable and plantarflexor strength, Achilles tendon stiffness and hip extensor moment as independent variables. Hip extensors moment was the only significant predictor of the models which predict between 35% and 38% of the variance.

Figure legends

Fig.1. Comparisons of medial gastrocnemius muscle-tendon length (A), tendinous tissue length (B), fascicle length (C) and fascicle velocity (D) during walking in young (blue lines) and older adults (red lines). Length were normalized respective to length measured during the static standing trial. Results for repeated measures ANOVA's with statistical non-parametrical mapping (SNPM) are depicted on the right (main effect group). No significant group differences were found at both speeds between young and older adults.

Fig.2. Comparisons of ankle moment (A) and hip moment (B) during walking in young (blue lines) and older (red lines) adults at 0.83 m.s^{-1} and 1.39 m.s^{-1} . Negative values indicate ankle plantarflexor moment and hip extensor moment. Results for repeated measures ANOVA's with statistical non-parametrical mapping (SNPM) are depicted on the right (main effect group). Larger peak ankle moment was found in young adults during the push-off phase. Larger hip extensor moment at early/mid-stance and hip flexor moment at pre-swing were observed in older adults.

Fig.3. Comparisons of ankle power (A) and hip power (B) during walking in young (blue lines) and older (red lines) adults at 0.83 m.s^{-1} and 1.39 m.s^{-1} . Negative values indicate ankle plantarflexor moment and hip extensor moment. Results for repeated measures ANOVA's with statistical non-parametrical mapping (SNPM) are depicted on the right (main effect group). Larger peak ankle power was found in young adults during the push-off phase. Larger hip peak power was observed in older adults.

Fig.4. Ankle total positive work (A) and hip total positive work (B) during walking in young (blue circles) and older (red circles) adults at 0.83 m.s^{-1} and 1.39 m.s^{-1} . * indicates significant difference between young and older adults. Larger total positive work at the ankle was found in young adults at slow and fast speed. Inversely, larger total positive work at the hip was found in older adults at both speeds.

211x280mm (300 x 300 DPI)

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