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ABSTRACT

Introduction: Rocker profile shoes with a proximally placed apex are currently one of the most prescribed shoe modifications for treatment and prevention of lower leg deficits. Three geometrical rocker design parameters apex position (AP), apex angle (AA) and rocker radius (RR) influence both plantar pressure redistribution and kinetic and kinematic alterations of the lower leg. In addition, longitudinal bending stiffness (LBS) of the outsole influences these parameters as well. This study aims to investigate the effects of the LBS in combination with different forefoot radii of rocker shoes on kinematics and kinetics of the lower limb.
Methods: 10 participants walked in standard shoes and six experimental shoe conditions with high and low LBS and three different forefoot rocker radii with the same (proximal) AP and AA. Lower extremity kinematics and kinetics were collected while walking on an instrumented treadmill at preferred walking speed and analysed with a repeated measures ANOVA and Statistical Parametric Mapping (SPM) ($\alpha = .05$; post hoc $\alpha = .05/6$).
Results: SPM analyses revealed no significant differences for LBS and interaction LBS*RR for most research variables in terminal stance (ankle angle, ankle moment, ankle power, foot-to-horizontal angle, shank-to-vertical angle, external ankle moment, ground reaction force angle). A significant LBS effect was found for anterior-posterior position of the centre of pressure during pre-swing and peak ankle dorsiflexion angle. No relevant significant differences were found in spatio-temporal parameters and total work at the ankle between low and high LBS.
Conclusion: This study showed that longitudinal bending stiffness does not affect the biomechanical working mechanism of rocker profile shoes as long as toe plantarflexion is restricted. Providing that the forefoot rocker radius supports at least a normal foot-to-horizontal angle at toe-off, there is no reason to increase sole stiffness to change ankle kinematics and kinetics.

1. Introduction

Rocker profile shoes with a proximally placed apex are currently one of the most prescribed shoe modifications for treatment and prevention of lower leg deficits [1]. Changes in the geometrical forefoot rocker design parameters: apex position (AP) [1], apex angle (AA) and rocker radius (RR) [2] induce both plantar pressure redistributions [3] and lower limb kinetic and kinematic alterations [4,5]. In addition, longitudinal bending stiffness (LBS) of the outsole influences these parameters as well [2,6–9]. LBS can be defined in different ways. Both linear stiffness (N/mm) [7,10] and angular stiffness (Nm/rad or Nm/deg) [8] are used to describe LBS which is defined as the amount of force divided by the deformation of the shoe. In this article angular stiffness (Nm/rad) is used, in line with literature [11,12]. A high LBS limits metatarsophalangeal joint movement, and therefore the toe contribution to gait, and redistributes forefoot plantar pressures [2,6]. Several studies on running have shown that kinematics and kinetics of the metatarsal phalangeal (MTP) joints are influenced by LBS [7,12]. Carbon-plated running shoes with a high LBS have a reported stiffness of 18.5 Nm/rad where low LBS running shoes have a stiffness of 7.0 Nm/rad [12]. High LBS shifts the application of the ground reaction force (GRF) more distally, resulting in an increased ankle plantarflexion moment (PFM) during terminal stance in running [7]. Research in LBS of shoes is, as opposed to walking performance, more focussed on running

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performance where increase in power and positive work are the main goals [9,12]. This contrasts with rocker profile shoes in a clinical setting where reducing PFM is often one of the main goals [5]. When the apex position is positioned proximal to the MTP joints, this results in a decrease in ankle moment arm and moment [5]. However, rocker profile shoes with a low LBS with little restriction of toe plantarflexion are not favourable for maintaining the shape of the forefoot rocker during the stance phase. Basically, the sole, and the toes will plantarflex when the centre of pressure (COP) crosses the apex which counteracts the working mechanism of rocker profile shoes [13,14]. On the other hand, a low LBS with only a restriction of toe plantarflexion and not of toe dorsiflexion allows toe dorsiflexion during terminal stance and pre-swing and maintains the abovementioned working mechanism of the rocker profile [2]. Since LBS controls toe dorsiflexion, the ankle moment arm is thought to be affected as well as depicted in Fig. 1.

The external ankle moment arm is directly related to the PFM and therefore affects the amount force and work needed of the muscle-tendon structures around the ankle. The leading role of the triceps surae and the Achilles tendon, are often seen as one of the major sources of positive work production for gait within the lower leg [15]. Besides the role of ankle biomechanics, the more distal muscle-tendon structures (like toe flexors) seem to be of great influence for the push-off as well [16]. Adjusting LBS affects the contribution of the distal muscle-tendon structures of the foot during the stance phase of gait in terms of kinematics [17], work and power [16], plantar fascia strain [18], and running energetics [7,11].

AP, AA, forefoot RR and LBS are known to affect the kinetics, kinematics, and plantar pressure during gait [2,3,19]. However, it remains unclear what the combined effect is. Subsequently, the size of the forefoot RR in rocker profile shoes with high LBS determines the foot-to-horizontal angle (FHA), and determines therefore ankle dorsiflexion angle (DFA) and PFM at terminal stance [19]; it remains unclear if the same results are found with a low LBS and restriction of plantarflexion. The moment arm that in theory would reduce with reduced LBS of a rocker profile with different radii compared to high LBS rocker profile has never been tested experimentally. This study aims to investigate the effects of LBS and different forefoot radii of rocker shoes with the same proximal AP and AA on kinematics and kinetics of the ankle (DF, PFM and power) and its underlying variables (FHA, shank-to-vertical angle, external ankle moment arm, anterior-posterior position of the COP and sagittal plane GRF vector angle). It was hypothesized that a larger LBS decreases DF and PFM. Furthermore, a larger LBS decreases FHA at toe-off, without an effect the shank-to-vertical angle and the GRF vector angle, and increases the external ankle moment arm and more distal COP at pre-swing.

2. Methods

2.1. Participants

Ten healthy adults participated in this pilot study. Inclusion criteria were female gender, age≥18 and fit the shoe sizes of the experimental female shoe (EU37). Exclusion criteria were the use of lower leg orthoses including custom inlays and self-reported diseases or injuries that influence gait. Written informed consent was obtained before starting the experiments. The local Medical Ethics Committee approved this research (METc 2018.060).

2.2. Shoe conditions

In this study we used neutral athletic shoes (Katy, Dr Comfort, Mequon, WI, USA) sizes EU 37 all with medium (M) width. The experimental shoes were fitted with 3D-printed 30 mm nylon (Polyamide 12) midsoles and the control shoes were the original shoes (not adapted). A total of seven shoe conditions were used, a control condition without toe restrictions, three high LBS shoes with complete restriction of the toe motion and three different radii (R16, R18.5 and R21) and three lower LBS shoes with toe plantarflexion restriction, but no dorsiflexion restriction, also with three radii (F16, F18.5 and F21). Fig. 2 shows a hinge-like structure making toe dorsiflexion possible where plantarflexion of the toes was restricted. The cable ties were applied to restrict toe dorsiflexion (high LBS). All shoes had neutral inlays (EVA; 25 durometers, shore A, thickness: 6 mm). The LBS of the control and low LBS shoes was 6.7 Nm/rad and 24.1 Nm/rad respectively. All low LBS shoes allowed toe dorsiflexion and restricted toe plantarflexion. The high LBS shoes were outside the measure limits (>180 Nm/rad). The apex position of the experimental shoes was located at 55% of the total shoe length of 260 mm with an apex angle of 90°. The bending axis was placed parallel to the apex at 66% of the total shoe length allowing sole dorsiflexion. The RR of the experimental shoes were 160 mm, 185 mm, and 210 mm respectively. The mean (SD) weight for each pair of shoes was 240, 588, 556 and 558 grams for control, R16, R18.5 and R21 respectively. Small weight differences are explained by a different number of weight reducing holes in the internal structure of the outsoles.

2.3. Set-up

Measurements were performed at the GRAIL (Gait Realtime Analysis Interactive Lab; Motekforce Link, Amsterdam) of the Department of Rehabilitation Medicine, University Medical Center Groningen. A ten-camera motion capture system (Vicon Bonita 10, Oxford, UK; Fs = 100 Hz) tracked marker trajectories of 22 reflective markers placed according to the lower-limb Human Body Model 2 [20] and two force plates (Motekforce Link; Amsterdam; Fs = 1000 Hz) recorded analogue data. Both kinetic and kinematic data were synchronised and normalised

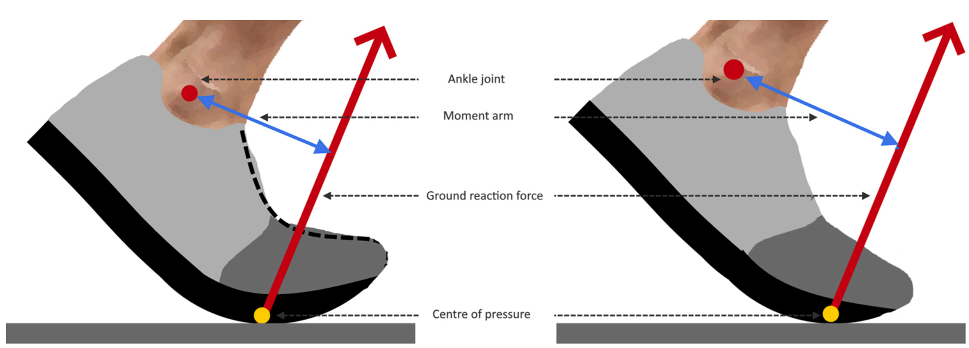

Fig. 1. Schematic view of the push-off phase of gait with a shoe with low LBS (left) and high LBS (right). The moment arm (blue double arrow) is affected by the position of the ankle joint, centre of pressure, the direction of the ground reaction force and toe dorsiflexion.

Fig. 2. Experimental shoe with a forefoot rocker radius of 210 mm. Cable ties were used to increase LBS of the rocker sole (graphically visualized in blue). Cable ties were fitted in the holes and did not stick out the bottom nor effect sole geometry. Shoe soles had a thickness of 35 mm and toe spring of 20 mm.

to 100% gait cycle and for body mass (kg).

2.4. Experimental procedure

Only the dominant foot (foot used to kick a ball) was taken into account in the data analysis. Preferred walking speed while walking on the control shoes was determined by taking the mean speed during 60 s self-paced walking on the GRAIL after a 60 s familiarization period. In all conditions this preferred walking speed was fixed. Measurements started with the control condition, followed by the experimental conditions in a randomized order of RR defined by a custom-made script in Matlab (R2016b). For each RR, first the high LBS condition was measured, followed by the low LBS condition by quickly cutting the cable ties. Each measurement contained 60 s of familiarization followed by 60 s of recording.

2.5. Data analysis

With a custom-made Matlab-script data of twelve steps per condition for each subject were averaged and used for further analysis of the selected gait parameters. Data were filtered with a fourth order zero-lag Butterworth low-pass filter with a 4 Hz cut-off frequency. Per condition spatio-temporal parameters, peak values for DFA, PFM and ankle power as well as ankle work and peak GRF values were determined during terminal stance and pre-swing (time region of interest). Ankle moment arm was determined based on the marker trajectories of the ankle and the GRF vector. Shank-to-vertical angle and FHA during the stance phase were determined based on the marker trajectories of the shank and the foot markers respectively and calculated. COP trajectories were transformed to the foot coordinate system.

2.6. Statistical analysis

Repeated measures analyses of variance (ANOVA) were used to estimate the effect shoe condition ($p < .05$) on the discrete gait parameters (spatio-temporal parameters, peak DFA, peak PFM and ankle work) and corrected Bonferroni post hoc procedure (IBM Statistics 23). Continuous gait parameters were analysed using 1-dimensional Statistical Parametric Mapping (SPM) [21,22]. This allowed us to detect both temporal and magnitude related changes simultaneously and side effects outside the region of interest. All SPM analyses were conducted with a custom-made Matlab-script (R2016b) and using the open-source software package spm1D 0.4 (www.spm1d.org). The focus of this manuscript is on the effect of LBS in rocker shoes, however comparisons of low LBS shoes with the control condition were performed as well. Therefore, six conditions were used: control versus low LBS ($n = 3$) and per RR low versus high LBS ($n = 3$). The significance level for all statistical tests was set a priori to .05. The significance level of the six post hoc paired t-tests was Bonferroni corrected and set at .0083 (.05/6). Post hoc tests were conducted when significant main effect of LBS or interaction effect LBSxRR were found. Main effects of rocker radius with high LBS were previously reported [19].

3. Results

The subjects had a mean ($\pm$SD) age of 20.8 ($\pm$2.4) years, bodyweight of 67.5 ($\pm$10.5) kg, and body height of 164.4 ($\pm$3.1) cm. The right foot was dominant in all participants. Per subject, treadmill speed was fixed at their self-selected walking speed in all 7 conditions. Average walking speed was 1.19 ($\pm$0.18) m/s. No significant effects of LBS were found in spatio-temporal parameters between the rocker conditions (Table 1). Analyses of the peak values show a significant main effect of LBS in peak DFA and peak ankle power. Low LBS increases both peak DFA ($p = .002$) and peak positive ankle power ($p = .002$). Both positive ankle work ($p = .004$) and total ankle work ($p = .011$) are increased as well with low LBS. No LBS effects were found on peak PFM ($p = .276$). Peak GRF during terminal stance are both increased for low LBS for the anterior-posterior ($p = .026$) and vertical ($p = .004$) component. For all parameters no interaction effects were found.

SPM analyses with repeated measures ANOVA showed a significant main effect of LBS for ankle angle (71–95%, $p < .001$) and ankle moment (5–7%, $p < .038$) (Fig. 3). No significant differences were found between rocker shoe conditions. Post hoc analyses show a significant decrease in DFA in terminal stance between the low LBS rockers and the control condition ($p < .01$) and an increase between F16 versus R16 and F18.5 versus R18.5 ($p < .01$). All low LBS rockers show a significant decrease of PFM in terminal stance ($p < .01$) compared to the control condition and an increase in F16 relative to R16 ($p < .01$). There is no significant main effect of LBS on ankle power in the whole stance phase. Therefore, no post hoc analyses were conducted for ankle power. SPM analyses with repeated measures ANOVA showed a significant main effect for LBS for shank-to-vertical angle (10-12%; $p = 0.047$) and FHA (69-84%; $p < .001$), but no interaction effect between rocker shoe conditions (Fig. 4). Post hoc analyses showed a significant increase for both shank-to-vertical angle and FHA in terminal stance between F18.5 and R18.5 ($p < .01$) and a decrease for the low LBS rockers and the control condition ($p < .01$).

SPM analyses with repeated measures ANOVA showed a significant main effect for LBS between rocker shoe conditions for ankle moment arm (4-7%; $p < .05$), and an interaction effect for ground reaction force angle (1-5%; $p < .01$) but not during terminal stance (Fig. 5). However, post hoc analyses showed a significant increase for moment arm in terminal stance between for F18.5 compared with R18.5 ($p < .01$). For the COP trajectory there was a significant main effect of LBS during pre-swing ($p < .01$). Post hoc analyses showed a more anterior COP location during terminal stance in F18.5 compared to R18.5 ($p < .01$).

4. Discussion

To the best of our knowledge this is the first study that evaluated the effects of LBS in rocker profile shoes with different forefoot radii on biomechanics of gait. We showed with SPM that rocker profile shoes with a proximal apex position reduce peak DFA and peak PFM in line with previous research [4,5], but that there is no main effect nor interaction effect of LBS on the research variables in DFA, PFM and ankle power in terminal stance. In other words, the theoretical decrease in PFM with the use of rocker shoes with low LBS compared to high LBS

Table 1
Mean values, standard deviations, and results of the repeated measures ANOVA for spatio-temporal parameters, kinetics and kinematics. Only main effects for LBS and interaction effect LBS*RR were reported (α ≤ .05). Post hoc significant differences (marked with a (F-shoe vs. Control shoe), b (vs. R16), c (vs. R18.5), d (vs. R21). DFA = Ankle dorsiflexion angle; PFM = Ankle plantarflexion moment; GRF = Ground reaction force; BW = body weight.

	Control shoe	R16	F16	R18.5	F18.5	R21	F21	p (LBS)	p (LBS*RR)
Spatio-Temporal									
Cadence (steps/min)	113.2 (9.1)	111.1 (9.2)	111.1 (9.7)	110.4 (8.9)	110.4 (9.3)	109.8 (9.3)	109.4 (9.9)	.215	.525
Step length (m)	0.63 (0.06)	0.64 (0.05)	0.64 (0.05)	0.65 (0.05)	0.65 (0.05)	0.65 (0.05)	0.65 (0.05)	.933	.270
Step width (m)	0.13 (0.03)	0.14 (0.02)	0.14 (0.02)	0.14 (0.02)	0.14 (0.02)	0.14 (0.02)	0.14 (0.02)	.083	.400
Stance time (s)	0.75 (0.06)	0.77 (0.06)	0.77 (0.06)	0.81 (0.08)	0.81 (0.10)	0.78 (0.05)	0.77 (0.06)	.706	.947
Swing time (s)	0.32 (0.05)	0.31 (0.05)	0.31 (0.06)	0.29 (0.10)	0.28 (0.11)	0.32 (0.07)	0.32 (0.07)	.869	.941
Stance percentage (%)	70.0 (3.7)	71.1 (3.4)	71.2 (3.8)	74.0 (8.2)	74.3 (9.4)	71.3 (4.9)	71.6 (5.0)	.847	.945
Ankle									
Max DFA (°)	13.89 (2.49)	9.80 (3.65)	10.49 (3.69)[a,b]	10.62 (2.95)	11.04 (3.02)[a]	10.91 (2.70)	11.39 (2.83)[a]	**.002**	.799
Max PFM (Nm/kg)	1.54 (0.15)	1.25 (0.15)	1.27 (0.15)	1.31 (0.16)	1.33 (0.17)	1.39 (0.16)	1.39 (0.17)	.276	.323
Max power generation (W/kg)	3.71 (0.93)	2.80 (0.79)	3.00 (0.86)[a,b]	2.81 (0.74)	2.97 (0.68)[a,c]	2.81 (0.68)	2.89 (0.75)[a]	**.002**	.211
Positive work (J)	0.37 (0.08)	0.30 (0.07)	0.32 (0.07)[a,b]	0.30 (0.06)	0.32 (0.06)[a,c]	0.30 (0.06)	0.31 (0.06)[a]	**.004**	.852
Negative work (J)	−0.16 (0.02)	−0.11 (0.02)	−0.11 (0.03)	−0.12 (0.02)	−0.12 (0.03)	−0.13 (0.02)	−0.13 (0.02)	.088	.503
Total work (J)	0.20 (0.05)	0.19 (0.05)	0.20 (0.06)	0.18 (0.05)	0.19 (0.05)[a,c]	0.17 (0.06)	0.18 (0.05)	**.011**	.879
Ground reaction Force									
Peak anterior GRF (x BW)	0.16 (0.02)	0.16 (0.02)	.016 (0.02)[a,b]	0.15 (0.02)	0.16 (0.02)	0.15 (0.02)	0.15 (0.02)	**.026**	.938
2nd Peak Vertical GRF (x BW)	1.07 (0.03)	1.05 (0.03)	1.07 (0.03)[a,b]	1.05 (0.03)	1.07 (0.03)[a,c]	1.06 (0.03)	1.06 (0.04)	**.004**	.112

were not substantiated. Analysis of the peak values (Table 1) showed a significant increase in peak DFA (+ <1°), peak ankle power (+3-7%) and positive (+3%) and total ankle work (+5%). Differences in outcomes derived from SPM and peak value analysis can be explained by temporal shifts between strides.

Increase of peak DFA for low LBS shoes can be seen as a negative point for people with AT, as it increases Achilles tendon strain, however differences were negligibly small and therefore prone to measurement errors. There was no significant effect on DFA found with SPM, probably because of temporal variability and the magnitude of the difference (<1°). As our aim was to decrease peak DFA, it is important to observe both statistical methods as outcomes might be different. As the significant increase in peak DFA was small and no significant differences in SPM were found, the effect of LBS on DFA was therefore interpreted as not relevant.

Toe dorsiflexion increases FHA in pre-swing when the LBS is low. A flat outsole with high LBS is very challenging to walk on and a rocker curvature is therefore essential to facilitate FHA at toe-off. The FHA at toe-off can be the explanation that no LBS effects were found in this study as this describes the rotation of the foot at the end of stance phase. In this study subjects had a mean FHA at toe-off (at 67% gait cycle) of 50.4° in the control shoes and 52.0/49.3/49.0° for the rigid rocker shoe configurations R16/R18.5/R21, respectively. This means that the radii already facilitated or almost facilitated sufficient FHA during the third rocker and that, therefore, no or only limited toe dorsiflexion was necessary for normal roll over. The mean FHA values in the low LBS conditions, F16/F18.5/F21 of 51.6/49.7/50.6° respectively, show a small increase for F18.5 and F21 when compared with R18.5 and R21, respectively. In theory this would mean that toe dorsiflexion in F18.5 and F21 was necessary to facilitate the FHA at toe-off of 50.4° as measured in the control condition. In other words, toe dorsiflexion restored a normal FHA, however the differences are marginal and should therefore be interpreted with caution. Further research with a larger RR and toe dorsiflexion measurement must confirm this theory as there would be larger difference in FHA at toe-off between the high and low LBS condition when the radius is larger. Another explanation might be that the LBS is still too high in the low LBS shoes and therefore limiting toe dorsiflexion. The LBS of 24.1 Nm/rad is higher than the control shoe (6.7 Nm/rad), but comparable with carbon plated running shoes [12]. Further research should use rocker profiled shoes with identical low LBS to the control shoe and with added toe plantar flexion restriction.

This study placed the bending axis of the sole at 66% of the shoe length and 90° rotated relative to the longitudinal axis of the shoe. As the MTP axis might be different between subjects, it might affect the amount of toe flexion as the bending axis is not individually adjusted. The 90° apex angle and bending axis are aligned with each other and not in line with the natural angle of the MTP axis. To the best of our knowledge there are no studies describing the effect of apex angle and the bending axis on sagittal ankle kinematics and kinetics of rocker profile shoes. Research in prosthetics, where the shape of the prosthetic feet (toe length, foot arch length and bending axis position) is manipulated, might help to optimize bending axis orientation, position and stiffness in rocker profile shoes [23]. However, we expect it does not affect the outcomes of our study. Further research should measure the effects of location and rotation of the bending axis of the shoe sole.

LBS aims to influence the forefoot kinematics. Multi-segment foot models are therefore an important tool to separate toe-, MTP joint and hindfoot biomechanics. However, measuring foot biomechanics by using multi-segment foot models in a walking or running shoe requires modifications to the shoe to place the markers directly on the skin. This study used markers attached to the shoes, which makes measuring the motion of the foot in the shoe impossible. However, modifications to the shoe may affect its structure and support and are therefore not applied for this study. Common marker templates like Plug-in Gait (Vicon, Oxford, UK) and HBM2 (Motek Forcelink, Amsterdam, NL) with a standard one-segmental foot model overestimate the ankle contribution in terms of power as power of the MTP joint is not measured [24]. In this study low LBS did not affect positive ankle power at terminal stance but measuring a multi-segment foot model might have revealed a (possible) decrease in positive ankle power within the low LBS rocker conditions.

Weaknesses in the used marker template in combination with the used shoes might have given flaws in the calculation of biomechanical parameters. A larger sole thickness might for instance increase the error of COP and ankle external moment arm calculation on the instrumented treadmill, as the distance of the ankle joint to the ground increases and the force measurement errors are amplified [25]. However, in this study we only focused on the effect of LBS within each RR, thus within marker set-up and with constant sole thickness. As marker set-up and sole thickness were constant for each RR, and twelve steps per measurement condition were used, the chance of random errors were decreased.

The used cut-off frequency of 4 Hz might have filtered out portions of the human data. However, it was important to make the data as smooth as possible to make it suitable for SPM. A higher filter frequency led to more ‘noise errors’ with an instrumented treadmill, which would generate large SPM-errors and therefore poor SPM interpretations. This would not have compensated for the potentially reduced amplitudes

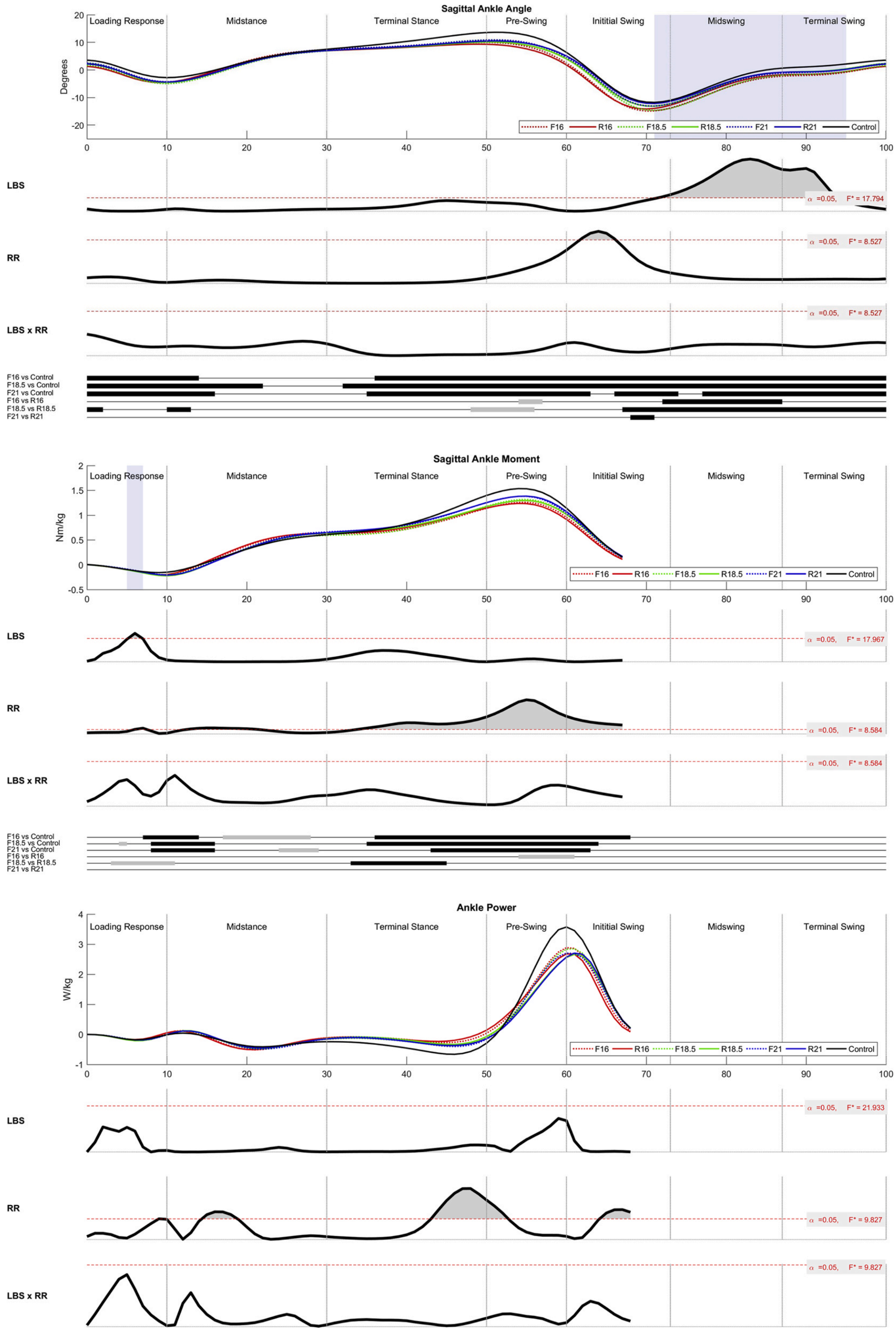

Fig. 3. Sagittal ankle angle, sagittal ankle moment and ankle power mean patterns for the control, flexible (F) and rigid (R), and radius 16, 18.5 and 21 cm conditions. For moment and power only the stance phase is plotted. The blue areas indicate statistical significantly differences between conditions for main effect longitudinal bending stiffness (LBS), followed by the time-dependent F-values of the SPM (main statistical test; analysis of variance) for all subjects (dashed red line; $\alpha \leq .05$) for LBS and rocker radius (RR) and the interaction effect LBSxRR. Grey areas indicate regions with significant differences. For the variables with a main effect of LBS (ankle angle and moment), post hoc tests between shoe conditions of interest were performed. Solid black (decrease) and grey bars (increase) span the region of the gait cycle where significant ($\alpha \leq .0083$) differences were observed.

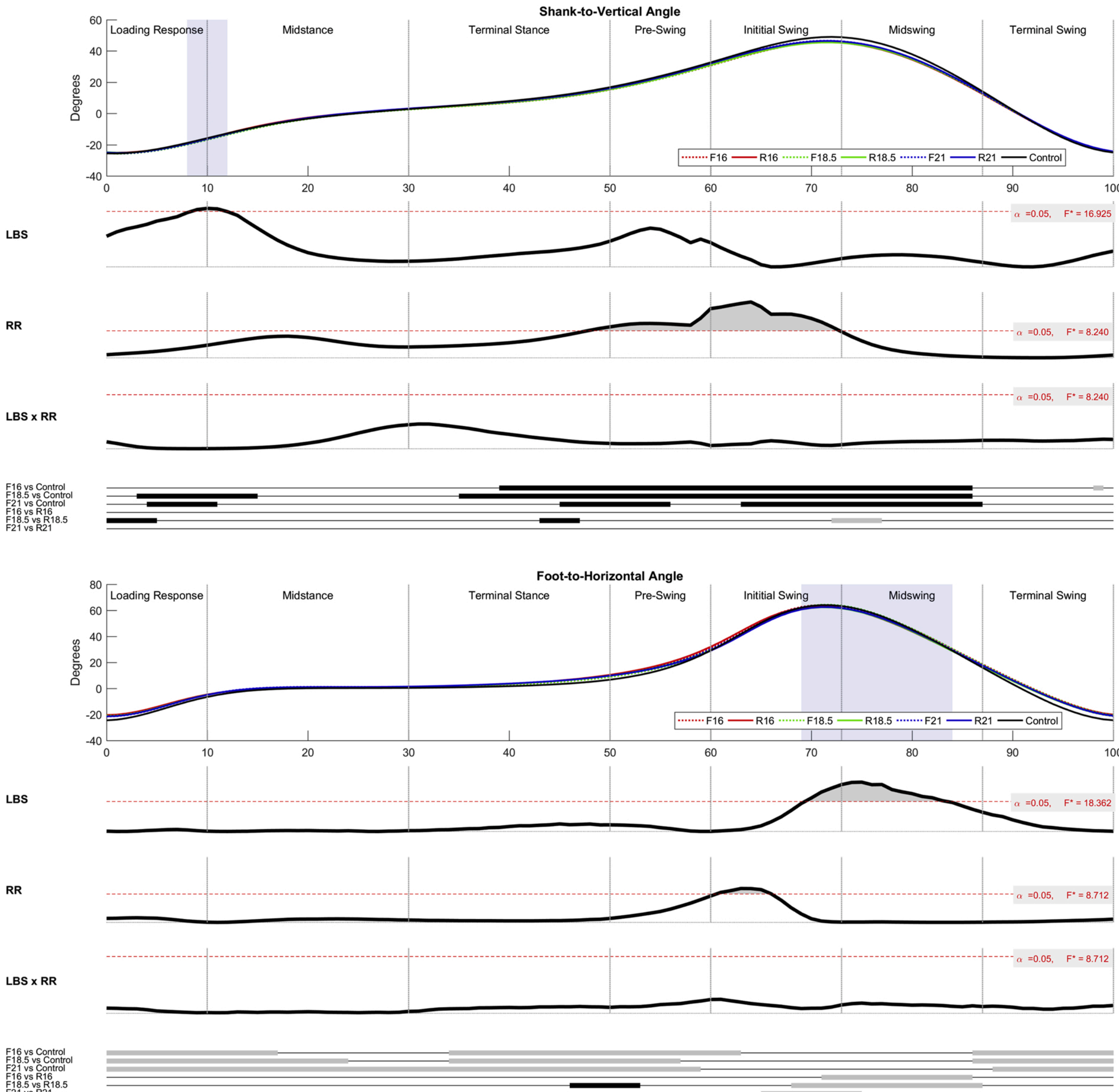

Fig. 4. Shank-to-vertical angle and foot-to-horizontal angle mean patterns for the control, flexible (F) and rigid (R), and radius 16, 18.5 and 21 cm conditions. The blue areas indicate statistical significantly differences between conditions for main effect longitudinal bending stiffness (LBS), followed by the time-dependent F-values of the SPM (main statistical test; analysis of variance) for all subjects (dashed red line; $\alpha \leq .05$) for LBS and rocker radius (RR) and the interaction effect LBSxRR. Grey areas indicate regions with significant differences. Both variables showed a main effect of LBS, therefore post hoc tests between shoe conditions of interest were performed. Solid black (decrease) and grey bars (increase) span the region of the gait cycle where significant ($\alpha \leq .0083$) differences were observed.

resulting from the 4 Hz filtering. Future research with an instrumented (movable) treadmill should consider more reliable raw data capturing that could be filtered with higher cut-off frequencies.

In this study, no LBS effects were found for rocker shoes with different radii at comfortable walking speed. Therefore, other parameters like plantar pressure [2], plantar fascia strain [18], MTP kinematics and kinetics [7] or comfort should determine the selection of the appropriate LBS. Additionally, other conditions like standing, stair walking, car driving, other daily life activities and even shoe material may determine whether rocker shoes should be stiffened or not. When RR does not allow for sufficient FHA at toe-off, as is the case in far most conventional shoes, low LBS induces toe dorsiflexion to allow a normal roll-over curve during gait. Optimally, LBS should be adjusted to the RR, AP, and the amount of toe dorsiflexion restriction of the patient. This will allow larger RR to facilitate sufficient FHA by using toe dorsiflexion. This will reduce the outsole thickness of the rocker shoe as smaller radii need more sole thickness. LBS should therefore not be dichotomous (high or low), but more a continuous rocker design parameter to meet individual LBS optimization. When there is no contra-indication for toe dorsiflexion, a uniform standardisation of LBS is essential to improve clinical prescription making.

5. Conclusion

This study showed that longitudinal bending stiffness does not affect the biomechanical working mechanism of rocker profile shoes if toe plantarflexion is restricted, and the rocker profile facilitated a normal FHA at toe off. Therefore, provided that the forefoot rocker radius and apex position support at least a normal foot-to-horizontal angle at toe-off, there is, without a contra-indication, no reason to increase sole stiffness to change ankle kinematics and kinetics.

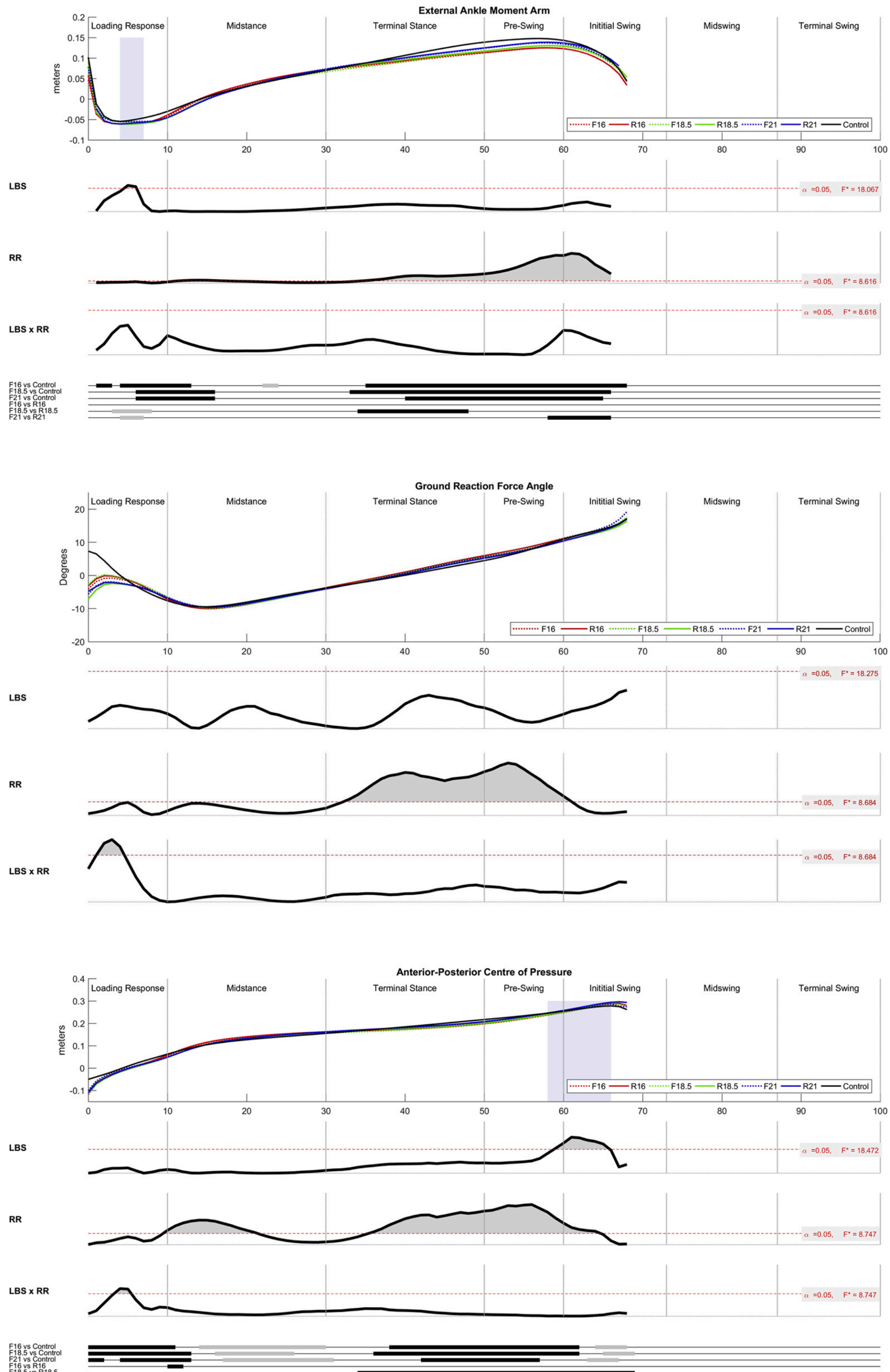

Fig. 5. External ankle moment arm, ground reaction force vector and anterior-posterior centre of pressure mean patterns for the control, flexible (F) and rigid (R), and radius 16, 18.5 and 21 cm conditions. The blue areas indicate statistical significantly differences between conditions for main effect longitudinal bending stiffness (LBS), followed by the time-dependent F-values of the SPM (main statistical test; analysis of variance) for all subjects (dashed red line; $\alpha \leq .05$) for LBS and rocker radius (RR) and the interaction effect LBSxRR. Grey areas indicate regions with significant differences. For the variables with a main effect of LBS (moment arm and centre of pressure), post hoc tests between shoe conditions of interest were performed. Solid black (decrease) and grey bars (increase) span the region of the gait cycle where significant ($\alpha \leq .0083$).

Declaration of Competing Interest

The authors report no declarations of interest.

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