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ORIGINAL RESEARCH

Optimal Starting Block Configuration in Sprint Running: A Comparison of Biological and Prosthetic Legs

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In the 2012 Paralympic 100 m and 200 m finals, 86% of athletes with a unilateral amputation placed their unaffected leg on the front starting block. Can this preference be explained biomechanically? We measured the biomechanical effects of starting block configuration for seven nonamputee sprinters and nine athletes with a unilateral amputation. Each subject performed six starts, alternating between their usual and unusual starting block configurations. When sprinters with an amputation placed their unaffected leg on the front block, they developed 6% greater mean resultant combined force compared with the opposite configuration (1.38 ± 0.06 vs 1.30 ± 0.11 BW, P = .015). However, because of a more vertical push angle, horizontal acceleration performance was equivalent between starting block configurations. We then used force data from each sprinter with an amputation to calculate the hypothetical starting mechanics for a virtual nonamputee (two unaffected legs) and a virtual bilateral amputee (two affected legs). Accelerations of virtual bilateral amputees were 15% slower compared with athletes with a unilateral amputation, which in turn were 11% slower than virtual nonamputees. Our biomechanical data do not explain the starting block configuration preference but they do explain the starting performance differences observed between nonamputee athletes and those with leg amputations.

Keywords: amputee, Paralympics, acceleration, track and field, 100 meter

In both the Olympic and Paralympic Games, the start of the 100 m running event is critical to overall performance. In nonamputee 100 m races, the start comprises about 5% of the total race time[1] and athletes need 30 to 40 m to reach their maximum speed.[2] However, for athletes with unilateral and bilateral transtibial amputations (T43/T44 classification) in the Paralympic 100 m, it is apparent that the acceleration phase is much longer, particularly for bilateral amputees. Slower accelerations likely reflect the lack of ankle muscle power combined with the compliance of the leg prosthesis. However, there are no published studies regarding the sprint start biomechanics of athletes with leg amputations. Thus, our overall goal was to quantify and better understand how leg prostheses and starting block configuration affect start performance.

In sprint running races, it is mandatory for athletes to use starting blocks.[3] Each athlete must position their feet against two adjustable plates anchored to the track. One foot is placed on the front block, typically ~0.5 m behind the start line, and one foot is placed on the back block, ~0.8 m behind the start line.[4,5] Although previous studies have tried to establish the best block configuration,[6,7] there is no biomechanically based best practice. Therefore, left/right foot positions and distances from the start line are typically based on the athlete's or coach's preference. Athletes fine-tune their starting block positions mostly by trial and error in the initial years of practice and typically converge on a final usual configuration.[6] Although nonamputee athletes have nearly symmetric legs, they typically place their dominant leg on the front block.[8] Further, Fortier et al[9] have shown that nonamputees generate greater forces and impulses with their front leg during a sprint start.

In contrast, the legs of athletes with a unilateral leg amputation are asymmetric. Their unaffected leg has both active (muscle) and passive-elastic (tendon) components, whereas the affected (amputated) leg of high-caliber athletes is typically fitted with a J-shaped carbon-fiber prosthesis that is elastic but completely passive. Once up to speed, these lower-leg running specific prostheses clearly store elastic energy during the first half of the contact phase, and return it during the second half of the contact phase. But, when using starting blocks, athletes are not able to use the elastic energy storage and return of a running-specific prosthesis, because in the "set" position, the athlete must be still, with no bouncing allowed.[3] Therefore, the force exerted by the affected leg onto the starting block is solely generated by the remaining proximal muscles. At constant speeds, sprinters using running-specific prostheses generate lower forces with their affected leg compared with their unaffected leg.[10] Thus, it is reasonable to expect the same force discrepancy during the start. However, no published studies have quantified the forces exerted on the starting blocks by sprinters using running specific prostheses.

Various authors have proposed different measures of start performance, including horizontal and resultant force applied on the starting blocks,[4] delay between the end of the rear and front force offset,[9] net horizontal external power[16] and knee joint power.[17] However, there is no universally agreed upon metric for start performance. Because horizontal acceleration equals the horizontal force divided by body mass, horizontal force can be used to quantify sprint start performance. For example, Morin et al[11]

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found a correlation of 0.834 between average horizontal ground reaction force (GRF) during a 6 s and 100 m performance. Thus, we reasoned that to optimize start performance, an athlete with a unilateral leg amputation should start with their unaffected leg in front, assuming the unaffected leg can generate greater force. Indeed, video recordings[12–15] of sprinters with a unilateral transtibial amputation competing in the 100 m and 200 m men's and women's T43/T44 finals of the 2012 London Paralympics revealed that 19 of 22 athletes placed their unaffected leg on the front block. Can this strong preference among Paralympic athletes be explained by biomechanical data? Moreover, how do running specific prostheses affect start performance?

The aim of the current study was to measure the effect of starting block configuration on starting performance of both nonamputee sprinters and athletes with a unilateral transtibial amputation. We hypothesized that sprinters with a unilateral transtibial amputation using a starting block configuration with their unaffected leg in front would have faster acceleration out of the blocks compared with using the configuration with their affected leg in front. Further, we aimed to understand the effect of running specific prostheses on start performance. We hypothesized that running specific prostheses impair start performance.

Materials and Methods

Subjects

A total of 16 subjects participated in this study: 7 (6 males and 1 female) nonamputee sprinters and 9 (7 males and 2 females) sprinters with a unilateral transtibial amputation (Table 1). Inclusion criteria were as follows: between 18 and 45 years old and currently competing in sanctioned competitions in 100 m and/or 200 m races. All subjects gave their written informed consent before participating as per the University of Colorado Institutional Review Board.

Experimental Design

Before experimental trials, each subject performed their normal prerace warm-up. Then, subjects practiced both their usual and unusual starting block configurations twice. Subsequently, each subject performed a total of six starts, alternating between their usual and unusual starting block configurations. The initial configuration was randomized. For the unusual blocks configuration, front and back blocks were switched while maintaining the same relative distance from the starting line. For each start, the investigator gave the standard verbal commands used in sprint races: "on your marks" and "set." After the subject assumed the set position, a computer-generated audio gunshot was provided by a nearby computer speaker. The audio signal also triggered force data collection. Each subject was instructed to run as fast as possible to a line placed 5 m from the start. Between each trial, eight minutes of rest were enforced to decrease the potential for fatigue. We performed a pilot study and found that after eight starts, athletes begin to experience performance deficits; thus, we limited our experiment to six starts.

Each athlete positioned custom-built individual foot starting blocks on two separate force platforms (LG6-4-2000, Advanced Mechanical Technology, Watertown, MA, USA) that were embedded in a 10 m runway that was covered with a rubber mat. The rubber mat allowed the subjects to use spiked shoes. The individual starting blocks were secured to the rubber mat via spikes on the bottom surface of each block. Each subject placed their feet against the blocks and their hands on a wide (1.22 m × 0.6 m) "winged" plate secured on top of the front force platform (Figure 1). The winged plate was wider than the force platform, and allowed us to measure the forces exerted by the front foot and both hands. Because the hands are immediately lifted at the starting signal, the hands exert a negligible amount of propulsive force.[5] We simultaneously measured the forces exerted by the back foot from a plate secured on top of the back force platform. We sampled the force signals at 1000 Hz via a data acquisition device (NI USB-6009, National Instruments Corp., USA) and a custom program (LabView, National Instruments Corp., USA). Then, we filtered the data with a recursive fourth-order Butterworth low-pass filter with a cut-off frequency of 30 Hz. To obtain the combined force exerted, we summed the signals from the front and back force platforms.

We used a custom program (Matlab, MathWorks, USA) to analyze the data. We identified the actual start of the push phase for each subject, (defined as the instant when the combined horizontal force exceeded a threshold of 5 N) and the end of the push phase (defined as the instant when the combined horizontal force crossed 0 N). Based on a pilot experiment, we chose a 5 N threshold to identify the start because it is clearly above the noise in the force signal that we measured in the set position (~2–3 N) and a 0 N threshold to define the end of the push phase based on previous research.[6] We calculated the reaction time as the interval between the start audio signal and the start of the push phase (Figure 2).

We calculated the following parameters to thoroughly quantify start performance: reaction time (s), push time (s), mean and peak combined (front + back) resultant force normalized to body weight (BW), mean push angle (degrees), mean and peak mass-specific combined horizontal power ($W \cdot kg^{-1}$), horizontal impulse (BW·s), final velocity (ie, horizontal velocity at the end of the push phase, $m \cdot s^{-1}$, obtained by integrating the combined horizontal force over the push time and dividing it by the subject's mass), mean horizontal acceleration of the push phase ($m \cdot s^{-2}$), front-leg and back-leg mean and peak horizontal force normalized to body weight (BW) and mass-specific mean and peak power ($W \cdot kg^{-1}$) in the horizontal direction. More specifically, we calculated the front and back leg horizontal instantaneous powers as per Donelan et al.:[18]

$$P_{h,\,front} = F_{h,\,front} \times v_{h,\,com} \tag{1}$$

$$P_{h,\,back} = F_{h,\,back} \times v_{h,\,com} \tag{2}$$

where $F_{h,\,front}$ and $F_{h,\,back}$ are the horizontal forces on the front block and on the back block respectively and $v_{h,\,com}$ is the horizontal velocity of the center of mass. We calculated the front and back leg mean forces and powers over the respective push phase of each leg and calculated the combined mean force and power over the entire push phase. We defined the mean push angle as the angle of the mean combined resultant force vector with respect to horizontal.

We used data from the unaffected leg and affected leg of each sprinter with an amputation to simulate their start mechanics as if they had (1) two unaffected legs ("a virtual non-amputee") or (2) two affected legs ("a virtual bilateral amputee"). Given that each sprinter switched front and back legs in the usual and unusual block configurations, we generated a virtual nonamputee by summing the impulse of their unaffected leg when it was placed on the front block and the impulse of their unaffected leg when it was placed on the back block. Then, we divided the combined impulse by the mean push time of the usual and unusual block configurations, to obtain mean force. We made the same calculations for the affected leg to simulate a virtual bilateral amputee. We performed all calculations in both the horizontal and vertical directions, obtaining

Table 1 Subject characteristics

	Subject	Sex (M/F)	Age (years)	Height (m)	Mass (kg)	100 m SB (s)	Usual front leg (L/R)	Training (Hours per Week)
	1	M	18	1.73	66.7	11.58	L	6
	2	M	18	1.76	81.8	11.70	L	5
	3	M	44	1.75	75.1	11.80	L	7
Nonamputee	4	M	33	1.75	65.8	12.22	L	7
	5	M	45	1.65	68.1	12.42	L	6
	6	M	19	1.82	96.0	12.96	L	7
	7	F	24	1.65	74.1	14.75	L	7
Average ± SD			30.3 ± 12.3	1.73 ± 0.06	75.4 ± 10.7	12.49 ± 1.11		6.5 ± 0.8

	Subject	Sex (M/F)	Age (years)	Height (m)	Mass (kg)	100 m SB (s)	Usual front leg (L/R)	Training (Hours per Week)	RSP Model	Time since amputation (years)	Experience on current RSP (years)
	1	M	33	1.91	112.3	12.14	L (UL)	8	Ottobock Sprinter	20.4	5.0
	2	M	24	1.75	74.5	12.40	L (UL)	7	Ossur Cheetah	2.1	1.5
	3	M	28	1.88	79.6	12.60	L (UL)	12	Ossur Flexfoot Sprint	3.1	2.5
	4	M	18	1.78	92.2	14.50	R (UL)	8	Ossur Cheetah	2.3	1.5
Amputee	5	F	21	1.70	59.0	15.63	L (UL)	7	Ottobock Sprinter	10.5	1.0
	6	M	29	1.83	73.9	11.90	R (AL)	9	Ottobock Sprinter	4.0	0.6
	7	M	22	1.77	73.9	12.60	R (AL)	12	Ottobock Sprinter	7.1	0.2
	8	F	27	1.70	64.2	13.61	L (AL)	11	Ottobock Sprinter	12.4	2.0
	9	M	28	1.70	60.7	26.33 (200 m)	R (AL)	8	Ossur Cheetah	6.1	0.3
Average ± SD			25.6 ± 4.7	1.78 ± 0.08	76.7 ± 16.8	13.17 ± 1.31		9.3 ± 2.1		7.6 ± 6.0	1.6 ± 1.5

Note. Demographic and anthropometric variables, 100 m seasons' best times (SB) and training hours per week of nonamputee and amputee sprinters. The front leg for the usual configuration is indicated for nonamputees and amputees (AL is the affected leg and UL is the unaffected leg). Each amputee subject's running-specific prosthesis (RSP) model, years since amputation, and experience on current running-specific prosthesis are reported. Mean values ± standard deviations (SD) are reported for both groups. Average ± SD of 100 m SB for Amputee group was calculated without taking in account subject 9, who competed only in 200 m events.

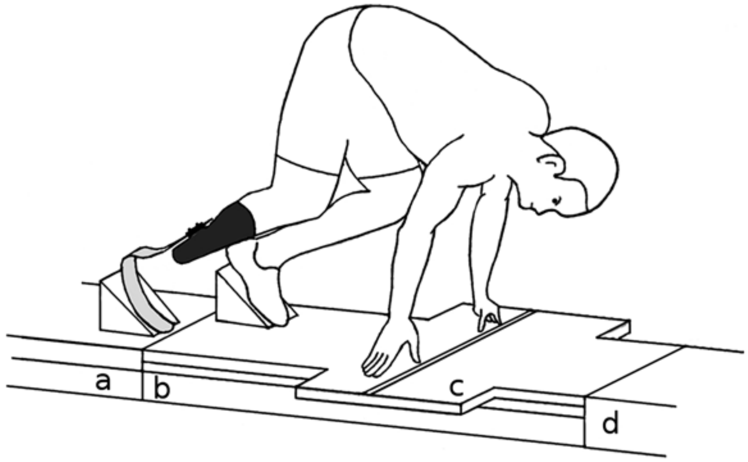

Figure 1 — Experimental setup of a starting blocks configuration. This athlete has his back affected leg, with his running-specific prosthesis, on force platform "a," his front unaffected leg and hands on the "winged" plate "c," secured to force platform "b." After the start, he runs as fast as possible on the elevated runway "d."

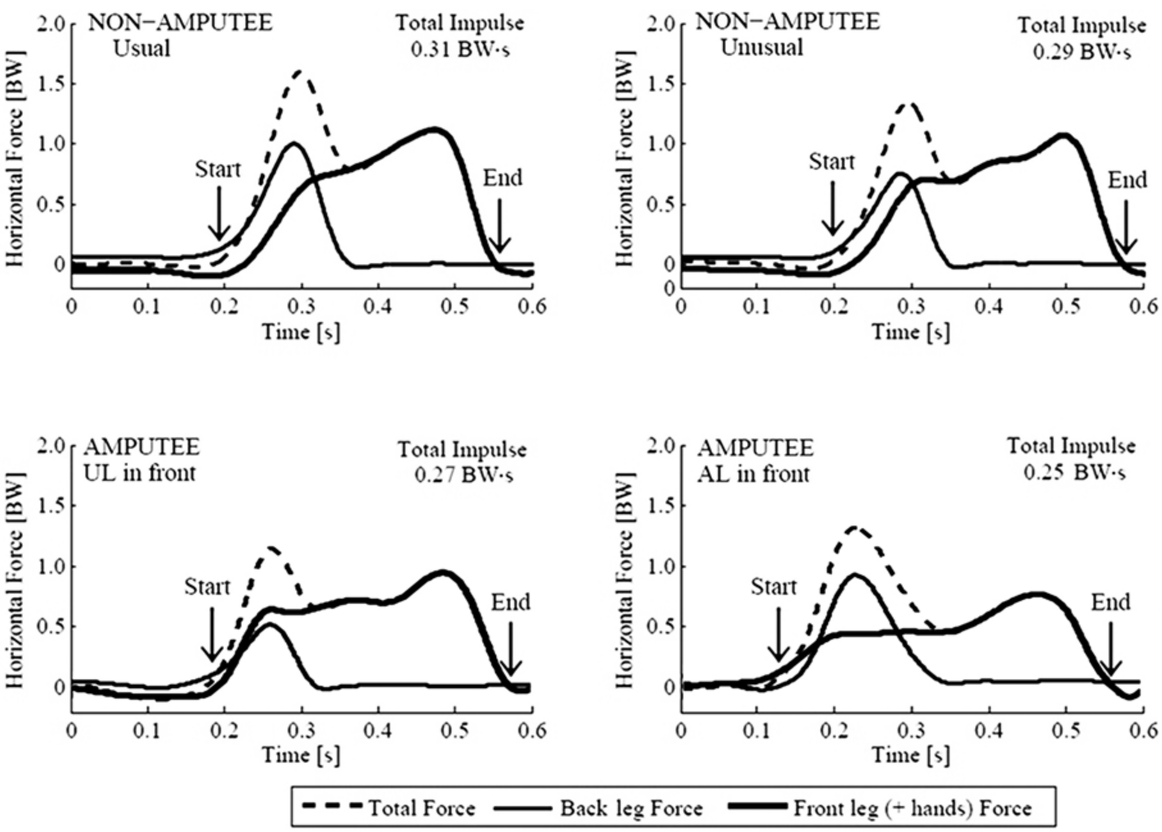

Figure 2 — Typical horizontal force traces for a nonamputee sprinter (upper panels) in the usual and unusual blocks configurations and for a sprinter with a unilateral transtibial amputation (lower panels) with the unaffected leg (UL) and affected leg (AL) on the front block: Time = 0 identifies the start signal, "Start" indicates the actual start (Horizontal Force > 5 N), "End" indicates the end of the push phase, the time between Start and End is the push time, and forces are expressed as multiples of body weight (BW).

mean resultant force and push angle. Lastly, we calculated the mean horizontal acceleration by dividing the mean horizontal force by the body mass of each athlete. For simplicity, we assumed the body mass of the virtual nonamputee and virtual bilateral amputee to be the same as the respective unilateral amputee, neglecting the small mass differences between the unaffected leg and affected leg.

Statistical Analysis

We checked the normality of the samples with the Shapiro-Wilk test and then used a repeated measures design to assess the differences between usual and unusual configurations in both groups and between the unaffected and affected front leg configurations in the amputee group. A P-value of less than or equal to 0.05 was accepted as significant. Although not statistically different, there was a 0.68 s average difference in 100 m personal best times between nonamputee sprinters and sprinters with a unilateral amputation, a large time difference in sprint performance. Therefore, we did not statistically compare nonamputee and amputee groups.

Results

We found no significant differences between or within groups for reaction times among the different configurations (Table 2). For the nonamputee sprinters, push time was slightly but significantly shorter in the usual configuration compared with the unusual configuration (0.407 ± 0.035 vs 0.416 ± 0.044 s, P = .043, n = 7). For the amputee sprinters, the push times were not significantly different between usual and unusual configurations or between unaffected leg in front and affected leg in front configurations.

We found no differences in mean or peak combined resultant forces or push angles between the usual and unusual configurations in the nonamputee sprinters (Table 2). However, we found that for sprinters with an amputation, the mean combined resultant force was 5% greater in the usual versus unusual configuration (1.37 ± 0.07 vs 1.30 ± 0.10 BW, P = .038, n = 9), but found no differences in peak combined resultant forces or push angles.

When comparing the unaffected leg in front versus the affected leg in front configurations, we found that the mean combined resultant force was 6% greater when the unaffected leg was placed in front (1.38 ± 0.06 vs 1.30 ± 0.11 BW, P = .015, n = 9). However, the peak combined resultant force was smaller for the unaffected leg in front vs affected leg in front configurations (2.10 ± 0.18 vs 2.36 ± 0.21 BW, P = .0010, n = 9). With the unaffected leg in front, the mean push angle was more vertical compared with having the affected leg in front (59.4 ± 2.2 vs 57.8 ± 2.7 deg, P = .029, n = 9). Because we found that there were much greater differences between the unaffected leg in front vs affected leg in front configurations, compared with the amputee sprinters' usual vs unusual configurations, the subsequent sections focus only on comparing the unaffected leg in front vs affected leg in front for the sprinters with an amputation.

We found that the non–amputee sprinters had slightly but significantly greater mean combined force in the horizontal direction for the usual vs unusual configuration (0.78 ± 0.09 vs 0.75 ± 0.09 BW, P = .042, n = 7). Therefore, the mean horizontal acceleration was greater for the usual versus unusual configuration (7.63 ± 0.91 vs 7.39 ± 0.84 $m \cdot s^{-2}$, P = .042, n = 7). Between the usual and unusual configurations there were no differences in peak horizontal force, horizontal power, final velocity, or horizontal impulse at the end of push time (Table 3).

In sprinters with an amputation, our most important finding was that the mean combined force, power, acceleration and final velocity were not significantly different between the unaffected leg in front compared with the affected leg in front configurations. Interestingly, sprinters with an amputation had significantly smaller peak combined horizontal force in the unaffected leg in front compared with the affected leg in front configuration (1.12 ± 0.13 vs 1.31 ± 0.18 BW, P = .008, n = 9).

In the nonamputee sprinters, there were no statistically significant differences between the front and back leg in the usual and unusual configurations (Table 4), but all values in the usual configuration resulted in numerically better performance, except for front leg mean forces, which were equal in the two configurations. The amputee sprinters had greater mean horizontal front leg force when their unaffected leg was in front (0.60 ± 0.13 vs 0.52 ± 0.11 BW, unaffected leg in front versus affected leg in front, respectively, P = .016, n = 9). However, the following key variables were superior for the affected leg in front configuration: mean back leg force (0.27 ± 0.13 vs 0.38 ± 0.13 BW, P = .025), mean back leg power (1.48 ± 1.04 vs 2.44 ± 1.40 W/kg, P = .048) and peak back leg power (3.53 ± 2.38 vs 6.76 ± 3.16 W/kg, P = .017), where all values refer to the unaffected leg in front versus affected leg in front, respectively and n = 9. (See Appendix for individual leg impluses.)

Our virtual nonamputee sprinter calculations demonstrated much greater mean combined resultant forces (1.47 ± 0.13 vs 1.16 ± 0.09 BW, P = .0003, n = 9) and mean horizontal accelerations (7.66 ± 0.94 vs 5.84 ± 0.69 $m \cdot s^{-2}$, P = .0001, n = 9) compared with virtual bilateral amputee sprinters. However, there were no differences in mean combined push angles.

Virtual nonamputee sprinters showed greater mean combined resultant forces (1.47 ± 0.13 vs 1.38 ± 0.06 BW, P = .004, n = 9) and lower mean push angles (58.0 ± 1.9 vs 59.4 ± 2.2 deg, P = .009, n = 9), which resulted in greater mean horizontal accelerations (7.66 ± 0.94 vs 6.76 ± 0.65 $m \cdot s^{-2}$, P = .002, n = 9) compared with sprinters with a unilateral amputation using their unaffected leg in front configuration (Table 5). In turn, sprinters with a unilateral amputation had greater mean combined resultant forces (1.38 ± 0.06 vs 1.16 ±

Table 2 Reaction times, push times, mean and peak resultant forces, and mean push angles

	Reaction Time (s)	Push Time (s)	Mean Resultant Force (BW)	Peak Resultant Force (BW)	Mean Push Angle (deg)
Nonamputee (n = 7)					
Usual	0.155 ± 0.043	**0.407 ± 0.035**	1.40 ± 0.07	2.28 ± 0.21	55.9 ± 3.1
Unusual	0.163 ± 0.036	**0.416 ± 0.044**	1.39 ± 0.07	2.30 ± 0.25	56.6 ± 3.0
Amputee (n = 9)					
Usual	0.173 ± 0.060	0.417 0.059	**1.37 ± 0.07**	2.26 ± 0.23	58.3 ± 3.2
Unusual	0.188 ± 0.083	0.424 0.052	**1.30 ± 0.10**	2.21 ± 0.24	59.0 ± 1.6
Amputee (n = 9)					
UL in front	0.169 ± 0.059	0.419 ± 0.046	**1.38 ± 0.06**	**2.10 ± 0.18**	**59.4 ± 2.2**
AL in front	0.192 ± 0.076	0.422 ± 0.059	**1.30 ± 0.11**	**2.36 ± 0.21**	**57.8 ± 2.7**

Note. Each value is reported as mean ± S.D. Highlighted gray indicates the value that results in the best performance between the two configurations. Statistically significant (P < .05) differences between the 2 configurations within each group are highlighted in **boldface type**. AL is the affected leg and UL is the unaffected leg; forces are expressed as multiples of body weight (BW).

Table 3 Mean and peak combined (front + back) forces, mean accelerations, final velocities, and impulses at the end of the push phase

	Horizontal					
	Mean Combined Force (BW)	Peak Combined Force (BW)	Mean Combined Power (W·kg^{-1})	Mean Acceleration (m·s^{-2})	Final velocity (m·s^{-1})	Impulse (BW·s)
Nonamputee (n = 7)						
Usual	**0.78 ± 0.09**	1.39 ± 0.24	11.94 ± 2.19	**7.63 ± 0.91**	3.09 ± 0.27	0.315 ± 0.028
Unusual	**0.75 ± 0.09**	1.39 ± 0.26	11.36 ± 1.72	**7.39 ± 0.84**	3.05 ± 0.22	0.310 ± 0.022
Amputee (n = 9)						
UL in front	0.69 ± 0.07	**1.12 ± 0.13**	9.59 ± 1.53	6.76 ± 0.65	2.81 ± 0.25	0.286 ± 0.026
AL in front	0.68 ± 0.10	**1.31 ± 0.18**	9.42 ± 2.52	6.67 ± 1.02	2.78 ± 0.37	0.283 ± 0.038

Note. All values are calculated in the horizontal direction and are reported as mean ± SD. Highlighted gray indicates the value that results in the better performance between the two configurations. Statistically significant ($P < .05$) differences between the 2 configurations within each group are highlighted in **boldface type**. AL is the affected leg and UL is the unaffected leg; forces and impulses are expressed as multiples of body weight (BW) and as multiples of body weight seconds (BW·s) respectively.

Table 4 Individual (front and back) leg forces and powers

	Horizontal							
	Mean Force Back (BW)	Mean Force Front (BW)	Peak Force Back (BW)	Peak Force Front (BW)	Mean Power Back (W·kg^{-1})	Mean Power Front (W·kg^{-1})	Peak Power Back (W·kg^{-1})	Peak Power Front (W·kg^{-1})
Nonamputee (n = 7)								
Usual	0.42 ± 0.11	0.62 ± 0.10	0.84 ± 0.22	1.02 ± 0.13	2.72 ± 1.10	11.30 ± 2.34	6.63 ± 2.75	26.51 ± 5.03
Unusual	0.39 ± 0.07	0.62 ± 0.07	0.77 ± 0.12	1.00 ± 0.09	2.33 ± 0.54	11.02 ± 2.01	5.86 ± 1.40	25.62 ± 3.86
Amputee (n = 9)								
UL in front	**0.27 ± 0.13**	**0.60 ± 0.13**	**0.48 ± 0.21**	0.92 ± 0.18	**1.48 ± 1.04**	9.43 ± 2.22	**3.53 ± 2.38**	21.37 ± 5.40
AL in front	**0.38 ± 0.13**	**0.52 ± 0.11**	**0.77 ± 0.25**	0.94 ± 0.13	**2.44 ± 1.40**	8.59 ± 2.41	**6.76 ± 3.16**	19.82 ± 6.56

Note. All values are calculated in the horizontal direction and are reported as mean ± SD. Highlighted gray indicates the value that results in the best performance between the two configurations. Statistically significant ($P < .05$) differences between the 2 configurations within each group are highlighted in **boldface type**. AL is the affected leg and UL is the unaffected leg; forces are expressed as multiples of body weight (BW).

Table 5 Resultant forces, push angles, and horizontal accelerations of virtual athletes and athletes with a unilateral amputation

	Mean Resultant Force (BW)	Mean Push Angle (deg)	Horizontal acceleration (m·s^{-2})
Virtual nonamputee	**1.47 ± 0.13**	**58.0 ± 1.9**	**7.66 ± 0.94**
Unilateral amputee	**1.38 ± 0.06**	**59.4 ± 2.2**	**6.76 ± 0.75**
Virtual bilateral amputee	**1.16 ± 0.09**	59.1 ± 2.4	**5.84 ± 0.69**

Note. We calculated values for a virtual nonamputee by merging 2 unaffected legs and calculated values for a virtual bilateral amputee by merging 2 affected legs from the unilateral amputee group. Values from the unilateral amputee group refer to the unaffected leg in front condition. Highlighted gray indicates the value that results in the best performance between the three conditions. Statistically significant ($P < .05$) differences between groups are highlighted in **boldface type**; forces are expressed as multiples of body weight (BW).

0.09 BW, $P = .0001$, n = 9) and mean horizontal accelerations (6.76 ± 0.65 vs 5.84 ± 0.69 m·s^{-2}, $P = .001$, n = 9) compared with virtual bilateral amputee sprinters (Figure 3). There were no differences in mean push angles.

Discussion

We hypothesized that sprinters with a unilateral transtibial amputation using a starting block configuration with their unaffected leg in front would have faster acceleration out of the blocks compared with using the configuration with their affected leg in front. However, we found that, despite a greater combined resultant force in the configuration with their unaffected leg in front, horizontal acceleration was not statistically different between the two configurations. We also hypothesized that running-specific prostheses would limit start performance. Our measurements and calculations indicate that, *ceteris paribus*, athletes using two running specific prostheses would have significantly slower horizontal accelerations of the CoM out of the

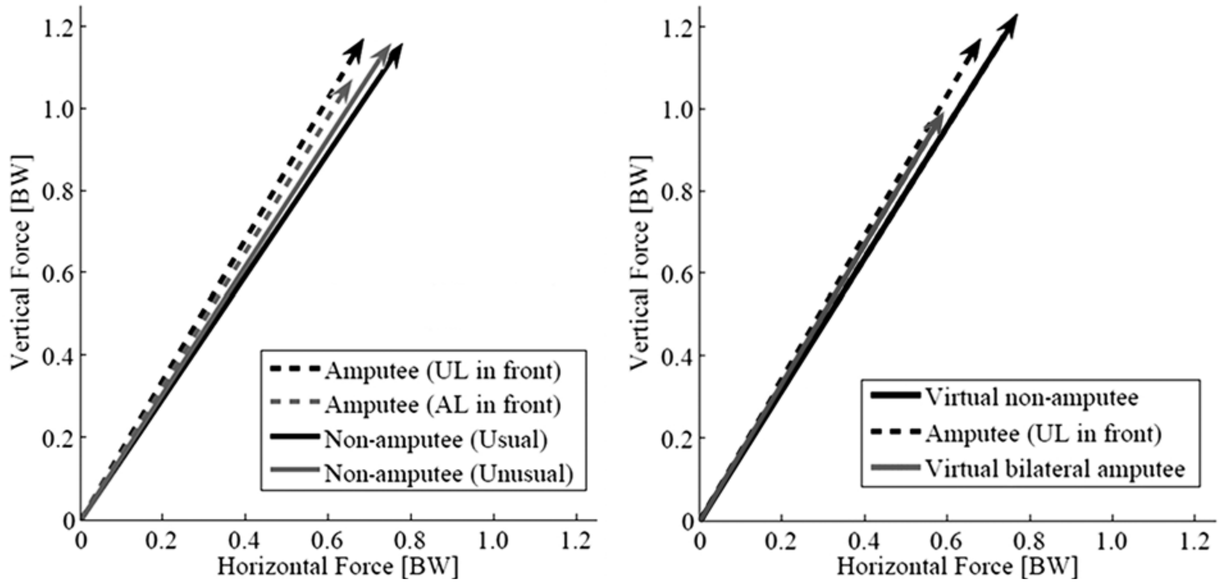

Figure 3 — Left panel: mean combined resultant force vectors (average values) for unilateral amputee (n = 9) and nonamputee (n = 7) sprinters in unaffected leg (UL) versus affected leg (AL) in front and usual versus unusual, respectively. Forces are expressed as multiples of body weight (BW). There were no differences between conditions for nonamputee sprinters. Unilateral amputee sprinters with their unaffected leg in front developed more force ($P = .015$) and directed it more vertically ($P = .029$) compared with the configuration with their affected leg in front. Right panel: mean resultant force vectors (average values) for virtual nonamputee (n = 9), unilateral amputee in the unaffected leg in front configuration (n = 9) and virtual bilateral amputee (n = 9) sprinters. Virtual nonamputee sprinters with 2 unaffected legs developed 7% more force ($P = .001$) compared with unilateral amputee sprinters. Unilateral amputee sprinters, in turn, developed 15% more force ($P = .0001$) than virtual bilateral amputees with 2 affected legs.

blocks compared with athletes with a unilateral amputation, who in turn would have slower accelerations compared with nonamputees.

As expected, nonamputee sprinters had better horizontal acceleration, and therefore better starts, in their usual compared with unusual configuration. In their unusual configuration, all sprinters reported that they felt slightly "uncomfortable" or "awkward." Despite this, combined resultant forces and push angles were not statistically different between configurations, indicating that it is the combination of both the resultant force and push angle that determines horizontal force and therefore forward acceleration. Although not surprising, we found that starting block configuration has a small but detectable effect on sprint start performance. Single-leg values, however, were not different between configurations, suggesting that block configuration is not, or at least not only, based on the small differences in terms of power and/or force capability between right or left legs in nonamputee athletes. Another factor that could influence start performance is the brain's laterality. Like in hand dominance, leg dominance seems to be the result of subtle differences in the brain hemispheres, rather than biomechanical differences between the right and left legs.[19] Eikenberry et al[6] cited different hemisphere specialization to explain performance asymmetries in subjects placing their left or right foot on the front block.

Our findings are similar to previous studies of nonamputee sprinters with their usual blocks configuration.[1,4] To the best of our knowledge, no previous studies have measured the force-time characteristics of sprinters adopting an unusual blocks configuration. Morin et al[11] demonstrated that the magnitude of the horizontal component of the ground reaction force during the acceleration was correlated with 100m performance, but the magnitude of the resultant force was not. Our data are in accordance with Morin et al[11] in that we found the slower start in the unusual configuration for nonamputees is a consequence of a suboptimal combination of force magnitude and push angle, rather than a force deficiency alone.

When subjects with an amputation placed their unaffected leg on the front block, they developed 6% greater mean combined resultant forces compared with when they placed their affected leg on the front block. However, in contrast with our hypothesis, horizontal acceleration was not statistically different between the unaffected leg and affected leg in front configurations. Unlike nonamputees, the combination of resultant force and push angle cancelled out any differences in horizontal acceleration between the unaffected leg and affected leg in front configurations. The greater mean resultant forces in the unaffected leg in front configuration were directed more upwards (+3%), causing a mere 1% nonsignificant improvement in horizontal acceleration, compared with the affected leg in front configuration. Thus, the observed strong preference of Paralympic athletes for the unaffected leg in front configuration does not seem to be supported by biomechanical data.

Surprisingly, peak combined resultant and horizontal forces were greater in the affected leg in front compared with the unaffected leg in front configuration. Nonamputee and amputee sprinters reach peak combined force when they are pushing with both legs on the blocks (Figure 2). Amputee sprinters, in particular, develop greater peak force with their back (unaffected) leg in the affected leg in front configuration; this configuration also results in a greater peak combined force than the unaffected leg in front configuration.

Apart from peak horizontal front block forces that were almost equal in the two configurations, the unaffected leg developed more force and power compared with the affected leg when it was placed on the front block (unaffected leg in front) and on the back block (affected leg in front). In particular, when switching from the unaffected leg in front to the affected leg in front configuration, there was a 13% reduction in mean horizontal force on the front block due to the affected leg that was compensated for by a 40% increase in horizontal force on the back block from the unaffected leg.

In previous Paralympic Games, athletes with a unilateral transtibial amputation (classified as T44) have competed in the same races as athletes with bilateral transtibial amputations (T43). Although it has not yet been systematically quantified, it appeared that in the 2012 Paralympics, athletes with a unilateral amputation had quicker starts than athletes with bilateral amputations, but athletes with bilateral amputations attained faster speeds in the final portion of the race.[12,15] Our virtual athlete calculations provide some insight into start performance differences between sprinters with unilateral and bilateral amputations and between nonamputee and amputee sprinters. The average values of mean combined resultant force, push angle and horizontal acceleration of virtual nonamputee sprinters showed no statistical difference (P = .225, P = .109, and P = .956 respectively) from our nonamputee subjects using their usual configuration. Virtual bilateral amputee sprinters had worse start performance compared with sprinters with a unilateral amputation; they had a 15% lower resultant force and maintained the same horizontal push angle, which produced a 15% slower horizontal acceleration. Sprinters with a unilateral amputation, in turn, had worse start performance compared with virtual nonamputees: a 7% lower resultant force and a 2% more vertical push angle, resulting in an 11% slower horizontal acceleration. These virtual athlete calculations support the impression that, among similar-level athletes (ie, same overall times), sprinters with bilateral transtibial amputations have slower starts than sprinters with unilateral amputations, and sprinters with unilateral amputations have slower starts than nonamputee sprinters.

We did not account for potential differences in body mass when we simulated virtual athletes. However, a running specific prosthesis and socket have an approximate mass of 1.5 kg, while a biological foot and partial shank for an 80 kg male have a mass of ~2.9 kg,[20] a difference of 1.4 kg. Therefore, we chose to neglect the effect of a mass difference in our calculations. In addition, we assumed that the unaffected and affected legs act independently when athletes with a unilateral amputation use starting blocks. Our measurements, however, indicate that a performance deficit in one leg can be at least partially compensated for by the other leg. Any dependence and adaptability between legs would not be reflected in our virtual athlete calculations.

In conclusion, we found that nonamputee sprinters have a better start performance in their usual configuration, while there is no consistent effect of overall block configuration on start performance of sprinters with a unilateral transtibial amputation. Athletes with amputations exerted less force with their affected leg in both configurations, thus, the use of running specific prostheses clearly impairs start performance.

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Appendix

In the nonamputee sprinters, there were no statistically significant differences in individual (front and back) leg impulses between usual and unusual configurations (Appendix Table 1). The amputee sprinters had greater front leg impulse when their unaffected leg was in front (0.250 ± 0.054 vs 0.218 ± 0.043, unaffected leg in front versus affected leg in front respectively, $P = .047$, n = 9) while no statistically significant differences were found in the back leg impulses between the two configurations.

Appendix Table 1 Individual (front and back) leg impulses

	Impulse Back (BW·s)	Impulse Front (BW·s)
Nonamputees (n = 7)		
Usual	0.066 ± 0.023	0.249 ± 0.026
Unusual	0.054 ± 0.011	0.256 ± 0.027
Amputees (n = 9)		
UL in front	0.037 ± 0.035	**0.250 ± 0.054**
AL in front	0.065 ± 0.034	**0.218 ± 0.043**

Note. All values are calculated in the horizontal direction and are reported as mean ± SD. Highlighted gray indicates the value that results in the best performance between the two conditions. Statistically significant ($P < .05$) differences between the 2 configurations within each group are highlighted in **boldface type**; impulses are expressed as multiples of body weight seconds (BW·s).