Gait & Posture 35 (2012) 255–260

Contents lists available at SciVerse ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Repeatability in the assessment of multi-segment foot kinematics

Kevin Deschamps [a,b,c,d,*], Filip Staes [b], Herman Bruyninckx [e], Ellen Busschots [a], Ellen Jaspers [f], Ameya Atre [e], Kaat Desloovere [c,f]

[a] Division of Musculoskeletal Disorders, University Hospital Leuven, Weligerveld 1, 3212 Pellenberg, Belgium
[b] Department of Rehabiliation Sciences-Research Center for Musculoskeletal Rehabilitation, Katholieke Universiteit Leuven, Tervuursevest 101, B-3001 Leuven (Heverlee), Belgium
[c] Clinical Motion Anaysis Laboratory, University Hospital Pellenberg, Weligerveld 1, 3212 Pellenberg, Belgium
[d] Institut d' Enseignment Supérieur Parnasse Deux-Alice, Division of Podiatry, 84 Avenue E. Mounier, 1200 Bruxelles, Belgium
[e] Department of Mechanical Engineering, Katholieke Universiteit Leuven, Celestijnenlaan 300B, B-3001 Leuven (Heverlee), Belgium
[f] Department of Rehabilitation Sciences - Research Center for Neuromotor, Pediatric and Pelvic Rehabilitation, Katholieke Universiteit Leuven, Tervuursevest 1001, B-3001 Leuven (Heverlee), Belgium

ARTICLE INFO

Article history:
Received 26 December 2010
Received in revised form 14 September 2011
Accepted 17 September 2011

Keywords:
Multi-segment foot kinematics
Reproducibility
Repeatability
Gait analysis

ABSTRACT

A recently published systematic review on 3D multi-segment foot models has illustrated the lack of repeatability studies providing evidence for appropriate clinical decision making. The aim of the current study was to assess the repeatability of the recently published model developed by Leardini et al. [10]. Foot kinematics of six healthy adults were analyzed through a repeated-measures design including two therapists with different levels of experience and four test sessions.

For the majority of the parameters moderate or good repeatability was observed for the within-day and between-day sessions. A trend towards consistently higher within- and between-day variability was observed for the junior compared to the senior clinician. The mean inter-session variability of the relative 3D rotations ranged between 0.9–4.2° and 1.6–5.0° for respectively the senior and junior clinician whereas for the absolute angles this variability increased to respectively 2.0–6.2° and 2.6–7.8°. Mean inter-therapist standard deviations ranged between 2.2° and 6.5° for the relative 3D rotations and between 2.8° and 7.6° for the absolute 3D rotations. The ratio of inter-therapist to inter-trial errors ranged between 1.8 and 5.5 for the relative 3D rotations and between 2.4 and 9.7 for the absolute 3D rotations. Absolute angle representation of the planar angles was found to be more difficult.

Observations from the current study indicate that an adequate normative database can be installed in gait laboratories, however, it should be stressed that experience of therapists is important and gait laboratories should therefore be encouraged to put effort in training their clinicians.

1. Introduction

Stereophotogrammetry has been used for two decades to represent shank and foot multi-segment kinematics. Review of the literature demonstrates that several 3D multi-segment foot models (3DMFMs) have been published [1–10], embodying the international consensus that the foot should be modeled as a number of segments [11]. However, a recent review highlighted a lack of standardization as well as adequate repeatability studies [12]. As a consequence, clinical utility of several 3DMFMs still has to be determined.

Leardini et al. [10] recently described an interesting 3D multi-segment foot model (Leardini Foot Model: LFM) protocol. Noteworthy differences with some other 3DMFMs are the specific marker placement at the calcaneus, the indirect calculation of landmarks (by mid-point calculation), the fact that the midfoot is considered as one rigid segment and the calculation of several planar angles. The latter two components enforce the clinical relevance of the LFM, especially with respect to foot pathologies [8,12].

With respect to the LFM protocol, reference waveforms for the stance phase of gait have been published for a group of 10 subjects, illustrating 'typical' patterns for all 3D rotations and planar angles [10]. Furthermore, the effect of walking cadence on the kinematic waveforms [13], as well as repeatability indices from a repeated measure design have been reported [14]. Results of the latter indicated that most of the 3D rotations can be measured in a repeatable way by experienced clinicians.

Although several studies contributed to an improved understanding of the LFM protocol, it should be stressed that a full

* Corresponding author at: Division of Musculoskeletal Disorders, University Hospital – Leuven, K.U. Leuven at Weligerveld 1, 3212 Pellenberg, Belgium. Tel.: +32 16 338 024.
E-mail address: kevin.deschamps@uz.kuleuven.ac.be (K. Deschamps).

0966-6362/$ – see front matter © 2011 Elsevier B.V. All rights reserved.
doi:10.1016/j.gaitpost.2011.09.016

picture of the clinical features has not yet been provided in literature. Moreover, kinematic waveforms have only been published for the stance phase of gait despite the fact that other studies have demonstrated the relevance to report the swing phase [15–17]. Still, replication of the study by a group independent from the authors of the LFM protocol is lacking, thereby limiting the strength of the conclusions on the reproducibility of the protocol [18]. Finally, intrinsic and extrinsic errors have only been quantified for the 3D rotations and not for the planar angles. Knowing consistency and variability of these planar angles is helpful, as it may guide the use and interpretation [19].

Consequently, the aim of the present study was to further explore the kinematic waveforms obtained from the LFM [10], to provide further insight in the repeatability of the protocol in the presence of different expertise levels, as well as to quantify the repeatability of all parameters (3D rotations and planar angles).

2. Methods

2.1. Participants

Six symptom-free adult volunteers (three men, three women, age range 22–54 years, BMI 22.2 ± 2.5) were recruited through advertisements at the institution's gait laboratory for a period of 3 months. Potential participants were excluded if they had a history of trauma of the foot and lower limbs, if they presented with a foot deformity (screened via standard clinical examination) or with a systemic or neurological disorder.

2.2. Design

A repeated measure design was used in which each participant underwent eight 3D gait analysis sessions; four sessions conducted by each of two staff podiatrists (Fig. 1). Two sessions per subject and therapist were performed on the same day, 1 h apart and again 1 week later. All markers were removed and reapplied for each session. Both therapists were blinded to the marker placement of the other therapist in order to limit potential bias.

Two staff podiatrists, with a different level of experience in gait analysis, were involved in this study. Therapist one (senior) had 5 years of clinical experience and 3 years in 3D gait analysis. Therapist two (junior) had 1 year of clinical experience and no experience with 3D gait analysis nor with 3DMFM protocols. Training of both therapists was organized prior to the initiation of the current study and involved the placement of markers at the anatomical landmarks related to the 3DMFM protocol in a number of colleagues. During this training a discrepancy was observed in marker placement and a 'consensus' session was organized in order to finalize the anatomical placement of the markers. In fact, certain markers have to be placed on a specific anatomical point (e.g. sustentaculum tali), whereas for other markers this distinct anatomical point is not present (e.g. base of the first metatarsal).

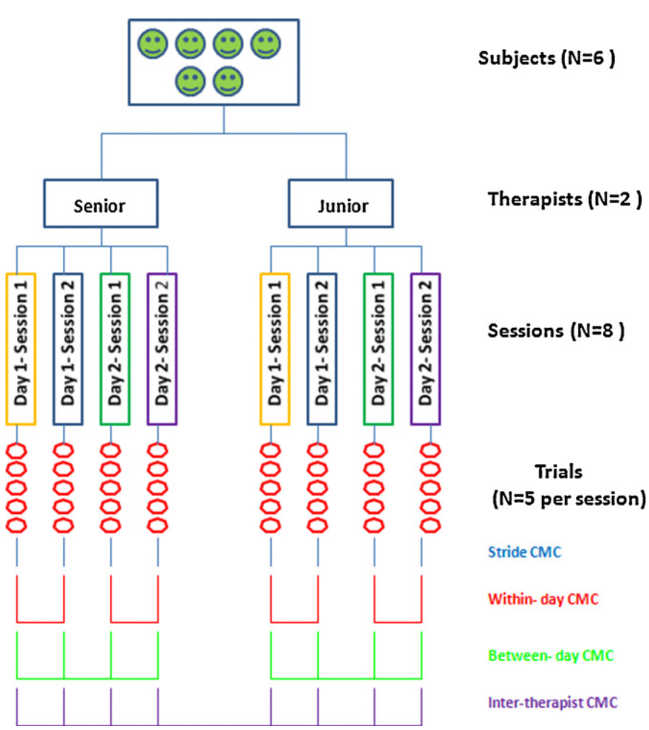

Fig. 1. Illustration of the repeated-measures experimental design applied in the current study and the trials that have been considered for the specific CMC calculations. A mean value was calculated for the stride- and within-day CMCs.

2.3. 3D multi-segment foot analysis

Fourteen 6 mm retro-reflective markers were mounted over the described anatomical landmarks according to the protocol of Leardini et al. [10]. After marker placement, a standing trial in a relaxed position was recorded. Subsequently, the measurements of the dynamic trials started, with individuals walking at a self-selected speed until five representative walking trials were recorded. Marker trajectories were acquired using an optoelectronic system with 10 T-10 cameras and Nexus motion capture software (Vicon Motion System Ltd., Oxford Metrics, UK). Temporal parameters of all gait cycles were determined using two force plates (Advanced Mechanical Technology Inc., Watertown, MA) and identification of gait events within the software. Kinematic data were sampled at 100 Hz, kinetic data at 200 Hz. The local Ethics Committee granted approval for the study and written informed consent was obtained from each subject.

2.4. Data analysis and statistics

Three-dimensional rotations were calculated between shank and calcaneus (Sha–Cal), calcaneus and midfoot (Cal–Mid), midfoot and metatarsus (Mid–Met) as well as calcaneus and metatarsus (Cal–Met). The planar angles of interest were: first metatarsus and ground (F2G), second metatarsus and ground (S2G), fifth metatarsus and ground (V2G), first metatarsal and proximal phalanx in transverse plane (F2Pt), second and first metatarsus (S2F), second and fifth metatarsus (S2V), first metatarsal and proximal phalanx in the sagittal plane (F2Ps) and calcaneus and first metatarsus (MLA) [10]. Kinematic data were computed throughout the Vicon Foot model Plug-in (Aurion Srl, Milano, Italy) using Nexus 1.5 software (Vicon Motion System Ltd., Oxford Metrics, UK).

Data from the standing trials were used to calculate offset values for both 3D rotations and planar angles. To fully explore the clinical utility of the 3DMFM, both relative and absolute angles were considered. In the former, offset values were subtracted from the corresponding dynamic values whereas in the latter this offset subtraction was not performed. Matlab® was used to time normalize the data to 100% gait cycle (GC) (1% interval between time points) and for further statistical analysis on both relative and absolute gait data.

Repeatability, defined as the variation in measurements performed by therapists on the same participant(s) and under the same test conditions, was analyzed in different ways.

Inter-trial coefficients of multiple correlation (CMC) values were calculated separately for each measurement session and subsequently an average was calculated [20]. Within-day CMC calculations were based on selecting all trials performed on the same day and subsequently taking an average (Additional File 1). Between-day CMC values were obtained by considering all trials over the four sessions of each therapist. The inter-trial errors (σ^{trial}) and the inter-session errors (σ^{sess}), as proposed by Schwartz et al. [21], were also calculated for all 3D rotations and planar angles. For the σ^{trial} calculations all trials of both therapists were considered whereas for the σ^{sess} calculations all trials of each therapist separately were considered.

Variability between both therapists was evaluated by calculating the inter-therapist CMCs, the inter-therapist errors ($\sigma^{\text{therapist}}$) and the inter-therapist/inter-trial error ratio ($\sigma^{\text{therapist/trial}}$) [21]. For the first two repeatability indices all trials of both therapists were considered. The mean range of motion occurring during five subphases of the gait cycle (loading response 0–10% GC, midstance 10–30% GC, terminal stance 30–50% GC, preswing 50–60% and swing 60–100% GC) were also compared between both therapists using the Mann–Whitney test.

To simplify data interpretation, a mean was calculated for the σ^{trial}, σ^{sess} and $\sigma^{\text{therapist}}$ as these are normally point-by-point error representations. All CMC values were interpreted following the benchmarks proposed by Garofalo et al. [22].

3. Results

The self-selected speed adopted by the subjects resulted in a mean stance phase duration of 0.69 s (±0.07 s) and a mean cadence of 107.4 steps/min (±10.6). Distinct patterns for all 3D inter-segmental rotations and planar angles were obtained.

Mean inter-trial CMCs were >0.820 for both therapists (Table 1). With respect to the relative angles, within-day CMC values ranged between 0.782 and 0.987 for the senior clinician, whereas for the junior clinician these indices ranged between

Table 1
Range of motion and mean standard deviations in degrees and inter-trial, within-day, between-day and inter-therapist coefficient of multiple correlation (CMC).

	ROM[a]			Relative angles					Absolute angles				
		Senior	Junior	Senior	Junior	Senior	Junior	Senior–junior	Senior	Junior	Senior	Junior	Senior–junior
		Inter-trial CMC[b]	Inter-trial CMC [b]	Within-day CMC [b]	Within-day CMC[b]	Between-day CMC	Between-day CMC	Inter-therapist CMC	Within-day CMC[b]	Within-day CMC[b]	Between-day CMC	Between-day CMC	Inter-therapist CMC
Sha–Cal Do/PF	33.2 ± 5.1	0.968	0.987	0.923	0.844	0.933	0.821	0.791	0.911	0.810	0.890	0.786	0.688
Sha–Cal Inv/Eve	22.5 ± 3.1	0.935	0.879	0.912	0.903	0.899	0.851	0.761	0.872	0.829	0.811	0.808	0.701
Sha–Cal Add/Abd	12.9 ± 1.2	0.907	0.853	0.877	0.848	0.854	0.731	0.710	0.884	0.808	0.840	0.749	0.633
Cal–Mid Do/PF	20.8 ± 6.2	0.981	0.960	0.952	0.943	0.842	0.787	0.891	0.814	0.855	0.733	0.833	0.471
Cal–Mid Inv/Eve	13.0 ± 4.2	0.964	0.947	0.933	0.891	0.831	0.744	0.811	0.797	0.823	0.712	0.746	0.725
Cal–Mid Add/Abd	15.9 ± 10.1	0.881	0.864	0.782	0.710	0.801	0.661	0.761	0.733	0.718	0.699	0.698	0.655
Mid–Met Do/PF	14.9 ± 8.1	0.928	0.962	0.837	0.821	0.741	0.701	0.516	0.786	0.841	0.819	0.823	0.490
Mid–Met Inv/Eve	15.9 ± 5.3	0.924	0.930	0.867	0.877	0.801	0.792	0.775	0.799	0.671	0.686	0.541	0.248
Mid–Met Add/Abd	4.3 ± 3.5	0.908	0.820	0.851	0.673	0.761	0.557	0.448	0.735	0.555	0.652	0.512	0.390
Cal–Met Do/PF	22.9 ± 7.8	0.974	0.939	0.870	0.821	0.861	0.755	0.801	0.688	0.789	0.619	0.718	0.499
Cal–Met Inv/Eve	20.1 ± 6.3	0.942	0.887	0.799	0.788	0.701	0.659	0.799	0.442	0.699	0.180	0.675	0.432
Cal–Met Add/Abd	12.2 ± 4.3	0.920	0.889	0.890	0.791	0.881	0.690	0.837	0.610	0.512	0.532	0.499	0.321
F2G	92.5 ± 8.5	0.992	0.996	0.963	0.969	0.966	0.955	0.932	0.861	0.951	0.712	0.912	0.765
S2G	88.4 ± 6.9	0.994	0.982	0.972	0.977	0.952	0.912	0.923	0.875	0.944	0.844	0.899	0.831
V2G	99.0 ± 5.3	0.992	0.993	0.987	0.991	0.971	0.982	0.955	0.888	0.831	0.851	0.799	0.809
F2Pt	11.9 ± 3.5	0.920	0.910	0.901	0.882	0.862	0.851	0.745	0.681	0.765	0.623	0.644	0.451
S2F	15.0 ± 9.7	0.923	0.891	0.880	0.762	0.744	0.666	0.634	0.851	0.781	0.767	0.731	0.645
S2V	36.8 ± 13.6	0.939	0.971	0.861	0.931	0.801	0.880	0.811	0.771	0.541	0.699	0.089	0.031
F2Ps	40.0 ± 13.6	0.987	0.960	0.954	0.823	0.851	0.730	0.752	0.515	0.510	0.246	0.267	0.181
MLA	28.3 ± 13.2	0.936	0.966	0.872	0.853	0.865	0.811	0.654	0.413	0.615	0.199	0.100	0.061

[a] ROM: range of motion, mean calculation for both therapists (senior and junior).
[b] A mean value has been calculated for the inter-trial and within-day CMC values. Inter-trial CMC values are only reported for the relative angles (no significant differences are expected with the absolute angles).

Table 2
Mean range of motion (°) measured by both therapists during five subphases of the gait cycle. Statistical differences were analyzed using Mann–Whitney test ($p < 0.05$).

	Loading response			Midstance			Terminal stance			Preswing			Swing		
	Senior	Junior	*p* value	Senior	Junior	*p* value	Senior	Junior	*p* value	Senior	Junior	*p* value	Senior	Junior	*p* value
Sha–Cal Do/PF	7.7 (2.1)	7.1 (1.5)	ns	13.4 (2.9)	12.1 (1.1)	ns	9.8 (3.7)	9.4 (3.1)	ns	30.3 (7.0)	28.1 (6.4)	ns	31.0 (4.6)	31.3 (7.9)	ns
Sha–Cal Inv/Eve	8.4 (2.0)	7.3 (2.0)	ns	10.0 (1.8)	9.5 (2.5)	ns	11.9 (1.8)	13.2 (2.9)	ns	16.6 (3.6)	16.9 (2.8)	ns	22.7 (3.6)	22.2 (4.1)	ns
Sha–Cal Add/Abd	9.3 (3.7)	9.7 (3.2)	ns	7.8 (2.0)	6.7 (1.0)	ns	11.5 (3.4)	9.0 (1.6)	0.035*	10.4 (1.5)	11.6 (1.8)	ns	14.1 (3.2)	17.3 (5.3)	ns
Cal–Mid Do/PF	7.0 (2.2)	7.5 (1.3)	ns	7.6 (1.9)	9.6 (2.4)	0.017*	8.5 (3.0)	11.2 (3.1)	ns	16.7 (2.6)	21.6 (3.3)	0.003*	14.2 (5.1)	16.8 (2.4)	ns
Cal–Mid Inv/Eve	5.3 (1.1)	7.6 (1.6)	ns	4.5 (0.8)	7.7 (3.0)	ns	6.1 (2.0)	9.0 (2.5)	ns	10.4 (3.7)	13.1 (3.1)	ns	12.2 (5.6)	15.3 (3.4)	ns
Cal–Mid Add/Abd	7.7 (3.4)	8.1 (2.3)	ns	5.9 (1.2)	8.0 (3.6)	ns	8.6 (3.1)	8.7 (2.6)	ns	8.4 (2.8)	12.6 (2.9)	ns	12.4 (4.2)	15.2 (4.7)	ns
Mid–Met Do/PF	6.6 (2.2)	8.8 (1.5)	ns	6.8 (1.8)	9.3 (1.6)	ns	10.8 (4.2)	12.2 (5.4)	ns	11.0 (4.7)	11.5 (6.2)	ns	12.5 (2.3)	14.2 (3.2)	ns
Mid–Met Inv/Eve	6.5 (2.6)	9.2 (2.6)	ns	6.1 (2.2)	9.9 (1.7)	ns	7.3 (1.9)	9.7 (2.8)	ns	12.5 (3.3)	14.2 (2.7)	ns	11.6 (4.5)	14.6 (2.7)	ns
Mid/Met Abd/Add	4.0 (0.5)	5.6 (2.1)	ns	4.2 (0.7)	6.1 (2.7)	ns	6.0 (1.8)	7.0 (2.8)	ns	4.2 (0.9)	5.6 (2.3)	ns	6.6 (2.5)	8.3 (4.6)	ns
Cal–Met Do/PF	7.1 (1.8)	9.5 (5.8)	ns	8.2 (1.4)	10.5 (7.8)	ns	11.1 (5.7)	15.7 (12.0)	ns	19.3 (5.4)	20.7 (5.1)	ns	21.1 (7.8)	20.6 (5.4)	ns
Cal–Met Inv/Eve	8.8 (2.3)	9.6 (2.3)	ns	9.4 (1.8)	10.8 (3.2)	ns	13.1 (6.1)	17.1 (5.0)	ns	15.4 (3.7)	16.8 (4.0)	ns	16.4 (6.1)	21.0 (6.5)	ns
Cal–Met Add/Abd	4.8 (1.6)	5.8 (1.5)	ns	4.4 (1.1)	6.3 (1.2)	0.042*	5.7 (1.9)	7.6 (1.4)	0.011*	11.0 (2.8)	13.2 (4.0)	0.047*	11.4 (3.4)	14.7 (7.0)	ns
F2G angle	23.6 (3.4)	23.2 (4.1)	ns	8.2 (3.5)	7.5 (3.5)	ns	23.1 (5.1)	26.7 (4.2)	ns	41.8 (7.4)	58.2 (5.7)	0.001*	91.9 (7.7)	104.8 (6.2)	0.015*
S2G angle	20.6 (2.2)	22.5 (2.7)	ns	5.4 (2.7)	6.9 (2.7)	ns	21.1 (3.3)	22.5 (2.0)	ns	47.0 (5.9)	51.3 (7.3)	ns	88.9 (6.1)	94.7 (6.0)	ns
V2G angle	23.4 (1.7)	23.2 (5.7)	ns	5.6 (3.1)	6.8 (3.4)	ns	21.5 (5.1)	22.8 (2.4)	ns	54.7 (8.9)	54.3 (5.9)	ns	104.1 (5.5)	103.6 (5.4)	ns
F2Pt angle	9.1 (4.5)	9.4 (2.9)	ns	7.6 (1.1)	8.2 (2.4)	ns	10.4 (2.4)	9.9 (2.4)	ns	10.9 (2.2)	11.1 (2.5)	ns	13.9 (1.7)	14.7 (2.2)	ns
S2F angle	9.3 (5.9)	10.9 (4.7)	ns	8.8 (3.5)	12.7 (8.6)	ns	10.5 (4.0)	16.1 (12.3)	ns	10.5 (4.1)	11.9 (9.6)	ns	11.9 (4.4)	16.5 (5.5)	ns
S2V angle	17.3 (5.2)	17.3 (6.3)	ns	16.7 (2.5)	17.0 (3.7)	ns	14.8 (3.8)	19.3 (8.7)	ns	20.1 (5.6)	22.6 (8.7)	ns	20.5 (7.7)	21.4 (9.4)	ns
F2PS angle	13.4 (3.9)	14.9 (5.3)	ns	10.9 (4.5)	14.4 (5.3)	ns	22.1 (4.1)	22.6 (6.5)	ns	31.4 (9.0)	29.6 (9.3)	ns	32.2 (10.4)	33.2 (8.6)	ns
MLA	14.4 (6.1)	20.0 (9.0)	ns	13.5 (2.7)	24.3 (15.1)	ns	16.3 (3.6)	25.3 (14.1)	ns	28.1 (8.3)	39.1 (9.5)	0.038*	30.5 (19.9)	39.0 (9.3)	ns

ns = not significant ($p > 0.05$).
* $p < 0.05$.

0.673 and 0.991. Between-day CMCs ranged between 0.701 and 0.971 for the senior clinician and between 0.557 and 0.982 for the junior clinician. Similar calculations for the absolute angles showed consistently lower within and between-day CMC values (Table 1).

The mean ROM measured by both therapists for all parameters throughout the five subphases of the gait cycle were in 91% of the cases similar and only in 9% of the cases some significant differences were observed (Table 2).

Mean σ^{trial} were <2.0° for all angles with exception of the planar angles F2G, S2G and V2G (Fig. 1). Mean σ^{sess} of the relative 3D rotations ranged between 0.9° and 4.2° for the senior therapist and between 1.6° and 5.0° for the junior therapist. With respect to the planar angles, mean relative σ^{sess} ranged between 2.0° and 3.5° for the senior clinician and between 2.4° and 4.4° for the junior clinician. In the majority of the cases the σ^{sess} of the absolute 3D rotations and planar angles were higher; particularly in certain planar angles a considerable variability was observed. A trend towards a lower inter-session variability was observed for the senior clinician with respect to the junior clinician.

Inter-therapist CMCs of the relative 3D rotations ranged from 0.448 to 0.891, whereas these indices ranged from 0.634 to 0.955 in the relative planar angles (Table 1). Again, a negative trend was observed in these repeatability indices when absolute angles were considered.

The mean $\sigma^{therapist}$ ranged between 2.2° and 6.5° for the relative 3D rotations and increased to a range between 2.8° and 7.6° in the absolute 3D rotations. With respect to the planar angles, this error also increased from 2.2° to 6.5° to a range of 3.6–18.2° for respectively relative and absolute angles (Fig. 2).

Eighty percent (16/20) of the relative parameters had a $\sigma^{therapist/trial} \leq 4.0$, while, only 35% of the absolute parameters had a ratio ≤4.0 (Table 3).

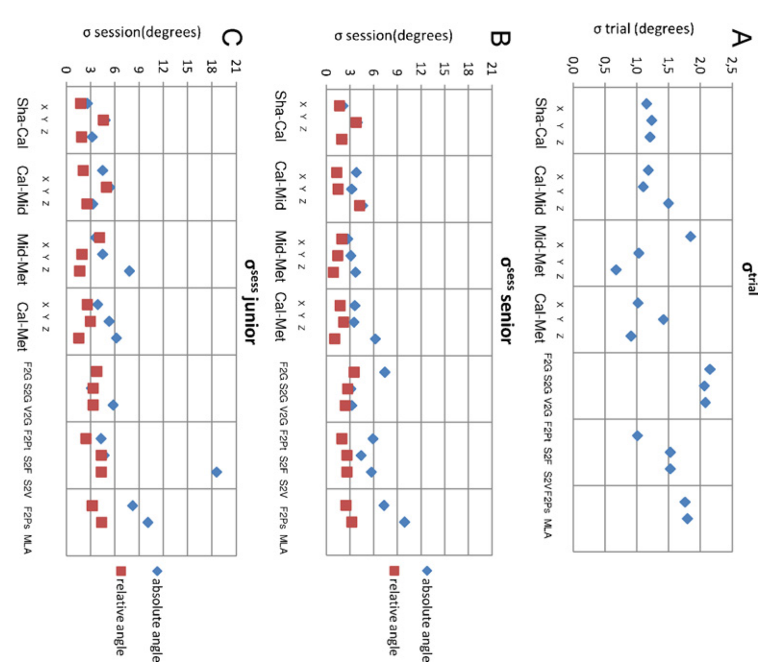

Fig. 2. Summary of the different experimental errors found for the inter-segmental and planar angles with (A) σ^{trial}, (B) σ^{sess} senior, and (C) σ^{sess} junior (X = DO/PF, Y = ADD/ABD, Z = INV/EVE). A mean was calculated for all errors as these were found to be constant over the complete gait cycle.

Table 3
Comparison of the mean inter-therapist σ and inter-therapist/inter-trial errors (as reported by Schwartz et al. [21]) for conventional lower limb model and those of the current study. A distinction is made for the relative angles and absolute angles.

	Mean σtherapist (Schwartz et al. [21])	σtherapist/trial (Schwartz et al. [21])		Relative angles		Absolute angles	
				Mean σtherapist	σtherapist/trial	Mean σtherapist	σtherapist/trial
Pelvic obliquity	1.5	1.4	Sha–Cal DO/PF	2.5	2.2	2.8	2.4
Knee flexion/extension	3.4	2.1	Sha–Cal Add/Abd	6.5	5.3	6.4	5.2
Pelvic rotation	1.2	2.6	Sha–Cal Inv/Eve	2.2	1.8	3.2	2.7
Hip flexion/extension	3.5	3.0	Cal–Mid DO/PF	3.1	2.6	6.1	5.2
Foot progression	5.3	3.2	Cal–Mid Add/Abd	6.0	5.5	7.1	6.5
Hip abduction/adduction	2.2	3.8	Cal–Mid Inv/Eve	5.1	3.4	6.1	4.1
Pelvic tilt	2.8	4.0	Mid–Met DO/PF	4.8	2.6	5.1	2.8
Hip transverse plane rotation	4.5	4.2	Mid–Met Add/Abd	3.2	3.1	4.4	4.3
Knee varus/valgus	0.5	5.0	Mid–Met Inv/Eve	3.5	5.2	6.5	9.7
			Cal–Met DO/PF	2.9	2.8	5.9	5.8
			Cal–Met Add/Abd	4.2	3.0	6.1	4.3
			Cal–Met Inv/Eve	4.1	4.5	7.6	8.4
			F2G angle	4.6	2.1	10.1	4.7
			S2G angle	3.5	1.7	3.6	1.7
			V2G angle	3.1	1.5	5.4	2.6
			F2Pt angle	2.5	2.5	7.4	7.3
			S2F angle	4.8	3.1	6.1	4.0
			S2V angle	4.8	3.1	16.1	10.5
			F2Ps angle	3.1	1.8	13.1	7.4
			MLA	5.2	2.9	18.2	10.1

4. Discussion

The present study was designed to independently assess the repeatability of the Leardini Foot Model protocol [10]. Measuring the extent to which gait measurements are consistent or free from variation is critical if appropriate clinical use is to be pursued [19,21]. Several earlier studies have focused on the repeatability of kinematic data from a 3DMFM, though, often with a less meticulous repeated-measure design. In fact, models with the highest level of scientific credibility are characterized by repeatability assessment of between-trial, between-day and between- and within-therapists studies [2,5,8,12,23–26]. Compared to previous studies, the results of the current study are promising (Table 1). Using the CMC benchmarks proposed by Garofalo et al. [22], it can be concluded that the majority of the parameters had moderate and good similarity for within- and between-day sessions. A trend towards consistently lower within- and between-day CMCs was observed in the junior clinician, indicating that training and experience are important factors to obtain repeatable results. The latter observation has also been stressed by other authors [19]. In addition, it can be observed that within-day CMCs are generally higher compared to between-day CMCs, which may indicate some 'memory effect' between both sessions of a day.

The σ^{trial} were small for all 3D rotations (range between 0.7° and 1.5°). This finding is similar to the σ^{trial} recently reported by Caravaggi et al. [14]. The current study also found comparable σ^{trial} for the planar angles, though, comparison with existing literature was not possible.

A trend towards a higher σ^{sess} for the junior therapist was observed in the current study. However, this phenomenon seems not to be of the same order as that reported in the study of Caravaggi et al. [14]. A plausible explanation may be related to the training that was given to the junior clinician prior to the start of the current study. Furthermore, it is also reasonable to assume that the specific background of the junior clinician, i.e. podiatry, results in more consistent marker placement and better palpation of anatomical reference points, as treatment of foot disorders is of primary interest in this profession. In this perspective, it is also reasonable to assume that additional training may further optimize the repeatability of the protocol. One should be careful when comparing the current σ^{sess} values to those proposed by Caravaggi et al. [14] as these authors omitted to report in their technical note whether offset subtraction, recorded from static trials, was performed on the dynamic trials. In fact, it should be stressed that such subtraction was initially proposed in the methodological paper of Leardini et al. [10]. It was decided in the current study to perform statistical analyses on both relative as well as absolute angles, as this would better illustrate the clinical utility of the LFM. In fact, it can be assumed that clinical decision making makes more sense based on absolute angles as this is easier to incorporate in their reasoning. Typical examples of the latter are the planar angles which often reflect specific radiographic angles [8].

In the current study, it was observed that the use of absolute angles did not have a critical impact on the variability of 3D rotations; however, for the planar angles this was found to be of greater impact. Especially the S2F angle, S2V and MLA angles were found to be difficult to measure when expressed as absolute values. In this perspective, it is reasonable to accept that one would, at this moment, only rely on range of motion data throughout his pathomechanical reflection. Evidence for this was provided in Table 2, which illustrates in a majority of the cases similar range of motion measurements throughout the different gait phases between both therapists. Meanwhile, additional efforts in optimizing the repeatability of the LFM protocol may, at a certain time point, shed new light on the use of absolute values.

Mean $\sigma^{therapist}$ were of greater magnitude when compared to reported data from a conventional lower limb model (Table 3), although this did not have a dramatic impact on the $\sigma^{therapist/trial}$. In general, similar $\sigma^{therapist/trial}$ were reported by Caravaggi et al. [14] when compared to the results obtained with the absolute angles. Interestingly, Caravaggi et al. [14] found that marker placement at the calcaneus was difficult in non-experienced clinicians. In particular, marker placement at the sustentaculum tali and tuberculum peroneum were found to cause larger deviations of the orientation of the calcaneus reference frame. In addition to these observations, it is of paramount importance to emphasize that a small shift (e.g. 5 mm) in marker position may result in considerable kinematic differences [27] and that anatomical landmarks in the foot may be difficult to palpate or even be absent [28,29].

A considerable part of this study focused on replicating the protocol proposed by Leardini et al. [10], in an independent study while using the commercially available Vicon Plug-in. Overall, distinct waveforms for all calculated angles have been observed throughout both the stance and swing phase. Moreover, visual comparison showed similar patterns with the reported waveforms in the original study [10] (Additional Files 1 and 2). Similar conclusions can be drawn when comparing the data of the current study with those reported by Caravaggi et al. [13] during their 'normal' cadence measurements. In fact, this study demonstrated the critical impact of walking cadence on multi-segment foot kinematics.

Despite the similar patterns between the studies, detailed analysis showed small differences. The Sha-Cal inversion and Cal-Mid plantarflexion during terminal stance and toe off were found to be more pronounced in our group. Calcaneus–Midfoot adduction during push off was also found to be more pronounced, which is probably associated with the increased inversion at the Sha–Cal during push-off. A marked dorsiflexion at the Mid–Met angle was observed during terminal stance, followed by a rapid plantarflexion at toe off. Finally, at the Cal–Met angle, a marked inversion during propulsion was observed. This was probably again related to the increased inversion of the calcaneus. The planar angles also showed typical patterns (Additional File 3), which was in good agreement with published data [10]. A distinct 'double bump' was observed at F2G and S2G during toe off and initial swing. Being associated with an adequate foot clearance, this insight was not fully provided by the original paper as they omitted to report on swing phase [10]. It should be emphasized that, for the current study, offset values were subtracted from each individual planar angle. The impact of this offset subtraction is most obvious in the MLA angle.

Even though the present study included an appropriate repeated measures design for testing repeatability, it might be interesting to perform a multi-center assessment of the 3DMFM. Such an investigation implies that the same participants are measured in different gait labs, ideally throughout a repeated-measure design at each site. To our knowledge, the Milwaukee Foot Model is the only 3DMFM protocol that is supported by such scientific evidence [23].

5. Conclusion

The current study has illustrated the repeatability of the 3DMFM protocol proposed by Leardini et al. [10]. Kinematic data can be estimated in a repeatable way by an individual therapist, hence illustrating its clinical utility for other gait laboratories. The repeatability was found to be adequate for a number of 3D rotations and planar angles, indicating that an adequate normative database can be installed in gait laboratories. However, it should be emphasized that experience seems to play a critical role. Moreover, it should be recognized that generalization of the current findings should be handled with care. In fact, future studies should focus on the repeatability of 3DMFM protocols in presence of foot deformities. Currently literature is scarce regarding relevant repeatability studies in patient groups, despite the observation of several clinical trials in literature.

Acknowledgments

The authors are grateful to the Agency for Innovation by Science and Technology Flanders for funding this project (Grant: 080659). They would also like to thank Christophe Meyer for his technical support during the initiation of the project.

Conflict of interest

The authors declare that no financial and personal relationship exist which could have influence (bias) their work.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.gaitpost.2011.09.016.

References

[1] Kidder SM, Abuzzahab FS, Harris GF, Johnson JE. A system for the analysis of foot and ankle kinematics during gait. IEEE Trans Rehabil Eng 1996;4(1):25–32.
[2] Rattanaprasert U, Smith R, Sullivan M, Gilleard W. Three dimensional kinematics of the forefoot, rearfoot, and leg without the function of tibialis posterior in comparison with normal during stance phase of walking. Clin Biomech 1998;14:14–23.
[3] Cornwall MW, McPoil TG. Three dimensional movement of the foot during the stance phase of walking. J Am Podiatr Med Assoc 1999;89(2):56–66.
[4] Leardini A, Benedetti MG, Catani F, Simoncini L, Giannini S. An anatomically based protocol for the description of foot segment kinematics during gait. Clin Biomech 1999;14:528–36.
[5] Carson MC, Harrington ME, Thompson N, O'Conner JJ, Theologis TN. Kinematic analysis of a multi-segment foot model for research and clinical applications: a repeatability analysis. J Biomech 2001;34:1299–307.
[6] Arampatzis A, Brüggemann G, Morey Klapsing G. A three-dimensional shank-foot model to determine the foot motion during landings. Med Sci Sports Exerc 2002;34(1):130–8.
[7] Mac Williams BA, Cowley M, Nicholson DE. Foot kinematics and kinetics during adolescent gait. Gait Posture 2003;17:214–24.
[8] Simon J, Doederlein L, McCintosh AS, Mataxiotis D, Block HG, Wolf SI. The Heidelberg foot measurement method: development, description and assessment. Gait Posture 2006;23:411–24.
[9] Tome J, Nawoczenski D, Flemister A, Houck J. Comparison of foot kinematics between subjects with posterior tibialis tendon dysfunction and healthy controls. J Orthop Sports Phys Ther 2006;3(9):635–44.
[10] Leardini A, Benedetti MG, Berti L, Bettinelli D, Nativo R, Giannini S. Rear-foot, mid-foot and fore-foot motion during the stance phase of gait. Gait Posture 2007;25:453–62.
[11] Baker R, Robb J. Foot models for clinical gait analysis. Gait Posture 2006; 23:399–400.
[12] Deschamps K, Staes F, Roosen P, Nobels F, Desloovere K, Bruyninckx H, et al. Body of evidence supporting the clinical use of 3D multisegment foot models: a systematic review. Gait Posture 2011;33(3):338–49.
[13] Caravaggi P, Leardini A, Crompton R. Kinematic correlates of walking cadence in the foot. J Biomech 2010;43:2425–33.
[14] Caravaggi P, Benedetti MG, Berti L, Leardini A. Repeatability of a multisegment foot protocol in adult subjects. Gait Posture 2011;33(1):133–5.
[15] Khazzam M, Long JT, Marks RM, Harris GF. Preoperative gait characterization of patients with ankle arthrosis. Gait Posture 2006;1(24):85–93.
[16] Ness ME, Long J, Marks R, Harris G. Foot and ankle kinematics in patients with posterior tibial tendon dysfunction. Gait Posture 2007;27:331–9.
[17] Khazzam M, Long JT, Marks RM, Harris GF. Kinematic changes of the foot and ankle in patients with systematic rheumatoid arthritis and forefoot deformity. J Othop Res 2007;3(25):319–29.
[18] McGraw KO, Wong SP. Forming inferences about some intraclass correlation coefficients. Psychol Methods 1996;1:30.
[19] McGinley JL, Baker R, Wolfe R, Morris ME. The reliability of three-dimensional kinematic gait measurements: a systematic review. Gait Posture 2009; 29:360–9.
[20] Kadaba MP, Ramakrishnan HK, Wooten ME, Gainey J, Gorton G, Cochran GVB. Repeatability of kinematic, kinetic and electromyographic data in normal gait. J Orthop Res 1989;7:849–60.
[21] Schwartz MH, Trost JP, Wervey RA. Measurement and management of errors in quantitative gait data. Gait Posture 2004;20:196–203.
[22] Garofalo P, Cutti AG, Filippi MV, Cavazza S, Ferrari A, Cappello A, et al. Inter-operator reliability and prediction bands of a novel protocol to measure the coordinated movements of shoulder-girdle and humerus in clinical settings. Med Biol Eng Comput 2009;47:475–86.
[23] Long JT, Eastwood DC, Graf AR, Smith AP, Harris GF. Repeatability and sources of variability in multi-center assessment of segmental foot kinematics in normal adults. Gait Posture 2009;31(1):32–6.
[24] Curtis DJ, Bencke J, Stebbins JA, Stansfield B. Intra-rater repeatability of the Oxford foot model in healthy children in different stages of the foot roll over process during gait. Gait Posture 2009;30:118–21.
[25] Sawacha Z, Cristoferi G, Guarneri G, Corazzo S, Dona G, Denti P, et al. Characterizing multisegment foot kinematics during gait in diabetic foot patients. J Neuroeng Rehabil 2009;23:37.
[26] Woodburn J, Nelson KM, Siegel KL, Kepple TM, Gerber LH. Multisegment foot motion during gait: proof of concept in rheumatoid arthritis. J Rheumatol 2004;31(10):1918–27.
[27] Stebbins J, Harrington M, Thompson N, Zavatsky A, Theologis T. Repeatability of a model for measuring multi-segment foot kinematics in children. Gait Posture 2006;23:401–10.
[28] Heller E, Robinson D. Traumatic pathologies of the calcaneal peroneal tubercle. The Foot 2010;9:6–98.
[29] Hofmeister EP, Juliano P, Lippert F. The anatomical configuration and clinical implications of the peroneal tubercle. The Foot 1996;6:138–42.