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Speed, age, sex, and body mass index provide a rigorous basis for comparing the kinematic and kinetic profiles of the lower extremity during walking

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ABSTRACT

The increased use of gait analysis has raised the need for a better understanding of how walking speed and demographic variations influence asymptomatic gait. Previous analyses mainly reported relationships between subsets of gait features and demographic measures, rendering it difficult to assess whether gait features are affected by walking speed or other demographic measures. The purpose of this study was to conduct a comprehensive analysis of the kinematic and kinetic profiles during ambulation that tests for the effect of walking speed in parallel to the effects of age, sex, and body mass index. This was accomplished by recruiting a population of 121 asymptomatic subjects and analyzing characteristic 3-dimensional kinematic and kinetic features at the ankle, knee, hip, and pelvis during walking trials at slow, normal, and fast speeds. Mixed effects linear regression models were used to identify how each of 78 discrete gait features is affected by variations in walking speed, age, sex, and body mass index. As expected, nearly every feature was associated with variations in walking speed. Several features were also affected by variations in demographic measures, including age affecting sagittal-plane knee kinematics, body mass index affecting sagittal-plane pelvis and hip kinematics, body mass index affecting frontal-plane knee kinematics and kinetics, and sex affecting frontal-plane kinematics at the pelvis, hip, and knee. These results could aid in the design of future studies, as well as clarify how walking speed, age, sex, and body mass index may act as potential confounders in studies with small populations or in populations with insufficient demographic variations for thorough statistical analyses.

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1. Introduction

While several studies have outlined isolated relationships between gait parameters and demographic measures of age (e.g., Kang and Dingwell, 2008; Nigg et al., 1994), sex (e.g., Boyer et al., 2008; Cho et al., 2004; Kerrigan et al., 1998), and body mass index (BMI) (e.g., Ko et al., 2010; Spyropoulos et al., 1991), the literature currently lacks a comprehensive study that simultaneously describes the relationships for lower-limb kinetic and kinematic features with respect to these three measures. Fully understanding which demographic measures are associated with a gait feature of interest is critically important when designing clinical studies. Indeed, the relationships between gait and demographics need to be considered when defining the inclusion/exclusion criteria, setting the target demographic distribution of the study population, and planning statistical analyses. A comprehensive understanding of the relationships between demographic measures and gait features will also improve the interpretation of gait data when the experimental setup does not allow for the controlling of all possible demographic confounding variables. For example, it would be difficult to find a height-matched control group for a gait study on basketball players, or to match BMI in a study on subjects with and without diabetes. In these cases, having an *a priori* understanding of the relationships between demographic measures and gait features will help evaluate if positive or negative findings may be attributed to confounding variables.

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The relationships between demographics and gait features are also important because they could help researchers understand the physiopathology of numerous musculoskeletal conditions such as knee osteoarthritis or hip impingement (Anderson and Loeser, 2010; Jung et al., 2011). Specifically, age, sex, and BMI are known to be risk factors for injuries and pathologies of the lower extremity (Bowers et al., 2005; Dillon et al., 2006; Uhorchak et al., 2003), but it is often difficult to outline the pathomechanisms without an overall understanding of the relationship between these demographic measures and gait mechanics. For example, the peak knee adduction moment is an important feature in the development of knee osteoarthritis (Baliunas et al., 2002) and has been shown to vary with demographics (Blazek et al., 2013; Kaufman et al., 2001). But it is only with a recent study (Blazek et al., 2014) that tested healthy subjects who varied in age, sex, and BMI, that it was possible to understand the relative role of each measure on the knee adduction moment and its possible implications in knee osteoarthritis. This prior study improved our understanding of the between-subject variations of frontal-plane knee kinetics and knee osteoarthritis, and calls for a general characterization of the relationships between demographic measures and the overall walking profile of the lower limb.

Ambulatory mechanics are also known to be associated with walking speed (Holden et al., 1997; Jordan et al., 2007; Lelas et al., 2003) such that between-subject and within-subject variations in walking speed could confound the associations between demographics and specific gait features (Astephen Wilson, 2012). In order to assess the effect of walking speed on gait, it is necessary to collect data at different speeds for each subject. Other studies have indeed analyzed the relationship with walking speed, but none has done so with a large dataset and in parallel with several other measures pertaining to subject demographics. There is therefore a need for a comprehensive gait study that simultaneously tests speed with all three major demographic measures in order to assess the extent to which variations in gait mechanics may be attributed to variations in walking speed, age, sex, or BMI.

The objective of this study was to measure gait profiles in a heterogeneous asymptomatic adult population in order to analyze the variations in lower limb kinetics and kinematics in relation to walking speed, age, sex, and BMI. Specifically, this study aimed to identify the features of the kinetic and kinematic walking profiles in all three anatomical planes that are correlated with walking speed and with the three demographic measures.

2. Methods

2.1. Study participants

121 subjects (58 female, 63 male) were selected as an asymptomatic subpopulation of a larger study (Blazek et al., 2014). The mean ± standard deviation age and BMI of the subjects were 37.0 ± 11.1 years and 26.9 ± 5.6 kg/m^2 (histograms in Supplementary Materials). The subjects reported no chronic pain or significant injury to the back or lower extremity. Subjects were recruited using a protocol approved by the Institutional Review Board; informed consent was obtained prior to testing. For each subject, the left or right leg was randomly selected for analysis (54 left, 67 right).

2.2. Gait analysis

Subjects' movement was collected using a 10-camera marker-based motion capture system at 120 Hz (Qualisys, Sweden) and ground reaction forces were measured with floor-embedded force plates at 1200 Hz (Bertec, Ohio). Before capturing gait trials, a calibration phase was achieved where the subjects stood still with arms abducted with 30 reflective markers placed on the side of interest (Fig. 1, Table 1). Fifteen of these markers were placed on anatomical landmarks identified by palpation and 15 were distributed on the thigh and shank segments according to prior publications (Alexander and Andriacchi, 2001; Cappozzo et al., 1995; Dyrby and Andriacchi, 2004; Rylander et al., 2013). Anatomical frames were defined from the positions of the markers placed on anatomical landmarks (Table 2) following biomechanical models in literature (Bell et al., 1990; Cappozzo et al., 1995; Dyrby and Andriacchi, 2004). The standing reference pose was also used to embed a technical frame in each segment using a previously described point cluster technique (Andriacchi et al., 1998; Dyrby and Andriacchi, 2004). The pelvis, thigh, shank, and foot technical frames included 6 markers (#1–3, #13–15), 10 markers (#4 and #16–24), 7 markers (#7 and #25–30), and 3 markers (#10–12), respectively. Finally, the transformations between the technical and anatomical frames were calculated for each segment (Favre et al., 2009). Transformations were also calculated to determine the position of the ankle (i.e., midpoint between the malleoli, #9 and #10), knee (midpoint between the tibial plateaus, #7 and #8), and hip (regression using the four iliac spine markers #1, #3, #13, and #15; Bell et al., 1990) joint centers based on the technical frames.

After recording the standing reference pose, the subjects were asked to walk 10-meter long trials at a self-selected normal speed, a self-selected "slower-than-normal" speed, and a self-selected "faster-than-normal" speed. The subjects took at least two practice trials before data were collected for each speed. The variations in walking speed between the three trials at a requested speed were limited by ensuring that the subjects started walking at the same location for the three trials and made the same number of footsteps (at least five) until they hit the force plate. Three successful trials were collected at each speed, with a trial being considered successful if the foot of the leg of interest fully stepped on a force plate. For each trial, the point cluster algorithm was used to calculate the position and orientation of the four technical frames and then the technical-to-anatomical transformations were used to calculate the position and orientation of the pelvis, thigh, shank, and foot anatomical frames. Next, the kinematic curves describing the pelvic rotation (i.e., rotation of the pelvis anatomical frame with respect to the global lab frame) during the entire trial were calculated according to Baker (2001). Similarly, the curves describing the hip, knee, and ankle rotation (i.e., rotation between two consecutive anatomical frames) were calculated following Grood and Suntay (1983). In addition to the kinematic assessment, a previously described inverse dynamic approach was used to calculate the external moment acting on the lower-limb joints (Andriacchi et al., 2005). This calculation was based on the position and orientation of the anatomical frames, the transformation to calculate the position of the joint centers based on the technical frames, the signals measured by the force plate, and classical modeling of the segment inertial properties using literature (Dempster and Gaughran, 1967). Moments were normalized to bodyweight and height (%BW∗Ht) to allow for comparison between subjects and expressed in the anatomical frame of the distal segment.

2.3. Description of kinematic and kinetic walking profiles

For each trial, the heel-strike and toe-off events for the gait cycle in the middle of the walkway were detected using a threshold of 10 N for the vertical force measured by the force plate. The gait cycle was then standardized to 101 points for the angle curves, and stance phase was normalized to 101 points for the moment curves.

In order to describe the kinetic and kinematic gait cycle curves using peak features, it was first necessary to confirm that each curve displays a characteristic pattern that exists despite subject-specific variations in curve means and amplitudes. To accomplish this, all curves of all subjects were individually standardized by shifting the curve's mean to 0 and scaling its standard deviation to 1; this adjusts for shifts in angles and moments that may exist between subjects and walking speeds, as well as differences in curve amplitudes. Next, these standardized curves were used to calculate the coefficient of multiple correlations (CMC) (Kadaba et al., 1989; Kutner et al., 2005) for each curve considering all speeds of all subjects in order to identify curves that have consistent patterns across subjects and speeds and can thus be described using peak features (Christe et al., 2016; Favre et al., 2010). This assessment results in a score between 0 (no common pattern among curves) and 1 (same pattern among curves). A CMC greater than 0.50 indicates that the curves share a common pattern and thus that the curves can be described using local maximum and minimum peaks. Consequently, the curves reporting a CMC above 0.50 were visually screened to list all major peak features. A hierarchical approach was then designed to automatically detect these features in the non-standardized curves, with some features being detected by looking for maximum or minimum peaks in predefined percentages of the gait cycle and other features being detected by looking for maximum or minimum peaks between already detected features (Table 3). In agreement with literature, the set of features used to describe the curves was completed with joint angles at heel-strike and ranges of motion. All processing was done with the software application BioMove (Stanford University, CA).

2.4. Statistical analysis

The characteristic gait features were identified for each trial and the features from the three trials with the same walking speed were averaged to obtain one value per feature, walking speed, and subject. Mixed-effects linear regression models were used with non-normalized data to test for associations between walking speed, age, sex, and BMI for each discrete gait feature. Each feature was fitted for fixed effects for walking speed, age, sex, and BMI, and a random intercept for subject due to the fact that three data points were used for each subject (one at each speed) and cannot be assumed to be independent using the following equation:

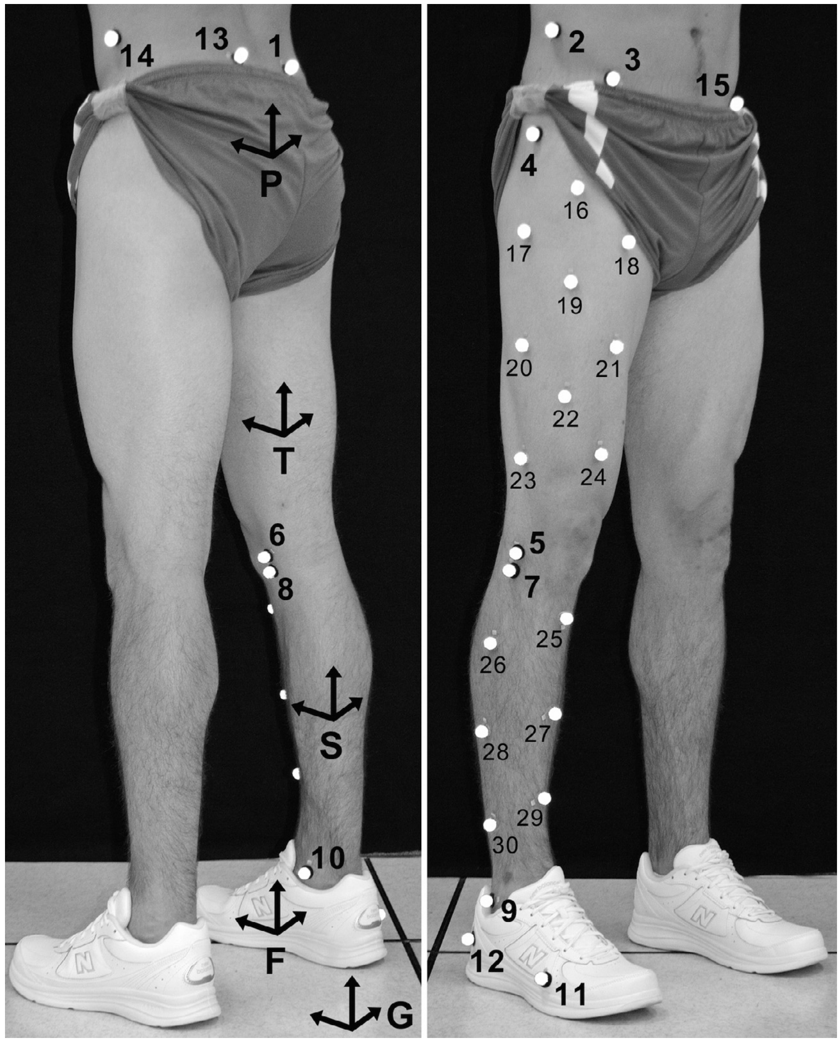

Fig. 1. Location of skin surface markers and reference frames. Bold numbers indicate anatomical bony landmarks and non-bolded numbers indicate point cluster markers, as described in Table 1.

Table 1
List of markers.

Anatomical markers	
Side of interest	
#1	Posterior superior iliac spine
#2	Tip of iliac crest
#3	Anterior superior iliac spine
#4	Greater trochanter
#5	Lateral femoral condyle
#6	Medial femoral condyle
#7	Lateral tibial plateau
#8	Medial tibial plateau
#9	Lateral malleolus
#10	Medial malleolus
#11	Fifth metatarsal
#12	Lateral calcaneus
Contralateral side	
#13	Posterior superior iliac spine
#14	Tip of iliac crest
#15	Anterior superior iliac spine
Non-anatomical markers	
Side of interest	
#16–24	Femur
#25–30	Tibia

$$\mathrm{GaitFeature} = \beta_S \cdot \mathrm{Speed} + \beta_A \cdot \mathrm{Age} + \beta_M \cdot \mathrm{Sex} + \beta_B \cdot \mathrm{BMI} + (1|\mathrm{Subject}) + \epsilon$$

where GaitFeature, Speed, Age, Sex, and BMI each correspond to a 363×1 vector that includes data points from each subject at each of the three speeds; (1|Subject) represents a random intercept term that is different for each subject. Females were coded as 0 and males were coded as 1 such that a positive regression coefficient for sex indicates a positive effect for males and a negative effect for females. The mixed-effects models were used to obtain a p-value for each fixed effect for each gait feature. Bonferroni *p*-value significance threshold correction was applied to adjust for using 78 regression models to test for statistical correlations with each of the discrete gait features, such that a significance threshold of $6.41 * 10^{-4}$ (0.05/78) was used. All statistical analyses were completed using MATLAB R2014b (The MathWorks, Natick, MA).

3. Results

The subjects' mean ± standard deviation walking speeds were slow: 0.93 ± 0.17 m/s; normal: 1.38 ± 0.16 m/s; fast: 1.83 ± 0.25 m/s (histograms in Supplementary Materials). The CMC was greater than 0.50 for all 12 of the angle and all 9 of the moment curves, allowing for the identification and definition of 78 characteristic features (Table 3) pertaining to heel-strike values, peaks, and ranges (Figs. 2 and 3). Walking speed was significantly correlated with 69 of the 78 features and its p-value was usually smaller than the other demographic measures (Table 3). Nevertheless, 28

Table 2
Markers used for the definition of the anatomical frames for the four lower-limb segments.

Pelvis	
Origin	Middle of the *anterior superior iliac spines* (#3 and #15)
x-axis	From the *left anterior superior iliac spine* (#3 or #15) to the *right superior iliac spine* (#3 or #15), pointing to the right
Temporary y-axis	From the middle of the *posterior superior iliac spines* (#1 and #13) to the middle of the *anterior superior iliac spines* (#3 and #15), pointing anteriorly
z-axis	Perpendicular to *x*-axis and temporary y-axis, pointing cranially
y-axis	Perpendicular to *z*-axis and *x*-axis, pointing anteriorly
Thigh	
Origin	Middle of the *femoral condyles* (#5 and #6)
z-axis	From the *lateral femoral condyle* (#5) to the *greater trochanter* (#4), pointing proximally
Temporary x-axis	From the *medial femoral condyle* (#6) to the *lateral femoral condyle* (#5) if right leg or from the *lateral femoral condyle* (#5) to the *medial femoral condyle* (#6) if left leg
y-axis	Perpendicular to *z*-axis and temporary *x*-axis, pointing anteriorly
x-axis	Perpendicular to *y*-axis and *z*-axis, pointing to the right
Shank	
Origin	Middle of the *tibial plateaus* (#7 and #8)
z-axis	From the middle of the *malleoli* (#9 and #10) to the middle of the *tibial plateaus* (#7 and #8), pointing proximally
Temporary x-axis	From the *medial tibial plateau* (#8) to the *lateral tibial plateau* (#7) if right leg, or from the *lateral tibial plateau* (#7) to the *medial tibial plateau* (#8) if left leg.
y-axis	Perpendicular to *z*-axis and temporary *x*-axis, pointing anteriorly
x-axis	Perpendicular to *y*-axis and *z*-axis, pointing to the right
Foot	
Origin	Middle of the *malleoli* (#9 and #10)
y-axis	From the *lateral calcaneus* (#12) to the *fifth metatarsal* (#11), pointing anteriorly
Temporary z-axis	Perpendicular to the floor, pointing upwards
x-axis	Perpendicular to *y*-axis and temporary *z*-axis, pointing to the right
y-axis	Perpendicular to *z*-axis and *x*-axis, pointing anteriorly

statistically significant Bonferroni-corrected correlations between gait features and age, sex, and BMI remained even with the strong correlations with walking speed (Table 3). Specifically, the demographic measures were highlighted by three clusters of significant correlations beyond the effects of gait variations due to walking speed, with age being correlated with sagittal-plane knee features, BMI being correlated with pelvis, hip and knee features, and sex being correlated with frontal-plane kinematics. All significant correlations between the demographic measures and characteristic gait features are also illustrated in Figs. 2 and 3 so the direction of changes may be more easily visualized (example 2D correlation plots in Supplementary Materials).

There were significant findings between the knee moments and all four tested measures; these correlations are described in more detail in the following paragraph to provide an example of reading the data from Table 3. Note that, in the text below, the gait features are presented relative to their anatomical directions. With this convention, when the value of a negative feature (minimum or maximum peak) increases/decreases, it is presented as a decrease/increase in the peak amplitude.

All discrete gait features for the knee moments (Table 3; Fig. 3) except the second peak adduction moment (KMFd) were correlated with walking speed. The positive β coefficients for the positive features (KMSb, KMFa, KMFc and KMTa) and the negative β coefficients for the negative features (KMSb, KMFa, KMFc and KMTa) indicated that the amplitudes of all eight features increased with faster walking speed. The positive coefficient for age at KMSb indicates that increased age was associated with an increase in the amplitude of the peak knee flexion moment. The negative coefficient for sex at KMFd indicates that on average, the amplitude of second peak knee adduction moment was larger in males. Finally, increased BMI was associated with a decrease in the amplitudes of the first peak knee adduction moment (KMFb), peak external rotation moment (KMTa), and peak internal rotation moment (KMTb). The unstandardized β coefficients in Table 3 can be used for a quantitative representation of the results. For example, an increase of walking speed by 0.5 m/s was associated with a 1.06% BW∗Ht increase in the amplitude of the first peak knee extension moment (KMSa). Similarly, a 10-year increase in age was associated with a 0.3% BW∗Ht increase in the amplitude of the peak knee flexion moment (KMSb); males had 0.41% BW∗Ht greater second peak knee adduction moment (KMFd) amplitude; and an increase of 5 kg/m^2 in BMI was associated with a 0.2% BW∗Ht decrease in the first peak knee adduction moment (KMFb) amplitude.

4. Discussion

This study showed that walking speed was significantly correlated with the ambulatory kinematics and kinetics at every joint and in each anatomical plane. The results obtained for 69 of the 78 discrete gait features analyzed in this study reinforce the need to account for walking speed when comparing gait features as it may act as a notable confounder in nearly all scenarios. But this study also highlighted correlations with demographic measures for over one third of the features, thus suggesting that inter-subject variations in demographics should be considered in addition to walking speed when analyzing gait. This may be achieved, for example, by recruiting populations of subjects that walk at similar speeds and are homogeneous in the demographic measures correlated with the feature of interest. This would reduce the risk of seeing variations in the gait feature that are due to differences in walking speed and subject demographics. Conversely, if a population is demographically heterogeneous and demographic-based effects on gait are a concern, one could include speed and the appropriate demographic measures as fixed effects in statistical models, as in this study.

This study found that 28 of the 78 gait features were correlated with at least one demographic measure, with three particular clusters of significant correlations between gait features and demographic measures that are worth further discussion. The first cluster of significant correlations showed that age had significant effects on sagittal-plane knee features during the weight acceptance portion of stance, which may be an indication of age-related decreases in neuromuscular control (McGibbon, 2003; Schmitz et al., 2009). These findings are in support of other studies reporting similar differences among subpopulations with different age (Begg and Sparrow, 2006). When considering that these differences further progress with the onset of pathological conditions such as knee osteoarthritis (Favre et al., 2014), the sagittal-plane knee features that were associated with age could possibly serve as early markers for some diseases, including knee osteoarthritis. Further studies are needed to assess this possibility.

The second cluster of significant correlations outlined the influence of BMI on specific gait features at the pelvis and hip. Previous studies have reported conflicting relationships between obesity and hip abduction, and did not report significant findings between obesity and hip flexion (Browning and Kram, 2007; Lai et al., 2008; Spyropoulos et al., 1991). However, these prior analyses did not directly account for walking speed, even when variations in speed were reported between groups with different BMI. This renders it difficult to assess whether the variations in gait features between subjects with different BMI could be attributed to demographic

Table 3
Definition of discrete gait features that characterize the kinematic and kinetic profiles, and their correlations with walking speed, age, sex, and BMI. Beta coefficients are calculated relative to the directions of the kinetic and kinematic curves, indicated by ± signs in the headers below (see Figs. 2 and 3). Features are labeled using a notation of joint (P/H/K/A), angle or moment (A/M), plane (S/F/T), and feature (a, b, c, etc.). β coefficients are unstandardized. Bolded *p*-values indicate statistical significance with the Bonferroni threshold.

Discrete point definitions		Correlations							
Pelvis angles (degrees)		Speed (m/s)		Age (yrs)		Sex (F → M)		BMI (kg/m^2)	
Sagittal Plane: ± *indicates pelvis forwards/backwards tilt*		β_S	*p*	β_A	*p*	β_M	*p*	β_B	*p*
PASa	Pelvis sagittal-plane angle at heel-strike	2.42	**9.0e−41**	0.06	6.6e−02	−1.11	1.5e−01	0.55	**1.3e−14**
PASb	Minimum pelvis forwards tilt angle during loading response	2.38	**3.8e−43**	0.08	2.4e−02	−1.14	1.3e−01	0.41	**1.8e−09**
PASc	Maximum pelvis forwards tilt angle during terminal stance	2.93	**3.3e−49**	0.07	4.2e−02	−1.48	4.9e−02	0.52	**4.1e−14**
PASd	Minimum pelvis forwards tilt angle after terminal stance	2.13	**4.7e−41**	0.08	2.0e−02	−1.15	1.3e−01	0.42	**1.4e−09**
PASe	Total pelvis sagittal-plane angular excursion	0.66	**3.5e−10**	−0.01	1.6e−01	−0.34	2.8e−02	0.13	**2.7e−18**
Frontal Plane: ± *indicates pelvis rise/drop (obliquity)*									
PAFa	Pelvis frontal-plane angle at heel-strike	−0.23	8.4e−03	0.02	3.8e−01	1.26	1.2e−03	0.00	1.0e + 00
PAFb	Minimum pelvis rise angle during terminal stance	−1.80	**3.1e−67**	0.01	7.2e−01	2.30	**1.1e−09**	0.07	3.8e−02
PAFc	Maximum pelvis rise angle after terminal stance	1.58	**4.4e−57**	−0.01	4.9e−01	−1.16	4.2e−03	−0.04	3.2e−01
PAFd	Total pelvis frontal-plane angular excursion	3.25	**5.2e−77**	−0.01	3.9e−01	−3.33	**4.1e−19**	−0.09	2.9e−03
Transverse Plane: ± *indicates pelvis anterior/posterior rotation*									
PATa	Pelvis transverse-plane angle at heel-strike	3.36	**1.4e−46**	0.01	6.7e−01	0.37	5.3e−01	0.05	3.3e−01
PATb	Maximum pelvis anterior rotation angle during stance phase	1.91	**2.0e−21**	−0.01	5.5e−01	−0.24	6.1e−01	0.06	1.9e−01
PATc	Minimum pelvis anterior rotation angle during swing phase	−1.25	**7.7e−09**	−0.01	7.9e−01	0.57	2.1e−01	−0.01	8.9e−01
PATd	Total pelvis transverse-plane angular excursion	3.54	**3.8e−21**	0.00	9.1e−01	−0.76	1.6e−01	0.07	1.4e−01
Hip Angles (degrees)		Speed (m/s)		Age (yrs)		Sex (F → M)		BMI (kg/m^2)	
Sagittal Plane: ± *indicates hip flexion/extension*		β_S	*p*	β_A	*p*	β_M	*p*	β_B	*p*
HASa	Hip sagittal-plane angle at heel-strike	7.94	**2.0e−134**	0.13	2.7e−03	−2.78	2.6e−03	0.55	**7.8e−11**
HASb	Minimum hip flexion angle during stance phase	−3.69	**5.3e−71**	0.09	5.1e−02	−1.85	7.9e−02	0.55	**6.5e−09**
HASc	Maximum hip flexion angle during swing phase	6.08	**8.2e−86**	0.11	4.6e−03	−3.35	**1.2e−04**	0.53	**2.3e−11**
HASd	Total hip sagittal-plane angular excursion	10.37	**3.8e−148**	0.03	3.9e−01	−1.31	7.0e−02	−0.02	7.2e−01
Frontal Plane: ± *indicates hip abduction/adduction*									
HAFa	Hip frontal-plane angle at heel-strike	1.18	**6.4e−12**	0.03	3.3e−01	5.09	**1.9e−17**	−0.19	**2.2e−04**
HAFb	Minimum hip abduction angle during stance phase	−0.92	**3.2e−15**	0.01	5.3e−01	4.95	**4.2e−25**	−0.10	1.1e−02
HAFc	Maximum hip abduction angle after terminal stance	1.66	**1.6e−35**	−0.03	2.3e−01	1.51	7.3e−03	0.01	8.7e−01
HAFd	Total hip frontal-plane angular excursion	2.63	**1.8e−40**	−0.04	5.0e−02	−3.48	**1.4e−11**	0.11	1.2e−02
Transverse Plane: ± *indicates hip external/internal rotation*									
HATa	Hip transverse-plane angle at heel-strike	−0.65	8.0e−04	0.06	2.9e−01	1.13	3.9e−01	0.23	4.8e−02
HATb	Minimum hip external rotation angle during stance phase	−0.91	**2.3e−07**	0.05	3.7e−01	2.16	9.6e−02	0.26	2.3e−02
HATc	Maximum hip external rotation angle after terminal stance	1.49	**1.4e−27**	−0.02	6.8e−01	2.09	1.1e−01	0.18	1.2e−01
HATd	Total hip transverse-plane angular excursion	1.65	**3.6e−15**	−0.08	1.9e−02	−0.27	7.1e−01	−0.03	5.9e−01
Knee Angles (degrees)		Speed (m/s)		Age (yrs)		Sex (F → M)		BMI (kg/m^2)	
Sagittal Plane: ± *indicates knee flexion/extension*		β_S	*p*	β_A	*p*	β_M	*p*	β_B	*p*
KASa	Knee sagittal-plane angle at heel-strike	4.77	**2.2e−81**	0.11	1.5e−03	−1.81	1.5e−02	0.04	5.4e−01
KASb	Maximum knee flexion angle during loading response	9.93	**2.2e−85**	0.22	**7.5e−07**	−0.06	9.5e−01	0.09	3.0e−01
KASc	Minimum knee flexion angle during terminal stance	−1.26	**1.5e−05**	0.12	2.2e−03	−0.50	5.7e−01	−0.04	6.2e−01
KASd	Maximum knee flexion angle during swing phase	1.27	**3.0e−05**	0.04	2.1e−01	−0.62	3.5e−01	−0.04	5.2e−01
KASe	Total knee sagittal-plane excursion	−1.30	**5.9e−04**	−0.04	2.3e−01	0.28	6.9e−01	0.00	9.5e−01
Frontal Plane: ± *indicates knee abduction/adduction*									
KAFa	Knee frontal-plane angle at heel-strike	−1.10	**1.5e−30**	−0.04	9.8e−02	−4.00	**2.9e−14**	0.32	**7.1e−12**
KAFb	Minimum knee abduction angle during loading response	−2.19	**3.3e−45**	−0.05	4.6e−02	−2.62	**2.2e−06**	0.24	**1.2e−06**
KAFc	Maximum knee abduction angle during terminal stance	0.83	**1.0e−28**	−0.04	1.0e−01	−3.02	**3.5e−08**	0.22	**6.6e−06**
KAFd	Minimum knee abduction angle during swing phase	0.08	6.7e−01	0.06	1.8e−01	−1.89	4.2e−02	0.00	9.9e−01
KAFe	Total knee frontal-plane angular excursion	0.28	1.3e−01	−0.09	1.4e−02	−1.87	2.7e−02	0.23	2.0e−03
Transverse Plane: ± *indicates knee external/internal rotation*									
KATa	Knee transverse-plane angle at heel-strike	1.99	**3.4e−17**	−0.04	4.0e−01	1.40	1.6e−01	0.15	1.0e−01
KATb	Minimum knee external rotation angle during stance phase	−0.62	**6.6e−05**	0.02	5.3e−01	0.26	7.4e−01	0.07	3.1e−01
KATc	Maximum knee external rotation angle during swing phase	−0.94	**4.8e−04**	−0.03	5.0e−01	−1.76	4.4e−02	0.27	**4.0e−04**
KATd	Total knee transverse-plane angular excursion	0.66	1.4e−02	−0.05	1.6e−01	−1.94	1.0e−02	0.17	1.2e−02
Ankle Angles (degrees)		Speed (m/s)		Age (yrs)		Sex (F → M)		BMI (kg/m^2)	
Sagittal Plane: ± *indicates ankle dorsiflexion/plantar flexion*		β_S	*p*	β_A	*p*	β_M	*p*	β_B	*p*
AASa	Ankle sagittal-plane angle at heel-strike	4.90	**2.3e−89**	0.01	7.7e−01	0.89	1.1e−01	0.05	3.6e−01
AASb	Minimum ankle dorsiflexion angle during loading response	4.21	**2.8e−87**	0.02	3.9e−01	1.13	3.6e−02	−0.05	2.9e−01
AASc	Maximum ankle dorsiflexion angle during stance phase	−2.04	**2.9e−18**	0.05	4.7e−02	0.33	5.7e−01	0.02	6.7e−01
AASd	Minimum ankle dorsiflexion angle after terminal stance	−4.54	**1.0e−42**	0.13	7.8e−04	2.92	6.5e−04	−0.19	1.4e−02
AASe	Total ankle sagittal-plane angle excursion	2.75	**9.4e−14**	−0.07	3.0e−02	−2.71	**4.0e−04**	0.22	1.3e−03
Frontal Plane: ± *indicates ankle eversion/inversion*									
AAFa	Ankle frontal-plane angle at heel-strike	2.35	**2.1e−40**	0.05	1.2e−01	0.08	3.5e−01	−0.13	2.8e−02
AAFb	Maximum ankle eversion angle during stance phase	0.06	7.1e−01	0.02	5.2e−01	1.61	9.0e−01	−0.10	5.9e−02
AAFc	Minimum ankle eversion angle after terminal stance	1.57	**5.9e−18**	0.00	9.1e−01	0.76	2.6e−02	−0.05	4.6e−01
AAFd	Maximum ankle eversion angle during late swing	2.35	**1.2e−30**	0.06	6.2e−02	−1.52	2.4e−01	−0.11	5.6e−02
AAFe	Total ankle frontal-plane angular excursion	−1.52	**2.6e−15**	0.02	6.2e−01	−0.53	2.6e−02	−0.05	3.9e−01

(*continued on next page*)

Table 3 (*continued*)

Ankle Angles (degrees)		Speed (m/s)		Age (yrs)		Sex (F → M)		BMI (kg/m^2)	
Transverse Plane: ± *indicates ankle external/internal rotation*									
AATa	Ankle transverse-plane angle at heel-strike	1.31	**4.9e−20**	0.05	3.2e−01	−0.33	6.4e−01	−0.15	1.4e−01
AATb	Maximum ankle external rotation angle during stance phase	0.43	1.8e−03	0.03	5.4e−01	1.98	7.9e−01	−0.20	6.5e−02
AATc	Minimum ankle external rotation angle during late swing	−0.29	1.9e−01	0.06	2.5e−01	−0.56	1.1e−01	−0.24	2.8e−02
AATd	Maximum ankle external rotation angle during late swing	1.54	**2.3e−26**	0.06	3.0e−01	−2.45	6.4e−01	−0.10	3.4e−01
AATe	Total ankle transverse-plane angular excursion	0.91	**8.0e−05**	−0.03	3.0e−01	−0.33	**3.0e−05**	0.06	2.4e−01
Hip Moments (%BW*Ht)		Speed (m/s)		Age (yrs)		Sex (F → M)		BMI (kg/m^2)	
Sagittal Plane: ± *indicates hip extension/flexion moment*		β_S	*p*	β_A	*p*	β_M	*p*	β_B	*p*
HMSa	Minimum hip extension moment during stance phase	−5.98	**2.2e−143**	−0.01	5.1e−01	0.23	2.6e−01	0.01	6.1e−01
HMSb	Maximum hip extension moment during stance phase	2.34	**6.1e−105**	−0.02	1.0e−02	−0.35	6.4e−02	−0.03	1.1e−01
Frontal Plane: ± *indicates hip abduction/adduction moment*									
HMFa	Maximum hip abduction moment during loading response	2.17	**2.8e−75**	0.00	8.9e−01	0.01	9.6e−01	−0.02	3.5e−01
HMFb	Minimum hip abduction moment during loading response	−1.31	**1.3e−65**	0.00	5.6e−01	0.03	8.6e−01	0.05	2.7e−03
HMFc	Maximum hip abduction moment between HMFb and HMFd	1.39	**8.3e−85**	−0.01	2.4e−01	0.18	7.5e−02	0.01	2.1e−01
HMFd	Minimum hip abduction moment during terminal stance	0.20	**1.3e−04**	0.01	3.9e−01	−0.21	2.0e−01	0.05	1.1e−03
Transverse Plane: ± *indicates hip external/internal rotation moment*									
HMTa	Minimum hip external rotation moment during loading response	−0.65	**9.9e−57**	0.00	6.7e−01	0.41	**7.0e−20**	−0.02	**1.7e−06**
HMTb	Maximum hip external rotation moment during loading response	0.36	**2.2e−58**	0.00	2.0e−02	−0.02	6.9e−01	−0.02	**2.6e−05**
HMTc	Minimum hip external rotation moment during terminal stance	−0.08	**1.5e−08**	0.00	9.5e−02	−0.10	2.9e−02	0.00	3.8e−01
Knee moments (%BW*Ht)		Speed (m/s)		Age (yrs)		Sex (F → M)		BMI (kg/m^2)	
Sagittal Plane: ± *indicates knee flexion/extension moment*		β_S	*p*	β_A	*p*	β_M	*p*	β_B	*p*
KMSa	Minimum knee flexion moment during loading response	−2.12	**6.5e−97**	0.01	2.0e−01	−0.04	7.7e−01	0.00	6.8e−01
KMSb	Maximum knee flexion moment during loading response	3.11	**3.7e−114**	0.03	**5.9e−04**	−0.02	9.3e−01	0.03	1.2e−01
KMSc	Minimum knee flexion moment during terminal stance	−0.60	**7.7e−22**	0.02	4.0e−02	−0.14	3.9e−01	0.03	4.5e−02
Frontal Plane: ± *indicates knee abduction/adduction moment*									
KMFa	Maximum knee abduction moment during loading response	0.59	**1.4e−44**	0.00	3.3e−01	−0.06	4.7e−01	0.00	8.2e−01
KMFb	Minimum knee abduction moment during loading response	−1.34	**6.8e−98**	−0.01	3.7e−02	−0.15	2.3e−01	0.04	**2.9e−04**
KMFc	Maximum knee abduction moment between KMFb and KMFd	0.66	**2.52e−61**	0.00	2.90e−01	−0.15	7.83e−02	0.00	5.59e−01
KMFd	Minimum knee abduction moment during terminal stance	−0.07	3.5e−02	0.00	9.9e−01	−0.41	**2.7e−04**	0.02	6.8e−02
Transverse Plane: ± *indicates knee external/internal rotation moment*									
KMTa	Maximum knee external rotation moment during loading response	0.07	**4.0e−18**	0.00	5.2e−01	−0.01	3.9e−01	−0.01	**1.5e−04**
KMTb	Minimum knee external rotation moment during stance phase	−0.17	**3.7e−31**	0.00	1.1e−02	−0.10	1.1e−02	0.01	**1.7e−04**
Ankle Moments (%BW*Ht)		Speed (m/s)		Age (yrs)		Sex (F → M)		BMI (kg/m^2)	
Sagittal Plane: ± *indicates ankle plantar flexion/dorsiflexion moment*		β_S	*p*	β_A	*p*	β_M	*p*	β_B	*p*
AMSa	Maximum ankle plantar flexion moment during loading response	0.54	**4.8e−61**	0.00	2.1e−01	0.00	9.6e−01	−0.01	4.7e−02
AMSb	Minimum ankle plantar flexion moment during stance phase	−1.40	**6.3e−61**	0.02	1.8e−03	−0.20	1.1e−01	0.05	**5.3e−05**
Frontal Plane: ± *indicates ankle eversion/inversion moment*									
AMFa	Maximum ankle eversion moment during loading response	0.12	**2.1e−18**	0.00	7.0e−01	−0.03	4.1e−01	0.00	4.3e−01
AMFb	Minimum ankle eversion moment during stance phase	−0.46	**8.9e−58**	0.00	5.4e−01	−0.14	3.9e−02	0.01	2.0e−01
Transverse Plane: ± *indicates ankle external/internal rotation moment*									
AMTa	Maximum ankle external rotation moment during loading response	0.10	**1.3e−29**	0.00	1.6e−02	−0.02	4.2e−01	0.00	5.9e−03
AMTb	Minimum ankle external rotation moment during stance phase	−0.21	**2.8e−19**	0.00	2.7e−01	−0.03	6.7e−01	−0.01	3.9e−01

measures without a comprehensive study design and analysis as in this study. Specifically, the present study found that increased BMI was associated with greater forward pelvic tilt (PASa-e) and increased hip flexion (HASa-c) independent of walking speed. These correlations between BMI and walking mechanics could be due to differences in the location of subjects' center of mass. Indeed, morphological variations could affect the pelvis and hip during both static and dynamic conditions by altering stability requirements and modifying the ground reaction force.

Even when accounting for variations in walking speed, age, and sex, it was found that subjects with greater BMI had more abducted knees during stance (KAFa-b), and a first peak adduction moment (KMFb) of reduced amplitude. These results again highlighted the value of comprehensive analyses. Indeed, these findings differed from previous studies outlining that obese subjects walk with greater knee adduction (Lai et al., 2008), suggesting a possible confounding effect of slower walking speeds. The present study specifically found that slower speeds were alone correlated with greater knee adduction. It is worth noting that variations in sagittal-plane kinematics and kinetics at the knee were not significantly correlated to variations in BMI, as supported by previous works (Freedman Silvernail et al., 2013; Lai et al., 2008). It should also be noted that although the BMI measure has been shown to be a reasonable characterization of a subject's relative body weight (Keys et al., 1972; Micozzi et al., 1986), possible variations in weight and adiposity for a given height (Nevill and Metsios, 2015) indicate that higher BMIs should not be absolutely interpreted as "overweight" or "obese".

The third cluster of significant correlations was related to the influence of sex on frontal-plane kinematics. The complexity of sex as a confounder to gait is often attributed to neuromuscular and anatomical variations (Chiu and Wang, 2007; Chumanov et al., 2008), which are rarely quantified in gait studies. In addition, males and females are commonly pooled together in gait studies such that sex differences are not considered in statistical analyses. However, the results from this study suggested that sex has a specific impact at the pelvis, hip, and knee, where females were found to have greater pelvic drop angles (PAFb), more adducted hip angles during stance (HAFa-b), more abducted knee angles during stance (KAFa-c) and greater frontal-plane excursions at

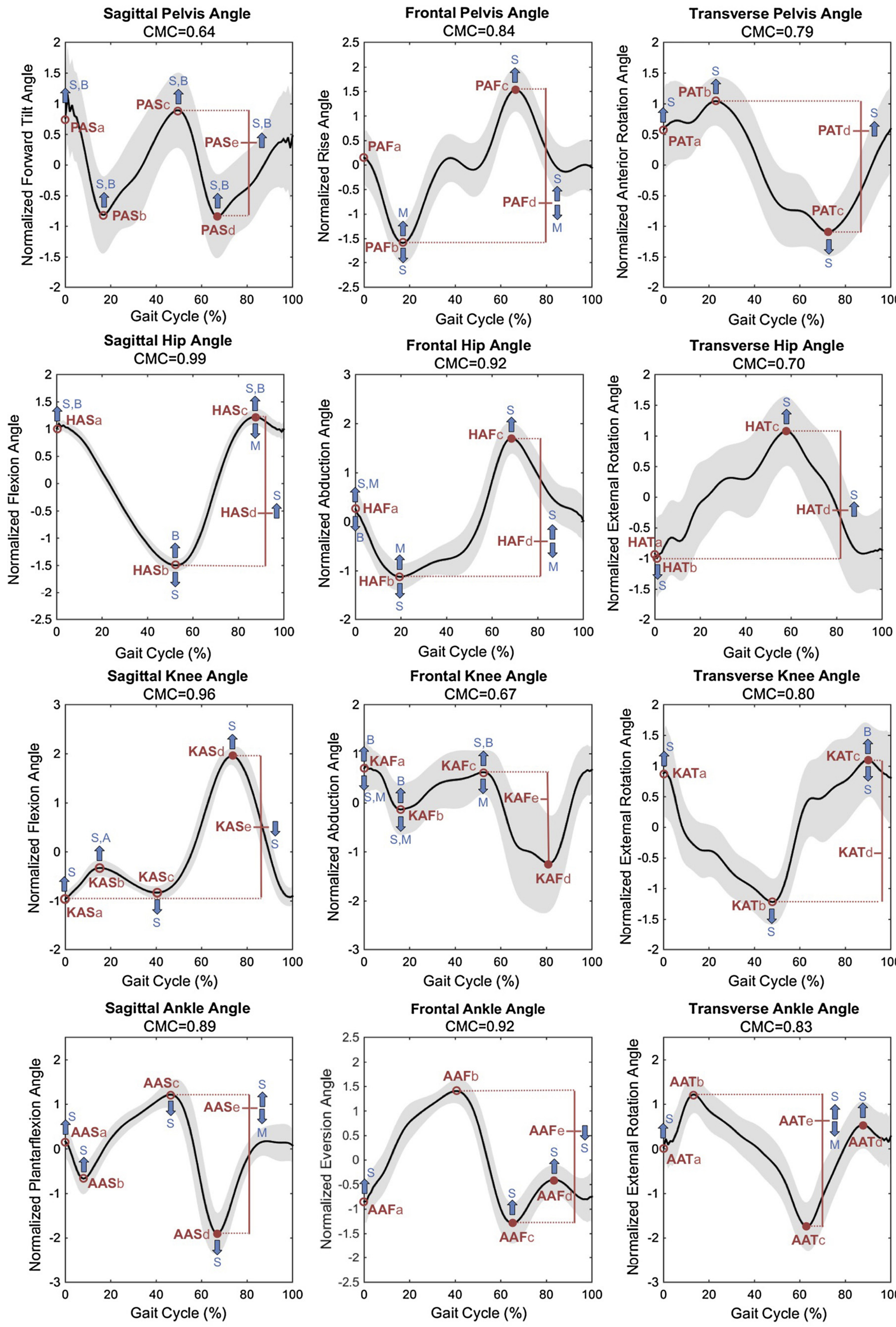

Fig. 2. Average normalized kinematic curves for all subjects at all speeds during the gait cycle. Between-curve standard deviation is shaded; the CMC represents the correlation of the curves between all subjects and speeds. Open points represent discrete features (labeled in red) during stance and filled points represent features during swing; features are labeled using a notation of joint (P/H/K/A), angle (A), plane (S/F/T), and feature (a, b, c, etc.). Blue arrows represent statistically-significant trends related to differences in speed (S), age (A), sex (where 'M' represents the trend for males), and BMI (B). (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

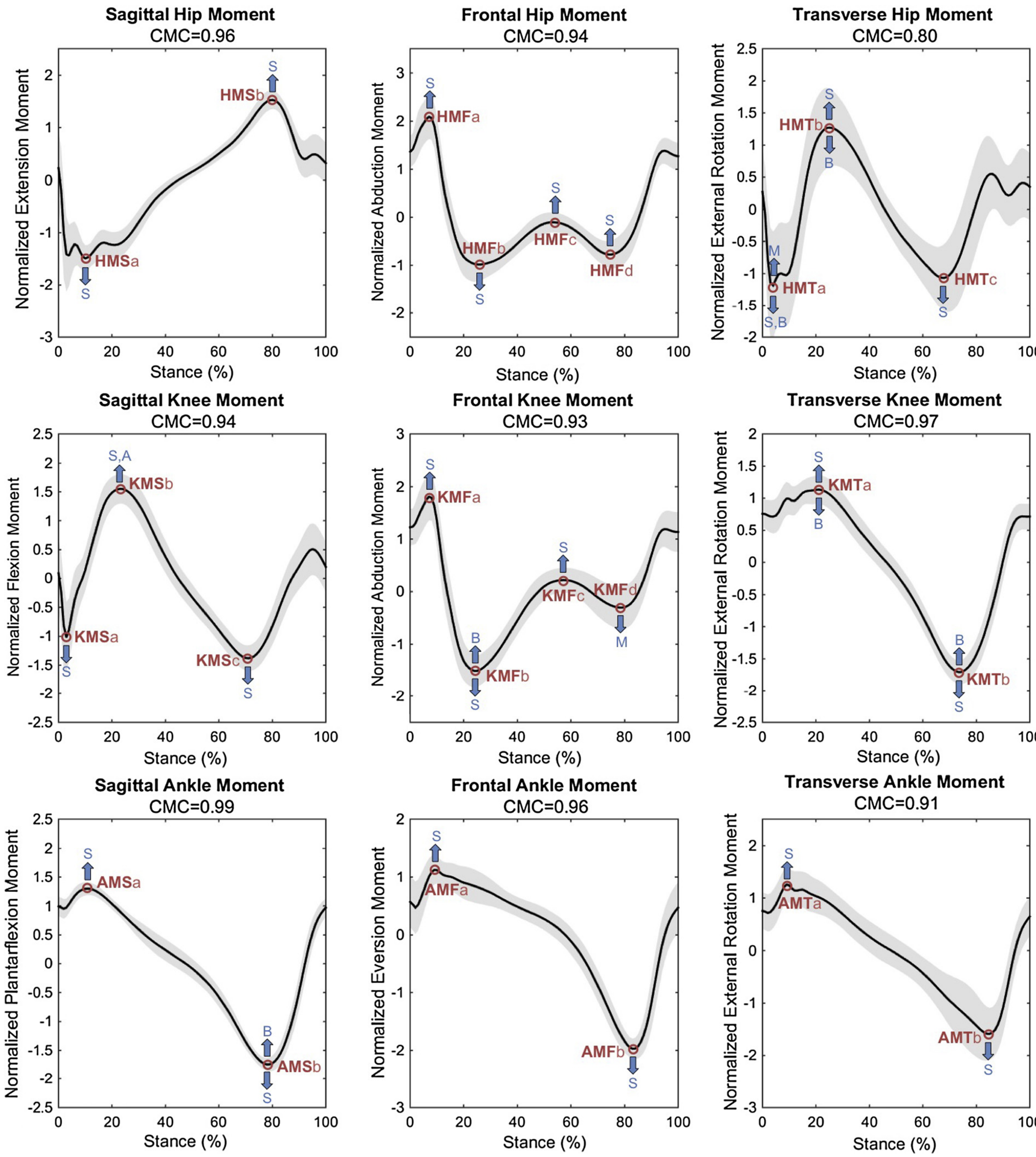

Fig. 3. Average normalized kinetic curves for all subjects at all speeds during stance. Between-curve standard deviation is shaded; the CMC represents the correlation of the curves between all subjects and speeds. Open points represent discrete features (labeled in red); features are labeled using a notation of joint (P/H/K/A), moment (M), plane (S/F/T), and feature (a, b, c, etc.). Blue arrows represent statistically-significant trends related to differences in speed (S), age (A), sex (where 'M' represents the trend for males), and BMI (B). (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

the pelvis (PAFd) and hip (HAFd). It is also worth noting that there was a lack of consistent correlations between sex and gait features in the sagittal and transverse planes.

A main strength of this study was the inclusion of a heterogeneous population and a study design that allows for a comprehensive analysis of the variations in lower limb walking mechanics. This study was able to systematically identify a comprehensive set of characteristic features of the 3-dimensional kinematics and kinetics curves that may be used to compare walking profiles. Furthermore, the standardization of kinematic and kinetic curves of each joint and in each anatomical plane allowed for the identification of features that would exist in any subpopulation of this study. Specifically, the standardization revealed that curves do indeed have characteristic features that may otherwise be obscured by variations in curve offsets and amplitudes between subjects and speeds. For example, pelvis angles during ambulation tend to vary greatly between subjects and speeds, but the standardization of the curves revealed discrete gait features that existed between subjects and walking speeds. This study is based on a rich dataset that could be useful to test other research questions; any investigator interested in further analyzing these data is invited to contact the authors.

This study contains a few limitations, including the use of a cross-sectional design that renders it difficult to identify the causal relationships between variations in demographic measures and variations in gait features. In addition, the large number of correlations tested greatly reduces the statistical threshold for significance with the conservative Bonferroni correction; however, several measures were still found to be significantly correlated with variations in age, sex, and BMI, indicating that such correlations were especially strong. This study excluded subjects at elevated risk for joint disease due to age, rendering these results applicable to asymptomatic subjects under the age of 60. Furthermore, the segment data from Dempster and Gaughran (1967) came from male cadavers with body mass under 72 kg, possibly leading to inertial segment property misestimates for subjects with different distributions in body mass. There may also be side-to-side differences associated with limb dominance (Sadeghi et al., 2000) that this study did not assess. Additionally, one cannot exclude errors related to gait measurement, especially, due to the difficulty of marker placements in overweight or obese individuals. Nevertheless, the large sample size and advanced statistical method in this study should have limited these possible errors. Furthermore, since gait features could be dependent on the method used for data collection and post-processing, the regression coefficients in this study should be used carefully if gait is analyzed with a drastically different method. Possible directions for future work includes comparing gait profiles and demographic effects at an anthropometrically scaled walking speed, for example, by using the Froude number (Moretto et al., 2007), or structuring an analysis to identify the discrete features that most simply characterize each curve for a given subpopulation. Finally, there would be value in future studies that test the effect of speed and demographics in an older population, subjects with pathological conditions such as osteoarthritis or a torn anterior cruciate ligament, as well as the stratification of subjects who are asymptomatic but have signs of joint disease. This would allow for a greater understanding of the effect of disease on gait mechanics independent of walking speed and other demographic measures.

In conclusion, this study provided an identification of characteristic gait features during walking, and was able to correlate many variations in lower limb kinematics and kinetics with walking speed, which can affect almost every aspect of the gait cycle and should therefore be considered in nearly any analysis of ambulation. In addition, this study identified three clusters of gait features where variations were correlated with age, sex, and BMI, suggesting that these demographic measures also play an important role in explaining some variations in gait between people with varying demographic identities. This provides the field of gait analysis with much-needed information that could allow future studies to better understand which gait features are affected by walking speed and when they may also be affected by other demographic measures. This could allow subsequent studies to select populations that are possibly more relevant to a hypothesis and less likely to confound analyses. These results could also provide an important background for analyses that need to test populations heterogenous in demographic measures, or for studies that are unable to perform extensive statistical analyses due to limited sample size. Finally, this study could be especially useful for the identification of specific gait features as clinical markers for musculoskeletal disease, or to assess the potential value of gait-based clinical interventions.

Conflict of interest

The authors do not have any conflicts of interest to disclose.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.jbiomech.2017.04.014.

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