FREE MOMENT APPLICATION BY ATHLETES WITH AND WITHOUT AMPUTATIONS IN LINEAR AND CURVED SPRINTING

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The purpose of the present study was to describe free moment (FM) application to the ground in high speed linear and curved running in athletes with unilateral amputations and non-amputee athletes. The results indicate that peak FM amplitudes are about three times higher in sprinting compared to running at distance running speeds. Curved running decreases internal rotation FM amplitudes of the right (outside) leg in non-amputee runners. The use of running specific prostheses (RSPs) is related to lower FM application in sprinters with unilateral amputations. This might be related to inertial asymmetries between legs or to constraints imposed at the RSP attachment interface. Monitoring FM patterns in sprint diagnostics might help athletes and coaches in the improvement of running and sprinting performance and overuse injury prevention.

KEY WORDS: Running mechanics, locomotion, Paralympic sport.

INTRODUCTION: Running at high speed requires powerful force application to the ground as well as rapid and coordinated motions of the arms and legs. The main function of the arms and the upper trunk in the sprinting gait cycle is to counteract the angular impulse about the vertical axis created by the motions of the legs and the lower trunk (Hinrichs, 1987). In case of insufficient angular impulse cancellation between upper and lower body within the transversal plane, a compensating moment needs to be applied. If sufficient rotational traction is available at the interface between foot and ground, this can be done by the application of a free moment (FM). The FM acts as a force couple about an axis normal to the running surface, by applying frictional shear forces of equal magnitude but opposite direction in the horizontal plane at the front and rear parts of the foot (Holden & Cavanagh, 1991). It is called FM, because the effects of a moment created by a force couple on a body are independent (free) of the point of application in rigid body dynamics. The FM plays an important role in controlling whole body angular momentum within the transversal plane in straight running (Willwacher, Eglitits, Heinrich, Sanno, & Brüggemann, 2014).
In addition, the FM is a basic variable for the determination of intra and inter-segmental load, as its amplitudes are related to torsional deformation of the tibia and to lower extremity joint moments in the transversal and frontal planes during human locomotion (Yang et al., 2014; Willwacher, Goetze, Fischer, & Brüggemann, 2016). Furthermore, FM amplitudes play a role in the development of tibial stress fractures (Milner, Davis & Hamill 2006).
Despite its important role in dynamic stability control and for the loading of the lower extremities, FM application has been studied only in comparably less dynamic motions like walking and distance running. Furthermore, no data exists on FM application strategies in athletes with lower extremity amputations, even though the constraints imposed by the use of running specific prostheses (RSPs) and the altered mass of their replaced legs might considerably affect the ability to control whole body angular momentum by means of FM application. A better understanding of FM application in faster movements might aid coaches in the evaluation of sprinting technique and in an efficient load management, but might also result in the improvement of RSP design. As FM signals are readily available from force plate measurements, they might also serve within direct feedback tools for the improvement of the running technique.
Therefore, the purpose of the present study was to describe the FMs applied to the ground by runners with and without amputations during high speed straight and curved running. It was hypothesized that that FM peaks were higher in sprinting than FM peaks reported for

slower speed running in the literature, because the FM angular impulses need to be created during much shorter ground contact times. We further hypothesized that FM patterns would differ between straight and curved running due to the task differences and the corresponding alterations in body orientation and horizontal ground force application. Thirdly, it was hypothesized that athletes with unilateral amputations would apply FMs in an asymmetric fashion to the ground.

METHODS: Six male sprinters without amputations (age: 20.2 ± 2.6 years; height: 1.86 ± 0.06 m; mass: 76.3 ± 8.2 kg) of national competitive level, two Paralympic athletes with unilateral transtibial amputation (age: 25 and 26 years; height: 1.89 and 1.83 m; mass: 89.0and 74.6 kg) and one Paralympic athlete with unilateral transfemoral amputation (age: 31 years; height: 1.78 m; mass: 80.4 kg) gave their written consent to participate voluntarily in this study. At testing day all athletes were pain free and without any physical impairment. During the testing procedure the participants wore their own sprint spikes.

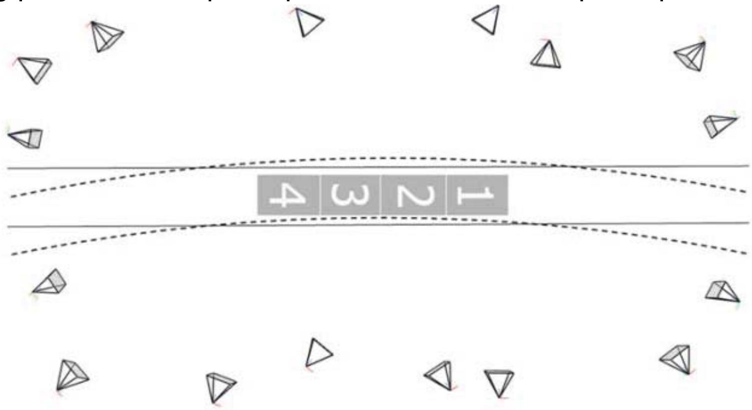

Figure 1: Experimental setup including 16 infrared motion capture cameras and four 90x60 cm force plates. Running direction is from the right to the left.

All non-amputee athletes were advised to perform straight and curve sprints with a constant individual submaximal sprinting speed which needed to be attained after an approach run of 40 m. Athletes with amputations performed all trials with their maximum effort. The radius of the curved track was 36.5 m which represents the first lane of an IAAF approved 400 m track.
A 16 camera (Vicon MX40) infrared motion capture system (250 Hz, Vicon, Oxford, UK) and four 90x60 cm force plates (Kistler AG, Winterthur, Switzerland) were used to collect kinematic and FM data (Figure 1).
The midpoint of four markers placed on the pelvis was used to calculate the average running speed within the transversal plane of the runner during each analysed stance phase. The stance phase was determined using a 20 N threshold of the vertical ground reaction force. Peak internal rotation FM and net FM impulse were determined from the FM waveform in the stance phase. FMs were calculated using the following formula:

$$FM = Mz - (r \times F_{Shear})$$

In this equation, FM is the free moment, Mz is the plate moment around a vertical axis crossing the midpoint of the force plate. F_{Shear} equals the resultant shear force vector of the horizontal ground reaction force components, while r represents the vector pointing from the centre point of the force platform to the point of force application of the ground reaction force. FMs were expressed as reaction moments normalised to body mass (Wannop, Worobets, & Stefanyshyn, 2012). Differences between curved and straight running were compared within legs using dependent sample t-tests. Level of significance was set to 0.05.

RESULTS: Non-amputee FM peak values were on average about three times higher compared to the largest FM amplitudes reported in a large (n = 222) sample of recreational runners at a speed of 3.5 m/s (Willwacher et al., 2016). Curved running resulted in lower contact times, FM peaks and net impulses in the right leg of non-amputee athletes (Table 1). In the left leg, only a significantly higher running speed was detected in curve sprinting compared to the straight sprinting condition (Table 1).

Table 1
Discrete parameter results for non-amputee athletes (n = 6).

		Non-amputees	
		Left	**Right**
Contact time (s)^R	Straight	0.107 ± 0.007	0.109 ± 0.008
	Curve	0.113 ± 0.009	0.106 ± 0.008
Speed (m/s)^L	Straight	9.142 ± 0.275	9.226 ± 0.441
	Curve	9.533 ± 0.184	9.506 ± 0.329
Peak FM (Nm/kg)^R	Straight	0.363 ± 0.110	0.334 ± 0.083
	Curve	0.415 ± 0.143	0.193 ± 0.087
FM Impulse (Nms/kg)^R	Straight	0.010 ± 0.007	0.008 ± 0.005
	Curve	0.010 ± 0.007	0.002 ± 0.006

L,R: significant ($p < 0.05$) effect between straight and curve for the right and left leg, respectively.

In straight running, non-amputee athletes applied FMs in a clearly more symmetrical manner than athletes with amputations (Figure 2). Athletes with amputations applied lower FM peaks in their affected compared to their non-affected legs (Table 2, Figure 2).

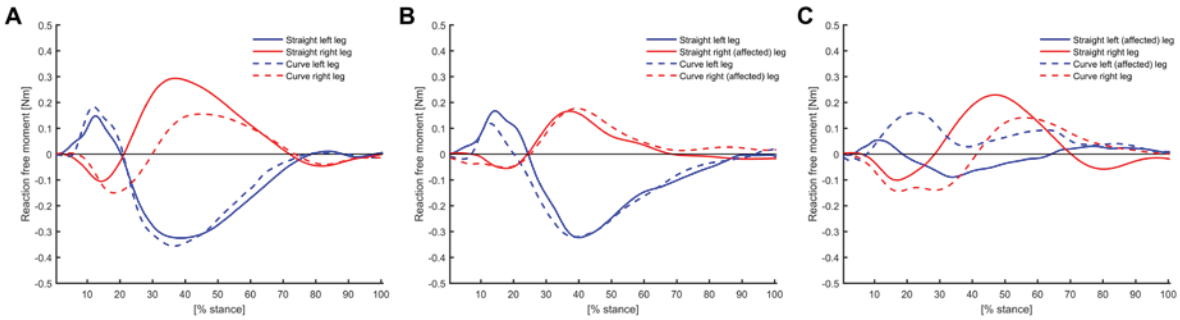

Figure 2: Stance phase normalized FM curves of the left and right legs for non-amputee (A) as well as athletes with transtibial (B) and transfemoral (C) unilateral amputations. FMs are visualized without laterality correction in order to better visualize the asymmetry between legs.

Table 2
Discrete parameter results for athletes with unilateral amputations.

		Transtibial amputees (n=2)		Transfemoral amputee	
		Left unaffected	**Right affected**	**Left affected**	**Right unaffected**
Contact time (s)	Straight	0.098 ± 0.000	0.113 ± 0.006	0.120	0.099
	Curve	0.104 ± 0.001	0.112 ± 0.001	0.139	0.098
Speed (m/s)	Straight	9.416 ± 0.406	9.520 ± 0.296	8.723	8.959
	Curve	8.360 ± 1.333	8.447 ± 0.799	7.994	8.105
Peak FM (Nm/kg)	Straight	0.356 ± 0.151	0.172 ± 0.046	0.130	0.281
	Curve	0.369 ± 0.219	0.186 ± 0.083	0.049	0.173
FM Impulse (Nms/kg)	Straight	0.008 ± 0.004	0.003 ± 0.004	-0.001	0.004
	Curve	0.011 ± 0.005	0.005 ± 0.006	-0.010	0.000

DISCUSSION: The purpose of the present study was to describe FMs applied to the ground in high speed linear and curved running in athletes with unilateral amputations and non-amputee athletes. To the knowledge of the authors, this is the first study to describe FM application for sprinting. The results clearly suggest a speed dependence of FM peak amplitudes in running, because average peak values were clearly higher than previously reported in the literature for distance running speeds (e.g. Willwacher et al., 2016). In non-amputee sprinters, FM application appears to be symmetrical between legs, while in athletes with amputations pronounced asymmetries were observed. This might be due to the fact that the inertial properties of the legs are asymmetrical due to the potentially lower mass and moment of inertia of RSPs compared to biological limbs. Furthermore, it could be possible that the attachment interface of the prosthesis and the residual leg is not capable of tolerating the higher transversal plane moments necessary for applying high FM amplitudes. This might result in a constrained FM application ability in athletes with unilateral amputations.
In curved sprinting, lower reaction FM net impulses were observed for non-amputee athletes, showing that FM impulse is not used to increase the whole body angular momentum in the direction of the curve (rotation to the left). Other mechanisms might be applied to achieve the change in body orientation that is necessary in curve sprinting. A contribution of the moment created by horizontal ground reaction forces with respect to the centre of mass is most conceivable to this respect and needs to be determined in future studies.
The application of higher FMs is related to increased joint moments at the lower extremities. Therefore, it can be speculated that running with higher FMs insures a greater metabolic cost, as greater muscle forces might be necessary to create the increased joint moments.

CONCLUSION: This study identified for the first time FM characteristics in high speed running. FM amplitudes are higher in high speed compared to typical distance running speeds. FM patterns could be a valuable target for a quick evaluation of left / right symmetry in sprinting. Future studies should address the relationship between FM amplitudes and running economy. Reducing excessive FM amplitudes might result both in a better sport performance and a decreased risk of sustaining overuse injuries.

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Acknowledgement

This study was supported by Nike Inc.