ELSEVIER

Journal of Biomechanics 36 (2003) 1159–1168

JOURNAL OF BIOMECHANICS

www.elsevier.com/locate/jbiomech
www.JBiomech.com

Repeatability of gait data using a functional hip joint centre and a mean helical knee axis

Thor F. Besier, Daina L. Sturnieks, Jacque A. Alderson, David G. Lloyd*

School of Human Movement and Exercise Science, University of Western Australia, 35 Stirling Highway, Crawley, WA 6907, Australia

Accepted 7 February 2003

Abstract

Repeatability of traditional kinematic and kinetic models is affected by the ability to accurately locate anatomical landmarks (ALs) to define joint centres and anatomical coordinate systems. Numerical methods that define joint centres and axes of rotation independent of ALs may also improve the repeatability of kinematic and kinetic data. The purpose of this paper was to compare the repeatability of gait data obtained from two models, one based on ALs (AL model), and the other incorporating a functional method to define hip joint centres and a mean helical axis to define knee joint flexion/extension axes (FUN model). A foot calibration rig was also developed to define the foot segment independent of ALs. The FUN model produced slightly more repeatable hip and knee joint kinematic and kinetic data than the AL model, with the advantage of not having to accurately locate ALs. Repeatability of the models was similar comparing within-tester sessions to between-tester sessions. The FUN model may also produce more repeatable data than the AL model in subject populations where location of ALs is difficult. The foot calibration rig employed in both the AL and FUN model provided an easy alternative to define the foot segment and obtain repeatable data, without accurately locating ALs on the foot.

Keywords: Kinematics; Kinetics; Repeatability; Gait analysis

1. Introduction

Efforts have been made to reduce errors associated with photogrammetric techniques and skin movement artefacts during motion analysis, to accurately determine the position and orientation of body segments. Skin movement artefact has been shown to be reduced by employing a 'CAST' technique, whereby three or more clusters of markers are placed on each segment to create technical coordinate systems (TCSs) (Cappozzo et al., 1995). Anatomical landmarks (ALs) are then defined relative to the TCSs in a static trial to reconstruct an anatomical coordinate system (ACS) during a dynamic trial (Cappozzo et al., 1995; Lucchetti et al., 1998). However, imprecise location of ALs can lead to mislocation of the ACS and subsequent joint centres, which propagates to errors in joint kinematics and kinetics (Holden and Stanhope, 1998; Della Croce et al., 1999; Stagni et al., 2000). Errors associated with the imprecise location of ALs have been noted as the greatest source of error in motion analysis, compared to instrument error or skin movement artefact (Della Croce et al., 1997). These findings raise concerns regarding the repeatability of models using ALs to define an ACS, which from a clinical perspective, is of paramount importance. Methods to reduce any variability in locating ALs and defining the ACS therefore warrant investigation.

Numerical methods can be used to determine joint centres and axes of rotation relative to marker clusters, without the need to accurately locate ALs. Techniques have been previously established to estimate joint centres for the hip (Cappozzo, 1984; Shea et al., 1997; Leardini et al., 1999), and the shoulder (Stokdijk et al., 2000), by moving the joint through a functional range of motion, assuming a true 'ball-and-socket' articulation. Methods have also been developed to determine 'optimal' axes or mean helical axes of rotation for the knee (Boyd and Ronsky, 1998; Churchill et al., 1998) and elbow (Chéze et al., 1998; Stokdijk et al., 1999)

*Corresponding author. Tel.: +61-9-380-3919; fax: +61-9-380-1039.

E-mail address: dlloyd@cyllene.uwa.edu.au (D.G. Lloyd).

0021-9290/03/$ - see front matter
doi:10.1016/S0021-9290(03)00087-3

throughout a dynamic range of motion. Defining joint centres and axes of rotation using these functional methods may improve the repeatability of kinematic and kinetic data, compared to traditional methods that rely on the accurate location of ALs to define an ACS.

The purpose of this paper was to compare within-tester and between-tester repeatability of gait data using two different methods of defining a lower limb ACS. The first model utilizes a functional hip joint centre (HJC) and a mean helical axis of the knee (FUN model), while the second model uses traditional ALs to define an ACS (AL model). It was hypothesized that the FUN lower limb model would demonstrate greater repeatability of kinematic and kinetic gait data than an AL model for both between-tester and within-tester conditions. Furthermore, in an effort to reduce the error associated with locating ALs on the foot (Della Croce et al., 1997), a new technique is demonstrated to define the orientation of the foot segment for gait analysis.

2. Methods

2.1. Subjects

Ten able-bodied subjects (6 males, 4 females; body mass 55–89 kg; height 1.64–1.90 m) participated in the study. Written informed consent was obtained prior to participation, as per requirement of the University of Western Australia ethics committee. Subjects had no current musculoskeletal injury or disease and were free of pain.

2.2. Gait analysis protocol

Gait analyses were performed on each subject in the morning and afternoon of the same day. Two examiners performed separate gait analyses in the morning (for between-tester comparisons), and one of these examiners performed a repeat analysis in the afternoon, at least 4 h following the first session (for within-tester comparisons). This procedure resulted in three gait analyses for each subject. Five examiners in total were used, three of which performed the test–retest sessions for within-tester comparisons.

A six-camera VICON motion analysis system (Oxford Metrics, Oxford, UK) was used in conjunction with two AMTI force-plates (AMTI, Watertown, MA) to collect motion data (50 Hz) and ground reaction force data (2000 Hz), respectively. Marker coordinate data were filtered using a GCVSPL routine (Woltring, 1986), which helped to reduce the error in reconstructing the helical axes parameters (de Lange et al., 1990) and in performing inverse dynamic calculations. The 7-segment FUN and AL kinematic/kinetic models were constructed using BodyBuilder software (Oxford Metrics, Oxford, UK). Gait data were retrieved using custom software and normalized to 51 points over stride using a cubic spline within MATLAB (Mathworks, Natick, MA).

During each session, subjects walked at a self-selected pace (observed speeds were between 1.2 and 1.5 m/s) across an instrumented 10 m walkway. A minimum of six trials with successful force-plate strikes were captured for both left and right legs. In addition to the walking trials, subjects performed a series of static and dynamic calibration trials to locate ALs and estimate the location of the HJCs, knee joint centres (KJC) and functional flexion/extension axes of each knee.

2.3. Marker set and definitions of segment and joint coordinate systems

To determine the three-dimensional position and orientation of each lower limb segment, clusters of three retro-reflective markers (20 mm) were firmly adhered to the subject's pelvis, thighs, shank, and feet (Fig. 1B). A TCS was defined using each thigh, shank, and foot segment cluster such that the ACS and joint centres could be defined relative to these TCSs. The pelvis, femoral, and tibial ACSs were defined the same as those

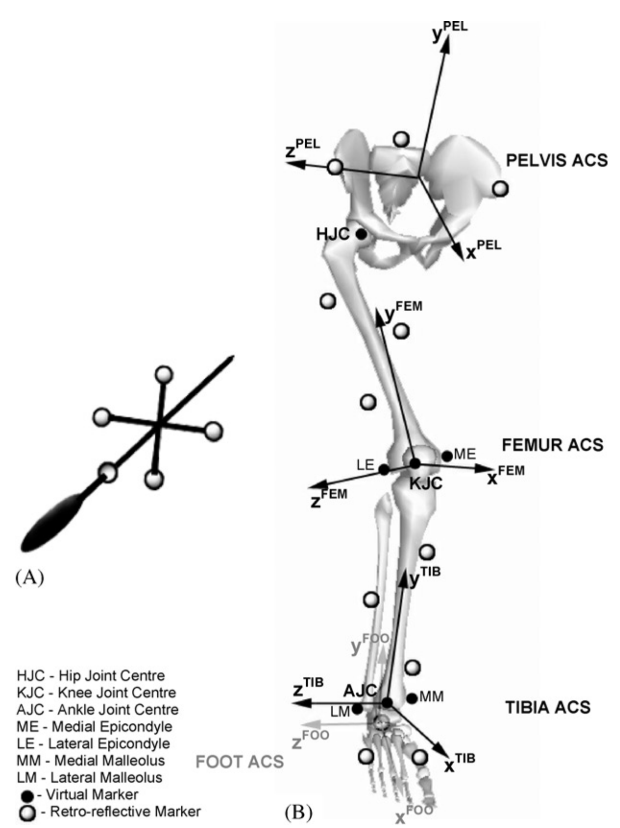

Fig. 1. (A) Pointer device used to locate the medial and lateral epicondyles of the femur. (B) ACSs of the lower limb model. Figure produced using SIMM (Musculographics Inc., Evanston, IL).

used by Kadaba et al. (1989), although a new ACS was established for the foot.

The pelvis ACS was defined using an origin midway between the antero-superior iliac spines (ASISs), a positive z-axis along the line of the left ASIS to the right ASIS, an x-axis along a line from the sacrum marker to the origin (positive being anterior), and a y-axis orthogonal to the $x-z$-plane (positive being superior) (Fig. 1B).

Two different methods were used to determine the HJCs. For the AL model, HJCs were defined relative to the pelvis ACS and estimated using a regression equation developed by Shea et al. (1997). For the FUN model, a functional method similar to that used by Piazza et al. (2001) was employed, whereby subjects were required to consecutively move the right and left thigh through a range of flexion, abduction, adduction, and extension (Piazza et al., 2001). These data were used in a constrained optimization program written in MATLAB (*Optimization Toolbox*, Mathworks Inc.; Natick, MA), where spheres were fit to each thigh marker to find a left and right HJC location relative to the pelvis ACS (x_p, y_p, z_p) and sphere radii. An initial estimation was obtained using the regression equation of Shea et al. (1997) and the optimization was constrained to be within a 100 mm cube surrounding this position. The location of the initial estimate of the HJC coordinates was randomly perturbed within the 100 mm cube during six consecutive optimizations to avoid finding local minima.

In four static trials, the position of the medial and lateral femoral epicondyles (ME and LE, respectively) of each femur were measured. A pointer with five markers was used to locate the ME and LE, with marker redundancy used to reduce error in locating the end of the pointer (Fig. 1A).

The femoral ACS, KJCs, and knee flexion/extension axis were defined differently for the AL and FUN models. In the AL model, the femoral ACS was defined using an origin at the KJC (midway between the ME and LE), a y-axis as the line passing through the KJC to the HJC (positive being superior), a z-axis being along a plane defined by the ME and LE and orthogonal to the y-axis (positive pointing from left to right), and an x-axis orthogonal to the $y-z$-axes (positive being anterior) (Fig. 1B).

For the FUN model, a mean helical axis was used to define the KJC and flexion/extension axis of each knee. To achieve this, subjects stood on one leg and flexed the contra-lateral thigh to enable the shank to freely flex and extend about the knee from full extension to $\sim 100°$ of flexion. This was performed for at least three cycles for each limb. Using the custom MATLAB program, the tibia markers were expressed in the femoral TCS and instantaneous helical axes calculated throughout the range of motion using a singular value decomposition method from Spoor and Veldpaus (1980) and Reinschmidt and van den Bogert (1997). A mean helical axis was then calculated for each knee, analogous to the method presented by Stokdijk et al. (1999) for the elbow joint. The mean helical axis was used to define the flexion/extension axis of the knee relative to the thigh TCS. The KJC in the FUN model was defined relative to the mean helical axis, at a point along the helical axis that intersected a plane that was normal to the transepicondylar line, midway between the epicondyles. As such, the location of the ME and LE in the FUN model were only used to define the medial and lateral side of the knee. The femoral ACS in the FUN model was then defined similar to the AL model, except using the FUN KJC and flexion/extension axis.

The tibial ACS was determined using the position of markers placed on the medial malleoli (MM) of the tibiae and lateral malleoli (LM) of the fibulae, collected during a static trial. Each tibial ACS was defined using an origin at the AJC (midway between the MM and LM), a y-axis as the line passing from the AJC to the KJC (positive being superior), a z-axis being along a plane defined by the flexion/extension axis of the knee, and an x-axis orthogonal to the $y-z$-axes (positive being anterior) (Fig. 1B). Subsequently, the tibial ACS was different for the AL and FUN models due to the respective difference in KJC and knee flexion/extension axis definitions.

The foot segment was defined the same way for both AL and FUN models. To overcome large errors in palpating and placing markers on ALs of the foot (Della Croce et al., 1997), an alignment rig was developed to define the orientation of the foot segment (Fig. 2). A TCS was defined for the alignment rig using four retro-reflective markers attached to the rig, such that any foot

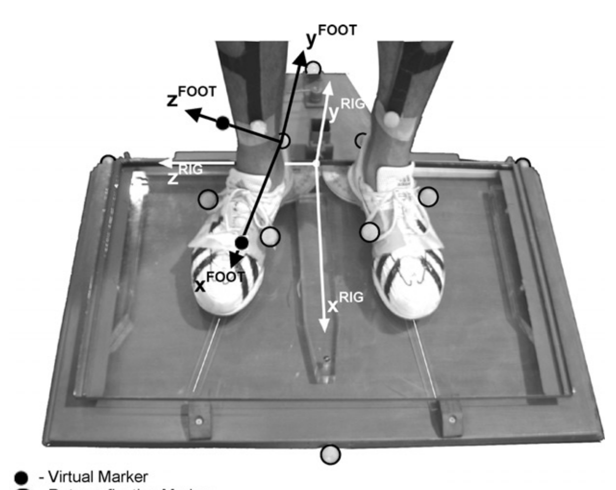

Fig. 2. Foot calibration rig and rig TCS. Note the goniometer beneath the subjects' foot to measure foot progression (abduction/adduction) relative to y^{RIG}. Rear-foot inversion/eversion was also taken whilst the subject was on the rig using an inclinometer.

measurements were made with respect to this coordinate system (Fig. 2). The subject stood on the alignment rig in a comfortable stance with their heels against a small metal plate at the rear of the rig. The long axis (x) of the foot segment was assumed to be parallel to the $x-z$ (horizontal) plane of the rig coordinate system (Fig. 2). It was assumed that the x-axis of the foot was rotated around the rig y-axis. This rotation was measured using a goniometer fixed to the alignment rig (Fig. 2), which was defined by the line bisecting the calcaneus and the midpoint between the 2nd and 3rd metatarsal heads. The foot was then assumed to be rotated in inversion/eversion about the x-axis of the foot, which was a rear-foot angle measured relative to the $x-z$ plane of the rig, perpendicular to the x-axis of the foot. This rear-foot inversion/eversion angle was taken using an inclinometer (Dasco Pro Inc., Rochford, IL). These sequences of rotations, in accordance with the ISB standard (Wu and Cavanagh, 1995), were used to define the foot ACS, from which two virtual markers were then created and expressed relative to the foot TCS (Fig. 2). In subsequent dynamic trials, the calcaneus-marker and the two virtual markers were used to define the foot ACS.

The convention used to describe the kinematics of the hip, knee, and ankle joints followed the ISB standard (Wu and Cavanagh, 1995). The sequence of rotations was: flexion/extension about the z-axis of the proximal segment; then adduction/abduction about a floating x-axis; followed by internal/external rotation about the y-axis of the distal segment. Joint kinetics were expressed in the ACS of the distal segment.

2.4. Statistics

Coefficients of Multiple Determination (CMD or r^2, Kadaba et al., 1989) were calculated between testing sessions 1–3, 1–2, and 2–3 using time normalized kinematic and kinetic curves. The CMD was reported, as opposed to the Coefficient of Multiple Correlation (CMC, or r), as an r^2 value indirectly refers to the percentage variance accounted for within the data. In this fashion, repeatability of each model was measured, between testers and within tester.

The systematic error of the two models was determined similar to the 'static daily offset' calculated by Kadaba et al. (1989) and Growney et al. (1997), which occurs due to the re-application of markers. This error term was calculated using the following:

$$\text{systematic error} = \frac{1}{N}\sum_{n=1}^{N}\sqrt{(x1_n - x2_n)},$$

where N is the number of datapoints, $x1_n$ is the mean of variable x at time point n from testing session 1, $x2_n$ is the mean of variable x at time point n from testing session 2. This error term was also calculated using the mean gait variables of testing sessions 1–3, and 2–3, thus giving an indication of the systematic error within tester and between testers.

Two-factor ANOVA's (model type: AL or FUN $\times$ examiner: within-tester or between-tester factor) with repeated measures on subject were then used to compare the CMD values and systematic error between models and examiner conditions ($p<0.05$). Tests were performed independently on data from each limb.

3. Results

3.1. Joint kinematics

Both AL and FUN models produced highly repeatable sagittal plane kinematic data with $r^2 > 0.85$ (Table 1). Frontal plane kinematics were highly repeatable for the hip joint ($r^2 > 0.85$), moderately repeatable for the knee (r^2 between 0.54 and 0.74) and less repeatable for the ankle ($r^2 > 0.42$). Rotations in the transverse plane were quite repeatable at the knee ($r^2 > 0.64$) and ankle ($r^2 > 0.53$) and less repeatable at the hip (r^2 between 0.28 and 0.47).

There were no significant interactions between model type and examiner conditions, except for left knee internal rotation kinematics of the left leg. This generally suggests that repeatability of the gait data was independent of the examiner administering the test (Tables 1 and 2).

The FUN model demonstrated better repeatability in frontal plane knee kinematics than the AL model, with an average r^2 of 0.73 compared to 0.57 (Table 1), although this result was only significant for the comparison of the right leg ($p<0.05$). Similarly, the FUN model produced greater repeatability of internal/external rotation of the knee compared to the AL model (mean r^2 of 0.74 and 0.67, respectively), but did not reach significance for individual limb comparisons (Table 1).

Repeatability of ankle joint kinematic and kinetic data was similar within tester and between testers, for both AL and FUN models (Tables 1 and 2), which was expected as the calibration rig was used to define the foot segment for both models. Ankle dorsi/plantar flexion angles were most repeatable with a mean r^2 of 0.89, whereas ankle inversion/eversion and abduction/adduction angles were moderately repeatable, with mean r^2 of 0.47 and 0.59, respectively.

3.2. Joint moments

Sagittal plane moment data were highly repeatable for the hip, knee, and ankle, with CMD's ranging from 0.81 to 0.97 (Table 2). Hip and knee joint moments in the frontal plane were also very repeatable with r^2 values

Table 1
Coefficients of multiple determination (CMD, r^2) of kinematic data using both anatomical landmark (AL) and functional (FUN) models

Joint angle	AL (left)	FUN (left)	AL (right)	FUN (right)
CMD between testers				
Hip flexion/extension	0.944±0.030	0.950±0.027	0.957±0.021	0.962±0.016
Knee flexion/extension	0.939±0.032	0.945±0.032	0.962±0.021	0.964±0.021
Ankle dorsi/plantar flexion	0.855±0.046	0.861±0.046	0.895±0.030	0.902±0.027
Hip abduction/adduction	0.846±0.078	0.848±0.094	0.862±0.050	0.881±0.038
Knee varus/valgus	0.635±0.272	0.606±0.252	0.541±0.205*	0.742±0.166
Ankle abduction/adduction	0.499±0.240	0.505±0.245	0.487±0.259	0.421±0.249
Hip internal/external rot^n	0.351±0.227	0.354±0.261	0.374±0.228	0.332±0.210
Knee internal/external rot^n	0.713±0.133	0.699±0.159	0.687±0.164	0.764±0.149
Ankle inversion/eversion	0.592±0.190	0.539±0.197	0.634±0.194	0.577±0.221
CMD within tester				
Hip flexion/extension	0.950±0.038	0.956±0.032	0.967±0.014	0.969±0.013
Knee flexion/extension	0.926±0.046	0.934±0.042	0.961±0.025	0.962±0.026
Ankle dorsi/plantar flexion	0.848±0.070	0.869±0.057	0.900±0.040	0.898±0.038
Hip abduction/adduction	0.871±0.056	0.851±0.077	0.846±0.089	0.868±0.069
Knee varus/valgus	0.642±0.254	0.679±0.159	0.589±0.156*	0.714±0.238
Ankle abduction/adduction	0.495±0.277	0.463±0.225	0.443±0.294	0.484±0.238
Hip internal/external rot^n	0.386±0.160	0.397±0.301	0.280±0.202	0.465±0.230
Knee internal/external rot^n	0.686±0.138	0.764±0.093	0.637±0.209	0.720±0.212
Ankle inversion/eversion	0.561±0.237	0.531±0.229	0.669±0.171	0.611±0.210

Note: Significant differences between AL and FUN model are denoted by *($p<0.05$).

Table 2
Coefficients of Multiple Determination (CMD, r^2) of joint moment data using both Anatomical Landmark (AL) and Functional (FUN) models

Joint moment	AL (left)	FUN (left)	AL (right)	FUN (right)
CMD between testers				
Hip flexion/extension	0.896±0.036	0.895±0.048	0.918±0.032	0.919±0.027
Knee flexion/extension	0.810±0.111	0.815±0.068	0.821±0.082	0.837±0.084
Ankle dorsi/plantar flexion	0.951±0.024	0.951±0.024	0.968±0.010	0.968±0.010
Hip abduction/adduction	0.885±0.042*	0.904±0.044	0.899±0.036*	0.933±0.020
Knee varus/valgus	0.826±0.090	0.801±0.125	0.804±0.096	0.769±0.149
Ankle abduction/adduction	0.365±0.236	0.371±0.240	0.534±0.269	0.535±0.269
Hip internal/external rot^n	0.621±0.131	0.643±0.127	0.683±0.081	0.711±0.084
Knee internal/external rot^n	0.747±0.097	0.736±0.096	0.763±0.126	0.748±0.135
Ankle inversion/eversion	0.622±0.242	0.624±0.240	0.704±0.193	0.704±0.193
CMD within tester				
Hip flexion/extension	0.887±0.039	0.903±0.028	0.916±0.027	0.920±0.022
Knee flexion/extension	0.810±0.103	0.831±0.062	0.813±0.095	0.823±0.091
Ankle dorsi/plantar flexion	0.945±0.025	0.946±0.026	0.964±0.013	0.964±0.013
Hip abduction/adduction	0.857±0.066*	0.910±0.032	0.906±0.025*	0.937±0.013
Knee varus/valgus	0.787±0.097	0.800±0.120	0.792±0.077	0.750±0.105
Ankle abduction/adduction	0.325±0.213	0.338±0.216	0.558±0.175	0.557±0.177
Hip internal/external rot^n	0.621±0.131	0.652±0.122	0.666±0.082	0.701±0.074
Knee internal/external rot^n	0.720±0.090	0.717±0.095	0.731±0.118	0.710±0.128
Ankle inversion/eversion	0.633±0.226	0.639±0.231	0.705±0.261	0.703±0.260

Note: Significant differences between AL and FUN models are denoted by *($p<0.05$).

ranging between 0.75 and 0.94. Internal/external rotation moments at the hip and knee were slightly less repeatable than ab/adduction moments, with mean r^2 of 0.66 and 0.73 Ankle joint moments had similar repeatability to the hip and knee joint for inversion/eversion, with a mean r^2 of 0.67, but displayed less repeatability abduction/adduction moments (mean r^2 of 0.48, Table 2).

The FUN model produced more repeatable hip moments than the AL model in the frontal plane, both between testers and within tester (FUN $r^2 = 0.92$; AL $r^2 = 0.89$, $p<0.05$) (Table 2). The FUN model also produced slightly more repeatable hip moments than the AL model in the transverse plane (mean r^2 of 0.68 and 0.65, respectively), although these were not statistically significant. As expected, ankle joint moments displayed similar repeatability between models, within tester and between testers.

Although the curves were repeatable for both the AL and FUN models, differences in the magnitude of the kinematic and kinetic data collected with both models were observed (Figs. 3 and 4). These differences were due to the altered location of the hip and KJCs and axes of rotation between models, and mostly affected data in the frontal and transverse plane. The magnitude of the ankle joint moments were not affected by the model employed (Fig. 4).

The systematic error due to the re-application of markers was similar for both AL and FUN models, with an average difference of $\sim 3.8°$ and ~ 0.03 Nm/kg between testing sessions for joint angle and moment data, respectively. Systematic error within tester was similar to the error between testers (Figs. 3 and 4).

4. Discussion

It was hypothesized that the numerical definition of joint centres and axes of rotation in the FUN model would improve the repeatability of kinematic and kinetic data, compared to traditional methods that rely on the location of ALs to define joint centres and axes of rotation (AL model). This hypothesis was tested using within-tester and between-tester conditions. Furthermore, in an effort to reduce the error associated with locating ALs on the foot (Della Croce et al., 1997), a new technique to define the orientation of the foot segment was investigated.

Both models presented in this paper produced highly repeatable frontal and transverse plane kinematic data compared to previous studies (Kadaba et al., 1989; Growney et al., 1997). For example, Kadaba et al. (1989) and Growney et al. (1997) reported between-day CMD values of ~ 0.40 for knee varus/valgus and ~ 0.24 for knee internal/external rotation angles. Repeatability of hip internal/external rotation angles measured between-days by Kadaba et al. (1989) and Growney et al. (1997) were also low, with average r^2 of 0.2 and 0.42, respectively. Reasons for such low between-day repeatability of non-sagittal plane angles included skin movement artefact and marker re-application, coupled with the effects of 'downstream' errors associated with Euler angle calculations (Growney et al., 1997). The use of marker clusters in the AL and FUN model appears to improve the repeatability of non-sagittal plane kinematic data, which may be partly due to reduced skin movement artefact. Furthermore, the AL and FUN model do not appear to have the same propensity for error from marker re-application as the model of Kadaba et al. (1989) and Growney et al. (1997). The use of mid-segment markers to define joint centres and axes of rotation for the knee and ankle in previous models may explain the low repeatability in non-sagittal plane data compared to the AL and FUN model presented here. An investigation is currently underway to directly compare the model employed by Kadaba et al. (1989) and Growney et al. (1997) with the current models.

With regard to the accuracy of frontal and transverse plane rotations using these two models, it is difficult to ascertain whether our results reflect the repeatability of segmental rotations, or the repeatability of motion artefact, for example, due to skin movement from underlying muscle activation or impact loading of the foot at heel strike. However, the knee varus/valgus angles from both models compare favourably to that of Lafortune et al. (1992), who used high speed motion capture and cortical pins embedded within the femur and tibia to measure knee joint rotations (see dashed lines in Fig. 3). Knee internal/external rotation kinematics were also similar in shape to that of Lafortune et al. (1992), although a difference in the ACS used by us and Lafortune et al. (1992) may be responsible for the obvious offset (Fig. 3). The difference in the ACSs may have also caused our models internal/external and varus/valgus rotations during swing to be slightly larger than that reported by Lafortune et al. (1992), possibly due to cross-talk from knee flexion/extension (Piazza and Cavanagh, 2000). Skin movement may also be responsible for these differences. Similar effects may be evident in the frontal and transverse plane rotations at the hip and ankle joints.

The FUN model produced slightly more repeatable gait curves than the AL model, in knee joint kinematics and hip joint moments, thus supporting the hypothesis of this study. However, the differences in repeatability between the AL and FUN model were not as predominant as expected. The FUN model was also expected to be more repeatable than the AL model in the between-tester condition, as the FUN model is not as reliant upon accurate location of ALs to define joint centres and axes of rotation. However, the AL and FUN models produced similar repeatability of gait data within tester as they did between testers (r^2 and systematic error), suggesting that both models produce repeatable results, regardless of whether the same examiner performs the test–retest gait analysis.

Several factors might explain the similarities in the repeatability of gait data between the AL and FUN model, and the repeatability of each model, within tester and between testers. Perhaps most importantly, the examiners performing the gait analyses in this study were all very experienced in locating lower limb ALs, such as the ASISs of the pelvis and epicondyles of the femur. An interesting analysis would be to compare the gait data from both models collected by an experienced and a novice examiner. Similar gait data would be expected between-examiners using the FUN model,

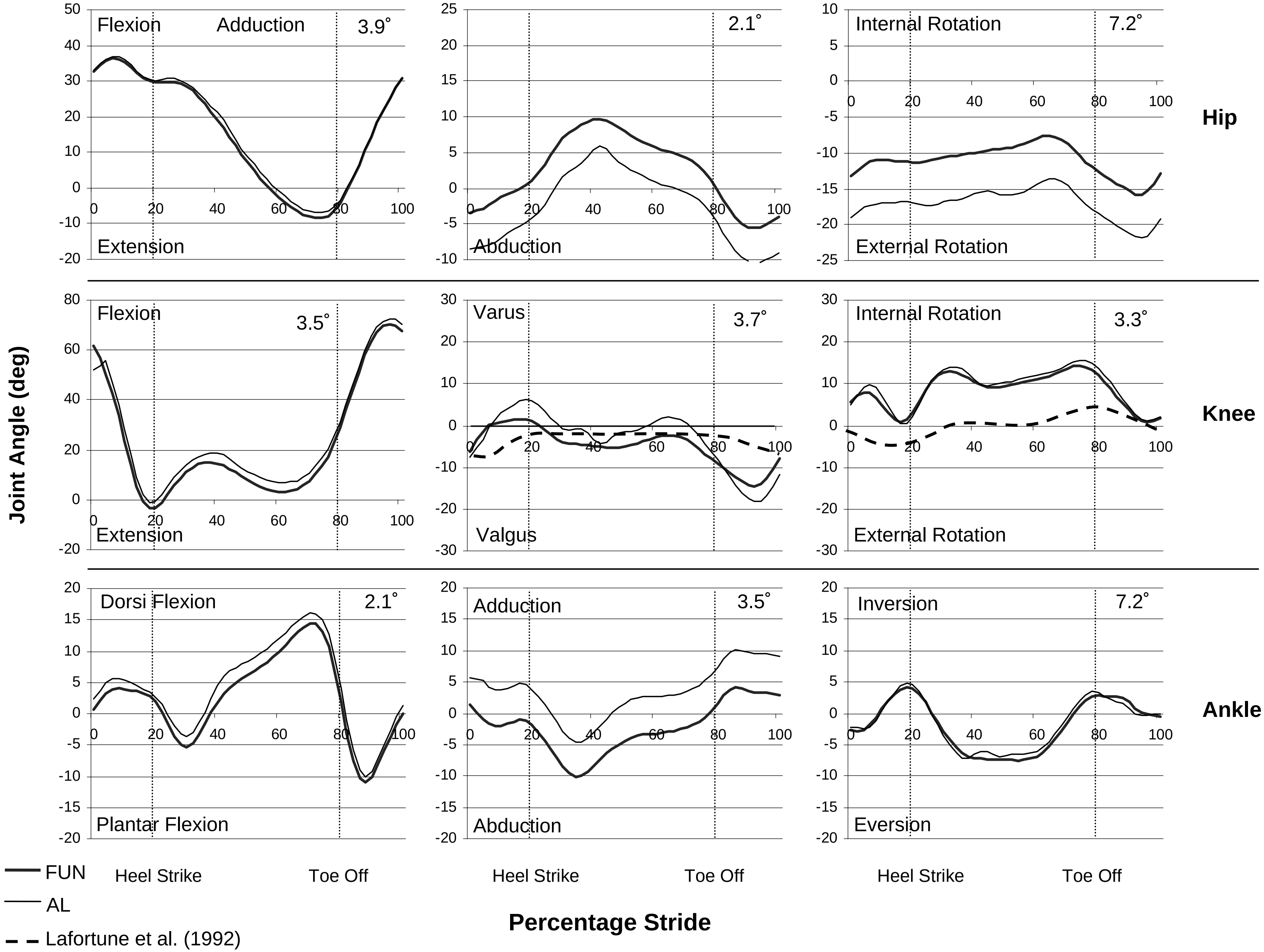

Fig. 3. Typical joint kinematic data obtained from a single gait analysis session using both AL and Optimized (FUN) joint models. The number on each graph indicates the average systematic error (in degrees) between gait sessions (between testers and within tester) of both models.

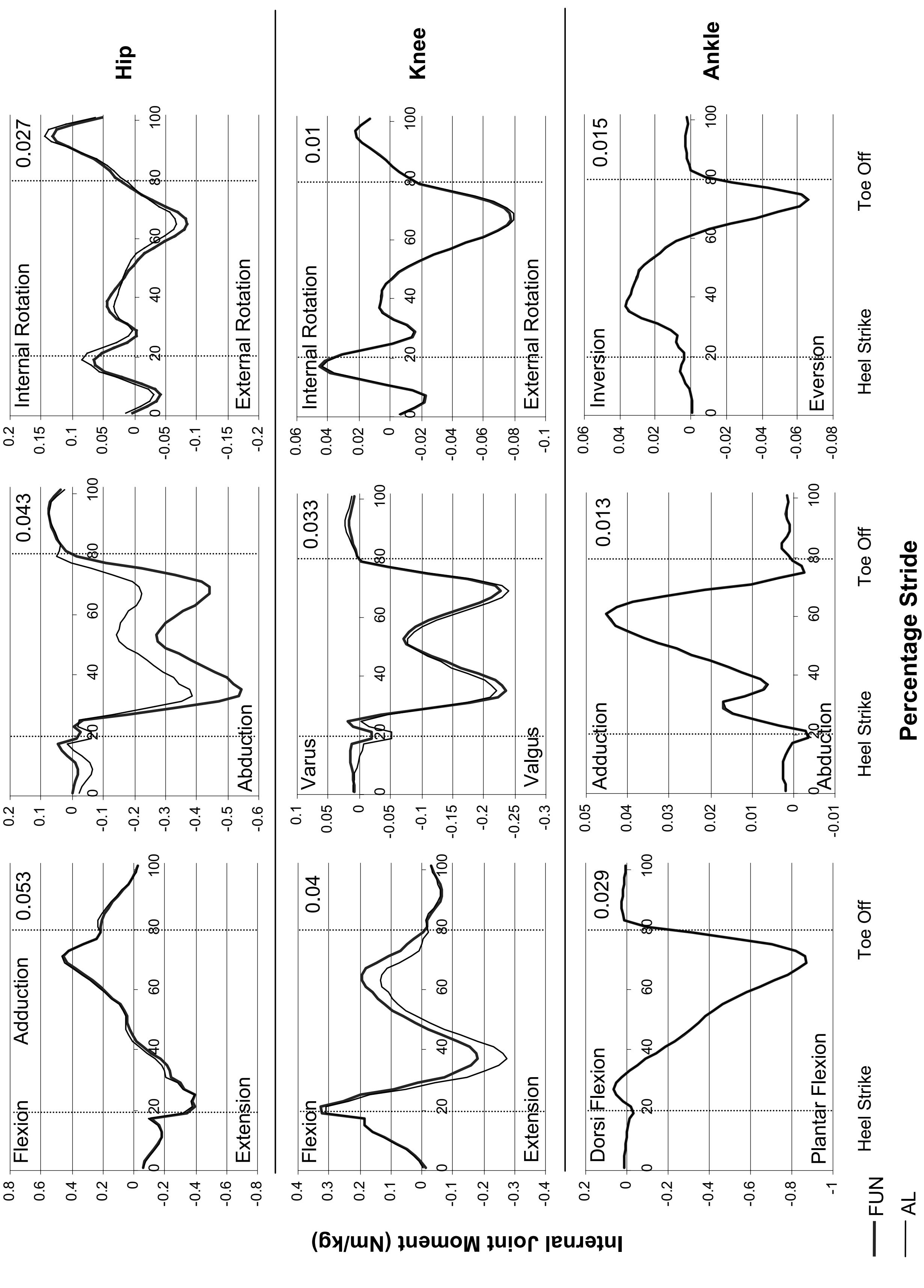

Fig. 4. Typical joint moment data obtained from a single gait analysis session using both AL and Optimized (FUN) joint models. The number on each graph indicates the average systematic error (in Nm/kg) between gait sessions (between testers and within tester) of both models.

whereas incorrect location of ALs might lead to significant differences in the gait data between-examiners using the AL model. Furthermore, the subjects in this study were young and healthy, with minimal body fat, which would improve the ability to locate ALs, particularly on the pelvis. In subject populations where palpation of pelvis ALs becomes difficult, the FUN model would be expected to produce more repeatable gait data than the AL model between testing sessions, as it does not rely on the accurate location of pelvis markers to define the HJCs. The FUN model might also produce more repeatable knee joint kinematics and kinetics than the AL model in subjects who have bony deformities of the knee joint, where location of the epicondyles of the femur might prove difficult.

Della Croce et al. (1997) suggested that the largest error in locating ALs of the lower limb occur at the foot, which might have a negative effect on the repeatability of ankle kinematic and kinetic data. Using ALs to define the foot segment, Growney et al. (1997) reported an average CMD of 0.41 for ankle inversion/eversion angles and 0.23 for ankle abduction/adduction angles. Hence, the foot calibration rig was developed to define the foot segment using clinical measures of rear-foot inversion/eversion, and foot abduction/adduction, rather than relying on the accurate location of ALs. Using the calibration rig to define the foot segment resulted in much higher r^2 values than those reported previously (Growney et al., 1997) for ankle inversion/eversion and abduction/adduction angles and moments.

Although the repeatability of the waveform data and systematic error were quite similar between AL and FUN models, differences in the magnitude of the joint angle and moment data (offsets) were still observed (Figs. 3 and 4). These were due to the different locations of the hip and KJCs in each model, and the different flexion/extension axes defined for the knee joint. The question must then be asked, which model provides the most clinically representative data? Previous investigations have shown the functional method of locating HJCs to be superior to definitions based on regression equations or models of 'best-fit' (Leardini et al., 1999). Therefore, the FUN model is likely to provide more accurate HJCs compared to the AL model. It is more difficult to ascertain the differences between the knee flexion/extension axes defined by the transepicondylar axis or a mean helical axis. Churchill et al. (1998) found no difference between an 'optimal' knee flexion/extension axis and the transepicondylar axis, which would suggest that either model of defining the knee axis would produce similar kinematic and kinetic data. However, the definition of the transepicondylar axis requires the accurate location of the epicondyles of the femur, which may be problematic given inexperienced examiners or knees with bony deformities. Although the helical axis provides an accurate and convenient frame of reference to describe motion of one segment relative to another, care should be taken when using it to describe pathological joint motion. Abnormal knee kinematics, for example, may not be easily identified from time histories of knee joint angles using a mean helical axis, as the orientation of the helical axis may change relative to the anatomy.

In summary, both models produced highly repeatable lower limb gait data compared to previous models (Kadaba et al., 1989; Growney et al., 1997). The FUN model produced slightly more repeatable kinematic and kinetic data than the AL model, with the advantage of not having to accurately locate bony landmarks. Improved repeatability of the FUN model compared to the AL model might be more pronounced with examiners who are less experienced, in overweight subject populations, where location of bony landmarks becomes more difficult, or with subjects who have large bony deformities at the knee. These issues require further investigation. The foot calibration rig provides an easy alternative to define the foot segment, without accurately locating ALs of the foot, producing repeatable results.

Acknowledgements

We would like to acknowledge the financial support of the Australian Football League (AFL) and the Australian National Health and Medical Research Council (NHMRC, ID# 991134).

References

Boyd, S.K., Ronsky, J.L., 1998. Instantaneous moment arm determination of the cat knee. Journal of Biomechanics 31, 279–283.

Cappozzo, A., 1984. Gait analysis methodology. Human Movement Science 3, 27–54.

Cappozzo, A., Catani, F., Della Croce, U., Leardini, A., 1995. Position and orientation in space of bones during movement: anatomical frame definition and determination. Clinical Biomechanics 10, 171–178.

Chèze, L., Fregly, B.J., Dimnet, J., 1998. Determination of joint functional axes from noisy marker data using the finite helical axis. Human Movement Science 17, 1–15.

Churchill, D.L., Incavo, S.J., Johnson, C.C., Beynnon, B.D., 1998. The transepicondylar axis approximates the optimal flexion axis of the knee. Clinical Orthopaedics and Related Research 356, 111–118.

de Lange, A., Huiskes, R., Kauer, J.M.G., 1990. Effects of data smoothing on the reconstruction of helical axis parameters in human joint kinematics. Journal of Biomechanical Engineering 112, 107–113.

Della Croce, U., Cappozzo, A., Kerrigan, D.C., Lucchetti, L., 1997. Bone position and orientation errors: pelvis and lower limb anatomical landmark identification reliability. Gait & Posture 5, 156–157.

Della Croce, U., Cappozzo, A., Kerrigan, D.C., 1999. Pelvis and lower limb anatomical landmark calibration precision and its propagation to bone geometry and joint angles. Medical & Biological Engineering & Computing 37, 155–161.

Growney, E., Meglan, D., Johnson, M., Cahalan, T., An, K., 1997. Repeated measures of adult normal walking using a video tracking system. Gait & Posture 6, 147–162.

Holden, J.P., Stanhope, S.J., 1998. The effect of variation in knee center location estimates on net knee joint moments. Gait & Posture 7, 1–6.

Kadaba, M.P., Ramakrishnan, H.K., Wootten, M.E., Gainey, J., Gorton, G., Cochran, G.V., 1989. Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. Journal of Orthopaedic Research 7, 849–860.

Lafortune, M.A., Cavanagh, P.R., Sommer, H.J., Kalenak, A., 1992. Three-dimensional kinematics of the human knee during walking. Journal of Biomechanics 25, 347–357.

Leardini, A., Cappozzo, A., Catani, F., Toksvig-Larsen, S., Petitto, A., Sforza, V., Cassanelli, G., Giannini, S., 1999. Validation of a functional method for the estimation of hip joint centre location. Journal of Biomechanics 32, 99–103.

Lucchetti, L., Cappozzo, A., Cappello, A., Della Croce, U., 1998. Skin movement artefact assessment and compensation in the estimation of knee-joint kinematics. Journal of Biomechanics 31, 977–984.

Piazza, S.J., Cavanagh, P.R., 2000. Measurement of the screw-home motion of the knee is sensitive to errors in axis alignment. Journal of Biomechanics 33, 1029–1034.

Piazza, S.J., Okita, N., Cavanagh, P.R., 2001. Accuracy of the functional method of hip joint center location: effects of limited motion and varied implementation. Journal of Biomechanics 34, 967–973.

Reinschmidt, C., van den Bogert, T., 1997. KineMat: a MATLAB Toolbox for Three-Dimensional Kinematic Analyses. Human Performance Laboratory, Calgary.

Shea, K.M., Lenhoff, M.W., Otis, J.C., Backus, S.I., 1997. Validation of a method for location of the hip joint center. Gait & Posture 5, 157–158.

Spoor, C.W., Veldpaus, F.E., 1980. Rigid body motion calculated from spatial coordinates of markers. Journal of Biomechanics 13, 391–393.

Stagni, R., Leardini, A., Cappozzo, A., Grazia, B.M., Cappello, A., 2000. Effects of hip joint centre mislocation on gait analysis results. Journal of Biomechanics 33, 1479–1487.

Stokdijk, M., Meskers, C.G., Veeger, H.E., de Boer, Y.A., Rozing, P.M., 1999. Determination of the optimal elbow axis for evaluation of placement of prostheses. Clinical Biomechanics 14, 177–184.

Stokdijk, M., Nagels, J., Rozing, P.M., 2000. The glenohumeral joint rotation centre in vivo. Journal of Biomechanics 33, 1629–1636.

Woltring, H.J., 1986. A Fortran package for generalized, cross-validatory spline smoothing and differentiation. Advanced Engineering Software 8, 104–113.

Wu, G., Cavanagh, P.R., 1995. ISB recommendations for standardization in the reporting of kinematic data. Journal of Biomechanics 28, 1257–1261.