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THE ROLE OF ANTHROPOMETRIC CHANGES DUE TO AGING ON HUMAN WALKING: MECHANICAL WORK, PENDULUM AND EFFICIENCY

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ABSTRACT: The aging process modifies body composition and the inertial properties of body limbs might change accordingly. Pendular energy exchange, mechanical work and locomotion efficiency should be affected by these changes. To check this hypothesis, seven elderly subjects were asked to walk on a treadmill at five speeds ranging from 0.55 to 1.66 $m \cdot s^{-1}$. The internal work is indeed reduced when calculated by using specific anthropometric tables for the elderly. The pendular recovery and external work are not affected by the anthropometric profile. Our results suggest that the mass-specific mechanical work based on anthropometric tables consistently decreases with age and consequently the greater metabolic cost is counterbalanced, in part, by decreased mechanical work, resulting in a similar locomotion efficiency.

KEY WORDS: locomotion, elderly, mechanical work

INTRODUCTION

Locomotion is one of the most relevant aspects in the evolution of the species. The way an animal moves is a decisive factor in the process of natural selection [1,23]. Thus, the mechanics and energetics of animal locomotion have been a target of study for a long time.

The mechanical energy and the consequent mechanical work generated during a step are the main aspects related to locomotion analysis, from a mechanical point of view [23,26]. The work done to elevate and accelerate the centre of body mass (COM) in relation to the environment (external work, W_{ext}) and the work associated with the body segment acceleration (mainly limbs) in relation to the COM (internal work, W_{int}) have been widely investigated in the literature [4,6,16,26].

W_{ext} can be easily calculated from the ground reaction forces (GRF), once the mass and average horizontal speeds are known [26]. On the other hand, W_{int} is very hard to measure, as the fluctuations of the mechanical energy for each segment related to the COM make it necessary to collect and process data with more associated artefacts.

In spite of its relevance, there are few researchers who routinely calculate W_{int} due to the methodological limitations. A model equation proposed by Minetti [16] presents a mathematical model of W_{int}, using as input variables speed, step frequency (*sf*) and duty factor (*d*). This model accepts as constant the anthropometric factors directly related to calculation of W_{int} (0.1, see equation 4 in [16]). More recently, W_{int} has been observed to be greater in the elderly than in young people [15]. Again, the anthropometric tables used for young and elderly were the same.

There is solid evidence to support the observation that body segment inertial parameters are altered with aging, primarily thigh, lower leg, upper arm and forearm [17,21], as well as the muscle mass distribution [10]. In addition, these differences imply that specific age-related regression equations may be required to provide accurate estimates of body segment parameters to be used in kinetic equations of motion [9].

Our hypothesis is that the anthropometric changes due to aging, such as lower body and percent segmental masses, would counterbalance the increased *sf* [12] and larger upper limb movements [15] leading to a lower W_{int} and consequently, a smaller total mechanical work (W_{tot}) in older women. To test this hypothesis, we measured: i) the external and internal mechanical work to maintain the motion of the COM and the centres of the segmental mass, respectively; ii) the mass specific cost of transport (*C*) in elderly females walking on the level at different speeds. Furthermore, in each subject we calculated the percentage efficiency (*eff*) of positive work production, percentage pendular recovery (*R*) and W_{int} according to two different equations available from the literature [6,16].

MATERIALS AND METHODS

Seven healthy elderly females (age: 70-75 years old; height: 1.63-1.70 m, body mass: 56.2-70.9 kg) walked on the treadmill for 3 minutes at 5 speeds: 0.55, 0.83, 1.11, 1.38, 1.66 $m{\cdot}s^{-1}$. Reflective markers were placed on the following anatomical landmarks: fifth metatarsal, calcaneus, lateral malleolus, femoral epicondyle, greater trochanter, acromion, lateral epicondyle of humerus, middle point ulnar-radius, temporal. At each speed they were videotaped (100 Hz) 3D during the last minute, using the movement analysis system Dvideow (Laboratory of biomechanics – FEF Institute of Computing UNICAMP, Campinas, Brazil) with two cameras. The routines in Matlab version 6.3 (Mathworks, Inc, USA) were developed for the automatic digitalization. Low-pass filtering of coordinate data was then performed using the procedure of residual analysis [27], which automatically selects optimal filter cut-off frequencies for each marker in x, y and z directions. Filtered coordinate data, together with classical [8] and specific (for elderly; [21]) regression equations providing anthropometric data of 11 rigid segments (head-trunk, upper arms, lower arms, thighs, lower legs, feet) were used to compute the position of the segments and the COM and the vertical height of the medial malleoli.

The study was approved by the institutional ethics committee (Federal University of Rio Grande do Sul, Brazil) and all participants were made aware of potential risks and discomforts before signing an informed consent form. Those admitted after screening were healthy community individuals, free from frailty or signs of gait impairment, and participated in a physical activity group for the elderly. They were also able to walk on a treadmill for sufficient time to complete a metabolic analysis and had no other major medical disorders. Subjects were excluded if they reported any history of neurological, orthopaedic, cardiovascular or psychiatric disorders; if they reported falls in the past 12 months; or if they were taking tranquillizers or other medications that could affect the treadmill walking.

From linear position data to mechanical work

The COM and body segment mass were estimated using 2 tables: i) classical table, from the predictive equations [8] and ii) specific table, from a method that estimates the body segment inertial parameters for the elderly [21]. We calculated internal and external mechanical energies as proposed by Cavagna & Kaneko [6], in summary: The external energy (E_{ext}), as a function of time, is the total amount from adding gravitational potential (E_p) and kinetic energies (E_k):

$$E_{ext}(t) = mgh(t) + 0.5mV_{ap}^2(t) + 0.5mV_v^2(t) \tag{1}$$

where *m* is body mass (kg), *g* is gravitational acceleration (9.81 ms^{-2}), *h* is vertical linear position of the COM, and V_{ap} and V_v are the antero-posterior and vertical components of the translational speed of COM.

The internal energy (E_{int}) as a function of time is the total amount from the addition of translational and rotational kinetic energies of each segment:

$$E_{\text{int}}(t) = 0.5mV_{ap,r}^2(t) + 0.5mV_{v,r}^2(t) + 0.5m\omega^2(t)K^2(t) \tag{2}$$

where $V_{ap,r}$ and $V_{v,r}$ are the antero-posterior and vertical components of the translational speed of the centre of a segmental mass in relation to analogue velocities (antero-posterior and vertical) of COM.

The total positive mechanical work (W_{tot}) was determined from the sum of the positive increments from the E_{int} and E_{ext} curves.

In addition, we also compared W_{int} calculated using classic and specific anthropometric tables by calculating it according to the equation proposed by Minetti [16], which assumes as constant the inertial properties of the limbs and the mass partitioning between the limbs and the remainder of the body.

$$W_{\text{int},theor} = 0.1 sf\,\bar{\dot{s}}\left[1 + \left(\frac{d}{1-d}\right)^2\right] \tag{3}$$

where *sf* is stride frequency ($strides.s^{-1}$), $\bar{\dot{S}}$ is average horizontal speed ($m.s^{-1}$) and *d* is duty factor.

Metabolic parameters

The cost of transport (*C*) was analysed as the O_2 consumption during the trial, subtracted by O_2 consumption at rest per kg carried and metre travelled (VO2000 Aerosport Medgraphics).

The degree of interchange of the mechanical energies of the COM (pendulum-like mechanism) was quantified using *R*, as proposed by [5] and in summary as follows:

$$R = 100 \times \frac{W_V + W_F - W_{EXT}}{W_V + W_F} \tag{4}$$

The efficiency (*eff*) of positive work production by the muscles was calculated as the ratio between W_{tot} and the energy expended, *C*.

$$eff = 100 \times \frac{Wtot}{C}, \tag{5}$$

W_{tot} and *C* were expressed in joules per kilogram body weight and per unit distance ($J{\cdot}kg^{-1}{\cdot}m^{-1}$). The value of *eff* is shown as a percentage.

Statistics

Statistical analysis was carried out using SPSS software, version 15.0. The mean and standard error were determined for each subject and then for each speed.

Significance was accepted when $\alpha = 0.05$ and $p < 0.05$. Data were tested with the Shapiro-Wilk test and were normally distributed. Significant differences between subject characteristics were tested with ANOVA for repeated measures followed by post hoc analysis (LSD).

RESULTS

The results are presented in three sections: i) spatio-temporal parameters; ii) energetic parameters; and iii) metabolic parameters.

Spatio-temporal parameters

Statistically significant differences ($p < 0.001$) were found for stride length (*sl*) and stride frequency (*sf*) of elderly females (table 1). The *sl* and *sf* increased directly with greater speed. But duty factor (*d*) was not different between speeds.

Energetic parameters

The behaviour of W_{int}, W_{tot} and *R* were similar, showing an upward trend across all speeds, while W_{ext} behaved homogeneously up to speed 1.66 ms^{-1}.

There was a significant interaction between speed and different anthropometric tables employed. The effects of interaction tables * speed indicate that with increasing speed, tables differ among

TABLE 1. SPATIO-TEMPORAL PARAMETERS

	Walking speed ($m \cdot s^{-1}$)				
	0.55	0.83	1.11	1.38	1.66
sl (m)	0.7±0.12^a	0.9±0.09^b	1.07±0.04^c	1.27±0.04^d	1.38±0.07^e
sf (steps · s^{-1})	0.81±0.15^a	0.93±0.09^a	1.03±0.04^b	1.07±0.04^b	1.20±0.06^c
D	0.70±0.12	0.62±0.01	0.60±0.03	0.59±0.03	0.57±0.03

Note: Values are mean±standard deviation. Different letters indicate significant differences among the speeds (p<0.001).

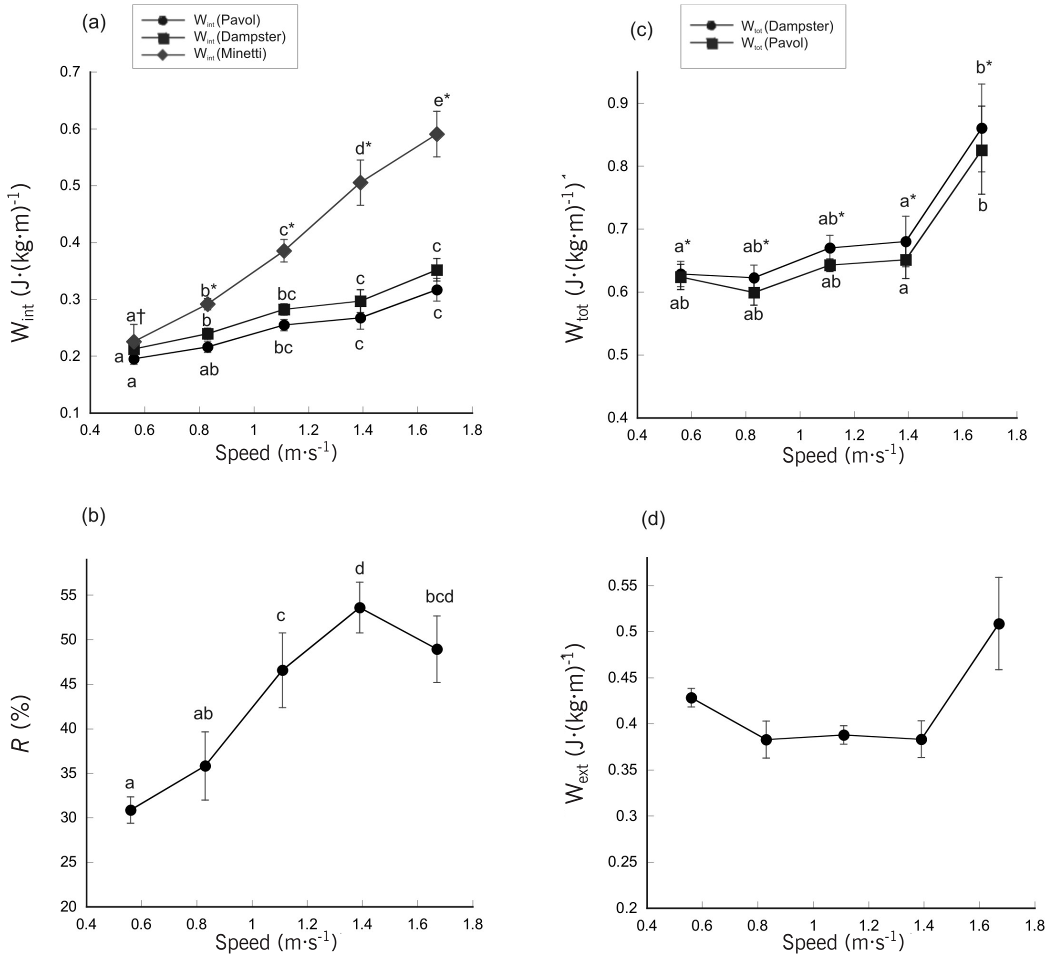

FIG. 1. INTERNAL MECHANICAL WORK (W_{int}) BASED ON ANTHROPOMETRIC TABLES AND MINETTI'S THEORETICAL EQUATION (a), RECOVERY DATA (R) (b), TOTAL MECHANICAL WORK (W_{tot}) BASED ON CLASSIC (DEMPSTER) AND SPECIFIC (PAVOL) ANTHROPOMETRIC TABLES (c) AND EXTERNAL MECHANICAL WORK (W_{ext}) (d) AT 5 SPEEDS WITHIN THE RANGE 0.55 – 1.66 $m \cdot s^{-1}$.

Note: *Indicates statistically significant differences between the anthropometric models of Pavol, Minetti and Dempster (p<0.001). † indicates significant differences between Pavol and Dempster. Different letters indicate significant differences among the speeds (p<0.001).

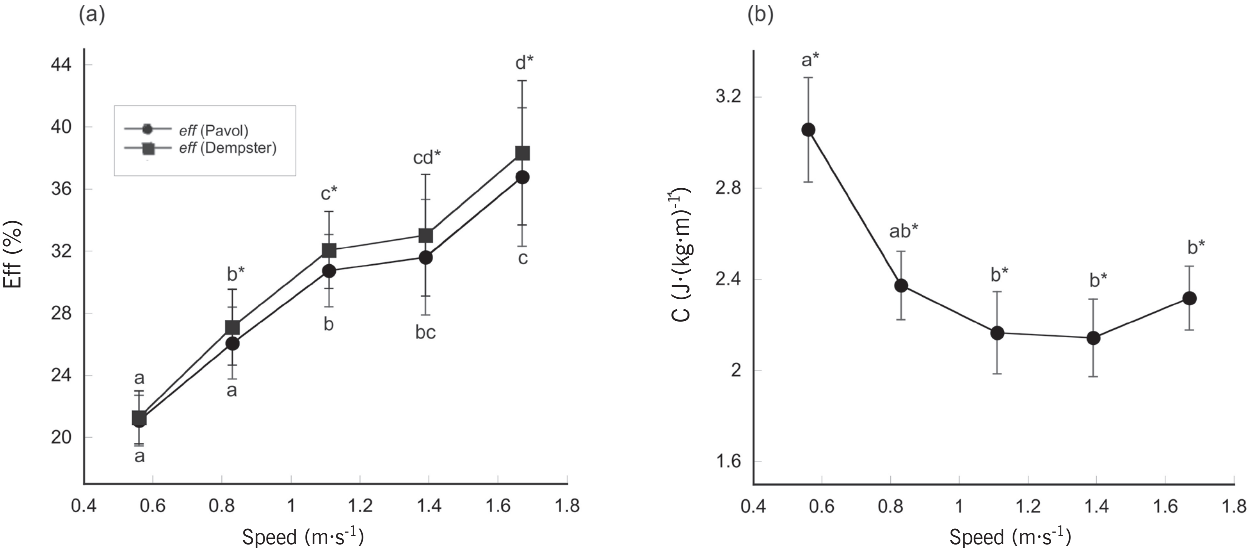

FIG. 2. EFFICIENCY (Eff) (a) AND COST OF TRANSPORT (C) (b) AT 5 SPEEDS IN THE RANGE 0.6 – 1.7 M.S-1. DIFFERENT LETTERS INDICATE SIGNIFICANT DIFFERENCES AMONG THE SPEEDS ($P<0.001$)

Note:* Indicates statistically significant differences between the anthropometric models of Pavol and Dempster ($p<0.001$)

themselves. Figure 1a shows that W_{int} was higher ($p<0.001$) than W_{int} as calculated according to both Pavol and Dempster data.

We also found statistically significant differences ($p < 0.001$) in W_{tot} based on anthropometric tables (Fig. 1c). Hence, anthropometric tables for specific populations may be required to provide accurate estimations of body segment parameters to be used in kinetic equations of motion.

As seen in figure 1d, W_{ext} was not statistically different with the increase in walking speed ($p <0.01$). A similar behaviour was found at speeds below 1.4 $m{\cdot}s^{-1}$.

The *R* data (Fig. 1b) showed significant differences ($p = 0.003$) among speeds. Recovery of mechanical energy attained a maximum close to 1.38 $m{\cdot}s^{-1}$ and the lowest value was obtained at the minimum speed tested (0.55 $m{\cdot}s^{-1}$).

Metabolic parameters

The cost of transport (*C*) also showed significant differences ($p <0.001$) especially between lower and higher speeds of walking (Fig 2b). As expected, minimum values were at intermediate speeds. Likewise, *eff* obtained by calculating W_{tot} from different anthropometric tables showed statistical differences between the speeds and among themselves (Fig 2a).

DISCUSSION

Our results show that the anthropometric changes due to aging led to a lower W_{int} and, as hypothesized, a decreased W_{tot} in the population studied. One possible explanation is that the increased stride frequency (*sf*) partially counterbalanced the change. Body segments' inertial parameters taken from different anthropometric tables eventually affected mechanical work calculations.

The inertial parameters obtained from anthropometric tables are widely used in the literature [8,21,27]. Nevertheless, these methods have several limitations and a main concern is the applicability of predictive equations to several different populations [9]. Beyond that, individual characteristics are not preserved (or cannot be taken into account) when the mass, COM and moments of inertia are derived from the total mass of the individual and segments' size. Accurate estimation reasonably requires to take into account age-related changes in anthropometry [24] and variations in body mass distribution dependent on gender and body type [11]. Thus, to calculate the mechanical work in locomotion (walking on the level, at different gradients, carrying a load or running) data from these anthropometric tables are necessary as input.

The optimization of interchange of the mechanical energies of the COM occurs when they are equal in amplitude and opposite in phase, as in an inverted pendulum [7] commonly quantified by *R*. In an ideal pendulum-like situation *R* would be 1, but when walking on Earth the maximum value reached is ≈0.70 at intermediate speeds (5.5 $km{\cdot}h^{-1}$). Our *R* data were similar to the results of Mian et al. [15], both for elderly and young men. Our data suggest that there is no substantial decrease in *R* in elderly women; thus, the major cost of transport in the elderly is not related to some detriment of the pendular mechanism.

One component of *R*, the W_{ext}, which represents the mechanical work performed to lift and accelerate the COM relative to the environment, showed an increase at the highest walking speeds, although it was not statistically significant (1.66 $m{\cdot}s^{-1}$ in Fig. 1d). So it suggests an altered COM displacement without the *R* index being affected. By contrast, W_{tot} (the sum of W_{ext} and W_{int}) could have been affected. Another study suggests a lack of a significant increase in W_{tot} in healthy elderly subjects [15]. Nevertheless, our W_{tot} data were smaller in elderly women and one possible explanation is that differences in W_{tot} were mainly determined by changes in W_{ext}.

On the other hand, W_{int}, which represents the mechanical work to accelerate the limbs relative to the COM, is directly related to the anthropometric tables (segment mass, COM of segments, radius of gyration of segments). In disagreement with other studies, our data of W_{int} were lower despite *sf* having increased. It is worth noting here that Minetti's W_{int} calculation method showed overestimation when comparing different (population specific *vs* classical) anthropometric tables. This difference may be explained by the independence of the anthropometric factor, i.e., to accept a constant value (=0.1) across different gaits, speeds and gradients (see equation 3 above). Another interpretation that should be considered is that *sf* is higher in old adults and this phenomenon could affect W_{int}. So it seems that a higher *sf* can be compensated by inertial parameters.

In relation to the metabolic parameters it is well described in the literature that elderly adults consume more metabolic energy during walking than young adults [12,15,25], but the reason is still unknown. However, it has already been established that a U-shaped relationship exists between *C* and speed [13] and this may explain why elderly and young adults consume the least metabolic energy to walk a given distance at intermediate speeds, whilst they consume more energy when they move at both faster and slower speeds [20].

In agreement with our data, Mian et al. [15] suggested that anthropometric changes in the elderly do not explain the change in whole body W_{tot} and *C*. Furthermore, Ortega & Farley [20] believe that biomechanical factors could affect *C* regardless of age. In our study *C* values were lower compared to Mian et al. [15], thus indicating a more economical pattern of gait. Despite the integrity of the pendulum-like interchange of mechanical energies, *C* remains higher in elderly subjects. Some authors have justified this elevated *C* in elderly adults as being due to other mechanical factors, such as stability during walking, muscular co-contraction and muscular efficiency [20]. Another factor that may contribute to high energetic demand during the aging process is osteoarthritis involving the hip or knee [28].

In the present study *eff* data presented higher values than other published studies evaluating walking in elderly adults (maximum of 36-38% at higher speeds [15,20]). In contrast, Cavagna & Kaneko [6] indicated that the maximum efficiency has to fall within the most economical speed range (intermediate speeds).

To better understand the discrepancy between our findings and data already available in the scientific literature we identified a series of potential explanations:

1) Our data came from a population-specific (elderly) table and being more accurate should bring more reliable results.

2) Some authors have pointed out W_{ext} as a contributor to higher *C* and associated with simultaneously positive (concentric) and negative (eccentric) muscle work in the stance phase, i.e., co-contractions [18,19]. This phenomenon may be partially explained by a new concept, recently proposed and called electromyographic (EMG) cost. This calculates the EMG activity as a function of the distance travelled (instead of the metabolic energy expenditure) and states that unstable gait patterns will generate greater muscular co-contraction and consequently low-efficiency locomotion [22]. Is known that healthy older adults experience changes in the dynamics of walking and are at risk of falls, i.e., they have a change of stability [14]. In Ortega & Farley's [20] study, elderly subjects presented *C* increasing with speed. Interestingly, age and participants' characteristics were similar to those of the subjects recruited in our study, while speeds were not.

3) The speeds chosen for our study were below those selected to test the elderly in the above cited studies [15,20] and may have caused a greater *C* at low walking speeds.

4) Elderly adults participating in the current study were healthy and active so more likely to be accustomed to exercise even if it is known that training status does not affect the optimal speed [2].

5) Our results might be affected by the so-called independence of size effect. The Froude number can be applied to locomotion to compare the speed of different sized individuals as previously suggested by Saibene & Minetti [23].

6) Our subjects were only females, who present different mass distribution in comparison to men. It seems that the similar gross and smaller standing metabolic rates of normal-weight women would presumably result in a greater net metabolic rate than normal-weight men during walking [3,10].

CONCLUSIONS

Further research should address the mechanics and energetics during walking in older adults. Our findings suggest that the inverted pendulum is maintained while metabolic parameters increase in the elderly. This costly compensation may be mainly related to maintenance of postural stability (co-contraction). Based on that, our suggestion for future studies is to apply the EMG cost to try to explain the greater energy expenditure in elderly.

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