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Measurement error of 3-D kinematic and kinetic measures during overground endurance running in recreational runners between two test sessions separated by 48 hours.

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Title: Measurement error of 3-D kinematic and kinetic measures during overground endurance running in recreational runners between two test sessions separated by 48 hours.

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Abstract

Objective: The purpose of this study was to quantify measurement error of 3-D kinematic and kinetic measures during overground endurance running between two sessions separated by 48 hours. Approach: 13 recreational runners were assessed on two occasions while running overground, over embedded force plates and through an array of 3-D cameras. Main results: In the sagittal, frontal and transverse planes, over the entire stance phase, the typical error of kinematic variables ranged from 1.33° – 6.16° for the hip, 1.38° – 6.01° for the knee and 0.48° – 7.36° for the ankle. Over the same time period and planes typical error of peak-joint moments ranged from 0.04 – 0.54 $Nm{\cdot}Kg^{-1}$ for the hip, 0.06 – 0.37 $Nm{\cdot}Kg^{-1}$ for the knee and 0.01 – 0.15 $Nm{\cdot}Kg^{-1}$ for the ankle. Significance: Results suggest 3-D kinematic and kinetic measures of the stance phase in overground-endurance running are reliable between sessions separated by 48 hours. The measurement error reported here could inform sample-size estimates for future studies and provide smallest-detectable changes for the interpretation of interventions performed over a similar time scale.

Key Words: Running; Reliability; Kinematic; Kinetic; Biomechanics.

Introduction

Running is one of the most popular forms of exercise for both adults and adolescents (Hulteen et al., 2017). Running USA (2014) report approximately 54 million people run in America at least once per week. However, injury rates are also high ranging between 19.4 and 79.3% (Taunton et al., 2002; van Gent et al., 2007). In attempts to understand and resolve this, research has investigated running technique and footwear, and their relationship with surrogate measures of injury and injury rates (Daoud et al., 2012; Sinclair, 2014). A common approach in these types of investigation is 3-D kinematic and kinetic analysis. It is therefore important to understand measurement error of these tools in their application to endurance running to aid interpretation of findings, and in particular the effects of interventions designed to reduce or eliminate factors associated with injury risk.

Currently, few studies report the reliability of kinematic and kinetic measures during overground-endurance running (Diss, 2001; Ferber, Davis, Williams, & Laughton, 2002). Of those that do, there is a trend for peak-joint angles and peak-joint moments to be more reliable within a single test session than between test sessions performed on separate days (Ferber, et al., 2002; Mason, Preece, Bramah, & Herrington, 2016; Queen, Gross, & Liu, 2006). Explanations for increased measurement error between days include wand alignment and anatomical landmark reapplication (Della Croce, Leardini, Chiari, & Cappozzo, 2005). Ferber, et al. (2002) suggest that differences between days effects absolute measures (e.g. joint angles at initial contact) more than relative measures (e.g. range of motion). Collectively, this suggests measurement error in marker reapplication has the potential to create an offset in joint kinematics and joint range of motion will be more reliable than absolute values.

Kinematics in the sagittal plane are suggested to be least effected by measurement error (Ferber, et al., 2002; Queen, et al., 2006). Della Croce, et al. (2005) reported that when joint anatomy dictates motion primarily in one plane (e.g. the knee joint), variability in rotations out of this plane are augmented by inaccurate marker reapplication. This suggests that increased measurement error in the frontal and transverse plane in endurance running might be underpinned by inaccurate marker application between days. Moreover, Kadaba, Ramakrishnan, & Wootten (1990) have reccomeneded that, in some cases, frontal and transverse kinematics at the knee joint are to be interprated with caution. Soft-tissue artefect also contributes to increased kinemetaic varability. Using bone pins as reference data, and an optimal marker configuration, Manal, McClay, Stanhope, Richards, & Galinat (2000) demonstrated that soft-tissue artifact was larger in the transverse plane than the sagittal plane (sagittal error ±2° and transverse ±4°). Together, this suggests that kinematics in the frontal and transverse plane exhibit larger measurement error than the sagittal plane. This has important implications for studies examining knee adduction moments, a loading pattern suggested as a risk factor for running-related injuries (Dudley, Pamukoff, Lynn, Kersey, & Noffal, 2017).

Similar to kinematic data, kinetic measures in the sagittal plane appear less variable between days than measures in the other planes (Ferber, et al., 2002; Kadaba et al., 1989). This seems logical given the effects of soft-tissue artefact and erroneous marker reapplication. Furthermore, as the calculation of joint moments require that trajectories are double differentiated, it follows that kinetic measurement error in each plane will be augmented by inverse dynamic calculations that rely on kinematic data. The augmentation of error resulting from inverse dynamic calculations, in particular that of the frontal and transverse planes, supports the observations of increased variability between days in frontal-and transverse-peak-joint moments. This could also explain why sagittal plane moments report the smallest measurement error between days.

Despite suggestions that, in some cases, frontal and transverse data of particular joints should be interpreted with caution (Kadaba, et al., 1990), comparisons between lower-limb joints are lacking. The thigh segment possesses an increased volume of soft tissue compared to the shank and foot segment. As such, markers attached to the thigh will likely have increased soft-tissue artefact. In a study investigating the effect of skin movement on the analysis of knee-joint motion during running, Reinschmidt, Van Den Bogert, Nigg, Lundberg, & Murphy (1997) reported that the thigh segment was responsible for the majority of the discrepancies between the skin and skeletal-mounted markers. In addition, inaccurate anatomical landmark identification is less likely at the lower limbs as a result of protruding bony landmarks and less soft tissue. A comparison of joint-specific reliability has yet to be performed between days.

Currently, few studies address multiple measures in the stance phase at once. A study addressing multiple time points across stance using a controlled method is important, as many kinematic variables are used to examine mechanisms of running injury. For example, ankle flexion at initial contact has previously been used to suggest kinematic adaptations over multiple running surfaces (Gruber, Silvernail, Brueggemann, Rohr, & Hamill, 2012). In addition, while there are data quantifying the relative reliability of peak-joint loading (Ferber, et al., 2002), absolute measurement error data expressed in international system of units (SI) are sparse. Absolute measurement error is essential for interpreting the effectiveness of an intervention against random noise, and deciding on its practical/clinical value if the change exceeds measurement error. Test-retest correlations (and other relative reliability metrics) are of little value in this regard.

The purpose of this investigation was to quantify absolute measurement error of 3-D kinematic and kinetic variables at key time points in the stance phase of overground running between two sessions separated by 48 hours.

Methods

Participants

With institutional ethics approved and consent attained, 13 volunteers participated. Mean and SD age, height and mass were 28 ± 6 yrs, 1.75 ± 0.07 m and 76.8 ± 10.0 Kg. Inclusion criteria were aged 18-45 and participation in endurance running more than once per week. Participants were excluded if they had an injury in the previous six months to the lower limbs such as tendon/ligament damage that precluded running activates for more than a week, or any neuromuscular condition that would affect normal running gait such as peripheral nerve damage.

Experimental design

3-D kinematic and kinetic measures were collected during two sets of running trials separated by 48 hours. Prior to data collection participants were provided with short-sleeved compression top and shorts to improve skeletal representation in biomechanical modelling. Participants were instructed to be well rested before both testing sessions and were instructed to return to the lab at the same time of day. Testing took place over 20 meters with participants running from an indoor running track through a calibrated biomechanics suite containing 12 infrared 3-D motion analysis cameras and four embedded force platforms. As time to habituate to new footwear is not known, participants were instructed to perform both sessions in their own habitual running shoes.

Procedures

Anthropometric measures were recorded for use in biomechanical modelling (stature (mm), mass (Kg), bilateral-leg length (mm), and knee and ankle joint width (mm)). Participants reported the leg they kicked a football with to identify limb dominance. 3-D trajectories were recorded using 12 infrared cameras (T20, Vicon, Oxford, UK).

Participants were given five minutes to establish a consistent speed they would describe as 'an endurance pace that could be comfortably sustained for 45 minutes'. A pair of timing gates (Brower timing gates, Utah, USA) placed 10 meters apart and set at hip height provided a split time from which running speed was calculated. A mean of five 20-m runs after the fifth minute was taken as the participant's mean running speed for subsequent data capture trials. During data collection participants were required to run at this speed ± 5%. Three successful trials were recorded from each session (Diss, 2001). Successful trials were defined as speed within 5% of the target, the entire stance phase occurring on a force platform, and no targeting of the force platforms. Mean and SD running speed for all trials was 2.81 ± 0.59 $m \cdot s^{-1}$.

Sixteen (Ø=14mm) reflective markers attached in a 'Plug-In gait' formation facilitated the assessment of lower-limb kinematics and kinetics (Vicon, 2010). A level one ISAK certified researcher with 12 months prior experience in gait analysis attached markers for all testing sessions. Markers were attached to skin where possible following participants wore mid-thigh compression cycling shorts. Anatomical locations were anterior-and posterior-superior iliac spine, the bilateral-distal-lateral thigh, bilateral-femoral-lateral epicondyle, the bilateral-distal-lateral shank, the bilateral-lateral malleoli, the left/right toe (dorsal aspect of the second metatarsal head) and the bilateral calcaneus at the same height as the toe marker. To assess the measurement error of peak-forefoot abduction angle while performing endurance running for a future study, the latter 10 participants had an additional markers attached to their dominant lower limb (and shoe) in an 'Oxford-Foot Model' formation as described and illustrated by Stebbins, Harrington, Thompson, Zavatsky, & Theologis (2006). Marker locations were as follows, sacrum, lateral head of the fibula, tibial tuberosity, anterior aspect of the shin, the medial malleoli, the proximal aspect of the calcaneus, a 'peg marker' extending from the most posterior aspect of the calcaneus, the inferior aspect of the calcaneus, sustaniculum tali, proximal and dorsal aspect of the first metatarsal, the medial and distal aspect of the first metatarsal head, the proximal-and distal-lateral aspects of the fifth metatarsal, the lateral calcaneus and the medial aspect of the first phalanx. For markers of the foot, footwear was palpated and markers where attached to corresponding anatomical landmarks. A static trial was recorded for biomechanical modelling. Kinematic data were captured at 200Hz. To avoid targeting of embedded force plates (OR6-7, AMTI, Watertown MA, USA), all four were activated and the force plate with a clear and dominant full foot contact was analysed. Kinetic data were captured at 1000Hz. Kinetic data were then processed by an amplifier (MiniAmp MSA-6, AMTI, Watertown MA, USA) with a gain of 1000 and imported into one of two available Giganet core-processing units (Vicon, Oxford, UK).

Data analysis

Only the dominant limb was analysed (12 right, 1 left). Initial contact and toe off events were defined when the ground-reaction force crossed a 20N threshold (Vicon, 2010). In line with the recommendations of Bisseling and Hof (2006) a 25Hz fourth-order Butterworth filter with zero lag filtered both the kinematic and kinetic data. Three-dimensional joint angles were calculated with respect to the proximal segment using a Cardan XYZ rotation sequence. Polygon Authoring Tool (3.5.1, Vicon, Oxford, UK) was used to normalise data to the stance phase and normalise joint moments to body weight. Data were then exported to Microsoft Excel (Microsoft, USA) where kinematic data were extracted for the ankle, knee and hip at key positions of initial contact (IC), mid-stance (50% of stance) (MS), toe off (TO), maximum (MAX) and minimum (MIN) excursions, and the subsequent range of motion (ROM). Peak-forefoot abduction angle relative to the tibia and peak-joint moments in the sagittal, frontal and transverse planes for the hip, knee and ankle were also exported for statistical analysis.

Statistical Analysis

After verification of underpinning assumptions of normality, linearity and equality of errors over data ranges, mean difference, intra-class correlations (ICC), least-products regression (LPR) and typical error (TE) were calculated. Multiple metrics of reliability were calculated due to debate in the literature (Atkinson and Nevill, 2000; Hopkins, 2000a; Ludbrook, 1997) and to facilitate comparison with previous studies. Mean difference assessed the systematic difference between days. Least-products regression was chosen over linear regression as it assumes error in both testing occasions and minimises the sums of squares for both x and y, whereas simple linear regression assumes x is measured without error and minimises the sum of squares only in Y therefore underestimating total error (Ludbrook, 1997). Interpreting LPR, a slope of one and intercept of zero represent perfect replication. Intra-class correlation was chosen because it is a common metric used in other studies assessing reliability of 3-D kinematics and kinetics (Ferber, et al., 2002), where correlations <0.5, 0.5 - 0.8 and >0.8 represent weak, moderate and strong, respectively (Newell, Aitchison, & Grant, 2014). The main study aim was to assess absolute measurement error.

Hopkins (2000a) suggests typical error is the most appropriate means to assess absolute error due to low bias (particularly in small samples) and ease of interpretation, given error is quantified in the units of the measurement device. Typical error was calculated as follows, TE = SD $_{diff}$ / √2. As such, typical error was taken as our primary expression of measurement error. Mean difference and LPR were calculated using SPSS 22.0 (SPSS, Inc., Chicago, IL, USA), TE and ICC were derived using and a custom built spreadsheet Microsoft Excel spreadsheet (Hopkins, 2000).

Results

Tables 1-3 show mean difference, typical error, intra-class correlations and least-products regression of kinematic and kinetic measures at initial contact, mid stance and toe off, maximum, minimum and ROM during overground running trials separated by 48hrs.

Table 1: Between-day kinematic and kinetic measurement error of the hip joint during the stance phase of overground endurance running, in 13 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal (°)	IC	0.26 ± 3.08	2.18 (1.64 - 3.30)	0.89 (0.72 - 0.96)	1.15 -5.12
	MS	0.07 ± 2.75	1.95 (1.47 - 2.95)	0.94 (0.84 - 0.98)	1.18 -4.38
	TO	0.18 ± 2.53	1.79 (1.35 - 2.71)	0.92 (0.79 - 0.97)	0.87 -0.47
	Peak flexion	-0.11 ± 2.90	2.05 (1.55 - 3.11)	0.92 (0.78 - 0.97)	1.15 -5.57
	Peak extension	0.20 ± 2.48	1.76 (1.33 - 2.66)	0.92 (0.79 - 0.97)	0.88 -0.42
	ROM	-0.31 ± 2.28	1.61 (1.22 - 2.44)	0.89 (0.72 - 0.96)	1.17 -7.72
Sagittal moments ($Nm \cdot kg^{-1}$)	Peak flexion	0.37 ± 0.77	0.54 (0.41 - 0.82)	0.81 (0.55 - 0.93)	1.16 -0.07
	Peak Extension	-0.25 ± 0.72	0.51 (0.39 - 0.77)	0.56 (0.12 - 0.82)	1.84 1.20
Frontal (°)	IC	-1.14 ± 2.65	1.88 (1.42 - 2.84)	0.81 (0.54 - 0.93)	0.95 -0.94

	MS	-1.15 ± 2.76	1.95 (1.48 - 2.96)	0.84 (0.61 - 0.94)	0.92 -0.46
	TO	-0.21 ± 2.63	1.86 (1.41 - 2.82)	0.69 (0.31 - 0.88)	0.78 -0.77
	Peak adduction	-0.21 ± 2.63	1.86 (1.41 - 2.82)	0.69 (0.31 - 0.88)	0.88 -0.13
	Peak abduction	-0.56 ± 2.38	1.68 (1.27 - 2.55)	0.74 (0.41 - 0.90)	0.79 -1.10
	ROM	-0.79 ± 1.88	1.33 (1.01 - 2.02)	0.86 (0.65 - 0.95)	0.93 0.20
Frontal moments ($Nm \cdot kg^{-1}$)	Peak adduction	0.08 ± 0.40	0.28 (0.21 - 0.42)	0.58 (0.14 - 0.83)	1.41 -0.58
	Peak abduction	0.05 ± 0.26	0.18 (0.14 - 0.28)	0.34 (-0.16 - 0.71)	1.41 -0.58
Transverse (°)	IC	-2.04 ± 8.56	6.05 (4.57 - 9.17)	0.44 (-0.05 - 0.76)	1.16 -1.39
	MS	0.10 ± 7.94	5.61 (4.24 - 8.51)	0.56 (0.11 - 0.82)	1.49 1.58
	TO	-2.30 ± 8.65	6.11 (4.62 - 9.27)	0.42 (0.08 - 0.75)	1.39 -0.28
	Peak internal rotation	-0.67 ± 8.35	5.90 (4.46 - 8.95)	0.60 (0.17 - 0.84)	1.33 -1.38
	Peak external rotation	-2.40 ± 8.71	6.16 (4.65 - 9.33)	0.37 (-0.13 - 0.72)	1.51 1.70
	ROM	1.73 ± 3.26	2.30 (1.74 - 3.49)	0.68 (0.30 - 0.87)	1.10 0.71
Transverse moments ($Nm \cdot kg^{-1}$)	Peak internal rotation	-0.01 ± 0.06	0.04 (0.03 - 0.06)	0.19 (-0.31 - 0.61)	0.76 0.01
	Peak external rotation	0.01 ± 0.13	0.09 (0.07 - 0.14)	0.57 (0.13 - 0.82)	1.09 0.05

Table 2: Between-day kinematic and kinetic measurement error of the knee joint during the stance phase of overground endurance running, in 13 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal (°)	IC	0.23 ± 2.40	1.70 (1.28 - 2.57)	0.76 (0.44 - 0.91)	1.23 -3.83
	MS	-0.24 ± 2.02	1.42 (1.08 - 2.16)	0.83 (0.59 - 0.94)	0.95 2.06
	TO	1.32 ± 2.76	1.95 (1.48 - 2.96)	0.89 (0.72 - 0.96)	0.84 4.38
	Peak flexion	0.62 ± 1.95	1.38	0.78	1.16

			(1.04 - 2.09)	(0.49 - 0.92)	-7.27
	Peak extension	0.76 ± 2.26	1.60 (1.21 - 2.42)	0.80 (0.52 - 0.92)	1.00 0.75
	ROM	-0.14 ± 2.21	1.56 (1.18 - 2.37)	0.90 (0.74 - 0.96)	1.05 -1.88
Sagittal moments ($Nm \cdot kg^{-1}$)	Peak flexion	-0.08 ± 0.34	0.24 (0.18 - 0.36)	0.80 (0.53 - 0.93)	1.19 -0.63
	Peak extension	0.02 ± 0.11	0.08 (0.06 - 0.12)	0.80 (0.53 - 0.93)	0.82 -0.08
Frontal (°)	IC	-0.12 ± 2.23	1.58 (1.19 - 2.39)	0.82 (0.56 - 0.93)	1.16 0.08
	MS	1.56 ± 5.46	3.86 (2.91 - 5.85)	0.59 (0.16 - 0.83)	1.44 2.49
	TO	0.17 ± 2.82	1.99 (1.51 - 3.02)	0.79 (0.50 - 0.92)	1.35 0.45
	Peak adduction	0.16 ± 4.48	3.17 (2.40 - 4.80)	0.66 (0.26 - 0.86)	1.26 -0.51
	Peak abduction	-0.85 ± 4.57	3.23 (2.44 - 4.90)	0.64 (0.23 - 0.85)	1.51 1.34
	ROM	1.01 ± 2.07	1.46 (1.10 - 2.21)	0.67 (0.29 - 0.87)	1.26 -0.78
Frontal moments ($Nm \cdot kg^{-1}$)	Peak adduction	-0.04 ± 0.53	0.37 (0.28 - 0.56)	0.49 (0.02 - 0.79)	1.18 -0.31
	Peak abduction	0.01 ± 0.10	0.07 (0.06 - 0.11)	0.63 (0.22 - 0.85)	1.48 0.12
Transverse (°)	IC	1.95 ± 5.67	4.01 (3.03 - 6.08)	0.82 (0.56 - 0.93)	0.96 2.16
	MS	0.91 ± 8.20	5.80 (4.38 - 8.79)	0.69 (0.32 - 0.88)	0.96 1.66
	TO	2.33 ± 6.31	4.46 (3.37 - 6.76)	0.75 (0.42 - 0.90)	1.26 1.87
	Peak internal rotation	1.31 ± 8.50	6.01 (4.54 - 9.10)	0.67 (0.28 - 0.87)	1.03 0.61
	Peak external rotation	2.19 ± 6.06	4.28 (3.24 - 6.49)	0.75 (0.43 - 0.91)	1.18 2.01
	ROM	-0.88 ± 4.14	2.93 (2.21 - 4.44)	0.50 (0.04 - 0.79)	1.01 -1.15
Transverse moments ($Nm \cdot kg^{-1}$)	Peak internal rotation	-0.04 ± 0.08	0.06 (0.04 - 0.08)	0.48 (0.01 - 0.78)	0.58 0.02
	Peak external rotation	0.02 ± 0.09	0.07 (0.05 - 0.10)	0.61 (0.18 - 0.84)	0.53 -0.04

Table 3: Between-day kinematic and kinetic measurement error of the ankle joint during the stance phase of overground endurance running, in 13 recreational runners.

Plane of motion	Time point	Mean diff ± SD	AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Sagittal	IC	0.59 ± 1.74	1.23 (0.93 - 1.87)	0.95 (0.86 - 0.98)	0.96 1.38
(°)	MS	0.81 ± 1.84	1.30 (0.98 - 1.97)	0.87 (0.68 - 0.95)	0.91 3.61
	TO	0.92 ± 4.75	3.36 (2.54 - 5.09)	0.82 (0.57 - 0.93)	0.94 0.23
	Peak dorsiflexion	0.63 ± 1.71	1.21 (0.91 - 1.83)	0.85 (0.63 - 0.95)	0.96 1.89
	Peak plantarflexion	0.97 ± 4.70	3.33 (2.51 - 5.04)	0.82 (0.57 - 0.93)	0.95 0.40
	ROM	-0.34 ± 5.01	3.54 (2.68 - 5.37)	0.81 (0.54 - 0.93)	0.99 0.30
Sagittal moments ($Nm{\cdot}kg^{-1}$)	Peak dorsiflexion	-0.07 ± 0.19	0.13 (0.10 - 0.20)	0.89 (0.71 - 0.96)	1.02 -0.13
	Peak plantarflexion	0.02 ± 0.09	0.06 (0.05 - 0.09)	0.91 (0.76 - 0.97)	1.26 0.09
Frontal (°)	IC	0.20 ± 1.49	1.05 (0.80 - 1.60)	0.69 (0.31 - 0.88)	0.93 0.31
	MS	0.28 ± 1.76	1.24 (0.94 - 1.89)	0.67 (0.29 - 0.87)	0.93 0.55
	TO	0.22 ± 1.41	1.00 (0.75 - 1.51)	0.69 (0.31 - 0.88)	1.08 0.26
	Peak adduction	0.22 ± 1.80	1.27 (0.96 - 1.93)	0.74 (0.41 - 0.90)	0.97 0.35
	Peak abduction	0.23 ± 1.42	1.00 (0.76 - 1.52)	0.68 (0.31 - 0.88)	1.07 0.26
	ROM	-0.01 ± 0.68	0.48 (0.37 - 0.73)	0.89 (0.73 - 0.96)	1.01 -0.03
Frontal moments ($Nm{\cdot}kg^{-1}$)	Peak adduction	-0.06 ± 0.17	0.12 (0.09 - 0.18)	0.21 (-0.30 - 0.62)	0.49 0.01
	Peak abduction	0.00 ± 0.21	0.15 (0.11 - 0.22)	0.43 (-0.06 - 0.75)	0.61 -0.10
Transverse (°)	IC	-0.53 ± 9.22	6.52 (4.93 - 9.88)	0.56 (0.11 - 0.82)	0.92 -1.43
	MS	-1.40 ± 10.05	7.11 (5.37 - 10.77)	0.41 (-0.09 - 0.74)	0.88 -3.96
	TO	-0.83 ± 8.63	6.10 (4.61 - 9.24)	0.66 (0.26 - 0.86)	1.17 -0.83
	Peak Internal rotation	-0.87 ± 8.66	6.12 (4.63 - 9.28)	0.65 (0.25 - 0.86)	1.17 -0.90

	Peak external rotation	-0.98 ± 10.41	7.36 (5.56 - 11.16)	0.46 (-0.02 - 0.77)	0.90 -3.54
	ROM	0.11 ± 3.15	2.23 (1.68 - 3.38)	0.81 (0.54 - 0.93)	0.74 6.68
Transverse moments ($Nm \cdot kg^{-1}$)	Peak internal rotation	0.02 ± 0.16	0.11 (0.08 - 0.17)	0.60 (0.17 - 0.84)	0.97 0.04
	Peak external rotation	0.01 ± 0.01	0.01 (0.01 - 0.02)	0.84 (0.60 - 0.94)	0.92 0.00

The largest typical error for the hip joint in the sagittal, frontal and transverse plane was 2.18°, 1.95° and 6.16°, respectively, at key positions of initial contact, mid stance and peak-external rotation, respectively. The largest typical error for the knee joint in the sagittal, frontal and transverse plane was 1.95° 3.86° and 6.01°, respectively, at key positions of toe off, mid stance and peak-internal rotation, respectively. The largest typical error for the ankle joint in the sagittal, frontal and transverse plane was 3.54°, 1.27° and 7.36°, respectively, for ROM, peak adduction and peak-external rotation, respectively.

The largest typical error for peak-joint moments at the hip joint in the sagittal, frontal and transverse plane was 0.54, 0.28 and 0.09 $Nm \cdot kg^{-1}$, respectively. The largest typical error for peak-joint moments at the knee joint in the sagittal, frontal and transverse plane was 0.24, 0.37 and 0.07 $Nm \cdot kg^{-1}$, respectively. The largest typical error for peak-joint moments at the ankle joint in the sagittal, frontal and transverse plane was 0.13, 0.15 and 0.11 $Nm \cdot kg^{-1}$, respectively.

Table 4 illustrates the mean difference, typical error, intra-class correlations and least-products regression for peak-forefoot abduction angle during stance between two testing occasions separated by 48 hours.

Table 4: Between-day measurement error of the peak forefoot abduction angle during the stance phase of overground endurance running in 13 recreational runners.

Research design	Variable	Mean diff ± SD		AVRG TE (CI)	AVRG ICC (CI)	LPR slope and intercept
Between-day (°)	Peak forefoot abduction	-0.32	5.12	3.62 (2.64 - 5.96)	0.51 (-0.06 - 0.83)	1.03 -2.07

Between two testing occasions separated by 48 hours the typical error of peak-forefoot abduction angle was 3.62°.

Discussion

The purpose of this study was to quantify absolute test-retest error of 3-D kinematic and kinetic measures between two sessions of overground-endurance running separated by 48 hours. Typical error for kinematic measures depending on lower-limb joint and plane of motion ranged from 0.48° to 7.36°. As expected, transverse plane kinematics showed a trend for larger measurement error when compared the sagittal and frontal planes. Depending on lower-limb joint and plane, the typical error for peak-joint moments ranged from 0.01 to 0.54 Nm·kg^{-1}.

Between-day kinematic error was small and comparable to previous work for majority of measures (sagittal: 1.21 – 3.54°; frontal: 0.48 - 3.86°; transverse: 2.30 - 7.36°). For example, when comparing self-selected to standardised running speed between days, Queen, et al. (2006) reported comparable coefficients of multiple correlations for peak knee joint angles in the sagittal (0.91), frontal (0.62) and transverse (0.76) planes. Queen, et al. (2006), Mason, et al. (2016) and Ferber, et al. (2002) all reported that within-session measurement error of kinematics was generally smaller than between-day error. Specifically, when observing the effects of self-selected running speed on the reliability of knee joint kinematics, Queen, et al. (2006) reported that measurement error was significantly higher in all planes between days than within trials on the same day. A possible explanation for increased error between days is marker reapplication. Osis, Hettinga, Macdonald, & Ferber (2016) demonstrated a 10mm anterior/posterior translation in marker location equated to a 7.59° change in peak-ankle joint angle; this demonstrates the sensitivity of measurement error to marker placement. As majority of measures in the sagittal and frontal plane were less than the maximum error reported in Osis, et al. (2016) and the largest reported error in the transverse plane was similar to current work, it can be concluded that measurement error was small. In summary, between-day measurement error for kinematic measures was small and comparable to previous work, therefore the described method is reliable for future injury risk assessment when running at similar speeds.

Absolute error in peak-joint moments were generally small between days (sagittal: 0.06 - 0.54 $Nm \cdot kg^{-1}$; frontal: 0.07 – 0.37 $Nm \cdot kg^{-1}$; transverse: 0.01 – 0.11 $Nm \cdot kg^{-1}$). Intra class correlations calculated for comparative purposes were less than previously reported between-day-peak-joint-moment measurement error (hip: 0.73 - 0.95; knee: 0.73 – 0.8; ankle 0.42 – 0.94, (Ferber, et al., 2002)). A possible explanation might be the choice of biomechanical model. In this study, the 'Plug-In gait' model used small wands that extend from the lateral aspects of the lower limbs. As there is no clear anatomical reference for the placement of wands, between-day measurement error could be larger as a result of problems identifying a consistent location between days. Taking this further, if wand location is inconsistent, it is likely that the reported effects of phasic muscle action on wands will differ between-days (Manal, et al., 2000). Remembering that inverse dynamic calculations exacerbate all sources of kinematic error, error in wand placement might explain the decreased intra-class correlations compared to previous work using a cluster marker system. Alternatively, from a statistical perspective, a reduced spread of scores would also decrease intra-class correlations; however, Ferber, et al. (2002) did not report standard deviations, therefore a comparison between spread of scores was not possible. Although intra-class correlations in this study were not as large as previous work, possibly because of differences in biomechanical models, when expressed in absolute terms, the reported kinetic error between days was small. In the context of applying 3-D kinetic measures to running injury, the absolute error reported was sufficiently small to detect magnitudes of difference associated with injury. For example, Dudley, et al. (2017) reported that a peak-knee adduction magnitude of 0.39 $Nm \cdot Kg^{-1}$ differentiated injured and uninjured runners. This value is larger than between-day measurement error found in this study (0.37 $Nm \cdot Kg^{-1}$). Collectively, absolute between-day peak-joint-moment error was sufficiently small to detect differences that distinguish injured from non-injured runners and demonstrated the method outlined in this study is sufficiently rigorous to detect the potential for injuries in a sample of recreational-overground endurance runners.

Between-day kinematic error was similar across the joints assessed (hip: 1.33 - 6.16°; knee: 1.38 – 6.01°; ankle: 0.48 – 7.36°). A possible explanation is that the combined effect of measurement error and physiological variability between days was large enough to supersede any joint-specific trend (Della Croce, et al., 2005). For example, a 10mm error in marker placement can induce a kinematic change of up to 7.59° at the ankle joint (Osis, et al., 2016). Notably, peak-joint moments at the ankle joint report the smallest measurement error between-days for majority of peak-joint moments (hip: 0.04 - 0.54 $Nm \cdot kg^{-1}$; knee: 0.06 – 0.37 $Nm \cdot kg^{-1}$; ankle: 0.01 – 0.15 $Nm \cdot kg^{-1}$). In contrast, Ferber, et al. (2002) reported larger average intra-class correlations for peak moment data between days at the hip (0.86) and knee (0.84) compared to the ankle (0.7). Factors that might explain these findings are the reduced effects of soft-tissue artefact and wand placement error at the ankle. Increased soft-tisue mass at the thigh relative to the foot segment is likley to contribute to increased soft-tissue artefact for the the hip-and knee-peak joint moments compared to ankle-peak joint moments (Reinschmidt, et al., 1997). Also, as previously disscussed, a wand based model could introduce additional hip and knee joint measurement error compared to ankle joint measurement error as a function of erronous wand placement between days (Leardini et al., 2007). The combination of soft-tissue artefact, marker application error and use of a wand based model are a likely explanation for the increased measurement error between days for peak-joint moments at the knee and hip. In general, kinematic error was similar across the joints assessed though there was a trend for the ankle to have the smallest meaurement error for the majority of peak-joint moments recorded between days. Measurement error in the transverse plane was larger than other planes for most measures. In context, the largest measurement error in the transverse plane was near twice that of the sagittal and frontal plane measurement error between-day data (transverse 7.36°; frontal and sagittal: 3.86° and 3.54°). These findings agree with the work of Della Croce, et al. (2005) who reported that when a joint predominantly performs in one plane, for example running, small rotations out of this plane are strongly influenced by small errors in marker placement. This goes some way to explain the trend for the transverse plane to produce the largest measurement error. In addition, Manal, et al. (2000) reports soft-tissue artefact effects the transverse plane more than the primary plane of motion (sagittal error ±2° and transverse ±4°). Taking this further, Manal, et al. (2000), proposed phasic muscle actions acting upon mid-segment wands as the mechanism responsible for increased transverse-plane measurement error. Collectively, future research should exercise caution when interpreting transverse plane data following data suggest the transverse plane has the largest measurement error from a combination of marker placement error, soft-tissue artefact, and phasic-muscle contractions.

Relative measures such as ROM (0.48 - 3.54°) were more reliable between-days than absolute measures (1 - 7.36°), such as joint angle at initial contact. This is consistent with previous work by Ferber, et al. (2002) who suggests erroneous marker placement between testing occasions introduces an offset in absolute kinematic data. Supporting this finding, work by Kadaba, et al. (1990) reported that when the hip joint centre was systematically translated by 10mm, the curve was displaced, however, the shape of curve was unchanged. This evidence suggests, between-day relative measures were more reliable, possibly because the error introduced by marker reapplication.

Limitations

The footwear used in this study was not consistent between participants. However, because the time taken to habituate to new footwear is yet to be reported, using participants' habitual footwear ensured results were not affected by novel adaptations. While some authors argue that more trials might have improved the likelihood of intra-participant variation (Bates, Osternig, Sawhill, & James, 1983), more recent work reports three trials are sufficient to provide a precise representation of overground running kinematic and kinetic data (Diss, 2001; Hopkins, 2000a). Noteably, the 'Oxford-Foot Model' data reflects the movement of the shoe, thus some foot-movement data is likely occluded by footwear. It is also noteable that the data in this study might be effected by the biomechanical model used, therefore, researchers should consider implications of model choice when conducting future biomechanical-running investigations.

Conclusion

The method and procedures undertaken in this study produce reliable measures of 3-D kinematic and kinetic variables that characterise overground-endurance running. Data reported suggests the outlined method is sufficiently reliable to capture injury mechanisms associated with knee joint loading. The typical error data reported here could be used to set thresholds defining minimal-detectable change for future comparative and intervention studies. These results could also be used to inform sample size estimates for studies of a similar timescale using recreational-endurance runners.

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