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Dynamic elastic response prostheses alter approach angles and ground reaction forces but not leg stiffness during a start-stop task

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Dynamic elastic response prostheses alter approach angles and ground reaction forces but not leg stiffness during a start-stop task

Abstract

In a dynamic elastic response prosthesis (DERP), spring-like properties aim to replace the loss of musculature and soft tissues and optimise dynamic movement biomechanics, yet higher intact limb (IL) loading exists. It is unknown how amputees wearing a DERP will perform in start-stop movements and how altering the prosthetic stiffness will influence the performance and loading. This study assessed movement dynamics through comparisons in spatiotemporal, kinematic and kinetic variables and leg stiffness of intact, prosthetic and control limbs. The effect of prosthetic stiffness on movement dynamics was also determined. Eleven male unilateral transtibial amputees performed a start-stop task with one DERP set at two different stiffness – Prescribed and Stiffer. Eleven control participants performed the movement with the dominant limb. Kinematic and kinetic data were collected by a twelve-camera motion capture system synchronised with a Kistler force platform. Selected variables were compared between intact, prosthetic and control limbs, and against prosthetic stiffness using ANOVA and effect size. Pearson's Correlation was used to analyse relationship between leg stiffness and prosthetic deflection. Amputees showed a more horizontal approach to the bound during the start-stop movement, with lower horizontal velocities and a longer stance time on the IL compared to controls. In both stiffness conditions, the IL showed selected higher anteroposterior and vertical forces and impulses when compared to the controls. Leg stiffness was not significantly different between limbs as a result of the interplay between angle swept and magnitude of force, even with the change in prosthetic stiffness. A main effect for prosthetic stiffness was found only in higher impact forces of the prosthetic limb and more horizontal touchdown angles of the I when using the prescribed DERP. In conclusion, amputees achieve the movement with a horizontal approach when compared to controls which may reflect difficulty of movement initiation with a DERP and a difficulty in performing the movement dynamically. The forces and impulses of the IL were high compared to control limbs. The consistent leg stiffness implies compensation strategies through other joints.

<u>Keywords:</u>

Transtibial amputee; Start-stop task; Stiffness; Dynamic elastic response prosthesis; Spring-mass model; Kinetics

1. Introduction

Physical activity is important for amputees to gain from the physiological and musculoskeletal health benefits associated with exercise. Though para-sports often require amputees to participate without their prosthesis (wheelchair tennis and basketball, sitting volleyball, football), the review by Deans et al. (2012) suggests that amputees would prefer to engage in inclusive, integrated sports wearing their prosthesis. However, amputees perceive that their prosthesis limits their performance (Deans et al., 2012).

Passive-elastic prostheses, commonly known as dynamic elastic response prostheses (DERP), have been developed to enable lower limb amputees to run (Nolan, 2008). The principle behind DERP is that the passive-elastic spring is deformed in loading and the stored energy is returned during unloading of the foot, contributing in forward propulsion (Zelik et al., 2011). The ideal prosthetic stiffness should vary according to the speed and dynamics of the movement performed. Currently, a DERP can only return the energy absorbed while in biological limbs, the powerful contraction of the ankle plantarflexors is the primary contributor to propulsive force production (Hamner et al., 2009) and running speed (Dorn et al., 2012). In dynamic non-locomotor movements, where the role of the ankle-foot complex in absorbing and generating the load is modulated depending on the task (Neptune et al., 2001; Zajac, 2002), amputee performance may be compromised due to the passive prosthetic spring, as evidenced in jumping (Schoeman et al., 2012; 2013). A prosthetic spring that modulates in response to the demands of the movement in a similar way to an intact active ankle is not viable due to weight and material limitations. Manufacturers recommend prosthetic spring stiffness to be based on the weight and activity requirements of the user. However, there is little research to indicate the effect of prosthetic stiffness on the performance of the athlete or on potential injury mechanisms (Butler et al., 2003), other than Wilson et al. (2009), who found that prosthetic stiffness has significant effect on leg stiffness of the intact limb (IL) in a case study of a transtibial amputee male sprinter.

If DERP are to be used in sports, they must allow amputees to complete a variety of dynamic, non-locomotor manoeuvres effectively. Since DERP have been designed for steady state running (Nolan, 2008) and are optimal at the natural frequency of the spring, their performance in other activities may be compromised resulting in undue strain on the IL. Past research on amputee walking and running has suggested that (i) the IL is overloaded compared to the prosthetic limb (PL) (Rossi et al., 1995; Snyder et al., 1995; Grabowski et al., 2009) and when compared to the biological limb of an able-bodied control group (Baum et al., 2016) as a result of the reduced push-off contralaterally (Morgenroth et al., 2011) and (ii) the increased force is responsible for increased leg stiffness on the IL (Hobara et al. 2013; McGowan et al. 2012). How and why the increased loading occurs is unclear due to a number of limitations. Past research has executed the analyses using many different types of prostheses (Hobara et al 2013, Baum et al 2016) which may influence the results. Research has been conducted on walking and running with variable step lengths between limbs which results in different velocities at touchdown (TD) and take off (TO). Consequently, the impulse required against the ground is adjusted to ensure ongoing stepping. Finally, the influence of manufactured prosthetic stiffness on the loading and leg stiffness may play a role and has not been assessed.

It is not known how amputees wearing a DERP will perform in start-stop movements common in court-based sports, how the limbs will experience the load to enable the movement and how altering the prosthetic stiffness will influence the performance. To investigate this, we chose a controlled start-stop movement to assess the ability of the amputees to initiate movement, bound forward dynamically on the assessed limb and then stop the movement in a stable erect stance (Fig 1). We chose to focus on the stance phase of the bound to assess the biomechanics on the intact and prosthetic limbs and the effect of stiffness. As this stance phase will be affected by the initial and terminal conditions, we chose to control these by defining the distances to be achieved in each part of the movement. The purpose of this research in analysing a controlled forward movement is to develop our understanding of mechanisms which underpin movement mechanics and loading in amputees during activities other than walking and running.

Thus, the first aim of the research was to determine spatiotemporal, kinematic and kinetic differences between the IL, PL and control limb (CL) when executing a start-stop dynamic movement using their Prescribed DERP. The second aim was to determine the effect of prosthetic spring stiffness on the movement performance. We hypothesised that there would be no between-limb difference in spatiotemporal (initiation time, stance time, flight time, and movement completion time) and kinematic variables (TD and TO angles, and pelvis velocity) since the controlled nature of the task would require a similar execution. We hypothesised that the kinetic variables (peak vertical and anteroposterior forces and impulses) and leg stiffness (K_{leg}) would be greater on the IL and less on the PL compared to each other and to the dominant biological limb of an able-bodied control group (CL) as a result of the presence of the prosthesis. Finally, we hypothesised that a change in prosthetic stiffness would have no effect on kinematic variables but would increase the loading and leg stiffness variables on the PL and IL.

2. Methods

2.1. Participants

Eleven male unilateral traumatic transtibial amputees (Mean [SD]: 36 [8] years, 83 [21] kg, 1.80 [0.09] m) for the study group (SG) and eleven male able-bodied individuals (35 [10] years; 77 [12] kg; 1.81 [0.06] m) for the control group (CG) were recruited. All participants provided informed consent to an Institutional Review Board-approved protocol and were asymptomatic of neurological or musculoskeletal disorders. Amputee participants were all involved in recreational physical activity and had at least six months' experience using the same brand of DERP (BladeXT; Blatchford & Sons Ltd., Hampshire, UK) without the requirement of additional mobility aids.

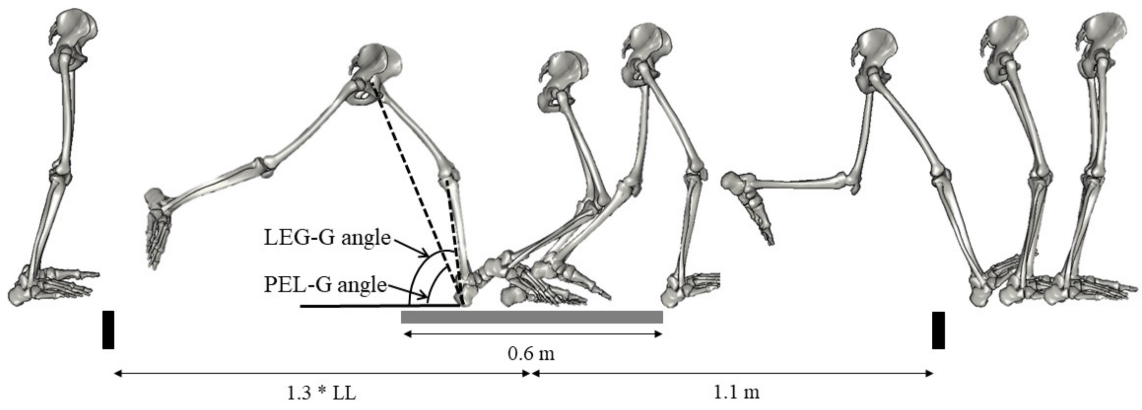

Figure 1: An illustration of the start-stop task performed by participants in this study. A mark on the floor indicated the initial take off position at a distanced normalized to the participant's leg length (1.3*LL) to a mark at the centre of the force platform. Distance between this mark and the stop zone, which was marked with tape was fixed at 1.10 m. The participants were asked to land with their heels beyond the stop-zone mark. The LEG-G and PEL-G angles at touchdown are shown.

2.2. Procedures

Spatiotemporal and kinematic data were collected by a twelve-camera motion capture system (Vicon, Oxford, UK) at 120 Hz using sixteen retro-reflective markers according to the lower body model created by Davies et al. (1991). Three extra markers (piston, blade, and toe) were used to model the elastic blade of the DERP (Fig 2). Kinetic data were collected in synchrony with kinematic data via a Kistler force platform (400x600mm) at 1000 Hz.

Figure 2: The dynamic elastic response prosthesis used by amputee participants in this study – BladeXT (Blatchford & Sons Ltd., Hampshire, UK; http://www.blatchford.co.uk/endolite/bladext/), with the three extra markers (*)

Following a five-minute self-directed warm-up, participants performed the start-stop task (Fig.1), which was demonstrated to the participants. The instruction given to all participants was the same: "Perform a two-step forward movement during which you need to be airborne for each step. You must start at the first mark, leap forwards to land with the middle of your foot on the second mark. Then, as quickly as possible, leap forwards again to land with the other foot beyond the third mark. Here you need to stop and bring the opposite foot down to balance in standing. Hold your balance until I say that you are finished". Although this particular movement is unlikely to be replicated during court games, imposing these controls allowed a mechanistic analysis of dynamic landing while maintaining some ecological validity. Initial take off occurred at a distance normalized to 1.3 times the participant's leg length, defined as the distanced from the greater trochanter to the ground measured in standing. This distance ensured that a flight phase was required to execute the task. This distance was adjusted for the both intact and prosthetic limbs. A minimum distance of 1.10 m from the force platform was set for the final landing, since some study participants found it difficult to target a mark that was set at a distance normalised to LL. The minimum threshold allowed all participants to execute the movement as instructed.

SG participants performed the start-stop task first with the DERP at the stiffness prescribed on prosthetic fitting (DERPp). Then, a certified prosthetist fit the participants with a stiffer prosthesis (DERPs) and assured proper component alignment and fit. The BladeXT has 9 spring set ranges and a particular spring set is prescribed according to the user's body weight and activity level. Prosthetic stiffness was increased by one spring set in the range above the prescribed stiffness, as per prosthesis manufacturer guidelines (Blatchford & Sons Ltd.), ensuring a standard relative increase in prosthetic stiffness amongst SG participants. In the absence of research on the effect of prosthetic stiffness on injury risk and prosthesis structural integrity, the prosthetic stiffness was only increased by one spring set range to ensure the safety of the task for study participants. Participants were allowed a one-hour adjustment period before repeating the start-stop task with the DERPs. For each condition, the movement was also performed with the intact limb leading. CG participants performed the movement with the dominant limb leading. Limb dominance was defined as the preferred limb to kick a ball. Repeated trials were collected until five valid trials were captured. A valid trial was based on the foot placement on the force platform, achieving flight in each step and the ability to control movement termination, that is, successfully stop beyond the threshold without stepping forward.

2.3. Data analyses

Kinematic and kinetic data were filtered with a fourth order low-pass Butterworth filter and cut off frequency of 6 Hz and 30 Hz, respectively in Vicon Nexus (Version 2.2). Spatiotemporal variables of initiation time (start of movement to TD on force platform), stance time (ground contact time on force platform), time in loading during stance, flight time (TO from force platform to second TD) and movement completion time (start to end of movement) were also calculated. The start and end of the movement were determined by the horizontal displacement of the PEL marker as these events did not occur on a force plate. A threshold of 2mm determined the start and end of the motion and the displacement-time curve was visually inspected to ensure the robustness of the result. Velocity of the PEL marker was differentiated to calculate velocity which was reported at TD and TO. The beginning and end of stance was defined by a vertical ground reaction force (GRF) threshold of 20 N. Stance was divided into loading and propulsion at the point where the anteroposterior (AP) GRF equalled zero (Fy0) (Fig. 3). GRFs were measured as discrete peak values. Impulses were calculated via the trapezoidal rule integration method. GRFs and impulses were normalized to body mass. The loading rate was calculated by the slope of the force-time curve from the time required for the vertical GRF to rise from 50 N to body weight plus 50 N, and vice versa for the decay rate (Munro et al., 1987).

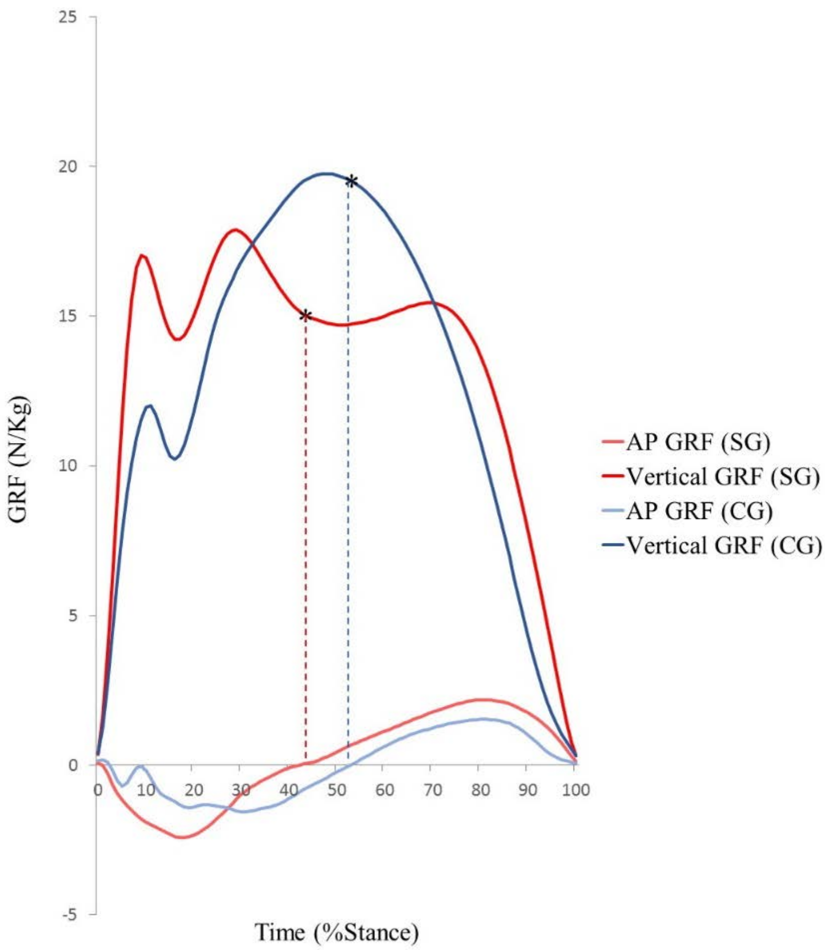

Figure 3: An example of the typical ground reaction force (GRF) time profiles seen in this study. Vertical GRF curves for the PL and the CL are illustrated in red and blue respectively. Fy0 is indicated by a dotted line, dividing the stance into loading and propulsion phase. Discrete measures for the vertical GRF at Fy0 are indicated with an *.

A pelvic marker (PEL) generated by Vicon's Plug-In-Gait which represents the origin point of the pelvis segment was used as a marker of the centre of mass (CoM) (Glaister et al., 2007). The global angle of attack was defined by the PEL to the heel (PEL-G) and right horizontal at TD and the angle of take-off was defined by the PEL to toe and right horizontal at TO. These were determined through coordinate data. The angle of the shank (LEG-G) against the heel and toe was also calculated through coordinate data using the knee marker to overcome any inaccuracies of a changing ankle joint in DERP. Angle and displacement data were restricted to the sagittal plane.

The leg stiffness during the loading phase (K_{leg}) was calculated using the spring mass model (Blickhan, 1989; McMahon & Cheng, 1990). Since an AP GRF of zero (Fy0) is chosen to represent the end of loading in this study, K_{leg} is calculated as the ratio of the vertical GRF to the lower body compression (ΔL) at Fy0 (Fig. 4).

$$K_{leg} = \text{vertical GRF at Fy0} / \Delta L \qquad (1)$$

ΔL was calculated from the maximum vertical displacement of PEL (Δz), the initial leg length (L_0) and the angle swept (θ) from initial contact till Fy0. L_0 was measured as the distance from the PEL to the Heel marker in standing. θ was defined by the angle formed between the PEL and Heel marker from heel strike until Fy0, measured via coordinate data.

$$\Delta L = \Delta z + L_0 (1 - \cos \theta) \qquad (2)$$

Prosthetic deflection was measured as the maximum displacement between the Piston and Toe extra markers during loading.

Averaged data from five valid trials were used for statistical analyses of all variables (mean and standard deviation [SD]).

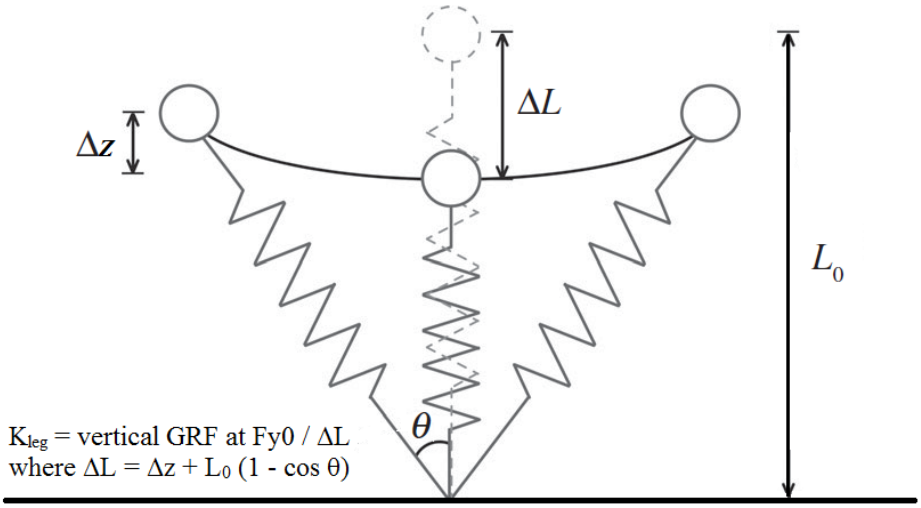

Figure 4: Diagram representation of the spring-mass model and the calculation of leg stiffness during the start-stop task.

2.4. Statistical analyses

Data was statistically analysed using SPSS version 22.0.0.1 (IBM Corp., USA). Upon meeting the assumptions of normality and homogeneity of variance, a three-way independent analysis of variance (ANOVA) was used to test for differences between the prosthetic, intact and control limbs for each condition (i.e. DERPp and DERPs). If a significant main effect was found ($\alpha = 0.05$), Planned Contrasts (Repeated & Simple) were used to determine which between-leg (IL, PL, CL) factors were significantly different when analysing ***movement dynamics***. A two-way repeated measures ANOVA was used to compare movement biomechanics in the SG when using two different levels of prosthetic stiffness (prescribed and stiffer) and thus analyse the effect of ***prosthetic stiffness*** on a start-stop movement. To determine clinical importance of significant differences, effect size (Hedges' *g*) was also calculated for each comparison according to Cohen (1988) using the pooled standard deviation, where by large, medium and small effect sizes are denoted by values greater than 0.8, 0.5 and 0.2 respectively. To determine if the change in mechanical stiffness of the DERP resulted in a change of deflection of the spring, Mann-Whitney Test was used to detect a statistical difference in prosthetic deflection between DERPp and DERPs, since deflection data was non-parametric. To determine if leg stiffness was associated with prosthetic deflection, Pearson's Correlation was used to analyse the relationship between K_{leg} and prosthetic deflection in each condition.

3. Results

3.1. Movement Dynamics

Our first aim was to determine any biomechanical differences between the IL, PL and CL when amputees execute a start-stop movement using their Prescribed DERP

3.1.1. Spatiotemporal & Kinematic variables

Within the SG, no significant differences or large effect sizes were found for the spatiotemporal and kinematic variables between the IL and PL (Table 1) but noteworthy differences were found between the SG and the CG. There was no difference in movement completion time when performing the start-stop movement. However, a shorter initiation time with a large effect (g=1.13) was seen for bounding onto the PL compared to the CL. A longer stance time, with a large effect (g=0.88), was seen when bounding on to the IL compared to the CL.

All angles of attack at TD (PEL-G & LEG-G) for the IL and PL were significantly lower (more horizontal) compared to the CL with a large effect size. For PL, the TO angle (LEG-G) was significantly lower with a very large effect (g=1.15, P=0.004) compared to the CL. Although TD and TO LEG-G angles tended to be lower on PL compared to IL, no significant or large effect was found within the SG participants (g=0.60-0.67).

The horizontal (AP) PEL velocity at both TD and TO was reduced when bounding onto the IL compared to the CL, with a large effect (TD: g=0.90; TO: g=1.05). No noteworthy results were found for the vertical PEL velocity at TD or TO.

3.1.2. Kinetic variables

Within the SG, no significant differences or large effect sizes were found for the peak GRFs and impulses between the IL and PL (Table 2). However, noteworthy differences were found when the IL and PL were compared to CL. The AP GRF braking peak for the IL was significantly higher compared to the CL with a large effect size (g=0.88, P=0.009). The vertical GRF impact peaks were higher with a large effect size for the both IL (g=0.80) and PL (g=0.97) compared to the CL. At Fy0, both the IL and PL showed significantly lower vertical GRF with very large effect sizes when compared to CL (IL: P=0.003, g=1.17; PL: P<0.001, g=1.27). There were no between-limb differences for the horizontal or vertical GRF peaks in the propulsion phase.

Braking AP impulses were greater for the IL compared to CL, with a large effect size (g=0.82). Horizontal and vertical propulsive impulses of the IL were higher compared to CL, with a large effect size (AP GRF: g=0.91; vertical GRF: g=0.80). The IL also showed a larger total vertical impulse (g=0.80) compared to the CL.

The IL tended to experience, on average, the higher loading rates while the PL experienced the lowest loading rates when compared to the CL. However, there was no significant difference or effect in this result. No noteworthy results were found for the decay rate.

3.1.3. Leg Stiffness (K_{leg})

Although K_{leg} of the PL was marginally higher than the K_{leg} of the IL or CL, no results were significantly different or indicated a large effect size.

3.2. Prosthetic Stiffness

Our second hypothesis assessed the effect of the prosthetic stiffness on the IL and PL. There was no difference in deflection (U = 39.0 p = 0.436 r = 0.23) between the prescribed prosthesis (Median: 0.025 m) and the stiffer prosthesis (Median: 0.024 m).

In the DERPs condition, a significantly longer initiation time with a large effect (g=0.87, P=0.031) is seen when bounding onto the IL compared to the PL. No main effect of prosthetic stiffness was found in spatiotemporal and kinematic variables except for TD angles. The IL showed a significantly less horizontal approach (PEL-G: P=0.007, g=0.29; LEG-G: P=0.005, g=0.49) on TD when using the DERPs compared to the DERPp, although the difference in TD angle is small (PEL-G: 1.3^{o}; LEG-G: 1.5^{o}). The vertical PEL velocity at TD was lower for PL in the DERPs condition compared to the CL, with a large effect (g=0.83).

Apart from impact GRF peaks which were significantly higher for the PL when using the DERPp compared to the DERPs (P=0.034, g=0.33), no main effect of prosthetic stiffness was seen in the selected kinetic variables. However, a large effect for the loading rate was found between the IL and PL in DERPs (g=0.90).

There is no main effect of prosthetic stiffness on K_{leg} of either the intact or prosthetic limb in the SG.

No significant correlations for prosthetic deflection with K_{leg} were present in the DERPp condition (Intact: $P = 0.060$, R = 0.522; Prosthetic: $P = 0.260$, R = 0.232) or the DERPs condition (Intact: $P = 0.246$, R = 0.227; Prosthetic: $P = 0.402$, R = 0.090).

Table 1: Mean and Standard Deviation [SD] of Spatiotemporal and Kinematic variables for the Control Group (CG) and Study Group (SG) participants when performing the start-stop task, and bounding onto the control, intact and prosthetic limbs respectively. Large Hedges' g Effect Size (*) and significant p values for between limb differences in both the Prescribed DERP and Stiffer DERP condition are reported. Only significant p values are reported for the main effect of stiffness (i.e. differences between DERPp and DERPs) since no large effect sizes (Hedges' g) were found.

	Control	**Prescribed DERP**			**Stiffer DERP**			**Stiffness**
		Intact	**Prosthetic**	**Large effect size**	**Intact**	**Prosthetic**	**Large effect size**	**Significant *P* value**
Initiation Time (s)	0.81 [0.08]	0.74 [0.10]	0.69 [0.08]	PvC	0.76 [0.10]	0.68 [0.06]	IvP^{+} PvC^{++}	
Stance Time (s)	0.44 [0.07]	0.50 [0.06]	0.50 [0.09]	IvC	0.49 [0.08]	0.49 [0.10]	/	
Flight Time (s)	0.13 [0.05]	0.10 [0.04]	0.12 [0.04]	/	0.11 [0.04]	0.12 [0.05]	/	
Movement completion time (s)	2.06 [0.12]	2.04 [0.16]	2.01 [0.16]	/	2.02 [0.16]	1.96 [0.17]	/	
Time in Loading (%Stance)	50.88 [7.09]	47.24 [7.10]	52.16 [9.35]	/	50.30 [9.07]	52.04 [9.69]	/	
TD angle – PEL-G (º)	72.60 [3.98]	66.89 [4.64]	66.55 [6.00]	IvC^{+} PvC^{+}	68.21 [4.34]	66.90 [6.37]	IvC PvC^{+}	I^{+}
TD angle – LEG-G (º)	88.66 [4.29]	82.27 [2.84]	78.47 [6.60]	IvC^{+} PvC^{++}	83.81 [3.06]	79.65 [7.73]	IvC^{+} PvC^{++}	I^{+}
AP PEL velocity at TD (m/s)	2.35 [0.16]	2.18 [0.17]	2.25 [0.19]	IvC	2.18 [0.17]	2.26 [0.16]	IvC	
Vertical PEL velocity at TD (m/s)	-0.48 [0.17]	-0.36 [0.21]	-0.34 [0.30]	/	-0.42 [0.19]	-0.29 [0.23]	PvC	
TO angle – PEL-G (º)	70.85 [5.21]	68.64 [5.35]	68.45 [3.60]	/	69.94 [4.81]	68.37 [4.24]	/	
TO angle – LEG-G (º)	71.91 [6.12]	68.35 [8.60]	64.05 [4.23]	PvC^{+}	67.78 [7.84]	63.52 [5.20]	PvC^{++}	
AP PEL velocity at TO (m/s)	2.44 [0.17]	2.26 [0.09]	2.35 [0.19]	IvC	2.27 [0.17]	2.41 [0.25]	IvC	
Vertical PEL velocity at TO (m/s)	0.32 [0.38]	0.11 [0.25]	0.27 [0.25]	/	0.21 [0.33]	0.30 [0.30]	/	

Abbreviations: I – intact limb; P – prosthetic limb; C – control limb; TD – touchdown; TO - take off; AP – anteroposterior

Note: The indicated limb represents the limb performing the bound.

$^{+}$ $P < 0.05$
$^{++}$ $P < 0.001$

*The magnitude of Hedges' *g* may be interpreted using Cohen's (1988) convention as small (0.2-0.5), moderate (0.5-0.8), and large (>0.8) effect sizes. Only large effect sizes are reported.

Table 2: Mean and Standard Deviation [SD] of Kinetic variables and Leg Stiffness for the Control Group (CG) and Study Group (SG) participants when performing the start-stop task, and bounding onto the control, intact and prosthetic limbs respectively. Large Hedges' g Effect Size (*) and significant p values for between limb differences in both the Prescribed DERP and Stiffer DERP condition are reported. Only significant p values are reported for the main effect of stiffness (i.e. differences between DERPp and DERPs) since no large effect sizes (Hedges' g) were found.

	Control	**Prescribed DERP**			**Stiffer DERP**			**Stiffness**
		Intact	**Prosthetic**	**Large effect size**	**Intact**	**Prosthetic**	**Large effect size**	**Significant *P* value**
Peak GRF – AP (N/Kg)								
Brake	-2.53 [0.99]	-4.18 [2.12]	-3.07 [1.45]	IvC^{+}	-4.08 [2.17]	-2.77 [1.05]	IvC^{+}	
Propulsion	1.84 [0.48]	2.11 [0.51]	1.95 [0.62]	/	2.26 [0.63]	1.97 [0.59]	/	
Peak GRF – vertical (N/Kg)								
Impact	12.77 [1.37]	16.00 [5.55]	16.00 [3.74]	IvC PvC	15.95 [4.92]	14.89 [2.65]	IvC PvC	P^{+}
Loading	15.77 [1.05]	16.09 [1.88]	15.59 [2.62]	/	16.05 [2.12]	15.67 [2.49]	/	
Propulsion	15.21 [1.51]	14.15 [1.86]	14.04 [1.75]	/	14.86 [1.56]	14.49 [2.06]	/	
At Fy0	16.38 [2.48]	12.90 [2.02]	12.33 [2.05]	IvC^{+} PvC^{++}	13.04 [1.98]	12.67 [3.37]	IvC^{+} PvC^{+}	
Impulse – AP (Ns/Kg)								
Brake	-0.26 [0.10]	-0.36 [0.13]	-0.35 [0.18]	IvC	-0.36 [0.13]	-0.33 [0.16]	IvC	
Propulsion	0.21 [0.05]	0.28 [0.08]	0.28 [0.14]	IvC	0.28 [0.10]	0.28 [0.13]	/	
Total	-0.04 [0.12]	-0.08 [0.11]	-0.07 [0.13]	/	-0.09 [0.12]	-0.06 [0.17]	/	
Impulse – vertical (Ns/Kg)								
Loading	2.87 [0.65]	2.97 [0.41]	3.15 [0.76]	/	3.10 [0.57]	3.02 [0.81]	/	
Propulsion	2.22 [0.42]	2.57 [0.39]	2.39 [0.71]	IvC	2.44 [0.56]	2.37 [0.73]	/	
Total	5.09 [0.54]	5.54 [0.46]	5.54 [1.08]	IvC	5.54 [0.64]	5.38 [0.95]	/	
Rate of Force (N/Kg/s)								
Loading Rate	445.55 [185.5]	508.12 [190.0]	392.99 [112.2]	/	533.24 [198.4]	351.45 [147.3]	IvP	
Decay Rate	138.37 [40.86]	121.91 [31.91]	139.07 [40.49]	/	135.95 [37.65]	135.85 [35.73]	/	
Leg Stiffness								
K_{leg} (N/Kg/m)	114.93 [33.72]	118.80 [39.25]	125.85 [74.66]	/	110.23 [43.28]	137.49 [77.66]	/	

Abbreviations: I – intact limb; P – prosthetic limb; C – control limb; AP – anteroposterior

Note: The indicated limb represents the limb performing the bound.

$^{+}$ $P < 0.05$
$^{++}$ $P < 0.001$

*The magnitude of Hedges' *g* may be interpreted using Cohen's (1988) convention as small (0.2-0.5), moderate (0.5-0.8), and large (>0.8) effect sizes. Only large effect sizes are reported.

4. Discussion

The present study aimed to assess the performance of transtibial amputees, whose ankle cannot adapt to specific movement requirements like an anatomical ankle, during a start-stop movement when using a dynamic prosthesis. The study aimed to determine between-limb differences in spatiotemporal, kinematic and kinetic variables and leg stiffness when the initial and terminal conditions which influence the loading were controlled. Finally, the research aimed to determine if a change in prosthetic stiffness, which may reflect an alteration to the ankle mechanism, affects movement performance. We controlled the start-stop movement by fixing the step lengths, to be able to compare the bounding phase loading patterns

We found that for amputees to achieve the start-stop task, there were some significant changes in performance between the SG and the CG. Results indicate that amputees had a more horizontal approach to the bound during the start-stop movement, when compared to controls. When bounding onto the IL the amputee participants also experienced longer stance times, and slower horizontal velocities at TD and TO. We found that there was an effect of limb on the loading variables but no effect on leg stiffness. On the IL, the higher braking forces and impulses compared to the CG in conjunction with the lower force at Fy0 suggests a less vertically dynamic movement. We had expected that altering the prosthetic stiffness would mimic an ankle adaptation and that this would influence the performance of the movement and limb loading. A main effect for prosthetic stiffness was found only in higher impact GRF peaks of the PL and more horizontal TD angles of the IL when using the DERPp. It must be noted that prosthetic stiffness was only increased by one spring set in the stiffer condition and no statistical difference in deflection was found between conditions. Thus, attention should be focused on dynamic movement execution rather than prosthetic stiffness to promote performance and prevent injury.

In controlling the step length to approach we had attempted to control the velocity at TD, thus ensuring the mechanisms of loading could be realised under controlled conditions. The amputees manipulated the TD velocity by altering the angle of attack at TD. It is likely that the reduced forward velocity and horizontal limb angle when bounding onto the IL compared to the CL reflects the difficulty in starting and stopping the movement on the prosthesis contralaterally. The lower velocity at TD is likely due to the reduced push-off from the prosthesis in the previous stance and is associated with difficulty to initiate movement and achieve flight in amputees (Vrieling et al., 2008; Strike & Diss, 2005). The lower velocity of the IL at TO suggests a controlled propulsion to enable the amputees to stop on the prosthesis. As indicated in the methods (Section 2.2), we had to adapt the second bound distance based on the first two amputee participants who found stopping on the prosthesis at the 1.3*LL target more difficult than control participants. In addition to the reduced horizontal velocity of the IL, the peak braking force and braking impulse were higher compared to the CL, suggesting a particular loading pattern is adopted by the amputees to control the movement. It seems that movement initiation and termination was more influenced by the prosthesis than the inability to maintain momentum in the bound. Future work investigating the initiation and termination steps is required to understand the influence of the contralateral limb on TD and TO velocity and thereafter on limb loading.

When bounding onto the PL, the more horizontal limb angle and tendency towards reduced horizontal velocity compared to the CL also suggests a more horizontal movement. Past research on running suggests that amputees land with a more vertical limb to ensure knee stability and loading of the prosthetic spring (Sanderson & Martin, 1996). Thus, we expected that the amputee would land in a more vertical position when bounding onto the prosthesis. In this study, the more horizontal motion in conjunction with the tendency towards higher horizontal forces and impulses may be a generally more cautious performance by the amputees or a strategy to obtain more energy return from the DERP. In past research running prostheses have been aligned to enable a “running on the toes” posture (Nolan 2008) using running specific prostheses (RSP) which do not have a heel spring. The different design of the Blade XT, with its heel component (Fig 2), may have both engaged the heel spring and allowed for a more heel-strike pattern. However, this may have resulted in higher impact vertical GRF peaks compared to CG participants.

There was an effect of limb on selected loading variables. Past research in walking and running (Baum et al., 2016) has indicated an increased loading on the IL, though it is unclear if this is as a result of the variability of the TD and TO velocities which are not controlled in locomotion analyses which require ongoing stepping. Though we tried to control the velocity at TD to assess the limb loading independent of velocity, the technique used by the amputees manipulated the velocity, suggesting that adapting loading is a priority for this population. The more vertical dynamic movement of the CG is underlined by the higher vertical force at Fy0 and the difference in force curves compared to the SG (Fig. 3). In the CG, the shorter stance time and the shape of the vertical GRF curve (i.e. a single active peak) is typical of plyometric activities such as running on the toes (Williams et al., 2000) and drop jumping (McKay et al., 2005). In the SG, the multiple active peaks for both the IL and PL in both conditions is more typical of walking (Geyer et al., 2006) and likely reflects the amputees moderating the use of the prosthetic spring or engaging a separate loading of the heel and toe springs. This, in conjunction with the consistent spring deflection with DERPp and DERPs, indicates the inability of the prosthesis to adapt the 'ankle' spring to this start-stop movement. During initial loading the amputees showed greater impact vertical force peaks than controls, but the loading rate of the PL tended to be lower than in the CL and IL, however there is large variability in the result. These results suggest that, as well as the prosthesis limiting force generation at Fy0 as previously found in running (Grabowski et al., 2009), the elastic properties of this DERP limit loading rates. This is likely to prevent injury of the residual limb (Dudek et al., 2005; Milner et al., 2005). The high impact peaks and tendency towards higher loading rates of the IL agrees with past research on running (Hobara et al., 2014) and implies that higher loading persists across many movements and supports the suggestion that amputees are at risk of greater chronic injury risk such as osteoarthritis of the IL (Norvell et al., 2005; Lloyd et al., 2010; Gailey et al., 2008).

Despite the reduced force at Fy0, K_{leg} did not differ between IL, PL and CL in this study. This contrasts with previous literature on running with an RSP which found that leg stiffness of IL to be greater than the PL during running in transtibial amputees (Hobara et al., 2013; Wilson et al.,2009), which was linked to the inability to generate GRFs with a RSP (McGowan et al., 2012). In walking and running research, the loading experienced by each of the IL and PL in comparison to CLs may be influenced by the asymmetrical step lengths and the reliance on step frequency to increase running speed (McGowan et al., 2012). In this study, the K_{leg} result may be explained by the fact that, although we controlled the step lengths it was changes in the angle swept during loading (i.e. significant TD angles) that maintained leg stiffness. The leg stiffness results need to be interpreted with caution as the IL and PL appear to have contravened the requirements of a spring, the execution of the task was less dynamic than the CL and the peak force did not coincide around the time of Fy0. Compensation via modulation of knee and hip joint stiffness may also be present to maintain a constant K_{leg} in the absence of an active biological ankle joint (Farley & Morgenroth, 1999). Future work should assess the changes in leg stiffness through the stance phase to investigate why the timing of the peak loading force did not coincide with Fy0 for the SG.

Contrary to expectation, systematically modifying the prosthetic stiffness (i.e. DERPp vs DERPs) showed no statistical effect on prosthetic deflection or on K_{leg} for either the IL and PL. It seems that the difference in prosthetic stiffness may have been too small to cause sufficient changes in the 'ankle' to evoke a performance response or the movement was too conservative to require change. We had expected that the stiffness of the prosthesis would influence the performance, given Farley & Morgenroth (1999) concluded that ankle joint stiffness is the primary mechanism for leg stiffness adjustment. Furthermore, the ankle plantarflexor fascicle shortening velocities and their corresponding force production are known to alter with movement demands (Adamczyk & Kuo, 2015). Yet, we found few variables affected by prosthetic stiffness. The intact limb showed more horizontal pelvis and leg TD angles when using the DERPp compared to the DERPs, possibly due to initiating movement with a softer prosthetic spring which likely allowed a greater roll-over. Higher vertical impact GRF peaks for the PL were observed with the DERPp than the DERPs. Future studies are recommended to evaluate the effect of changing prosthetic stiffness at different prescription ratings on dynamic movement biomechanics. Nevertheless, ethical considerations are crucial since a large change in prosthetic stiffness may compromise the ability to complete designated task and increase risk of injury.

In this study, GRFs and impulses were utilized to evaluate movement biomechanics since the biomechanical representation of the prosthetic foot by the link segment model contains inaccuracies (Prince et al., 1994), leading to overestimation of joint forces, moments and powers (Geil et al., 2000). Future research on more appropriate modelling of dynamic prosthesis with a changing centre of rotation is warranted. The start-stop movement in this study was relatively slow, likely slower than that experienced in a game situation. Nonetheless, it was a useful task to assess the ability of the amputees to perform a controlled game-type movement that is dynamic and non-locomotion, and enabled an analysis of limb loading under controlled conditions. Further research needs to build on this study to determine the mechanisms that enable agility in the absence of an active ankle joint, particularly in investigating the influence of the initiation and termination steps on the movement TD and TO velocity and limb loading. Increasing our knowledge will enhance our understanding of the mechanisms of limb loading and thereafter in the development of training programmes and the design of multi-direction games type prostheses.

To conclude, the results of the present study suggest that transtibial amputees could perform a relatively dynamic start-stop movement but that they exhibit differences in spatiotemporal, kinematic and kinetic variables in comparison to able-bodied individuals, regardless of the prosthetic stiffness (DERPp vs DERPs) utilized during the movement. Amputee participants demonstrated a more horizontal approach with lower TD and TO angles, lower forward velocities and greater braking and propulsive peak forces and impulses compared to the more vertical, dynamic movement in control participants. The more horizontal movement of amputees, particularly when loading the intact limb may reflect a priority to adapt the load or a difficulty with movement initiation with a DERP, which warrants further research. Peak braking forces and impulses of the IL were significantly higher compared to the CL. Leg stiffness remained constant between limbs and with change in prosthetic stiffness, which may reflect compensation strategies through knee and hip joints. Further research should aim to enhance understanding of amputee movement biomechanics in relation to rehabilitation and prosthetic design.

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