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Effects of low-pass filter combinations on lower extremity joint moments in distance running

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ABSTRACT

Inverse dynamics is a standard tool in biomechanics, which requires low-pass filtering of external force and kinematic signals. Unmatched filtering procedures are reported to affect joint moment amplitudes in high impact movements, like landing or cutting, but are also common in the analysis of distance running. We analyzed the effects of cut-off frequencies in 94 rearfoot runners at a speed of 3.5 m/s. Additionally, we investigated whether the evaluation of footwear interventions is affected by the choice of cut-off frequencies. We performed 3D inverse dynamics for the hip, knee and ankle joints using different low-pass filter cut-off frequency combinations for a recursive fourth-order Butterworth filter. We observed fluctuations of joint moment curves in the first half of stance, which were most pronounced for the most unmatched cut-off frequency combination (kinematics: 10 Hz; ground reaction forces (GRFs): 100 Hz) and for more proximal joints. Peak sagittal plane hip joint moments were altered by 94% on average. We observed a change in the ranking of subjects based on joint moment amplitude. We found significant ($p < 0.001$) footwear by cut-off frequency combination interaction effects for most peak joint moments. These findings highlight the importance of cut-off frequency choice in the analysis of joint moments and the assessment of footwear interventions in distance running. Based on our results, we propose to use matched cut-off frequencies around 20 Hz in order to avoid large artificial fluctuations in joint moment curves while at the same time avoiding a severe removal of physiological high-frequency signal content from the GRF signals.

1. Introduction

Inverse dynamics is a standard tool to calculate net joint moments and forces for a known motion using segmental inertial properties, kinematics and external forces acting on the bodies (Bresler, 1950). Net internal joint moments represent the summed action of all forces creating moments across a joint. These internal moments are generated by muscle-tendon, ligament and articular contact forces; hence a correct estimation of joint moments helps to understand the underlying causes of movement and the mechanical loading of biological structures. Joint moment variables are used to estimate traumatic or overuse injury risk (Hewett et al., 2005; Padua et al., 2009; Stefanyshyn et al., 2006). However, a problem of inverse dynamics calculations is that signal noise is amplified when segment position and orientation data is differentiated twice to calculate segmental accelerations. Noise is also inherent in the measurement of external forces such as the ground reaction force (GRF), even though to a smaller extent compared to kinematic data. Therefore, low pass filtering is commonly applied to remove noise from segment kinematics and force signals. However, the signal to noise ratios and frequency contents of GRF and kinematics data are typically very different, which implies different filtering strategies for the two signal types.

In biomechanics, a recursive fourth order low pass Butterworth filter with specific cut-off frequency is often used. However, there is no consistency in the choice of cut-off frequencies for force and kinematic data, even though it has been shown that the choice of cut-off frequencies affects joint moment calculations, which sometimes can make comparisons across studies difficult (Kristianslund et al., 2012). In previous studies, some authors chose their filter frequencies for kinematic and GRF data based on the different signal to noise ratios or based on a frequency content analysis of their data. This approach typically results in different filter frequencies for kinematic and kinetic data. Other authors highlighted the inconsistency in the equations of motion introduced by removing the high frequency component from the kinematic data while keeping high frequency content in the GRF data. Consequently,

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these authors have filtered kinematics and GRF data at the same cut-off frequency while sacrificing the high frequency components of the signals.

First attempts to address the issue of different low-pass filter cut-off frequencies have been made by Van den Bogert and de Koning by using simulated running movement data with added white noise for investigating the effects of different cut-off frequencies on inverse dynamic results. The authors concluded that each segment should be filtered with a different cut-off frequency to get either the best force or the best joint moment results (Van den Bogert and De Koning, 1996). Bisseling and Hof suggested that in analyzing movements with high accelerations like, e.g., landing maneuvers, marker trajectories, and external force data should be filtered with the same cut-off frequency to avoid filter artifacts in the assessment of joint moments (Bisseling and Hof, 2006). Bezodis and colleagues supported these findings for sprinting while using manual marker digitization and 2D video analysis to determine kinematics. The authors observed that once the cut off frequency for kinetics was higher than the kinematics cut-off frequency, fluctuations in joint moment curves after initial ground contact were amplified (Bezodis et al., 2011, 2013). In another study aiming at determining the anterior cruciate ligament (ACL) injury risk of female handball players, Kristianslund and colleges also concluded that marker trajectories and force plate data should be filtered with the same cut-off frequency when analyzing knee joint moments in cutting maneuvers. These authors could show that the ranking of players based on the external knee abduction moment, which is considered to be a sensitive parameter for ACL injury risk, changed with different cut-off frequencies (Kristianslund et al., 2012). Nonetheless, another study analyzing drop jumping movements failed to show that filter frequencies influence the ranking of participants even though external knee abduction moments were also systematically affected by cut-off frequency combinations (Roewer et al., 2014).

Other than traumatic ACL injuries in team sports, in distance running, chronical overuse injuries are more common. Several studies have identified a relationship between altered joint moment amplitudes and the risk of sustaining an overuse injury in distance running. In a retrospective analysis, it was found that runners with a history of Achilles tendinopathy show altered joint moment amplitudes at the distal tibia compared to runners with no history of overuse injuries (Williams et al., 2008). Furthermore, it was found that female runners who had suffered from iliotibial band syndrome were characterized by greater peak internal rearfoot invertor moments and greater peak internal knee abduction moments (Ferber et al., 2010, Stickley et al., 2018). In another study, Stefanyshyn and colleagues have shown that an increased internal knee abduction impulse is related to the development of patellofemoral pain in distance runners (Stefanyshyn et al., 2006).

Consequently, technical interventions (like e.g. footwear, insoles, and orthoses) often attempt to alter the frontal plane joint moment requirements during locomotion (Radzimski et al., 2012). Despite the importance of joint moment variables for injury risk in running, and even though running might also be considered a high impact movement, the effect of different filtering strategies on joint moment calculations in distance running have only been addressed in one study so far (Van den Bogert and De Koning, 1996). However, this study was performed on only one simulated running trial and limited to the sagittal plane. A comprehensive study including data from multiple male and female runners, which includes frontal and transverse plane joint moments, is currently missing in the literature. Moreover, the effects of cut-off frequency combinations in the assessment of technical interventions (like e.g. footwear) have not been systematically analyzed, yet.

Therefore, the purpose of this study was to explore the effects of applying different cut-off frequencies to marker trajectory and GRF signals on resulting 3D internal joint moments in a comprehensive data set. Furthermore, we wanted to assess whether filter cut-off frequency combinations can affect the results and the interpretation of a footwear intervention with respect to peak joint moments. We hypothesized that (1) unmatched filter frequencies for GRFs and marker trajectories result in more pronounced fluctuations of joint moment curves within the initial stance phase. Further, we hypothesized that (2) the effects of unmatched filtering are more pronounced for more proximal joints since the effects of filtering on segment accelerations accumulate for each added joint in the kinematic chain. Finally, we hypothesized (3) that peak joint moment differences between footwear conditions are affected by the choice of cut-off frequency combinations.

2. Methods

2.1. Participants

For this study, were-analyzed a data set consisting of 104 participants (Fischer et al., 2017). The local ethics committee approved all methods. Participants were injury-free within the year before the study. All participants gave their informed written consent before participation in the study. All participants repeated a running protocol in 4 different footwear conditions: a neutral running shoe (Brooks Glycerin), a motion control shoe (Brooks Adrenaline), a stability shoe (Brooks Beast) and barefoot running on a 1 cm thick Ethylene Vinyl Acetate (EVA) foam. We collected five valid trials per participant; a trial was considered to be valid when the runners achieved an average running speed of 3.5 ± 0.2 m/s. We controlled running speed with two double photocells, placed 4 m before and behind the force platform in the direction of running. We only included participants with a distinct local maximum in the vertical GRF in initial stance in the neutral running shoe condition in four of their five trials. This resulted in a subject sample of ninety-four participants (female: 48; male: 46; age: 40 ± 13.6 years; height: 1.75 ± 9.95 m; mass: 69.3 ± 11.4 kg). We used this inclusion criterion because we expected that the relative amount of high frequency components is higher in runners with a distinct local maximum of the vertical GRF in initial stance. Consequently, we expected that GRF filter frequencies would affect the calculated joint moments of these runners more compared the joint moments of other runners.

2.2. Experimental procedures

Ten infrared cameras (MX–F40, Vicon Motion Systems, Oxford, UK) recorded 22 retro-reflective markers (diameter 10 mm) with a sampling frequency of 250 Hz. We attached markers at the following anatomical landmarks; both anterior superior iliac spine and posterior superior iliac spine; medial and lateral femoral condyles; medial and lateral malleoli; medial, lateral and posterior aspect of the calcaneus; basis and head of the first metatarsals. We attached rigid shell carbon fiber marker clusters (mass between 31 g and 40 g) over the distal lateral aspect to the leg using an underwrap attachment (Fischer et al., 2017). We captured GRFs using a force plate sampling at 1250 Hz (Kistler AG, Winterthur, Switzerland, 0.9 m × 0.6 m). We calculated segmental inertial properties based on anthropometric data derived from de Leva (1996). We defined knee and ankle joint centers as the midpoints between medial and lateral femur condyles and both malleoli markers, respectively. We defined hip joint centers and coordinate system according to Bell et al. (1990) and Seidel et al. (1995). We defined the stance phase as the time points where the unfiltered vertical GRF component exceeded 20 N.

2.3. Data analysis

For filtering of marker trajectory and GRF data, we applied in total six combinations of cut-off frequencies. These included four combinations of unmatched cut-off frequencies for kinematic (K) and force (F) data (K10F50; K10F100; K20F50 and K20F100). Further, we included two matched filter combinations one with 10 Hz and one with 20 Hz (K10F10 and K20F20). We applied all cut-off frequencies on a recursive fourth order low pass Butterworth filter.

We determined lower extremity resultant internal joint moments with the explicit expression provided by Hof (1992) using a rigid body model of the lower extremities including a forefoot, rearfoot, shank, thigh and pelvis segment (see Willwacher et al., 2016b). We normalized all internal joint moments to body mass; time normalization was performed to stance time by interpolating to 201 data points. All model calculations were made using custom made Matlab algorithms (R2018a, The Mathworks, Natick, USA).

2.4. Statistical analysis

To assess whether filtering affected the joint moment curves in the time domain, we used repeated-measurement ANOVAs of one-dimensional statistical parametric mapping (SPM) to compare the time series of internal joint moment curves of ankle, knee and hip joints in each plane of motion for the six different filter combinations in the neutral running shoe condition. We implemented all SPM analyses using the open-source spm1dcode (SPM, v.M0.4, www.spm1d.org) in Matlab. Further, we conducted one factor (cut-off frequency combination) repeated-measurement ANOVAs in the neutral running shoe condition for peak values (discrete parameters) for each joint in each plane to identify potential filter combination main effects. If the ANOVA revealed a significant result, we performed post hoc testing using Bonferroni corrected alpha levels. We computed partial eta squared (η_p^2) for repeated-Measurement ANOVAs as effect size. Further, we calculated effect sizes for pairwise comparisons (Cohen, 1988) for discrete parameters for each possible filter combination using the following equation:

$$d = \frac{mean(X_1 - X_2)}{sd(X_1 - X_2)} \quad (1)$$

with X_1 and X_2 being the data vectors at of filter condition (1) and filter condition 2, respectively. In order to quantify the filter-related effects on the ranking of the participants within the neutral running shoe condition, we calculated Spearman's rank correlation coefficients (ρ) for the discrete parameters (Altman, 1991).

In order to assess whether peak joint moment differences between footwear conditions are affected by the choice of cut-off frequency combinations, we performed two factor (cut-off frequency combination, footwear) repeated measures ANOVAs. For all discrete parameter analyses, we chose peak values; either maxima or minima; based on the dominant load direction. We set the alpha level to 0.05 for all tests.

3. Results

The SPM analysis revealed significant filter frequency main effects on the time-series of internal joint moment curves across all planes and joints in the neutral running shoe condition (Fig. 1). Most pronounced fluctuations in joint moments were observed for the hip in the sagittal and transversal plane and the knee joint in the frontal plane within the initial stance phase. (Fig. 1B–D). The duration of significant differences between filter combinations during initial stance increased from distal to proximal joints (Fig. 1A–I). Except for the knee and ankle joint in the frontal plane, the SPM revealed significant differences also during the push-off phase (Fig. 1A–C, E, F, H, I).

Post-Hoc SPM analyses indicated differences between the lowest matched filter combination (K10F10) in comparison to all other filter combination within all planes of motion and joints in the early stance phase ($p < 0.003$). In general, we identified the most pronounced differences between the K10F10 and the K10F100 condition (Figs. 2–4).

When performing repeated measures ANOVA on peak joint moment amplitudes in the neutral running shoe condition, we found significant main effects of cut-off frequency combinations for all planes and joints ($p < 0.05$; Tables 1–3). Matched cut-off frequencies lead to lower peak values compared to unmatched conditions. For every joint and in every plane of motion, post hoc analyses indicated significant differences between K10F10 in comparison to all other filter combinations. The largest differences between individual conditions observed in any plane of motion were 94%, 22% and 24% for the hip, knee, and ankle joint, respectively (Figs. 2–4).

We also computed normalized effect sizes (partial eta squared for repeated measures ANOVA and Cohen's d for repeated measures for post hoc comparisons; Tables 1–3; Figs. 2–4). Partial eta squared was medium for all peak values, expect maximal sagittal hip moment ($\eta_p^2 = 0.810$) and ankle sagittal moment ($\eta_p^2 = 0.708$) where it was large. For every analyzed peak moment, effect sizes at the hip were higher compared to knee and ankle joints in the respective plane (Tables 1–3, Figs. 2–4).

We found the largest normalized effect sizes across filter combinations during post hoc testing between K10F10 and K10F100 ($d = 2.54$) for the peak hip sagittal plane moment.

The ranking of participants based on peak joint moments changed for different cut-off frequency combinations (Fig. 5). This effect was most pronounced when the ranking was performed based on peak hip joint moments. Most substantial differences in the ranking were observed for peak sagittal plane hip moment between K10F10 and K10F100 (Spearman $\rho = 0.64$). Spearman rank correlation also indicated differences in ranking for K10F100 and K20F20 ($\rho = 0.70$). The rank order also changed in the comparisons between the two matched cut-off frequency combinations for the hip in the sagittal plane ($\rho = 0.88$). We observed the general trend that the higher the differences in cut-off frequency was, the more different was the ranking.

When considering the different footwear conditions, we identified significant filter frequency by footwear interaction effects ($p < 0.001$; Fig. 6) for all analyzed peak joint moments except for the peak ankle joint moment in the frontal plane ($p = 0.36$; Fig. 6). We computed effect sizes (Cohen's d) for the pairwise comparisons between the different footwear conditions for all peak joint moments. Between filter conditions, the maximum effect size observed between the two most differing footwear conditions varied most substantially for the peak hip moment in the sagittal plane (K20F20 = 0.29; K10F50 = 1.71), but also for the peak knee moment in the frontal plane (K10F100 = 0.23; K10F10 = 0.56). Smaller differences in peak footwear effect sizes were also observed for other joints and planes of motion (Table 4).

4. Discussion

The main purpose of this explorative study was to assess the effect of different filter combinations on the resulting internal joint moments at the lower limb joints when running with a constant, common distance running speed. Since no consistent low-pass filtering strategy exists across studies in the literature, we tried to

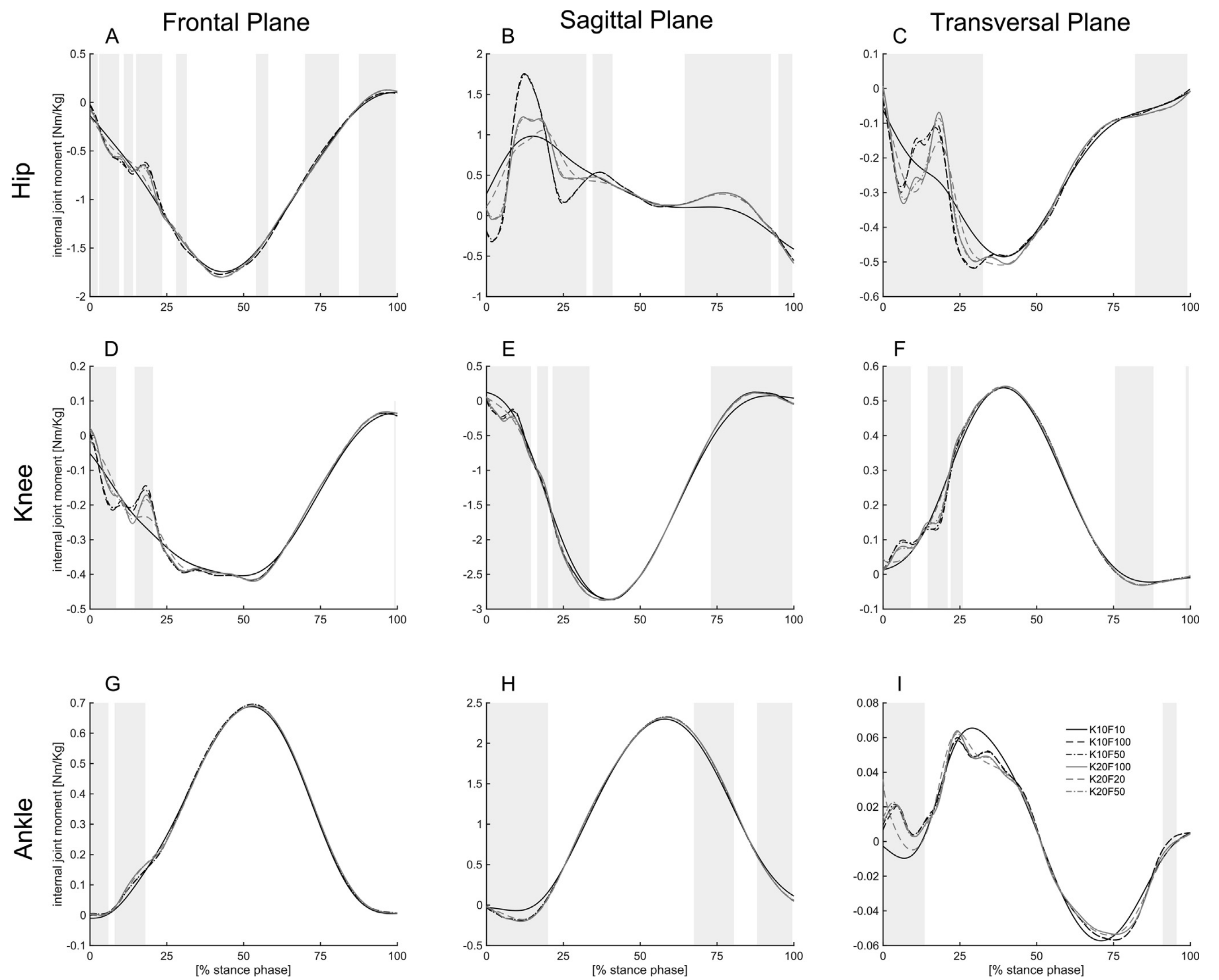

Fig. 1. Mean internal joint moment curves [Nm/Kg] of the hip (A–C), the knee (D–F) and the ankle joint (G–I) during the stance phase of running in the neutral running shoe condition for each filtering cut-off frequency combination (N = 94). Gray areas indicate a significant difference obtained from one-dimensional statistical parametric mapping. Excessive fluctuations for unmatched filtering in sagittal and transversal plane of the hip (B, C) and frontal plane for the knee joint (D).

assess the typical range of filter frequency combinations and analyzed results in all planes of motion. Previous studies already have shown fluctuations of joint moment curves within the initial stance phase in high impact movements like landing, cutting or in simulated running data (Bisseling and Hof, 2006; Edwards et al., 2011; Kristianslund et al., 2012). However, no study so far has quantified this problem systematically for distance running using comprehensive sample size and considering all planes of motion.

Our results show that different cut-off frequency combinations lead to differences in the time course of joint moments in the initial, but also later parts of the stance phase. Our results indicate further that fluctuations of joint moment curves arise due to unmatched cut-off frequencies of marker and force data, while these fluctuations are more pronounced when the differences in cut-off frequencies increase. Therefore, our first hypothesis can be accepted.

Unmatched filter combinations are common in running biomechanics research (Stefanyshyn et al., 2006; Willwacher et al., 2016a, 2014b, 2014a, 2013b, 2013a). Typically, researchers filter marker data with relatively low cut-off frequencies and GRF data with relatively higher frequencies. The problem of this unmatched filter frequency approach is that peak segment accelerations are suppressed (Fig. 7B–D), while the impact peak in the GRF is retained (Fig. 7A). This discrepancy results in an inconsistency in the equations of motion and leads to fluctuations in joint moment curves (Bezodis et al., 2013; Bisseling and Hof, 2006; Kristianslund et al., 2012). Consequently, peak joint moment values in distance running obtained using unmatched filtering need to be interpreted with caution, in particular for more proximal joints.

Furthermore, we found that the ranking of the participants based on joint moments was affected by filter combinations. This effect was also more pronounced for proximal than distal joints. We observed the highest-ranking distortion for the hip in the sagittal plane. Hence, filter procedures need to be considered when analyzing the loads imposed on the joints of distance runners. Further, when comparing different studies which have addressed joint loads or have assessed the effects of technological or running style interventions (Dunn et al., 2018; Willy et al., 2016, 2012), filter procedures need to be considered.

The peak values obtained for unmatched filtering were higher in proximal joints compared to distal joints when considering percent wise differences between the mean values of the different conditions. This pattern can be explained by the cumulative effects

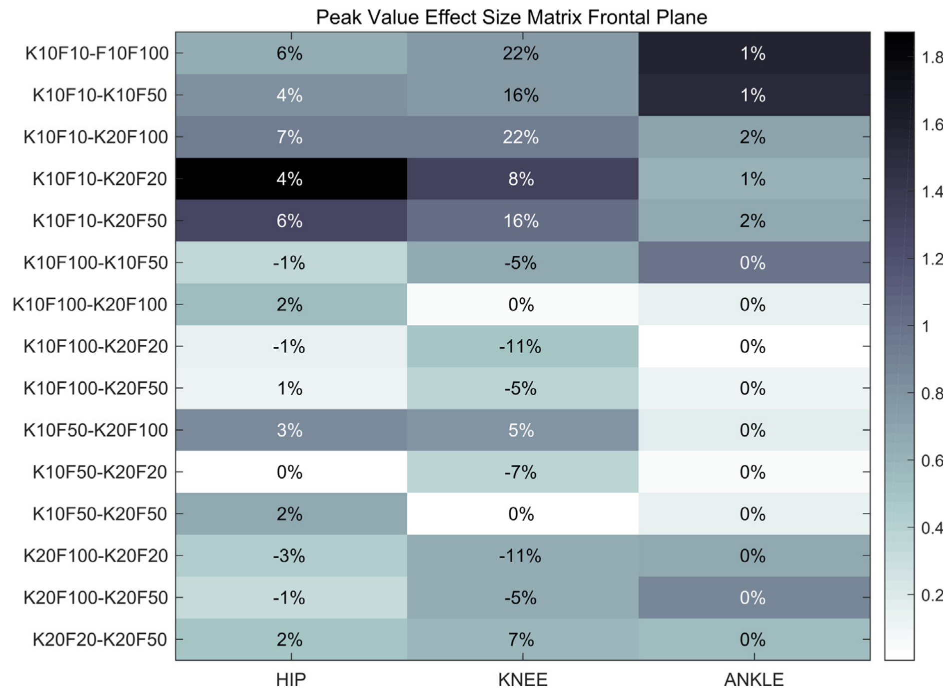

Fig. 2. Effect size matrix for the frontal plane peak moment values of the hip, knee and the ankle in the neutral running shoe condition for each possible filter combination. Effect sizes according Cohen's d are indicated via gray color gradients corresponding to the color bar on the right-hand side. Percentage differences between the means for all filter combinations are displayed in the respective fields. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

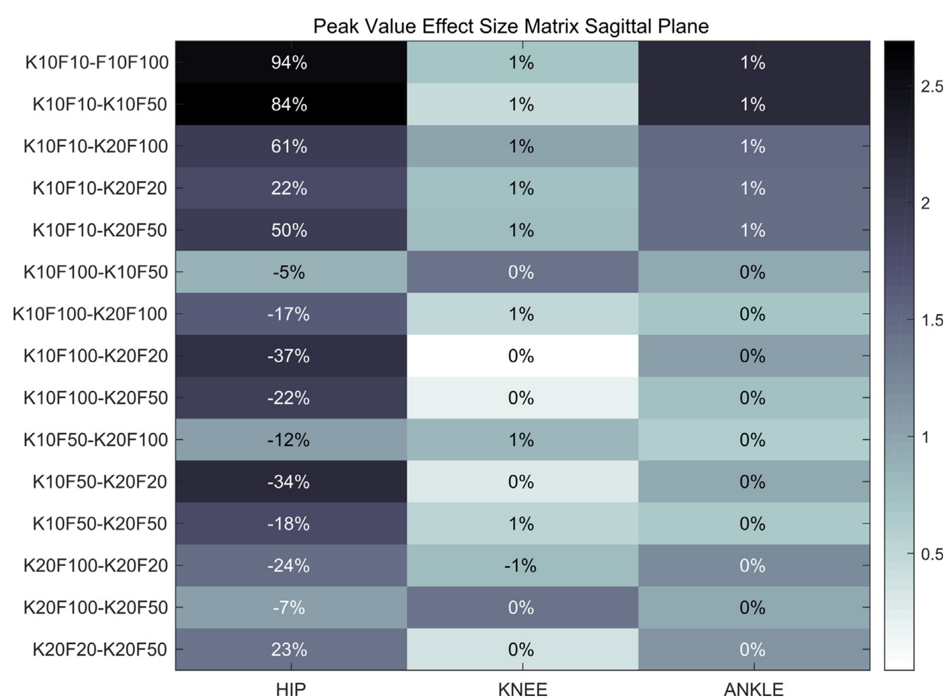

Fig. 3. Effect size matrix for the sagittal plane peak moment values of the hip, knee and the ankle in the neutral running shoe condition for each possible filter combination. Effect sizes according Cohen's d are indicated via gray color gradients corresponding to the color bar on the right-hand side. Percentage differences between the means for all filter combinations are displayed in the respective fields. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

of suppressing more segment accelerations when analyzing more proximal joints of the kinematic chain, which likely leads to an accumulation of artificial fluctuations. Furthermore, impact-related accelerations propagating through the leg reach proximal joints later compared to distal joints. Consequently, we found increased durations of joint moment fluctuations for unmatched filter frequencies from distal to proximal joints (Fig. 1A–I). These results lead us to accept our second hypothesis.

Paradoxically, the percent wise differences between mean values of different cut-off frequency conditions within the neutral

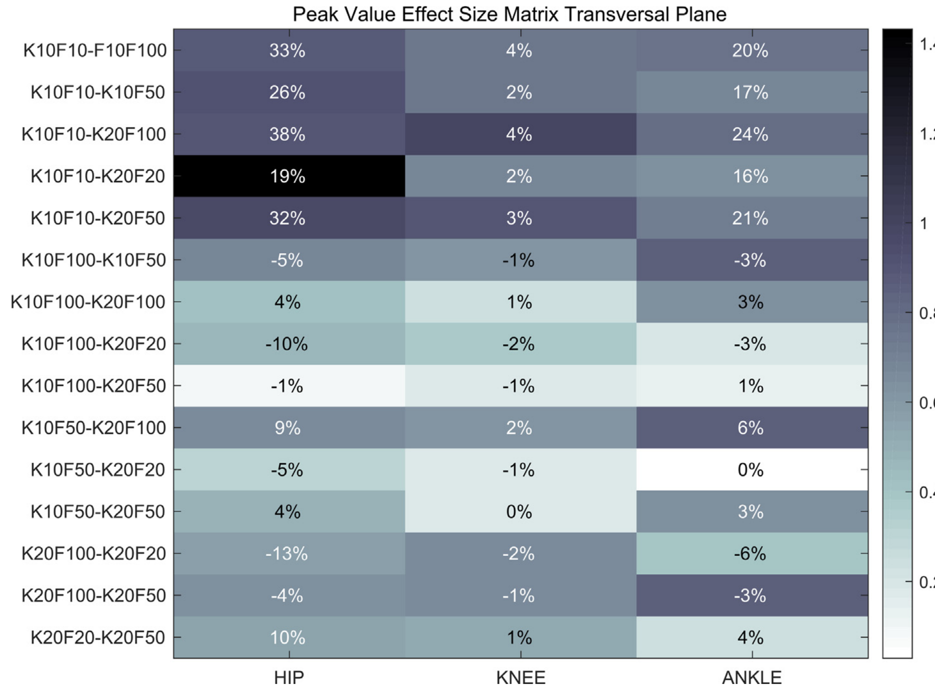

Fig. 4. Effect size matrix for the transversal plane peak moment values of the hip, knee and the ankle in the neutral running shoe condition for each possible filter combination. Effect sizes according Cohen's d are indicated via gray color gradients corresponding to the color bar on the right-hand side. Percentage differences between the means for all filter combinations are displayed in the respective fields. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

Table 1
Mean internal joint moment peak values for the hip, knee and ankle in the frontal plane. Peak moments are normalized to body mass. Standard deviations are reported in brackets. Bonferroni Post- hoc results are marked with superscripted letters for the respective filter combination. * indicates a significant difference to all other conditions obtained during post-hoc testing.

Frontal Plane							
Filter	K10F10^{A}	K10F100^{B}	K10F50^{C}	K20F100^{D}	K20F20^{E}	K20F50^{F}	η_p^2
Hip [Nm/Kg]	-1.756^{*} (0.248)	-1.856^{D} (0.289)	$-1.833^{D, F}$ (0.263)	-1.887^{E} (0.286)	-1.834^{F} (0.263)	-1.867 (0.267)	0.314
Knee [Nm/Kg]	-0.503^{*} (0.336)	$-0.614^{C, E}$ (0.351)	-0.585^{D} (0.338)	-0.612^{*} (0.340)	-0.544^{F} (0.336)	-0.583 (0.332)	0.311
Ankle [Nm/Kg]	0.700^{*} (0.202)	0.710^{C} (0.202)	0.709 (0.202)	0.712^{*} (0.204)	0.710^{F} (0.205)	0.716 (0.205)	0.205

Table 2
Mean internal joint moment peak values for the hip, knee and ankle in the sagittal plane for the neutral running shoe condition. Peak moments are normalized to body mass. Standard deviations are reported in brackets. Bonferroni Post- hoc results are marked with superscripted letters for the respective filter combination. * indicates a significant difference to all other conditions obtained during post-hoc testing.

Sagital Plane							
Filter	K10F10^{A}	K10F100^{B}	K10F50^{C}	K20F100^{D}	K20F20^{E}	K20F50^{F}	η_p^2
Hip [Nm/Kg]	1.041^{*} (0.256)	2.017^{*} (0.537)	1.920^{*} (0.488)	1.681^{*} (0.491)	1.271^{F} (0.330)	1.657 (0.444)	0.810
Knee [Nm/Kg]	-2.885^{*} (0.419)	$-2.910^{C,D}$ (0.423)	$-2.899^{D,F}$ (0.422)	-2.926^{*} (0.415)	-2.910 (0.413)	-2.916 (0.414)	0.291
Ankle [Nm/Kg]	2.312^{*} (0.282)	2.342^{*} (0.285)	2.342^{*} (0.285)	2.337^{*} (0.286)	2.334^{F} (0.285)	2.336 (0.286)	0.708

shoe condition were often in disagreement with the amplitudes of the normalized effect size measures (partial eta squared and Cohen's d, Figs. 2–4, Tables 1–3). Between cut-off frequency conditions, we only varied the details (i.e. cut-off frequencies) of the inverse dynamics analysis method and did not perform repeated measurements for each condition. Consequently, we found very systematic changes with little biological variation between conditions. Therefore, even if the absolute differences in mean values between conditions were small, the variances of these differences were even smaller, leading to partly large normalized effect sizes.

The second purpose of this study was to assess whether different cut-off frequency combinations can change the results and the interpretation of a technological intervention, like e.g. the use of different types of running shoes. We identified significant footwear

Table 3
Mean internal joint moment peak values for the hip, knee and ankle in the transversal plane for the neutral running shoe condition. Peak moments are normalized to body mass. Standard deviations are reported in brackets. Bonferroni Post- hoc results are marked with superscripted letters for the respective filter combination. * indicates a significant difference to all other conditions obtained during post-hoc testing.

Transversal Plane							
Filter	K10F10^{A}	K10F100^{B}	K10F50^{C}	K20F100^{D}	K20F20^{E}	K20F50^{F}	η_p^2
Hip [Nm/Kg]	−0.499* (0.163)	−0.661^{C,D,E} (0.266)	−0.628^{D,F} (0.229)	−0.686* (0.288)	−0.594^{F} (0.178)	−0.656 (0.121)	0.381
Knee [Nm/Kg]	0.544* (0.120)	0.563^{C} (0.115)	0.557^{D} (0.117)	0.567* (0.119)	0.553 (0.123)	0.559 (0.121)	0.274
Ankle [Nm/Kg]	0.099* (0.080)	0.119^{C,D} (0.079)	0.116^{D,F} (0.078)	0.123^{F} (0.079)	0.115 (0.077)	0.120 (0.079)	0.281

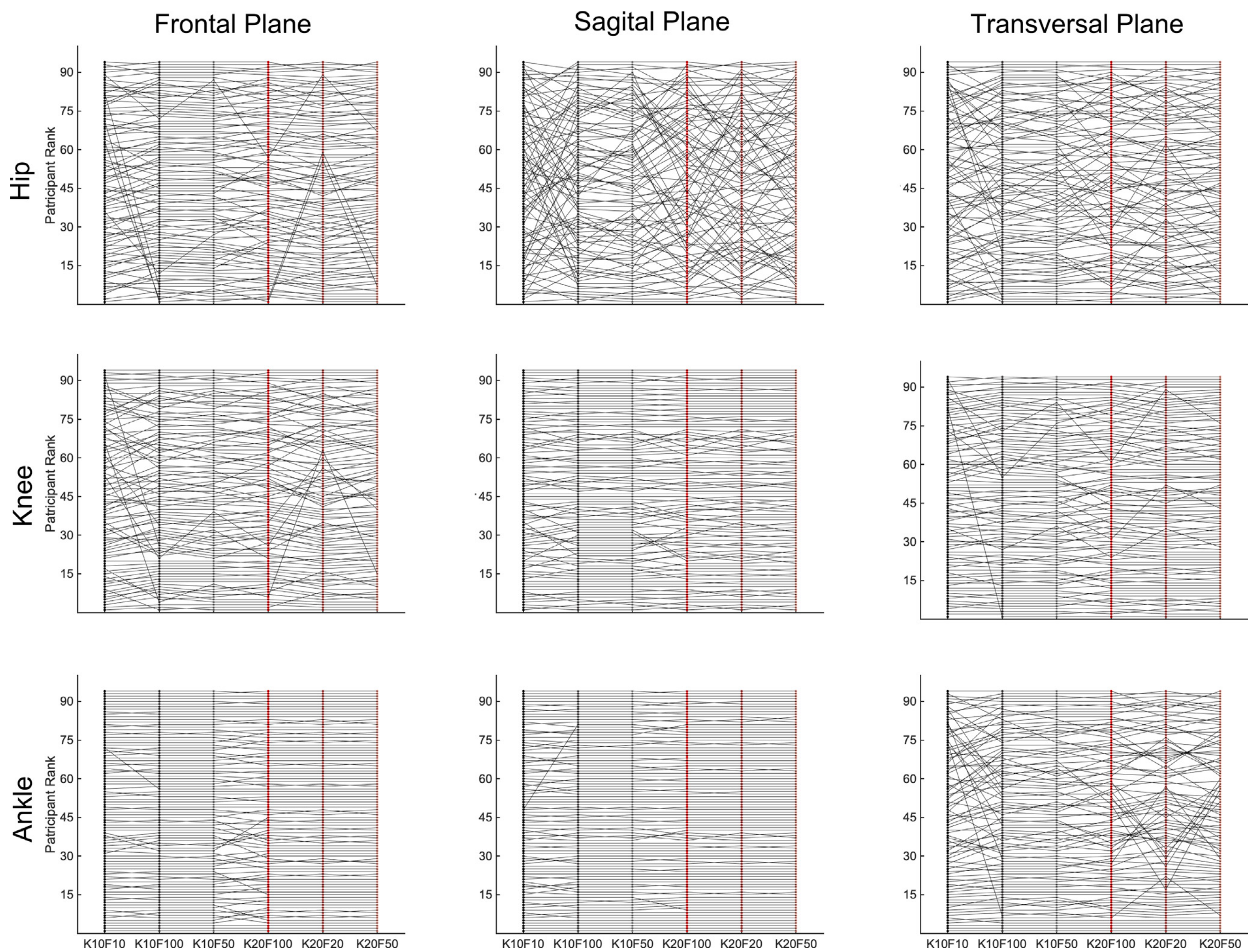

Fig. 5. Ranking of the participants based on peak internal joint moments in the respective plane and joint in the neutral running shoe condition; each data point corresponds to one of the 94 participants in each filter condition. The lines connect the participants in the different filter combinations. Parallel connecting lines indicate no effect of the filter conditions on the ranking of participants. Consequently, more crossing over between lines indicate a stronger effect of filter conditions on the ranking of participants.

by cut-off frequency combination effects for all peak joint moments, except for the peak ankle joint moment in the frontal plane ($p = 0.36$). These results indicate that footwear effects on joint moments can be different when using different filter cut-off frequency strategies during the inverse dynamics calculations. We determined the maximum effect size between the most differing footwear conditions for the different cut-off frequency combinations. We found substantial alterations of these effect sizes, in particular for the hip joint in the sagittal plane and for the knee joint in the frontal plane. Furthermore, the footwear conditions showing the highest or lowest peak joint moments were not always identical between the different cut-off frequency combinations (Table 4). Therefore, our third hypothesis can also be accepted. This finding has important implications for the evaluation of footwear interventions, since the hip joint has an important motor function during distance running (Farris and Raiteri, 2017) and the internal knee abduction moment has been related to knee pain and injuries in runners (Stickley et al., 2018, Stefanyshyn et al., 2006) as well as to the progression of knee osteoarthritis (Andriacchi et al., 2004).

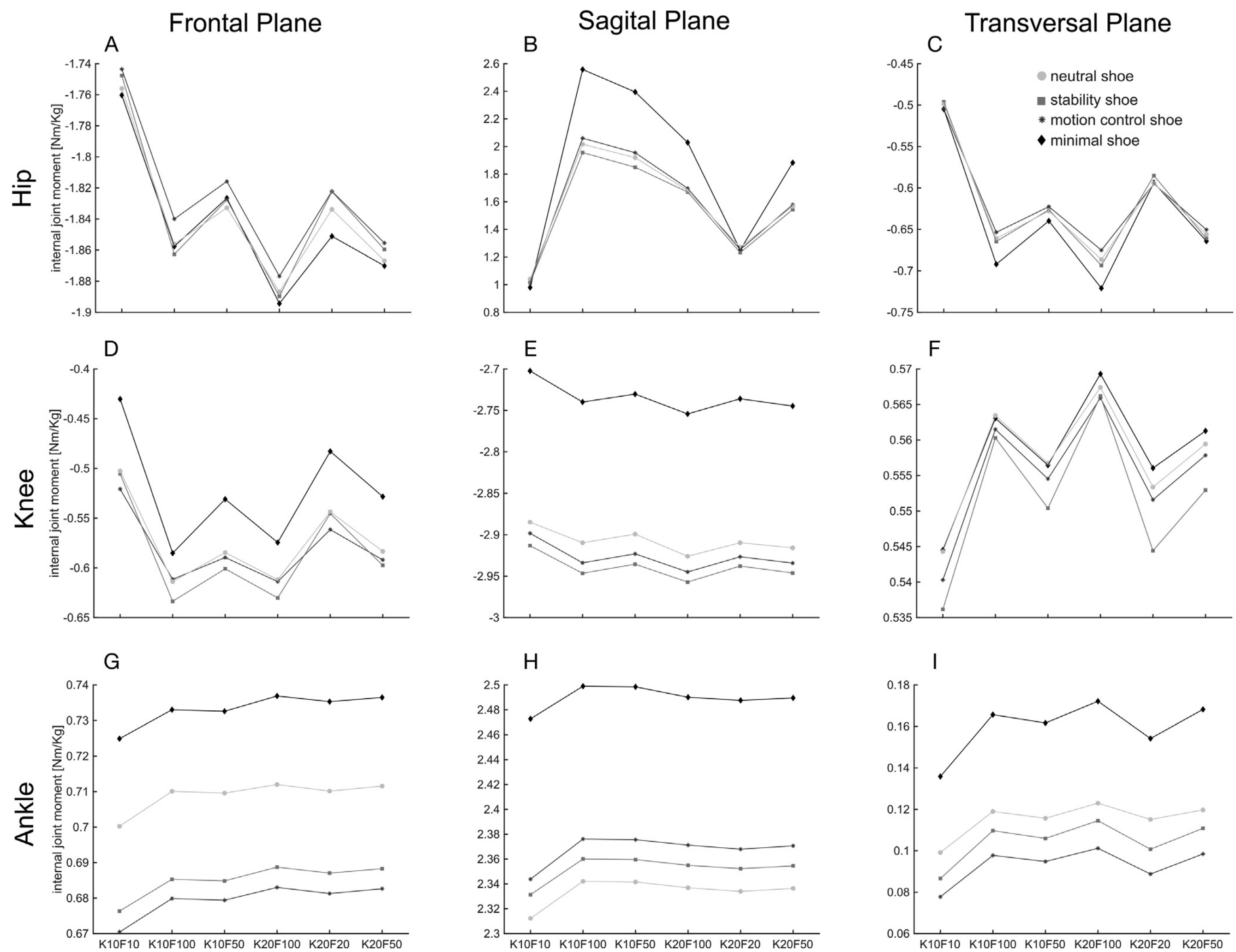

Fig. 6. Mean values of peak joint moments for all cut-off frequency combinations and all footwear conditions.

Table 4
Maximum effect sizes (Cohen's d for repeated measures) observed between the two most different footwear conditions for every joint and every plane of motion between filter conditions. Superscript letter combinations indicate the two most different footwear conditions, for which effects sizes are reported.

Joint	Plane	K10F10	K10F100	K10F50	K20F100	K20F20	K20F50
Hip	Frontal	$0.19^{B\text{-}M}$	$0.22^{M\text{-}S}$	$0.19^{M\text{-}N}$	$0.14^{B\text{-}M}$	$0.30^{B\text{-}M}$	$0.13^{B\text{-}M}$
	Sagittal	$0.56^{B\text{-}N}$	$1.62^{B\text{-}M}$	$1.71^{B\text{-}S}$	$0.99^{B\text{-}S}$	$0.29^{M\text{-}N}$	$1.14^{B\text{-}S}$
	Transversal	$0.22^{M\text{-}S}$	$0.22^{B\text{-}M}$	$0.13^{B\text{-}M}$	$0.25^{B\text{-}M}$	$0.20^{M\text{-}S}$	$0.12^{M\text{-}S}$
Knee	Frontal	$0.56^{B\text{-}M}$	$0.23^{B\text{-}S}$	$0.36^{B\text{-}S}$	$0.27^{B\text{-}S}$	$0.46^{B\text{-}M}$	$0.36^{B\text{-}M}$
	Sagittal	$1.17^{B\text{-}M}$	$1.13^{B\text{-}M}$	$1.12^{B\text{-}M}$	$1.11^{B\text{-}M}$	$1.12^{B\text{-}M}$	$1.10^{B\text{-}M}$
	Transversal	$0.19^{M\text{-}N}$	$0.07^{M\text{-}N}$	$0.14^{M\text{-}N}$	$0.05^{B\text{-}M}$	$0.21^{M\text{-}N}$	$0.15^{M\text{-}N}$
Ankle	Frontal	$0.40^{B\text{-}M}$	$0.38^{B\text{-}M}$	$0.38^{B\text{-}M}$	$0.39^{B\text{-}M}$	$0.39^{B\text{-}M}$	$0.39^{B\text{-}M}$
	Sagittal	$1.10^{B\text{-}N}$	$1.07^{B\text{-}N}$	$1.07^{B\text{-}N}$	$1.04^{B\text{-}N}$	$1.05^{B\text{-}N}$	$1.04^{B\text{-}N}$
	Transversal	$1.20^{B\text{-}M}$	$1.23^{B\text{-}M}$	$1.22^{B\text{-}M}$	$1.24^{B\text{-}M}$	$1.26^{B\text{-}M}$	$1.24^{B\text{-}M}$

^{B,N,M,S} refer to the barefoot (B), neutral (N), motion-control (M), and stability (S) footwear conditions, respectively.

Our findings for distance running are consistent with the results observed for other high impact movements (Bezodis et al., 2013; Bisseling and Hof, 2006; Kristianslund et al., 2012). Matched cut-off frequencies lead to smoother joint moment curves compared to unmatched filtering, which is considered to be more physiological. However, low cut-off frequencies for GRFs also lead to the removal of the existing first peak of the vertical GRF commonly observed in rearfoot strikers. Therefore, either smooth joint moment curves are generated by matched filtering or some physiological information of the GRF signal is removed, which might also lead to altered joint moment curves.

Our study has several limitations. We only included runners who were characterized by an initial peak of the vertical GRF signal during the early ground contact phase. Therefore, our results do not extend to (forefoot) runners that are lacking a clear first peak of the vertical GRF. We only included one running velocity (3.5 m/s) in our analysis. Future studies need to replicate our findings for lower or higher distance running speeds. We only varied filter frequencies for a recursive Butterworth filter with fixed cut-off frequencies for all markers and all force components, respectively. Other digital filter procedures might be less or more prone to the effects introduced by different filtering strategies.

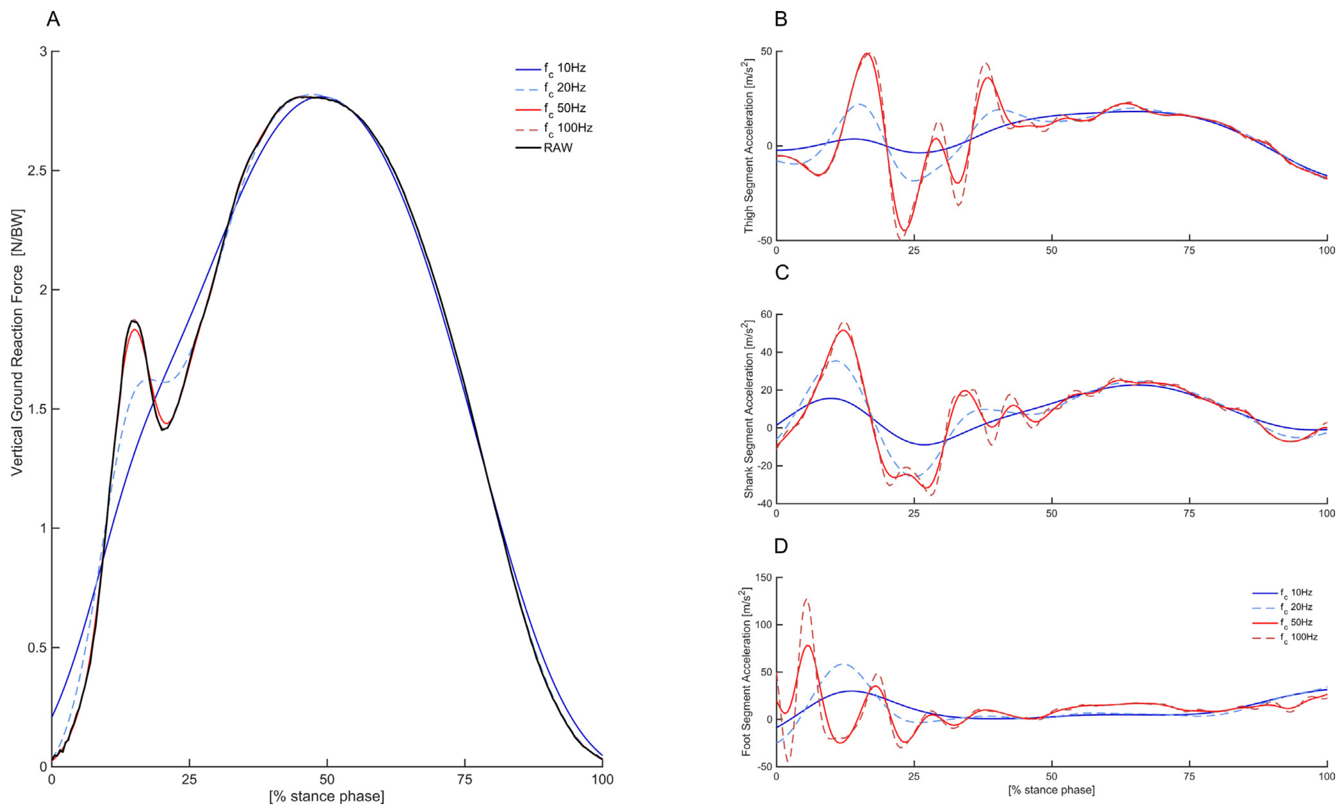

Fig. 7. Vertical GRF [N/BW] (A) and segment accelerations [m/s^2] (B–D) of one trial from a representative participant for different cut-off frequencies. Data are time normalized to stance time.

For example, Edwards et al. have developed an inverse dynamics processing technique in which initially they perform the calculations on raw kinematic and force platform data (Edwards et al., 2011). Subsequently, they filter joint reaction forces and moments based on the frequency content of the distal reaction force. With this approach, they could minimize the attenuation of high frequency components within the joint reaction forces while at the same time avoid artefacts in the joint moment curves.

We used rigid shell marker clusters to track the motion of the shank and thigh segments. Future studies need to replicate our findings for approaches where tracking markers are attached directly to the skin.

In conclusion, our study underlines the importance of cut-off frequency choice when performing inverse dynamics based analyses in distance running. Considering filter strategies is essential to enhance the comparability of studies within the literature and to assess injury risk for runners. Furthermore, different filter cut-off frequency combinations should be considered when evaluating the effects of a footwear intervention on 3D lower extremity joint moments. Ideally, researchers should use sophisticated filtering approaches (like e.g. Edwards et al., 2011) in order to consider high frequency components that are essential to impact movements while avoiding filtering artefacts in the joint moment calculations. However, this method has, to the knowledge of the authors, not been implemented to popular software tools, which are used to perform inverse dynamics calculations. Further, the method is computationally more expensive than the traditional approach to filter kinematic data and GRF data at fixed cut-off frequencies, which might limit its applicability in real time feedback applications with current computer processing power. Therefore, if the traditional approach of filtering data for inverse dynamics needs to be applied, based on our results we recommended the following: Avoid lower cut-off frequencies than 20 Hz for GRFs in order to retain some physiological high frequency information of the first GRF peak, but also use the same or similar cut-off frequencies for marker trajectories and GRFs in order to avoid artificial fluctuations of joint moments during the first half of stance.

Declaration of Competing Interest

No author has any financial and personal relationships with other people or organizations that could inappropriately influence their work.

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