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Effect of age and speed on the step-to-step transition phase during walking

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ABSTRACT

Gait is a powerful measurement tool to evaluate the functional decline throughout ageing. Falls in elderly adults happen mainly during the redirection of the center of mass of the body (CoM) in the transition between steps. In young adults, this step–to–step transition begins before the double contact phase (DC) with a simultaneous forward and upward acceleration of the CoM. We hypothesize that, compared to young adults, elderly adults would exhibit unbalanced contribution of the back leg and the front leg during the transition. We calculated the mean vertical push-off done by the back leg (F_{BACK}) and the mean impact force on the front leg (F_{FRONT}) during the transition. Eight young (mean ± SD; age: 24 ± 2 y) and 19 elderly (age: 74 ± 6 y) healthy adults walked on a force-measuring treadmill at five selected speeds ranging from 0.56 to 1.67 m·s^{-1}. Results show that, at mid and high speeds, elderly adults exhibit a smaller F_{BACK} compared to young adults, possibly linked to the decreased plantar flexion of the back foot. As a consequence, F_{FRONT} is significantly increased and the transition begins lately in the step, at the beginning of DC. Also, elderly adults show an inability to accelerate the CoM upward and forward simultaneously. Our findings show a different adaptation of the step–to–step transition with speed in elderly adults and identify two potential indicators of gait impairment with age: the F_{FRONT}/F_{BACK} contribution and the synchronization between the upward and forward acceleration of the CoM during the transition.

1. Introduction

Gait is a powerful measurement tool to evaluate the rate of functional decline throughout old age and falling is the most frequent reason of unintentional injuries in elderly adults. More than 25% of elderly adults fall each year (Tromp et al., 2001) and these falls are often associated with serious injuries and hospitalizations (Sterling et al., 2001). Knowing that the percentage of worldwide population over 60 year-old will continue to rise, the prevention of falls is a major issue for our society.

It is well known that the spontaneous speed decreases with age (Alcock et al., 2013; Bohannon, 1997), but it is still unclear if the modifications of the walking pattern that has be observed in elderly adults (Boyer et al., 2017) come from the reduction of the spontaneous speed or is a direct effect of age or a combination of both. For instance, the reduced speed could be the consequence of other differences in mechanics such as limitations in joints' mobility, e.g. a decrease of the plantar flexion as mentioned by Boyer et al. (2017). The need to understand how differences in age and walking speed affect the dynamic stability is obvious (Kang and Dingwell, 2008; Monaco et al., 2009). Most of the falls occur during the body weight transfer from one leg to the other when elderly adults are simply walking forward (Blake et al., 1988; Robinovitch et al., 2013). During that phase, the downward trajectory of the centre of mass of the body (CoM) has to be redirected to an upward trajectory (Bastien et al., 2003); this part of the stride is often referred as the step–to–step transition (Kuo et al., 2005). The influence of the walking speed on this transition has already been studied in young subjects (Adamczyk and Kuo, 2009) and in elderly adults at various but uncontrolled speeds, i.e speeds relative to individual preferred speed (Hernandez et al., 2009). Interestingly, the latter authors observed a greater use of the front leg compensating a reduced vertical support of the back leg at spontaneous and fast speeds in elderly adults. In addition, the transition seems to be initiated later in the gait cycle with age at least at a medium speed (Fig. 4 in Franz and Kram, 2013). In order to isolate the effect of age on the step–to–step transition, a comparison of both populations walking at matched speeds, and on a wide range of speeds, is required.

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The aim of this study is to investigate the step–to–step transition during gait in young and aged populations. Data are collected with an instrumented treadmill allowing measurement at same constant walking speeds for both populations. The forces exerted by each leg, the CoM trajectory and the kinematics of the lower limb are measured and especially analysed during the step–to–step transition.

We hypothesize that, compared to young adults, elderly adults would exert less vertical force on the back leg and more on the front leg during the transition. At a given speed we also anticipate a delayed step–to–step transition within the gait cycle and a decreased range of motion of the ankle with age. Finally, we expect that these modifications will offer insight into the mechanisms by which gait becomes hazardous for elderly people and provide clues to explain and prevent falls.

2. Materials and methods

2.1. Subjects and experimental procedures

Eight young (age: 21 to 28 yo, weight: 71.9 ± 16.7 kg; height: 1.76 ± 0.08 m, BMI: 23.0 ± 3.8 kg.m^{-2}) and nineteen elderly (age: 67 to 86 yo, weight: 65.9 ± 14.5 kg; height: 1.68 ± 0.09 m, BMI: 23.0 ± 3.6 kg.m^{-2}) healthy adults were enrolled in the study. There was no significant difference between the two groups in terms of body mass, height and BMI. The inclusion criteria were no history of falling, ability to walk a kilometre, no current locomotor system injury complaints, no history of neurological disorder and a body mass index between 15 and 35 kg/m^2. The additional specific criteria for elderly adults were to be over 65 years and present no risk of falling according to the Tinetti Gait Balance Examination (score > 25/28). Before the experiments, the purpose and the nature of the study were explained to the subjects. The experiments were performed according to the Declaration of Helsinki and were approved by the local ethics committee ("Commission d'éthique Biomédicale Hospitalo-Facultaire de l'Université catholique de Louvain", 2015/18MAI/245, Belgian Registration Number: B403201524765). All subjects gave their written informed consent to participate to the study.

The subjects were asked to walk on an instrumented treadmill at five different selected speeds of 0.56; 0.83; 1.11; 1.39 and 1.67 m·s^{-1}. The order of speeds was randomized and the speed could not exceed 150% of the spontaneous speed of the subject, previously measured in a corridor. The mean spontaneous speeds were equal to 1.32 ± 0.10 and 1.20 ± 0.15 m·s^{-1} for young and elderly adults, respectively. There was no significant difference of spontaneous speed between the two groups. Elderly subjects were equipped with a harness during tests and all subjects performed a warm up trial on the treadmill before data acquisitions. For each speed, data were recorded for 15 s after 3 to 5 min of waking at constant speed.

2.2. Experimental set-up and signals processing

The ground reaction forces. The ground reaction forces (GRF) were measured by means of four force transducers (Arsalis, Belgium) located under each corner of a modified commercial treadmill (h/p/Cosmos-Stellar, Germany, belt surface: 1.60x0.65 m). The forces were filtered with an 8th order Bessel dual low pass filter with a cut-off frequency of 20 Hz. The non-linearity was <1% of full scale, and the crosstalk between vertical and horizontal GRF <1%.

The GRF signal was digitized with a 16-bit A/D convertor at a sampling rate of 500 Hz. The amplified GRF signals were processed by means of a computer with dedicated software (LABVIEW 2010, National Instruments, Austin, TX, USA). A decomposition algorithm proposed by Meurisse et al. (2016) was used to determine the beginning and the end of the double contact phase, the front foot contact time and the back toe-off time, and the vertical component of the GRF, expressed in body weight, acting upon each limb during the double contact phase.

The GRFs were used to compute the velocity and the displacement of the CoM (Cavagna, 1975). Data were normalised relative to the stride duration, 0% corresponding to the initial contact of the right foot and 100% corresponding to the next contact of the same foot (Fig. 1).

The kinematics of the lower segments. The angular displacements in the sagittal plane were measured by means of a high-speed video camera placed on the right side of the subject (BASLER piA640–210gc, resolution of 2.4 pixels/cm). The camera was fixed 3 m to the side of the treadmill. Images were sampled at a rate of 200 fps and synchronized with the GRF signals by means of a digital trigger signal. Five reflective markers were taped on the skin of the subject, at the chin-neck intercept, the greater trochanter, the external femoral condyle, the lateral malleolus and the fifth metatarsal phalangeal joint.

Coordinates of the reflectors were measured using a dedicated tracking software (LABVIEW 2010, National Instruments, Austin, TX, USA) that allowed measuring the position of the different joints. From the coordinates of the reflectors, the joint angle of the hip, the knee, the ankle and the trunk forward inclination angle were calculated at each frame (maximal error: 1.75°). Angular signals were interpolated with a cubic spline function to obtain a 500 Hz signal. The data were filtered with a 2nd order Bessel dual low pass filter with a cut-off frequency of 20 Hz. Positive values correspond to flexion/dorsiflexion and negative values correspond to extension/plantar flexion.

2.3. Temporal, kinetic and kinematic parameters

Temporal parameters. The stride duration was calculated as the duration between two successive right heel strikes and the stride frequency was obtained by taking the inverse of the stride duration. The double contact (DC) duration was calculated as the duration between the front foot contact and the back foot toe-off. The step–to–step transition phase was calculated as the duration between the time where the CoM has reached its minimal vertical velocity before the front foot contact and its maximal vertical velocity after the back foot toe-off (Fig. 1) as defined by Franz and Kram (2013). The beginning of DC was reported relative to the duration of the step–to–step transition (DC timing, in %, Fig. 2).

Kinetic parameters. The mean values of the vertical push off done by the back leg (F_{BACK}) and of the impact force on the front leg (F_{FRONT}) were calculated during the step–to–step transition. The force ratio between F_{FRONT}/F_{BACK} was calculated to illustrate the asymmetry in the force generation between the two legs. The mean values were preferred over the peak values as they are more representative of the contribution of each leg during the step–to–step transition.

CoM hodograph. The vertical and horizontal components of the CoM velocity are plotted against each other over the course of the step to form a CoM hodograph (Fig. 3) as described by Adamczyk and Kuo (2009).

Kinematic parameters. The range of motion of the plantar flexion, the peak extension of the hip of the back leg and the trunk inclination were calculated during the step–to–step transition (Fig. 4).

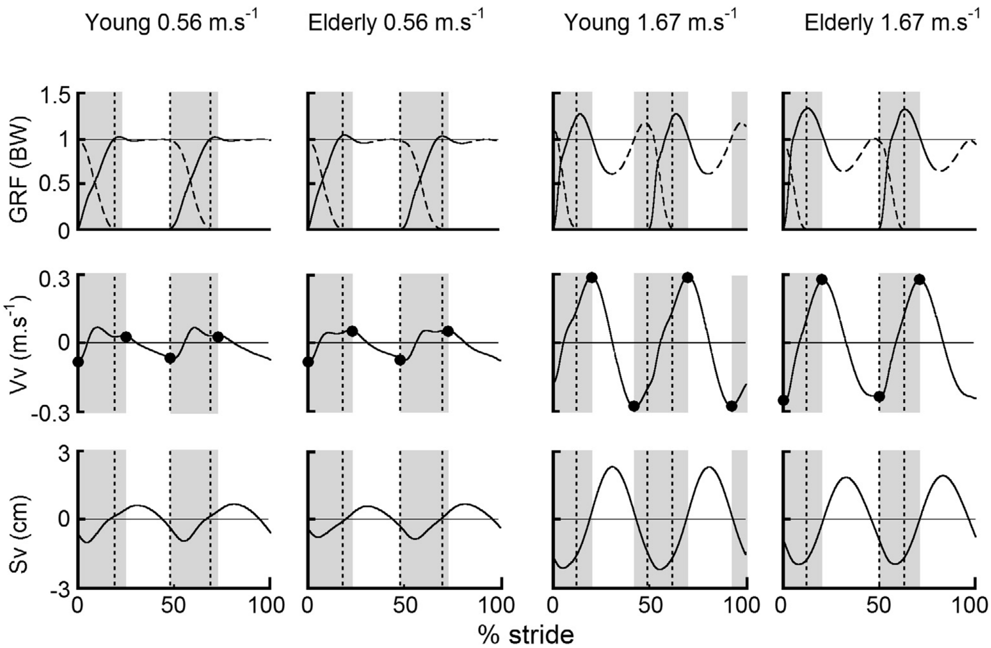

Fig. 1. Kinetic traces for young and elderly adults at slow and fast speeds. Mean traces of the vertical component of the ground reaction force (GRF, expressed in body weight) acting upon each leg separately (continuous line: front leg; dotted line: back leg), of the vertical component of the velocity of the CoM (Vv, expressed in $m{\cdot}s^{-1}$) and its displacement (Sv, expressed in cm) during a stride for both groups walking at slow ($0.56\ m{\cdot}s^{-1}$, left) and fast speeds ($1.67\ m{\cdot}s^{-1}$, right). The vertical dotted lines indicate the double contact phase (DC); the grey areas represent the step-to-step transition phase between the minimum of Vv before DC and the maximum of Vv after DC.

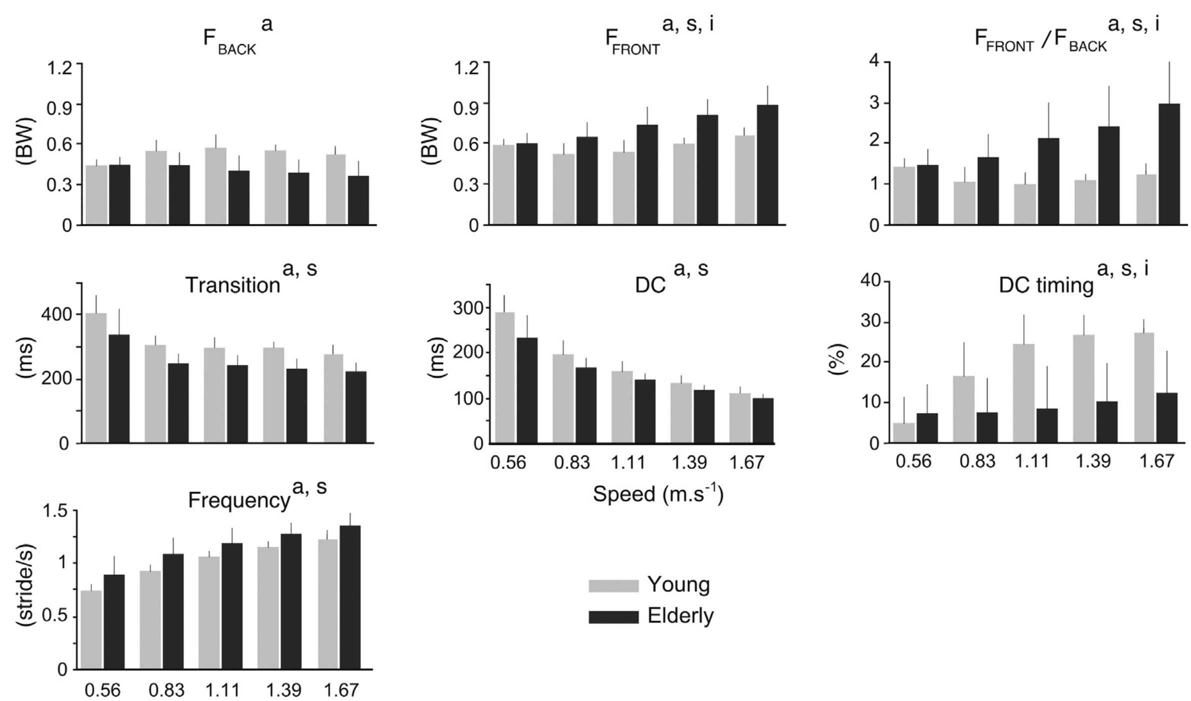

Fig. 2. Temporal and kinetic results. Mean value (bar) + 1 SD (tick) of F_{BACK} (BW), F_{FRONT} (BW), ratio F_{FRONT}/F_{BACK}, DC duration (ms), step-to-step transition duration (ms), DC timing (% of the step-to-step transition) and the stride frequency (stride/s) for each walking speed. Age groups are differentiated by the colours (grey: young subjects; black: elderly subjects). *a*: significant age effect; *s*: significant speed effect; *i*: significant interaction of age and speed effects. N = 228 strides (young: 58 and elderly: 170) at $0.56\ m{\cdot}s^{-1}$, 282 strides (Y: 78 and E: 204) at $0.83\ m{\cdot}s^{-1}$, 311 strides (Y: 88 and E: 223) at $1.11\ m{\cdot}s^{-1}$, 327 strides (Y: 98 and E: 229) at $1.39\ m{\cdot}s^{-1}$ and 314 strides (Y: 108 and E: 206) at $1.67\ m{\cdot}s^{-1}$.

2.4. Statistics

For each subject, the variables were measured and averaged during 5 to 16 walking strides depending on the speed before computing the mean values for the two groups of subjects (as presented in the result section and figures).

A linear mixed model for repeated measures (PASW Statistics 19, SPSS, IBM, Armonk, NY, USA) was used to analyse the variables

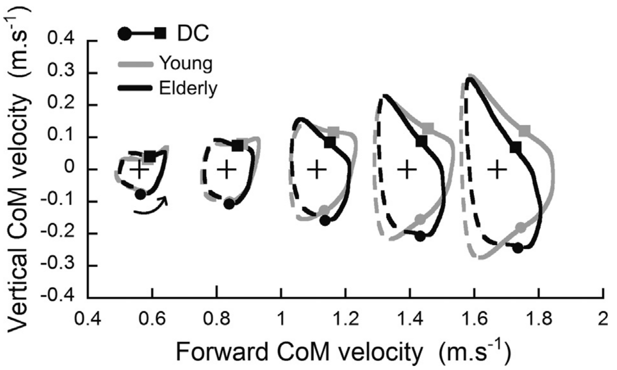

Fig. 3. Hodograph. Mean plot of the vertical (Vv) and forward (Vf) components of the CoM velocity during a step at each walking speed, from 0.56 to 1.67 m·s^{-1} (grey: young subjects; black: elderly subjects). Mean speeds are represented by a +. The step-to-step transition phase is illustrated by the continuous line. The DC phase is delimited by the circle (front foot contact) and the square (back foot toe-off).

for a fixed effect of the age group, speed class and their interaction. The normality of the residuals was checked with a Kolmogorov-Smirnov test.

3. Result

3.1. Kinetics

The Fig. 1 presents the mean traces of the vertical ground reaction force (GRF) acting upon each leg separately, the vertical component of the velocity (Vv) and the vertical displacement (Sv) of the CoM during a stride for young and elderly adults walking at slow and fast speeds. The grey areas delimit the step–to–step transition and the dotted lines delimit the double contact phase (DC). At slow speeds, GRF, Vv and Sv amplitudes are very similar between both groups. For elderly as for young adults, the step–to–step transition is initiated by the DC and is mainly done during DC; DC represents more than 3/4th of the transition phase. At faster speeds, the GRF, Vv and Sv amplitudes increase largely and the step–to–step transition is now much larger that DC; the DC period now represents less than half of the transition phase. Differences between groups become evident: (a) the GRF pattern of the front vs back foot is clearly asymmetric for elderly compared to young adults and (b) the step–to–step transition is still initiated by DC for elderly adults while the transition begins well before DC for young adults.

The Fig. 2 illustrates the differences between both groups. At all speeds, elderly adults have a significant smaller F_{BACK} and a significant greater F_{FRONT} than young adults. Moreover for elderly adults, F_{FRONT} is always greater than F_{BACK} and increases with speed so that the force ratio F_{FRONT}/F_{BACK} is greater than 1. In addition F_{FRONT} increases to reach a value three times greater than F_{BACK} at the fastest speed (Fig. 2, speed effect $p < 0.05$; age effect $p < 0.001$; interaction $p < 0.001$). In comparison for young adults, the force ratio F_{FRONT}/F_{BACK} remains close to 1 whatever the walking speed.

For the temporal parameters, it can be observed that, as speed increases, the DC represents a smaller fraction of the step–to–step transition; the DC duration decreases relative to the transition duration (Fig. 2). Noteworthy at all speeds, elderly adults have significant shorter transition duration compared to young adults (Fig. 2, age effect $p < 0.001$). Another difference is the relation between the DC phase and the transition phase: as speed increases, the DC occurs later in the transition for young adults, while DC remains in the first 15% of the transition for elderly adults (Fig. 2, speed effect $p < 0.001$; age effect $p < 0.001$; interaction $p < 0.01$).

The stride frequency increases with speed for both groups. However at all speeds, elderly adults have significant greater stride frequency compared to young adults (Fig. 2, age and speed effects $p < 0.001$). The difference between groups is such that the stride

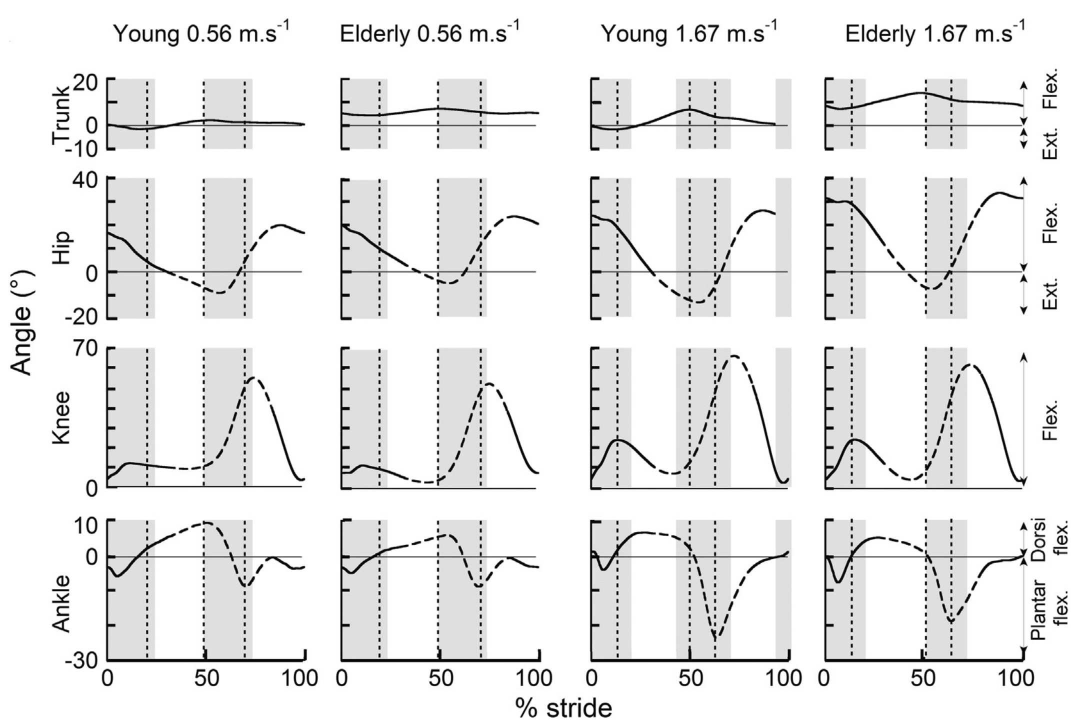

Fig. 4. Kinematic traces for young and elderly adults at slow and fast speeds. Mean traces of the kinematics taken from the right side of the subjects during a stride for both groups walking at slow (0.56 m·s^{-1}, left) and fast speeds (1.67 m·s^{-1}, right). Continuous line: front leg; dotted line back leg. From top to bottom: the trunk inclination, angles of the hip, knee and ankle joints. Positive values correspond to flexion/dorsiflexion and negative values correspond to extension/plantar flexion. Other indications as in Fig. 1.

frequency of elderly adults at one speed is comparable to the stride frequency of young adults but at one speed slower.

3.2. CoM hodograph

The Fig. 3 presents the mean plot of the vertical (Vv) and forward components (Vf) of the CoM velocity during a step at each walking speed, from slow to fast speeds. The continuous lines graphically illustrate the step–to–step transition during which the CoM is redirected upward. At slow speeds, the restoration of the upward and forward velocity is similar for both groups characterised by a successive increase of the forward and vertical speeds: first Vf increases before the transition and then Vv increases during the transition. At speeds at and above 1.11 $m \cdot s^{-1}$, a different strategy is used in young adults to restore the speed of the CoM: the transition starts before DC and there is a simultaneous increase of Vf and Vv. In elderly adults the strategy does not change: the dichotomic pattern is used at all speeds; Vf increases before the transition and Vv increases during the transition.

3.3. Kinematic

The Fig. 4 presents the mean traces of the hip, knee, ankle angles and trunk inclination during a stride for both groups at slow and fast speeds. The range of motion of the ankle is smaller for elderly compared to young adults at the slow speeds, and the difference increases with speed. Another difference between both groups is the more bent forward inclination of the trunk as well as a hip more flexed for elderly adults. These differences are significant during the step–to–step transition: for elderly adults, the trunk is more inclined forward and this inclination increases with speed (Fig. 5, age effect $p < 0.001$; speed effect $p < 0.01$). For elderly adults also, the peak of the hip extension is reduced independently of speed (Fig. 5, age effect $p = 0.002$). Finally, the plantar flexion of the back foot during the step–to–step transition increases with speed but is systematically lower for elderly adults compared to young adults (Fig. 5, age effect $p < 0.001$; speed effect $p < 0.001$).

4. Discussion

The aim of this study is to determine the step–to–step transition differences between young and aged populations as a function of speed. Our results show that there is an effect of age on the kinetic, the temporal and the kinematic parameters. Indeed the forces exerted by the back or the front leg, the duration of the transition, the stride frequency but also the inclination of the trunk, the angles of the hip and ankle show an age-dependent significant effect. However, the step–to–step transition of young and elderly adults diverges clearly only at speeds equal to or greater than 1.11 $m \cdot s^{-1}$, corresponding to the speed at which young adults change the strategy to redirect the trajectory of the CoM. The discussion will focus on these speeds, equivalent to spontaneous walking speed and above. Even though walking at high speed on a treadmill is not representative of the everyday life of elderly adults, our results highlight that this stress testing allows detecting degradations of the walking pattern.

During normal walking, adult subjects initiate the step–to–step transition during the single support phase with a back leg's push-off produced by the hip extension and the ankle plantar flexion. Indeed, Schloemer et al. (2017) identified the plantar flexor muscles as the principal muscular component contributing to the push-off. And Giest and Chang (2016) showed with increasing speed that unilateral transtibial amputees increase F_{BACK} on the intact leg, but not on the impaired leg, obviously with no active plantar flexion.

In young adults, the step–to–step transition seems to be smooth, anticipated and balanced. Indeed, the CoM is accelerated and redirected upwards thanks to push-off and almost a quarter of the transition phase happens before the other foot contacts the ground (Fig. 1, Fig. 2). During this first part of the transition, the vertical and the forward velocity of the CoM both increase simultaneously (Fig. 3). And then the transition effort moves on progressively from the back to the front leg until the latter supports alone the last part of the transition. In this way, the front and the back legs contribute equally to the transition as is illustrated by the F_{FRONT}/F_{BACK} force ratio (Fig. 2).

The equal contribution of the front and the back legs is no longer present for elderly adults as shown by the F_{FRONT}/F_{BACK} force ratio (Fig. 2). Indeed, the vertical push-off produced by the hip extension and the ankle plantar flexion is reduced with age at all speeds (Fig. 1, Fig. 2). Franz and Kram (2013) measured older adults while walking uphill on a treadmill at 1.25 $m \cdot s^{-1}$ and showed a decrease of this push-off at the different slopes. The same authors, in 2014, also observed a decrease of the ankle plantar flexion and of the hip extension for elderly adults walking with a slope. In general, the reduction of the plantar flexion with ageing is commonly observed in the literature (Boyer et al., 2017). From those different sources, we can reasonably hypothesize that the increased hip extension and the reduced planter flexion, described in the literature (DeVita and Hortobagyi, 2000; Franz and Kram, 2014; Judge et al., 1996; Monaco et al., 2009), and that we also observed (Fig. 5) are linked to the decrease of the push-off of the back leg. The reduction of the plantar flexion in elderly adults can be set in relation to a reduced capacity of the plantar flexors to generate force or to a reduced force generated for a same speed as young subjects. Indeed, Judge et al. (1996) and Monaco et al. (2009) have described a reduction of the force produced by the plantar flexors with age during walking. This reduced plantar flexors activity has been described as a limiting factor of the walking speed for elderly adults as they were forced to modify their walking pattern to reach the fastest speeds of young adults (Anderson and Madigan, 2014; Kerrigan et al., 1998). Stenroth et al. (2017) have suggested a less effective use of tendinous tissue elasticity of the ankle plantar flexors. However, Franz (2016) noticed that

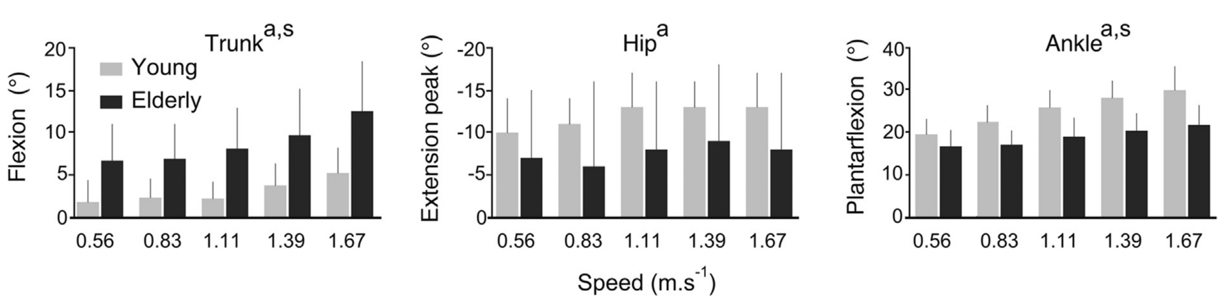

Fig. 5. Kinematic results. The mean values of the trunk inclination, the peak of the hip extension and of the plantar flexion of the back leg during the step-to-step transition are presented for each walking speed. Other indications as in Fig. 2.

elderly adults underutilize their available muscular capacity of the ankle flexors and were, in fact, able to increase their generated propulsive power during the uphill walk for example. According to this author, the reduced plantar flexors activity may reflect a trade-off between propulsive force generation and walking balance control.

The weak force production of the plantar flexors could be compensated by a more flexed hip position as illustrated in the Figs. 4 and 5. Indeed as mentioned by others, this position allows increasing the force generation by the hip extensors in order to counteract the decrease of force produced by the ankle (DeVita and Hortobagyi, 2000; Franz and Kram, 2014; Judge et al., 1996; Monaco et al., 2009). In addition in elderly adults, the increased flexion of the hip can be linked to the more forward inclination of the trunk in elderly adults (Fig. 5). This compensation allows elderly adults to sustain and increase the walking speed but nevertheless, the force distribution between the front and back leg for the step–to–step transition remains impaired. A relationship between the decreased plantar flexion of the ankle, the more flexed position of the hip and the "unbalanced" F_{FRONT}/F_{BACK} force ratio has emerged, but it's difficult to untangle the direction of the causation.

The step–to–step transition temporal differences between groups are also possible causes of the decrease of the vertical push-off in elderly adults. Indeed, elderly adults have higher stride frequency and shorter step–to–step transition at any given speed compared to young adults (Fig. 2). In our view, elderly adults combine two disadvantages; they have less time to perform the push-off as well as weak or underused ankle plantar flexors.

In elderly adults the reduced vertical push-off performed by the back leg affects the transition: the DC initiates the transition (Fig. 2). Chong et al. (2009) described this initiation of the transition by DC at high speed as the inability of elderly adults to accelerate upward the CoM before DC. Consequently, the CoM accelerates forward, only, before the transition and then upward when the transition begins, thanks to the force exerted by the front foot. The incapacity to increase simultaneously the vertical and forward components of the CoM velocity could be an early sign of the deterioration of the walking pattern with age. This observation leads us to consider the CoM hodograph as a helpful tool for examining abnormal or impaired gaits, as proposed by Adamczyk and Kuo (2009), particularly in elderly walking.

Another consequence of the weak push-off of the back leg is the necessary compensation to be done by the front leg in order to redirect and accelerate upward the CoM. Indeed in elderly adults the force developed by the front leg is significantly greater during the transition, for example, at mid speed the solicitation of the front leg reaches two times the one on the back leg (Fig. 2). This overuse of the front leg and the underlying higher muscular control may be challenging and could increase the risk of fall in a weakened subject, particularly at the end of the transition when the leg supports all the bodyweight.

We acknowledge that the present study focuses on the modification of the vertical ground reaction force with speed in elderly adults, while the anterior-posterior force could also be relevant. Indeed, the decrease of the vertical push-off has been shown to be associated with a decrease of the anterior-posterior force in elderly adults (Boyer et al., 2012; Franz and Kram, 2013). Also, according to Browne and Franz (2017), a reduced anterior-posterior push-off affects the dynamic stability. In line, the reduced width of the hodograph loops that we observe for our elderly adults reflects less variation of the CoM forward velocity and therefore a diminished anterior-posterior push-off. Further investigations are necessary to investigate the age and speed effect on the anterior-posterior and mediolateral ground reaction force.

In conclusion, our observations on non-falling elderly adults show that they have, at medium and high speeds, a lower vertical back leg's push-off, probably due to a reduced force generated during the plantar flexion movement of the ankle. This reduction could come from either a decreased time to perform the push-off and/or weak/underused ankle plantar flexor muscles. The incomplete push-off affects the step–to–step transition: the transition phase begins at the front foot contact, the force generated by the front leg is increased and the velocity of the CoM increases only upward during this phase. These modifications could influence their equilibrium during walking.

The F_{FRONT}/F_{BACK} force ratio and the CoM hodograph seem rich indicators of the modification of the walking patterns and are helpful to detect gait impairment with age. Moreover, as proposed by Franz (2016), rehabilitative approaches to preserve mobility in aging population should target walking exercise with online biofeedback. In our view, at least the force ratio can be used for this purpose.

Future studies should focus on elderly adults with risk or history of falling and confirm that these parameters could provide clues to avoid falls.

Conflict of interest

The authors have no conflict of interest to declare.

Acknowledgments

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