Ground Reaction Forces and Lower Extremity Kinematics When Running With Suppressed Arm Swing

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The role of arm swing in running has been minimally described, and the contributions of arm motion to lower extremity joint kinematics and external force generation are unknown. These contributions may have implications in the design of musculoskeletal models for computer simulations of running, since previous models have usually not included articulating arm segments. 3D stance phase lower extremity joint angles and ground reaction forces (GRFs) were determined for seven subjects running normally, and running under two conditions of arm restraint. When arm swing was suppressed, the peak vertical GRF decreased by 10–13% bodyweight, and the peak lateral GRF increased by 4–6% bodyweight. Changes in peak joint angles on the order of 1–5 deg were observed for hip flexion, hip adduction, knee flexion, knee adduction, and ankle abduction. The effect sizes (ES) were small to moderate ($ES<0.8$) for most of the peak GRF differences, but large ($ES>0.8$) for most of the peak joint angle differences. These changes suggest that suppression of arm swing induces subtle but statistically significant changes in the kinetic and kinematic patterns of running. However, the salient features of the GRFs and the joint angles were present in all conditions, and arm swing did not introduce any major changes in the timing of these data, as indicated by cross correlations. The decision to include arm swing in a computer model will likely need to be made on a case-by-case basis, depending on the design of the study and the accuracy needed to answer the research question.
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1 Introduction

Studies of human locomotion often focus on the lower extremity. Although the arms and the muscles that control them have no direct mechanical linkage to the lower extremity, arm actuation can influence lower extremity joint kinematics and ground reaction forces (GRFs) through the principles of dynamic coupling [1]. This effect can be demonstrated by standing upright on a force platform, swinging the arms, and observing the fluctuations in the GRF (Fig. 1). Previous studies have quantified the effects of arm swing on lower extremity kinematics and external force generation during walking [2–5]. Umberger [5] recently showed that lower extremity joint angles and GRFs during walking were nearly identical with or without arm swing, but the peak magnitude of the free moment at the point of GRF application increased when arm swing was suppressed.

The role of the arms in running has been investigated in only a few studies. Mann [6] reported that shoulder moments during sprinting were small relative to lower extremity joint moments, and concluded that the arms were only involved in maintaining balance. Hinrichs et al. [7] measured the center of mass trajectories and linear momenta of the whole body, the arms, and the lower extremities during the stance phase of running. The arms increased the vertical range of motion of the center of mass, but reduced the anterior-posterior and medial-lateral ranges of motion, and also accounted for 5–10% of the total vertical linear momentum and impulse. Hinrichs [8] also compared the angular momenta of the arms, the lower extremities, and the whole body. In the transverse plane, the angular momenta of the arms and lower extremities were similar in magnitude and opposite in direction, which kept the net transverse angular momentum small and steady, and allowed the legs to propel the body horizontally. These findings indicate that the arms are important in supporting the center of mass and maintaining an anterior direction of progression, contrary to Mann's [6] earlier suggestion.

The question remains as to the extent that arm motion affects lower extremity joint kinematics and GRFs during running. Along with providing insights into the control of human running, an answer to this question could suggest strategies for defining musculoskeletal models in computer simulations, an increasingly popular research strategy. When studying human running, the dynamics of the lower limbs are usually of primary interest, and models of the arms have, in most cases, been greatly simplified or neglected altogether. In muscle-actuated forward dynamics computer simulations of running, the arms are often combined with the head and trunk into a single segment representing the net dynamics of the upper body, with no explicit consideration of the shoulder and elbow musculature [9–11]. In order to understand if models that lack explicit consideration of arm actuation are suitable for accurate simulations, it is important to gain a greater understanding of the role of the arms in running, and the influence of arm swing (or a lack of arm swing) on the lower extremity kinematics and GRF variables studied during running.

Therefore, the purpose of the present study was to determine the effects of suppressing arm swing on GRFs and lower extremity kinematics during running. We hypothesized that the peaks of the medial-lateral GRF and the free moment would increase when arm swing was suppressed, due to loss of angular impulse control provided by arm swing in the transverse plane [8,5]. We also expected the peak vertical GRF to decrease, and lower extremity joint flexion peaks to increase, due to the reduction of the vertical impulse normally provided by arm swing [7].

2 Methods

2.1 Subjects. A power analysis ($\alpha=0.05$, $\beta=0.80$) of pilot data suggested that at least five subjects were needed for a 10% decrease in the vertical GRF peak with arm swing suppression to be statistically significant. Seven subjects (four males, three females; mean ± SD: age = 28 ± 3 years, height = 177 ± 6 cm, mass = 73 ± 14 kg) participated. Protocols were approved by the local institutional review board. Subjects gave informed written consent prior to participating.

2.2 Experimental Setup. Coordinates of retroreflective markers were sampled at 500 Hz using an eight-camera optical motion capture system (Oqus 300, Qualisys, Gothenburg, Sweden). GRFs were sampled synchronously at 5000 Hz using a strain gauge force platform (model OR6–5, AMTI, Watertown,

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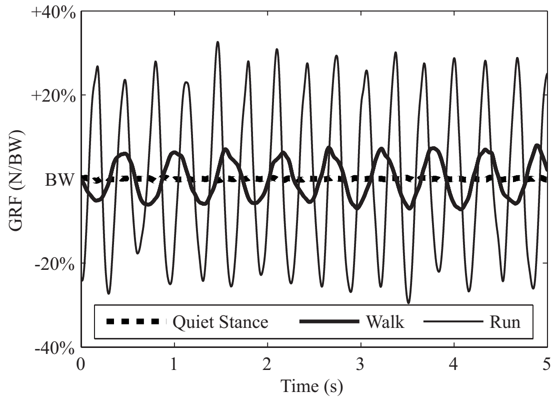

Fig. 1 Magnitude of GRF relative to bodyweight (BW, 756 N) during two-legged upright stance while standing quietly (dashed line), swinging the arms in a "walking" style (elbows straight, 1 Hz swing; thick line), and swinging the arms in a "running" style (elbows bent, 2 Hz swing; thin line)

MA, USA) situated at the center of a 20-m runway. The running speed was determined using two photoelectric timing devices on either side of the force platform.

2.3 Protocol. Plates of four retroreflective markers (15 mm in diameter) were attached to the thigh and calf using velcro wraps and athletic tape. A triad of markers was taped to the heel of the shoe. Individual markers were taped to the left and right anterior-superior iliac spines and the L5/S1 lumbosacral joint. Subjects performed a standing calibration trial, then ran in their own shoes along the runway so that their right foot fully contacted the force platform. Subjects performed three conditions with five trials each (Fig. 2):

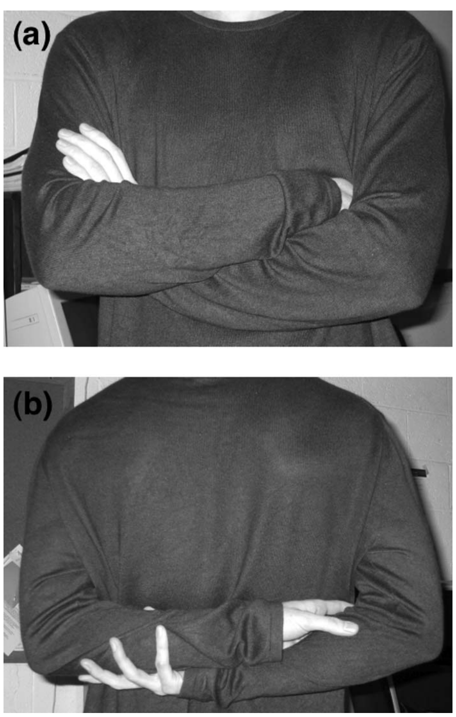

Fig. 2 Photograph of a subject demonstrating the restrained arm conditions: (*a*) arms restrained across the chest (condition RC), and (*b*) arms restrained behind the back (condition RB)

- Condition N: running normally, with the arms unrestrained.
- Condition RC (Fig. 2(*a*)): running with the arms held across the chest.
- Condition RB (Fig. 2(*b*)): running with the arms held behind the back.

All trials were performed at the subject's self-reported 5-km training pace (mean: 4.0 m/s; range: 3.1–4.5 m/s). Condition N was always performed first, with the order of conditions RC and RB subsequently randomized. During conditions RC and RB, subjects were instructed to relax their arm muscles as much as possible, and were allowed numerous practice trials to become comfortable with these conditions before data were collected. The pilot testing demonstrated no differences between running with voluntary arm restraint versus running with a physical restraint (velcro strap). The two different arm swing suppression conditions (RC and RB) were performed to determine if differences relative to the normal condition were due to the suppression of arm swing, which was common to both conditions, or to the relocation of the center of mass, which differed between conditions RC and RB.

2.4 Data Analysis. Marker coordinates were filtered digitally using a fourth-order lowpass recursive Butterworth filter. Cutoff frequencies (11–14 Hz) were selected by residual analysis [12]. A four-segment kinematic model (foot, calf, thigh, pelvis) was defined from the marker coordinates. Stance phases were isolated using the vertical GRF component (threshold=15 N). Three-dimensional lower extremity joint angles were calculated with a Cardan rotation sequence of flexion/extension-adduction/abduction-internal/external rotation [13]. Stance phase data were interpolated to 101 points, and averaged across trials for each subject. GRFs were scaled by bodyweight. Free moments acting about a vertical axis through the center of pressure of the GRF were calculated by subtracting the transverse ground reaction shear moments due to the medial-lateral and anterior posterior GRF from the total transverse ground reaction moment about the vertical axis at the force platform origin [14]. Free moments were scaled by the product of bodyweight and height, then amplified by 1000 for presentation.

Outcome variables (peak scaled GRF magnitudes, peak scaled free moment magnitudes, peak joint angles) were compared between conditions by repeated measures of ANOVA with post-hoc matched-pair t-tests and a false discovery rate adjustment for multiple comparisons [15], resulting in a critical $\alpha=0.0167$. To further assess differences between conditions, effect sizes (ES) were calculated and classified as "small" ($\text{ES}<0.2$), "moderate" ($0.4<\text{ES}<0.6$), or "large" ($\text{ES}>0.8$), based on suggestions by Cohen [16]. ES between 0.2 and 0.4 were deemed "small-to-moderate," and ES between 0.6 and 0.8 were deemed "moderate-to-large." The temporal similarity of the time series between conditions was assessed with zero-lag cross correlations. Cross correlations were z-transformed prior to averaging over subjects.

Finally, the time series from each condition were averaged across all trials of all subjects, and the within-subject (*i.e.* between-trials) standard deviations (SDs) of condition N were calculated as the SD between all trials for one subject averaged across all subjects. Conditions RC and RB were deemed similar to condition N if their time series fell within two SDs of the mean of condition N. This criterion is similar to the validation criterion used with musculoskeletal models that track experimental data from human subjects [10,17].

3 Results

3.1 GRFs. The three GRF components for each condition are shown in Fig. 3, with peak magnitudes compared between conditions in Table 1. Compared with the normal condition, when the arms were restrained across the chest (RC), the lateral GRF peak increased by 6.3% bodyweight (BW) ($p=0.005$), and the second vertical GRF peak decreased by 12.8% BW ($p=0.0001$). When

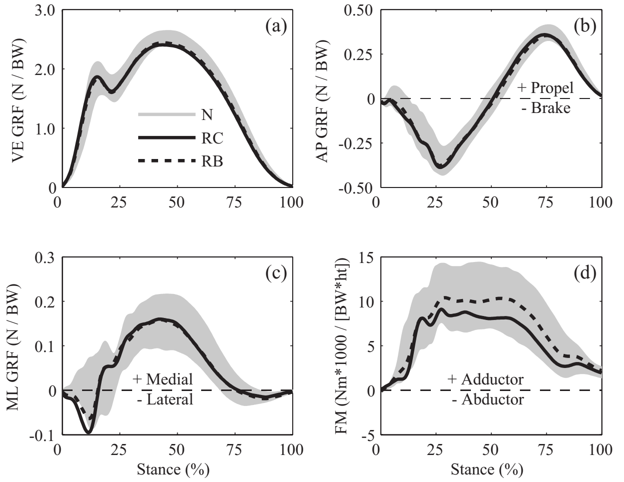

Fig. 3 (*a*) Vertical, (*b*) anterior-posterior, and (*c*) medial-lateral GRF components, and (*d*) the vertical free moment during normal running (shaded), running with the arms held across the chest (RC, solid line), and running with the arms held behind the back (RB, dashed line). Shaded areas are ±2 within-subject standard deviations for condition N.

the arms were restrained behind the back (RB), the second vertical GRF peak decreased by 9.7% BW ($p=0.002$). The only large effect size was for the difference in lateral GRF peaks between conditions N and RC (ES=1.02).

3.2 Free Moment. All seven subjects exhibited adductor-biased free moments, regardless of the condition, based on the sign convention of Holden and Cavanagh [14]. Figure 3(*d*) shows the free moments, and Table 1 shows the differences in peak free moments between conditions. With the arms restrained across the chest (RC), the scaled free moment peak decreased, although this change was not statistically significant ($p=0.04$), and effect sizes were small to moderate.

3.3 Joint Angles. Lower extremity joint angles for each condition are shown in Fig. 4. Changes in peak joint angles between conditions are shown in Table 1. Compared with the normal condition, in condition RC, peak hip flexion increased by 2.2 deg ($p=0.008$), peak hip adduction decreased by 1.7 deg ($p=0.0013$), peak knee flexion increased by 4.0 deg ($p=0.003$), peak knee adduction decreased by 2.6 deg ($p=0.009$), and peak ankle abduction increased by 2.6 deg ($p=0.015$). The same joint angle peaks were significantly different between conditions N and RB, when the arms were restrained behind the back (Table 1). Of the 18 joint angle peaks compared with condition N, six had large effect sizes: peak hip adduction, knee flexion, and knee adduction for condition RC, and peak hip internal rotation, knee flexion, and knee adduction for condition RB.

3.4 Timing. All cross correlations were greater than 0.93 (Table 2), suggesting that all data patterns were temporally similar between conditions when scaled to 0–100% of stance. However, the duration of stance increased slightly (+12 ms on average) when arm swing was suppressed.

3.5 Variability. Most GRF components from conditions RC and RB were within two between-trial SDs of condition N throughout stance (Fig. 3). The vertical and lateral GRF peaks for condition RC fell slightly outside this range by less than 5% and 1% BW, respectively. When arm swing was suppressed, the hip ab/adduction, hip internal/external rotation, knee flexion/extension, knee ad/abduction, knee internal/external rotation, and ankle dorsi/plantarflexion angles all fell outside two SDs of condition N at particular times during stance (Fig. 4).

3.6 Arm Swing Suppression Method. When comparing conditions RC and RB, there were no significant differences in any of the GRFs or joint angle peaks (Table 1), with the exception of the lateral GRF peak, which was 2.6% BW greater when the arms were restrained across the chest ($p=0.006$). Of the 16 comparisons of peak magnitudes between conditions RC and RB, only the peak hip internal rotation angle (ES=0.64) had an effect size greater than moderate.

Table 1 Changes in magnitudes of GRF peaks, free moment peaks, and lower extremity joint angle peaks between normal running (N), running with the arms held across the chest (RC), and running with the arms held behind the back (RB)

	RC-N	ES	RB-N	ES	RC-RB	ES
			GRF peak			
ML lateral	6.3(4.5)*	**1.02**	3.7(4.2)	0.74	2.6(2.2)*	0.35
ML medial	0.0(2.0)	0.10	0.0(1.9)	0.12	0.0(2.0)	0.20
AP braking	3.5(5.3)	0.64	2.3(3.4)	0.47	1.2(3.4)	0.27
AP propelling	−0.8(1.4)	0.17	−1.4(1.2)	0.29	0.6(1.4)	0.13
VE first	5.6(13.2)	0.35	1.6(6.0)	0.14	4.3(11.7)	0.19
VE second	−12.8(4.4)*	0.60	−9.7(5.8)*	0.46	−3.1(2.9)	0.19
Free moment	−1.8(2.3)	0.44	−0.7(1.5)	0.19	−1.1(1.6)	0.27
			Joint angle peak			
Hip flexion	2.2(1.7)*	0.56	3.5(5.3)*	0.78	−1.3(1.9)	0.30
Hip adduction	−1.7(1.1)*	**0.81**	−1.9(1.5)*	0.69	0.2(0.8)	0.00
Hip internal rotation	1.1(1.5)	0.18	5.1(4.4)	**0.81**	−4.0(3.5)	0.64
Knee flexion	4.0(1.7)*	**0.89**	3.6(1.8)*	**0.80**	0.3(1.4)	0.00
Knee adduction	−2.6(1.7)*	**0.99**	−2.6(1.7)*	**1.03**	−0.1(0.8)	0.01
Knee internal rotation	−2.7(2.3)	0.43	−3.8(4.6)	0.51	1.1(2.5)	0.16
Ankle dorsiflexion	2.1(1.9)	0.43	2.2(1.7)	0.48	−0.2(1.1)	0.09
Ankle eversion	0.7(1.7)	0.23	0.7(1.4)	0.27	−0.3(0.7)	0.04
Ankle abduction	2.6(2.1)*	0.31	2.4(1.3)*	0.36	0.2(1.3)	0.06

ES refer to the comparison left of the ES value. GRFs are in percent bodyweight. Free moments are scaled by bodyweight and height, then multiplied by 1000. Angles are in deg. ML=medial-lateral, AP=anterior-posterior, and VE=vertical. Data are mean (between-subjects SD) for seven subjects. $^*=p<0.0167$. "Large" effect sizes (ES≥0.80) are in boldface.

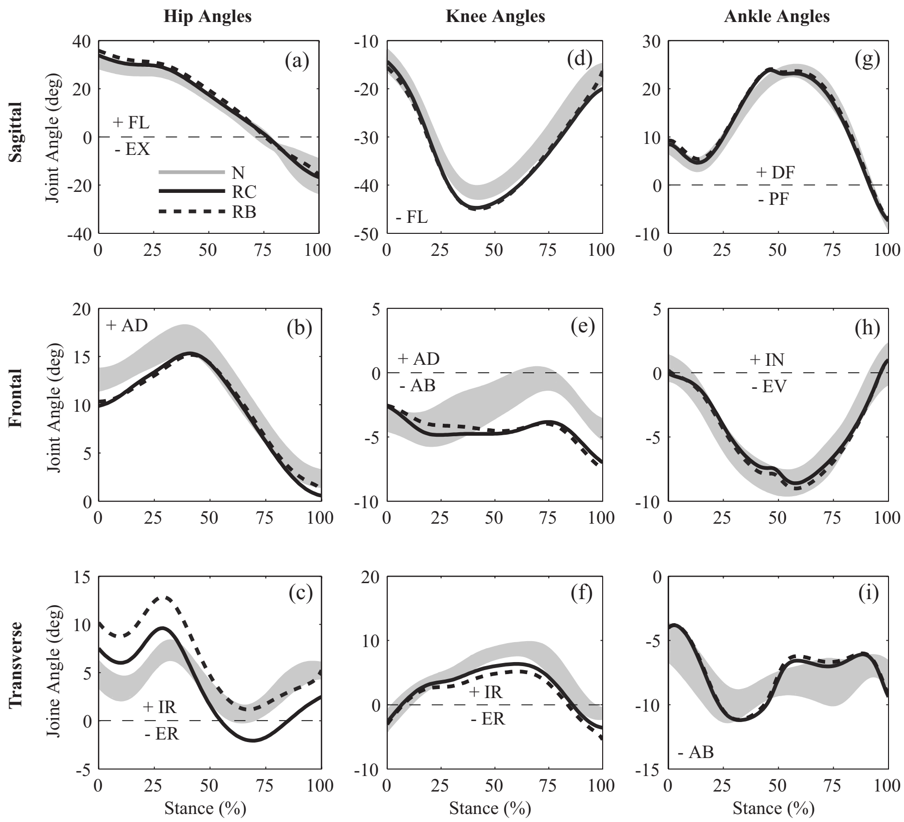

Fig. 4 (*a–c*) Hip, (*d–f*) knee, and (*g–i*) ankle joint angles during normal running (shaded), running with the arms held across the chest (RC, solid line), and running with the arms held behind the back (RB, dashed line). Shaded areas are ±2 within-subject standard deviations for condition N. FL/EX=flexion/extension; AD/AB=adduction/abduction; IR/ER=internal/external rotation; DF/PF=dorsiflexion/plantarflexion; IN/EV =inversion/eversion.

4 Discussion

The purpose of the study was to quantify the effects of suppressing arm swing on GRFs and lower extremity joint angles during the stance phase of running. In support of our hypotheses, the peak vertical GRF decreased, and the peak hip and knee flexion angles increased when arm swing was suppressed. The peak ankle dorsiflexion angle also increased as hypothesized, although this change was not statistically significant. In the frontal plane, the peak lateral GRF and the hip and knee adduction angles were also altered when arm swing was suppressed (Table 1). Arm swing suppression did not appear to alter the general timing of GRFs and lower extremity joint angles, as shown by zero-lag cross correlations, although the duration of stance increased when arm swing was suppressed. These findings were consistent for all seven subjects studied, although most effect sizes were not large, suggesting that the biological significance of these differences (identified by effect sizes) was small to moderate at best. Peak hip and knee joint angles in the transverse plane were not significantly affected by arm swing, although by visual inspection, these angles appear to have differed dramatically when arm swing was suppressed (Fig. 4). The lack of significance was due to the greater variance in these data, compared with the sagittal plane joint angles and the GRFs. These subtle rotations are difficult to measure accurately with skin-mounted markers [18], and it is possible that the presented data do not accurately reflect the underlying skeletal motions.

Table 2 Zero-lag cross correlations between normal running (N), running with the arms held across the chest (RC), and running with the arms held behind the back (RB). Data are averages for seven subjects.

Time series	RC versus N	RB versus N	RC versus RB
Medial-lateral GRF	0.95	0.94	0.97
Anterior-posterior GRF	0.98	0.98	0.98
Vertical GRF	0.99	0.98	0.99
Free moment	0.94	0.93	0.97
Hip flexion/extension	0.98	0.98	0.99
Hip adduction/abduction	0.98	0.97	0.98
Hip internal/external rotation	0.94	0.95	0.95
Knee flexion/extension	0.97	0.96	0.97
Knee adduction/abduction	0.96	0.95	0.97
Knee internal/external rotation	0.96	0.96	0.98
Ankle dorsi/plantarflexion	0.95	0.95	0.96
Ankle inversion/eversion	0.98	0.98	0.99
Ankle adduction/abduction	0.93	0.93	0.99

Regarding the control of the lower extremity, center of mass trajectory, and external force generation, the results indicate that arm swing suppression alters the mechanism of the center of mass support (the vertical GRF and sagittal joint flexion) and the control of the frontal plane motion (the medial-lateral GRF and hip and knee joint adduction/abduction) during stance. Hinrichs et al. [7] suggested that arm swing accounts for 5–10% of the vertical impulse during running. In the present study, the increase in stance duration when arm swing was suppressed may be reconciled as an adjustment made by the subjects to maintain the vertical impulse when the peak vertical GRF decreased with arm swing suppression.

The suppression of arm swing induced the same direction of change in all significant differences, regardless of whether the arms were restrained across the chest or behind the back (Table 1; Figs. 1 and 2). This similarity suggests that the differences between the normal and suppressed arm swing conditions were, in fact, due to the loss of arm swing, which was common to both suppression conditions, and not due to the relocation of the center of mass, which differed between conditions.

Contrary to our hypothesis, free moments were not statistically different between conditions. The peak free moment actually decreased slightly when arm swing was suppressed, although this change was not statistically significant. This finding contrasts previous research on arm swing and walking, which has demonstrated a clear increase in free moment magnitude when arm swing is suppressed [3,5]. It is possible that during running, trunk rotation (which was not restrained), rather than arm swing, plays the dominant role in counterbalancing the transverse angular momentum of the legs, whereas the converse may be true in walking. Since trunk kinematics and segment angular momenta were not reported here, this suggestion is made cautiously.

These results have direct implications for the design of musculoskeletal models in computer simulations of running. Most simulations of running have not included the arms as separate segments [9,19,10,11]. An exception is the swing phase sprinting model of Thelen et al. [20] and Chumanov et al. [21], who defined elbow and shoulder joints that moved according to prescribed kinematic measurements. Human movement simulations are increasingly formulated on a subject-specific basis, where a model tracks a template of experimental data from a specific individual. In these simulations, the model is often deemed valid if the simulated kinematics and kinetics fall within two between-trial (within-subject) standard deviations of the experimental means (e.g., Refs. [10,17]). In the present study, the suppression of arm swing caused seven of the nine kinematic variables to fall outside two between-trial standard deviations of the mean from normal running for at least part of the stance phase (Fig. 4). The peak vertical and lateral GRFs were also slightly outside this range when arm swing was suppressed (Fig. 3). Based on the two-standard-deviations criterion, these results indicate that GRFs and lower extremity joint motions during running are affected by the suppression of arm swing, and suggest that the mechanisms underlying the generation of these dynamics are affected when running with restrained arm swing. If we assume that humans running with suppressed arm swing approximate the capabilities of a model that lacks separate arm segments, these results suggest that researchers should carefully consider whether the effects of arm swing need to be directly accounted for when modeling running.

The decision on whether these differences are great enough to warrant the inclusion of articulating arm segments in a computer model will likely need to be made on a case-by-case basis. For bottom-up inverse dynamics models, the arms will, in most cases, be unnecessary, since their influence will be contained implicitly in the experimental data. For forward dynamics models, additional segments would increase the model's degrees of freedom and control variables, the required simulation time, and the difficulty in locating an optimal solution to a dynamic optimization problem. However, with continual increases in affordable computational power and efficient simulation algorithms, these concerns are diminishing. Rather than including explicit arm segments and shoulder/elbow muscles, one option would be to apply forces and moments representing arm swing dynamics at virtual "shoulder" joints on the model's trunk segment. Alternatively, data could be collected from humans running with suppressed arm swing, and simulated using a model with no arm swing, although the results of these simulations may not be representative of normal human running.

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