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The gait initiation process in unilateral lower-limb amputees when stepping up and stepping down to a new level

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Abstract

Background. Unilateral lower-limb amputees lead with their intact limb when stepping up and with their prosthesis when stepping down; the gait initiation process for the different stepping directions has not previously been investigated.

Methods. Ten unilateral amputees (5 transfemoral and 5 transtibial) and 8 able-bodied controls performed single steps up and single steps down to a new level (73 and 219 mm). Duration, a-p and m-l centre of mass and centre of pressure peak displacements and centre of mass peak velocity of the anticipatory postural adjustment and step execution phase were evaluated for each stepping direction by analysing data collected using a Vicon 3D motion analysis system.

Findings. There were significant differences (in the phase duration, peak a-p and m-l centre of pressure displacement and peak a-p and m-l centre of mass velocity at heel-off and at foot-contact) between both amputee sub-groups and controls ($P < 0.05$), but not between amputee sub-groups. These group differences were mainly a result of amputees adopting a different gait initiation strategy for each stepping direction.

Interpretation. Findings indicate the gait initiation process utilised by lower-limb amputees was dependent on the direction of stepping and more particularly by which limb the amputee led with; this suggests that the balance and postural control of gait initiation is not governed by a fixed motor program, and thus that becoming an amputee will require time and training to develop alternative neuromuscular control and coordination strategies. These findings should be considered when developing training/rehabilitation programs.

Keywords: Lower-limb amputation; Amputee; Gait initiation; Stepping; Locomotion; Centre of mass; Centre of pressure

1. Introduction

The transition from upright standing to stepping or walking is a common locomotor task. Such transitions, commonly referred to as gait initiations, involve complex and synergist neuromuscular control and coordination (Dichgans and Diener, 1989; Paulus et al., 1984). In attempting to understand the balance and postural control strategies used in executing a gait initiation, previous research has determined the relationship between the horizontal trajectories of the body centre of mass (CoM) and the point of application of the ground reaction force vector (the so-called centre of pressure, CoP). The gait initiation process in healthy able-bodied individuals is characterised by stereotypical displacements of the CoM and CoP that result from a consistent pattern of lower limb muscle activity. For example, during the early part of initiation, from movement initiation up

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to swing limb heel-off (the so-called period of anticipatory postural adjustment, Gélat and Brenière, 2000), ankle plantar- and dorsi-flexor activity effects a backward and towards the intended swing limb displacement of the CoP, and an accompanying forward and towards the intended stance limb displacement of the CoM (Brenière et al., 1981; Crenna and Frigo, 1991; Elble et al., 1994; Jian et al., 1993). During the subsequent step execution phase (from swing limb heel-off up to swing limb foot-contact, Gélat and Brenière, 2000), swing and stance limb hip abductor muscle activity cause the CoP to move rapidly across to the heel region of the stance limb, while the CoM continues to move forwards (Jian et al., 1993). Following swing limb toe-off, stance limb ankle plantar–flexor activity helps control the forward progression of the CoP beneath the stance foot (Jian et al., 1993). However, while much is known about the gait initiation process used by able-bodied individuals, there is a paucity of information regarding how amputees initiate gait, particularly so when stepping to a new level. The present study was conducted to address this lack of knowledge.

Muscular activity of the lower limbs plays an important role in controlling the displacement trajectories of the CoM and CoP during the gait initiation process. Thus it follows that, because feedback originating from cutaneous receptors within the foot and leg is only present unilaterally (Geurts et al., 1992; Isakov et al., 1992), or at best is available via re-mapped receptors within the stump (Schwenkreis et al., 2003), individuals with a unilateral lower limb amputation will need to adopt alternative, or new, neuromuscular control and coordination strategies.

It has been reported that both transtibial and transfemoral amputees take significantly longer to load their stance limb when initiating gait leading with their intact limb compared to when leading with their prosthetic limb (Michel and Do, 2002; Rossi et al., 1995; Tokuno et al., 2003). This temporal difference was suggested to be a strategy aimed at counteracting the stability and propulsion limitations of the prosthetic (stance) limb (Michel and Do, 2002; Rossi et al., 1995; Tokuno et al., 2003). In two of these studies the horizontal trajectory of the CoP was found to be similar to that seen in able-bodied controls, regardless of the limb with which the amputee subjects led with (Michel and Do, 2002; Rossi et al., 1995). Conversely, the findings of Tokuno et al. (2003) indicated that, while the horizontal trajectory of the CoP during gait initiation when leading with the prosthetic limb was similar to that seen in able-bodied controls, when leading with the intact limb the lateral displacement of the CoP, from its early position towards the intended swing limb back towards the stance limb, also tended to shift anteriorly so that, rather than be located beneath the heel region of the stance foot at swing limb toe-off, it became located beneath the mid-foot or metatarsals head region (Tokuno et al., 2003). These findings indicate that the choice of lead limb will affect the temporal aspects of gait initiation in unilateral lower-limb amputees, but it is unclear the influence such a choice has on the balance and postural control adaptations adopted.

The gait initiation process in unilateral amputees when performing a single step up to a new level (leading with their intact limb) has also been investigated (Jones, 2004). Findings indicated that the initial displacement of the CoP towards the intended swing limb had greater posterior displacement and less lateral displacement in amputee subjects compared to able-bodied controls. In the subsequent step execution phase, the lateral displacement of the CoP back towards the stance limb, tended to move more posteriorly towards the heel region in able-bodied subjects, whereas in amputees it tended to move anteriorly towards the mid-foot region of the stance foot. It was suggested that this adaptation ensured the CoP was kept anterior of the knee joint centre, thereby ensuring the ground reaction force vector helped keep the knee of the prosthetic limb fully extended during the period when body weight was being supported solely by this limb (Jones, 2004).

Given that unilateral lower-limb amputees are taught, during rehabilitation and prosthetic training, to lead with their intact limb when stepping up to a new level, and with their prosthetic limb when stepping down to a new level, amputees may employ a different gait initiation process for each direction of stepping. The aim of the present study was to determine the balance and postural control adaptations adopted by unilateral lower-limb amputees when stepping up and when stepping down to a new level.

2. Methods

2.1. Subjects

Ten male unilateral lower-limb amputees (5 transfemoral and 5 transtibial) and 8 able-bodied controls (subject characteristics are listed in Table 1) participated in the study. All subjects self-reported they were currently moderately active (undertaking exercise bouts such as walking to and from local shops, gardening and engaging in household chores on a daily basis). Details of the prosthetic limb and socket used by each amputee subject are shown in Table 2. The study was performed in accordance with the Declaration of Helsinki and ethical approval for the study was obtained from the University of Bradford Research Strategy Committee. All subjects gave written informed consent to participate in the study. Amputee or able-body individuals with orthopaedic and/or neurological problems or diseases known to affect equilibrium (other than that of their primary pathology, in the case of the amputees) were excluded from the study.

Table 1
Subject characteristics

	Control (n = 8)			Transfemoral (n = 5)			Transtibial (n = 5)		
	Mean	SD	Range	Mean	SD	Range	Mean	SD	Range
Age (years)	33.7	11.8	25.0–61.6	42.1	18.2	22.8–67.8	48.9	5.6	40.2–55.8
Body mass (kg)	77.5	11.5	59.0–92.2	81.1	9.8	69.7–91.6	108.0	14.1	90.5–120.0
Height (cm)	175.3	5.7	166.5–187.0	175.0	7.6	164.0–183.0	180.0	9.4	169.0–193.2
Length of residuum (cm)	–	–	–	33.2	4.1	29.0–38.0	17.2	1.9	15.5–20.0
Time since amputation (years)	–	–	–	9.3	7.0	4.0–21.0	8.6	11.4	3.0–29.0
Time with current prosthesis (years)	–	–	–	1.8	1.6	0.5–4.0	1.1	0.9	0.5–2.5

Table 2
Prosthetic details for each amputee

Subject	Level of amputation	Prosthetic details
1	Transfemoral	Suction socket, 200 Series hydraulic knee, with C-Walk foot
2		Suction socket, 200 Series hydraulic knee, with Otto-Bock Torque Absorber and SACH foot
3		Iceross socket, Adaptive-Knee prosthesis, with Mercury Foot (integral TT Pylon)
4		Suction socket, Hi-activity CATECH swing-control knee, with Mercury Foot (integral TT Pylon)
5		Quad socket, Endolite Intelligent Knee Prosthesis, with TT Multiflex foot/ankle
6	Transtibial	Neoprene suspension socket, Total Shock Pylon, with Uni-Axle ankle
7		Shuttle lock suspension socket, ICON Shock Absorber prosthesis, with Variflex Foot
8		Shuttle lock suspension socket, Otto-Bock prosthesis, with Dynamic Foot
9		Cuff suspension PTB socket, Total Shock Pylon, with Seattle-Light Foot
10		Shuttle lock suspension socket, Seattle Flex-Pylon prosthesis, with Seattle-Light Foot

2.2. Data collection

Data collection was carried out using a Vicon 5-camera Motion Analysis System (Vicon 250; Oxford Metrics Ltd., Oxford, UK) and two AMTI force platforms (model OR6-7; Advanced Medical Technology, Inc., Boston, MA, USA). The two force platforms were positioned side by side; this enabled bilateral and simultaneous measurement of the ground reaction force for each limb to be undertaken. Kinematic and kinetic data were collected simultaneously at a sampling frequency of 50 and 100 Hz respectively. Subjects wore shorts and a t-shirt during testing and their usual 'everyday' footwear.

Thirty-one retro-reflective markers were attached, either directly to the skin or to clothing, to the following body locations; anterior-lateral and posterior-lateral aspects of the head, vertebrae C7 and C10, jugular notch, xiphoid process, acromions, lateral epicondyles of the elbows, lateral and medial styloid processes of the wrists, anterior superior iliac spines (ASIS), sacrum, lateral surface of the thighs and shanks, lateral femoral condyles, lateral maleoli, calcanei, second metatarsal heads. Markers were placed on corresponding sites on the prosthetic limb of each amputee subject.

Subjects completed repeated single steps up and single steps down to heights of 73 and 219 mm (low and high step respectively). These step heights approximated those that would be encountered when stepping up to or from a kerb (low step) and up to or from the upper limit set out for a household stair riser (high step). Each stepping platform was made of two separate blocks constructed from solid sections of medium density fibreboard with a surface area of 464×508 mm. The two blocks were positioned side-by-side to give a total surface (step) area of 928×508 mm. During data collection the two blocks (high or low) were either positioned directly in front of the two force platforms (stepping up) or directly over the two force platforms (stepping down). Amputee subjects completed four trials stepping up and four trials stepping down, at each of the step heights. However, because we assumed their movement pattern would be more consistent, able-bodied controls completed only two trials per step height and direction. The order of step height (low or high) and stepping direction (up or down) was randomised.

Subjects started from a stationary standing position with feet comfortably apart and placed separately on the two adjacent force-platforms (stepping up) or separately on the two blocks placed adjacently over the two force-platforms (stepping down). A verbal instruction ('step up', 'step down') was given to initiate each trial, and data collection began approximately 5 s prior to the verbal instruction and ended approximately 5 s after the movement had finished. Each subject performed a single practice step to each of the step heights and in each direction. Although not instructed to do so, all amputee subjects stepped down leading with their prosthesis and stepped up leading with their intact limb.

Using the Plug-In Gait software (Oxford Metrics Ltd.) 3-D marker trajectory data were filtered using the cross-validatory quintic spline smoothing routine and were then processed to define a 3-D linked-segment model of the subject. The markers, along with specific anthropometric measurements (subject height, mass, leg length, knee width, ankle width, inter-ASIS distance and ASIS-trochanter distance) were used to define a 3-D linked-segment model of the human body (Eames et al., 1999). Each body segment parameter was calculated using the anthropometric regression equations supplied by Dempster (1955). This approach is commonly used to determine the body segment parameters in able-bodied individuals, and because of the relative difficulty of accurately assessing the mass and CoM properties of the combined prosthetic limb and residuum, is also often used to estimate the mass characteristics of the prosthetic limb (e.g. Buckley, 2000; Miller, 1987; Miller and Verstraete, 1999; Winter and Sienko, 1988). Whole body CoM was calculated as the weighted sum of the centres of mass of all linked-segments.

Body CoM and heel marker co-ordinate data (50 Hz), along with the ground reaction force data and CoP data (50 Hz) from each force-platform, were exported in ASCII format for further analysis.

2.3. Data analysis

Using the following formula (Winter, 1995), force and CoP data from each force-platform were combined to provide global CoP co-ordinates:

$$\mathrm{CoP} = \mathrm{CoP}_\mathrm{l}\frac{Fz_\mathrm{l}}{Fz_\mathrm{l}+Fz_\mathrm{r}} + \mathrm{CoP}_\mathrm{r}\frac{Fz_\mathrm{r}}{Fz_\mathrm{l}+Fz_\mathrm{r}},$$

where CoP_l and CoP_r are the CoPs under the left and right feet respectively. Fz_l and Fz_r are the vertical reaction forces under the left and right feet respectively.

The stepping movement was divided into two phases (Gélat and Brenière, 2000): (i) anticipatory postural adjustment phase—from initiation of movement, defined as the instant when the CoP first moved beyond 10 mm laterally, up to the instant of swing limb heel-off, defined as the point at which the height of the swing limb heel marker increased by 1 mm for 3 consecutive frames, and (ii) step execution phase—from the instant of swing limb heel-off, up to the instant of swing limb foot-contact, defined as the instant when the swing limb toe trajectory ceased to travel downwards. This coincided with the instant the vertical ground reaction force on the stance limb dropped below one bodyweight.

The variables examined were: duration of the anticipatory postural adjustment phase, expressed as a percentage of the overall movement time (therefore, changes found also highlighted changes in the step execution phase duration), and m-l and a-p peak CoM and CoP displacements and peak CoM velocities. As foot placements and subject height were not standardised, m-l and a-p displacements were normalised to stance width and subject height respectively.

Data were analysed using a random effects population averaged model (Stata vers 7.0, Stat Corp., College Station, USA). This approach can loosely be thought of as an extension of the repeated measures ANOVA method (see Everitt, 2003). This multivariate model was obtained using the Generalised Least Squares (GLS) random-effects estimator, which produces a matrix-weighted average of between-subjects and within-subjects results. Given the experimental design an exchangeable correlation structure was judged to be appropriate, and due to the exploratory nature of the study no 'type I' error adjustment of the alpha level was deemed necessary. Thus, level of significance was set at $P < 0.05$. Factors of interest were incorporated sequentially and their statistical significance were tested using a likelihood ratio test. Factors with a P-value less than 0.1 were provisionally retained, whereas those above 0.1 were dropped. The final model adopted was the most parsimonious one that was felt to adequately explain the data. The P-values quoted in the text of the paper are those associated with the specific terms in the final regression model, which were:

1. Direction: a fixed factor with 2 levels (upwards and downwards).
2. Phase: a fixed factor with 2 levels (phase one and phase two).
3. Step Height: a fixed factor with 2 levels (low and high step).
4. Group: a fixed factor with three levels (transfemoral amputees, transtibial amputees, able-bodied controls).
5. Repetition: a fixed factor with 2 (able-bodied subjects) or 4 (amputee subjects) levels.

CoM and CoP trajectory data were also compared qualitatively. Using quintic spline multiple point interpolation (Biomechanics Toolbox, Infomedia Systems Ltd., UK) CoM and CoP trajectory data for the total movement time (i.e. initiation of movement up to swing limb foot-contact) were time normalised to 100 points for each subject. To reduce intra-group variability these CoM and CoP m-l and a-p displacements were again normalised to stance width and subject height respectively. 'Ensembled average' CoM and CoP trajectories for each subject group were then computed, and to highlight differences in the interaction of the CoM and CoP between each subject group these data were presented graphically (see Figs. 1 and 2).

3. Results

Figs. 1 and 2 show, respectively, the ensembled average CoP trajectories for the transfemoral amputee subjects versus controls and the transtibial amputee sub-

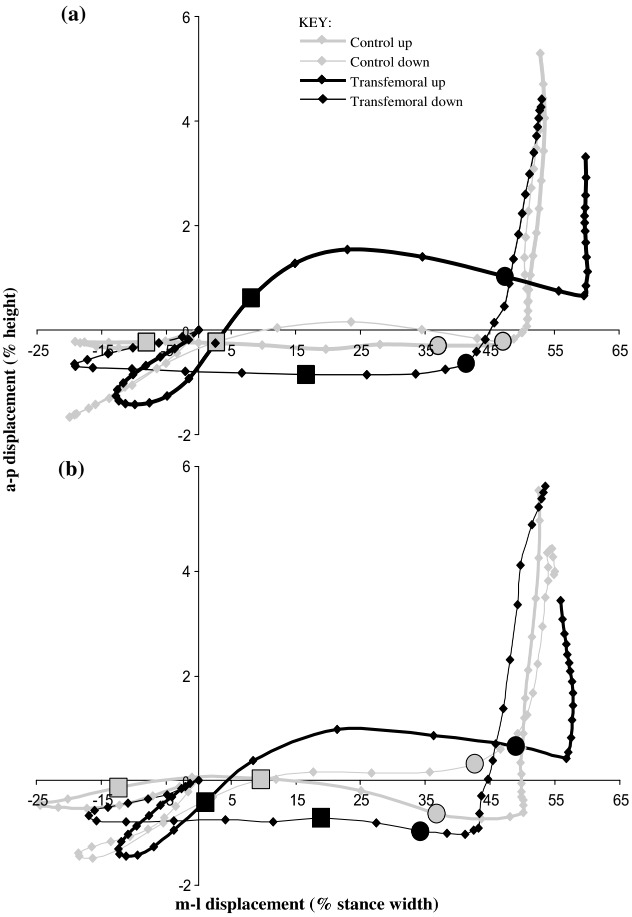

Fig. 1. 'Ensembled average' CoP trajectories for transfemoral amputees versus controls when stepping up to and down from the low (a) and high (b) step heights. Distance between points represents speed of movement (data point every 0.06 s). Data are shown from initiation of movement up to swing limb foot-contact. The zero reference represents the average CoP position when standing prior to movement initiation. □ and ○ indicates the instant of heel-off and toe-off of the swing limb respectively.

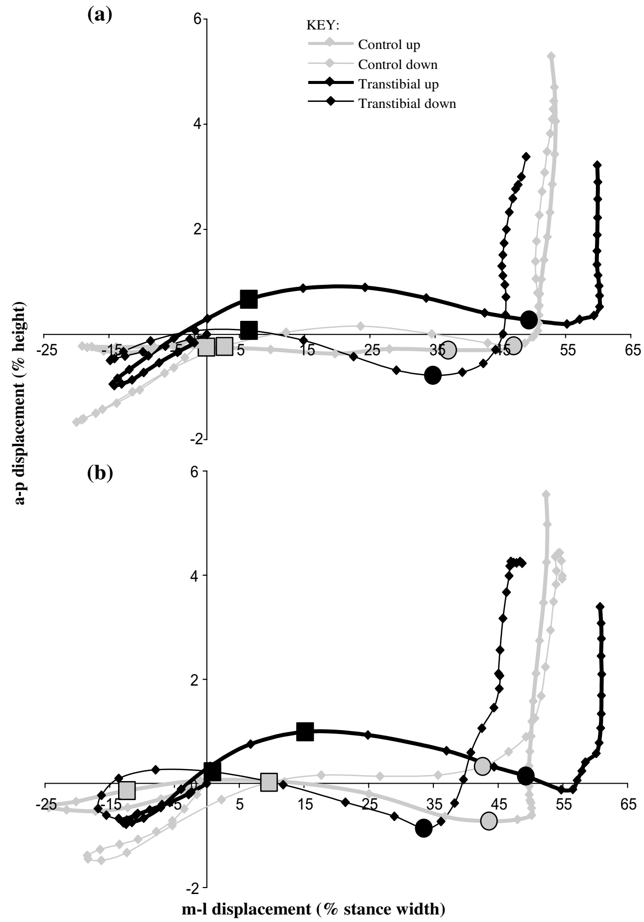

Fig. 2. 'Ensembled average' CoP trajectories for transtibial amputees versus controls when stepping up to and down from the low (a) and high (b) step heights. Distance between points represents speed of movement (data point every 0.06 s). Data are shown from initiation of movement up to swing limb foot-contact. The zero reference represents the average CoP position when standing prior to movement initiation. □ and ○ indicates the instant of heel-off and toe-off of the swing limb respectively.

jects versus controls, when stepping up to and down from each of the step heights. These figures indicate there were marked differences in the CoP trajectory between stepping up and stepping down for all subject groups during the anticipatory postural adjustment phase, and that these stepping direction differences became considerable for each amputee sub-group during the subsequent step execution phase. Table 3 presents the group mean temporal and kinematic parameters determined for each of the stepping phases. As the pattern in the data was consistent across the two step heights, to avoid repetition, the values shown are the average value across step height.

There was a significant group effect for many of the variables analysed ($P < 0.05$), which highlighted there were significant differences between transfemoral amputee subjects and controls and transtibial amputee subjects and controls, but not between amputee sub-groups. Group differences were highlighted further by significant group–direction interactions ($P < 0.05$ to $P < 0.001$) in all of the measured variables (except m-l CoM velocity). The following sections highlight intra-group differences between stepping up and stepping down.

3.1. Differences in anticipatory postural adjustment phase between stepping up and stepping down

The duration of the anticipatory postural adjustment phase (as a percentage of the overall movement time) when stepping up compared to when stepping down was significantly longer in both amputee sub-groups ($P < 0.001$) but was significantly shorter in the control group ($P < 0.001$). In both amputee sub-groups the posterior displacement of the CoP was significantly greater when stepping up compared to when stepping down

Table 3
Mean (SD) temporal and kinematic parameters averaged across step height when stepping up and stepping down

		Control		Transfemoral		Transtibial	
		Up	Down	Up	Down	Up	Down
APA phase	APA duration (% movement time)	30.7 (7.8)	39.1 (7.1)**	47.5 (4.3)**	37.0 (5.4)	42.2 (7.5)**	31.9 (8.2)
	Peak a-p CoP displacement (% height)	−0.8 (−0.5)	−1.8 (−0.7)**	−1.5 (−0.8)**	−1.0 (−0.5)	−1.1 (−0.5)*	−0.8 (−0.4)
	Peak m-l CoP displacement (% stance)	22.8 (7.9)	20.8 (8.5)	13.9 (4.4)**	20.1 (6.7)	16.3 (4.3)	17.9 (4.8)
	m-l CoM velocity at heel-off (mm s^{-1})	190.1 (56.8)**	178.1 (44.2)	159.9 (32.7)**	150.1 (44.6)	212.8 (34.3)**	175.2 (38.4)
	a-p CoM velocity at heel-off (mm s^{-1})	91.1 (31.7)	129.6 (37.4)**	112.0 (49.8)*	64.7 (14.3)	86.3 (37.7)*	54.5 (18.9)
SE phase	Peak m-l CoM displacement (% stance)	26.9 (3.4)*	23.6 (4.8)	23.2 (4.8)	24.0 (7.6)	31.0 (5.9)**	27.0 (4.5)
	m-l CoM velocity at foot-contact (mm s^{-1})	334.9 (73.9)**	301.4 (74.0)	303.9 (47.8)**	240.0 (46.2)	360.2 (45.5)**	318.8 (54.6)
	a-p CoM velocity at foot-contact (mm s^{-1})	511.0 (29.1)	622.1 (106.0)**	462.4 (75.7)	457.3 (90.8)	485.7 (64.4)	511.9 (69.7)
	Overall movement time (s)	1.29 (0.14)*	1.19 (0.16)	1.18 (0.23)	1.31 (0.21)*	1.38 (0.23)	1.47 (0.20)*

APA = anticipatory postural adjustment, SE = step execution. Difference between stepping direction are indicated as $^{*}P < 0.05$, $^{**}P < 0.001$. NB, m-l CoP displacements during the APA phase were directed towards the intended swing limb, whereas during the SE phase they were directed towards the stance limb.

(transfemoral amputees, $P < 0.001$; transtibial amputees, $P < 0.05$), whereas in the control group the trend was reversed and the posterior displacement of the CoP was significantly greater when stepping down compared to stepping up ($P < 0.001$). Peak lateral displacement of the CoP during stepping up compared to stepping down was significantly reduced in transfemoral amputees ($P < 0.001$) but was not significantly different in transtibial amputees ($P = 0.32$) or controls ($P = 0.22$). All subject groups had a higher m-l CoM velocity when stepping up compared to when stepping down ($P < 0.001$). Differences in the resulting CoM forwards velocity between stepping direction reflected those found in the posterior displacement of the CoP, i.e. a-p CoM velocity at swing limb heel-off was significantly greater in both amputee sub-groups when stepping up compared to when stepping down (transfemoral amputees, $P < 0.05$; transtibial amputees, $P < 0.05$), whereas in the control group the a-p CoM velocity at swing limb heel-off was significantly greater when stepping down compared to when stepping up ($P < 0.001$).

3.2. Differences in step execution phase between stepping up and stepping down

Differences in anticipatory postural adjustment phase duration, which were determined as a percentage of the overall movement time, indicate the duration of the step execution phase when stepping up compared to when stepping down was significantly shorter in both amputee sub-groups ($P < 0.001$) but was significantly longer in the control group ($P < 0.001$). Peak lateral CoM displacement was more or less the same when stepping up as when stepping down in transfemoral amputees ($P = 0.37$), but was significantly greater when stepping up compared to when stepping down in transtibial amputees ($P < 0.001$) and control subjects ($P < 0.05$). All subject groups had a higher m-l CoM velocity at swing limb foot-contact when stepping up compared to stepping down ($P < 0.001$). Finally, in both amputee sub-groups differences in a-p CoM velocity at swing limb foot-contact between stepping up and stepping down were not significant ($P > 0.05$), whereas in the control group, a-p CoM velocity at swing limb foot-contact was significantly reduced when stepping up compared to stepping down ($P < 0.001$).

4. Discussion

By evaluating the relationship between the horizontal trajectories of the CoM and CoP to describe the gait initiation process, the aim of the present study was to determine the balance and postural control adaptations adopted by unilateral lower-limb amputees when stepping up and when stepping down to a new level. Although the subjects in the present study used various different prosthetic foot/leg combinations, which is likely to have affected the specific way in which the gait initiations were performed, the data presented represent the general trends for the different sub-groups. Findings indicate the gait initiation process utilised by lower-limb amputees was dependent on the direction of stepping and more particularly by which limb the amputee led with.

When stepping up (leading with their intact limb), amputee subjects had significantly greater posterior displacement of the CoP compared to when stepping down leading with their prosthesis (53% and 39% greater in transfemoral and transtibial amputees respectively, $P < 0.001$). Posterior displacement of the CoP is associated with the generation of forward momentum when initiating gait on the level (Brenière et al., 1987). Thus, the increased backward shift of the CoP during stepping up compared to when stepping down may have been a result of amputee subjects attempting to create greater forward thrust whilst their intact (intended swing) limb was still in contact with the ground. Posterior displacement of the CoP has also been shown to be proportional to the progression velocity of the CoM at the end of the first step (Brenière et al., 1987), which may explain why the a-p CoM velocity at heel-off was greater when stepping up compared to stepping down in both amputee sub-groups (73% greater in transfemoral amputees, $P < 0.01$; and 58% greater in transtibial amputees, $P < 0.001$, Table 3). This finding is in agreement with those reported by Michel and Do (2002), indicating that the forwards velocity of the CoM at swing limb toe-off was significantly higher in transfemoral amputees when gait was initiated leading with the intact limb compared to leading with the prosthetic limb. This suggests that when leading with their intact limb, lower-limb amputees develop much of their forward velocity whilst their swing (intact) limb is still in contact with the ground. In the present study, the greater forward momentum (a-p CoM velocity) created during the anticipatory postural adjustment phase when stepping up compared to stepping down, would have allowed amputees to use their prosthesis as a rigid lever about which their CoM was rotated forwards over during the subsequent step execution phase. Using the prosthesis in this way may partly explain why the duration of the step execution phase, when stepping up, was proportionally shorter in both amputee sub-groups compared to the able-bodied control group.

The increased a-p CoM velocity at heel-off in amputee subjects when stepping up compared to stepping down, may also have been as a result of them reducing their a-p CoM velocity when stepping down. In both amputee sub-groups there was a significant increase in step execution phase duration when stepping down compared to when stepping up. As the increase in step execution phase duration indicates that amputee subjects spent longer in single-support on their intact (stance) limb, findings suggest that amputee subjects, when landing on their prosthesis, lowered their CoM in a controlled manner. From a subjective (observational) viewpoint, the stepping down movement in amputee subjects could be described as being 'staggered', unlike in able-bodied controls who completed the stepping down task confidently and in a 'smooth' manner. Indeed, in able-bodied controls the a-p CoM velocity at swing limb foot-contact was significantly higher when stepping down compared to when stepping up ($P < 0.001$), whereas in both amputee sub-groups there was no difference in a-p CoM velocity at swing limb foot-contact between the two stepping directions ($P > 0.05$). These findings suggest amputees adopted a slower 'cautionary' approach when stepping down onto their prosthesis.

During the step execution phase, when the stance limb became fully loaded and the swing limb unloaded, the CoP moved rapidly across to the stance limb (see Figs. 1 and 2). For able-bodied controls this rapid movement of the CoP tended also to move posteriorly when stepping in each direction, so that during the initial part of single-support its location beneath the stance limb was towards the heel region. Similar results have been reported for gait initiation on the level in the young, elderly and parkinsonian patients (Halliday et al., 1998), in healthy adult subjects (Brenière et al., 1987; Jian et al., 1993; Malouin and Richards, 2000) and in children (Malouin and Richards, 2000). In the present study a lateral-posterior shift (towards the heel region) in the CoP was also seen to occur in both amputee sub-groups when stepping down. In contrast, when stepping up the rapid movement of the CoP towards the stance limb tended to also move anteriorly in both amputee sub-groups, so that at the beginning of single limb support its location under the stance limb was away from the heel region and towards the toes (see Figs. 1 and 2). This adaptation was likely a result of the increase in CoM a-p velocity created during the anticipatory postural adjustment phase and ensured the CoP moved anterior of the knee joint centre, thereby ensuring the ground reaction force vector helped to keep the knee of the prosthetic limb, in transfemoral amputees, fully extended and, therefore, 'locked'. The incorporation of microprocessor technology into hydraulic knee units to automatically control the amount of stance flexion has supposedly enabled lower-limb amputees to ascend/descend ramps and stairs in a way that closely resembles healthy able-bodied individuals, i.e. 'leg over leg'. It would be interesting to determine whether the use of such devices would alter the gait initiation process adopted when leading with the intact limb, e.g. when stepping up to a new level. In transtibial amputees this altered movement strategy may have reduced the moment required by the knee extensors (which would have also reduced the force of the prosthesis against the distal end of the tibia).

The initial m-l displacement of the CoP towards the intended swing limb when stepping up compared to stepping down was significantly reduced in transfemoral amputees ($P < 0.001$). Additionally, the duration of the step execution phase (representing most of the single-support time) was significantly reduced when stepping up was compared to stepping down in both amputee groups. As a lateral (towards the intended swing limb) shift in the CoP during the anticipatory postural adjustment phase is usually associated with a lateral shift in the CoM (towards the intended stance limb), the reduced lateral CoP displacement in transfemoral amputees may reflect an adaptive stepping strategy, whereby instead of shifting the CoM over to the stance limb, the CoM falls forwards as the intact limb is rapidly lifted from the support surface and is quickly thrust forward onto the step. Apart from limiting the initial lateral displacement of the CoM, such a stepping strategy would reduce the amount of time spent in single-support on the prosthetic limb. The reduced lateral CoP displacement may have also been partly due to a limb loading asymmetry when standing, whereby more weight was borne by the intact limb and thus the CoP position was already located towards this limb (Isakov and Mizrahi, 1997; Lord and Smith, 1984; Summers et al., 1987). It has been suggested that a lack of confidence in the prosthetic limb, resulting from poor balance and discomfort, leads to reduced weight bearing on the prosthetic side (Summers et al., 1987). Thus, the significantly reduced lateral CoP displacement found in transfemoral amputees in the present study during stepping up may have been partly due to a reluctance to use the prosthesis. As it is likely that the type of prosthesis and/or fitting used will affect not only the comfort but also the amount of afferent information from the prosthetic limb, future research should consider how type of prosthesis and/or type of socket affects the gait initiation process utilised.

5. Conclusion

The present study described the gait initiation process adopted by unilateral lower-limb amputees when stepping up and stepping down to a new level. Results indicate that the horizontal trajectories of the CoM and CoP and the temporal parameters that define the gait initiation process were dependent upon the direction of stepping. Findings also revealed that, for both stepping directions, the gait initiation process adopted by unilateral lower-limb amputees was different to that adopted by able-bodied controls.

In the able-bodied control group anticipatory postural adjustments tended to be greater for stepping down compared to stepping up. However, in both amputee subgroups anticipatory postural adjustments tended to be more pronounced when stepping up (leading with the intact limb) than when stepping down (leading with the prosthesis). This suggests that amputees developed much of their forward velocity whilst their intact limb was still in contact with the ground. During the step execution phase, the lateral movement of the CoP from its early position towards the intended swing limb over towards the stance limb tended also to move posteriorly in able-bodied control subjects when stepping in each direction and in amputee subjects when stepping down, so that during the initial part of single-support its location beneath that stance limb was towards the heel region. However, in both amputee sub-groups the strategy adopted when stepping up ensured that the position of the CoP at the beginning of single-(prosthesis)support was located towards the toes. This was seen as a way of ensuring the ground reaction force vector was anterior of the knee joint centre, which in transfemoral amputees would have helped keep the knee of the prosthetic limb fully extended and thus 'locked'. In transtibial amputees this altered movement strategy may have been an adaptation to reduce the knee extensor moment, thereby reducing the force of the prosthesis against the distal end of the tibia. A reduction in anticipatory postural adjustments, and consequently a reduction in CoM velocity and an increase step execution phase duration when stepping down, suggests amputees adopted a cautionary stepping strategy when landing on their prosthesis.

Findings suggest that the balance and postural control strategy relating to the gait initiation process is not a fixed motor program, and thus that becoming an amputee will require time and training to develop alternative neuromuscular control and coordination strategies.

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