THE INFLUENCE OF SOLE LONGITUDINAL BENDING STIFFNESS ON PUSH-OFF BIOMECHANICS IN FULL EFFORT LINEAR ACCELERATION

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The purpose of this study was to identify the effect of longitudinal bending stiffness (LBS) alterations of running shoes in the initial push-off of the front leg in a linear acceleration task. 14 male sport students were analysed using a full body 3D motion analysis in combination with ground reaction force measurements during the initial step of a full effort 5 m sprint. Increasing LBS did not lead to a significant increase of normalized average acceleration power, but affected MTP and ankle biomechanics. Push-off time was systematically increased with increased LBS. Average ankle joint moments were reduced, even though average GRF lever arms were increased with higher LBS. Increasing the power generation capacities of ankle plantar flexors combined with LBS increase might be a promising combination to improve acceleration performance.

KEY WORDS: MTP joint, lever arm, GRF, acceleration performance, athletic footwear

INTRODUCTION: Improving acceleration performance (AccP) is a major target in the training of sprinters and in strength and conditioning programs in many team sports. In addition to improvements through training interventions, AccP might be improved by optimized sport equipment. The longitudinal bending stiffness (LBS) of sport shoes, for example, has been shown to affect athletes' performance in sprinting and cutting moves (Enders et al., 2015; Stefanyshyn & Fusco, 2004; Tinoco, Bourgit, & Morin, 2010; Stefanyshyn & Nigg, 2000) but also in endurance tasks (Roy & Stefanyshyn, 2006). With increased LBS, the energy dissipation at the metatarsophalangeal (MTP) joint was decreased and considered to be the causing mechanism for performance improvement (Stefanyshyn & Nigg, 2000; Roy & Stefanyshyn, 2006). In footwear science and manufacturing, optimizing LBS of running and sprinting shoes is an important issue (Nigg, 2010) to improve athletes' performance. But what the literature is still lacking of at the moment is a more detailed description of the mechanisms underlying the identified performance alterations.
Therefore, the purpose of this project was to elaborate if and how the athletes' front leg acceleration biomechanics are influenced by LBS of footwear during the initial push-off in a full effort linear acceleration motion.

METHODS: For the purpose of this study, 14 male subjects (age: 22.5 ± 1.31 years, height: 181.925 ± 4.52 cm, mass: 74.255 ± 3.95 kg) without injury and pain for at least 6 month prior to the study were recruited and analyzed. A further prerequisite was to be active in team or racket sports which include rapid acceleration motion. Written informed consent was signed by every athlete. Athletes were measured during 3 full effort 5 m sprints with 3 different insole conditions. A flexible condition was used as baseline (BL) condition. For this purpose, additional cuttings in the outsole of the test shoe Brooks Pure Connect were added. A custom-made plastic insole was used to create a medium stiff (MS) condition and a custom-made fiberglass insole was used to facilitate a highly stiff (HS) condition. A three point bending test according to the protocol in Willwacher et al. (2013) was utilized to quantify LBS, which was 1.44 ± 0.03 N/mm for the BL condition, 13.92 ± 0.08 N/mm for the MS condition and 21.82 ± 0.21 N/mm for the HS condition. Independent of personal habits, all athletes needed to perform all 5 m sprints with the right leg in front. Subjects started with the front (right) and rear (left) leg placed on two separate force plates (90x60 cm, 1000Hz, Kistler Instrumente AG, Winterthur, Switzerland). The position of the frontal foot was marked to ensure the same starting position. After 3 valid trials (standing-start from a resting position, push-off from the frontal foot) the insole condition was randomly changed for every participant. To analyze the athletes' motion, 3D analysis was performed with a full-body marker set with a total of 78 markers using 17 infrared cameras (200Hz, Vicon MX40, Vicon Motion Systems, Oxford, UK). The center of mass (CoM) of each athlete was calculated by using a 15 segment rigid body model with anthropometric values of de Leva (1996). Furthermore, lower extremity joint angles as well as power and moments were examined by the use of a rigid body model of fore- and rearfoot, shank, thigh and pelvis segment (Willwacher et al., 2014). Using the equations of Hof (1992), the lower extremity resulting internal joint moments were calculated. MTP, ankle and knee joint power was determined by multiplication of internal joint moments and angular velocities. These were analyzed in the sagittal plane because the major goal of linear acceleration is achieving a high increase of kinetic energy in the antero-posterior direction in the shortest possible time. The analysis was done for the front leg during the push-off time. Push-off time started when the horizontal CoM velocity exceeded 0.4 m/s for the first time and ended when vertical GRF fell below 10 N. The equation of Bezodis et al. (2010) for the normalized average horizontal block power was adopted for the calculation of the normalized average horizontal acceleration performance (NAHAP). Initial and final horizontal CoM velocities were determined from the whole body CoM at the beginning and end of the push-off. The horizontal distance of the vertical projection of the CoM to the point of force application (PFA) of the GRF (CoM-PFA-distance) and the trunk angle was used to describe an athletes' overall body configuration. Only the best trial (based on NAHAP) of each subject was used for the purpose of this study. One factor (LBS) repeated measures ANOVA was performed for LBS conditions. A post hoc pairwise t-test followed by Fisher's least significant difference approach was used in case of significant effect ($p < .05$) or a trend towards that ($p < .01$).

RESULTS: NAHAP did not change significantly between conditions and 10 out of 14 participants decreased their performance with MS. For 9 out of 14 athletes lower NAHAP was recorded for HS in comparison to BL. Vertical GRF impulse was significantly increased for HS. Although changes in horizontal GRF impulse were not significantly different, they still raised systematically from BL to HS (table 1). Contact times slightly increased systematically with increasing stiffness, which resulted in no significant differences in NAHAP. Strongest effects were observed for MTP joint biomechanics. GRF lever arm amplitudes, joint moments, average (negative) power and absorbed energy systematically increased with higher LBS (table 1). Ankle joint GRF lever arm amplitudes increased with higher LBS as well, but average ankle joint moments were decreased for higher LBS levels (table 1).

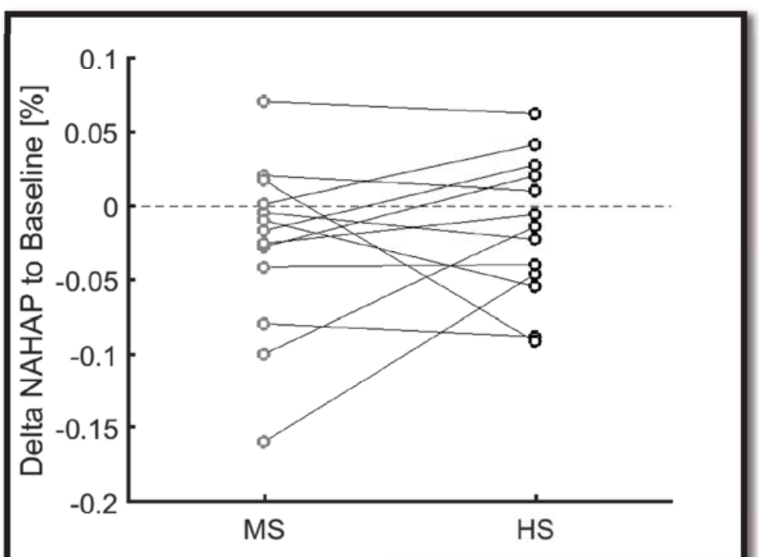

Figure 1: Graphical representation of the relative individual response in acceleration performance, quantified by NAHAP in relation to the baseline footwear condition.

DISCUSSION: The main purpose of this study was to identify, if a manipulation of the LBS in sport shoes has an effect on AccP during the push-off from a resting position in a linear acceleration task. Based on the results, it can be concluded that AccP (quantified by NAHAP) did not change with varying LBS. Even though horizontal GRF impulses were systematically increased by 2.1% and 6.8% for the MS and HS condition, respectively, NAHAP values were not altered significantly, most likely due to the fact that push-off time was increased by 1.8 % and 4.2 %, respectively. These findings are in incongruity with those of Stefanyshyn and Fusco (2004) and Tinoco et al. (2010) who found a positive relation of higher LBS and performance.

The changes in MTP joint kinetics are in line with previous literature, aside from the fact that average MTP joint power and negative MTP joint work were altered in direction to more energy absorption. Therefore, no reduction of absorbed energy was found with higher LBS in the present study while performance was not altered significantly, which is no contradiction to the theory that performance might be improved by reducing the amount of energy absorbed at the MTP (Stefanyshyn & Nigg, 2000; Roy & Stefanyshyn, 2006). Nonetheless, comparing the results of different studies is a difficult task as these studies differ partly in the motions which are analyzed, the parameters to specify performance and also in the mechanical tests to quantify LBS of the tested footwear conditions. There is further the possibility that other design features of different types of footwear can affect the biomechanical outcome of a LBS manipulation.

Table 1

Parameters of interest. Values are described as mean ± std for the BL, MS and HS conditions. Further, effect sizes (ES) are presented for MS and HS in comparison to BL. Results of the repeated measures ANOVA are displayed in the first column. ^{BL}, ^{MS} and ^{HS} indicate a significant difference in the post hoc test, with respect to the BL, ML and HS conditions, respectively.

	ANOVA p-Wert	Baseline (BL)	Medium Stiff (MS)	ES (to BL)	Highly Stiff (HS)	ES (to BL)
NAHAP	0,221	0,466 ± 0,036	0,439 ± 0,050	0,46	0,451 ± 0,054	0,32
Contact Time (s)	0,219	0,457 ± 0,042	0,465 ± 0,047	0,22	0,476 ± 0,046	0,43
Hor. GRF Impulse (kg*m*s^{-1})	0,086	1,869 ± 0,158	1,908 ± 0,195	0,26	1,994 ± 0,262	0,60
CoM to PFA Distance (m)	0,574	0,277 ± 0,047	0,285 ± 0,039	0,31	0,278 ± 0,036	0,04
Trunk Angle (°)	0,987	35,561 ± 6,919	35,538 ± 5,871	0,01	35,395 ± 7,339	0,04
Ratio Horizontal to Resultant Impulse	0,214	0,708 ± 0,032	0,687 ± 0,050	0,43	0,694 ± 0,041	0,36
Vertical GRF Impulse (kg*m*s^{-1})	**0,068**	**3,237 ± 0,305^{HS}**	**3,465 ± 0,620**	**0,43**	**3,552 ± 0,580^{BL}**	**0,63**
Average Knee Joint Power (W/kg)	0,552	2,701 ± 0,674	2,553 ± 0,545	0,31	2,637 ± 0,720	0,12
Average Ankle Joint Power (W/kg)	0,290	2,032 ± 0,313	2,260 ± 0,546	0,36	2,071 ± 0,313	0,11
Average MTP Joint Power (W/kg)	**0,039**	**-0,267 ± 0,087^{HS}**	**-0,267 ± 0,101^{HS}**	**0,01**	**-0,312 ± 0,101$^{BL, MS}$**	**0,69**
Average Lower Extremity Power (W/kg)	0,539	8,872 ± 0,984	8,674 ± 1,481	0,14	8,469 ± 1,525	0,32
Average Horizontal GRF (N/kg)	0,752	4,465 ± 0,435	4,475 ± 0,312	0,02	4,540 ± 0,460	0,21
Average Vertical GRF (N/kg)	0,144	7,692 ± 0,719	8,042 ± 0,825	0,41	8,018 ± 0,728	0,48
Average Resultant GRF (N/kg)	0,214	8,897 ± 0,807	9,210 ± 0,804	0,36	9,219 ± 0,808	0,45
Average Knee Joint Moment (Nm/kg)	0,991	-0,510 ± 0,200	-0,514 ± 0,224	0,03	-0,510 ± 0,207	0,00
Average Ankle Joint Moment (Nm/kg)	**0,003**	**1,155 ± 0,142$^{MS, HS}$**	**1,254 ± 0,131^{BL}**	**0,82**	**1,285 ± 0,139^{BL}**	**0,86**
Average MTP Joint Moment (Nm/kg)	**0,007**	**0,171 ± 0,038^{HS}**	**0,195 ± 0,075**	**0,40**	**0,227 ± 0,071^{BL}**	**1,09**
Average Knee Joint Lever (m)	0,762	0,063 ± 0,022	0,060 ± 0,022	0,17	0,061 ± 0,022	0,15
Average Ankle Joint Lever (m)	**0,001**	**-0,130 ± 0,010$^{MS, HS}$**	**-0,137 ± 0,010^{BL}**	**1,38**	**-0,140 ± 0,010^{BL}**	1,35
Average MTP Joint Lever (m)	**0,019**	**-0,019 ± 0,004^{HS}**	**-0,022 ± 0,008**	**0,31**	**-0,025 ± 0,007^{BL}**	0,90
Negative Work MTP Joint (J/kg)	**0,008**	**-0,122 ± 0,042^{HS}**	**-0,122 ± 0,045^{HS}**	**0,01**	**-0,149 ± 0,052$^{BL, MS}$**	0,83

In the present study LBS ha d a gearing effect for the ankle and MTP, which had previously been shown for distance running mechanics (Willwacher, König, Braunstein, Goldmann & Brüggemann, 2014). Despite the fact that GRF lever arms at the ankle were higher for increased LBS, the athletes showed reduced ankle joint moments in the sagittal plane. It seems that they were not capable of using the potential of the increased lever arms to actually increase power generation at this joint. This might be attributed to an insufficient capacity of the ankle plantar flexors to create a higher power output. Future studies should address this issue by analyzing the effect of LBS increase after a specific strengthening program, designed to improve the power output at the ankle joint. The subjects in the present study did not change their overall body configuration / forward leaning behavior in response to the LBS manipulation. This might be attributed to the fact that forward lean might be constrained by dynamic stability requirements during subsequent steps or simply by the fact that athletes were not used to the new LBS levels of the shoes. As this lack of adaptation time is one of the main limitations of the studies it might be worth addressing the change in acceleration biomechanics during an adaptation period using a longitudinal study design.

CONCLUSION: This study showed that increasing LBS does not always have an enhancing effect on performance. Reasons for that could rather be assumed in the individual muscular capacities of the analyzed athletes. Strengthening the calf and plantar flexor muscle might be a valuable combination with increasing LBS to achieve performance improvements.

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