Metabolic cost and co-contraction during walking at different speeds in young and old adults

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Title

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ABSTRACT

Background: The net metabolic cost of walking (*NCw*) and the co-activation of leg muscles are both higher in old adults (OG) than in young adults (YG). Nevertheless, the relation between the two remains unresolved, mainly due to the controversial co-activation measurement method used in previous studies. **Research question**: To compare ankle and knee co-contraction (CCI), calculated using an EMG-driven method, between the groups and to examine their relationship with *NCw*. **Methods**: Nine young (YG = 25.2 +/- 3.3 years old) and nine older (OG = 68.7 +/5.9 years old) adults walked on a treadmill at five speeds (YG: 1 ; 1.2 ; 1.4 ; 1.6 ; 1.8 m/s ; OG: 0.6 ; 0.8 ; 1 ;1.2 ; 1.4 m/s) while electromyography (sEMG) and oxygen consumption were measured. CCI were calculated around the ankle and knee for different parts of the gait cycle (entire gait cycle 0–100%, stance phase 0–60%, swing phase 60–100%). **Results**: *NCw* was significantly higher (25%, averaged over the walking speeds) in OG as were *Knee_CCI*, *Knee_CCI_swing* and *Knee_CCI_stance*. Multiple regression models in YG, OG and YG+OG highlighted *Ankle_CCI* as the main contributor in *NCw* (β = 0.08-0.188, $p<0.05$) with a positive relation between the two variables. **Significance**: The present findings provide a better understanding of the association between muscle co-contraction and metabolic cost in older adults. It may help scientists and clinicians to further develop strategies aimed at neuromuscular rehabilitation as a means of improving mobility and independence among older adults.

Keywords: ageing, co-contraction, energy cost, gait biomechanics, EMG-driven model

1 INTRODUCTION

Aging is associated with a decline in mobility and balance, which may lead to falls and a loss of independent function [1]**.** To increase postural stability, older adults are likely to develop compensatory strategies especially during locomotion tasks. Muscle co-activation, which is the concurrent activation of agonist and antagonist muscles that cross a joint, is one strategy to enhance joint stability [2]. Estimation of co-activation during gait with surface electromyographic analysis (*sEMG*) has demonstrated that older adults have higher muscle co-activation, particularly in antagonist muscles, than their younger counterparts [3–6]. However, the age-related increase in antagonist muscle co-activation would in turn require each agonist muscle to produce additional force, recruiting a greater portion of muscle mass and, thus, consume more metabolic energy, to offset the opposing force of the antagonist muscle. This process could be a contributor to the higher metabolic cost (*NCw*) in old adults compared to their younger counterparts [7,8].

Results of studies of the association between co-activation and *NCw* are contradictory and offer a limited understanding. For example, Ortega et al [9] showed a low correlation (r=0.23) between *NCw* and co-activation while other studies [8,10,11] have found higher correlations (r=0.5–0.69). Also, Marques et al. [11] highlighted that *NCw* is only correlated with co-activation at the knee joint, whereas Peterson et al. [8] emphasized a correlation at the ankle joint. Mian et al. [7] explained that their results were highly dependent on the method used to determine co-activation and found that their method, which normalizes EMG with a maximal voluntary contraction (MVC), could be inadequate when applied in older adults. Indeed, this population may have trouble consistently reaching MVC, leading to a miscalculation in co-activation. Moreover, Souissi et al. 2017 [12] show that co-activation methods tend to misestimate the simultaneous action of agonist and antagonist contractions, which can lead to an underestimation of energy cost and erroneous clinical conclusions during pathological gait analysis. New methods based on EMG-driven models have been developed and may provide a more reliable description of muscle action than co-activation methods. First, Co-Activation methods are based on one agonist and one antagonist muscles when the CCI includes several agonist and antagonist muscles. This can be particularly important at the ankle joint where three muscles represent the plantar flexor muscles group and one muscle for the dorsiflexor muscle group. In addition, the normalized EMG used to compute the Co-Activation leads to equivalent contribution of agonist and antagonist. It is important to note that the maximal muscle moment production which is fourfold stronger in the plantar flexors than in the dorsiflexor cannot be included when normalized EMG or normalized moment [13] are used. Finally, the EMG data does not include the non-neural factors that influence muscle moment [13] such as the contraction mode (eccentric vs concentric) that lead to different muscle force**.** These new methods could significantly help to determine the most prominent factor in the prediction of metabolic cost, which is still a challenging task. Indeed, previous studies, most using co-activation methods, have only demonstrated a correlation between co-activation and metabolic cost without explaining which joints (knee, ankle or hip) or other parameters contribute the most. The use of a more reliable method to estimate co-contraction could provide a further understanding of this relationship.

The aim of this study was, thus, to compare knee and ankle co-contraction index (CCI) levels between young and old adults and to identify their contributions to metabolic cost (*NCw*) with a new consistent method. The detailed hypotheses are (1) older adults exhibit higher *NCw* than young adults, (2) older adults have higher levels of co-contraction in the knee and ankle than young adults across a range of walking speeds, and (3) higher levels of co-contraction are associated with higher *NCw* in both older and young individuals.

2 Materials and methods

2.1 Participants

9 healthy subjects (25.2 +/- 3.3 years) were included in the Young Group (YG) and 9 older subjects (68.7 +/- 5.9 years) constituted the Old Participants' group (OG) (Table 1). Approval of the study protocol was obtained (*n° 2015-A01188-41*). Inclusion criteria were the ability to walk independently without walking aids, no use of antispastic medications and no orthopedic surgery in the last 6 months. Prior to enrollment in the study, all participants provided written informed consent.

[Please insert Table 1]

2.2 Experimental protocol

Participants were asked to walk on a Bertec instrumented treadmill (Bertec, Columbus, OH, USA) where kinematics, kinetics, surface electromyography (EMG) and oxygen consumption were recorded. Before acquisition, practice sessions were performed to familiarize participants with the procedure. Then, five trials were recorded at different speeds (OG: 0.6 m/s, 0.8m/s, 1 m/s, 1.2 m/s and 1.4 m/s and YG: 1 m/s, 1.2 m/s, 1.4 m/s, 1.6 m/s and 1.8 m/s) for 2 min each, in a random order. Only the last 30 s was analyzed. The five speeds were selected to provide a broad range of walking speed encompassing the preferred walking speeds of both older and young adults [14]**.** The procedure was based on the statement that older adults have lower PWS than their younger counterparts [15].

The whole-body segmental motion was recorded at 100 Hz using 9 optoelectronic cameras (Prime 13, Optitrack, NaturalPoint, USA) based on a full-body marker set [16]. Ground reaction forces were recorded from two force plates integrated in the treadmill. Skin surface electrodes (EMG) were placed longitudinally on different muscle bellies with respect to their muscle fiber direction following SENIAM conventions [17] and were recorded at 1000 Hz using a Bagnoli 16-channel bipolar surface electrode system (Delsys, Boston, USA). Finally, oxygen uptake (V02) was recorded by a COSMED K5 portable gas exchange system (COSMED, Rome, Italy) in breath-by-breath mode [18,19].

2.3 ESTIMATION OF AGONIST/ANTAGONIST MUSCLE MOMENT

Muscle forces at the ankle and knee joints were estimated using an EMG-driven approach with two degrees of freedom, which has been described in detail elsewhere [20]. OpenSim (version 3.1, SimTK, Stanford, USA) was used to create the anatomical model based on a full-body model [21] including eight muscle-tendon units: biceps femoris long-head (*BicFemLH*), semi-tendinosus (*SeTend*), gastrocnemius medialis (*GasMed*), soleus (*Sol*), rectus femoris (*RecFem*), vastus lateralis (*VasLat*), vastus medialis (*VasMed*) and tibialis anterior (*TibAnt*). Muscle activation patterns were derived from the raw EMG data after band-pass filtering (30–400 Hz), full-wave rectification, low-pass filtering (4th order, 6 Hz cut-off frequency), and normalization by the maximal value of the muscle concerned. The transformation from normalized EMG to muscle activation was obtained by including second-order dynamics, electromechanical delay and a nonlinear relationship between EMG and muscle activation [22].

EMG-driven model was calibrated on the left side in knee and ankle joint by minimizing the difference between knee and ankle joint moments. Moments at hip, knee and ankle joints were calculated by multiplying each individual muscle forces of the relative joint by the muscle-tendon moment arms and summed to obtain the total moment of the joint. Then the CCI was computed as:

$$CCI(t) = \frac{2*(common\ area\ Anta\ and\ Ago)}{area\ Anta + area\ Ago} * 100$$ [23]

The antagonist moment ($M(t)_{antagonist}$) was defined as the moment of lowest absolute value and the agonist moment ($M(t)_{agonist}$) as the highest absolute value. The moment of the muscle group was computed as the sum of all the muscle moments.

2.4 METABOLIC COST CALCULATION

VO2 and VCO2 (ml/min) were averaged over the final 30 s of each trial, when they were stabilized. As the respiratory exchange ratio (QR) was always below 1 during the walking trial, confirming that aerobic metabolism was the main metabolic pathway, an energetic equivalent of 20.1 J/mL 02 (*EEQ*) [7] was used to convert to Joules. The resting metabolic rate was subtracted from the gross metabolic rate (*GMR*) and divided by walking speed to determine net metabolic cost (*NCw*).

The equation is as follows: $NCw = \frac{EEQ*GMR - RMR}{speed}$ [24] [25].

where EEQ is the Energetic Equivalent of the oxygen consumed (VO2(J/min)), GMR is the gross metabolic rate (energy expenditure measured during the trial in ml/kg/min) and RMR the resting metabolic rate (J/kg/min).

2.5 DATA ANALYSIS AND STATISTICAL CALCULATIONS

Mean values and standard deviations (SD) were calculated for each variable. The distribution of the data was checked using the Shapiro-Wilk normality test. A two-way (2*3, group * speed) ANOVA with repeated measures was used to determine the effects of age and walking speed on metabolic cost and CCI. Significant ANOVA results were followed by post hoc comparisons using Tukey's HSD test except for NCw that failed the normality test where a Dunn post-hoc was used. A Tukey p-correction was applied for multiple association. Concerning *NCw*, a one-tailed Welch test was used to test the difference between YG and OG at each speed. Cohen's d coefficient was used to measure the effect size. Pearson's *r* correlation coefficient was used to test the bivariate association between the dependent variable (i.e., *NCw*) and the covariate factors (CCI) for the entire sample (*n*= 18) at different speeds (1m/s ; 1.2 m/s ; 1.4 m/s) and in YG and OG separately. Then, a multiple regression model was built with backward selection to determine the overall fit of the model and the relative contribution of each of the predictors (CCI) to the total variance explained (*NCw*). The criterion for statistical significance was set at $p < 0.05$. Statistical analysis was performed using JASP (JASP Team (2020). JASP (Version 0.14.1)).

3 RESULTS

Older adults exhibit significantly higher *NCw (M= 3.5 J/kg/m ; SD=1)* than young adults *(M=2.8 J/kg/m ; SD=0.4)* with an average across speeds of 25% (Table 2, Figure 1) confirming the first hypothesis. No significant difference in *NCw* between speed were found ($F_{speed} = 0.409, p = 0.368, d = 0.014$) (Table 2) but it reflects a curvilinear response (Figure 1).

Concerning CCI at knee and ankle joint, no significant higher level of CCI at ankle joint were found in OG compared to YG (Table 2, Figure 2). However, at the knee joint (*Knee_CCI*, *Knee_CCI_stance* and *Knee_CCI_swing),*significant higher values were found for OG (Table 2, Figure 2). The walking speed where the difference in CCI between groups is significative is for 1m/s (Figure 2).

[Please insert Table 2]

[Please insert Figure 1]

[Please insert Figure 2]

In YG+OG, all the CCI except for *Ankle_CCI_swing* and *Knee_CCI_swing* were moderately correlated with *NCw* [r= 0.251–0.304] (Table 3). Higher level of CCI are associated with higher NCw only in OG (r=0.475, p=0.012 ; r=0.493, p=0.009 respectively for Ankle_CCI and Ankle_CCI_stance) while in YG no significant correlation between *NCw* and CCI variables (Table 3).

A multiple linear regression model was built with the 6 CCI variables to quantify their contribution to *NCw* in YG+OG and YG and OG separately. They are preliminary results regarding the number of parameters (6) and the number of participants (9 when divided into groups). Regarding YG + OG, the model was significant and explained 29.6% of the variability of *NCw* (F (5.48) = 4.027, p=0.004). *Ankle_CCI* and *Ankle_CCI_stance* were found as the main contributors in *NCw* with a slope coefficient of 0.188 and -0.139, respectively ($p<0.05$) (Table 4). The positive slope coefficient for *Ankle_CCI* means that *NCw* increase with *Ankle_CCI*. Conversely, an increase in *NCw* is associated with a decrease in *Ankle_CCI_stance*. The model in YG is also significant and very similar to YG+OG with 62.3% of the variability of *NCw* explained (F (5.21) = 6.946, $p<0.001$). The main contributors are *Ankle_CCI* and *Ankle_CCI_stance* with a slope coefficient of 0.138 and -0.126 respectively (p=0.003) (Table 4). Finally, the model for OG is significant with only 2 variables that explained 33.2% of the variability of *NCw* (F (2.24) = 5.972, p = 0.008).

[Please insert Table 3]

[Please insert Table 4]

4 DISCUSSION

The aim of this study was to compare knee and ankle CCI levels between young and old adults and to identify their contributions to metabolic cost using a new consistent method. *NCw* was significantly higher in OG for each speed, confirming the hypothesis. The hypothesis was also confirmed for *Knee_CCI*, *Knee_CCI_swing* and *Knee_CCI_*stance where it is significantly higher in OG. While we hypothesized that both ankle and knee CCI could play a role in *NCw*, results highlighted that *NCw* was largely explained by *ankle_CCI*, with a positive relationship between them.

The *NCw* average was 2.77± 0.43 J/kg/m in YG and 3.5 ± 1 J/kg/m in OG. A meta-analysis established that the average *NCw* in YG was 2.4 ± 0.4 J/kg/m while it was 2.8 ± 0.5 J/kg/m in OG [25]. It should be noted that walking on a treadmill leads to a higher metabolic cost than walking overground [26] which could contribute to our slightly higher values. The preferred walking speed was around 1 m/s for OG and 1.2 m/s for YG which is also in the range of habitual walking speeds in community-dwelling older adults (1.0–1.3 m/s) as for young adults (1.3–1.5 m/s) [27]. Our results confirmed the significantly higher *NCw* in older adults compared to their younger counterparts with an average difference across all speeds of 25%. This is similar to the 23% increase seen in a similar study (M= 71 years) [8] and slightly lower than the 31% increase in *NCw* seen in a previous study (M= 74 years) [7]. Thus, the decline of aerobic capacity with aging and the concurrent increase in *NCw* likely contribute to a reduced ability to maintain the activities of daily living, and potentially increase fatigue and fall risk. It is therefore important to determine the underlying contributing factors to the higher *NCw* in OG.

Significantly higher CCI values were found in OG for all the gait phases (entire gait cycle, swing, and stance phase) at the knee joint but not at the ankle joint. This result was the same as previous work [8,28], where the authors found that co-activation for *biceps femoris/ vastus lateralis* (knee joint) was 35% to 60% higher in the elderly, and no significant difference between groups for *tibialis anterior/Gastrocnemius lateralis* (ankle joint). Also, for the knee joint, the greatest difference between groups was during the stance phase. During initial stance, the quadriceps muscles display a burst of activation as they stabilize the pelvis and the thigh to reduce the displacement of the center of mass in the sagittal plane. During terminal stance, when the anterior acceleration of the center of mass occurs, these muscles exhibit a second burst of activation, again to stabilize the lumbopelvic and thigh regions [29]. We theorize that older adults increased co-contraction of knee muscles during stance to a greater extent than young adults to maintain stabilization of the ankle and hip, as a consequence of lower knee extensor strength. In general, increased muscle co-activation during locomotion is most commonly described as a compensatory mechanism to increase joint stiffness and thereby enhance stability [6]. Previous studies found similar results [6,30] [31] concluding that old adults may not be able to control the amount of muscle co-contraction according to performance speed. Instead of increasing muscle co-contraction, they may stiffen their joints and lessen the movement in a more challenging task [31]. Perhaps, in old adults, the level of co-activation reached a ceiling and could not further increase. Such insensitivity could limit the functional level of joint stiffness at ankle and knee joint during gait.

Although our results were in agreement with prior research that showed older adults walk with a higher metabolic cost and use greater co-activation, CCI at ankle and knee joint and *NCw* were only moderately correlated ($r = 0.296$, $r = 0.273$, $p<0.05$ respectively) with the strongest positive correlation for *Knee_CCI_stance* ($r = 0.304$, $p= 0.025$) in YG+OG similarly to Peterson et al. ($r=0.39–0.46$) [8]. When collapsed across age groups, no bivariate correlation was found in YG while *Ankle_CCI* and *Ankle_CCI_stance* were correlated with *NCw* in OG ($r=0.475$, $p=0.012$; $r=0.493$, $p=0.009$, respectively). Previous studies also found that ankle joint co-activation was positively correlated with *NCw* in old adults ($r=0.23–0.52$, $p < 0.05$), while no correlation appeared in young adults [8,11]. These results do not offer a window on the relative contributions and degree of involvement of the ankle and knee joint co-contraction in *NCw*. However, multiple regression revealed *Ankle_CCI* as the most contributive value on *NCw* ($\beta = 0.08\text{-}0.188$, $p<0.008$), whether the analysis was made on separate or mixed groups. The soleus muscles has one of the largest physiological cross-sectional areas in the lower limb and, thus, may contribute greatly to changes in metabolic cost. Finally, from a purely mechanical point of view, muscle co-contraction is an inefficient use of muscle forces, does not contribute to the useful work output of muscles, and requires higher metabolic cost.

Two reasons why the results of the various studies may differ is that different methods were used to calculate co-activation and different antagonist muscle pairs were evaluated. Indeed, while CCI were obtained from the estimation of muscle moments using the EMG-driven model in Lo et al. [23], other studies calculate the co-activation index derived from EMG data [4,8]. Methods derived from EMG data have been found to underestimate the simultaneous action of agonist and antagonist contraction while CCI from EMG-driven models may provide a more reliable description of muscle action [12]. Also, different combinations of muscles can be considered to compose the joint. While Marques et al. [11] defined the ankle joint with rectus femoris/biceps femoris, Lee et al. [4] preferred to use the vastus medialis oblique and biceps femoris. As each muscle plays different roles during walking, if the combination of muscle pairs is not exactly similar, slight differences could be found in CCI calculation and therefore comparisons between studies should be made carefully.

One limitation of the present study is its sole focus on the role of muscle activation in *NCw* instead of using a multifactorial approach. There are several other age-related adaptations, including differences in gait kinematics or proximal redistribution of lower extremity joint torque, and *NCw* may be affected to some degree by all of these factors. To fully understand why *NCw* is elevated in older adults, it is important to explore in more detail each factor and its effect on *NCw*. Silder et al [32] used a multifactorial approach and their linear regression model explained 96% of the variance in *NCw*, while our best model only explained 62.3%. However, caution should be made to the multiple linear results as the analysis was overfitted. Indeed, the rule of thumb is usually 2 subject per variable (SPV) [33] and when groups are divided into 2 parts (9 participants per group) with 6 variables as input of the model, the SPV is not greater than 1.5. When the number of SPV is low, Austin et al [33] suggest to take the adjusted R2 over the conventional R2 for quantifying the proportion of variance explained by the model which was applied in this study. However, as a result of overfitting, these preliminary results may be poorly reproducible in other samples [33]. Also, it is unclear how the use of a treadmill has affected gait and whether our results would be reproducible in gait tasks other than on a treadmill. To minimize its potential impact, we used 6 minutes of familiarization, which, according to Meyer et al [34], was sufficient to assure the reliability of lumbar, hip and knee kinematics. It might also be suggested that older adults feel more apprehensive on a treadmill, particularly when asked to walk at speeds faster than they are habitually accustomed to, resulting in an artificially elevated *NCw*. It is not clear the degree to which this influenced the relation between co-contraction and metabolic cost.

5 PERSPECTIVES

The present study has demonstrated a higher metabolic cost in old compared to their younger counterparts. This is closely related to the age-related decrease efficiency of the muscle itself, which would then require more metabolic energy to perform a given amount of mechanical work. The findings of the study also highlight that older adults had higher knee co-contraction compared to young adults while ankle co-contraction was found to be the most important contributor to the higher metabolic cost in old adults. This is very interesting as ankle co-contraction has been shown to be associated with an increased fall risk in older adults [35]. By providing a better understanding of how muscle co-contraction contributes to the greater metabolic cost of walking in older adults, the results of this study may help scientists and clinicians to further develop strategies aimed at neuromuscular rehabilitation and improving muscle efficiency as a means of improving mobility and independence among older adults.

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Figures

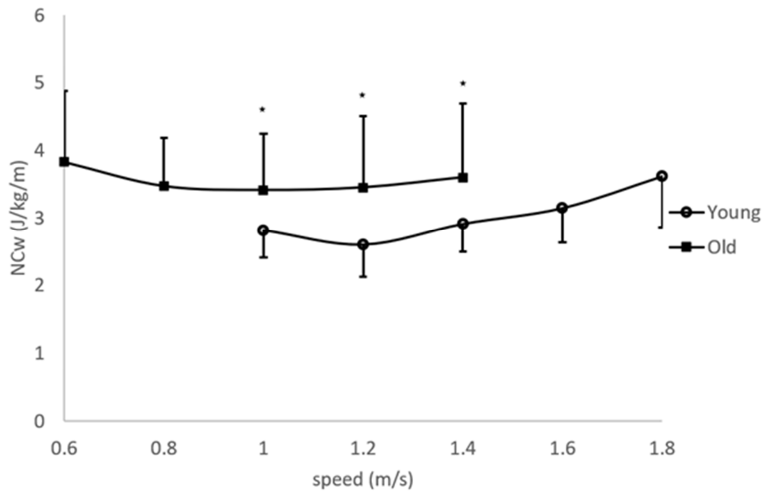

Figure 1*: The NCw-speed relationship, comparison of the metabolic cost between old (solid squares) and young (open circles) with standard deviation. * p<0.05 for difference between OG and YG.*

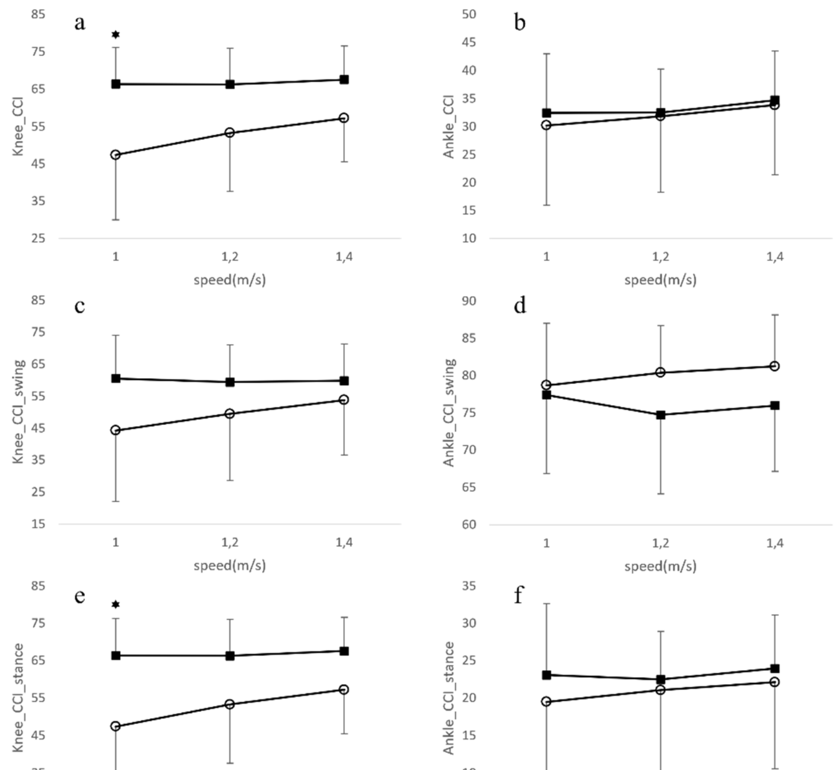

Figure 2*: Knee and Ankle CCI during the entire gait cycle (a-b), in swing (c-d) and stance phase (e-f) at different speeds (1 m/s, 1.2 m/s and 1.4 m/s) in YG (open circles) and OG (solid squares). *p<0.05 for significant ANOVA results (group*speed).*

Tables

***Table 1**: Participant's characteristic*

	Young (YG) (n=9)	Old (OG) (n=9)	P values
Age (years)	25.2 +/- 3.3	68.7 +/- 5.9	<0.001*
Height (cm)	172.3 +/- 5.2	162.8 +/- 8.5	0.011*
Weight (kgs)	66.3 +/- 6.2	66.4 +/- 12.7	0.982
BMI (kg/m2)	22.3+/- 1.3	25 +/- 4	0.072
Resting metabolic rate (J/kg/min)	122.4 +/- 21.2	91.1 +/- 17.2	0.003*

*Values are mean +/- SD. The p-values were from two-tailed unpaired t-tests between the groups. For all tests, the alternative hypothesis specifies that OG is greater than YG. * means significant difference (p<0.05)*

*Table 2: Mean +/- SD for each parameters (average across speed) in YG and OG and ANOVA outputs (group, speed, group*speed). * p<0.05, ** p<0.001. a : non parametric*

Parameters	YG	**OG**	**Model**	F	**p**	**Effect size d**
Knee_CCI (%)	52.1 +/- 13	65.5 +/- 9.3	group	18.33	<0.001 **	0.267
			speed	0.69	0.506	0.02
			Group*speed	0.526	0.594	0.015
Ankle_CCI (%)	32 +/- 13	33.2 +/- 8.8	group	0.16	0.691	0.003
			speed	0.315	0.731	0.013
			Group*speed	0.026	0.974	0.001
Knee_CCI_swing (%)	49.3 +/- 19.9	59.8 +/- 11.9	group	5.545	0.023 *	0.101
			speed	0.312	0.734	0.011
			Group*speed	0.429	0.654	0.016
Ankle_CCI_swing (%)	80.1 +/- 7	76.1 +/- 9.7	group	2.897	0.095	0.056
			speed	0.067	0.935	0.003
			Group*speed	0.344	0.711	0.013
Knee_CCI_stance (%)	52.6 +/- 15.1	66.6 +/- 9.3	group	16.854	<0.001 **	0.249
			speed	0.864	0.428	0.026
			Group*speed	0.580	0.58	0.016

Ankle_CCI_stance (%)	20.9 +/- 11.9	22.5 +/- 6.7	group	0.66	0.421	0.013
			speed	0.139	0.87	0.006
			Group*speed	0.057	0.945	0.002
NCw^a **(J/kg/m)**	2.8 +/- 0.4	3.5 +/- 1	group	11.457	0.008 *	0.189
			speed	0.409	0.368	0.014
			Group*speed	0.124	0.884	0.004

Table 3*: Correlation matrix and bivariate association between NCw and CCI parameters in YG, OG and YG+OG. Significant results are in bold*

	YG		**OG**		**YG+OG**	
	r	**p**	**r**	**p**	**r**	**p**
NCw – Knee CCI	-0.176	0.379	0.331	0.091	**0.296**	**0.03***
NCw – Ankle CCI	0.022	0.914	**0.475**	**0.012***	**0.273**	**0.046***
NCw–Knee_CCI_SWING	0.201	0.314	0.205	0.306	0.251	0.067
NCw–Ankle_CCI_SWING	-0.136	0.498	-0.018	0.929	-0.082	0.556
NCw-Knee_CCI_STANCE	-0.104	0.607	0.354	0.07	**0.304**	**0.025***
NCw-Ankle_CCI_STANCE	-0.052	0.796	**0.493**	**0.009***	**0.271**	**0.048***
Note: * p<0.05, r = Pearson's correlation						

Table 4: *Multiple linear regression model (backward) in YG, OG and YG+OG. The most predictive parameters are in bold.*

Model		β	t	p
YG	Knee_CCI	-0.015	-1.763	0.092
	Ankle_CCI	**0.138**	3.360	0.003*
	Ankle_CCI_stance	**-0.126**	-3.421	0.003*
	Knee_CCI_swing	0.014	2.073	0.051
	Ankle_CCI_swing	-0.064	-3.320	0.003*
OG	Ankle_CCI	**0.08**	3.454	0.002*
	Ankle_CCI_swing	-0.039	-1.96	0.062
YG + OG	Ankle_CCI	**0.188**	2.761	0.008*
	Knee_CCI_stance	0.031	2.137	0.038*
	Ankle_CCI_stance	**-0.139**	-2.117	0.04*
	Knee_CCI_swing	-0.030	-2.111	0.04*
	Ankle_CCI_swing	-0.069	-3.546	<0.001*
Note : β = magnitude of the slope of the relationship ; p = probability of whether or not β is significantly different from 0 (* indicates values that were statistically significant)				