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Inertial motion capture validation of 3D knee kinematics at various gait speed on the treadmill with a double-pose calibration

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ABSTRACT

Background: Inertial motion capture (IMC) is rapidly gaining in popularity to evaluate gait in clinical settings. Previous examinations of IMC knee kinematics were often limited to the sagittal plane and IMC calibration has not been thoroughly investigated.

Research question: The objective was to validate IMC 3D knee kinematics calibrated with a double-pose during gait with reference to optical motion capture (OMC). The hypotheses are that IMC can estimate adequately knee kinematics and that both systems will detect similarly the changes with gait speed.

Methods: Twenty-four healthy participants walked on the treadmill at gait speed of 0.6, 0.8, 1.0 and 1.2 m/s. Knee kinematics were obtained simultaneously with two magnetic and inertial measurement units and passive markers fixed on the KneeKG system. OMC was calibrated with a functional anatomical approach and the IMC with a double-pose.

Results: Root mean square differences of the two systems yielded 3-6° for knee flexion, adduction and external rotation. Knee kinematics were more similar during the stance phase than the swing phase. Gait speed showed a significant progressive effect on the three knee angles that was similarly detected by the two systems.

Significance: IMC 3D knee kinematics can be obtained independently with a simple calibration and only two magnetic and inertial measurement units at an acceptable level of error especially during stance.

1. Introduction

Gait analysis has a long history of clinical applications and remains a relevant approach to evaluate patients. While most of these analyses use optical motion capture (OMC), a clear trend is oriented toward the use of inertial motion capture (IMC) instead [1–5]. The latter technology has the advantages of an affordable cost, a broader field of acquisition and being easily portable to conduct gait analyses [1,2,4]. While the validation remains to be determined in different settings, many studies have shown that IMC can adequately measure joint kinematics in the laboratory [1,5–7]. Since the positions of anatomical landmarks are not directly available with IMC, the calibration approach differs from OMC. Several approaches exist including single-pose [3,4], double-pose [2], functional [8], kinematical constraints [9], technical placement [8,10] or combination of approaches [1,10,11]. The choice of a calibration method for the knee remains difficult as most approaches have not been extensively evaluated in terms of complete 3D kinematics. In addition, most of the latter calibration approaches were evaluated on small sample sizes [1,2,8–11], which hampers the generalization of the findings and prevents from assembling normative data on healthy subjects. Evaluations of IMC knee kinematics are often limited to the sagittal plane where only knee flexion angle with 3-4° of error generally reported [2,3,7,9,12]. Nevertheless, the knee frontal and transverse planes of motion have shown clinical interest to characterize patients [13,14]. A few studies reported differences between IMC and OMC on knee adduction and external rotation angles of 9-16° with a combination of calibration approaches [15], 4-8° with a functional approach [8] and 3-4° with a single-pose [6]. The double-pose approach has the advantages of simplicity and rapidity to calibrate the IMC system, which could ease the clinical deployment, but has only

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been evaluated for knee flexion [2]. Hence, the capacity of IMC to measure complete 3D knee kinematics accurately during gait with a double-pose calibration remains unclear.

Another problematic aspect in the gait evaluation of patients with knee pain such as osteoarthritis is the choice of a walking speed, which can be limited by the severity of the symptoms [16]. Since patients generally walk at a slower pace, many studies report significantly different walking speeds between healthy controls and patients [14]. Hence, the IMC system should properly detect kinematics changes with gait speed for an appropriate clinical use. The main findings observed with increasing gait speed from previous studies were increases in step length [17–19], knee range of motion (ROM) [18–21] and ground reaction force [17]. However, the effect of gait speed is often limited to spatio-temporal parameters [17], knee kinematics of only the sagittal plane [19,19,20,21] or two contrasted gait speeds of 0.56 m/s and 0.97 m/s [18].

The main objective of the study was to validate 3D knee kinematics at various gait speed from an IMC system calibrated with a double-pose with reference to an OMC system calibrated anatomically. The first hypothesis was that the IMC system can measure 3D knee kinematics during gait with an acceptable level of error under 5° in the three planes. A specific aim was to determine the similarity of IMC to detect 3D knee kinematics changes according to gait speed in comparison to OMC. The second hypothesis was that gait speed influence on knee kinematics measured with IMC will be detected similarly to OMC.

2. Methods

2.1. Subjects

Twenty-four participants (10 women, 14 men, 24.8 ± 2.3 years, height of 174.5 ± 7.1 cm and bodyweight of 69.6 ± 10.8 kg) completed a consent form prior to participation in the study approved by the Ethics Committee from TÉLUQ University (#2017-244), École de Technologie Supérieure (#H20171103) and Centre Hospitalier de l'Université de Montréal (#17.333). Inclusion criteria were the capacity to walk 20 min without pain or discomfort. Exclusion criteria were self-reported lower limb musculoskeletal disorders within the last year.

2.2. Materials and instrumentation

The KneeKG system (Emovi Inc., Montreal, Canada) was used to measure 3D knee kinematics on the right knee for all participants (Fig. 1). The KneeKG has a broad clinical use and has showed reliability to reduce knee soft tissues artefacts [22,23]. The KneeKG is composed of a tibial tracker attached on the anteromedial aspect of the tibia with straps, a femoral tracker including an exoskeletal brace and a sacral belt positioned between the posterior superior iliac spines. The femoral tracker is adjusted with abutments between the biceps femoris tendon and iliotibial band laterally and between the adductor tubercle and posterior aspect of the adductor magnus tendon medially. Clusters of three passive markers are designed with the KneeKG and the two MTw magnetic and inertial measurement units (MIMUs) from the IMC system (Xsens Technologies, Enschede, Netherlands) were rigidly fixed on the tibial and femoral trackers (Fig. 1). For an independent clinical use, the MIMUs could be mounted anywhere on the trackers as long as they are rigidly fixed.

Knee kinematics were recorded simultaneously with an OMC system (Polaris Spectra, Northern Digital Inc., Waterloo, Canada) at 60 Hz and an IMC system comprised of two MTw MIMUs at 100 Hz. The signal was recorded during the complete session for the IMC system, while the signal of the OMC system consisted of separate trials. To synchronize the two systems, a resampling was applied to adjust the time series to a corresponding frequency. In the second step of the synchronization, a cross-correlation between the norms of the angular velocities of the tibia from the two systems with a cut-off threshold of 0.2 rad/s [11,24] was used on every trial to identify where each OMC trial was in the complete IMC signal. Similarly, the norms of the angular velocities of the femur served as a verification approach of the OMC trial identification within the IMC signal.

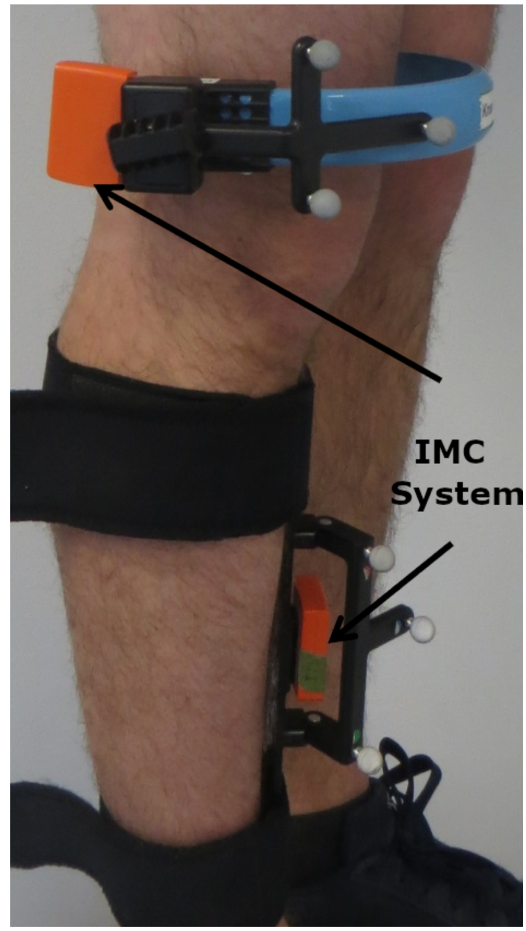

Fig. 1. Illustration of the KneeKG system installed on a subject with passive markers from the OMC system and the two MIMUs from the IMC system.

2.3. Experimental protocol

A previously validated calibration protocol of the OMC system was executed [22]. It consisted of the determination of the direction of progression on the treadmill, the identification of anatomical landmarks and execution of functional movements. The identified anatomical landmarks were the medial and lateral aspect of the femoral condyles and the medial and lateral malleoli. The functional movements were composed of hip circumduction to estimate the hip joint center location, knee flexion-extension to estimate the orientation of the medial-lateral axis and maximal knee extension reached from a small knee flexion to estimate a personalized neutral posture. The anatomical longitudinal axes were determined by linking the estimated joint centres, medio-lateral axes of the femur and tibia were aligned with the flexion-extension functional axis and the antero-posterior axes were determined orthogonally [22]. The clusters of markers of the OMC are used to measure the orientation of the femur and tibia with respect to the global coordinate system using the mathematical formalism of quaternions. These orientations are aligned to the anatomical reference systems based on the functional calibration with transformation matrices [22]. During walking, the unit quaternion of the femur relative to the tibia is expressed with respect to the personalised neutral posture by three independent knee clinical angles. The X-Y'-Z'' sequence of Cardan angles was used to obtain the knee flexion (+)-extension (-), adduction

(+)-abduction (-) and external (+)-internal (-) rotation.

The IMC system combines data from accelerometers, gyroscopes and magnetometers with a fusion algorithm to estimate orientation of the MIMU and needs time to gather information and stabilize as data history is used in the algorithm. Hence, the participant was asked to stay in static standing posture for one minute on the treadmill followed by one minute of hip flexion and extension at low velocity in the field of measurement. The calibration approach of the IMC was based on a double-pose including two static postures: 1) standing upright and 2) the lower limb being elevated in hip flexion with full knee extension maintained with the foot resting on a step [2]. The transition of the leg between the two postures is intended to stay in a sagittal plane. From the upright standing posture, the gravitational acceleration vector is used to define the longitudinal axis. Then from the elevated leg position, the gravitational acceleration vector is used to define a plane in combination with the first standing posture, and the medial-lateral axis is defined perpendicular to this plane. Finally, the anterior-posterior axis is determined orthogonally to the other two axes. The orientations of the femur and tibia MIMUs with respect to the global coordinate system were expressed with quaternions. These orientations are aligned to reference systems based on the double-pose calibration with transformation matrices. Knee joint angles were obtained with the same sequence of angles as the OMC system where the upright position was acting as a personalized neutral posture for the IMC system.

Afterwards, the treadmill was activated at 0.6 m/s to familiarise the subject to walking with the KneeKG on the treadmill for approximately three minutes. Several consecutive walking trials of 45 s were recorded at four treadmill speeds in the following order: 0.6, 0.8, 1.0 and 1.2 m/s. Three trials at each gait speed were recorded for a total of twelve walking trials. The participant was warned prior to the increase of the treadmill speed and asked if comfortable before recording the trial. At the end of the walking trials, the subject executed random movements of the instrumented leg in the three planes of motion separately and combined to obtain angular velocities along the three axes.

2.4. Data analysis

The normalisation of the gait cycle was achieved with minimum knee flexion angle after the swing phase, which was assumed to correspond to the foot strike [25]. Several discrete clinical parameters (Table 1) were extracted from the three rotation curves and included in the analysis as they are commonly used to characterise gait [5,14,18,19]. These were based on the gait events observed with kinematics and kinetics [25]. The random movements were used to align the local coordinate systems of the OMC and IMC systems with angular velocities as recommended [24]. This procedure evaluates the error of orientation estimation from the IMC system.

The root mean square difference (RMSD) between OMC and IMC was calculated on the complete gait cycle of the three knee angles at each gait speed and for the stance and swing phases. Parametric statistics were used after verification of the normality of the distribution with Lilliefors tests. A one-way repeated measures multivariate analysis of variance was conducted on the three knee angles RMSD with gait speed as a four level factor. A two-way repeated measures multivariate analysis of variance was conducted on the thirteen clinical parameters with factors system (OMC and IMC) and gait speed (0.6, 0.8, 1.0 and 1.2 m/s). To compare the gait speed factor for the OMC and IMC system, separate one-way repeated measures analysis of variance were conducted on clinical parameters on each system. In the case of multivariate significance according to Wilks' Lambda test, univariate tests were applied to each dependent variable. When a gait speed main effect was observed, Bonferroni post hoc tests were conducted to identify where the differences occurred between the levels of gait speed. When sphericity was not met according to Mauchly's test, the Huynh-Feldt correction was applied. Alpha level was set to 0.05 for all statistical analyses.

3. Results

The RMSD between the OMC and IMC is reported for knee angles at each gait speed for the complete gait cycle, the stance phase and the swing phase in Table 2. A multivariate significant main effect of gait speed was observed on RMSD ($F_{9,\ 15} = 4.7$, $p = .004$). Univariate tests indicated that gait speed was only significantly different for adduction ($F_{3,\ 69} = 15.2$, $p < .001$). Post hoc tests showed significant differences ($p < .05$) between all gait speed combinations, with the exception of 1.0 compared to 1.2 m/s. Knee kinematics normalised to gait cycle involving flexion, adduction and external rotation angles were contrasted by system (Fig. 2A) and by gait speed (Fig. 2B). The mean ± SD RMSD between the OMC and IMC after alignment of the local coordinate systems averaged for the four gait speeds was 1.1 ± 0.4° 1.2 ± 0.7° and 2.5 ± 1.3° for the flexion, adduction and external rotation respectively.

Mean clinical parameters measured with OMC and IMC showed a relative absolute difference under 5° for twelve of the thirteen parameters (Table 3). A multivariate significant main effect of system was observed on the clinical parameters ($F_{13,\ 11} = 8.10$, $p = .001$). Univariate main effects of system were observed on five of the thirteen clinical parameters (Table 3).

The gait speed analyses separated for IMC and OMC showed similar statistical conclusions for the two systems (Table 4). Nine clinical parameters were significantly different ($p > .05$) for the OMC system compared to eight for the IMC system. Post hoc tests of the pairwise differences between the four gait speeds also indicated similar results for the OMC and IMC system (Table 4).

Table 1
Definitions of the calculation of the clinical parameters.

Clinical parameter	Definition
Initial flexion	Flexion at foot strike
Flexion during loading	Difference between the maximal flexion from 0 % to 20 % of the gait cycle and the initial flexion
Flexion during stance	Difference between the maximal flexion from 0 % to 20 % of the gait cycle and the minimal flexion from 20 % to 68 % of the gait cycle
Maximal flexion	Maximal flexion of the complete gait cycle
Flexion ROM	Flexion range of motion over the complete gait cycle
Initial adduction	Adduction at foot strike
Varus thrust	Difference between the maximal adduction from 0 % to 20 % of the gait cycle and the initial adduction
Valgus thrust	Difference between the initial adduction and the minimum adduction from 0% to 20% of the gait cycle
Adduction during stance	Mean adduction from 20 % to 54 % of the gait cycle
Adduction ROM	Adduction range of motion over the complete gait cycle
Initial tibial rotation	Tibial rotation at foot strike
Tibial rotation during loading	Mean tibial rotation from 0 % to 20 % of the gait cycle
Tibial rotation ROM	Tibial rotation range of motion over the complete gait cycle

Table 2
Mean (SD) root mean square difference (RMSD) between the OMC and IMC for knee angles at each gait speed for the complete gait cycle, the stance phase and the swing phase.

RMSD (degrees) Gait phase	Knee angle	Gait speed (m/s) 0.6	0.8	1.0	1.2
Complete gait cycle (0–100 %)	Flexion-extension	3.3 (2.3)	3.2 (2.2)	3.4 (2.2)	3.4 (2.2)
	Adduction-abduction	5.1 (3.2)	5.3 (3.2)	5.6 (3.3)	5.7 (3.3)
	External-internal rotation	5.2 (5.2)	5.3 (5.0)	5.5 (5.3)	5.6 (5.1)
Stance phase (0–62 %)	Flexion-extension	3.2 (3.9)	3.1 (3.9)	3.1 (3.9)	3.2 (4.0)
	Adduction-abduction	3.2 (4.4)	3.3 (4.5)	3.6 (4.8)	3.8 (5.0)
	External-internal rotation	4.4 (6.8)	4.3 (6.6)	4.5 (7.0)	4.6 (6.8)
Swing phase (63–100 %)	Flexion-extension	3.4 (4.0)	3.5 (4.0)	3.6 (4.2)	3.6 (4.2)
	Adduction-abduction	6.3 (7.3)	6.6 (7.6)	6.9 (8.0)	7.0 (8.0)
	External-internal rotation	5.6 (7.8)	6.0 (7.9)	6.2 (8.2)	6.3 (8.2)

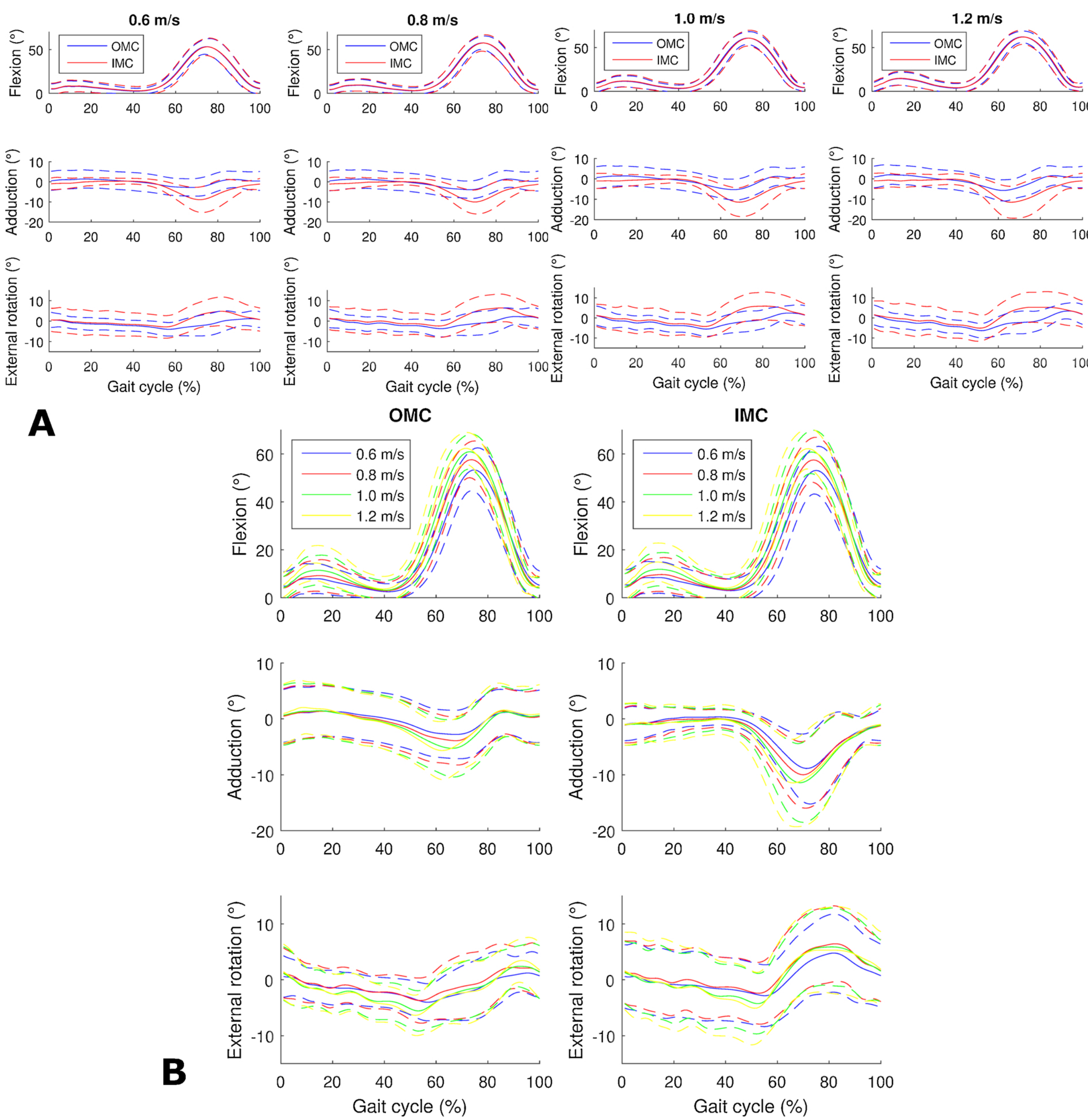

Fig. 2. Mean (SD) knee kinematics normalised to gait cycle involving flexion, adduction and external rotation angles contrasted by system (A) for optical motion capture (OMC, blue) and inertial motion capture (IMC, red) and contrasted by gait speed (B) of 0.6 m/s (blue), 0.8 m/s (red), 1.0 m/s (green) and 1.2 m/s (yellow) (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

Table 3
Mean clinical parameters (degrees) pooled for gait speed measured with the OMC and IMC system and their absolute difference, and associated univariate main effects on the factor system (OMC and IMC) from the repeated measures multivariate analysis of variance where significant P values were identified in bold.

Clinical parameter	OMC	IMC	Difference	$F_{1, 23}$	P value
Initial flexion	4.6	4.6	3.0	0.01	.947
Flexion during loading	7.2	7.6	0.5	**10.68**	**.003**
Flexion during stance	9.5	9.4	0.3	0.06	.816
Maximal flexion	60.7	60.4	3.2	0.07	.790
Flexion ROM	59.8	59.2	0.9	**7.52**	**.012**
Initial adduction	0.4	−1.3	3.0	3.83	.063
Varus thrust	1.7	1.6	1.2	0.10	.756
Valgus thrust	0.4	0.8	0.5	**8.91**	**.007**
Varus during stance	−0.2	−0.4	3.2	0.50	.825
Adduction ROM	9.2	14.3	6.4	**17.14**	**< .001**
Initial tibial rotation	1.9	2.0	4.7	0.001	.970
Tibial rotation during loading	−0.6	0.1	4.7	0.20	.657
Tibial rotation ROM	11.7	13.4	2.6	**7.02**	**.014**

Table 4
Statistics from the one-way repeated measures analysis of variance on the factor gait speed separated for the OMC and IMC system where significant P values were identified in bold.

	OMC		IMC	
Clinical parameter	$F_{3, 69}$	P value	$F_{3, 69}$	P value
Initial flexion	1.40	.257	2.45	.118
Flexion during loading	49.95	$< .001^{a, b, c, d, e, f}$	56.52	$< .001^{a, b, c, d, e, f}$
Flexion during stance	22.84	$< .001^{b, c, d, e, f}$	24.10	$< .001^{b, c, d, e, f}$
Maximal flexion	63.43	$< .001^{a, b, c, d, e}$	69.27	$< .001^{a, b, c, d, e}$
Flexion ROM	40.96	$< .001^{a, b, c, d, e}$	44.95	$< .001^{a, b, c, d, e}$
Initial adduction	1.50	.237	0.21	.774
Varus thrust	6.66	$.007^{c, f}$	0.62	.505
Valgus thrust	5.39	.008	8.87	$.001^{c, e}$
Adduction during stance	8.32	$.001^{a, b, c}$	5.33	$.012^{a, b}$
Adduction ROM	40.29	$< .001^{a, b, c, d, e, f}$	12.13	$< .001^{b, c, d, e}$
Initial tibial rotation	0.52	.597	1.93	.154
Tibial rotation during loading	1.25	.293	0.88	.419
Tibial rotation ROM	73.48	$< .001^{a, b, c, d, e, f}$	41.14	$< .001^{a, b, c, d, e, f}$

Post hoc tests significant differences were identified with: [a] 0.6 vs 0.8 m/s, [b] 0.6 vs 1.0 m/s, [c] 0.6 vs 1.2 m/s, [d] 0.8 vs 1.0 m/s, [e] 0.8 vs 1.2 m/s and [f] 1.0 vs 1.2 m/s.

4. Discussion

RMSD values between the OMC and IMC for the three knee angles remained under an acceptable level of 5° during the stance phase. The RMSD slightly overpassed 5° for the adduction and external rotation over the complete gait cycle. RMSD of approximately 3° obtained on the knee flexion was very similar to previous studies [1–3,6,7,9]. Adduction and external rotation RMSD were much lower than those reported during walking of 9.2–16.1° with a combination of calibration approaches [15], slightly higher than those reported during carrying tasks of 3.3–4.1° with a single-pose calibration [6] and similar to those reported during gait of 4.0–6.2° with a functional calibration [8]. Knee adduction and external rotation during the swing phase are overestimated by the IMC in comparison to the OMC. However, mean kinematics patterns are much more similar during the stance phase for both adduction and external rotation. This latter result is important, because the knee load is obtained during stance and is more relevant during the evaluation of patients [13], especially in the context of knee pain or prosthesis.

The total differences between OMC and IMC are a combination of technological error and differences due to the kinematical model [1,6]. The technological error obtained after aligning local coordinate systems of OMC and IMC showed a modest contribution to the total differences. The technological error arises from the orientation estimation of each MIMU obtained with a fusion algorithm of the accelerometers, gyroscopes and magnetometers where duration, type of motion and magnetic disturbances can affect accuracy. In comparison to literature, the technological error of 1.1–2.5° on the three knee angles during walking were inferior to those of 1.2–3.4° reported during handling tasks [6,7]. The model component accounts for most of the total difference between the two systems and the main factor affecting the kinematical model is the calibration where the anatomical calibration of the OMC serves as a reference. Since the IMC system uses a double-pose calibration, the joint coordinate systems are not perfectly aligned to those obtained anatomically, which explains most of the RMSD.

Gait speed showed a significant increase in RMSD, which indicates that motion speed can affect the accuracy of the IMC. Speed had also showed increases in the reported error on a pendulum [26,27], on an experimental gimbal setup [28] or on knee flexion during treadmill walking [12]. However, the relative increases in RMSD suggest that this effect remains fairly small for the tested gait speeds. Nevertheless, the effect of speed on the accuracy of the IMC system could potentially explain the higher RMSD observed during the swing phase when the leg is accelerated to the front.

Clinical parameters of initial values at heel strike, kinematics during stance and maximal knee angles were not significantly different as the knee kinematics were similar during the stance phase. The highest differences were observed on adduction and external rotation ROM, because the IMC overestimated these knee angles during the swing phase of the gait cycle. It must be noted that many significant differences observed on the clinical parameters remain of absolute amplitude under 1°. In addition, the 95 % confidence intervals from the two systems were always overlapping. The ROM differences between OMC and IMC are similar to a previous study during level walking of 0.03–1.45° of grand mean error [5], although the absolute difference was not reported.

The statistical analyses on gait speed indicated that both systems detect similarly the differences amongst the four tested speeds. The observation of the mean gait cycle curves shows that from 0.6 to 1.2 m/s, the changes in knee kinematics are more progressive than sudden. Both knee flexion at foot flat and midswing increased slightly with treadmill speed. The progressive changes with gait speed had also been observed on knee flexion during treadmill walking [21]. The maximal abduction observed after the preswing also increased progressively with gait speed. The maximal external rotation during terminal swing slightly increased from 0.6 to 0.8 m/s, but stayed more similar for higher gait speeds. The maximal internal rotation during double support was more similar between 0.6 and 0.8 m/s, but small increases were observed at 1.0 and 1.2 m/s. Similarly, a recent study comparing a slow speed to a self-selected speed had shown reduced ROM on the three knee angles [18]. The observations of gait speed influence on knee kinematics were independent of the OMC or IMC system. It indicates that the IMC system has the capacity of contrasting subtle changes in knee kinematics similarly to an OMC system.

A few limitations are present in the study. First, results from the IMC are dependent on the calibration approach used. The difference between OMC and IMC is mostly attributed to kinematical model where the calibration approach is the major aspect [1,6,7]. Further studies should elucidate the most appropriate calibration methods to estimate joint kinematics similarly to those obtained with anatomical calibration. A few studies evaluated various calibration approaches, but they were dedicated to the upper limb [10,11]. Second, the predetermined speeds on the treadmill may not be fully representative of natural progression of gait. The evaluation of the IMC done in this study will help to obtain more gait analyses in natural settings as this technology is not limited by the field of acquisition and easily portable. Third, walking on the treadmill can induce a certain level of error of the IMC system due to magnetic disturbances [29,30]. Since the fusion algorithm of the MIMUs to estimate orientation relies on magnetometers, the proximity to ferromagnetic objects may affect accuracy of the IMC system [30]. However, the laboratory environment is representative of the intended clinical use and gait evaluation in a natural setting could potentially increase the accuracy of the IMC system.

5. Conclusion

The study showed the capacity of evaluating complete 3D knee kinematics with only two magnetic and inertial measurement units and a simple double-pose calibration approach. Knee joint angles from the inertial motion capture system were closely related to those from an optoelectronic system especially during the stance phase to an acceptable level of error under 5° as hypothesized. The progressive changes in knee kinematics with increasing gait speed on the treadmill were similarly detected for both systems, which confirms the second hypothesis. The provided IMC knee kinematics on healthy subjects can serve as normative data for clinical comparison.

Declaration of Competing Interest

The authors have no conflicts of interest related to this manuscript.

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