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Original article

Can high-functioning amputees with state-of-the-art prosthetics walk normally? A kinematic and dynamic study of 40 individuals

Hannah L. Jarvis [a,b,*], Neil D. Reeves [a], Martin Twiste [b,c], Rhodri D. Phillip [d], John Etherington [d], Alexander N. Bennett [d,e,f]

[a] Research Centre for Musculoskeletal Science and Sports Medicine, Department of Life Sciences, Faculty of Science and Engineering, Manchester Metropolitan University, Manchester, UK
[b] School of Health Sciences, University of Salford, Salford, UK
[c] United National Institute for Prosthetics & Orthotics Development (UNIPOD), University of Salford, Salford, UK
[d] Academic Department of Military Rehabilitation, Defence Medical Rehabilitation Centre, Headley Court, Epsom, Surrey, UK
[e] Leeds Institute of Rheumatic and Musculoskeletal Medicine, University of Leeds, Leeds, UK
[f] National Heart and Lung Institute, Faculty of Medicine, Imperial College London, London, UK

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ABSTRACT

Background: Previous work has highlighted the highly functional post-rehabilitation level of military individuals who sustained traumatic amputation. Understanding how these individuals walk with their prosthesis could be key to setting a precedent for what is realistically possible in the rehabilitation of individuals with amputations.
Objective: The aim of this paper is to answer how "normal" should the gait of an individual with an amputation(s) be and can we aspire to mimic able-bodied gait with the most advanced prosthetics in highly functioning individuals?
Methods: This was a cross-sectional study comparing the gait of severely injured and highly functional UK trans-tibial ($n = 10$), trans-femoral ($n = 10$) and bilateral trans-femoral ($n = 10$) military amputees after completion of their rehabilitation programme to that of able-bodied controls ($n = 10$). Joint kinematics and kinetics of the pelvis, hip, knee and ankle were measured with 3-D gait analysis during 5 min of walking on level ground at a self-selected speed. Peak angle, moment or range of motion of intact and prosthetic limbs were compared to control values.
Results: Joint kinematics of unilateral trans-tibial amputees was similar to that of controls. Individuals with a trans-femoral amputation walked with a more anterior tilted pelvis ($P = 0.006$), with reduced range of pelvic obliquity ($P = 0.0023$) and ankle plantarflexion ($P < 0.001$) than controls. Across all amputee groups, hip joint moments and power were greater and knee and ankle joint moments were less than for controls.
Conclusions: This is the first study to provide a comprehensive description of gait patterns of unilateral trans-tibial, trans-femoral and bilateral trans-femoral amputees as compared with healthy able-bodied individuals. The groups differed in joint kinematics and kinetics, but these can be expected in part because of limitations in prosthesis and socket designs. The results from this study could be considered benchmark data for healthcare professionals to compare gait patterns of other individuals with amputation who experienced similar injuries and rehabilitation services.

1. Introduction

Previous work [1] by our research team and others [2] demonstrated that military personnel who sustained traumatic amputation(s) walk more efficiently (their cost of walking is less) for their amputation level than previously reported, most likely due to their high level of physical fitness before the injury and comprehensive rehabilitation programme after it [1]. Understanding how these individuals walk with their prosthesis could be key to setting a precedent for what is realistically possible in the rehabilitation of individuals with amputation.

Numerous studies have reported the biomechanical function of individuals walking with a prosthesis after amputation(s). Several

* Corresponding author.
E-mail address: Hannah.Jarvis@mmu.ac.uk (H.L. Jarvis).

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of these, although seminal research studies of their time, do not reflect modern care and up-to-date prosthetic provision for individuals with amputation(s). Also, the studies placed greater emphasis on comparing particular joints [3–5], the performance of specific prosthetic components [6], or the effect of altering alignment [7,8]. Several studies focused on the stance [3,8–10] or swing phase [3,11] or reported data only in the sagittal plane [9,12,13].

Many research papers base inclusion criteria on amputation level and the absence of other conditions or common problems individuals with amputation(s) face such as residual limb discomfort that might adversely affect walking ability [8,12,14]. In reality, in clinical practice, there is a broad spectrum of functional performance depending on the nature of the original amputation, any concomitant injuries or deformities, the intensity and duration of rehabilitation, the design of the prosthetic, and the quality of prosthetic or socket fitting, alignment, components and the motivation of the individual.

Although several studies have demonstrated that for individuals wearing advanced prostheses, joint kinematics and kinetics more closely resemble able-bodied individuals [3], the prostheses do not perfectly replicate the features associated with normal joint anatomy or the muscles' ability to generate force under fine neural control. Proprioception between the prostheses and the ground the individual is walking on is diminished, and additional constraints are imposed by the requirement to transmit loads between the socket and residual limb. A gait similar to the able-bodied person is not likely possible considering the sensory and motor deprivation induced by the amputation; instead, it may be necessary to use alternative kinematic (and kinetic) strategies than what is preconceived to be "normal" gait [15]. Therefore, it may be more realistic to set prosthetic goals in relation to the gait pattern of highly functional individuals with amputation such as military personnel who have undergone comprehensive rehabilitation with an optimally fitted, state-of-the-art prosthesis(s).

The aim of this paper was to ask how "normal" should the gait of an individual with amputation(s) be and can we aspire to mimic able-bodied gait with the most advanced prosthetics in highly functioning individuals? We envision that our data could be used as a benchmark dataset for representing the highest functioning individuals for future comparisons of gait in individuals with amputation using optimally fitted prostheses of varying design and specification.

2. Methods

2.1. Study design and setting

This was a cross-sectional study comparing the joint kinematics and joint kinetics of unilateral trans-tibial, trans-femoral and bilateral trans-femoral amputees at the end of their rehabilitation at the Defence Medical Rehabilitation Centre (DMRC) Headley Court, UK, to healthy able-bodied military controls during over-ground walking at self-selected speed. Participants were recruited from October 2013 to August 2014. This study is reported according to the Strobe checklist for observational cross-sectional studies. The study was approved by the Ministry of Defence Research Ethics Committee (272/PPE/11) and the University of Salford ethics panel (HSCR 13/12). Informed written consent was obtained from each participant.

2.2. Participants

We recruited 40 individuals ($n = 10$ unilateral trans-tibial, $n = 10$ unilateral trans-femoral, $n = 10$ bilateral trans-femoral) after completion of their rehabilitation pathway at the DMRC Headley Court and 10 healthy able-bodied military controls. This sample size is comparable to other studies of similar design [16,17]; no formal sample size calculation was undertaken. Inclusion criteria for amputees were that they could walk continuously for at least 12 min and had been wearing their prostheses for at least 6 months before testing. Individuals with amputation who had sustained a traumatic brain injury were excluded. Controls were 10 military personnel who had been asymptomatic for at least 6 months before testing and without previous major joint or soft tissue surgery and thus might be considered a match for the pre-injury status.

2.3. Outcome measures

2.3.1. Demographic data

Demographic data collected included age, weight, mass, New Injury Severity Score (NISS) [18], duration of rehabilitation (total time spent at DMRC Headley Court for intensive rehabilitation), time since injury and prosthetic foot and knee prescription.

2.3.2. Measurement of joint kinematics and joint kinetics

All data were collected simultaneously, including the metabolic energy expenditure data and temporal and spatial parameters reported previously [1]. Measurements involved using an optoelectronic motion capture system (Vicon, Oxford, UK) with 10 T-Series Vicon cameras and 4 strain gauged force plates (AMTI, Watertown, MA, USA) embedded within a 10-m walkway. Retro-reflective markers were attached to the skin or prosthesis to define a virtual model for anatomical coordinate systems and to track the movement of the pelvis, thigh, shank and foot segments during walking as per Table 1 and Fig. 1. Kinematic data were collected at 120 Hz and ground reaction forces at 1200 Hz. A static standing trial was recorded for each participant so to calculate the location of joint centres. Participants were instructed to walk at a comfortable self-selected speed up and down the gait laboratory (approximately 15 m) for 5 min.

Table 1
Marker placement for amputee individuals with amputation and control participants.

Segment	Marker Placement
Pelvis	Markers were placed on the right and left anterior superior iliac spine and right and left posterior superior iliac spine. These were used to define and track this segment
Thigh	To track the thigh segment 3 markers were placed on the mid-point of the anterior aspect of the thigh in a triangle cluster formation and another marker placed on the mid-point of the posterior aspect of the thigh To define the thigh segment, the hip joint centre was created using recommendations by Harrington et al. [31] and a marker was placed on the medial and lateral condyles of the femur or on the knee joint centre of a prosthetic knee.
Shank	To track the shank segment 4 markers were placed in a square cluster formation on the lateral distal aspect of the shank, the socket for trans-tibial amputees or the prosthetic knee for trans-femoral amputees. To define the shank segment, markers were placed on the medial and lateral condyles of the femur or the knee joint centre of a prosthetic knee and the medial and lateral malleoi or the equivalent for the prosthetic foot
Foot	To track the foot segment, a marker was placed on top of the shoe overlaying the mid-point of the posterior and lateral aspect of the calcaneus and on top of the 1st, 2nd and 5th metatarsal heads. To define the foot segment, markers were placed on the medial and lateral malleoli and metatarsal heads 1 and 5.

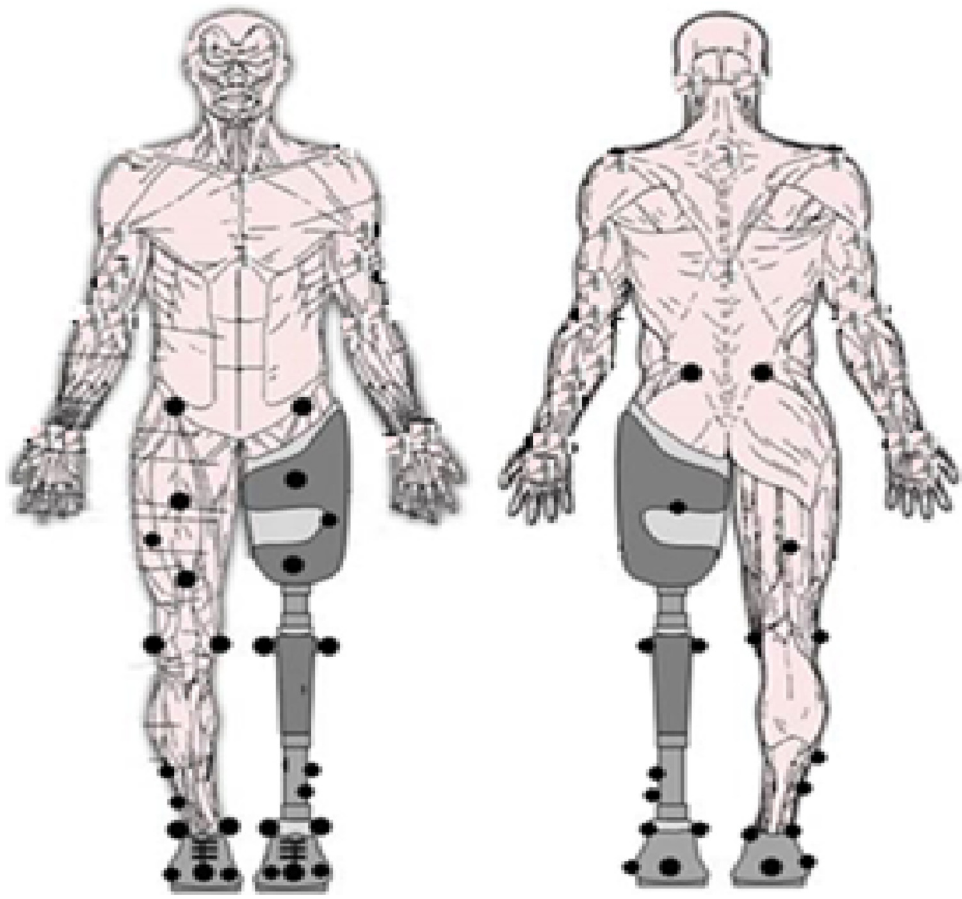

Fig. 1. Marker placement for amputee and control participants. Black dots represent marker placement. Refer to Table 1 for exact placement position.

2.3.3. Data analysis of joint kinematics and joint kinetics

All data were digitised within Vicon, then exported for modelling and analysis within Visual 3D (C-Motion, Rochelle, IL, USA). A model specific to the height and mass of each participant was created. The inertial parameters for each segment are based on the recommendations of De Leva et al. [19]. Joint kinematics were calculated for the pelvis, hip, knee and ankle by using inverse dynamics. Specific constraints were applied at the joints of the virtual model to limit rotation and or translation. The pelvis permitted 6 degrees of freedom; sagittal, coronal and transverse plane rotation were permitted at all other joints. Gait events (initial contact, toe off and initial contact after swing phase) were defined from contact with the force plates. All data were normalised to 0-100% of the gait cycle. Key parameters from Benedetti et al. [20] and Winter [21] and relevant to amputee gait are reported. For controls, only data from the right leg are presented.

2.4. Statistical methods

All data are reported as mean (SD) unless otherwise stated and were compared to controls. Individual parameters were checked with the Kolmogorov Smirnov test. Unilateral trans-tibial, unilateral trans-femoral and bilateral transfemoral and controls were compared relative to each other. For post-hoc analysis, each individual with amputation(s) was compared only to the control with significance at $P < 0.05$. Parametric data were compared by one-way ANOVA with post-hoc Least Significant Difference. Non-parametric data were compared by Kruskall-Wallis test with post-hoc analysis with individual Mann-Whitney tests with Bonferroni correction. Statistical analysis involved using SPSS 25 (IBM Corp., New York, NY, USA). No formal corrections were applied to p-values with respect to the multiple parameters considered.

3. Results

3.1. Demographic data

Individuals with amputation and controls were of similar age, height, and body mass (Table 2). Table 2 presents the cause of injury, injury severity, length of rehabilitation and prosthetic prescription for all individuals with amputation(s). The cause of amputation was mostly an explosion type injury pattern (e.g., improvised explosive device or mine). Other causes were road traffic accidents, crush injury and gunshot wound. Individuals with a bilateral trans-femoral amputation required significantly longer rehabilitation (mean 24 months, $n = 5$) than those with unilateral trans-tibial amputation (5 months, $n = 3$) or unilateral trans-femoral amputation (6 months, $n = 2$).

Individuals with unilateral trans-tibial amputation were fitted with a total surface-bearing or a patella tendon-bearing socket. Most were fitted with an Echelon VT ($n = 5$), reflex shock ($n = 2$), Variflex XC ($n = 2$) or VSP ($n = 1$) prosthetic foot. For individuals with a unilateral trans-femoral amputation, the type of socket seemed to depend on the type of amputation. Individuals with a knee-disarticulation amputation were fitted with a distal end-bearing socket, whereas those with a trans-femoral amputation were fitted with an ischial bearing socket. Nearly all individuals with unilateral trans-femoral amputation were fitted with a KX06 prosthetic knee ($n = 7$) or a microprocessor knee joint ($n = 2$) (C-Leg or Genium) and a dynamic elastic

Table 2
Characteristics of participants.

Group	Age (years)	Mass (kg)	Height (m)	NISS	Cause of amp.	Duration of rehab. (months)	Time from injury (months)	Socket type	Socket liner	Torque Adaptor	Prosthetic foot	Prosthetic knee
Unilateral trans-tibial	23	78.2	1.81	N/A	Crush	3.6	12	TSB	Iceross sport pin	Yes	Echelon VT	N/A
	29	88.5	1.86	17	IED	11.8	61	PTB	No	No	VSP	N/A
	24	119.6	1.86	12	IED	4.2	8	PTB	No	No	Variflex XC	N/A
	28	84.9	1.86	29	IED	13.4	33	PTB	cushion	Yes	Echelon VT	N/A
	32	94.1	1.85	12	Mine	2.1	69	TSB	Iceross pin	No	Reflex Shock	N/A
	28	89.5	1.75	17	IED	5.7	19	TSB	Iceross synergy	No	Echelon VT	N/A
	28	84.5	1.80	17	IED	4.9	20	PTB	Pin	No	Reflex Shock	N/A
	35	103.7	1.80	5	IED	5.9	19	TSB	Pin	Yes	Echelon VT	N/A
	26	87.8	1.90	21	IED	6.3	20	TSB	Pin	Yes	Echelon VT	N/A
	24	66.5	1.74	N/A	Crush	6.0	7	TSB	Pin	No	Variflex XC	N/A
Mean (SD)	28 (4)	90 (14)	1.82 (0.05)	16 (7)		5 (3)	27 (22)					
Unilateral trans-femoral	32	88.8	1.69	24	IED	4.5	39	IBS	Seal in liner	No	Axtion	C-LEG
	29^{KD}	85.3	1.78	22	RTA	6.8	8	DEB	Seal in	No	Variflex xc	KX06
	35	83.5	1.81	43	Mine	5.2	32	IC	Seal in	No	Reflex shock	KX06
	26^{KD}	98.6	1.89	18	IED	4.8	44	DEB	Seal in	No	LPRR	Plie
	27^{KD}	94.3	1.80	16	GSW	4.9	26	DEB	No	No	Variflex xc	KX06
	27	89.9	1.75	18	IED	6.9	71	IBS	No	No	LPRR	KX06
	30^{KD}	83.8	1.87	29	IED	3.5	23	DEB	No	Yes	Elite VT	KX06
	27	96.9	1.87	18	Mine	6.7	98	n/n	Seal in	No	Triton HD	Genium
	27	80.3	1.75	34	RTA	11.4	22	DEB	Guardian	No	Echelon	KX06
	35	81.1	1.71	16	RTA	3.8	29	IBS	Seal in	No	Variflex xc	KX06
Mean (SD)	29 (3)	89 (6)	1.81 (0.06)	24 (9)		6 (2)	39 (27)					
Bilateral trans-femoral	29	86.7	1.91	36	IED	6.7	32	IC (r) DEB (l)	Seal in (r) Alpha cushion (l)	No	Low profile triton	Genium
	24	85.5	1.85	59	IED	7.8	33	Quad	Seal in	No	Axtion	C-leg
	28	68.1	1.67	57	IED	12.3	40	n/n	Seal in	No	Axtion	C-leg
	28$^{TF/TT}$	88.8	1.85	50	IED	17.2	27	TSB (l) IC (r)	Seal in (r) Activa (l)	Yes (l)	Triton shock +LPRR	Genium
	29	72.9	1.83	41	IED	13.4	46	IC	Seal in	No	LPRR	Genium
	34	78.9	1.75	57	IED	12.5	39	IC	Seal in	No	LPRR	Genium
	37^{BKD}	88.7	1.89	41	IED	15.9	28	DEB	No	No	LPRR	Genium
	29	90.4	1.81	48	IED	10.9	24	IBS	Seal in	No	LPRR	Genium
	27	136.2	1.82	50	IED	17.7	32	DEB	Seal in (r) Sock fit (l)	No	LPRR	Genium
	25	70.7	1.76	54	IED	22.1	43	IC	Seal in	Yes	Triton shock	Genium
Mean (SD)	29 (4)	90 (20)	1.82 (0.07)	49 (8)		24 (5)	35 (7)					
Control Mean (SD)	30 (6)	78 (8)	1.84 (0.07)									

^{KD}: knee disarticulation (rather than true trans-femoral amputation). ^{BKD}: bilateral knee disarticulation. $^{TF/TT}$: trans-femoral and trans-tibial amputation (rather than bilateral trans-femoral). LPRR: low profile reflex rotate; RTA: road traffic accident; IED: improvised explosive device; Crush, crush injury; GSW: gunshot wound; PTB: patella tendon bearing; TSB: total surface bearing; IC: ischial containment; IBS, ischial bearing socket; DEB: distal end bearing; N/A: Not applicable. Duration of rehabilitation represents time spent attending a rehabilitation programme at DMRC Headley Court. Time from injury represents the time from injury to when the person attended data collection for the study. n/n: Not known

response prosthetic foot, although the model of the latter varied greatly among participants. Socket type varied greatly among individuals with bilateral trans-femoral amputation, with 2 participants wearing different socket types on either leg. Most participants with bilateral trans-femoral amputation were fitted with ischial bearing or distal end-bearing sockets. Nearly all individuals with bilateral trans-femoral amputation were fitted with a micro-processor knee joint such as a Genium prosthetic knee or C-leg, and most were fitted with a low-profile reflex rotate prosthetic foot (LPRR).

3.2. Joint kinematics and kinetics

All individuals walked at their self-selected walking speed: unilateral trans-tibial, mean 1.36 m/s; unilateral trans-femoral, 1.22 m/s; bilateral trans-femoral, 1.12 m/s; and able-bodied participants, 1.29 m/s. Other temporal and spatial parameters and metabolic energy expenditure values in the same cohort are reported in Jarvis et al. [1].

3.2.1. Individuals with unilateral trans-tibial amputation

Pelvis and hip joint kinematics were similar to that in controls ($P > 0.06$) (Tables 3 and 4 and Figs. 2 and 3). For the prosthetic leg, during early stance, maximum hip joint extension moment was greater [mean (SD)$-$1.3 N.m/Kg (0.4)] than for the intact leg. Hip power generation and absorption at H1 and H3 were greater than for controls (Supplemental Fig. 1). During early stance, maximum knee joint flexion [8.5° (7.4°), $P < 0.001$] and maximum knee joint extension moment [0.1 (0.3) N.m/Kg, $P = 0.006$] were lower than for controls (Fig. 1).

3.2.2. Individuals with unilateral trans-femoral amputation

Pelvis and hip joint kinetics were similar to controls, although hip power generation during early stance was reduced, but we reported several differences in joint kinematics. During the stance phase, the pelvis posterior tilted less on the prosthetic leg [mean (SD) 8.3° (4.3°), $P = 0.009$] (the pelvis was more anterior tilted throughout the gait cycle) (Fig. 2), then posterior tilted toward the end of the stance phase. The range of pelvic obliquity was less for

Table 3
Mean joint kinematics (degree°) of the pelvis, hip, knee and ankle for individuals with unilateral trans-tibial, unilateral trans-femoral and bilateral trans-femoral amputation(s) compared to controls.

Joint	Parameter (Anova p)	Control	Unilateral trans-tibial		Unilateral trans-femoral		Bilateral trans-femoral	
			Intact	Prosthetic	Intact	Prosthetic	Right	Left
Pelvic tilt°	Max post tilt (0.006)	2.5 (4.5)	2.6 (3.2)	2.4 (4.2)	3.6 (6.1)	8.3 (4.3)[b]	8.4 (5.3)[b]	7.2 (5.5)[a]
	Range (0.007)	4.9 (1.7)	5.9 (1.9)	6.7 (2.6)	8.7 (4.7)[b]	8.7 (4.7)[b]	9.0 (1.8)[b]	9.6 (2.5)[b]
Pelvic obliquity	Range (0.023)	12.1 (4.1)	13.3 (4.1)	13.3 (3.9)	8.2 (3.3)[a]	9.0 (3.7)	1.3 (3.6)	11.1 (3.1)
Pelvic rotation°	Max int rotation (0.007)	4.2 (2.6)	3.9 (5.3)	4.8 (3.2)	7.1 (4.3)	4.1 (2.2)	3.8 (4.8)	9.8 (5.9)[b]
Hip flexion°	Max ext (0.380)	−14.1 (6.7)	−13.7 (4.9)	−12.4 (6.7)	−12.4 (6.1)	−11.3 (7.4)	−8.4 (5.7)	−9.8 (7.2)
Hip adduction°	Max adduction stance (0.001)	11.1 (2.5)	10.3 (2.5)	8.2 (4.1)	6.3 (3.5)[b]	5.8 (4.3)[b]	6.0 (2.3)[b]	0.9 (2.3)[b]
Hip rotation°	Max int rotation (0.01)	1.5 (3.7)	−0.1 (6.1)	0.3 (6.6)	−0.2 (3.9)	1.4 (6.3)	−10.4 (3.8)	−15.8 (7.5)[b]
	Range (< 0.001)	12.1 (2.0)	11.7 (2.5)	10.1 (3.1)	10.7 (3.7)	12.5 (5.0)	14.3 (4.3)[b]	17.3 (7.1)[a]
Knee flexion°	Max flex early stance (0.001)	17.9 (5.9)	17.2 (5.5)	8.5 (7.4)**	20.1 (6.2)	1.8 (3.3)[b]	8.7 (4.6)[b]	8.6 (4.1)[b]
Ankle flexion°	Max late stance (< 0.001)	10.0 (2.8)	10.4 (3.1)	11.3 (3.1)	5.4 (1.5)[b]	7.5 (2.2)[a]	5.6 (2.6)[b]	6.8 (2.2)[b]

Data are mean (SD). Parametric data were compared by one-way ANOVA with least significant difference post-hoc analysis. Non-parametric data were compared by Kruskal-Wallis test with post-hoc analysis with Mann–Whitney tests with Bonferroni correction. Int, internal; ext, external; max, maximal; add, adduction; flex: flexion. Positive angle indicates anterior tilt, up and internal rotation at the pelvis, flexion, adduction and external rotation at the hip and knee joints.
[a] $P < 0.05$.
[b] $P < 0.01$.

Table 4
Joint moment (N.m/Kg) and joint power (W/Kg) at the hip, knee and ankle for individuals with unilateral trans-tibial, unilateral trans-femoral and bilateral trans-femoral amputation(s) compared to controls.

Joint	Parameter	Control	Unilateral trans-tibial		Unilateral trans-femoral		Bilateral trans-femoral	
			Intact	Prosthetic	Intact	Prosthetic	Right	Left
Hip ext moment, N.m/Kg	Max ext (< 0.001)	−0.9 (0.3)	−1.2 (0.4)	−1.3 (0.4)[a]	−1.1 (0.7)	−1.3 (0.4)	−1.1 (0.3)[b]	−1.0 (0.3)
	Max flex (< 0.001)	1.0 (0.4)	1.7 (0.5)[b]	1.1 (1.2)	1.2 (0.9)	1.2 (0.9)	0.9 (0.2)[b]	1.0 (0.2)
Knee ext moment, N.m/Kg	Max ext early stance (< 0.001)	0.6 (0.2)	0.8 (0.3)	0.1 (0.3)[a]	0.5 (0.6)	0.3 (0.3)[a]	0.2 (0.2)[b]	0.1 (0.2)[b]
Ankle plantar moment, N.m/Kg	Max late stance (0.005)	1.6 (0.3)	2.1 (0.5)	1.9 (0.7)	1.7 (0.4)	1.7 (0.5)	1.4 (0.2)	1.5 (0.2)
Hip power, W/Kg	H1 Max gen (0.002)	0.5 (0.3)	1.2 (0.8)[a]	1.3 (0.8)[b]	0.9 (0.4)	1.1 (0.5)[a]	1.6 (0.4)[b]	1.5 (0.6)[b]
	H2 Max abs (0.025)	−0.8 (0.3)	−1.2 (0.8)	−1.1 (0.4)	−0.9 (0.5)	−1.1 (0.6)	−0.7 (0.2)	−0.6 (0.3)
	H3 Max gen (0.028)	1.2 (0.5)	1.9 (0.5)[b]	1.8 (0.7)[a]	1.4 (0.6)	1.4 (0.6)	1.4 (0.3)	1.8 (0.4)8
Ankle power, W/Kg	A2 Max gen (< 0.001)	3.2 (0.6)	3.9 (1.5)	2.7 (1.5)	2.2 (1.8)[a]	1.3 (1.6)[b]	0.8 (0.4)[b]	0.9 (0.7)[b]

Data are mean (SD). Positive moment indicates extensor, abductor and external rotator at the hip and knee, plantarflexor, evertor and external rotator at the ankle. Ext: extensor; Abd: abductor; Rot: rotator; Gen: generation; Abs: absorption
[a] $P < 0.05$
[b] $P < 0.01$. Parametric data were compared by one-way ANOVA with least significant difference post-hoc analysis. Non-parametric data were compared by Kruskal-Wallis test with post-hoc analysis with Mann–Whitney tests with Bonferroni correction.

the intact leg [8.2° (2.3°) $P = 0.026$] and less but not significantly for the prosthetic leg [9.0° (3.7°), $P = 0.08$] (Supplemental Fig. 2, 3).

Maximum adduction was reduced at the hip joint for the intact leg [mean (SD) 6.3° (3.5°), $P = 0.002$] and prosthetic leg [5.8° (4.3°) ($P = 0.001$] (Supplemental Fig. 2, 3). For the prosthetic leg during early stance, knee joint flexion [1.8° (3.3°), $P < 0.001$] and knee joint extension moment [0.1 (0.3) N.m/Kg, $P = 0.116$] were reduced. Maximum ankle joint dorsiflexion was significantly reduced on both the intact leg [5.4°(1.5°), $P < 0.001$] and prosthetic leg [7.4° (2.2°), $P = 0.028$] during late stance, as was maximum power generation at the ankle joint during A2 [1.3 (1.6) W/Kg, $P < 0.001$].

3.2.3. Individuals with bilateral trans-femoral amputations

Nearly all joint kinetic parameters at the hip and knee of individuals with bilateral trans-femoral amputations significantly differed from control values, more so for the left than right leg. Maximum posterior tilt was increased (the pelvis was more anterior rotated throughout the gait cycle) [right: mean (SD) 8.4°(5.3°), $P = 0.008$ and left: 7.2° (5.5°), $P = 0.04$] (data from hereafter are for right and left, respectively). At the hip joint, extension moment was greater. Power generation during H1 was increased (Supplemental Fig. 1). Maximum adduction at the hip joint was reduced [6.0° (2.3°), $P < 0.001$; 0.9° (2.3°), $P < 0.001$] (Supplemental Fig. 2, 3). At the knee joint during early stance phase, we reported reduced maximum flexion [8.4° (4.1°), $P < 0.001$] and maximum extension moment [0.2 (0.2) N.m/Kg, $P = 0.004$; 0.1 (0.2) N.m/Kg, $P = 0.001$]. At the ankle joint, maximum dorsiflexion was reduced, as was maximum power generation at A2 [0.8 (0.4) W/Kg, $P < 0.001$ and 0.9 (0.7) W/Kg, $P < 0.001$] (Supplemental Fig. 1).

4. Discussion

4.1. Individuals with trans-tibial amputation

We reported small differences in joint kinematics and kinetics between individuals with unilateral trans-tibial amputation and controls. As expected and in agreement with previous work [8,22,23], maximum knee joint flexion and knee joint extension moment were reduced for the prosthetic leg during the early stance phase. In agreement with Lee et al. [23] and Beyaert et al. [8], this could be a protective mechanism to reduce stress at the stump-socket interface around the tibial tuberosity, which can cause discomfort and therefore reluctance to load onto the knee joint of the amputated limb. Others suggest that that limited dorsiflexion at prosthetic ankle joints limits tibial inclination reducing knee flexion [8,22]. In our cohort, peak dorsiflexion at the ankle joint during early stance was similar to that in controls, which may reflect the type of prosthetic feet worn by our cohort: all were wearing dynamic elastic response feet, which aim to facilitate dorsiflexion.

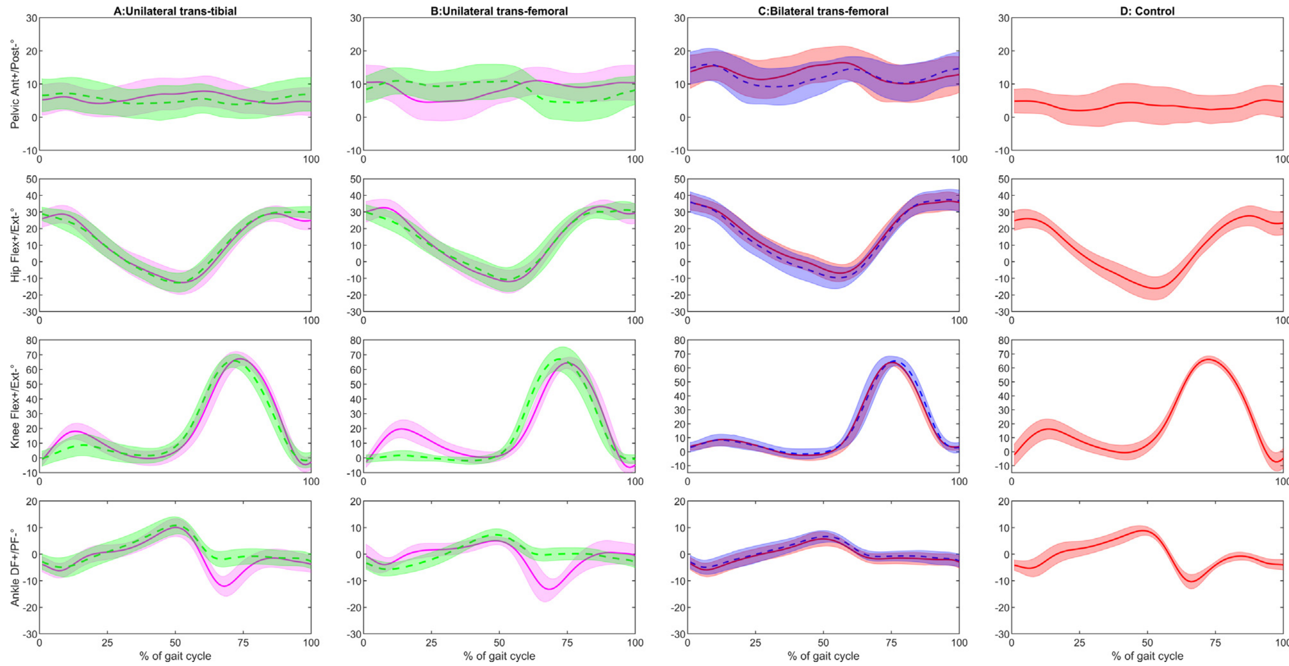

Fig. 2. Sagittal joint kinematics of the pelvis, hip, knee and ankle for individuals with unilateral trans-tibial, unilateral trans-femoral, bilateral trans-femoral amputation and controls. Positive angle indicates anterior tilt, up and internal rotation at the pelvis. Solid black line with dark grey shading: intact limb (or control limb), dashed black line and light grey shading: prosthetic leg. Solid line or dashed line represents mean, shading represents standard deviation.

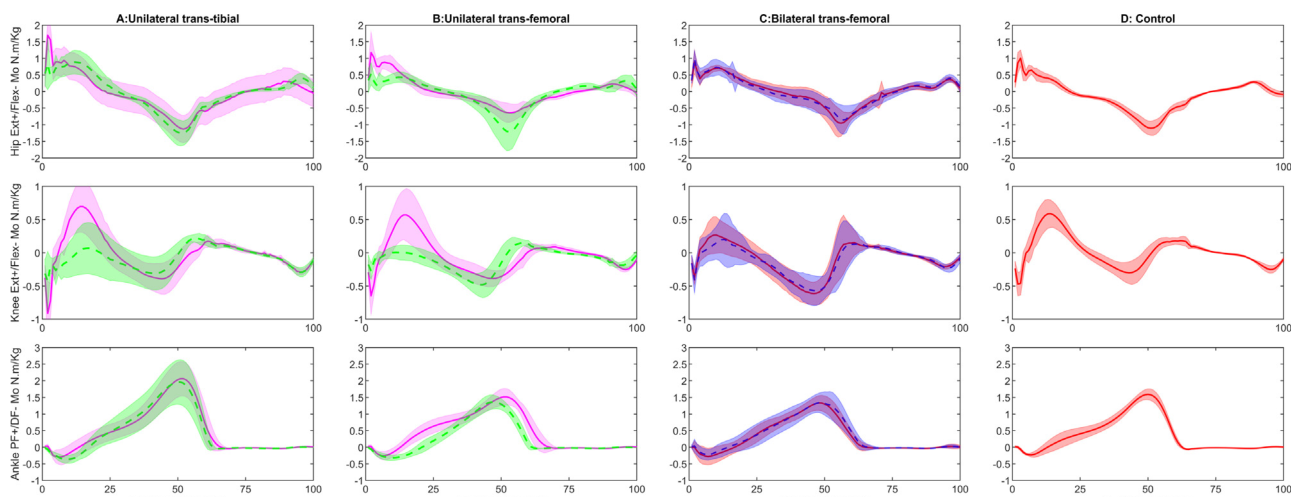

Fig. 3. Sagittal joint kinetics of the pelvis, hip, knee and ankle for individuals with unilateral trans-tibial, unilateral trans-femoral, bilateral trans-femoral amputation and controls. Positive angle indicates anterior tilt, up and internal rotation at the pelvis. Solid black line with dark grey shading: intact limb (or control limb), dashed black line and light grey shading: prosthetic leg. Solid line or dashed line represents mean, shading represents standard deviation.

Peak sagittal plane ankle joint moment was greater than in controls, which could be attributed to the greater walking speed of our trans-tibial cohort as compared with controls (mean 1.36 vs. 1.29 m/s). However, the most likely cause is that the angle of the ground reaction force is directed further away from the joint centre (perhaps due to prosthetic setup or the adapted walking pattern of the individual walking with a prosthesis), which creates a larger external moment arm that is balanced by an equal internal moment, the ankle plantarflexor moment.

5. Individuals with unilateral trans-femoral amputation

It is generally agreed and observed in our cohort that most individuals with unilateral (or bilateral) trans-femoral amputation will walk with a more anterior tilted pelvis than able-bodied individuals [10,24,25]. There are several reasons for this. Shortening of hip flexors and gluteal weakness due to prolonged sitting (often in a wheelchair to help initial mobility post-amputation) and the surgical technique used to attach hamstrings particularly where the residual limb is short can place the pelvis in a more anterior rotated position. Ischial bearing sockets commonly used for many individuals with trans-femoral amputation (and 4 of our participants were wearing that type of socket) can make it difficult to facilitate hip extension because they push against the ischium for control, maintaining the pelvis in an anterior rotated position [26].

There was a notable posterior tilt of the pelvis toward the end of the stance phase on the prosthetic leg, which predictably can occur to facilitate hip flexion from its hyperextended position. Although the cause of low back pain (a common problem for many individuals with amputation) is multi-factorial, from a biomechanical perspective, Esposito et al. [10] suggest that the counter rotation movement from anterior to posterior tilt may cause alternating compressive and then tensile stresses on the soft tissues of the lower back.

In the coronal plane, our study and others [25,27] report reduced up and down movement of the pelvis across the gait cycle as compared with controls. Michaud et al. [27] and Sjodahl et al. [25] suggest that the downward movement of the pelvis is reduced during the loading response because hip joint adduction is reduced due to the proximal rim of a trans-femoral socket making adduction of the hip joint uncomfortable. This is most likely evident in our cohort because hip adduction was reduced. Goujon-Pillet et al. [24] reported that for every additional millimetre of residual limb length, coronal plane movement was reduced by 0.03°. The hip stabilising muscles can become more atrophied with a shorter residual limb, causing difficulty stabilising the femur medio-laterally inside the socket and resulting in a noticeable downward movement (e.g., a Trendelburg gait pattern) of the pelvis at the end of the stance phase [24,25]. Four of our unilateral trans-femoral participants had knee disarticulation, which may explain in part the reduced range of pelvic obliquity.

It is a limitation of our study that we included participants with knee disarticulation among individuals with trans-femoral amputation(s). It is appreciably a highly topical issue of aesthetics versus function when discussing trans-femoral amputation versus knee disarticulation amputation that warrants further investigation. There is a strong trend when performing an amputation in military personnel to preserve as much limb as possible (certainly for trans-femoral amputation) [28], whereas in a civilian setting, sometimes limb length is determined for cosmetic preferences (so when an individual is sitting, the respective knee joints are the same distance from the body).

At the hip joint, maximum abduction was similar to controls, but maximum adduction was significantly less and was further highlighted by a small increase in stride width. This finding can have various explanations: a reduction in hip adductor strength post-surgery or even failure at the time of amputation to re-attach the hip adductors as close to their functional line as possible. In consideration that nearly all individuals in this study sustained limb loss from a traumatic blast injury and thus underwent immediate care in a battlefield hospital. Other factors often not discussed in research publications include discomfort from the socket, in particular from ischial bearing sockets. Although these sockets often offer better functional control, they can be uncomfortable, particularly for men, due to compression of genitalia if the leg is brought toward the midline of the body [24].

Somewhat predictable because of the design constraints of prosthetic knee devices, we reported reduced knee flexion and knee extension moment during early stance as compared with controls on the prosthetic side, although the latter was not significant. Nine of the 10 individuals with unilateral trans-femoral amputation were wearing a prosthetic knee joint that did not offer stance-phase knee flexion; only one was wearing a microprocessor knee joint, which permits a small amount of knee flexion [3] and explains the minimal flexion curve in Fig. 2. At the ankle joint, maximum plantarflexion and power generation during A2 were significantly reduced during push off. This observation is most likely due to the stiffness of the prosthetic feet used by our cohort reducing the range of motion available to increase stability.

5.1. Individuals with bilateral trans-femoral amputation

This is the first study to comprehensively report gait patterns of individuals with bilateral trans-femoral amputation. Only 2 other relatively basic studies are in the literature and one is a case study [5]. The data we present are considerably closer to control data than in either of these studies across the range of kinematic and kinetic variables studied, with little difference from individuals with unilateral trans-femoral amputation. Participants in our study were also considerably younger and predictably more active pre-injury than in previous studies.

Discussions regarding pelvis and hip joint kinematics and moments are best referred to the section describing individuals with a unilateral trans-femoral amputation because they demonstrate the same pattern of movement. Generation of hip power during H1 was significantly increased, most likely to aid propulsion forward in the direction of travel, which agrees with McNealy et al. [6]. Increased power absorption at the hip joint during H2 is likely a compensatory function to facilitate push off due to reduced power generation at the ankle joint from the limitations of prosthetic feet design [5]; indeed, all individuals with amputation in this study were fitted with dynamic elastic response feet (e.g., Variflex XC). However, the peak plantarflexion moment at the ankle joint was the same as control values, which suggests that the prosthetic foot limits the angular velocity of the ankle potentially to provide more stability but in turn will reduce the power generated.

The range of early stance phase knee flexion was reduced, which may be largely due to the type and design of prosthetic knee the participants were wearing (e.g., Genium knee, which permits stance-phase knee flexion). Participants in McNealey et al. [6] and Perry et al. [5] studies were wearing prosthetic knee devices, which do not permit stance-phase knee flexion and explains the difference between these cohorts and ours. Often fixed knee devices are used to improve stability for individuals with trans-femoral amputation, but devices such as the Genium knee can provide a much more natural gait pattern with a reduction in falls [29]. However, specific training with structured rehabilitation and an acclimation period is needed to be able to use the Genium knee optimally [30].

6. Conclusions

The results from this study present a thorough up-to-date description of how a lower-limb amputation changes an individual's gait pattern. It demonstrates that with current high-specification prosthetic provision and intensive rehabilitation, individuals with amputation can achieve a gait pattern very similar to able-bodied individuals. However, the results also demonstrate that because of the limitations of current prosthetic designs and discomfort from the socket, we cannot expect an individual with amputation(s) to walk exactly as an able-bodied individual. Therefore, we suggest that our results be used as benchmark data to represent gait patterns of the highest functioning individuals with amputation to guide clinicians in what is possible. We also propose that in conjunction with our and others' work, this research should help support prioritising high-quality rehabilitation and prosthetic provision in civilian healthcare and maintaining this standard in the care of military personnel.

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Author contributions

The study was designed by H.J., A.B., M.T., R.P., and J.E. H.J and N.R wrote the paper with substantial contribution from A.B. and M.T and some contribution from R.P. and J.E. Experimental data were collected and analysed by H.J.

Disclosure of interest

The authors declare that they have no competing interest.

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Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.rehab.2020.04.007.

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