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A strathclyde cluster model for gait kinematic measurement using functional methods: a study of inter-assessor reliability analysis with comparison to anatomical models

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ABSTRACT

A major source of error in reliability of gait analysis arises from the palpation of anatomical landmarks (ALs). The purpose of this study was to investigate whether less reliance on manually identifying ALs could improve inter-assessor reliability of joint kinematics compared to two anatomical models. It was hypothesised that the Strathclyde functional cluster model (SFCM), in which the hip, knee and ankle joint centres and knee and ankle flexion axes were determined by functional methods, would obtain greater inter-assessor reliability. Ten able-bodied participants and seven assessors were recruited. Each participant completed three trials conducted by different assessors on non-consecutive days. Agreement and inter-assessor reliability between the models were compared and analysed, whilst factor effects of assessor experience and body mass index (BMI) were investigated. The SFCM obtained excellent agreement with anatomical models for all sagittal angles and hip ab/adduction angle, and it showed slightly higher inter-assessor reliability with smaller variations in the knee and ankle. The assessor experience was not a significant factor, but the BMI had a significant effect on the inter-assessor reliability. The results demonstrate that the SFCM may be more beneficial for less experienced assessors.

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Gait; functional method; reliability; kinematics; motion analysis

Introduction

Three-dimensional gait analysis is an important clinical tool for understanding pathological movement patterns and evaluating the efficacy of therapeutic interventions. As the correct interpretation of data relies on the ability of the model to detect small variations between measurements, reliable gait assessment is critical for clinical investigations or research studies, especially when multiple measurements are performed by different assessors (McGinley et al. 2009).

A major source of kinematic errors in human gait analysis is the identification of superficial anatomical landmarks (ALs) (Della Croce et al. 2005; Leardini et al. 2005). Inaccurate location of ALs can result in errors in the location of joint centres, alteration of the orientation in anatomical coordinate systems (ACSs) and consequent inaccuracy in joint kinematics and kinetics (Della Croce et al. 1999; Stagni et al. 2000). The inter-assessor reliability of joint kinematics is largely dependent on the precision of AL position determination. Reduction of these errors related to palpation protocols in identifying the ALs can be obtained by improving the AL identification procedure or using functional methods for the determination of the joint location.

The estimation of the joint centres and axes based on the functional methods does not require accurate location of ALs and therefore is less sensitive to variability in identification of ALs using manual palpation. The methods have been established to estimate the hip joint centre (HJC) (Camomilla et al. 2006; Siston and Delp 2006; Lopomo et al. 2010; Kratzenstein et al. 2012; Zuk et al. 2014; Kainz et al. 2015) and mean helical axes of rotation for the knee (Ehrig et al. 2007; Van Campen et al. 2011). Studies on the validation of the functional methods have reported promising accuracy and repeatability over trials. The intra- and inter-examiner repeatability of gait kinematics obtained applying a functional model to determine the HJC location and the knee axis was compared to those obtained using a traditional anatomical model (Besier et al. 2003). It was shown that

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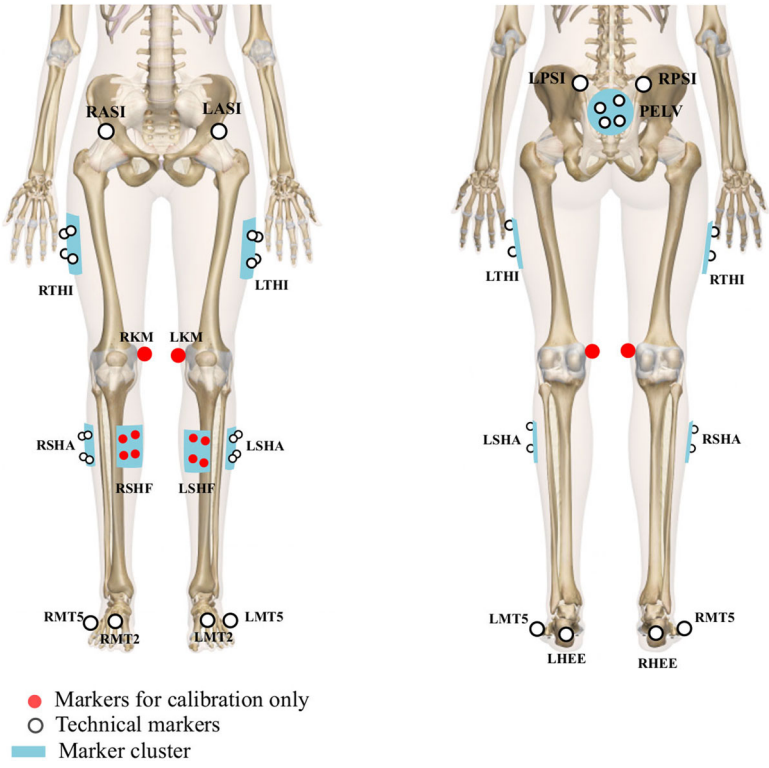

Figure 1. Marker layout of Strathclyde functional cluster model (SFCM).

the functional method provides hip and knee variables with slightly higher repeatability. Comparable results were also found in a reliability study of running (Pohl et al. 2010). However, these studies demonstrated that functional methods did not significantly improve the within- and between-tester reliability of gait kinematics. While the functional models still rely heavily on ALs identification for knee joint centre and ankle joint parameters, concerns regarding the dependency of the knee and ankle joint location estimations on the kinematic data might be raised.

The accuracy of determination of knee and ankle location is not studied in published functional models. It may be due to difficulties in the precise location of the knee joint centre (KJC) when the joint is considered a modified hinged joint with reduced movement. Meng et al. (2019) proposed an optimised functional method-based protocol for the determination of the KJC location during knee flexion/extension. Based on the previous work, the novel Strathclyde Functional Cluster Model (SFCM) was developed, in which joint kinematics were obtained with the joint centres (hip, knee and ankle) and helical axes of the knee and ankle joints determined functionally. We hypothesised that less reliance on manually identifying ALs could further improve the inter-assessor reliability of gait kinematic measurement. Therefore, the purpose of this article was to compare the inter-assessor reliability of SFCM based on functional methods to two anatomical clinical gait analysis models: Plug in Gait (PiG, Vicon, Oxford, UK) and Human Body Model Gait (HBM2, Motek Medical, Amsterdam, the Netherlands). Agreement and reliability indices were used for comparison. Factor effects of assessor experience and participants' body mass index (BMI) were also analysed in this study.

Materials and methods

Strathclyde functional cluster model

The SFCM was created in MATLAB (R2017b, MathWorks Inc., MA, USA) and can be run in Vicon Nexus (Nexus 2.6, Vicon Motion system, Oxford, UK) via the Vicon Nexus and MATLAB interface. The marker layout for SFCM is shown in Figure 1.

Two types of functional methods are utilised in the SFCM. The Centre transformation technique (CTT) was used to localise the joint centres with transformations between adjacent segments during the corresponding movement (Ehrig et al. 2006; Siston and Delp 2006). The sphere fitting algorithm (SFA) was applied to calculate the functional axes of flexion by optimising an objective function assuming that markers trace out a circle around the estimated axis

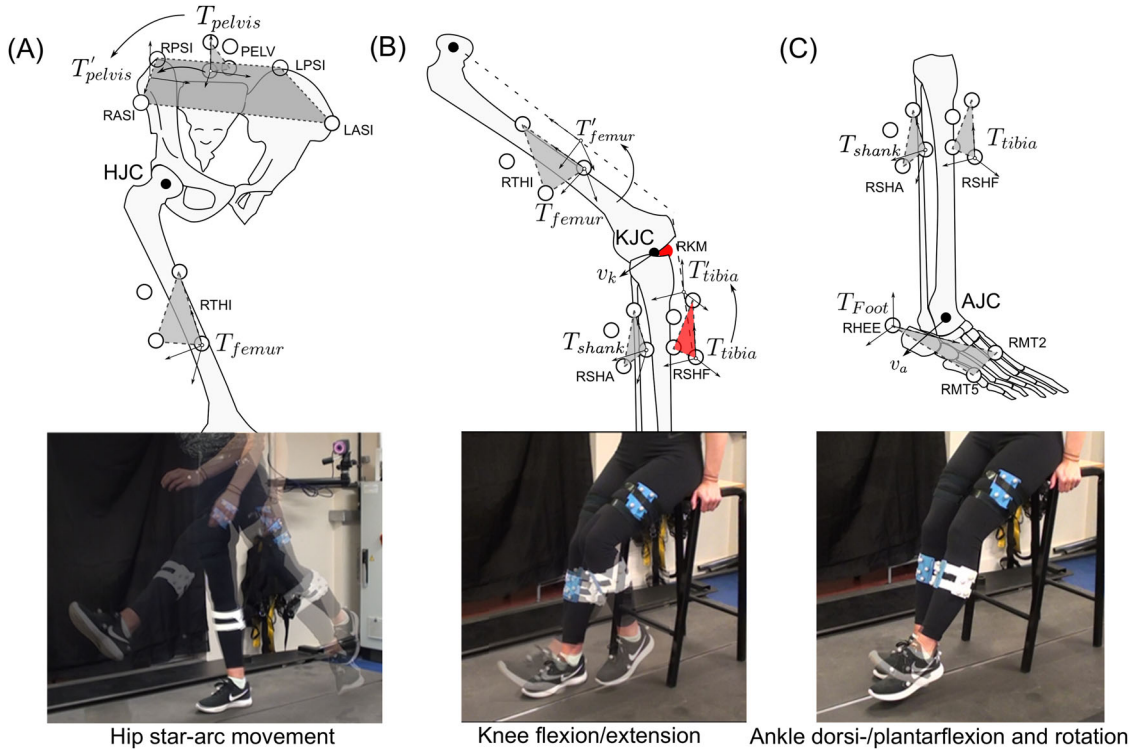

Figure 2. The procedure of functional joint parameters determination. The participant performed the hip star-arc movement, knee flexion/extension, ankle dorsi/plantarflexion and rotation for the (A) hip, (B) knee and (C) ankle joint calibration. HJC: hip joint centre; KJC: knee joint centre; AJC: ankle joint centre; v_k : knee flexion axis; v_a : ankle flexion axis.

of rotation (Gamage and Lasenby 2002; Van Campen et al. 2011).

The functional calibration trial was processed for defining joint parameters. Hip 'star-arc' movement was performed for the HJC calibration (Lopomo et al. 2010). As shown in Figure 2(A), the pelvis coordinate origin was located at the midpoint of L/RASI (left/right anterior superior iliac spine) and L/RPSI (left/right posterior superior iliac spine) markers and the CTT was used to calculate the functional HJC. The KJC was determined using an optimised protocol from our previous study (Meng et al. 2019). The origin of the proximal coordinate system T_{femur} was shifted to the midpoint between the estimated HJC and the medial knee marker. The origin of the distal coordinate system T'_{tibia} was relocated to the midpoint between the origin of T_{tibia} and medial knee marker before the KJC estimation (Figure 2(B)). The knee flexion axis was determined using the SFA. The ankle joint centre (AJC) was estimated using the CTT during the ankle rotation movements. The ankle flexion axis was calculated using the SFA during ankle dorsi/plantarflexion movements as shown in Figure 2(C). The hip flexion axis was not considered in this study because this axis was not included in the International Society of Biomechanics (ISB) recommendation on definitions of ACSs (Baker 2003).

Locations of all joint parameters were related to the corresponding distal cluster coordinate systems as shown in Figure 2. It should be noted that the functional knee parameters needed to be transferred to the technical coordinate system created based on the marker cluster attached to the lateral side of the shank. During walking trials, the marker clusters were tracked, and the joints were localised using the calibration results for each time instant. The ACSs in the SFCM were created following the ISB recommendations and their definitions were detailed in the Supporting Information document A. The ACSs were constructed and used to calculate joint kinematics (Wu et al. 2002).

The definition of joint parameters in the anatomical models

Markers and methods used to define the joint parameters in the SFCM, PiG and HBM2 are summarised in Table 1. All models shared the pelvis ALs. However, the PiG and HBM2 utilised different regression models to estimate the position of HJC based on the pelvis markers; Newington–Gage model for the PiG (Davis et al. 1991) and Harrington model (Harrington et al. 2007) for the HBM2. Lateral ALs on the joints and segments were required in the PiG

Table 1. Markers and methods for the determination of lower-limb joint parameters in the Strathclyde functional cluster model (SFCM), plug in gait (PiG) and human body model gait (HBM2).

	PiG		HBM2	SFCM
Hip	Markers	L/RPSI, L/RASI	L/RPSI, L/RASI	L/RPSI, L/RASI, L/RTHI and PELV cluster
	Method	Newington–Gage Model	Harrington Model	CTT
Knee	Markers	R/LLEK, R/LTHI	R/LLEK R/LMEK	R/LMEK L/RTHI, L/RSHA and L/RSHF clusters
	Method	A chord function is used, where known HJC, R/LLEK and R/LTHI are used to define a plane.	KJC: midpoint between R/LLEK and R/LMEK Joint axis: pointing to R/LLEK from R/LMEK	KJC: CTT Joint axis: SFA
Ankle	Markers	R/LLM, R/LTIB	R/LLM, R/LMM	L/RSHA and L/RSHF clusters, foot cluster consisting of RMT2, RMT5, RHEE
	Method	Same with knee	Same with knee	Same with knee

L/RASI: left/right anterior superior iliac spine; L/RPSI: left/right posterior superior iliac spine; L/RLEK: left/right lateral epicondyle of knee; L/RMEK: left/right medial epicondyle of knee; L/RLLM: left/right lateral malleolus of the ankle; L/RLMM: left/right medial malleolus of the ankle; L/RTHI: left/right thigh; L/RTIB: left/right tibia; HJC: hip joint centre; KJC: knee joint centre; CTT: centre transformation technique; SFA: sphere fitting algorithm.

for identifying knee and ankle joints. As shown in Table 1, a chord function was used to localise the joint centres and axes of the knee and ankle joints by defining a plane with the known proximal joint centre, lateral joint marker and lateral segmental marker. Medial and lateral ALs were all needed to determine the knee and ankle joint parameters in the HBM2. The SFCM requires only the medial knee AL for the determination of the KJC.

Reliability experiment

The study was approved by the ethics committee of the Department of Biomedical Engineering at the University of Strathclyde. Seven researchers volunteered to take part in the study as assessors. All assessors have good practical experience in the operation of PiG and HBM2 while three assessors did not have any knowledge about the SFCM before the experiment as detailed in Supporting Information Document B. An introduction session was taken in which experimental procedures and marker placement for each model were outlined to the assessors.

Ten participants (4 males and 6 females, age = 20-40 year old, BMI = $25.13 \pm 4.22\,kg/m^2$) were enrolled in the study. In each session, the assessor applied the combined marker set on the participant as shown in Figure 3. A static trial was recorded for 5 s with the participant standing in a natural posture. In the functional trial, the participant was asked to perform the required movements, including the hip star-arc movement, knee flexion/extension, ankle dorsi-/plantarflexion and foot rotation. In the walking trial, the participant walked on the treadmill at his/her comfortable speed (1.23 ± 0.13 m/s). Thirty seconds of data were captured after a two-minute familiarisation

Figure 3. A combined marker set-up for the Strathclyde functional cluster model (SFCM), plug-in gait (PiG) and human body model gait (HBM2).

period. Marker trajectories were captured by a 12 camera Vicon motion capture system at a sampling rate of 100 Hz. Each participant completed three trials conducted by different assessors on nonconsecutive days.

Data analysis

Three-dimensional marker trajectories were filtered using a fourth-order low-pass Butterworth filter at a cut-off frequency of 10 Hz and processed using the SFCM, PiG and HBM2 models. The SFCM was run using the Vicon Nexus-MATLAB interface. The

processing pipelines in the Vicon Nexus were used to generate the PiG model outputs. The HBM2 was applied offline using the D-Flow software (Motekforce Link B.V., Amsterdam, The Netherlands). Joint angles were calculated at the hip, knee and ankle angles using the standard Cardan sequence of rotations. The sagittal, coronal and transverse plane angles were calculated for the hip and knee (the HBM2 only had sagittal plane knee angle) as well as the sagittal plane angle for the ankle. Twenty gait cycles were extracted for each walking session. All kinematic outputs were time-normalised to a gait cycle (0–100%). The same researcher carried out all post-processing.

Relative agreement between models was examined with Pearson correlation coefficients (PCC). The differences of mean, maximum and range of motion (RoM) between the models were also calculated for model comparisons. Standard deviation (*SD*) and intraclass correlation coefficient (ICC, 2-1) across gait cycles were used to compare the inter-assessor reliability. The 95% confident interval of the ICC estimate was used as the basis to evaluate the level of reliability. The ICC value (α) was interpreted into four levels of reliability (Koo and Li, 2016): $\alpha < 0.5 =$ poor, $0.5 < \alpha < 0.75 =$ moderate, $0.75 < \alpha < 0.9 =$ good, $\alpha > 0.9 =$ excellent.

Results

The results of the agreement between the models for all joint angles are shown in Table 2 and Figure 4. The kinematic outputs had excellent agreements between the models for all flexion/extension angles and hip ab/adduction angle ($r > 0.9$). The hip rotation angle and non-sagittal knee angles had poor agreements between the PiG and SFCM ($r < 0.5$) whilst the agreement between SFCM and HBM2 was moderate for the hip rotation angle ($r = 0.63$). Table 2 shows that the PCC values between the SFCM and HBM2 tended to be slightly lower compared with those between the SFCM and PiG, except for the hip rotation angle. However, the absolute values of mean, maximum and RoM differences between the SFCM and HBM2 (mean difference $< 2°$, maximum difference $< 3°$, RoM difference $< 4°$) were much smaller compared with those between the SFCM and PiG (mean difference $< 5°$, maximum difference $< 8°$, RoM difference $< 14°$). The largest differences were found in non-sagittal knee angles ($>10°$) when the SFCM and PiG were compared.

Inter-assessor *SD* results are shown in Figure 5. The SFCM demonstrated significantly smaller inter-

Table 2. Comparison in kinematic variables between the Strathclyde functional cluster model (SFCM) and both plug in gait (PiG) and human body model gait (HBM2).

		Mean difference		Max difference		RoM difference		*r*	
		SFCM-PiG	SFCM-HBM2	SFCM-PiG	SFCM-HBM2	SFCM-PiG	SFCM-HBM2	SFCM-PiG	SFCM-HBM2
Hip	Flexion/extension (°)	4.13(3.08)	1.55(2.19)	4.64(2.96)	2.44(2.07)	−0.15(2.58)	0.14(3.03)	1.00**	1.00**
	Ab/adduction (°)	−0.06(3.07)	−0.23(2.53)	−0.1(3.18)	−2.42(2.95)	0.54(1.37)	−3.33(2.24)	0.98**	0.96**
	Rotation (°)	−6.33(15.77)	−1.76(5.04)	−8.07(18.63)	−2.71(6.45)	−3.55(11.85)	−0.97(5.31)	0.11	0.63*
Knee	Flexion/extension (°)	4.15(5.03)	1.16(2.37)	5.97(6.69)	1.37(1.6)	2.15(5.11)	−0.57(2.82)	0.99**	1.00**
	Ab/adduction (°)	−2.46(5.94)		−3.79(14.43)		−12.51(10.97)		0.14	
	Rotation (°)	6.18(12.11)		−2.94(13.51)		−13.41(7.96)		0.22	
Ankle	Dorsi/plantarflexion (°)	−0.23(2.61)	−0.1(1.91)	0.81(4.06)	1.61(2.37)	0.97(8.25)	2.04(3.58)	0.97**	0.96**

* = $p < .05$, ** = $p < .001$, RoM: range of motion.

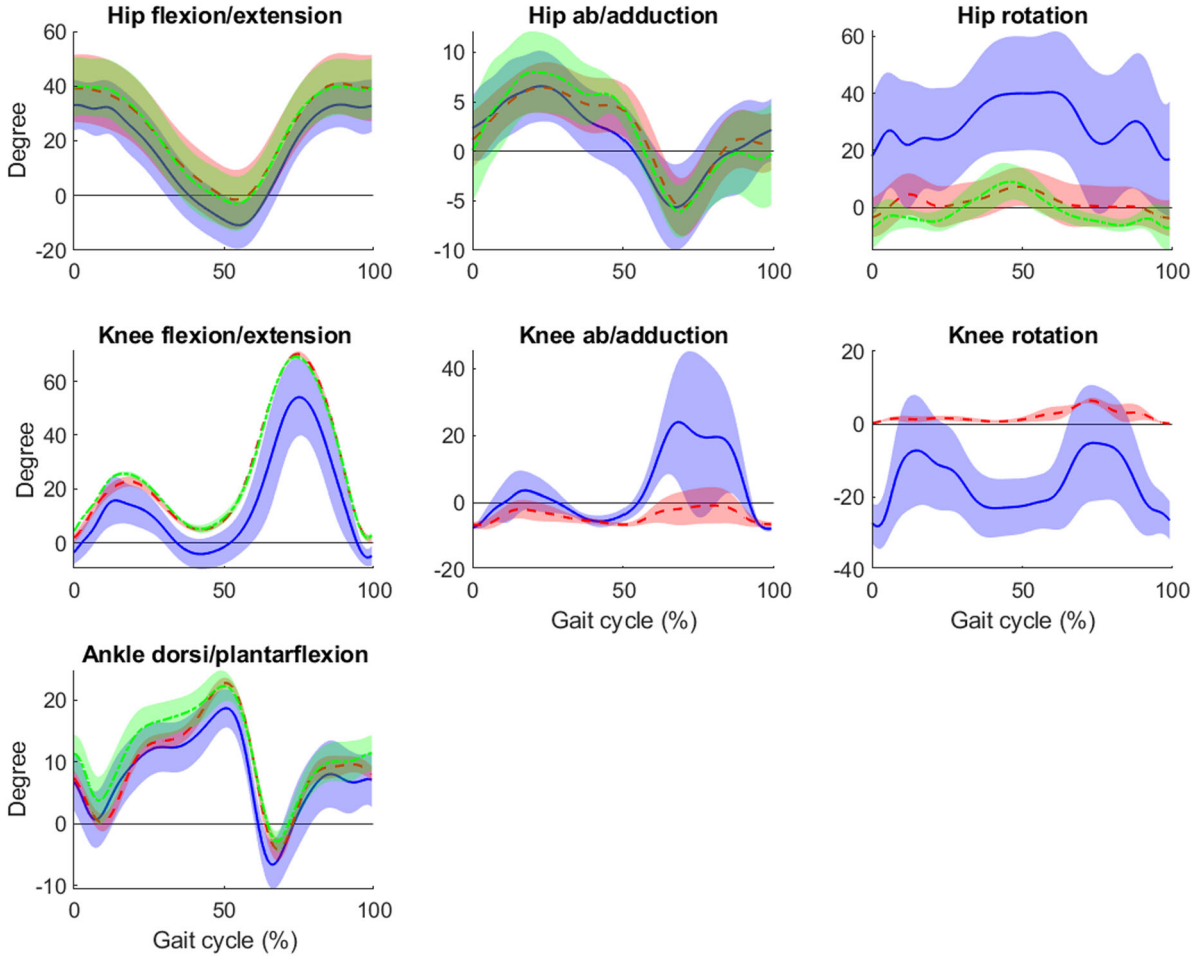

Figure 4. Inter-assessor mean (±*SD*) joint angles at the hip, knee and ankle obtained by the plug-in gait (PiG, blue), Strathclyde functional cluster model (SFCM, red) and human body model gait (HBM2, green) over normalised gait cycles of one participant.

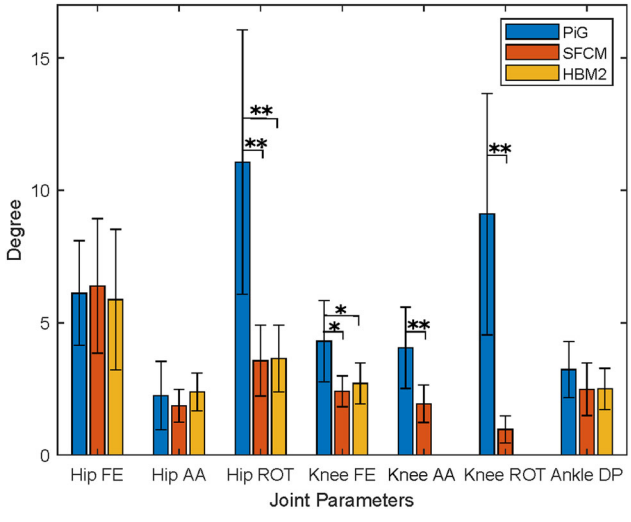

Figure 5. Inter-assessor standard deviation (*SD*) across normalised gait cycles at the hip, knee and ankle of the Plug-in Gait (PiG), Strathclyde functional cluster model (SFCM) and human body model gait (HBM2). * = $p < .05$, ** = $p < .001$. FE: flexion/extension; AA: Ab/adduction; ROT: rotation; DP: Dorsi/plantarflexion.

assessor *SD* values compared to the PiG for the hip rotation angle and knee angles while no significant difference in inter-assessor *SD* values was observed between the SFCM and HBM2. Results of the inter-assessor ICC values are detailed in Table 3. All three models showed good to excellent inter-assessor reliability for the sagittal joint angles (ICC> 0.75). The SFCM demonstrated larger inter-assessor ICC values than both the PiG and HBM2 at most joint angles, except the hip flexion/extension angle ($ICC_{PiG} = 0.86$, $ICC_{HBM2} = 0.86$, $ICC_{SFCM} = 0.84$) and hip rotation angle ($ICC_{PiG} = 0.17$, $ICC_{HBM2} = 0.52$, $ICC_{SFCM} = 0.43$). The inter-assessor reliability of the hip rotation angle, knee flexion/extension and rotation angles and ankle dorsi/plantarflexion angle were significantly higher with the SFCM than that with the PiG model. Although, no significant differences were found between the SFCM and HBM2, SD values of ICC among participants in the SFCM were smaller than those in the HBM2 for all joint angles.

Table 3. Inter-assessor intraclass correlation coefficient (ICC) values for the Strathclyde functional cluster model (SFCM) and anatomical models: plug in gait (PiG) and human body model gait (HBM2).

		PiG			HBM			SFCM		
		ICC (*SD*)	LB (*SD*)	UB (*SD*)	ICC (*SD*)	LB (*SD*)	UB (*SD*)	ICC (*SD*)	LB (*SD*)	UB (*SD*)
Hip	Flexion/extension	0.86 (0.08)	0.80 (0.11)	0.90 (0.06)	0.86 (0.10)	0.81 (0.13)	0.91 (0.07)	0.84 (0.09)	0.78 (0.12)	0.89 (0.07)
	Ab/adduction	0.73 (0.24)	0.67 (0.27)	0.79 (0.21)	0.78 (0.15)	0.71 (0.17)	0.83 (0.12)	0.80 (0.13)	0.74 (0.16)	0.86 (0.11)
	Rotation	0.17 (0.22)	0.14 (0.20)	0.22 (0.24)	0.52 (0.23)	0.44 (0.22)	0.59 (0.22)	**0.43*** (0.20)	0.36 (0.20)	0.50 (0.19)
Knee	Flexion/extension	0.94 (0.05)	0.92 (0.07)	0.96 (0.03)	0.98 (0.01)	0.97 (0.01)	0.99 (0.01)	**0.99*** (0.01)	0.98 (0.01)	0.99 (0.01)
	Ab/adduction	0.36 (0.30)	0.31 (0.29)	0.41 (0.31)				0.40 (0.21)	0.33 (0.20)	0.47 (0.22)
	Rotation	0.28 (0.21)	0.22 (0.17)	0.35 (0.23)				**0.57*** (0.19)	0.51 (0.20)	0.64 (0.18)
Ankle	Dorsi/Plantarflexion	0.82 (0.10)	0.76 (0.13)	0.87 (0.08)	0.86 (0.11)	0.82 (0.13)	0.90 (0.08)	0.89* (0.09)	0.85 (0.12)	0.92 (0.06)

ICC values above 0.9, between 0.75 and 0.9, between 0.5 and 0.75 and below 0.5, indicate excellent, good, moderate and poor reliability, respectively. Standard deviation (*SD*) is between participant means. LB and UB refer to the lower and upper bounds of the ICC.
*represents there are significant difference between inter-assessor ICC values of the PiG and SFCM models ($p < .05$).

A general linear analysis of variance (ANOVA) model was used to investigate the effects of the participant BMI and assessor experience on the inter-assessor reliability as shown in Table 4. The assessor experience had no significant influence on the inter-assessor reliability for all the models. The BMI had a significant effect on the inter-assessor ICC values of the hip flexion/extension angle for all the models. The inter-assessor reliability of the hip rotation and ankle dorsi/plantarflexion angles in the HBM2 were also significantly affected by the BMI as well as the reliability of the knee flexion angle in the SFCM (Table 4).

Discussion

In this article, we proposed a novel cluster-based model in which the hip, knee and ankle joint parameters were determined using functional methods. It was hypothesised that less reliance on the ALs localisation in a gait analysis model using functional methods would improve the inter-assessor reliability compared to anatomical models. This hypothesis was tested in a reliability experiment where the inter-assessor reliability of the SFCM was compared with two different anatomical models (the PiG and HBM2). Results demonstrate that all models showed a 'good to excellent' inter-assessor reliability (ICC $\geq$ 0.75) for all flexion/extension angles and hip ab/adduction angle but performed 'poor to moderate' inter-assessor reliability (ICC $<$ 0.75) for other non-sagittal angles. The SFCM obtained higher reliability with less variation compared with the anatomical models.

The SFCM produced more reliable gait parameters than the anatomical models in most joints. It can be concluded from Table 2 that the HBM2 has the most reliance on ALs, followed by the PiG and the SFCM has the least. The higher inter-assessor reliability obtained by the HBM2 for joint kinematics indicates that the determination of joint parameters based on the medial and lateral ALs is more reliable than the chord function used in the PiG (Table 3). The inter-assessor reliability of knee and ankle kinematics in the SFCM was significantly improved compared with the PiG, but the improvement was not significant when compared to the HBM2. The results are consistent with previous studies (Besier et al. 2003; Pohl et al. 2010) in which anatomical models similar to the HBM2 were used. However, the SFCM produced slightly higher ICCs with smaller variations among the participants compared to the HBM2, suggesting

Table 4. Results of a general linear analysis of variance (ANOVA) model that analysed factor effects of the participant BMI and assessor experience on the inter-assessor ICC for the Strathclyde functional cluster model (SFCM), plug in gait (PiG) and human body model gait (HBM2).

			Assessor experience			BMI		
			PiG	HBM	SFCM	PiG	HBM	SFCM
Hip	Flexion/extension	SumSq	0.33	0.45	0.43	198.74	217.18	230.39
		F	1.31	0.50	0.41	**19.71****	**23.05****	**25.73****
	Ab/adduction	SumSq	0.02	0.56	0.34	45.25	16.39	3.53
		F	0.07	0.63	0.31	2.91	0.99	0.21
	Rotation	SumSq	0.08	0.77	0.19	53.71	126.89	51.02
		F	0.29	0.88	0.18	3.52	**10.03***	0.08
Knee	Flexion/extension	SumSq	0.03	2.60	0.04	10.9	0.03	290.88
		F	0.10	3.18	0.04	0.65	0.00	**42.83****
	Ab/adduction	SumSq	0.00		0.16	11.07		15.44
		F	0.01		0.15	0.67		0.93
	Rotation	SumSq	0.09		0.88	1.89		37.12
		F	0.33		0.83	0.11		2.34
Ankle	Dorsi/plantarflexion	SumSq	0.01	0.17	0.05	29.64	92.88	25.44
		F	0.04	0.19	0.05	1.84	**6.70***	1.56

$^{*} = p < .05$, $^{**} = p < .001$, SumSq: Sum of Squares; DF equals to 1 for all ANOVA analysis.

that our model has the potential to reduce measurement variations caused by inaccurate AL palpation (Baudet et al. 2014). Moreover, we observed that the inter-assessor reliability of the ankle dorsi/plantarflexion angle was further improved with the SFCM ($ICC_{PiG} = 0.82$, $ICC_{HBM2} = 0.86$, $ICC_{SFCM} = 0.89$) compared to results from (Besier et al. 2003) where the difference of the coefficient of multiple determination between the anatomical and functional models was less than 0.01 for the ankle dorsi/plantarflexion. The results indicate that the application of functional methods for estimating the knee joint centre and ankle joint parameters may contribute to improve the reliability of the ankle joint. This would support our hypothesis that a functional model would further improve the inter-assessor reliability with less reliance on AL localisation.

Agreement between the SFCM and anatomical models was excellent for all sagittal angles and hip frontal angle. In particular, the SFCM and HBM2 had higher agreement compare to that between the SFCM and PiG as shown in Table 2. But the SFCM and PiG had poor agreement in knee ab/adduction and rotation angles (Note that the HBM2 was not compared for non-sagittal knee angles). Regarding the large difference of knee frontal and transverse plane rotations between the PiG and SFCM, it is difficult to ascertain the accuracy of knee angles in the two models. The angles were compared to those of (Lafortune et al. 1992) in which cortical pins were embedded in the femur and tibia to measure knee joint rotations. Lafortune et al. (1992) reported features of knee rotations include: (1) The average pattern of the knee ab/adduction angle is limited to 5° and reaches a peak after the maximum knee flexion angle; (2) The knee has internal rotation in stance and external rotation after toe-off with the RoM below 10°. As shown in Figure 4, the PiG produced higher amplitude and variability in knee ab/adduction and rotation. This result is likely due to crosstalk from knee flexion/extension as a consequence of misaligned coordinate system through misplacement of thigh markers (Piazza and Cavanagh 2000; Baudet et al. 2014). The non-sagittal knee kinematics obtained by the SFCM were much closer to the features of knee rotation compared to the PiG. The results indicate that our model would demonstrate higher accuracy and better reliability in the knee frontal and transverse angles.

Results showed that the assessor experience was not a significant factor for any joint kinematics in all models as shown in Table 4. All the assessors were experienced in the PiG and HBM2 which may explain the experience factor does not significantly affect the reliability of these lower limb anatomical models in the experiment. On the other hand, the assessors with different experience levels on the SFCM had no significant influence on the reliability of the SFCM, indicating that functional methods alleviate the requirement of assessor experience in the inter-assessor reliability of gait kinematic analysis. Therefore, the high reliability of the SFCM might be more pronounced with assessors who are less experienced.

Participant BMI is a significant factor on the inter-assessor reliability as seen in Table 4 and significantly influences the hip flexion/extension angle of all three models. Despite the hip flexion/extension angle of all three models were significantly influenced, the BMI had a significant effect on the reliability of the hip rotation and ankle dorsi/plantarflexion in the HBM2 as well as the knee flexion/extension angle in the

SFCM. As the STA is another major source of kinematic variability (Della Croce et al. 2005; Leardini et al. 2005), the difficulties of accurately localising the pelvis ALs on participants with high BMIs and large STA amplitudes during movement may explain the effect of BMI on the anatomical models. The HBM2 was more significantly influenced by the BMI factor due to its high reliance on the locations of ALs (Table 1). On the other side, the STA also plays a significant role in causing inaccuracies in the functional joint centre estimation (Kratzenstein et al. 2012). The use of rigid marker clusters may improve the accuracy of functional methods by eliminating the marker cluster deformation, especially for the CTT method (Meng et al. 2019). However, the STAs caused by the skin sliding cannot be ignored. Further development of models for removing STAs should be considered to enhance the performance of functional methods.

This study proposed a novel functional model that requires less ALs for the determination of the hip, knee and ankle joint parameters, although, ALs were still used in the SFCM for defining the pelvis and foot segments while the medial knee ALs were utilised in the functional knee calibration. These markers cannot be removed using functional methods. Studies have proposed a technical cluster set attached to the sacrum where the pelvis ALs can be transferred as markers in the technical coordinate system using a calibration wand (Borhani et al. 2013; Millar et al. 2019). Borhani et al. (2013) showed that the use of the cluster set had a higher repeatability in non-sagittal hip angles compared to the regression models, especially for the group of obese participants (BMI $>$ 28). As the method may alleviate the STA issue for the data collection of overweight and obese participants, an implementation of the technical cluster set for the pelvis will be undertaken in further study.

Conclusion

The SFCM produced slightly more reliable gait kinematic data with smaller variations than the anatomical models. The improvement is superior compared to previous functional models in which only the hip joint centre and the knee flexion axis were determined using functional methods. Anatomical models requiring precise AL locations, such as the HBM2, could achieve a high inter-assessor reliability when performed by experienced assessors. However, with the advantage of not having to accurately locate ALs for the knee and ankle joints, the improved reliability of the SFCM compared to the HBM2 might be more pronounced with less experienced assessors.

Disclosure statement

No potential conflict of interest was reported by the authors.

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References

Baker R. 2003. ISB recommendation on definition of joint coordinate systems for the reporting of human joint motion—part I: ankle, hip and spine. J Biomech. 36(2): 300–302.

Baudet A, Morisset C, d'Athis P, Maillefert J-F, Casillas J-M, Ornetti P, Laroche D. 2014. Cross-talk correction method for knee kinematics in gait analysis using principal component analysis (PCA): a new proposal. PLoS One. 9(7):e102098.

Besier TF, Sturnieks DL, Alderson JA, Lloyd DG. 2003. Repeatability of gait data using a functional hip joint centre and a mean helical knee axis. J Biomech. 36(8): 1159–1168. English.

Borhani M, McGregor AH, Bull AM. 2013. An alternative technical marker set for the pelvis is more repeatable than the standard pelvic marker set. Gait Posture. 38(4): 1032–1037.

Camomilla V, Cereatti A, Vannozzi G, Cappozzo A. 2006. An optimized protocol for hip joint centre determination using the functional method. J Biomech. 39(6): 1096–1106.

Davis RB, Õunpuu S, Tyburski D, Gage JR. 1991. A gait analysis data collection and reduction technique. Hum Movement Sci. 10(5):575–587.

Della Croce U, Cappozzo A, Kerrigan DC. 1999. Pelvis and lower limb anatomical landmark calibration precision and its propagation to bone geometry and joint angles. Med Biol Eng Comput. 37(2):155–161.

Della Croce U, Leardini A, Chiari L, Cappozzo A. 2005. Human movement analysis using stereophotogrammetry. Part 4: assessment of anatomical landmark misplacement

and its effects on joint kinematics. Gait Posture. 21(2): 226–237.
Ehrig RM, Taylor WR, Duda GN, Heller MO. 2006. A survey of formal methods for determining the centre of rotation of ball joints. J Biomech. 39(15):2798–2809.
Ehrig RM, Taylor WR, Duda GN, Heller MO. 2007. A survey of formal methods for determining functional joint axes. J Biomech. 40(10):2150–2157.
Gamage SS, Lasenby J. 2002. New least squares solutions for estimating the average centre of rotation and the axis of rotation. J Biomech. 35(1):87–93.
Harrington ME, Zavatsky AB, Lawson SEM, Yuan Z, Theologis TN. 2007. Prediction of the hip joint centre in adults, children, and patients with cerebral palsy based on magnetic resonance imaging. J Biomech. 40(3): 595–602.
Kainz H, Carty CP, Modenese L, Boyd RN, Lloyd DG. 2015. Estimation of the hip joint centre in human motion analysis: a systematic review. Clin Biomech (Bristol, Avon)). 30(4):319–329.
Koo TK, Li MY. 2016. A Guideline of Selecting and Reporting Intraclass Correlation Coefficients for Reliability Research. Journal of Chiropractic Medicine. J Chiropr Med. 15(2):155–163.
Kratzenstein S, Kornaropoulos EI, Ehrig RM, Heller MO, Popplau BM, Taylor WR. 2012. Effective marker placement for functional identification of the centre of rotation at the hip. Gait Posture. 36(3):482–486.
Lafortune MA, Cavanagh PR, Sommer HJ, Kalenak A. 1992. Three-dimensional kinematics of the human knee during walking. J Biomech. 25(4):347–357.
Leardini A, Chiari L, Croce UD, Cappozzo A. 2005. Human movement analysis using stereophotogrammetry. Part 3. Soft tissue artifact assessment and compensation. Gait Posture. 21(2):212–225.
Lopomo N, Sun L, Zaffagnini S, Giordano G, Safran MR. 2010. Evaluation of formal methods in hip joint center assessment: an in vitro analysis. Clin Biomech (Bristol, Avon). 25(3):206–212.
McGinley JL, Baker R, Wolfe R, Morris ME. 2009. The reliability of three-dimensional kinematic gait measurements: a systematic review. Gait Posture. 29(3):360–369.
Meng L, Childs C, Buis A. 2019. Evaluation of functional methods of joint centre determination for quasi-planar movement. PLoS One. 14(1):e0210807.
Millar LJ, Meng L, Rowe PJ. 2019. Routine clinical motion analysis: comparison of a bespoke real-time protocol to current clinical methods. Comput Methods Biomech Biomed Eng. 22(2):149–158.
Piazza SJ, Cavanagh PR. 2000. Measurement of the screw-home motion of the knee is sensitive to errors in axis alignment. J Biomech. 33(8):1029–1034.
Pohl MB, Lloyd C, Ferber R. 2010. Can the reliability of three-dimensional running kinematics be improved using functional joint methodology? Gait Posture. 32(4): 559–563.
Siston RA, Delp SL. 2006. Evaluation of a new algorithm to determine the hip joint center. J Biomech. 39(1):125–130.
Stagni R, Leardini A, Cappozzo A, Grazia Benedetti M, Cappello A. 2000. Effects of hip joint centre mislocation on gait analysis results. J Biomech. 33(11):1479–1487.
Van Campen A, De Groote F, Bosmans L, Scheys L, Jonkers I, De Schutter J. 2011. Functional knee axis based on isokinetic dynamometry data: comparison of two methods, MRI validation, and effect on knee joint kinematics. J Biomech. 44(15):2595–2600.
Wu G, Siegler S, Allard P, Kirtley C, Leardini A, Rosenbaum D, Whittle M, D'Lima DD, Cristofolini L, Witte H, et al. 2002. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. J Biomech. 35(4):543–548.
Zuk M, Swiatek-Najwer E, Pezowicz C. 2014. Hip joint centre localization: evaluation of formal methods and effects on joint kinematics. Adv Intell Syst. 284:57–67.