Human Movement Science 66 (2019) 9–21

Contents lists available at ScienceDirect

Human Movement Science

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

Common and specific gait patterns in people with varying anatomical levels of lower limb amputation and different prosthetic components

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ARTICLE INFO

Keywords:
Lower limb amputation
Prosthetic gait
Kinematic
Kinetic
Energetic gait parameters

ABSTRACT

The present study's aim was to identify the kinematic and kinetic gait patterns and to measure the energy consumption in people with amputation according to both the anatomical level of amputation and the type of prosthetic components in comparison with a control group matched for the gait speed. Fifteen subjects with unilateral transtibial amputation (TTA), forty with unilateral transfemoral amputation (TFA) (9 with mechanical, 17 with CLeg and 14 with Genium prosthesis) and forty healthy subjects were recruited. We computed the time-distance gait parameters; the range of angular motion (RoM) at hip, knee and ankle joints, and at the trunk and pelvis; the values of the 2 peaks of vertical force curve; the full width at half maximum (FWHM) and center of activity (CoA) of vertical force; the mechanical behavior in terms of energy recovery (R-step) and energy consumption. The main results were: i) both TTA and TFA show a common gait pattern characterized by a symmetric increase of step length, step width, double support duration, pelvic obliquity, trunk lateral bending and trunk rotation RoMs compared to control groups. They show also an asymmetric increase of stance duration and of Peak1 in non-amputated side and a decrease of ankle RoM in amputated side; ii) only TFA show a specific gait pattern, depending on the level of amputation, characterized by a symmetric reduction of R-step and an asymmetric decrease of stance duration, CoA and FWHM and an increase of Peak1 in the amputated side and of hip and knee RoM, CoA and FWHM in the non-amputated side; iii) people with amputation with Genium prosthesis show a longer step length and increased hip and knee RoMs compared to people with amputation with mechanical prosthesis who conversely show an increased pelvic obliquity: these are specific gait patterns depending of the type of prosthesis. In conclusion, we identified both common and specific gait patterns in people with amputation, either regardless of, or according to their level of amputation and the type of prosthetic component.

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https://doi.org/10.1016/j.humov.2019.03.008
Received 30 October 2018; Received in revised form 8 March 2019; Accepted 11 March 2019
Available online 16 March 2019
0167-9457/

1. Introduction

Prosthesis use in persons with lower limb amputation at different anatomic levels requires complex adaptation strategies, both in the prosthesis and in the sound sides, during gait (Bateni & Olney, 2002; Tura, Rocchi, Raggi, Cutti, & Chiari, 2012) and other daily life locomotor tasks (Ventura, Segal, Klute, & Neptune, 2011; Vrieling et al., 2008). Indeed, prosthetic gait reflects a mixture of deviations from normal gait and adaptive and compensatory motions dictated by residual limb function after amputation. For this reason, quantifying and characterizing the gait of persons with a prosthesis is an essential prerequisite to improve our ability to develop new and ergonomic prosthetic devices, as well as to optimize the rehabilitation programs (Schafer, Perry, & Vanicek, 2018).

Many previous studies have been performed on gait kinematics and kinetics of people with amputation in order to characterize their typical walking patterns.

It has been reported that the prosthetic limb shows a longer stride than the intact limb; however, the prosthetic limb's stance phase lasts less than that of the unaffected one (Bateni & Olney, 2002; Sanders, Bell, Okumura, & Dralle, 1998; Smidt, 1990). In addition, a greater hip flexion in early stance of the prosthetic limb and a higher than normal knee flexion in the early stance phase of the intact limb have been reported to probably improve the overall stability and energy expenditure, respectively (Bateni & Olney, 2002; Esquenazi, 2014).

Regarding the force interaction, when increasing walking speed, the vertical ground reaction force increases, particularly in the intact limb (Nolan et al., 2003). This tendency to load the intact limb more than the prosthetic limb has been also reported during gait initiation (Rossi, Doyle, & Skinner, 1995).

The two factors influencing the gait in people with amputation are the level of the amputation and the type of prostheses. Regarding the former factor, the gait in people with transfemoral amputation seems to be more asymmetric than that in people with transtibial amputation: they show wider steps of longer duration compared to people with transtibial amputation (Highsmith et al., 2011). Concerning the latter factor, in recent years the prostheses have improved in design, materials, and technology (Chitragari, Mahler, Sumpio, Blume, & Sumpio, 2014) to be more effective in terms of efficiency of ambulation, minimization of the asymmetries, and reduction of compensatory movements, which, over time, may prove damaging to individuals. Actually, subjects with lower limb amputation wear different type of prostheses, such as the old concept mechanical prostheses or the most recent and technologically advanced prostheses (Microprocessor Controlled Knees (MPKs)), i.e. CLeg and Genium (Cutti et al., 2017; Highsmith et al., 2016; Kannenberg, Zacharias, Mileusnic, & Seyr, 2013). These differences are related to different aspects, including individuals' preference and adaptation, time from the amputation/prosthesis implantation, local insurance laws for the prosthesis reimbursement.

To date, no consistent study has evaluated the impact of the different types of prostheses on the gait function according to the anatomical level of amputation. Such analysis would allow better understanding of the advantages and disadvantages of the different types of prostheses. In addition, some limitations are present in many studies, since most of them included a small sample of subjects (Segal, Orendurff, Czerniecki, Schoen, & Klute, 2011b; Shell, Segal, Klute, & Neptune, 2017; Sturdy, Gates, Darter, & Wilken, 2014), had no control group (Highsmith et al., 2011; Howcroft, Lemaire, Kofman, & Kendell, 2015; Sturdy et al., 2014), and did not match controls for the gait speed (Fradet, Alimusaj, Braatz, & Wolf, 2010; Sturk et al., 2017). Since many kinematic and kinetic variables are speed-dependent (Kluge, Krinner, Lochmann, & Eskofier, 2017; Stoquart, Detrembleur, & Lejeune, 2008), not controlling for the speed (by not including a control group walking at matched gait speed) may create uncertainty in the interpretation of the pathologic gait pattern and may not allow detection of a reliable and specific gait pattern in order to distinguish what is unique from what is common. Furthermore, the lack of a matched speed comparison between people with amputations and healthy subjects do not allow discriminating the primary deficits from compensatory mechanisms.

The aim of the present study was to identify the kinematic and kinetic gait patterns and to measure the energy consumption in a large sample of people with amputations according to both the anatomical level of amputation and the type of prosthetic components in comparison with a control group matched for the gait speed.

2. Materials and methods

2.1. Subjects

Fifty-five subjects with lower-limb amputation from the prosthetics center of Italian Workers' Compensation Authority (INAIL) of Rome were enrolled in this study between September 2015 and September 2018. All patients had a unilateral transtibial or transfemoral amputation as a consequence of a workplace traumatic accident. Among subjects with transfemoral amputation, mechanical prosthesis (Kannenberg et al., 2013; Tura et al., 2012), and two types of MPKs prosthesis: CLeg and Genium (Ottobock, Duderstadt, Germany) (Highsmith et al., 2016) were used. INAIL provided the same type prosthetic foot (Ossur Variflex foot), as well as the same sockets and suspensions for all participants (Ossur, Reykjavík, Iceland). All patients wore their prosthesis daily at least since 2 years, and were able to ambulate independently along level surfaces without mobility aids. None of the subjects had any chronic disease, cardiac complication, uncontrolled asthma or diabetes mellitus, severe osteoporosis, or cognitive disorder.

The study group included 15 subjects with transtibial amputation (TTA) (15 men; age, 52.81 ± 14.51 years; height, 176.44 ± 5.40 cm; mass, 87.44 ± 11.08 kg) and 40 subjects with transfemoral amputation (TFA) (37 men; age, 54.94 ± 12.31 years; height, 172.85 ± 7.95 cm; mass, 83.47 ± 15.69 kg). Among the 40 subjects with TFA, 9 wore a mechanical prosthesis (TFA_M), 17 a CLeg prosthesis (TFA_C), and 14 a Genium prosthesis (TFA_G).

Forty healthy subjects were recruited as the control group (C) and were age-sex-speed matched with subjects with TFA (C_{mTFA}). A subgroup of 15 healthy subjects was age-sex-speed matched with subjects with TTA (C_{mTTA}) and a subgroup of 13 age-sex-speed matched subjects with TFA (TFA_m) was age-sex-speed matched with 13 subjects with TTA (TTA_m) to analyze the effect of different anatomical levels of amputation.

2.2. Gait analysis

"Gait was analyzed by using an optoelectronic motion analysis system (SMART-DX 6000 System, BTS, Milan, Italy), consisting of six infrared cameras (sample frequency, 340 Hz) used to detect the movement of twenty-seven passive spherical markers placed over prominent bony landmarks (Davis, Õunpuu, Tyburski, & Gage, 1991; Wu et al., 2002, 2005). In detail, the markers were placed over the head, the cutaneous projections of the spinous processes of the seventh cervical vertebra and sacrum, and bilaterally over the acromion, olecranon, ulnar styloid process, anterior superior iliac spine, and greater trochanter for all the subjects with TTA and TFA and C. In addition, markers were placed, bilaterally in C and unilaterally on the amputated side of subjects with amputation, over the lateral femoral condyle, fibula head, lateral malleoli, fifth metatarsal head, and heel (for these last two points the markers were placed on the shoes). In subjects with TTA and TFA, amputated limb markers were placed over symmetrical points (no anatomical landmarks) with respect to the homologous marker's position on the non-amputated limb. Furthermore, wand markers were placed bilaterally on femurs and legs (Davis et al., 1991). Two dynamometric platforms (Kistler 9286AA, Winterthur, Switzerland), adjacent to each other in the longitudinal direction, but displaced by 0.2 m in the lateral direction, were used to acquire ground reaction forces (sampling rate, 680 Hz). Data acquisition from the optoelectronic cameras and dynamometric platforms was integrated and synchronized.

2.3. Experimental procedure

C and subjects with TTA and TFA underwent an initial training session to become familiar with the assessment procedures. Patients were asked to walk with their shoes at comfortable self-selected speeds along a walkway while looking forward. Because we were interested in natural locomotion, only qualitative instructions were provided, and each subject was free to choose his own cadence. On the other hand, C were requested to walk with their shoes at their preferred speed and at a lower speed to match the speed between groups (see 2.4.1). At least ten trials were recorded for each subject and for each speed. To avoid fatigue, groups of three trials were separated by 1-min rest periods in subjects with amputation.

2.4. Data analysis

After each acquisition performed by Smart Capture (BTS, Milan, Italy), data were processed using SMART Tracker and Analyzer software (BTS, Milan, Italy) and Matlab software (version 7.10.0, MathWorks, Natick, MA, USA). The gait cycle was defined as the interval between two successive foot contacts of the same leg. Kinematic and kinetic data were time-normalized to the duration of the gait cycle and interpolated to 101 samples using a polynomial procedure. In this study, heel strike and toe off instants were calculated from kinematic data. After this preprocessing procedure, time-distance, kinematic, and kinetic parameters were calculated. For people with amputations, each parameter was calculated for the amputated (A) and non-amputated (NA) side, while for C the parameters were evaluated without distinguishing between sides.

2.4.1. Speed matching procedure

Walking speed was matched between groups as follows: only the walking trials of C (preferred speed or at a lower speed) whose speed was near to a corresponding subject with amputation were considered. Furthermore, only a subgroup of subjects with TFA (TFA_m) was age-sex matched with subjects with TTA. To have the speed match between the groups of TFA_m and TTA, 2 subjects with TTA were excluded, leaving only 13 subjects (TTA_m), because their speeds were far from those of the age-sex matched subjects with TFA_m. Paired two-sample *t*-test was used to investigate the differences in walking speed between patients and C, both TTA vs. C_{mTTA} and TFA vs. $C_{mTFA,}$ and between TTA_m and TFA_m. In this way, the mean speed values were not statistically different (the p value for statistical significance was set at 0.05) between groups (TTA 1.08 ± 0.16 m/s, C_{mTTA} 0.97 ± 0.20 m/s, p = 0.116; TFA 0.92 ± 0.20 m/s, C_{mTFA} 0.93 ± 0.25 m/s, p = 0.813; TTA_m 1.05 ± 0.15 m/s, TFA_m 0.94 ± 0.16 m/s, p = 0.084).

2.4.2. Time-distance parameters

The following time-distance gait parameters were calculated for each subject: walking speed (m/s); cadence (step/s); step width (m); step length (m) (from the heel strike of a limb and the subsequent heel strike of the other limb); stance, swing, and double support phases durations (expressed as percentages of the gait cycle duration). Step length and the step width were normalized to the limb length of each subject.

2.4.3. Kinematic data

The anatomical and prosthetic joint angles for the hip, knee, ankle, trunk, and pelvis (frontal, sagittal, and transverse plane) were computed. Based on these variables, the joint range of motion (RoM) was calculated as the difference between the maximum and minimum values during the gait cycle.

2.4.4. Kinetic data

The vertical component of the ground reaction forces (Vertical Force, VF) provided by the dynamometric platforms, was normalized to the subject's body weight (Serrao et al., 2016).

For each subject and gait speed, the values of the 2 peaks of VF ($Peak1_{VF}$ and $Peak2_{VF}$) were computed. Furthermore, the full width at half maximum (FWHM) and the center of activity (CoA) of VF ($FWHM_{VF}$ and CoA_{FV}) were computed (Varrecchia et al., 2018). FWHM was calculated as the sum of the time durations during which the VF curves were higher than their half maximum. The CoA was calculated using circular statistics as follow:

$$A = \sum_{i=1}^{200} (\cos \theta_t \times VF_i) \tag{1}$$

$$B = \sum_{i=1}^{200} (\sin \theta_t \times VF_i) \tag{2}$$

$$CoA = \tan^{-1}\left({}^{B}\!/_{A}\right) \tag{3}$$

where θ_t is the angle that varies from 0° to 360° to plot in polar coordinates (polar direction denoted the phase of the gait cycle).

2.4.5. Energy consumption measurement

The mechanical behavior was measured in terms of energy recovery and energy consumption in relation to the whole-body center of mass (CoM). This methodology has been validated both in normal (Cavagna, Thys, & Zamboni, 1976; Cavagna, Willems, Legramandi, & Heglund, 2002) and abnormal gait patterns (Detrembleur, Dierick, Stoquart, Chantraine, & Lejeune, 2003; Don et al., 2007), and gives information on the mechanical energy expenditure involving the whole skeletal muscle system during walking.

The whole-body CoM was calculated as the centroid of a set of elements composed by 13 body segments (head, trunk, arms, forearms, pelvis, thigh, shanks, and feet). The computation was performed by considering kinematic and anthropometric data together with the body segment parameters (de Leva, 1996; Zatsiorsky, Seluyanov, & Chugunova, 1990), according to the weighted average of the individual body segments' center of mass (Ranavolo et al., 2017):

$$CoM_x = \frac{1}{m}\sum_{i=1}^{n} x_i \times m_i \tag{4}$$

$$CoM_y = \frac{1}{m}\sum_{i=1}^{n} y_i \times m_i \tag{5}$$

$$CoM_z = \frac{1}{m}\sum_{i=1}^{n} z_i \times m_i \tag{6}$$

where CoM_x, CoM_y and CoM_z are, respectively, the instantaneous x, y, and z components of the CoM position, m is the whole-body mass of the subject, $n = 13$ is the number of parts being considered, x_i, y_i, and z_i are the components of the CoM position of the i_{th} part, and m_i is the mass of the i_{th} segment or residuum or prosthesis components.

The kinetic energy (E_k) associated with CoM displacements during the gait cycle was calculated as the sum of the kinetic energy on the x (E_{kx}), y (E_{ky}), and z (E_{kz}) axes as follows:

$$E_k = E_{kx} + E_{ky} + E_{kz} = \frac{1}{2}m(v_x^2 + v_y^2 + v_z^2) \tag{7}$$

where m and v_x, v_y, and v_z are, respectively, the mass and the three velocity components of the CoM evaluated starting from its components (CoM_x, CoM_y and CoM_z). Furthermore, the potential energy (E_p) associated with CoM was calculated as follows:

$$E_p = mgh \tag{8}$$

where h is the vertical (y) CoM component and g is the gravity acceleration.

The total mechanical energy (E_{tot}) associated with CoM was computed as the sum of E_k and E_p.

The fraction of mechanical energy recovered during each walking step (R-step) (Cavagna et al., 2002) was calculated as follows:

$$R-step = \frac{W_p^+ + W_{kf}^+ - W_{tot}^+}{W_p^+ + W_{kf}^+} = \left(1 - \frac{W_{tot}^+}{W_p^+ + W_{kf}^+}\right) \tag{9}$$

where W_p^+, W_{kf}^+, W_{tot}^+ represent the positive work (sum of the positive increments over one step) produced by the gravitational potential energy, by the kinetic energy of forward motion, and by the total mechanical energy, respectively. Additionally, the total energy consumption (TEC) was calculated as follows (Cavagna et al., 1976; Varrecchia et al., 2018):

$$TEC = \frac{W_{tot}^+}{0.21} \tag{10}$$

TEC was then normalized to the body weight and step length. For each subject, R-step and the normalized TEC values of all the steps were averaged.

2.5. Statistics

The Shapiro-Wilk test was used to analyze the normal distribution of the data. The dependent *t*-test or the Wilcoxon's signed-rank test were used to evaluate the differences between the A and NA sides of subjects with TTA and TFA for each time-distance, kinematic, kinetic, and energy consumption measurements. Furthermore, paired two-sample *t*-test or the Mann-Whitney test (two-tailed) was used for each parameter to test for between-group differences (TFA vs. C_{mTFA}, TTA vs. C_{mTTA}, and TFA_m vs. TTA_m). To evaluate the effect of the gait speed on the CoA across the gait cycle, Watson-Williams test was used to test for circular data (Varrecchia et al., 2018).

Furthermore, one-way ANOVA was used also to evaluate the differences among the three subgroups of subjects with TFA for each evaluated parameter. Post hoc analyses (with Bonferroni's corrections) were performed when ANOVA showed significant differences.

Significance level was set at $p < 0.05$. All analyses were performed using SPSS 20.0 (*SPSS Inc., Chicago, IL, USA*) and Matlab (*version 8.3.0.532, MathWorks, Natick, MA, USA*) software.

3. Results

3.1. Time-distance parameters

The means, standard deviations, and statistical results for each time-distance parameter and for each group are presented in Table 1.

3.1.1. People with amputation vs. controls

Significantly increased step width, step length, and double support duration in both sides were found in both TTA and TFA groups compared to the C (C_{mTTA} and C_{mTFA}). Stance duration was significantly increased in the NA side in both TTA and TFA groups, and significantly decreased in the A side in the TFA group. Conversely, the swing duration was significantly decreased in the NA side in both TTA and TFA groups, and significantly increased in the A side in TFA group (Table 1).

Significantly shorter stance duration and longer swing duration were found in the A side than in the NA side (Table 1, Fig. 2) in TTA and TFA groups.

3.1.2. Type of prostheses (TFA_M, TFA_C and TFA_G)

A significant effect of the type of prosthesis on the step length of the NA side was detected. Post hoc analysis revealed higher values for the Genium prosthesis compared to mechanical prosthesis (Table 1).

A significantly shorter step length in A side than in the NA one (Table 1) was found in TFA_G subgroup. Furthermore, significantly shorter stance duration and longer swing duration in the A side than in the NA one (Table 2) were found in all three TFA_M, TFA_C and TFA_G subgroups.

3.1.3. TTA_m vs. TFA_m

A significant effect of the type of amputation (TTA_m vs. TFA_m) on the stance and swing duration was found in both sides, with the stance significantly increased and the swing significantly decreased in the NA side in TFA_m group compared to TTA_m group, (Table 1). Conversely, the stance significantly decreased and the swing significantly increased in the A side in TFA_m group compared to TTA_m group (Table 1).

Significantly shorter stance duration and longer swing duration in the A side than in the NA one (Table 1) were found both in TTA_m and TFA_m.

3.2. Kinematic data

The means, standard deviations and statistical results of RoM for the hip, knee, ankle, pelvic, and trunk for each group are shown in Fig. 1.

3.2.1. People with amputation vs controls

Significantly increased hip and knee RoMs in NA side were found in TFA compared to C_{mTFA} (Fig. 1). Furthermore, significantly decreased ankle RoMs in A side were detected in both TTA and TFA compared to the speed matched C (C_{mTTA} and C_{mTFA}) (Fig. 1).

Significantly increased pelvic obliquity, trunk lateral bending, and trunk rotation RoMs of both sides were found in both TTA and TFA groups compared to C (Figs. 1 and 2). Moreover, pelvic tilt, pelvic rotation, and trunk flection-extension RoMs of both sides were significantly increased in TFA group compared to C_{mTFA} group (Figs. 1 and 2). Fig. 2 also shows that people with amputation walked with the pelvis and trunk ante-flexed (flexed in a forward direction) compared to C.

A significantly shorter hip and knee RoMs were found in the A side than in the NA side (Fig. 1) in TFA group. Furthermore, a significantly shorter ankle RoMs were found in the A side than in the NA side (Fig. 1) in both TTA and TFA groups.

3.2.2. Type of prostheses (TFA_M, TFA_C, and TFA_G)

A significant effect of the type of prosthesis on the hip and knee RoMs in the A side and on the pelvic obliquity RoM was found in both sides. Post hoc analysis revealed higher values of the hip and knee RoMs in the A side of TFA_G subgroup compared to TFA_M

Table 1
The means, standard deviations, and statistical results (p value) of walking speed, cadence, step width, step length, stance duration, swing duration, and double support duration.

Time-distance parameters		People with amputation vs controls						Type of prostheses				TTA_m vs TFA_m		
		C_{mTTA}	TTA	p group	C_{mTFA}	TFA	p group	TFA_M	TFA_C	TFA_G	p group	TTA_m	TFA_m	p group
Walking speed (m/s)		0.97 ± 0.20	1.08 ± 0.16	0.1163	0.93 ± 0.25	0.92 ± 0.20	0.813	0.82 ± 0.20	0.91 ± 0.23	1.01 ± 0.12	0.064	1.05 ± 0.15	0.94 ± 0.16	0.084
Cadence (cycle/s)	A	0.84 ± 0.07	0.83 ± 0.07	0.52	0.81 ± 0.12	0.78 ± 0.08	0.289	0.75 ± 0.07	0.79 ± 0.09	0.80 ± 0.06	0.309	0.82 ± 0.10	0.77 ± 0.07	0.077
	NA		0.82 ± 0.07	0.481		0.78 ± 0.08	0.397	0.74 ± 0.07	0.79 ± 0.09	0.80 ± 0.06	0.16	0.81 ± 0.07	0.77 ± 0.07	0.11
	p side		0.083			0.439		0.269	0.158	0.619		0.114	0.973	
Step width (% limb length)		0.20 ± 0.05	0.28 ± 0.06	< **0.001**	0.19 ± 0.06	0.30 ± 0.09	< **0.001**	0.33 ± 0.10	0.32 ± 0.08	0.27 ± 0.09	0.214	0.27 ± 0.05	0.29 ± 0.09	0.538
Step length (% limb length)	A	0.61 ± 0.12	0.74 ± 0.09	**0.002**	0.60 ± 0.10	0.66 ± 0.09	**0.014**	0.63 ± 0.11	0.65 ± 0.12	0.66 ± 0.08	0.729	0.74 ± 0.10	0.64 ± 0.11	0.058
	NA		0.75 ± 0.07	**0.001**		0.67 ± 0.09	**0.003**	0.64 ± 0.12	0.65 ± 0.09	0.72 ± 0.05	**0.048**	0.74 ± 0.07	0.70 ± 0.08	0.182
	p side		0.903			0.241		0.681	0.877	**0.046**		0.946	0.38	
Stance duration (% cycle)	A	61.78 ± 1.78	61.89 ± 1.79	0.871	62.38 ± 2.88	59.63 ± 2.57	< **0.001**	60.14 ± 2.31	60.09 ± 3.02	58.74 ± 2.02	0.285	61.68 ± 1.80	58.71 ± 0.88	< **0.001**
	NA		64.17 ± 3.44	**0.009**		67.99 ± 3.31	< **0.001**	67.21 ± 3.02	69.27 ± 3.92	66.94 ± 2.14	0.107	64.68 ± 3.42	67.99 ± 3.15	**0.017**
	p side		**0.014**			< **0.001**		**0.004**	< **0.001**	< **0.001**		**0.01**	< **0.01**	
Swing duration (% cycle)	A	38.85 ± 3.29	38.09 ± 1.80	0.648	38.30 ± 2.99	39.88 ± 2.98	< **0.001**	39.9 ± 2.93	39.01 ± 2.33	40.91 ± 2.05	0.205	38.28 ± 1.82	40.58 ± 2.20	**0.002**
	NA		35.82 ± 3.43	**0.009**		31.91 ± 3.30	< **0.001**	31.6 ± 2.53	31.11 ± 4.24	33.08 ± 2.04	0.25	35.3 ± 3.4	31.64 ± 3.18	**0.011**
	p side		**0.015**			< **0.001**		**0.004**	< **0.001**	< **0.001**		< **0.01**	< **0.01**	
Double support duration (% cycle)	A	23.17 ± 3.15	26.07 ± 4.46	**0.049**	23.54 ± 5.76	27.72 ± 5.01	**0.001**	28.54 ± 3.73	28.97 ± 6.36	25.66 ± 3.12	0.161	26.38 ± 4.74	27.08 ± 3.54	0.677
	NA		26.08 ± 4.50	**0.037**		27.44 ± 6.58	< **0.001**	24.28 ± 9.03	30.26 ± 6.47	26.04 ± 2.98	0.05	24.4 ± 4.77	27.41 ± 4.09	0.568
	p side		0.670			0.367[b]		0.125	0.081	0.298		0.672	0.698	

C_{mTTA}: healthy subjects age-sex-speed matched with TTA; TTA: subjects with transtibial amputation; C_{mTFA}: healthy subjects age-sex-speed matched with TFA; TFA: subjects with transfemoral amputation; TFA_M: subjects with transfemoral amputation with mechanical prosthesis; TFA_C: subjects with transfemoral amputation with CLeg prosthesis; TFA_G: subjects with transfemoral amputation with Genium prosthesis; TFA_m: a subgroup of 13 age-sex-speed matched subjects with a subgroup of TTA; TTA_m: a subgroup of 13 age-sex-speed matched subjects with a subgroup of TFA.

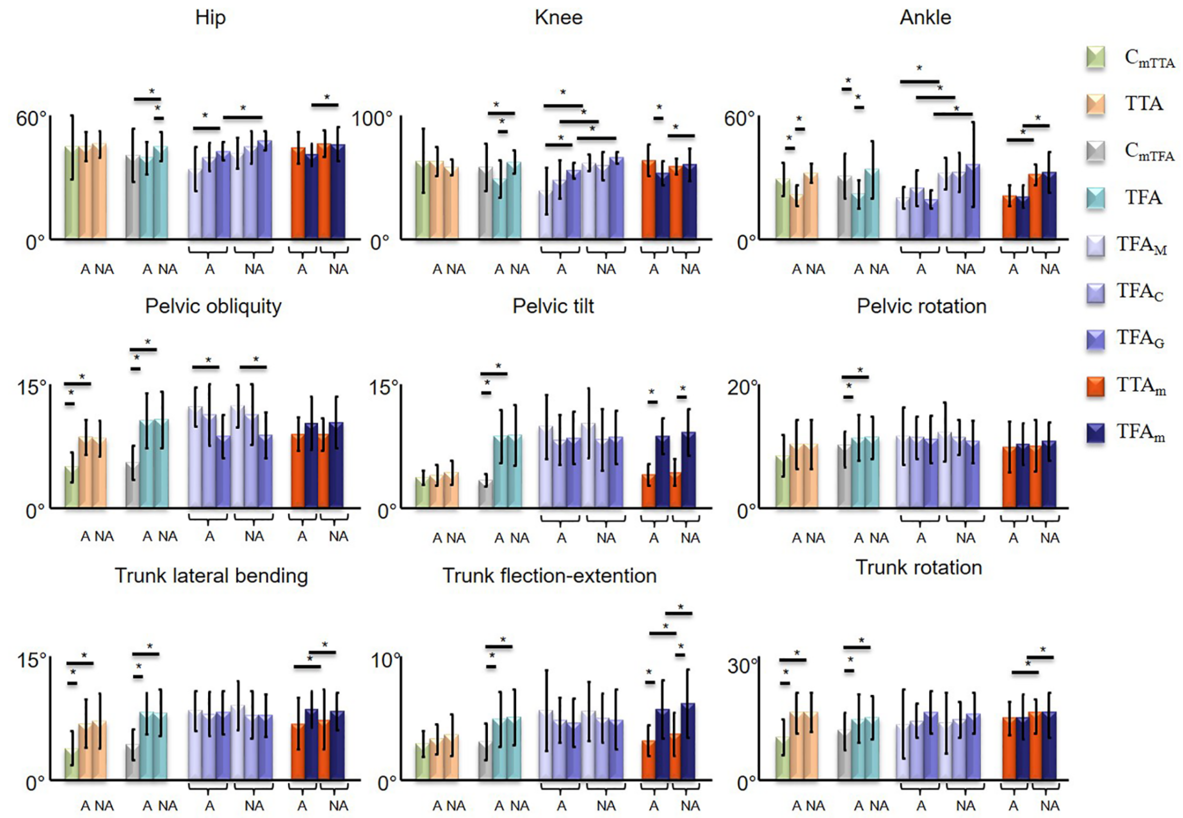

Fig. 1. The means, standard deviations, and statistical results of range of motion for the hip, knee, ankle, pelvic, and trunk for each group. C_{mTTA}: healthy subjects age-sex-speed matched with TTA; TTA: subjects with transtibial amputation; C_{mTFA}: healthy subjects age-sex-speed matched with TFA; TFA: subjects with transfemoral amputation; TFA_M: subjects with transfemoral amputation with mechanical prosthesis; TFA_C: subjects with transfemoral amputation with CLeg prosthesis; TFA_G: subjects with transfemoral amputation with Genium prosthesis; TFA_m: a subgroup of 13 age-sex-speed matched subjects with a subgroup of TTA; TTA_m: a subgroup of 13 age-sex-speed matched subjects with a subgroup of TFA.

subgroup (Fig. 1) and lower values of the pelvic obliquity RoMs for the Genium prosthesis (TFA_G) compared to mechanical prosthesis (TFA_M) in both sides (Fig. 1).

Significantly decreased knee and ankle RoMs in the A side than in NA side were found in TFA_M, TFA_C, and TFA_G subgroups

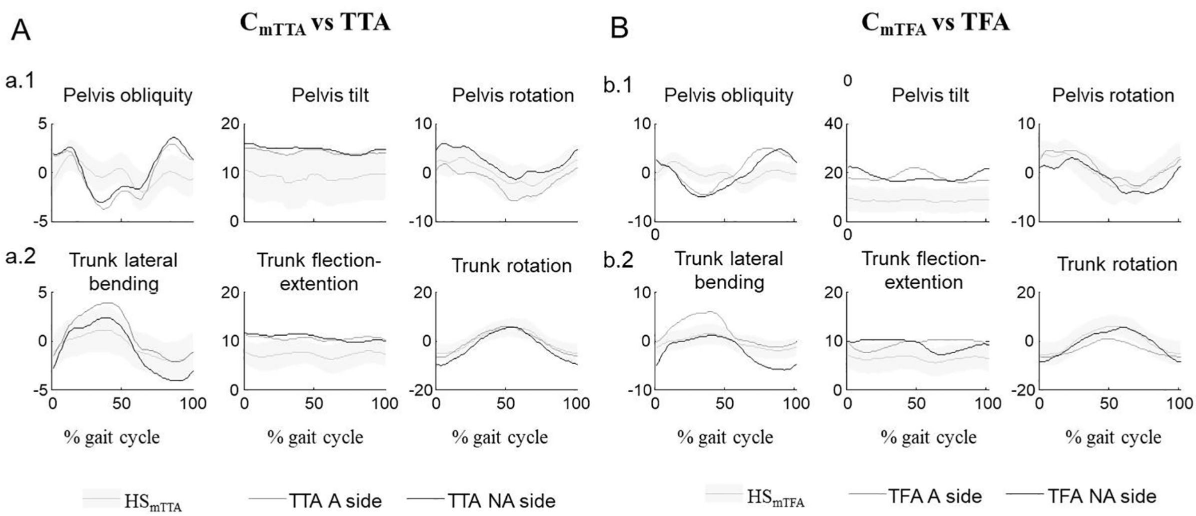

Fig. 2. Pelvic obliquity, pelvic tilt, pelvic rotation, trunk lateral bending, trunk flection-extension, and trunk rotation in both sides were found in both TTA and TFA groups compared to controls. C_{mTTA}: healthy subjects age-sex-speed matched with TTA; TTA: subjects with transtibial amputation; C_{mTFA}: healthy subjects age-sex-speed matched with TFA; TFA: subjects with transfemoral amputation.

Table 2
The means, standard deviations, and statistical results (p value) of parameters evaluated on vertical force (VF) curves ($Peak1_{VF}$ and $Peak2_{VF}$: 2 peaks, CoA_{VF}: center of activity and $FWHM_{VF}$: full width at half maximum).

		People with amputation vs controls					
Kinetic parameters		C_{mTTA}	TTA	p group	C_{mTFA}	TFA	p group
$Peak1_{VF}$	A)	1.014 ± 0.061	1.049 ± 0.083	0.125	0.973 ± 0.114	1.04 ± 0.064	< **0.001**
	NA)		1.109 ± 0.10	**0.005**		1.016 ± 0.062	**0.1**
	p side		**0.029**			**0.011**	
$Peak2_{VF}$	A)	1.054 ± 0.051	0.985 ± 0.032	< **0.001**	1.00 ± 0.124	0.989 ± 0.048	0.355
	NA)		1.063 ± 0.079	0.733		0.992 ± 0.071	0.379
	p side		< **0.001**			**0.008**	
CoA_{VF} (% gait cicle)	A)	31.25 ± 1.599	30.87 ± 1.68	0.654	31.43 ± 1.77	30.09 ± 1.67	< **0.001**
	NA)		31.40 ± 2.45	0.475		33.57 ± 1.74	< **0.001**
	p side		0.281			< **0.001**	
$FWHM_{VF}$	A)	50 ± 1.604	47.667 ± 3.457	0.066	47.179 ± 1.798	44.051 ± 3.809	< **0.001**
	NA)		50.47 ± 2.446	0.652		56.72 ± 2.611	< **0.001**
	p side		0.076			< **0.001**	

	Type of prostheses				TTA_m vs TFA_m		
Kinetic parameters	TFA_M	TFA_C	TFA_G	p group	TTA_m	TFA_m	p group
$Peak1_{VF}$	1.065 ± 0.068	1.052 ± 0.051	1.03 ± 0.075	0.423	1.021 ± 0.039	1.043 ± 0.071	0.336
	1.002 ± 0.056	1.008 ± 0.043	1.03 ± 0.083	0.421	1.094 ± 0.103	1.001 ± 0.07	**0.014**
	0.098	**0.013**	0.883		**0.02**	0.1	
$Peak2_{VF}$	0.997 ± 0.054	0.986 ± 0.043	0.987 ± 0.054	0.838	0.982 ± 0.032	0.976 ± 0.046	0.538
	0.999 ± 0.075	1.042 ± 0.066	1.011 ± 0.074	0.288	1.0584 ± 0.083	1.007 ± 0.063	0.084
	0.82	**0.001**	**0.288**		**0.002**	0.168	
CoA_{VF} (% gait cicle)	29.98 ± 1.62	30.38 ± 1.99	29.84 ± 1.31	0.666	30.81 ± 1.78	29.60 ± 1.56	0.091
	33.73 ± 1.86	34.08 ± 1.72	32.85 ± 1.56	0.142	31.86 ± 2.28	33.57 ± 1.62	0.095
	< **0.001**	< **0.001**	< **0.001**		0.07	< **0.001**	
$FWHM_{VF}$	43.222 ± 3.961	43.941 ± 4.322	44.712 ± 3.148	0.659	47.153 ± 3.412	43.231 ± 3.678	**0.01**
	57.11 ± 2.315	56.18 ± 3.005	57.12 ± 2.318	0.542	50.85 ± 2.267	57.46 ± 2.696	< **0.001**
	< **0.001**	< **0.001**	< **0.001**		**0.03**	< **0.001**	

C_{mTTA}: healthy subjects age-sex-speed matched with TTA; TTA: subjects with transtibial amputation; C_{mTFA}: healthy subjects age-sex-speed matched with TFA; TFA: subjects with transfemoral amputation; TFA_M: subjects with transfemoral amputation with mechanical prosthesis; TFA_C: subjects with transfemoral amputation with CLeg prosthesis; TFA_G: subjects with transfemoral amputation with Genium prosthesis; TFA_m: a subgroup of 13 age-sex-speed matched subjects with a subgroup of TTA; TTA_m: a subgroup of 13 age-sex-speed matched subjects with a subgroup of TFA.

(Fig. 1). Furthermore, a significantly decreased hip RoM in A side than in NA side (Fig. 1) was detected in TFA_G subgroup.

3.2.3. TTA_m vs. TFA_m

A significant effect of the type of amputation (TTA_m vs. TFA_m) on the knee, pelvic tilt, and trunk flexion-extension RoMs was found, with the knee RoM significantly decreased in the A side in TFA_m subgroup compared to TTA_m subgroup (Fig. 1) and the pelvic tilt and trunk flexion-extension RoMs significantly increased in both sides in TFA_m subgroup compared to TTA_m subgroup (Fig. 1).

A significantly shorter hip and knee RoMs were found in the A side than in the NA side (Fig. 1) in TFA_m subgroup. Furthermore, a significantly shorter ankle RoM was found in the A side than in the NA side (Fig. 1) in both TTA_m and TFA_m subgroups. Significantly shorter trunk lateral bending, trunk flexion-extension, and trunk rotation RoMs were found in the A side than in the NA side (Fig. 1) in both TTA_m and TFA_m subgroups.

3.3. Kinetic data

The curves of the vertical force for C_{mTTA} and TTA (Fig. 3A) and for C_{mTFA} and TFA (Fig. 3B) are shown in Cartesian coordinates as mean curves (a.1 and b.1), and in polar coordinates as mean curve (a.2 and b.2), as well as single and mean CoA values (a.3 and b.3), all expressed as percentage of gait cycle.

The means, standard deviations, and statistical results of VF for each group are reported in Table 3.

3.3.1. People with amputation vs. controls

A significantly increased $Peak1_{VF}$ value in NA side and a significantly decreased $Peak2_{VF}$ value in A side were found in TTA group compared to C_{mTTA} group (Table 2). Significantly increased $Peak1_{VF}$, CoA_{VF}, and $FWHM_{VF}$ values in NA side were found in TFA compared to C_{mTFA} (Table 2). Furthermore, significantly increased $Peak1_{VF}$ value and significantly decreased $FWHM_{VF}$ and CoA_{VF} values in A side were found in TFA compared to C_{mTFA} (Table 2).

$Peak1_{VF}$ was significantly lower in the A side than in the NA side (Table 2) in TTA group. A significantly lower $Peak2_{VF}$ value was found in the A side than in the NA side (Table 2) in both TTA and TFA groups. Furthermore, significantly higher $Peak1_{VF}$ and lower $FWHM_{VF}$ and CoA_{VF} values were found in the A side than in the NA side (Table 2) in TFA.

3.3.2. Type of prostheses (TFA_M, TFA_C, and TFA_G)

No significant effects of the type of prosthesis on the VF values were detected for both sides (Table 2).

A significantly increased $Peak1_{VF}$ value was found in A side than in the NA side (Table 2) in TFA_C and significantly decreased ($p < 0.05$) $Peak2_{VF}$ values were found in A side than in the NA side (Table 2) in TFA_C and in TFA_G subgroups. Furthermore, significantly decreased $FWHM_{VF}$ and CoA_{VF} values were found in A side than in the NA side (Table 2) in all three TFA_M, TFA_C, and TFA_G subgroups.

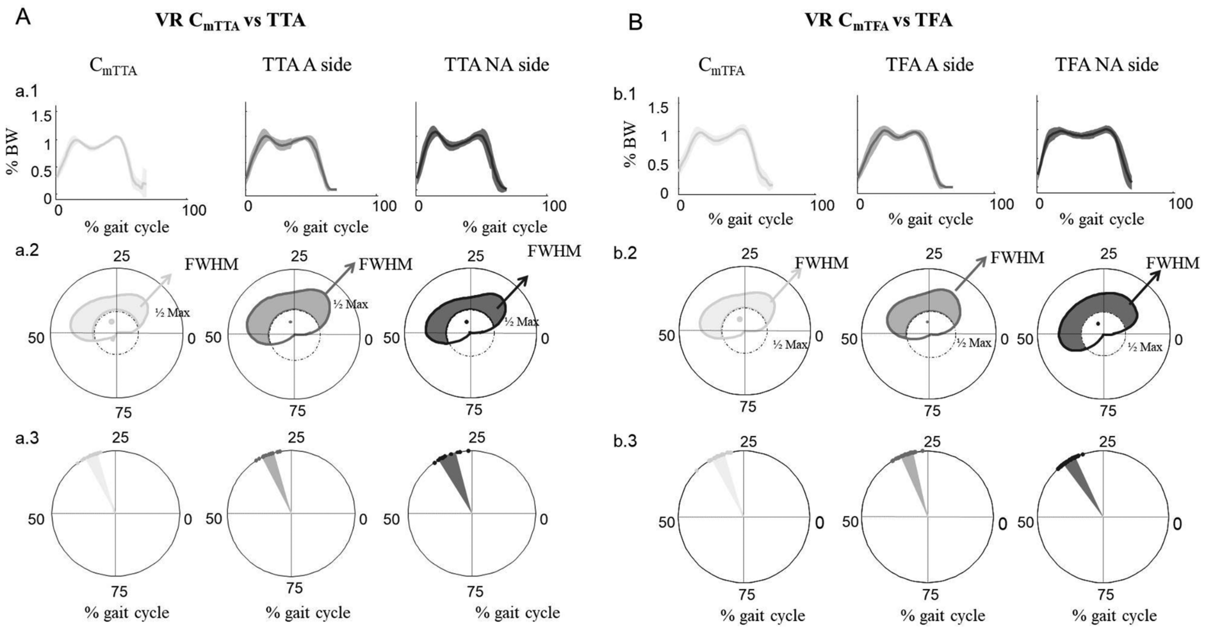

Fig. 3. Curves of the vertical force for C_{mTTA} and TTA and for C_{mTFA} and TFA shown in Cartesian coordinates as mean curves (a.1 and b.1), and in polar coordinates as mean curve (a.2 and b.2), as well as single and mean CoA values (a.3 and b.3), all expressed as percentage of gait cycle. C_{mTTA}: healthy subjects age-sex-speed matched with TTA; TTA: subjects with transtibial amputation; C_{mTFA}: healthy subjects age-sex-speed matched with TFA; TFA: subjects with transfemoral amputation.

Table 3
Common and specific gait patterns in people with amputation.

Gait parameters	Common gait pattern* TTA and TFA compared to controls			Specific gait pattern (amputation level) TFA compared to both control and TTA groups			Specific gait pattern (type of prosthesis) TFA_G vs TFA_M		
	Symmetric	Asymmetric		Symmetric	Asymmetric		Symmetric	Asymmetric	
		A side	NA side		A side	NA side		A side	NA side
Time-distance	Increased step width Increased step length Increased double support duration		Increased stance duration Reduced swing duration		Decreased stance duration Increased swing duration				Increased step length in TFA_G compared to TFA_M
Kinematics	Increased pelvic obliquity Increased trunk lateral bending and rotation Increased ante-flexion of the pelvis and trunk	Decreased ankle RoM				Increased hip and knee RoM	Increased pelvic obliquity in TFA_M compared to TFA_G	Increased hip e knee RoMs in TFA_G compared to TFA_M	
Kinetics			Increased Peak1		Increased Peak1 Decreased CoA, FWHM	Increased CoA and FWHM			
Energetic				Reduced Rstep					

* Irrespective of the amputation level.

3.3.3. TTA_m vs. TFA_m

A significant effect of the type of amputation (TTA_m vs. TFA_m) on the VF values was detected. $Peak1_{VF}$ was significantly decreased in the NA side in TFA_m group compared to TTA_m group (Table 2). $FWHM_{VF}$ was significantly increased in the NA side and significantly decreased in the A side in TFA_m group compared to TTA_m group (Table 2).

Significantly lower values were found in the A side than in the NA side for $Peak1_{VF}$ and $Peak2_{VF}$ in TTA_m, for $FWHM_{VF}$ in both TTA_m and TFA_m group, and for CoA_{VF} in TFA_m (Table 2).

3.4. Energy consumption measurement

Fig. 4 shows means, standard deviations, and statistical results of R-step and TEC values.

3.4.1. People with amputation vs controls

A significantly lower value of R-step in TFA subgroup compared to C_{mTFA} subgroup was found (Fig. 4). No significant differences of TEC values were detected.

3.4.2. Type of prostheses (TFA_M, TFA_C and TFA_G)

No significant effects of the type of prosthesis were found on both R-step and TEC (Fig. 4).

3.4.3. TTA_m vs. TFA_m

A significant effect of the type of amputation on R-step was found, with R-step value of TTA_m subgroup being significantly higher than that of TFA_m. Instead, no significant effect of the type of amputation on TEC was detected (Fig. 4).

4. Discussion

This study aimed to identify both common and specific gait patterns in people with amputation, either regardless of, or according to their level of amputation and the type of prosthetic component. Furthermore, this study was focused on the symmetric and asymmetric aspects of these patterns.

To have a global picture of all the gait deficits for both the common and the specific gait patterns, the data were summarized in Table 3.

In general, regardless of the level of amputation and type of prosthesis, subjects with TTA and with TFA showed a common gait pattern characterized by a symmetric increase of step length, step width, double support duration, pelvic obliquity, trunk lateral bending, and trunk rotation range of motions with increased pelvis and trunk ante-flexed (flexed in a forward direction) posture. Almost all these gait deficits reflect compensatory mechanisms adopted by people with amputation presumably to increase their stability in the frontal plane (increased step width), to maintain the most stable configuration (increased double support duration), to assist the lift of the affected limb (increased trunk lateral bending), while increasing the time of the stance and the force production during weight acceptance in the unaffected limb. Conversely, the reduced ankle joint range of motion in the prosthetic limb, which is the common prosthetic joint in both subjects with TTA and with TFA, is directly linked to the use of the prosthesis.

The increased step length is likely related to a compensatory increased movement of the trunk and pelvis (Fig. 1), since both the

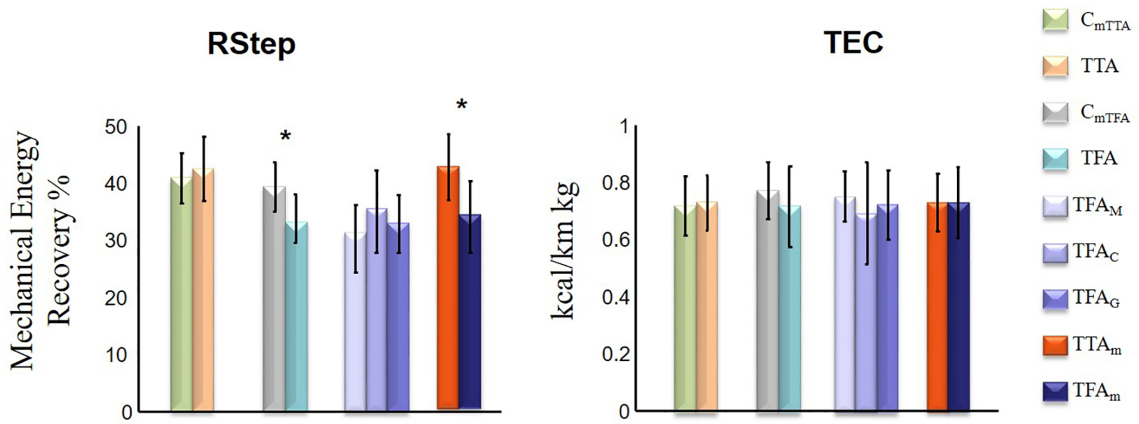

Fig. 4. Means, standard deviations, and statistical results of fraction of mechanical energy recovered during each walking step (R-step) and total energy consumption (TEC) values for each group. C_{mTTA}: healthy subjects age-sex-speed matched with TTA; TTA: subjects with transtibial amputation; C_{mTFA}: healthy subjects age-sex-speed matched with TFA; TFA: subjects with transfemoral amputation; TFA_M: subjects with transfemoral amputation with mechanical prosthesis; TFA_C: subjects with transfemoral amputation with CLeg prosthesis; TFA_G: subjects with transfemoral amputation with Genium prosthesis; TFA_m: a subgroup of 13 age-sex-speed matched subjects with a subgroup of TTA; TTA_m: a subgroup of 13 age-sex-speed matched subjects with a subgroup of TFA.

knee and hip joint range of motions of the prosthetic limb were either not increased, in the TTA group, or even reduced, in TFA group. However, it is not possible to exclude that the lack of sensory feedback (Fan et al., 2008) might have played a role in determining a hypermetric foot placement in the prosthetic limb, which, in turn, would have influenced the foot placement of the unaffected limb, as adaptive mechanism of the new support base schema (Head & Holmes, 1911; Ivanenko et al., 2011).

The subjects with TFA showed a specific gait pattern that differed from that of C and subjects with TTA in terms of kinematic, kinetic, and energetic behavior. The subjects with TFA reduced the duration of the stance and increased the duration of the swing in the prosthetic limb. Moreover, they increased the hip and knee joint range of motions in the unaffected limb. Interestingly, in the prosthetic limb of subjects with TFA, the Peak1 was increased, while the full width at half maximum was reduced. In general, the Peak and full width at half maximum parameters express two different spatio-temporal aspects of the force production. The first represents the maximal force produced in a given instant during the loading response subphase, while the second represents the amount of the force production (> 50% of the maximum) maintained during the whole duration of the stance. In this view, the subjects with TFA seem to be unable to control the prosthetic limb during the heel strike, likely caused by a reduced deceleration of the prosthetic limb from the late swing to the initial contact, leading to an increase in the Peak1. At the same time, they are unable to produce and maintain an adequate force during the whole stance phase, leading to a decrease of the full width at half maximum. Such behavior is reflected by the shift of the center of activity toward the initial contact event (initial part of the stance). Conversely, the full width at half maximum was increased in the unaffected limb, which compensated by producing a stronger force maintained for a longer time, determining, in this case, a shift of the center of activity toward the toe off event (final part of the stance). Altogether these findings deeply reflect the essence of the asymmetric gait (Iosa et al., 2014) revealing the greater effort achieved by subjects with TFA to compensate for the reduced motor performance by increasing both motion and force production in the unaffected limb. As a final result, this specific gait pattern makes the subjects with TFA unable to recover energy during the stance phase (Fig. 4).

People with amputation with Genium prosthesis (TFA_G) showed a longer step length in NA side and increased hip and knee range of motions in the prosthetic side compared to subjects with mechanical prosthesis, who, conversely, showed a symmetric increased pelvic obliquity. These findings indicate that the type of prosthesis influences the gait pattern of people with amputation both in terms of gait performance and adaptation (Highsmith et al., 2016). In this view, the increased step length and hip and knee ranges of motion, together with the trend of gait speed (Table 1), might reflect a better gait performance for the Genium vs mechanical prostheses. Conversely, the increased pelvic obliquity seems to reflect a greater compensatory effort in subjects with mechanical prostheses, likely aimed to lift the limbs during the gait progression.

In conclusion, in spite of common gait pattern in subjects with lower limb prostheses, both the anatomical level of amputation and type of prostheses determine a specific gait pattern that should be taken into account when developing new and ergonomic prosthetic devices and when planning the rehabilitation programs aimed at improving the physiology of gait and reducing the gait asymmetries.

Acknowledgments

This research was funded by INAIL, “Bando Ricerche in Collaborazione 2016”.

References

Bateni, H., & Olney, S. J. (2002). Kinematic and kinetic variations of below-knee amputee gait. *JPO Journal of Prosthetics and Orthotics, 14*(1), 2–10. https://doi.org/10.1097/00008526-200203000-00003.

Cavagna, G. A., Thys, H., & Zamboni, A. (1976). The sources of external work in level walking and running. Retrieved from *The Journal of Physiology, 262*(3), 639–657. http://www.ncbi.nlm.nih.gov/pubmed/1011078.

Cavagna, G. A., Willems, P. A., Legramandi, M. A., & Heglund, N. C. (2002). Pendular energy transduction within the step in human walking. Retrieved from *The Journal of Experimental Biology, 205*(Pt 21), 3413–3422. http://www.ncbi.nlm.nih.gov/pubmed/12324550.

Chitragari, G., Mahler, D. B., Sumpio, B. J., Blume, P. A., & Sumpio, B. E. (2014). Prosthetic options available for the diabetic lower limb amputee. *Clinics in Podiatric Medicine and Surgery, 31*(1), 173–185. https://doi.org/10.1016/j.cpm.2013.09.008.

Cutti, A. G., Lettieri, E., Del Maestro, M., Radaelli, G., Luchetti, M., Verni, G., & Masella, C. (2017). Stratified cost-utility analysis of C-Leg versus mechanical knees: Findings from an Italian sample of transfemoral amputees. *Prosthetics and Orthotics International, 41*(3), 227–236. https://doi.org/10.1177/0309364616637955.

Davis, R. B., Õunpuu, S., Tyburski, D., & Gage, J. R. (1991). A gait analysis data collection and reduction technique. *Human Movement Science, 10*(5), 575–587. https://doi.org/10.1016/0167-9457(91)90046-Z.

de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. Retrieved from *Journal of Biomechanics, 29*(9), 1223–1230. http://www.ncbi.nlm.nih.gov/pubmed/8872282.

Detrembleur, C., Dierick, F., Stoquart, G., Chantraine, F., & Lejeune, T. (2003). Energy cost, mechanical work, and efficiency of hemiparetic walking. Retrieved from *Gait & Posture, 18*(2), 47–55. http://www.ncbi.nlm.nih.gov/pubmed/14654207.

Don, R., Serrao, M., Vinci, P., Ranavolo, A., Cacchio, A., Ioppolo, F., ... Santilli, V. (2007). Foot drop and plantar flexion failure determine different gait strategies in Charcot-Marie-Tooth patients. *Clinical Biomechanics, 22*(8), 905–916. https://doi.org/10.1016/j.clinbiomech.2007.06.002.

Esquenazi, A. (2014). Gait analysis in lower-limb amputation and prosthetic rehabilitation. *Physical Medicine and Rehabilitation Clinics of North America, 25*(1), 153–167. https://doi.org/10.1016/j.pmr.2013.09.006.

Fan, R. E., Culjat, M. O., King, C.-H., Franco, M. L., Sedrak, M., Bisley, J. W., ... Grundfest, W. S. (2008). A prototype haptic feedback system for lower-limb prostheses and sensory neuropathy. Retrieved from *Studies in Health Technology and Informatics, 132*, 115–119. http://www.ncbi.nlm.nih.gov/pubmed/18391269.

Fradet, L., Alimusaj, M., Braatz, F., & Wolf, S. I. (2010). Biomechanical analysis of ramp ambulation of transtibial amputees with an adaptive ankle foot system. *Gait & Posture, 32*(2), 191–198. https://doi.org/10.1016/j.gaitpost.2010.04.011.

Head, H., & Holmes, G. (1911). Sensory disturbances from cerebral lesions. *Brain, 34*(2–3), 102–254. https://doi.org/10.1093/brain/34.2-3.102.

Highsmith, M. J., Kahle, J. T., Carey, S. L., Lura, D. J., Dubey, R. V., Csavina, K. R., & Quillen, W. S. (2011). Kinetic asymmetry in transfemoral amputees while performing sit to stand and stand to sit movements. *Gait & Posture, 34*(1), 86–91. https://doi.org/10.1016/J.GAITPOST.2011.03.018.

Highsmith, M. J., Kahle, J. T., Miro, R. M., Cress, M. E., Lura, D. J., Quillen, W. S., ... Mengelkoch, L. J. (2016). Functional performance differences between the Genium and C-Leg prosthetic knees and intact knees. *Journal of Rehabilitation Research and Development, 53*(6), 753–766. https://doi.org/10.1682/JRRD.2014.06.0149.

Howcroft, J., Lemaire, E. D., Kofman, J., & Kendell, C. (2015). Understanding dynamic stability from pelvis accelerometer data and the relationship to balance and

mobility in transtibial amputees. *Gait & Posture, 41*(3), 808–812. https://doi.org/10.1016/J.GAITPOST.2015.03.001.
Iosa, M., Paradisi, F., Brunelli, S., Delussu, A. S., Pellegrini, R., Zenardi, D., ... Traballesi, M. (2014). Assessment of gait stability, harmony, and symmetry in subjects with lower-limb amputation evaluated by trunk accelerations. *Journal of Rehabilitation Research and Development, 51*(4), 623–634. https://doi.org/10.1682/JRRD.2013.07.0162.
Ivanenko, Y. P., Dominici, N., Daprati, E., Nico, D., Cappellini, G., & Lacquaniti, F. (2011). Locomotor body scheme. *Human Movement Science, 30*(2), 341–351. https://doi.org/10.1016/J.HUMOV.2010.04.001.
Kannenberg, A., Zacharias, B., Mileusnic, M., & Seyr, M. (2013). Activities of daily living. *JPO Journal of Prosthetics and Orthotics, 25*(3), 110–117. https://doi.org/10.1097/JPO.0b013e31829c221f.
Kluge, F., Krinner, S., Lochmann, M., & Eskofier, B. M. (2017). Speed dependent effects of laterally wedged insoles on gait biomechanics in healthy subjects. *Gait & Posture, 55*, 145–149. https://doi.org/10.1016/j.gaitpost.2017.04.012.
Nolan, L., Wit, A., Dudziñski, K., Lees, A., Lake, M., & Wychowañski, M. (2003). Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. Retrieved from *Gait & Posture, 17*(2), 142–151. http://www.ncbi.nlm.nih.gov/pubmed/12633775.
Ranavolo, A., Varrecchia, T., Rinaldi, M., Silvetti, A., Serrao, M., Conforto, S., & Draicchio, F. (2017). Mechanical lifting energy consumption in work activities designed by means of the "revised NIOSH lifting equation". *Industrial Health, 55*(5), https://doi.org/10.2486/indhealth.2017-0075.
Rossi, S. A., Doyle, W., & Skinner, H. B. (1995). Gait initiation of persons with below-knee amputation: The characterization and comparison of force profiles. Retrieved from *Journal of Rehabilitation Research and Development, 32*(2), 120–127. http://www.ncbi.nlm.nih.gov/pubmed/7562651.
Sanders, J. E., Bell, D. M., Okumura, R. M., & Dralle, A. J. (1998). Effects of alignment changes on stance phase pressures and shear stresses on transtibial amputees: Measurements from 13 transducer sites. Retrieved from *IEEE Transactions on Rehabilitation Engineering: A Publication of the IEEE Engineering in Medicine and Biology Society, 6*(1), 21–31. http://www.ncbi.nlm.nih.gov/pubmed/9535520.
Schafer, Z. A., Perry, J. L., & Vanicek, N. (2018). A personalised exercise programme for individuals with lower limb amputation reduces falls and improves gait biomechanics: A block randomised controlled trial. *Gait & Posture, 63*, 282–289. https://doi.org/10.1016/j.gaitpost.2018.04.030.
Segal, A. D., Orendurff, M. S., Czerniecki, J. M., Schoen, J., & Klute, G. K. (2011). Comparison of transtibial amputee and non-amputee biomechanics during a common turning task. *Gait & Posture, 33*(1), 41–47. https://doi.org/10.1016/j.gaitpost.2010.09.021.
Serrao, M., Rinaldi, M., Ranavolo, A., Lacquaniti, F., Martino, G., Leonardi, L., & Pierelli, F. (2016). Gait patterns in patients with hereditary spastic paraparesis. *PLoS ONE, 11*(10), https://doi.org/10.1371/journal.pone.0164623.
Shell, C. E., Segal, A. D., Klute, G. K., & Neptune, R. R. (2017). The effects of prosthetic foot stiffness on transtibial amputee walking mechanics and balance control during turning. *Clinical Biomechanics, 49*, 56–63. https://doi.org/10.1016/j.clinbiomech.2017.08.003.
Smidt, G. L. (1990). *Gait in Rehabilitation.* Churchill Livingstone.
Stoquart, G., Detrembleur, C., & Lejeune, T. (2008). Effect of speed on kinematic, kinetic, electromyographic and energetic reference values during treadmill walking. *Neurophysiologie Clinique = Clinical Neurophysiology, 38*(2), 105–116. https://doi.org/10.1016/j.neucli.2008.02.002.
Sturdy, J., Gates, D. H., Darter, B. J., & Wilken, J. M. (2014). Assessing preparative gait adaptations in persons with transtibial amputation in response to repeated medial-lateral perturbations. *Gait & Posture, 39*(3), 995–998. https://doi.org/10.1016/J.GAITPOST.2013.12.006.
Sturk, J. A., Lemaire, E. D., Sinitski, E. H., Dudek, N. L., Besemann, M., Hebert, J. S., & Baddour, N. (2017). Maintaining stable transfemoral amputee gait on level, sloped and simulated uneven conditions in a virtual environment. *Disability and Rehabilitation. Assistive Technology, 1–10*. https://doi.org/10.1080/17483107.2017.1420250.
Tura, A., Rocchi, L., Raggi, M., Cutti, A. G., & Chiari, L. (2012). Recommended number of strides for automatic assessment of gait symmetry and regularity in above-knee amputees by means of accelerometry and autocorrelation analysis. *Journal of NeuroEngineering and Rehabilitation, 9*(1), 11. https://doi.org/10.1186/1743-0003-9-11.
Varrecchia, T., Rinaldi, M., Serrao, M., Draicchio, F., Conte, C., Conforto, S., ... Ranavolo, A. (2018). Global lower limb muscle coactivation during walking at different speeds: Relationship between spatio-temporal, kinematic, kinetic, and energetic parameters. *Journal of Electromyography and Kinesiology, 43*, 148–157. https://doi.org/10.1016/j.jelekin.2018.09.012.
Ventura, J. D., Segal, A. D., Klute, G. K., & Neptune, R. R. (2011). Compensatory mechanisms of transtibial amputees during circular turning. *Gait & Posture, 34*(3), 307–312. https://doi.org/10.1016/j.gaitpost.2011.05.014.
Vrieling, A. H., van Keeken, H. G., Schoppen, T., Otten, E., Halbertsma, J. P. K., Hof, A. L., & Postema, K. (2008). Gait initiation in lower limb amputees. *Gait & Posture, 27*(3), 423–430. https://doi.org/10.1016/j.gaitpost.2007.05.013.
Wu, G., van der Helm, F. C. T., Veeger, H. E. J. D., Makhsous, M., Van Roy, P., Anglin, C., ... International Society of Biomechanics (2005). ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion – Part II: Shoulder, elbow, wrist and hand. *Journal of Biomechanics, 38*(5), 981–992 Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/15844264.
Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., ... Standardization and Terminology Committee of the International Society of Biomechanics (2002). ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—Part I: Ankle, hip, and spine. *Journal of Biomechanics, 35*(4), 543–548. https://doi.org/10.1016/S0021-9290(01)00222-6.
Zatsiorsky, V., Seluyanov, V., & Chugunova, L. (1990). Methods of determining mass-inertial characteristics of human body segments. *Contemporary Problems of Biomechanics* Retrieved from https://ci.nii.ac.jp/naid/10022345842/.