A multiple-task gait analysis approach: Kinematic, kinetic and EMG reference data for healthy young and adult subjects

Article *in* Gait & posture · January 2011

DOI: 10.1016/j.gaitpost.2010.08.009 · Source: PubMed

4 authors, including:

Gabriele Bovi
Fondazione Don Carlo Gnocchi

Maurizio Ferrarin
Fondazione Don Carlo Gnocchi

Gait & Posture 33 (2011) 6–13

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

A multiple-task gait analysis approach: Kinematic, kinetic and EMG reference data for healthy young and adult subjects

Gabriele Bovi, Marco Rabuffetti, Paolo Mazzoleni, Maurizio Ferrarin *

Polo Tecnologico, IRCCS S. Maria Nascente, Fondazione Don C. Gnocchi, Italy

ARTICLE INFO

Article history:
Received 8 April 2010
Received in revised form 5 August 2010
Accepted 12 August 2010

Keywords:
Gait analysis
Toe-walking
Heel-walking
Step negotiation
Clinical decision making

ABSTRACT

Standard clinical gait analysis protocols usually limit to test self-selected speed gait: this approach is generally valid and permits time and cost saving. Yet, the literature evidences suggest that some pathologies (especially at onset or subclinical level) may not primarily affect plain gait, but more demanding locomotor tasks. In the present study we therefore propose a multiple-task gait analysis protocol including: self-selected, increased and decreased speed gait; walking on toes; walking on heels; step ascending and step descending, and apply it to 40 healthy subjects (20 aged 6–17, 20 aged 22–72) thus building extensive reference data set. Published studies already report normative data for some of these tasks, but inhomogeneously (due to different collecting methods and biomechanical models, population characteristics, nature of data). We verify a good correlation between our results and those presented by Schwartz et al. (2008) [12] in their study providing extensive data on the effect of walking speed on the gait of healthy children. In discussing the results, the rationale and effectiveness of each task is confirmed, and we supply an electronic addendum with comprehensive kinematic, kinetic and electromyographic normative data for the considered population, along with a set of reference parameters and related statistical analysis, as a premise for further applications on pathological subjects.

1. Introduction

Instrumented gait analysis has become a valid tool for clinicians in the last decade, even though some factors – like data variability, data interpretation, time needed and expenses – limit its potential application [1]. Although there is evidence that even pathologies at subclinical phase may be detectable by a standard gait analysis test (considering only plain gait at self-selected speed) [2,3], such an approach may be unsatisfactory in a number of cases: some early-stage diseases may not primarily affect plain gait, but more challenging or specific tasks like steering or initiating gait [4,5], toe-walking, and heel-walking [6]. For this reason, we think that adding properly other selected locomotor tasks to standard gait analysis could be helpful for many borderline or subclinical cases.

Tasks of interest are toe-walking and heel-walking, as they both challenge balance, and imply a stronger distal activity of respectively plantaflexor and dorsiflexor muscles [7–9]. Also noteworthy is stair negotiation, which implies larger amounts of energy to be produced (ascending) or dissipated (descending) proximally, and increased angular excursion and joint moment at hip and knee level [10,11]. Moreover, it is well documented that walking speed has noticeable effects on kinematics, kinetics and muscle recruitment [12,13] and gait at increased speed is able to highlight abnormal locomotor patterns [14].

Many published papers separately address such locomotor tasks: walking speed influence on gait features has been studied by several authors [12,13,15], and correspondingly stair negotiation, both in its biomechanical and clinical aspects [10,11,16,17], and toe-walking [7–9,18–21]; less attention has been reserved to heel-walking [22]. As Schwartz et al. observed [12] about studies on the effects of walking speed on gait, there is a common limitation in currently published data: they come from populations inhomogeneous in age, gender, number, health condition, origin; are collected with different methods and biomechanical models; are sometimes limited to the particular study design.

On this basis, it seems justified our present effort to provide homogeneous data from the same populations of young and adult subjects, inclusive of kinematics, kinetics and EMG, of a multiple-task gait analysis protocol which could serve as a general framework, to be customized case-wise or pathology-wise.

The tasks we include are: walking at self-selected, diminished and increased speeds; walking on toes and on heels; ascending and descending a double step. Extensive reference data on different speed walking have already been provided, for children, by Schwartz et al. [12]: the marker set and protocol used in that study (Vicon Plug-in Gait, VPiG) is different from ours (LAMB), so we provide a

* Corresponding author. Tel.: +39 02 40308305; fax: +39 02 4048919.
E-mail address: mferrarin@dongnocchi.it (M. Ferrarin).

0966-6362/$ – see front matter © 2010 Elsevier B.V. All rights reserved.
doi:10.1016/j.gaitpost.2010.08.009

comparison between the two data set to prove good data correspondence, similarly to Ferrari et al. [23].

As a future intention, we are applying the present multiple-task protocol to populations of pathological subjects.

2. Methods

2.1. Subjects

Forty healthy subjects participated to the study, 20 included in the *adult* group (aged from 22 to 72 years, mean 43.1 ± 15.4; body mass 68.5 ± 15.8 kg; height 1.71 ± 0.10 m; 9 males, 11 females) and 20 in the *young* group (aged from 6 to 17 years, mean 10.8 ± 3.2; body mass 41.4 ± 15.5 kg; height 1.47 ± 0.20 m; 9 males, 11 females). We explained each subject for the study purpose and asked to sign an informed consent. The study was approved by our institution's Ethical Committee.

2.2. Instrumentation and protocol

3D kinematics was measured using a 9-cameras SMART-E motion capture system (BTS, Milano, Italy), acquiring at 60 Hz. The total-body LAMB marker set [24] was adopted, which included 29 retroreflective markers (12 mm diameter) positioned on head, upper limbs, trunk, pelvis and lower limbs. Two force plates (Kistler, Winterthur, Switzerland), at 960 Hz sampling frequency, provided ground reaction forces (GRFs). Surface electromyography signals (EMG) were registered with an 8-channel wireless electromyograph, ZeroWire (Aurion, Milano, Italy), using 10 mm diameter adhesive electrodes. The muscles selected (unilaterally) were: tibialis anterior (TA), soleus (SOL), gastrocnemius medialis (GAM), peroneus longus (PL), rectus femoris (RF), vastus medialis (VM), biceps femoris (BF) and gluteus maximus (GM).

Five locomotor tasks were included in the protocol: walking at different speeds, toe-walking (T), heel-walking (H), step ascending (U) and step descending (D).

For the walking task, we initially asked the subjects to walk five trials at their natural speed (N). Hence we asked to perform the following ten trials progressively increasing (first 5 trials) or decreasing (latter 5 trials) their speed. We gave no precise indications about gait speed in order not to induce gait alterations.

Step ascending task consisted in climbing a two-step stair (standard step: height, 180 mm, step depth 300 mm), starting from ground level and stopping on the second (640 mm × 500 mm wide) step. Correspondingly, in step descending the subjects started standing on the second step, and stopped on the ground before the first step. Only the first step (a rigid custom-made wooden structure positioned on one force plate and whose weight was implicitly removed by the platform reset procedures preceding the tasks) provided GRF data: we instructed the subjects to step on it with the leg carrying EMG equipment. Since in the described set-up the stair has only two steps, the tasks are analogous but not exactly corresponding to proper stair negotiation, as at second foot-strike the foot lands on the same level as the controlateral limb, and not on the following one as occurs in ascending/descending a longer stair.

The subjects performed toe-walking, heel-walking and step negotiation at their self-selected speed.

We asked the subjects to repeat each task five times.

2.3. Data analysis

We classified walking trials *a-posteriori* as follows: at first we defined class N including the trials in which we instructed the subjects to walk at their natural speed; then we divided all the walking trials (including those of class N) into four classes based on walking speed normalized to subject's height (v/h): very slow (XS) for $v/h < 0.6$; slow (S) for $0.6 \leq v/h < 0.8$; medium (M) for $0.8 \leq v/h < 1$; fast (L) for $v/h \geq 1$.

For comparison between our data and the corresponding ones of Schwartz et al. [12], we adopted their inclusion criterion to create a group named *free speed* (FS), selecting those trials whose normalized gait speed, rendered dimensionless following the method proposed by Hof [25], ranged from 0.363 to 0.500. We calculated Pearson's correlation coefficient between mean curves of the two FS groups (from present work and from Schwartz et al. [12]), as proposed by Ferrari et al. [23].

3D kinematics, dynamics and EMG were considered if complete data were available over at least one stride (indifferently right or left), otherwise the trial was discarded. Trials, which presented evident artifacts due to technical contingencies (missing detection of some markers or improper foot-strike on plates), were excluded.

We normalized, dynamic data (ground reaction forces, joint moments and powers) to body mass (BM), and spatio-temporal parameters gait speed, stride length and step length to subject's height (BH) for age group comparisons.

To post-process EMG signals we performed rectification and low-pass filtering (Butterworth 5th order, 3 Hz cutoff frequency). We operated intra-subject normalization similarly to *peak dynamic method* [26]: for each subject's channel independently, we normalized to the maximum of the root mean square values of 2% intervals across all the subject's trials. This approach allows inter-subject and inter-task comparisons.

3. Results

Due to the vast amount of data we collected, in the following paragraphs we will limit to account for global trends without focusing on any specific parameter, except the spatio-temporal ones. We provide complete data in the electronic addendum.

The number of valid strides for each task was (adult|young): $N_N = 120|108$, $N_{XS} = 125|34$, $N_S = 110|72$, $N_M = 60|71$, $N_L = 54|100$, $N_T = 125|82$, $N_H = 83|51$, $N_U = 71|75$, $N_D = 69|67$.

3.1. Data-set validation

Present study population of young subjects closely matched Schwartz et al. one [12], with mean (SD) age respectively 10.8(3.2) and 10.5(3.5) years. We found good correlation (Pearson's ρ coefficients) between our data, collected with LAMB protocol, and the ones of Schwartz et al. [12], collected with Vicon Plug-in Gait (VPiG) protocol, for walking trials in the speed range (Hof normalization method) from 0.363 to 0.500. Joint moments ρ ranged from 0.810 to 0.989, kinematics and powers ρ respectively from 0.743 to 0.992 and from 0.693 to 0.916 (all significant at $p < 0.001$). Also EMG presented significant high correlation (0.826–0.915).

3.2. Spatio-temporal parameters

Table 1 reports spatio-temporal parameters for different tasks, comparing adult and young groups. We evaluated differences between tasks using one-way ANOVA and Tukey HSD post hoc test, and between age groups using t-test. Significance was for $p < 0.05$.

When considering all speed classes, stride length, step length and cadence showed a positive trend associated with speed increase; conversely, stance time and double support time showed a negative trend, both for young and adults (Fig. 1).

For natural speed class all parameters except step length showed significant differences between age groups: in particular young subjects had higher gait speed, cadence and stride length, but lower stance time and first double support time.

Comparing toe-walking to N, stride length and gait speed were significantly reduced for both adults and young, and stance time was increased for young. Young presented higher gait speed, cadence and first double support time than adults.

For heel-walking, gait speed was significantly lower than for N as a consequence of shorter stride length, even though cadence was increased. Stance time was reduced, and similarly first double support time. Adults presented significantly lower gait speed than young, as a consequence of shorter normalized stride length and lower cadence.

For step ascent and descent we noticed significantly reduced gait speed and cadence if compared to N; stance phase was prolonged in both cases. First double support time was significantly longer for U and slightly for D with respect to N. Gait speed and cadence were significantly higher xfor young than for adults, while stride time was reduced, both for U and D.

3.3. Kinematics and kinetics

When considering kinematics and kinetics for different gait speed compared to natural speed gait, evidences provided by other authors are confirmed (e.g. Schwartz et al. [12]); we provide a figure in the electronic addendum.

In Fig. 2 we present kinematics and kinetics for T and H compared to N, for adults (A) and young (Y), and in Fig. 3 for U and D compared to N. Curve patterns were similar for both age groups, even though many parameters differed between groups (see electronic addendum).

Table 1
Mean (SD) of spatio-temporal parameters categorized by task and age group. Italicized values indicate $p < 0.05$ significant differences between adult and young groups (single variable independent *t*-tests). We omit stride length and step length for U and D, as they are constrained by the step dimensions. Task coding: natural speed walking (N), toe-walking (T), heel-walking (H), step ascending (U), step descending (D).

Parameters	Age group	Tasks									
		N		T		H		U		D	
Gait speed/BH [%BH s^{-1}]	Adult	*71.4*	*(10.2)*	*63.6*	*(10.9)**	*42.2*	*(8.7)**	*27.6*	*(2.9)**	*27.3*	*(4.6)**
	Young	*87.7*	*(15.8)*	*71.1*	*(17.0)**	*53.6*	*(15.8)**	*36.1*	*(7.6)**	*39.5*	*(13.2)**
Cadence [steps min^{-1}]	Adult	*110*	*(9)*	*111*	*(13)*	*120*	*(16)**	*72*	*(6)**	*88*	*(10)**
	Young	*124*	*(14)*	*120*	*(20)*	*136*	*(25)**	*81*	*(10)**	*99*	*(19)**
Stride length/BH [%]	Adult	*77.8*	*(6.4)*	68.6	(8.6)*	*43.3*	*(9.8)**		–		–
	Young	*84.9*	*(7.5)*	70.3	(10.1)*	*46.9*	*(9.0)**		–		–
Step length [% stride length]	Adult	50.0	(1.3)	50.0	(2.1)	50.2	(4.2)		–		–
	Young	50.1	(1.3)	49.6	(3.1)	51.3	(4.3)		–		–
Stance time [% of gait cycle]	Adult	*63*	*(2)*	62	(2)	59	(4)*	*69*	*(4)**	*70*	*(3)**
	Young	*61*	*(2)*	62	(3)*	58	(3)*	*66*	*(4)**	*68*	*(4)**
First double support time [% of gait cycle]	Adult	*13*	*(2)*	*12*	*(2)*	10	(4)*	*17*	*(3)**	15	(3)*
	Young	*11*	*(2)*	*13*	*(2)**	9	(3)*	*16*	*(3)**	15	(3)*

* $p < 0.05$ significant (ANOVA and Tukey HSD post hoc test) difference from natural speed walking.

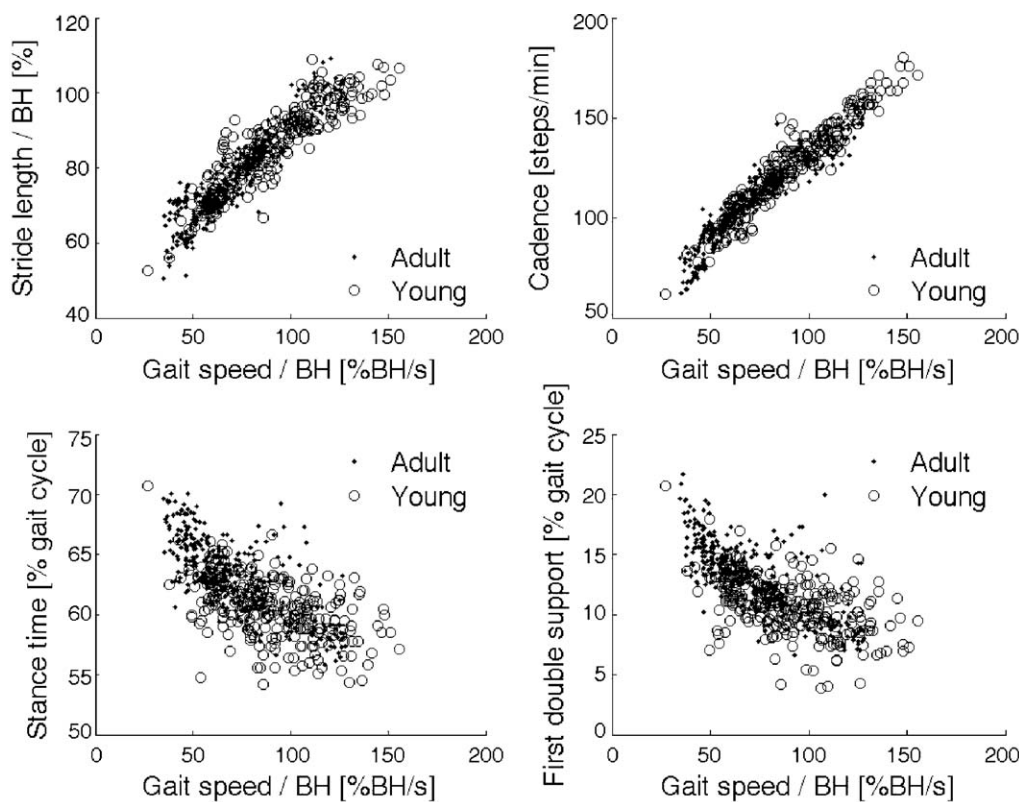

Fig. 1. Scatter plot representing the relationship between gait speed (*x*-axis) and spatio-temporal parameters stride length, cadence, stance time and first double support time, for adult and young subjects. Data from classes XS, S, M and L are merged.

With respect to N, pelvic rotation ROM was increased in task H ($p < 0.001$ for A and Y), and slightly increased in task T ($p < 0.05$ only for A). At hip level, flexion ROM was reduced both for H and T ($p < 0.001$ for A and Y). For H, positive work (generated power) was reduced for adults ($p < 0.05$), while negative work (absorbed power) was reduced for both adults and young ($p < 0.01$); no significant difference in power was observed for T. At knee level, swing flexion peak and flexion ROM were reduced for T and particularly for H ($p < 0.001$ for A and Y). Knee external extensor moment, which for N presented the highest peak corresponding to opposite foot-off, had maximum peak at opposite foot-strike for T, and kept on high value during all single support phase for H. Knee positive work showed no difference for both T and H, while absorbed power was reduced ($p < 0.001$ negative work difference from N, for both A and Y) and had a positive peak for T at foot-strike. The ankle was plantar-flexed for T with reduced ROM for adults ($p < 0.001$), while for H was almost blocked in a dorsiflexed position with reduced ROM both for young and adults ($p < 0.001$). Ankle moment had a double dorsiflexion peak in stance phase for T, the first one approximately corresponding to opposite foot-off. For task H, ankle power was almost null, while for T it denoted a noticeable absorption during first double support phase and a following generation in mid-stance and pre-swing ($p < 0.001$ increased negative work, for A and Y; positive work globally

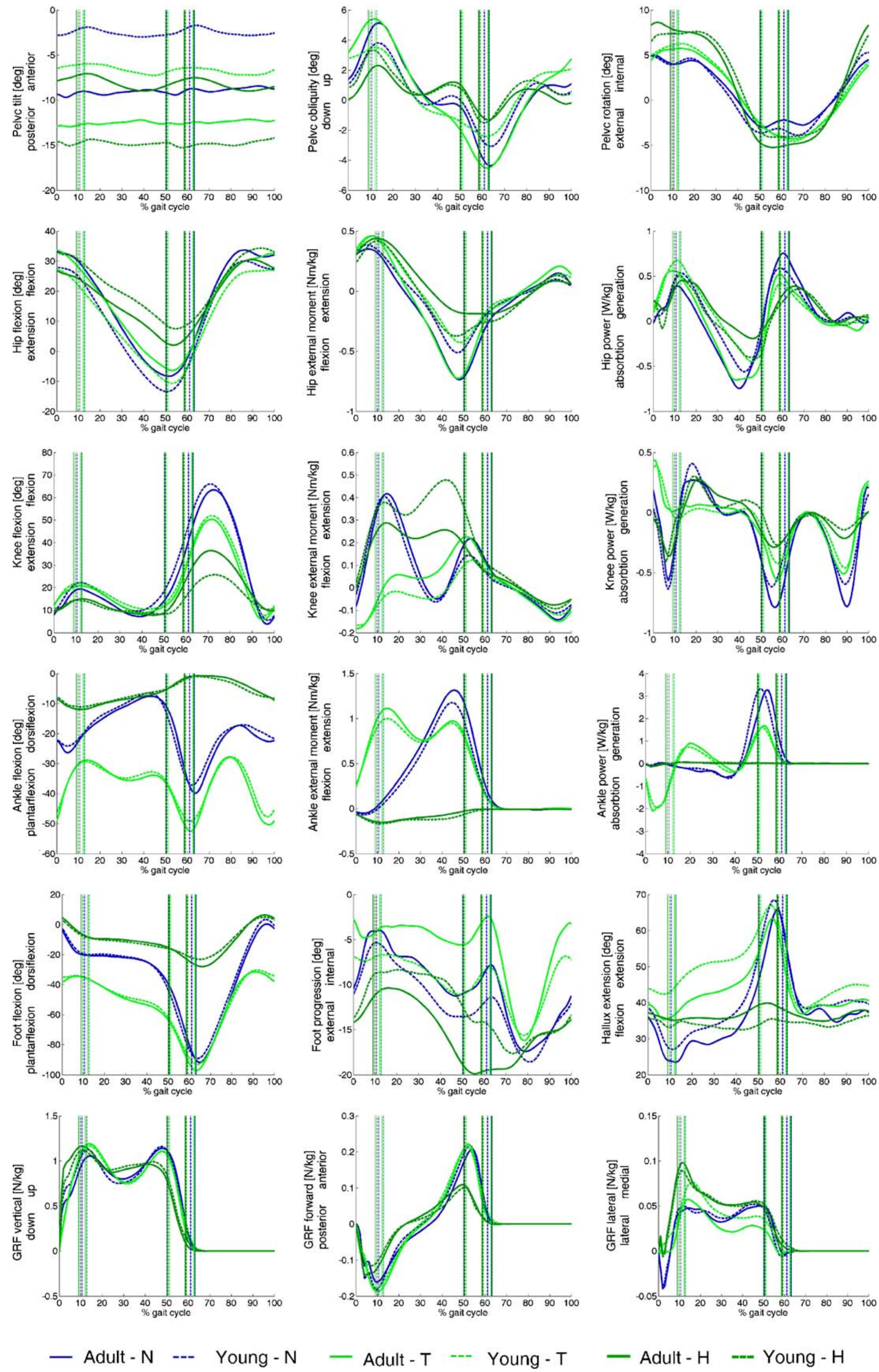

Fig. 2. Adult and young kinematics and kinetics, for self-selected speed walking (N, black lines), toe-walking (T, dark grey lines) and heel-walking (H, light grey lines). Solid lines represent adults, dashed lines young subjects. Each curve represents the average of trials for each speed class, and vertical lines indicate (from right to left) average ipsilateral foot-off, controlateral foot-strike and controlateral foot-off percents of gait cycle time.

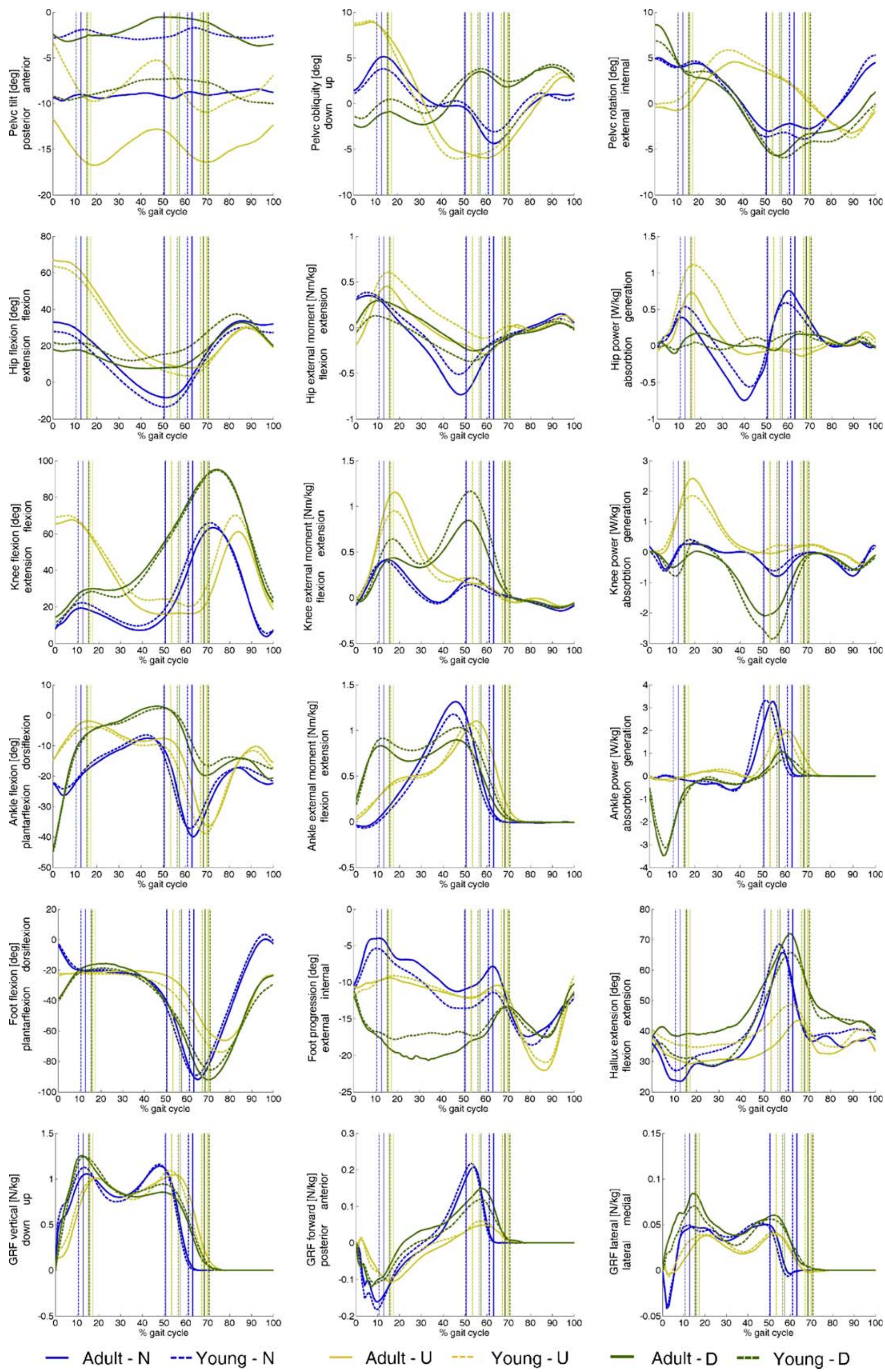

Fig. 3. Adult and young kinematics and kinetics, for self-selected speed walking (N, black lines), step ascending (U, dark grey lines) and step descending (D, light grey lines). Solid lines represent adults, dashed lines young subjects. Each curve represents the average of trials for each speed class, and vertical lines indicate (from right to left) average ipsilateral foot-off, controlateral foot-strike and controlateral foot-off percents of gait cycle time.

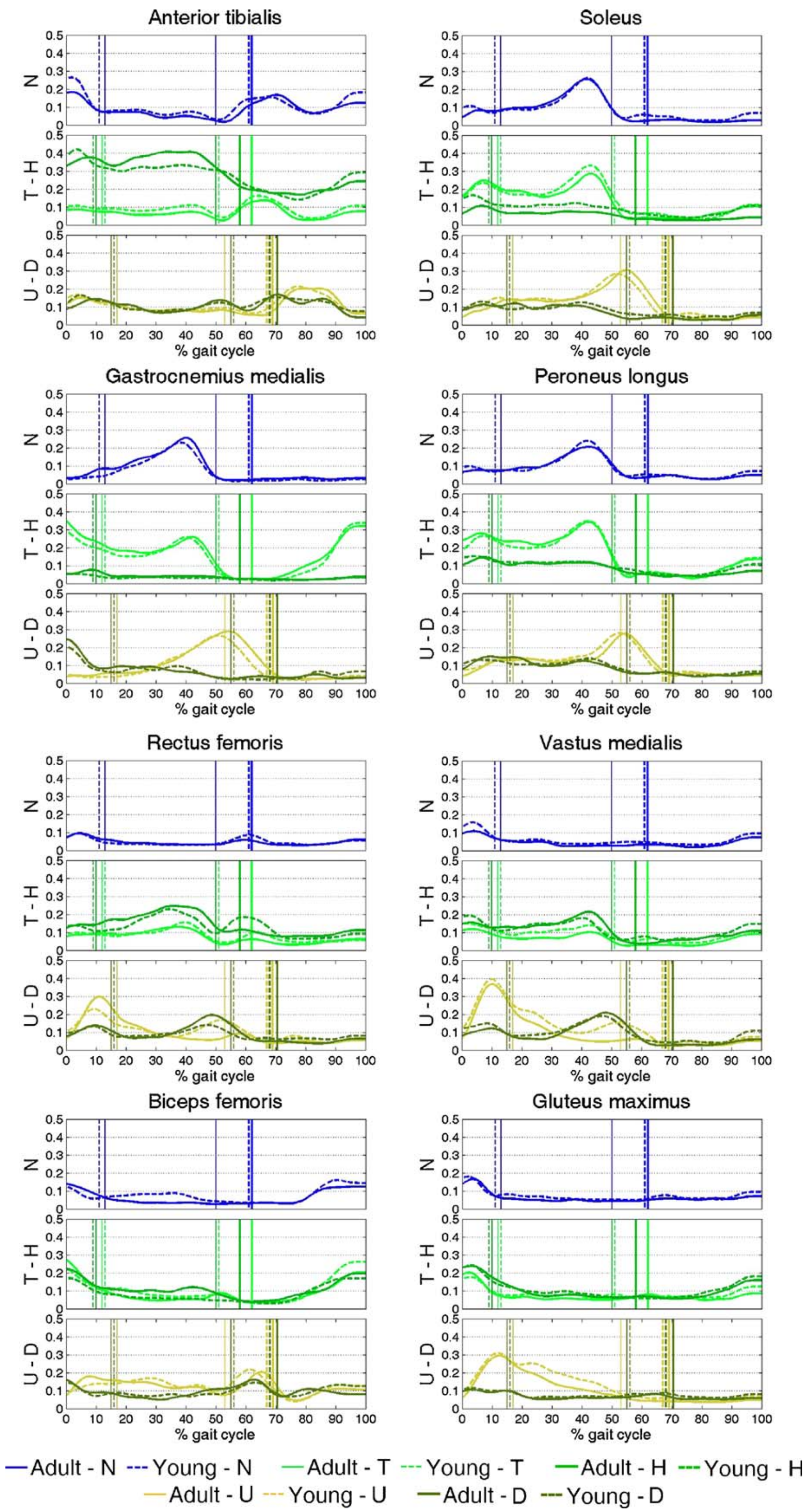

Fig. 4. EMG profiles of tibialis anterior, soleus, gastrocnemius medialis, peroneus longus, rectus femoris, vastus medialis, biceps femoris and gluteus maximus (*x*-axis, gait cycle time percent; *y*-axis normalized amplitude, scale from 0 to 0.5). Solid lines represent adults, dashed lines young subjects. Vertical lines indicate (from right to left) average ipsilateral foot-off, controlateral foot-strike and controlateral foot-off percents of gait cycle time. For each muscle, the upper graph reports self-selected speed walking, the middle one compares toe-walking (T) and heel-walking (H) and the lower one step ascending (U) and step descending (D).

reduced, $p < 0.001$ for A and Y). Considering foot kinematics, absolute sagittal angle at foot contact was more negative for T (plantarflexion, $p < 0.001$ for A and Y) and positive for H (dorsiflexion, $p < 0.001$ for A and Y). In task H, GRF showed an increase in medio-lateral component corresponding to a decrease in antero-posterior one ($p < 0.01$).

Pelvis ROM was increased in frontal plane for task U ($p < 0.001$ for A and Y), and in coronal plane for task D ($p < 0.001$). Hip flexion ROM was increased for U and reduced for D ($p < 0.001$ for A and Y). Higher power generation at hip level was required in early stance for U for young subjects ($p < 0.001$), perhaps due to the fixed step height, proportionally higher for young than for adults. Task D presented limited power dynamic (reduced positive and negative work, $p < 0.001$ for A and Y). At knee level, high external extension moment peaks occurred at controlateral foot-off for U (in association with power generation, $p < 0.001$ difference in positive for A and Y), and at opposite foot-strike for D (corresponding to power absorption, $p < 0.001$ difference in negative work for A and Y). At ankle level, dorsiflexion ROM was increased for U and specially for D ($p < 0.001$ for a and Y). For D, external moment presented a dorsiflexion peak preceding controlateral foot-off and corresponding to an absorption power peak. Negative work was increased for D ($p < 0.001$ for a and Y), while positive work was increased for U and reduced for D (both $p < 0.001$ for A and Y). Absolute sagittal foot angle showed a more plantar-flexed contact for both tasks, specially for D ($p < 0.001$ for A and Y).

3.4. EMG

Fig. 4 shows EMG envelope profiles for tasks T, H, U, D compared to N, both for adult and young. No marked difference was present between age groups.

For task T, plantaflexors (GAM and SOL) and PL were active through initial stance to mid-stance and part of terminal stance, while TA dorsiflexor activity was reduced, particularly in initial stance phase. RF, VM and BF seemed to increase their activity.

For task H, very evident was the increase in the activity of TA, corresponding to a decrease for GAM, SOL and PL. RF and VM also showed an evident activity increase, slightly noticeable also for BF and GM.

As principal observation for tasks U and D, we noticed a stronger activation of proximal muscles (RF, VM, BF and GM) in comparison to N.

4. Discussion

Correlation analysis proved good agreement between LAMB and VPiG protocols for all the considered variables. Comparison with data from Ferrari et al. [23] showed a similar figure (differences from their ρ values < 10%).

Considering different walking speeds, our results are in accordance with the literature findings above cited (e.g. [12,13]).

Comparing adult and young subjects at self-selected speed, we found that normalized gait speed was higher for young subjects, in contrast with Ganley et al. [27] who found that 7-year-olders had same natural speed as adults. The same was true for additional tasks T, H, U and D. Also Ganley's findings of higher ankle dorsiflexion moment peak and power generation peak for adults could not be confirmed: actually our data showed the opposite ($p < 0.001$ and $p < 0.05$ respectively). However, this was not true for tasks T and D (higher moment peak for adults, no significant difference for power peak). Hence, young's higher gait speed could not be the only cause of higher moment and power peak for N.

T and H tasks primarily challenged distal segments, requiring a wide ankle ROM and power generation/absorption skills, involving TA, SOL, GAM and PL muscles. Conversely, power required for U was mainly generated at hip and knee level in early stance, and at ankle level in late stance; energy dissipation for D was achieved by power absorption mainly at ankle level in early stance and at knee level in late stance. Accordingly, proximal muscles contribution (GM, RF, VM, BF) to power dynamics was enhanced in step ascending and descending tasks. These findings generally agreed with already cited papers (e.g. [20,11,17]).

We found similarities between tasks T and D at knee and chiefly at ankle level for joint moments and power, since both require toe ground contact.

Each of the proposed tasks presented peculiar traits which challenge the subject's skills beyond plain gait level, allowing a sort of spotlight analysis on target districts or functions of interest, in relation to specific pathologies. Patients presenting proximal deficits should undergo step negotiation which demonstrated to require noticeable power dynamics at proximal joints. Conversely, patients suffering diseases which mainly affect distal lower limbs should perform toe-walking and heel-walking as additional tasks to self-selected speed gait, since these tasks need good functionality of ankle-associated muscles.

For subjects at preclinical stage or slightly affected by pathology, additional tasks may help to let appear or quantify a latent deficit, which would not be manifest in self-selected speed gait.

Acknowledgement

This work was supported in part by a grant from the Italian Ministry of Health (Ricerca Finalizzata RF-FCG-2007-666049).

Conflict of interest

None of the authors reports a conflict of interest.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.gaitpost.2010.08.009.

References

[1] Simon SR. Quantification of human motion: gait analysis-benefits and limitations to its application to clinical problems. J Biomech 2004;37:1869–80.
[2] Benedetti MG, Piperno R, Simoncini L, Bonato P, Tonini A, Giannini S. Gait abnormalities in minimally impaired multiple sclerosis patients. Mult Scler 1999;5:363–8.
[3] Rao AK, Louis ED, Marder KS. Clinical assessment of mobility and balance impairments in pre-symptomatic Huntington's disease. Gait Posture 2009;30:391–3.
[4] Carpinella I, Crenna P, Calabrese E, Rabuffetti M, Mazzoleni P, Nemni R, et al. Locomotor function in the early stage of Parkinson's disease. IEEE Trans Neural Syst Rehabil Eng 2007;15(December (4)):543–51.
[5] Crenna P, Carpinella I, Rabuffetti M, Calabrese E, Mazzoleni P, Nemni R, et al. The association between impaired turning and normal straight walking in Parkinson's disease. Gait Posture 2007;26(July (2)):172–8.
[6] Burns J, Ryan MM, Ouvrier RA. Evolution of foot and ankle manifestations in children with CMT1A. Muscle Nerve 2009;39:158–66.
[7] Neptune RR, Burnfield JM, Mulroy SJ. The neuromuscular demands of toe walking: a forward dynamics simulation analysis. J Biomech 2007;40:1293–300.
[8] Sasaki K, Neptune RR, Burnfield JM, Mulroy SJ. Muscle compensatory mechanisms during able-bodied toe walking. Gait Posture 2008;27:440–6.
[9] Westberry DE, Davids JR, Davis RB, de Morais Filho MC. Idiopathic toe walking: a kinematic and kinetic profile. J Pediatr Orthop 2008;28:352–8.
[10] Andriacchi TP, Andersson GB, Fermier RW, Stern D, Galante JO. A study of lower-limb mechanics during stair-climbing. J Bone Joint Surg Am 1980;62:749–57.
[11] Riener R, Rabuffetti M, Frigo C. Stair ascent and descent at different inclinations. Gait Posture 2002;15:32–44.
[12] Schwartz MH, Rozumalski A, Trost JP. The effect of walking speed on the gait of typically developing children. J Biomech 2008;41:1639–50.

[13] Stoquart G, Detrembleur C, Lejeune T. Effect of speed on kinematic, kinetic, electromyographic and energetic reference values during treadmill walking. Neurophysiol Clin 2008;38:105–16.
[14] Jonsdottir, Recalcati M, Rabuffetti M, Casiraghi A, Boccardi S, Ferrarin M. Functional resources to increase gait speed in people with stroke: strategies adopted compared to healthy controls. Gait Posture 2009;29(3):355–9.
[15] Stansfield BW, Hillman SJ, Hazlewood ME, Robb JE. Regression analysis of gait parameters with speed in normal children walking at self-selected speeds. Gait Posture 2006;23:288–94.
[16] Mian OS, Thom JM, Narici MV, Baltzopoulos V. Kinematics of stair descent in young and older adults and the impact of exercise training. Gait Posture 2007;25:9–17.
[17] Protopapadaki A, Drechsler WI, Cramp MC, Coutts FJ, Scott OM. Hip, knee, ankle kinematics and kinetics during stair ascent and descent in healthy young individuals. Clin Biomech (Bristol Avon) 2007;22:203–10.
[18] Romkes J, Brunner R. An electromyographic analysis of obligatory (hemiplegic cerebral palsy) and voluntary (normal) unilateral toe-walking. Gait Posture 2007;26:577–86.
[19] Alvarez C, De Vera M, Beauchamp R, Ward V, Black A. Classification of idiopathic toe walking based on gait analysis: development and application of the ITW severity classification. Gait Posture 2007;26:428–35.
[20] Armand S, Watelain E, Mercier M, Lensel G, Lepoutre FX. Identification and classification of toe-walkers based on ankle kinematics, using a data-mining method. Gait Posture 2006;23:240–8.
[21] Armand S, Watelain E, Roux E, Mercier M, Lepoutre FX. Linking clinical measurements and kinematic gait patterns of toe-walking using fuzzy decision trees. Gait Posture 2007;25:475–84.
[22] Segal LS, Thomas SE, Mazur JM, Mauterer M. Calcaneal gait in spastic diplegia after heel cord lengthening: a study with gait analysis. J Pediatr Orthop 1989;9:697–701.
[23] Ferrari A, Benedetti MG, Pavan E, Frigo C, Bettinelli D, Rabuffetti M, et al. Quantitative comparison of five current protocols in gait analysis. Gait Posture 2008;28:207–16.
[24] Rabuffetti M, Crenna P. A modular protocol for the analysis of movement in children. Gait Posture 2004;20:S77–8.
[25] Hof AL. scaling data to body size. Gait Posture 1996;4:222–3.
[26] Burden AM, Trew M, Baltzopoulos V. Normalisation of gait EMGs: a re-examination. J Electromyogr Kinesiol 2003;13:519–32.
[27] Ganley KJ, Powers CM. Gait kinematics and kinetics of 7-year-old children: a comparison to adults using age-specific anthropometric data. Gait Posture 2005;21:141–5.