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Spring-mass behavioural adaptations to acute changes in prosthetic blade stiffness during submaximal running in unilateral transtibial prosthesis users

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ABSTRACT

Background: Individuals with lower-limb amputation can use running specific prostheses (RSP) that store and then return elastic energy during stance. However, it is unclear whether varying the stiffness category of the same RSP affects spring-mass behaviour during self-selected, submaximal speed running in individuals with unilateral transtibial amputation.
Research question: The current study investigates how varying RSP stiffness affects limb stiffness, running performance, and associated joint kinetics in individuals with a unilateral transtibial amputation.
Methods: Kinematic and ground reaction force data were collected from eight males with unilateral transtibial amputation who ran at self-selected submaximal speeds along a 15 m runway in three RSP stiffness conditions; recommended habitual stiffness (HAB) and, following 10-minutes of familiarisation, stiffness categories above (+1) and below (-1) the HAB. Stance-phase centre of mass velocity, contact time, limb stiffness' and joint/RSP work were computed for each limb across RSP stiffness conditions.
Results: With increased RSP stiffness, prosthetic limb stiffness increased, whilst intact limb stiffness decreased slightly ($p<0.03$). Centre of mass forward velocity during stance-phase ($p<0.02$) and contact time ($p<0.04$) were higher in the intact limb and lower in the prosthetic limb but were unaffected by RSP stiffness. Intact limb hip joint positive work increased for both the +1 and -1 conditions but remained unchanged across conditions in the prosthetic limb ($p<0.02$).
Significance: In response to changes in RSP stiffness, there were acute increased mechanical demands on the intact limb, reflecting a reliance on the intact limb during running. However, overall running speed was unaffected, suggesting participants acutely adapted to an RSP of a non-prescribed stiffness.

1. Background

Individuals with lower limb amputation (LLA) who wish to be physically active use a variety of prosthetic feet which are often described as dynamic elastic response or energy storing and returning. Running specific prostheses (RSP) [1] act in a similar fashion to a mechanical spring, deforming under load and then recoiling, first storing and then returning elastic energy [2]. In individuals with LLA, RSP stiffness is a key component in determining limb spring stiffness as per a spring-mass model [3]. In individuals with unilateral transtibial amputation (UTA) research has reported that prosthetic limb stiffness decreases [4] or remains constant [5] as running speed increases. Prosthetic limb spring stiffness remains largely unchanged across step frequencies [2,6]. Vertical ground reaction force impact peaks and loading rates have been shown to increase with increasing running speed, while rotational joint stiffness' remains unaffected [5,7–9].

As RSP stiffness is a result of the RSP design (material properties/thicknesses, shape), limb spring stiffness is modulated by adapting joint kinetics proximal to the RSP. For example, when UTA switch from using a daily use prosthesis to a RSP, hip joint positive work is increased on the prosthetic limb [10]. Therefore, differences in RSP stiffness' may result in different neuromotor adaptations. Selecting an optimum RSP stiffness

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is important for running performance e.g. running speed. This relates to the observation that reduced RSP stiffness can lead to a reduced metabolic cost of running in individuals with bilateral transtibial amputation [10,11] although not in UTA [11]. Given the well documented asymmetries and subsequent compensatory adaptations in this population [12], changing RSP stiffness may have implications for the intact limb in individuals with LLA. The selection of RSP stiffness is typically, pragmatically based upon the user's mass, levels of physical activity, or an individual's preference for, or prosthetist experience of, a particular set-up [11]. The extent to which this approach fits the individual requirements of an RSP user is not clear.

Some previous investigations of running with an RSP report findings from groups of highly trained individuals [4,11,13–15]. This potentially limits how far such results can be generalised to less well trained individuals using an RSP for recreational activity. Recreational running is typically self-paced and 'sub-maximal', and thus associated spring-mass behaviour may be considered as self-selected. Therefore, it is relevant to explore how changing RSP stiffness effects spring-mass behaviour in UTAs during submaximal or self-selected speed running. This allows for acute adaptations to prosthetic device set-up to be explored, similar to trialling different prosthetic devices during initial fitting. This could provide a specific evidence base to assist in RSP stiffness selection.

The primary aim of the current study was to investigate how varying the stiffness category of RSPs influenced the spring-mass behaviour of UTA during self-selected, submaximal speed running. To address this aim, the following specific objectives were explored: (1) to assess the effect of varying RSP stiffness' on prosthetic and intact limb spring-mass behaviour (limb stiffness) and running performance; and (2) to explore the changes in joint kinetics that occur with varying RSP stiffness.

2. Methods

2.1. Participants

Eight recreationally active male UTA (Table 1) provided informed consent to participate in the current study, which was approved by the Nottingham Trent University Human Research Ethics committee. All participants were otherwise free from illness, injury or pathology that could have affected gait. All participants were habitual users of an RSP (Blade XT, Chas A Blatchford and Sons, Basingstoke, UK; Fig. 1).

2.2. Experimental design and protocol

Following a warm-up period involving light jogging and stretching, participants performed overground running trials along a fifteen-metre runway and were instructed to run at their comfortable, habitual, self-selected running speed, contacting a force plate located approximately mid-way. Participants completed running trials under three conditions; using their prosthetist prescribed habitual RSP stiffness (HAB, Table 1), and using an RSP one stiffness category above (+1) and below (-1) the HAB, mirroring a prosthetic fitting session. The stiffness categories quoted (1 – 9; 9 being most stiff) are from the manufacturer provided guidelines. Participants completed trials using the HAB RSP, followed by trials using RSPs of varying stiffness (+1, -1), the order of which was counterbalanced across participants. All prosthetic adjustments were performed by the same highly experienced prosthetist familiar with setting up the Blade XT. A single type of RSP was used to eliminate the effects of differing prosthetic design on our research question. There were no other differences in the prosthetic componentry between conditions and thus prosthetic alignment and length were maintained across conditions. Participants completed practice trials to ascertain a consistent start position at the end of the runway, thus avoiding 'targeting' of the force plate. When the RSP stiffness was non-habitual, participants were provided with an accommodation period of ~10 mins to become familiar with RSP stiffness at the end of which they ascertained their start position as described above. During familiarisation time, participants completed practice trials until comfortable with the new prosthesis. Participants completed six trials (three force plate contacts per foot) for each RSP condition. Rest periods between all trials were offered to avoid any fatigue effects.

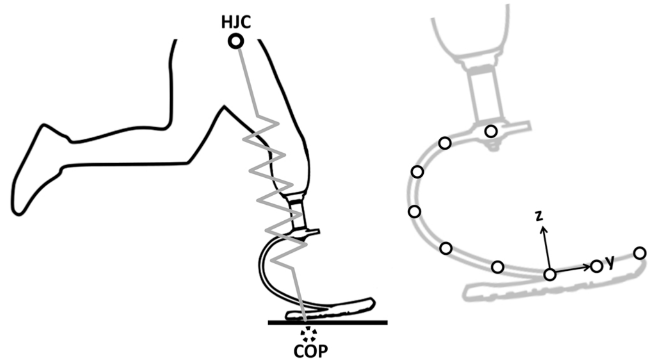

Fig. 1. Schematic representation of the running specific prosthesis (RSP) used in the current study (Blade XT, Chas A Blatchford and Sons, Basingstoke, UK. (https://www.blatchford.co.uk/products/bladext/). The Blade XT is a 'C' shaped RSP designed to also allow for dynamic physical activity, involving walking and running, and all participants reported using their device for recreational running, the target activity of the current study. Marker locations for the RSP and representations of the functional hip joint centre (HJC) and centre of pressure (COP) used to calculate changes in limb spring length (ΔL_{limb}) are also illustrated.

2.3. Data collection

To define an eight-segment model (head, trunk, pelvis, thigh, shank and intact foot), reflective markers (14 mm) were attached to participants as follows; four-marker crown headband, C7, T8, sternum, sacrum,

Table 1
Participant characteristics. The k-level, rated from 0-4, is a system used to categorise prosthesis users' functional capacity.

Participant	Age (years)	Height (m)	Mass (kg)	Prosthetic Limb (R/L)	Residuum length (cm)	K-level	Time Since Amputation (Years)	Cause of Amputation	Suspension	Habitual Stiffness Category (currently used)
1	37.4	1.82	92.8	R	17.5	3	10.9	Trauma	Suction	7
2	40.8	1.65	56.2	R	13	4	4.3	Trauma	Pin	3
3	29.6	1.86	89.3	R	16.5	3	17.4	Trauma	Suction	5
4	23.8	1.70	59.0	R	14	3	4.7	Elective	Suction	2
5	26.6	1.71	84.4	L	15	3	23.7	Trauma	Suction	5
6	52.9	1.74	66.8	L	12	3	26.6	Trauma	Suction	4
7	41.2	1.79	68.0	L	16	3	10.0	Trauma	Suction	3
8	44.5	1.72	70.9	R	15.5	3	10.1	Trauma	Suction	4
Mean	37.1	1.75	73.4				13.5			
SD	9.8	0.07	13.8				8.3			

Summary statistics are mean and standard deviation (SD); The K-level, rated from 0-4, is a system used to categorise prosthesis users' functional capacity.

acromion process, xiphoid process, apex of iliac crest, greater trochanter, femoral epicondyle, medial and lateral malleolus, hallux, calcaneus, 1st metatarsal head, 5th metatarsal head and the medial and lateral aspects of the rear foot. Four marker clusters were placed on the upper and lower legs and were used for segment tracking [16]. The RSP was modelled as eight linked segments (Fig. 1). Eight pairs of serially arranged markers were affixed on the medial and lateral borders of the RSP, except the last pair, which were placed on the RSPs superior surface. A single marker was placed at the distal end of the superior aspect of the RSP to represent the 'toe' (Fig. 1). The first pair of markers defined the distal end of the 'shank'/proximal end of the RSP, with the 'toe' marker representing the distal end of the eighth RSP segment. Determining RSP centre of mass (COM) location was incorporated in the calculation of the whole-body COM.

A static calibration file was collected from participants during quiet standing. Bilateral hip, knee and intact ankle joint centres were determined using a functional joint centre location method [17]. Data were collected using a nine-camera motion capture system (Oqus 400, Qualisys AB, Gothenburg, SE) and a single ground embedded force plate (OR6-7-2000, AMTI, Watertown, USA). Kinematic and ground reaction force (GRF) data were collected at 200 Hz and 1000 Hz, respectively.

2.4. Data processing and analysis

Raw marker position data were labelled and then interpolated using a cubic spline algorithm, with subsequent marker data and raw force data being filtered using a zero-lag low-pass Butterworth filter with a cut-off frequency of 15 Hz. Touchdown and foot-off events were defined at 20 N ascending and descending vertical GRF threshold. The COM position was calculated using defined model segments and weighted averages of each segments' COM properties [18]. For each of the eight linked-segments forming the RSP, the mass was assumed to be divided proportionally along the device and lie at the geometric centre of each linked-segment. All data processing was conducted in Visual 3D (v6: C-Motion Inc, Germantown, MD, USA).

To assess objective one i.e. spring-mass behaviour, limb stiffness was computed for the ground contact phase. [19]. In the current study, the following definition of limb stiffness was used to closely represent the prosthetic lower limb in UTA. Limb stiffness (k_{limb}) was defined as:

$$k_{limb} = \frac{Peak \quad F_{limb}}{\Delta \quad L_{limb}} \tag{1}$$

Where limb spring length (L_{limb}) is the three-dimensional Euclidean distance between the calculated functional hip joint centre and centre of pressure (COP). ΔL_{limb} represented the change in L_{limb} from touchdown to peak limb force (F_{limb}). F_{limb} was defined as the scalar magnitude of GRF directed along the line between the hip joint centre and COP and was calculated as the dot product of the unit vector L_{limb} ($\hat{U}_{limb}$) and the GRF vector, v(GRF):

$$F_{limb} = \widehat{U}_{limb} \bullet v(GRF) \tag{2}$$

F_{limb} was made dimensionless via multiplying it by the ratio of limb length over body weight [4]. Previous studies have acknowledged the need for limb length definitions to be adapted in spring-mass model calculations in UTA [5,20,21] as estimation of limb length from a standing posture in UTA using an RSP would be influenced by deformation of the RSP. Equally, the functional length of the limb changes during stance phase and is more accurately represented using the COP location as the distal end of the limb [22,23]. The three-dimensional nature of the model proposed in the current study overcomes any issues present in a planar definition, such as non-sagittal plane compliance from the stump socket interface, frontal or transverse plane movement [2] or variability in medial-lateral foot placement [13]. To estimate the contribution of RSP deformation to changes in limb spring length (L_{limb}), RSP compression (ΔL_{RSP}) was quantified as the change in three-dimensional Euclidean distance between the proximal end of the most proximal RSP segment and the COP. To assess the effects of RSP stiffness on running performance, running speed (m.s^{-1}) was calculated as the average COM anterior-posterior velocity through the calibrated volume of the runway (ca. 7 m). The stance phase COM velocity (m.s^{-1}), was calculated as the first derivative of the anterior-posterior COM displacement, averaged across stance phase for each limb. Contact time, was determined as the time between touchdown and foot-off respectively, using the kinetic thresholds described above.

To address objective two i.e. to explore joint kinetic changes with varying RSP stiffness, biological joint and RSP work were calculated following previously outlined procedures [24]. For the RSP and associated prosthetic components (socket, pylon etc.) power absorption and generation to and from the RSP was assessed using a previously described unified deformable segment approach [25].

2.5. Statistical analyses

To assess the effect of RSP stiffness (HAB, +1, -1) across the limbs (intact, prosthetic) on variables specific to a given limb (e.g. limb stiffness), two-way repeated measures analyses of variance (ANOVA) were conducted. For variables not specific to a given limb (e.g. running speed), a one-way repeated measures ANOVA was conducted. The normality of data distribution was assessed using a Shapiro-Wilk test. Where the assumption of sphericity was violated, a Greenhouse-Geisser correction factor was applied. Multiple post-hoc comparisons were adjusted for using a Sidak correction and effect sizes (partial eta squared)

were calculated for each statistical comparison. The alpha level of statistical significance was set at $p=0.05$. To reduce the potential for a high false positive rate arising from multiple ANOVA models, a study-wide false discovery rate method was implemented using the Benjamini-Hochberg procedure and a threshold of 20% for all p values produced from all ANOVA models [26,27]. All statistical analysis was conducted in IBM SPSS v.26 (IBM, Portsmouth, UK).

3. Results

Group mean ± 95% confidence intervals, along with individual participant mean outcome measures are presented in Figs. 2 and 3, with full statistical analysis reported in Table 2.

3.1. Spring mass behaviour

All participants made initial ground contact on the 'heel' section of the RSP in all trials. This was verified by determining that the COP y-coordinate, at the instant of ground contact, was posterior to the y-coordinate of the marker attached on the RSP above the point where the heel-keel meets the forefoot keel (Fig. 1).

Limb stiffness' (k_{limb}) showed a significant limb by RSP category interaction effect ($p<0.03$, Table 2), indicating that prosthetic limb stiffness increased with increasing RSP stiffness, but intact limb stiffness' was reduced in HAB and +1 compared to -1 RSP stiffness (Fig. 2). Peak limb force (peak F_{limb}) was significantly higher in the intact limb compared to the prosthetic limb, irrespective of RSP stiffness ($p<0.05$, Table 2). Changes in limb spring length (ΔL_{limb}) produced a significant limb by RSP category interaction ($p<0.01$, Table 2). This indicated that ΔL_{limb} decreased for the prosthetic limb as RSP stiffness increased, but remained unchanged with increasing RSP stiffness in the intact limb. RSP compression (ΔL_{RSP}) was not different across RSP stiffness conditions ($p=0.58$, Fig. 2).

3.2. Running performance

Stance-phase COM velocity and contact time were significantly increased ($p<0.02$) and decreased ($p<0.04$) respectively in the intact

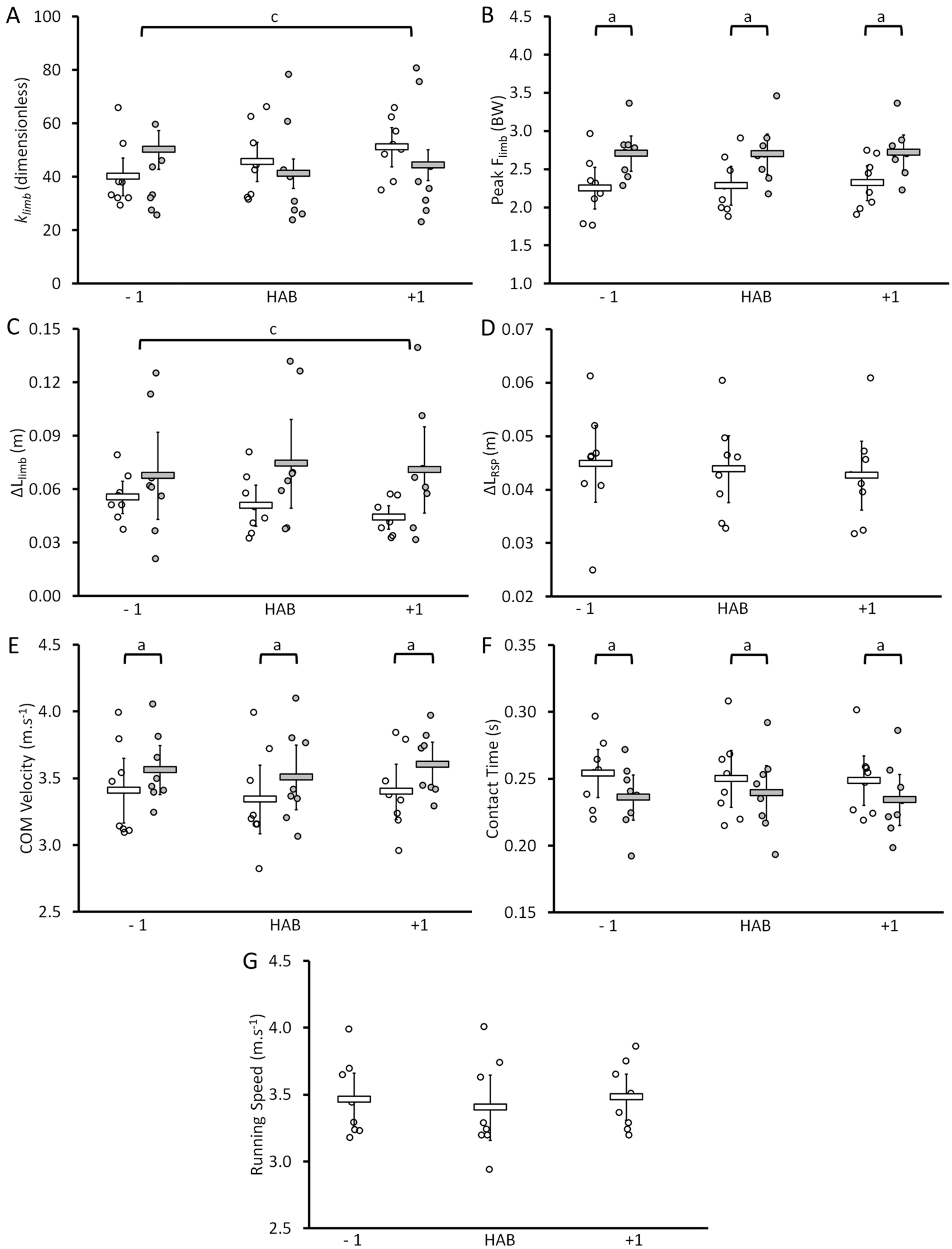

Fig. 2. Group mean ± 95% confidence intervals and individual participant mean data representing the prosthetic (white) and intact (grey) limbs for each RSP condition. Data presented and abbreviations are as follows: limb stiffness (k_{limb}) (A), peak limb force (Peak F_{limb}) (B), change in limb spring length (ΔL_{limb}) (C), and RSP compression (ΔL_{RSP}) (D), centre of mass velocity during stance phase (E), contact time (F) and running speed (G). Statistically significant main effects for limb (a), RSP stiffness (b) and statistically significant limb*RSP stiffness interaction effects (c) are denoted on the figure and in Table 2.

compared to the prosthetic limb (Table 2, Fig. 2). These effects were not influenced by RSP stiffness. Running speed was unaffected by RSP stiffness (Table 2, Fig. 2).

3.3. *Joint kinetics*

Negative work at the ankle/unified deformable segment and hip joint were similar between limbs and not significantly affected by RSP stiffness (Fig. 3). Irrespective of RSP stiffness, the intact limb ankle ($p<0.02$) and knee ($p<0.03$) joints did significantly more positive work,

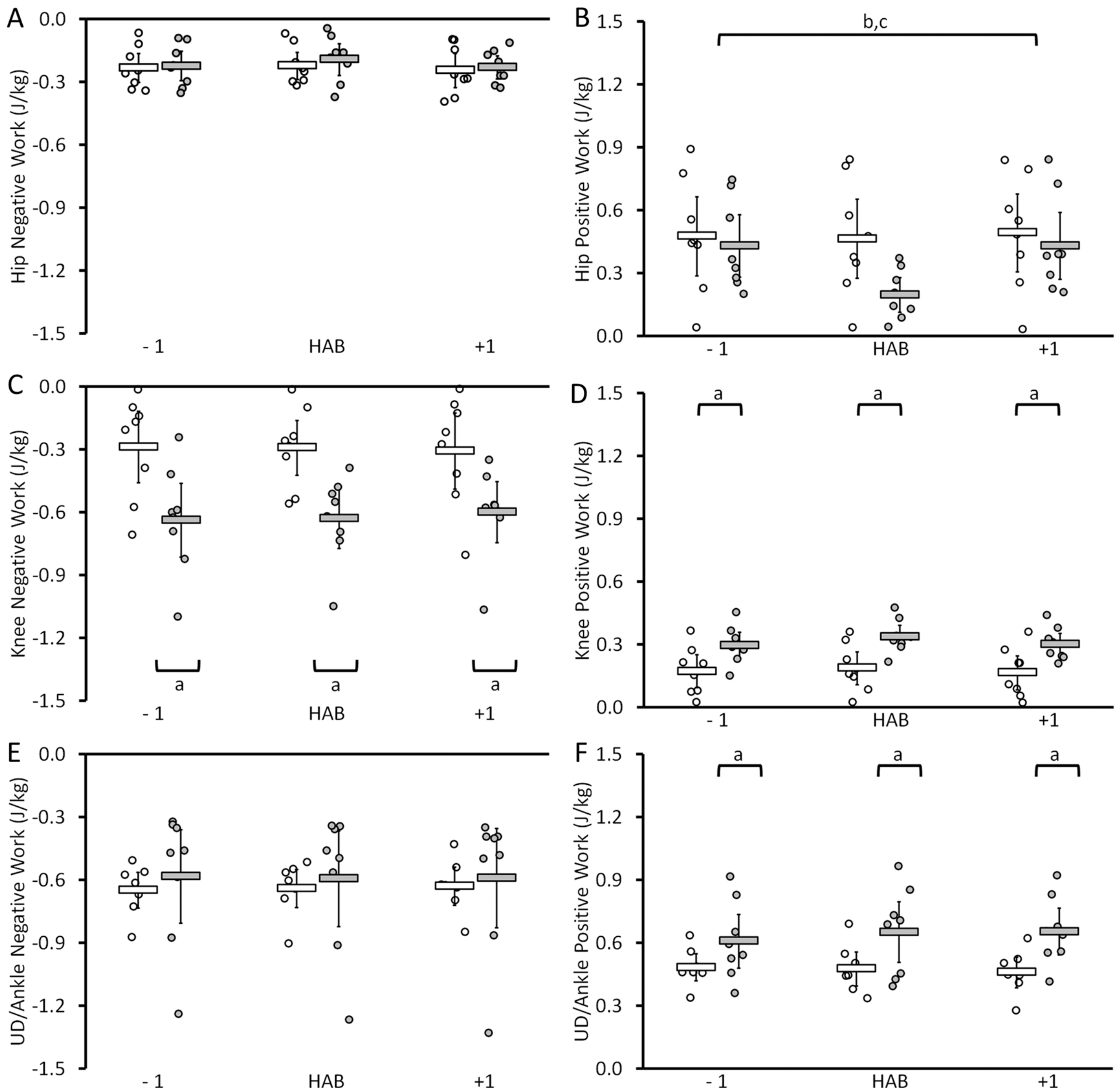

Fig. 3. Group mean ± 95% confidence intervals and individual participant mean data representing the prosthetic (white) and intact (grey) limbs for each RSP condition. Negative (A, C and E) and positive (B, D, F) work done by the hip and knee joints, as well as the intact ankle joint and RSP are presented. Statistically significant main effects for limb (a), RSP stiffness (b) and statistically significant limb*RSP stiffness interaction effects (c) are denoted on the figure and in Table 2.

with the intact limb knee joint also doing significantly more negative work ($p<0.03$), than the corresponding joints in the prosthetic limb (Table 2, Fig. 3). There was a significant limb by RSP category interaction for hip positive work ($p<0.02$); values were reduced for the intact compared to prosthetic limb across all stiffness conditions, with differences between limbs being greater for the HAB RSP condition compared to +1 and -1.

4. Discussion

The aim of the current study was to assess how varying RSP stiffness affected spring-mass behaviour and consequently, sub-maximal running performance in individuals with UTA. We provide the first quantifiable evidence that acute changes to RSP stiffness differentially affect the mechanical demands and running performance on the prosthetic and intact limbs during running at submaximal, self-selected running speeds in UTA.

Increasing RSP stiffness led to an increased limb stiffness in the prosthetic limb, but differentially affected intact limb stiffness. The finding that peak limb force remained unchanged across RSP stiffness' and was consistently higher in the intact vs. prosthetic limbs, agrees with previous work [9,10,28]. During prosthetic limb stance, limb spring length decreased as RSP stiffness increased, whilst, intact limb spring length was greater as RSP stiffness increased. Prosthetic limb stiffness changes occurring with changes in running speed originate from changes in limb spring length [4,29]. Therefore, changes in RSP stiffness seem to affect spring-mass behaviour by altering spring compression. In the prosthetic limb, increasing limb stiffness and decreasing limb spring compression were noted, as RSP stiffness increased. Coupled with no change in RSP compression, this implied that some adaptation occurred at the knee and hip joints in this limb, across RSP stiffness conditions. The adaptation in 'spring length' may potentially represent a control feature of UTA running. The merits of this require further exploration however, they may include maintaining a COM sagittal plane arc-trajectory during the ground contact phase that minimises inter-limb asymmetries in whole body COM height changes and/or in step length. This would minimise the joint kinetic adaptations that would otherwise be required, which will be physiologically beneficial in terms of running energetics. However, these assertions are speculative and require empirical verification.

Table 2
Outcomes of statistical analyses: F statistic, p value and effect size (partial eta squared, η_{p^2}) for each outcome measures' main and interaction effects. Alpha (p) values are reported from the ANOVA models. Those above the stated false discovery rate threshold (20%; indicating statistical significance) are presented in their unadjusted form in shaded grey boxes.

Outcome measure	Limb*RSP Stiffness Interaction	RSP Stiffness	Limb
Running Speed ($m.s^{-1}$)		$F(2,14) = 1.55, p=0.247, \eta_p^2 = 0.18$	
Centre of mass velocity ($m.s^{-1}$)	$F(2,14) = 0.38, p=0.690, \eta_p^2 = 0.05$	$F(2,14) = 1.35, p=0.291, \eta_p^2 = 0.16$	$F(1,14) = 11.78, p=0.011, \eta_p^2 = 0.63$
Contact Time (s)	$F(2,14) = 1.91, p=0.185, \eta_p^2 = 0.21$	$F(2,14) = 0.76, p=0.486, \eta_p^2 = 0.10$	$F(1,14) = 6.51, p=0.038, \eta_p^2 = 0.48$
K_{limb} (dimensionless)	$F(1.21,8.47) = 6.74, p=0.026, \eta_p^2 = 0.49$	$F(1.13,8.47) = 0.70, p=0.512, \eta_p^2 = 0.09$	$F(1,14) < 0.01, p=0.955, \eta_p^2 < 0.01$
Peak F_{limb} (BW)	$F(2,14) = 0.50, p=0.618, \eta_p^2 = 0.066$	$F(2,14) = 1.04, p=0.381, \eta_p^2 = 0.13$	$F(1,14) = 6.10, p=0.043, \eta_p^2 = 0.47$
ΔL_{limb} (m)	$F(2,14) = 13.85, p<0.001, \eta_p^2 = 0.66$	$(2,14) = 1.70, p=0.219, \eta_p^2 = 0.20$	$F(1,14) = 3.85, p=0.090, \eta_p^2 = 0.36$
ΔL_{RSP} (m)		$F(2,14) = 0.57, p=0.578, \eta_p^2 = 0.08$	
UD/Ankle Negative Work (J/kg)	$F(2,12) = 0.49, p=0.625, \eta_p^2 = 0.08$	$F(2,12) = 0.24, p=0.792, \eta_p^2 = 0.04$	$F(1,12) = 0.02, p=0.889, \eta_p^2 < 0.01$
UD/Ankle Positive Work (J/kg)	$F(2,12) = 2.87, p=0.096, \eta_p^2 = 0.32$	$F(2,12) = 0.62, p=0.555, \eta_p^2 = 0.10$	$F(1,12) = 12.81, p=0.012, \eta_p^2 = 0.68$
Knee Negative Work (J/kg)	$F(2,14) = 0.87, p=0.439, \eta_p^2 = 0.11$	$F(2,14) = 0.05, p=0.948, \eta_p^2 = 0.01$	$F(1,14) = 7.48, p=0.029, \eta_p^2 = 0.52$
Knee Positive Work (J/kg)	$F(2,14) = 0.67, p=0.527, \eta_p^2 = 0.09$	$F(1.23,12.38) = 3.16, p=0.074, \eta_p^2 = 0.31$	$F(1,14) = 8.00, p=0.025, \eta_p^2 = 0.53$
Hip Negative Work (J/kg)	$F(2,14) = 0.15, p=0.861, \eta_p^2 = 0.02$	$F(2,14) = 1.17, p=0.340, \eta_p^2 = 0.14$	$F(1,14) = 0.19, p=0.678, \eta_p^2 = 0.03$
Hip Positive Work (J/kg)	$F(2,14) = 5.44, p=0.018, \eta_p^2 = 0.44$	$F(2,14) = 27.98, p<0.01, \eta_p^2 = 0.80$	$F(1,14) = 1.95, p=0.206, \eta_p^2 = 0.22$

Body Weights (BW); Unified deformable segment (UD).

The current study found that varying RSP stiffness did not influence overall running speed, as reported previously [30]. This was achieved, despite contact time being shorter and stance-phase COM velocity being higher on the intact limb compared to prosthetic limb. Thus, it appears the consistency in running speed was achieved by modulating limb stiffness, which corroborate previous reports [4].

In the current study, the intact limb hip joint performed more positive work during running with a non-habitual stiffness RSP (+1 and -1 conditions), whilst positive hip work in the prosthetic limb was unchanged across RSP stiffness'. Increased positive work suggests greater positive musculo-tendon work. This supports the idea that in UTA individuals the intact limb, specifically the hip joint musculature, plays a key role in modulating running performance when using an RSP, which agrees with previous research [10]. The intact limb's ankle and knee did more stance-phase positive work and the knee more negative work, than that done on the prosthetic limb, which is again in agreement with previous reports [10] but the amount of work was unaffected by RSP stiffness condition.

The amount of negative work done at the hip and ankle/RSP were similar between limbs and across RSP stiffness' suggesting the active ability (i.e. power generation) of the intact limb was mainly responsible for the inter-limb differences in spring-mass behaviour. This may relate to the propulsive requirements placed on the intact limb [9]. The current study shows that physically active individuals with UTA rely heavily on the intact limb to do positive work during stance phase to achieve the desired running speed, regardless of RSP stiffness category. This effect is amplified when using an RSP with non-habitual stiffness i.e. an RSP stiffness outside of the range prescribed using current prescription guidelines based on user weight and physical activity. Therefore, selection of an RSP stiffness category outside of this range and its effect on hip joint work could, particularly because of already increased involvement of this joint when using an RSP, lead to earlier fatigue of hip musculature. Whilst speculative, this may also relate to reports of the development of hip joint osteoarthritis [31] and the asymmetries in dynamic balance of running [28] in this population.

The familiarisation period given to the non-habitual prosthetic stiffness category and the order of stiffness categories assessed were limitations as participants may not have been equally accustomed to all RSP stiffness conditions. Insights from long-term adaptation are worthy of further investigation, as are studies drawing from a larger sample size, to increase confidence in the reported statistical effects. The focus on self-selected running speed, means that the adaptations to RSP stiffness category described in the current study cannot be extended to other running speeds. Finally, it is not clear if the self-selected running speeds reported in the current study were a valid representation of participants' self-selected running speeds during prolonged running.

5. Conclusion

Acute changes in RSP stiffness in individuals with UTA running at self-selected submaximal speed led to significant changes in limb spring compression, and in hip joint positive work done but had little effect on running speed, stance-phase COM velocity and contact time. The amount of joint work done in the intact limb was greater than that done on the prosthetic limb particularly when using non-habitual RSP stiffness' and this reflects a reliance on the intact limb during running.

CRediT authorship contribution statement

CTB, ADA, TKS, JGB, RJF: Conceptualization. **CTB, ADA, TKS, JGB, RJF**: Methodology. **CTB, ADA**: Software. **CTB**: Validation. **CTB, ADA**: Formal analysis. **CTB, ADA, TKS, RJF**: Investigation. **CTB, RJF**: Resources. **CTB**: Data curation. **CTB**: Writing – original draft. **CTB, ADA, TKS, JGB, RJF**: Writing – review & editing. **CTB**: Visualization. **CTB, RJF**: Supervision. **CTB**: Project administration, Funding acquisition.

Conflict of Interest Statement

Authors report no conflicts of interest associated with the study.

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