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Short communication

Effects of suppressing arm swing on kinematics, kinetics, and energetics of human walking

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Abstract

Human walking is characterized by pronounced arm movement, yet computer simulation models of walking usually lump the mass of the arms with the head and torso. The implications of this simplification have not been thoroughly documented in the literature. Thus, the purpose of this study was to establish the dependence of several biomechanical and energetic variables on suppressing arm swing (AS) in walking. Eight healthy adult subjects walked with and without normal AS, with speed and stride frequency/length matched between trials. Metabolic data were collected during walking on a treadmill, while kinematic and kinetic data were collected during overground walking. Gross and net energy expenditure were significantly higher during walking without AS, with the mean differences being less than 10%. Joint angles, angular velocities, and ground reaction forces were nearly identical for walking with and without AS. Most joint moments and powers were also similar between AS conditions; however, some kinetic variables (e.g., knee joint power) exhibited larger differences, primarily during the stance phase. The variable that differed most between walking with and without AS was the free vertical moment between the foot and ground. In summary, most variables differed by less than 10% and were highly correlated ($r \geqslant 0.90$) between walking with and without normal AS. Thus, researchers may be justified in using walking models without articulated arms. However, a few variables exhibited larger differences, which might be of relevance based on the specific research question being addressed.

Keywords: Gait; Arm swing; Biomechanics; Locomotion; Metabolic cost

1. Introduction

During human walking, the arms normally swing in opposition to the legs, which helps balance angular momentum generated in the lower body (Elftman, 1939; Hinrichs, 1990). Despite this overt motion of the arms, computer simulation models of walking typically have the mass of the arms lumped into a single segment, along with the head and some or all of the trunk (e.g., Anderson and Pandy, 2001; Neptune et al., 2001; Umberger et al., 2003). This simplification improves computational efficiency, but it is appropriate only if arm swing (AS) has a relatively minor impact on the biomechanics and energetics of walking. The existing literature on restraining AS in walking is either narrow in focus (Eke-Okoro et al., 1997; Jackson et al., 1983; Li et al., 2001) or exists only in the form of technical reports and published abstracts (Chapman and Ralston, 1964; Park et al., 2000), making the overall impact of AS difficult to evaluate. Therefore, the purpose of this study was to determine the effects of suppressing AS on several common gait variables.

2. Methods

Five male and three female subjects (M ± SD: age 27.3 ± 3.8 yr; mass 65.7 ± 10.2 kg; height 1.73 ± 0.10 m; leg length 0.90 ± 0.05 m) completed this study after providing informed consent in accordance with local regulations. All testing was completed at a speed of 1.3 m/s as subjects walked with and without AS, using the preferred stride frequency (SF) and stride length (SL) determined during normal walking. Preferred SF (54.4 ± 3.2 stride/min) was computed from the time required to complete 50 strides, and preferred SL (1.43 ± 0.08 m) was determined by dividing

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speed by preferred SF (in Hz). Metabolic data were collected on a motorized treadmill, while kinematic and kinetic data were collected during overground walking. For trials, where AS was suppressed, subjects folded their arms across their chest, with the forearms interlocked such that each hand was supported on the flexed elbow of the contralateral arm. Subjects were instructed to let their folded arms hang in a relaxed manner, and to avoid "tensing" their shoulder or arm muscles. The SL and SF used to walk at any particular speed may be affected by restraining AS (Eke-Okoro et al., 1997), thus they were matched between conditions in this study. During trials on the treadmill, SF was matched to a metronome, and for overground trials, subjects matched their SL to marks on the floor (Laurent and Pailhous, 1986). For overground walking, speed was monitored using a timing device, and the trial in which each subject came closest to the target speed (typical error 1–2%), while also hitting the marks on the floor, was selected for analysis.

2.1. Metabolic data

Rates of oxygen consumption and carbon dioxide production were recorded using a metabolic measurement system (TrueMax 2400, Parvo Medics, Sandy, UT, USA). Baseline metabolic values were quantified while subjects stood quietly, following 15 min of seated rest. Gross and net (gross−standing) rates of metabolic energy expenditure were estimated from pulmonary gas exchange (Weir, 1949), and were normalized to body mass.

2.2. Kinematic and kinetic data

The locations of reflective makers placed over anatomical joint centers were recorded using a S-VHS video camera (60 Hz, Panasonic, Secaucus, NJ) while subjects walked along a 12 m walkway containing an embedded force platform (600 Hz, AMTI, Watertown, MA). Coordinates of the reflective markers were obtained using Peak Motus software (Vicon-Peak, Centennial, CO) and were smoothed using a Butterworth digital filter (Winter, 1990), with cutoff frequencies (3–6 Hz) determined separately for each marker (Jackson, 1979). The force plate and kinematic data were combined with estimates of body segment inertial parameters to compute sagittal plane joint moments and powers for the hip, knee, and ankle, using an inverse dynamics approach (Winter, 1990). The free vertical moment acting on the foot was also computed from the force platform data (Holden and Cavanagh, 1991). Ground reaction forces were scaled to body weight, and moments were scaled to body weight and leg length (Hof, 1996).

2.3. Analysis

The discrete metabolic variables were evaluated for differences using paired t-tests (two-tailed $\alpha = 0.05$). The continuous kinematic and kinetic data series were evaluated for similarity in shape and differences in magnitude using the coefficient of cross-correlation (r) and root-mean-square difference (RMSD), respectively. The r and RMSD values were computed separately for each variable for each subject, and then averaged across subject.

3. Results

The gross and net rates of metabolic energy expenditure were 5.0% and 7.7% higher ($p = 0.004$) for walking without AS than for walking with AS, respectively (Table 1). Overall, the kinematic (Fig. 1) and kinetic (Figs. 2 and 3) results were similar for walking with and without AS. The variables that exhibited the highest RMSD and lowest r values were the free vertical moment (Fig. 2C), knee joint moment (Fig. 3B), knee joint power (Fig. 3E), and hip joint power (Fig. 3D).

Table 1
Gross and net rates of metabolic energy expenditure (W/kg) during walking

	Arm swing	No arm swing	t	p
Gross	4.35 ± 0.26	4.57 ± 0.29	4.42	0.004
Net	2.85 ± 0.20	3.07 ± 0.24	4.42	0.004

Values are mean and standard deviation ($n = 8$).

4. Discussion

Overall, the kinematics and kinetics of walking without AS were found to be similar to walking with normal AS, while metabolic energy expenditure was significantly higher (5–8%) for walking without AS than during normal walking. Chapman and Ralston (1964) and Park et al. (2000) reported no significant differences in gross energy expenditure between AS conditions at speeds in the range of 1.1–1.2 m/s; however, Park et al. found energy expenditure to be 6% higher for walking without AS at 1.7 m/s. A simple linear interpolation of their results to the speed used in the current study (1.3 m/s) suggests an increase in gross energy expenditure of about 2% for walking without AS, which was still lower than the 5% increase in the current study. The reason that energy cost was greater for walking without AS is unknown, but is consistent with earlier reports that AS decreases the vertical excursion of the whole body center of mass (Hinrichs, 1990; Murray et al., 1967). Chapman and Ralston (1964) also found that restraining torso rotation increased gross energy cost by about 10%, independent of AS, suggesting that these two effects might be additive.

Most of the kinematic and kinetic variables exhibited mean differences between walking with and without AS that were 10% or less, with high correlations that mostly exceeded $r = 0.90$. The sagittal plane variable with the largest mean difference was the knee joint moment (Fig. 3B), with the difference confined largely to the stance phase. The larger knee extension moment and smaller knee flexion moment during walking without AS may be due to the subjects maintaining slightly more knee flexion during stance (Fig. 1B). The present kinematic results are consistent with Park et al. (2000), who found no significant effect of AS, expect in certain pelvis and hip variables at their fastest walking speed (1.7 m/s). Kinetic data comparable to the present results could not be found in the literature.

This study focused on sagittal plane variables commonly used to describe human gait. However, one transverse plane variable (free vertical moment) was also included, as AS could be expected to have an effect in that plane. This measure was more variable than the others (Fig. 2C), but

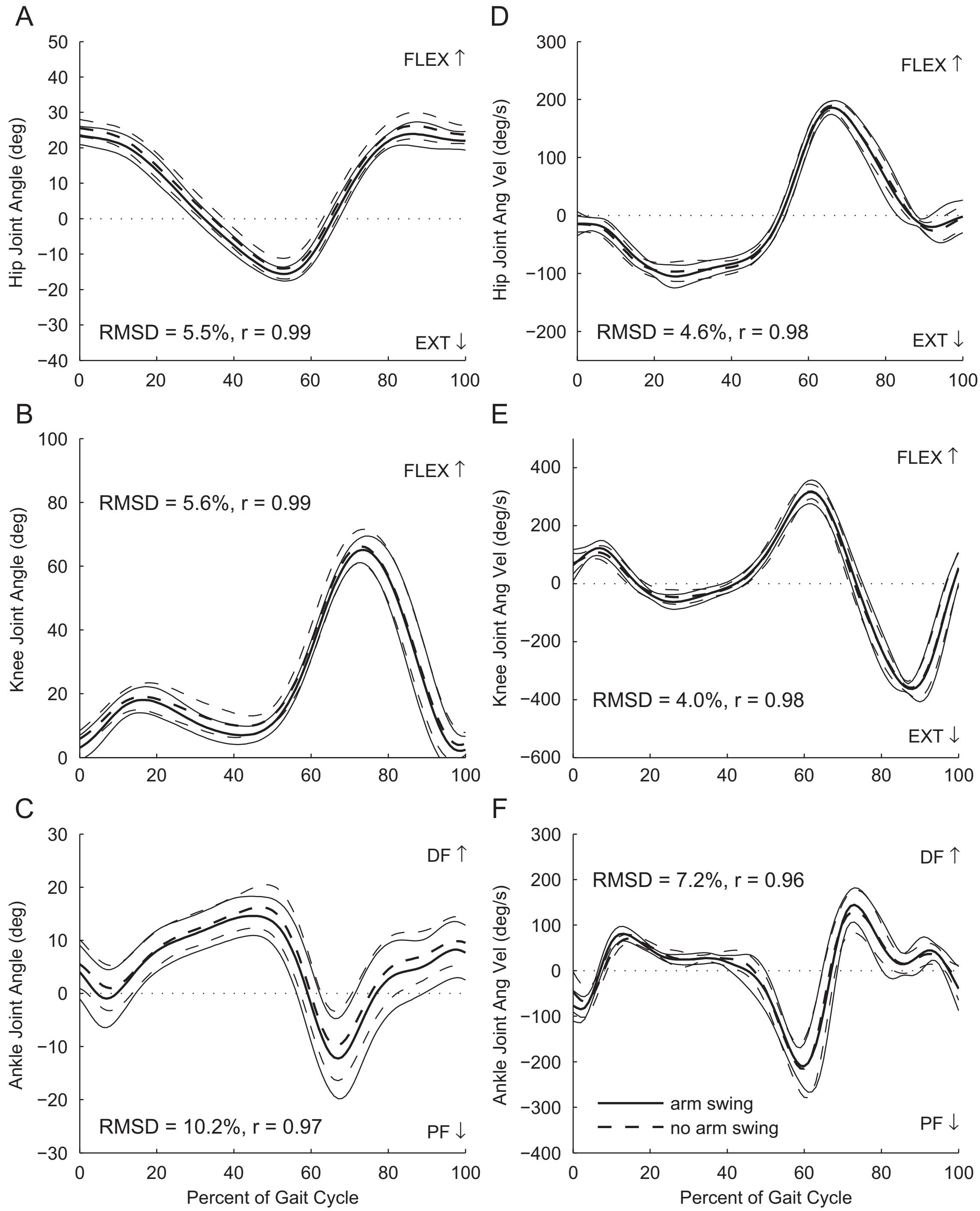

Fig. 1. Sagittal plane joint angles (A–C) and angular velocities (D–F) for the hip, knee, and ankle during walking with (solid lines) and without (dashed lines) normal AS. Bold lines are group means ($n = 8$) and thin lines indicate the first standard deviation envelope. In general, there was a good agreement between the shapes and magnitudes of the curves for walking with and without AS. The transition from stance phase to swing phase occurred at 64% of the gait cycle.

the mean response agreed with Li et al. (2001), exhibiting a larger external rotation moment during mid-to-late stance when walking without AS. This finding should be related to differences in the transverse plane movements of the body segments. Jackson et al. (1983) did report decreased transverse plane rotation of the shoulder girdle when AS was suppressed, however, they did not control for differences in speed and SF/SL, making direct comparisons difficult. Among all of the variables, the free moment exhibited the largest RMSD, however, the mean curves for walking with and without AS were always within 1–2 SD of each other. This variable also exhibited the weakest correlations, yet the patterns were still quite similar for walking with and without AS ($r = 0.76$).

An issue to consider when comparing results across studies is that some investigators used passive devices to restrict AS (Chapman and Ralston, 1964; Park et al., 2000), whereas subjects folded their arms across their chest in the current study. This might be expected to explain some of the discrepancies between studies, however, Jackson et al. (1983) found that passive and voluntary restraint of AS actually resulted in similar suppression of both sagittal plane shoulder rotation and rotation of the shoulder girdle in the transverse plane. An additional issue to consider is that speed and SF/SL were matched, when walking with and without AS in the present study. Most other investigators did not control for these factors, and Eke-Okoro et al. (1997) found that when AS was constrained, subjects adopted shorter, quicker steps at most speeds. Thus, the results of the present study may only apply in the case where speed, SF, and SL are matched between conditions.

Simulation studies of human walking typically rely on a set of experimental data. If a model lacking articulated arms is used, the researcher is left to choose between using data collected during walking with (e.g., Neptune et al., 2001) or without (e.g., Anderson and Pandy, 2001) AS. In either case, it is important to know how restraining the arms affects the biomechanics and energetics of gait. The results of this study, combined with earlier data from Chapman and Ralston (1964), indicate that the gross metabolic cost of walking predicted in a model with a single rigid head–arms–torso segment might be expected to exceed that recorded during normal walking by about 15% (~5% for AS and ~10% for torso restraint). The present results also indicate that sagittal plane kinematic and kinetic variables are only marginally affected by AS. This suggests that it is acceptable to use kinematic and kinetic data collected during normal walking in conjunction with a model that has a single head–arms–torso segment. However, our free moment data suggest that there might be greater differences in the transverse plane, which will be relevant in three-dimensional modeling studies. The importance of these differences will likely be context dependant, and will need to be evaluated by researchers on a case-by-case basis.

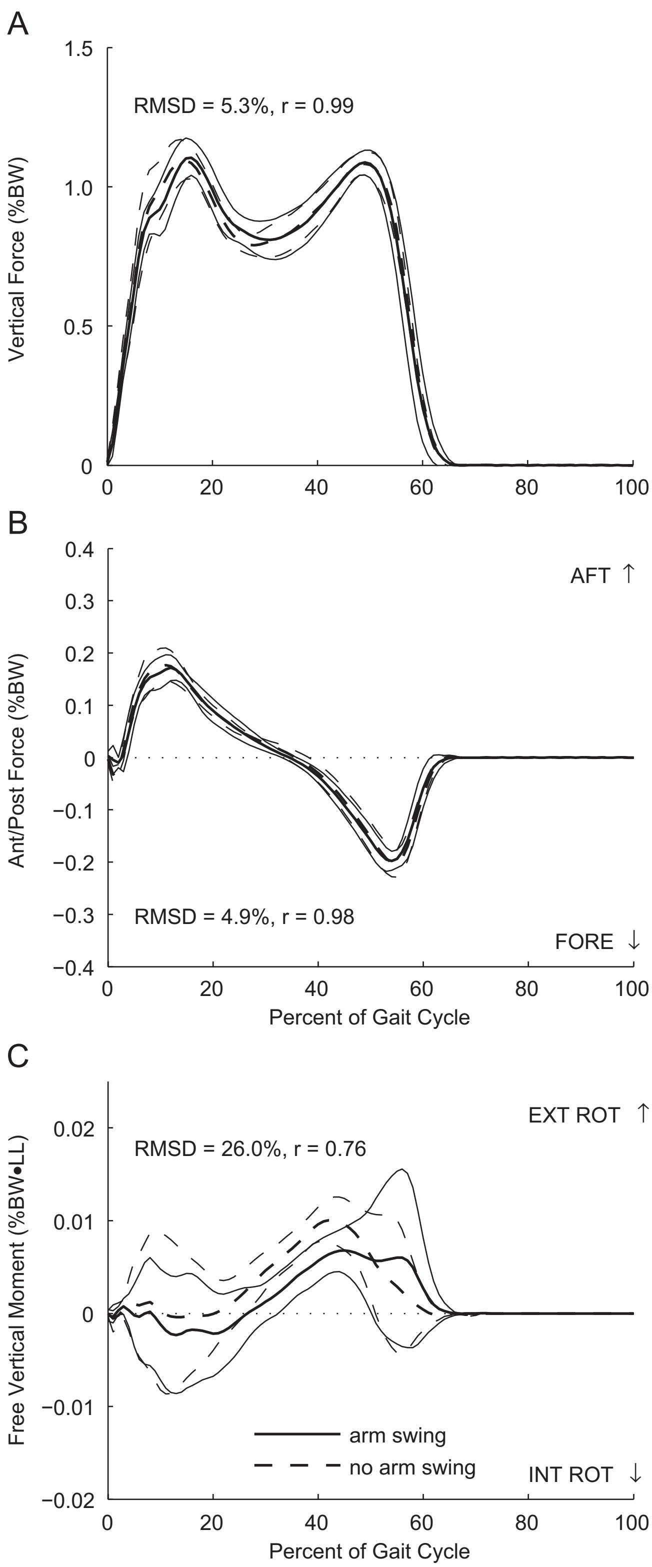

Fig. 2. Vertical (A) and anterior–posterior (B) components of the ground reaction force and the free moment exerted on the surface of the plate (C) during walking with (solid lines) and without (dashed lines) normal AS. Bold lines are group means ($n = 8$) and thin lines indicate the first standard deviation envelope. There was a good agreement for the forces between walking with and without normal AS. However, the free moment had a lower magnitude during the first half of stance and a larger peak magnitude during the second half of stance for walking without AS. Forces were normalized to body weight and the free moment was normalized to body weight and limb length. The transition from stance phase to swing phase occurred at 64% of the gait cycle.

Conflicts of interest

There are no conflicts of interest to report.

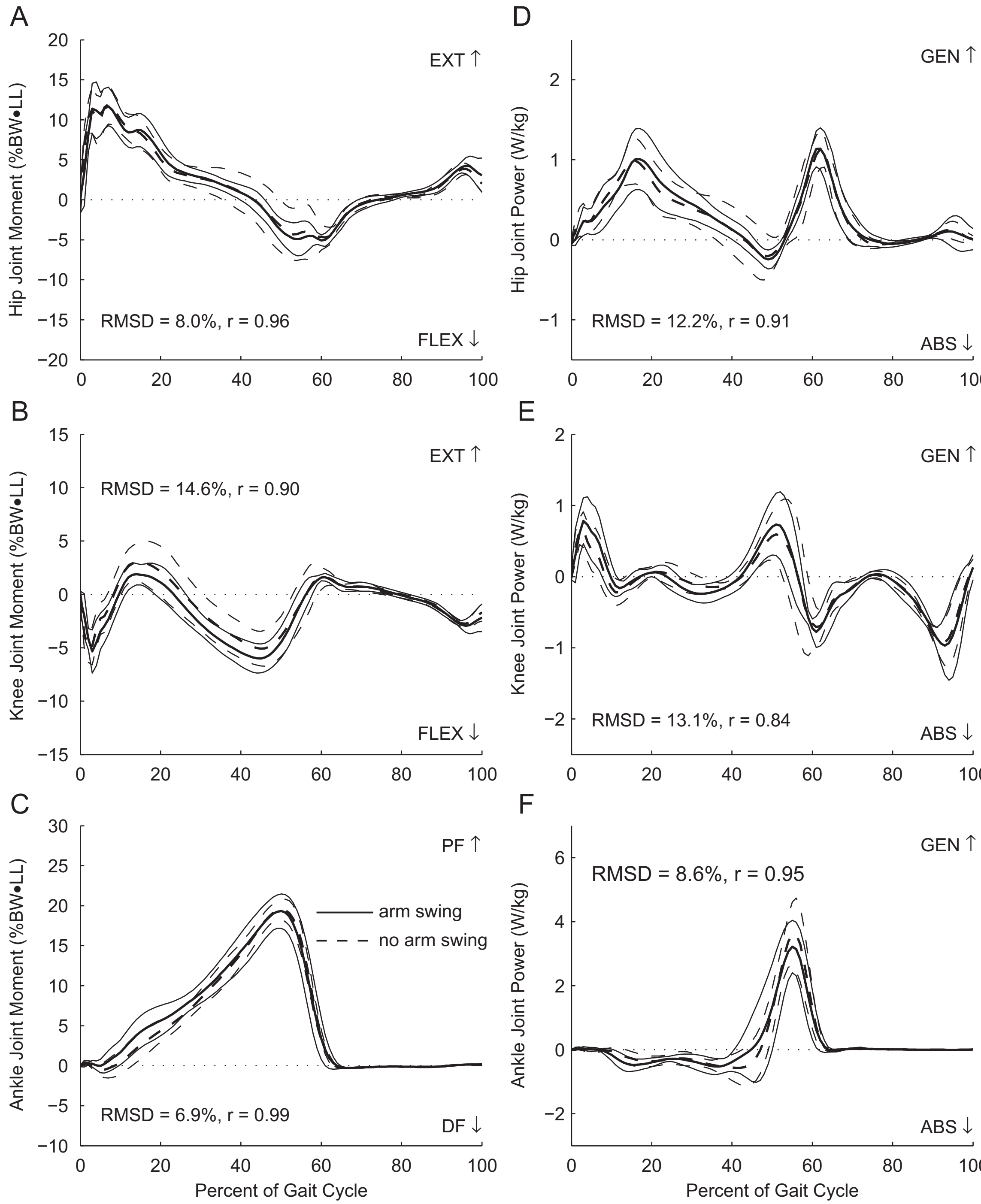

Fig. 3. Sagittal plane joint moments (A–C) and powers (D–F) for the hip, knee, and ankle during walking with (solid lines) and without (dashed lines) normal AS. Bold lines are group means ($n = 8$) and thin lines indicate the first standard deviation envelope. In general, there was a good agreement between the shapes and magnitudes of the curves for walking with and without AS. However, for most variables, the agreement tended to be better during the swing phase than the stance phase. The transition from stance phase to swing phase occurred at 64% of the gait cycle.

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