

# Gait analysis in children with cerebral palsy via inertial and magnetic sensors

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**Abstract** 3D kinematic measurements in children with cerebral palsy (CP) to assess gait deviations can only be performed in gait laboratories using optoelectronic systems. Alternatively, an inertial and magnetic measurement system (IMMS) can be applied for ambulatory motion-tracking. A protocol named Outwalk has recently been developed to measure the 3D kinematics during gait with IMMS. This study preliminary validated the application of IMMS, based on the Outwalk protocol, in gait analysis of six children with CP and one typically developing child. Reference joint kinematics were simultaneously obtained from a laboratory-based system and protocol. On average, the root mean square error (RMSE) of Outwalk/IMMS, compared to the reference, was less than 17° in the transversal plane, and less than 10° in the sagittal and frontal planes. The greatest differences were found in offsets in the knee and ankle rotation, and in the hip flexion. These offset differences were mainly caused by a different anatomical calibration in the protocols. When removing the offsets, RMSE was always less than 4°. Therefore, IMMS is suitable for gait analysis of major joint angles in a laboratory-

free setting. Further studies should focus on improvement of anatomical calibrations of IMMS that can be performed in children with CP.

**Keywords** Cerebral palsy · Gait · Inertial sensors · Rehabilitation · Joint kinematics

## 1 Introduction

Cerebral palsy (CP) is the most common cause of motor disability in childhood [3, 15, 27]. In clinical practice, gait analysis is performed to assess gait deviations in patients with CP. Walking patterns and gestures that are usually expressed by children affected with CP, that is, at different levels of severity, are equinus, crouch knee, stiff knee, pelvis anti-retroversion, hip abduction and rotations, ankle dorsiflexion and overall ROM limitation due to the presence of spasticity [15, 33].

Conventionally, the measurement of joint kinematics is performed by means of optoelectronic marker systems in gait laboratories [6]. Different protocols have been developed for anatomical calibration of the optoelectronic systems, e.g. the Calibration Anatomical System Technique [5] (CAST), which is based on technical clusters of markers attached to body segments, anatomically calibrated by palpation of bony landmarks with a pointer with markers attached. Although optoelectronic-based movement analysis is accurate [19], its use in clinical practice is limited, due to its complexity, costs, and lack of availability of well-equipped gait laboratories. Optoelectronic systems are barely portable, and only a small number of steps can be recorded, due to a measurement volume that is restricted by camera positioning and line of sight problems, resulting in missing data. Moreover, patients adapt their

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gait pattern, due to the limited space in the laboratory and the feeling that they are being observed, resulting in an over-performance with respect to their everyday life ability [2, 7, 31, 32].

Recently, inertial and magnetic measurement systems (IMMSs) have been applied for ambulatory measure of 3D body segment orientations [7, 20, 21, 36, 37]. The IMMS has the potential to overcome the limitations of laboratory-based optoelectronic systems. It consists of small, light-weight sensor units comprising miniaturized 3D accelerometers, gyroscopes, and magnetometers [21, 25]. Through sensor fusion algorithms, the 3D orientation of each sensor unit is measured with respect to a global, earth-based coordinate system (CS) [24, 29]. With the IMMS kinematics can be measured in a laboratory-free setting. Hence, gait analysis can be improved by measurement of a large number of consecutive gait cycles during spontaneous walking, out of specific contexts, and in a physiological unconditioned way.

When a sensor unit of the IMMS is attached to a body segment, anatomical segment orientation must be obtained from the definition of an anatomical CS with respect to its technical sensor CS. Anatomical calibration of the IMMS can be based on functional movements, static reference postures, and/or careful alignment of sensor units with anatomical structures. A protocol has been proposed for the application of the IMMS in gait analysis [7]. This protocol, called Outwalk, was developed for use in clinical practice. Outwalk has been validated with respect to the CAST protocol in healthy subjects [13] and in transtibial amputees [8], however it has not been assessed in typically developing children or in children with CP.

The aim of this study was a preliminary validation of the application of IMMS, based on the Outwalk protocol, for ambulatory gait analysis in children with CP. Reference joint kinematics were simultaneously obtained from the conventional laboratory-based optoelectronic system and the CAST protocol. We hypothesized that the differences in joint kinematics would be similar to the differences that have been reported in a previous study in which the IMMS was applied in healthy adults [13].

## 2 Methods

### 2.1 Subjects

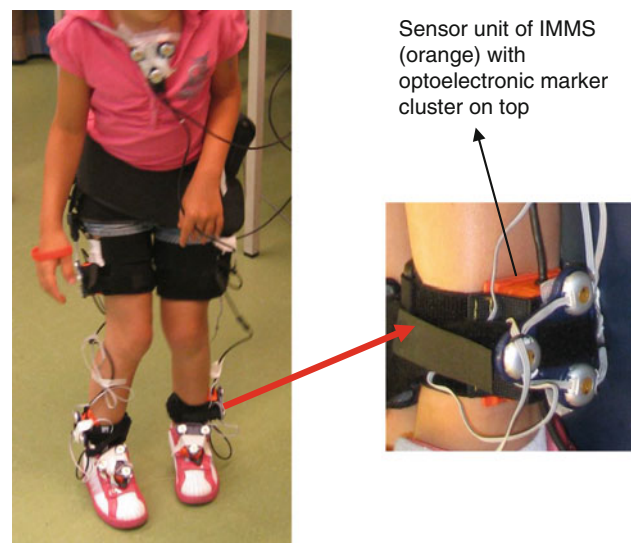
Seven children participated in the study: six children with spastic CP (all girls, four with GMFCS I (Gross Motor Function Classification System), and 2 with GMFCS II [28], age  $11.6 \pm 1.7$  years [mean  $\pm$  standard deviation (SD)], body weight  $42.2 \pm 11.2$  kg, body height  $1.50 \pm 0.15$  m], and 1 typically developing child (boy, age 12 years, body

weight 39 kg, body height 1.64 m). The children with CP were recruited from the Department of Rehabilitation Medicine of the VU University Medical Center in Amsterdam, The Netherlands. The study was approved by the Medical Ethics Committee of the VU University Medical Center. Informed consent was obtained from all parents of all the children, and additionally from children over 12 years of age.

### 2.2 Procedure

Gait measurements of the children were performed in a gait laboratory. The children walked on a 10 m walkway at self-selected walking speed. Kinematic data were simultaneously measured with the IMMS and an optoelectronic marker system. Sensor units (SUs) of the IMMS (MTx, Xsens Technologies B.V., The Netherlands) were attached to the feet, shanks, thighs, and pelvis with customized elastic straps (Xsens Technologies B.V.). The SU on the thorax was attached using skin-friendly double-sided tape. The SUs were positioned according to the Outwalk protocol [7]. In addition, optoelectronic marker clusters (OptoTrak 3020, Northern Digital Instruments, Waterloo, Canada) were rigidly attached over the SUs with double-sided tape (Fig. 1).

For the computation of joint kinematics, anatomical calibration of the optoelectronic system was according to the CAST protocol [5]. This included the calibration of bony landmarks relative to the marker clusters. A reference measurement was also performed in a static upright posture. The anatomical calibration of the IMMS was



**Fig. 1** A child with cerebral palsy wearing the IMMS sensor units (MTx, Xsens Technologies, The Netherlands) and marker clusters of the optoelectronic system (Optotrak 3020, Northern Digital Instruments, Waterloo, Canada)

according to the Outwalk protocol [7], which includes a static posture in either upright or supine position, and a passive knee flexion/extension movement, imposed by the examiner, to define the functional knee axis. In this study, the static posture and calibration movements were performed in an upright position. To assess the reliability of the functional calibration movement, we used the root mean square of the instantaneous helical axes coming from the computation of the mean helical axis from the flexion/extension movement. From practical experience, it is a sensitive parameter able to address the actual accuracy of the flexion/extension axis estimation, and should be as low as possible, i.e. allow minimal rotation in the frontal and transversal planes.

Data-acquisition with the IMMS and the optoelectronic marker system was synchronized using the synchronization pulse from the OptoTrak system (analog signal) recorded in the IMMS. Data were collected at a sample frequency of 100 Hz. Data on at least five successful gait trials were collected for each child.

### 2.3 Data analysis

The laboratory-based data were processed in BodyMech [<http://www.BodyMech.nl>, custom-made software based on MATLAB (R2009b, the Mathworks)] to obtain joint kinematics. The outcomes are further referred to as CO (i.e. CAST/Optoelectronic).

IMMS data were processed in MATLAB software, based on the MT Software Development Kit (Xsens Technologies B.V.) [7]. The orientations of the SUs on the foot, shank and thigh were calculated with the Kinematic Coupling algorithm (KiC [26]), to avoid any influence of a non-homogenous earth magnetic field in the gait laboratory, that could be caused by ferromagnetic materials in the surroundings [9]. The anatomical CSs were defined according to the Outwalk protocol. The outcomes are further referred to as OI (i.e. Outwalk/IMMS).

For three children, the optoelectronic data were also processed according to the Outwalk protocol. Anatomical CSs were defined according to the Outwalk protocol, e.g. the knee functional flexion/extension axis and use of static posture, instead of bony landmark data. In this way, the effects of a difference in protocol, excluding any effects of measurement system, could be evaluated. The outcomes are further referred to as OO (i.e. Outwalk/Optoelectronic).

The joint kinematics (3D ankle, knee, and hip angles) of the gait cycles of all successful trials per child were averaged. The differences in kinematics (averaged over the gait cycles) of OI versus CO were assessed. In addition, the difference between OO and CO was assessed. In this way, differences generated by the protocols (Outwalk and

CAST) could be isolated from the differences generated by the systems (IMMS and Optoelectronic) [13].

Parameters used in the evaluation of the joint kinematics were: (i) the difference in range of motion (ROM: maximal minus minimal joint angle), (ii) the offset (i.e. a mean difference over the entire gait cycle), (iii) the maximal difference, (iv) the root mean square error (RMSE), and (v) the RMSE with the offset removed (RMSE-offset; offset was removed by subtracting the mean difference between the waveforms), expressed in degrees (mean and SD over all children). In addition, the difference in gait parameters concerning the ankle, knee and hip kinematics that are clinically important in CP (as described by Schutte et al. [33]) were calculated. They included peak ankle dorsiflexion in stance and swing, knee flexion at initial contact, timing of peak knee flexion in the gait cycle, minimal hip flexion, peak hip abduction in swing, and mean hip rotation in stance.

The non-parametric Wilcoxon signed-rank test was used to test the significance ( $P < 0.05$ ) of the difference in ROM, the offset and the additional gait parameters. The differences were also expressed as a percentage of the SD of the CO trials (SD-CO) and as a percentage of the ROM of the CO trials. A difference of more than 100 % SD-CO means that the difference between OI and CO is greater than the intra-subject variability of CO. A difference of more than 100 % ROM-CO means that the difference is higher than the ROM of the CO of that particular joint angle.

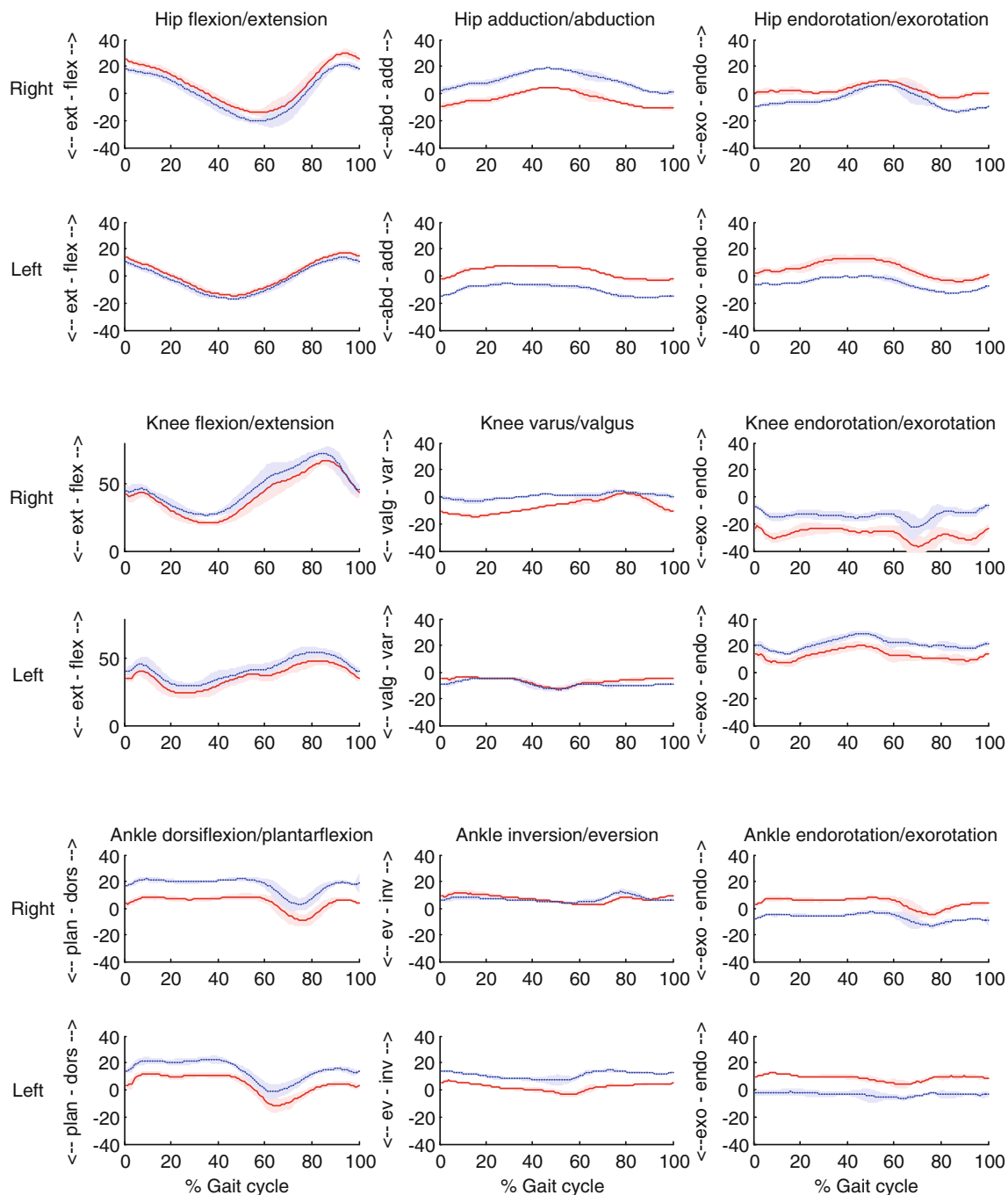
Furthermore, the adjusted coefficient of multiple correlations (CMC) was calculated for the joint kinematics (i.e. a variation on the Kadaba within-day CMC, and the new CMC proposed by Ferrari et al. [12, 13] that also takes into account the offset [16]).

## 3 Results

Figure 2 presents an example of the OI (solid red lines) and CO (dashed blue lines) kinematics during gait of a child with CP. The mean and standard deviation of the ankle, knee and hip kinematics of ten gait cycles of both legs are shown. In this example, the child with CP had a limited knee extension throughout the gait cycle in both the left and right knee due to the presence of spastic muscles at the lower limbs.

The differences in OI versus CO kinematics of all children are shown in the box and whisker plots in Fig. 3 and in Tables 1, 2, 3.

On average, the differences in ROM were less than  $3^\circ$  (SDs  $< 5^\circ$ ) and only significant for the ankle angle in the frontal plane (in/eversion). For the ankle and knee angle in the frontal plane, the difference in ROM was greater than



**Fig. 2** Typical example of joint angles of the right and left leg of a child with CP during gait, measured with Outwalk/IMMS (solid red lines) and CAST/Optoelectronic (dashed blue lines)

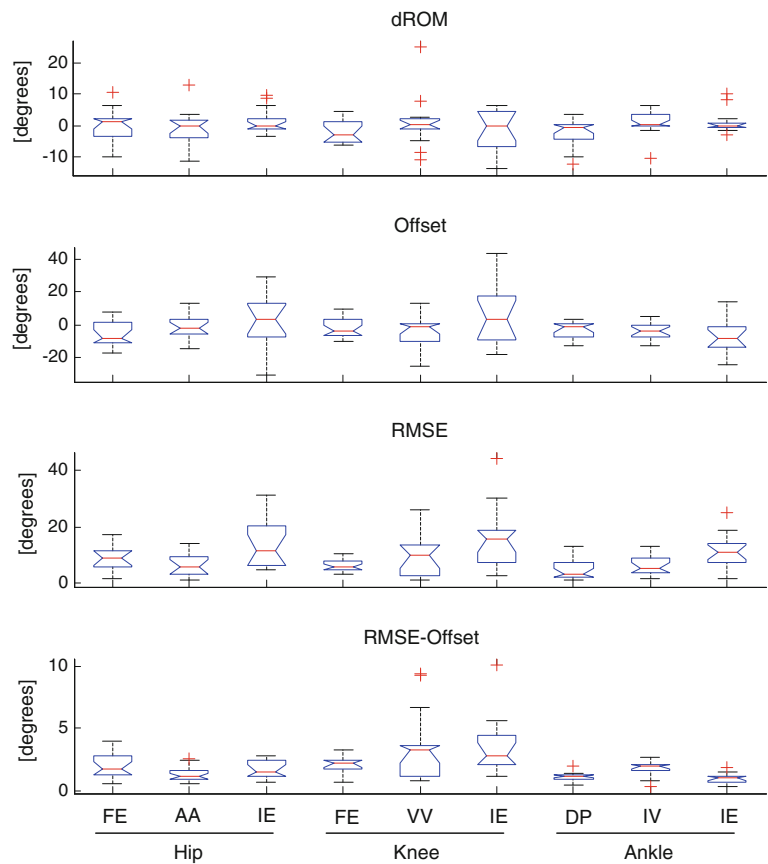
the intra-subject variability of the CO (i.e. the SD-CO was more than 100 %).

The transversal plane angles were mainly affected by offsets, i.e. a constant difference over the entire gait cycle. This is shown by the high offset values, the poor to moderate CMC values ( $<0.75$  or a complex number and therefore excluded in CMC Ferrari [12]), and the large effect of removing the offset from the RMSE. On average, the RMSE values were less than  $17^\circ$  in the transversal

plane (with SDs less than  $10^\circ$ ; and  $>500$  % SD-CO and  $>100$  % ROM-CO for the knee and ankle), and less than  $10^\circ$  in the sagittal and frontal planes. When the offset was removed, the RMSE values were less than  $4^\circ$ .

The OI measured on average less external knee rotation and more external ankle rotation. However, the inter-subject variability was high, as observed from the SDs, and the differences were not significant. In five of the seven children, less hip flexion was measured with the OI compared

**Fig. 3** Box and whisker plots for Outwalk/IMMS versus CAST/Optoelectronic; the dROM, Offset, RMSE and RMSE-Offset of the hip, knee and ankle joint angles of seven children are shown. *FE* flexion/extension, *AA* ab/adduction, *IE* in/external rotation, *VV* varus/valgus, *DP* dorsal/plantar flexion, *IV* in/eversion



to the CO. The offsets appeared to be greater than the SD of the CO (i.e.  $SD-CO > 100\%$ ), except for the hip angle in the transversal plane and the knee angle in the sagittal plane. The differences in peak ankle dorsiflexion in the stance and swing phase were significant ( $P = 0.04$ , on average  $3-4^\circ$  and  $>100\%$  SD-CO), mainly caused by the  $3^\circ$  of offset difference.

Figure 4 presents the box and whisker plots of OO kinematics compared to CO kinematics for three children. Similar to the comparison between OI and CO, the most prominent differences were found in the frontal and transversal knee angles, showing that these differences were mainly due to the protocol.

#### 4 Discussion

The aim of this study was a preliminary validation of the application of IMMS, based on the Outwalk protocol, in the gait analysis of children with CP. Six children with spastic CP and one typically developing child were included in the study to reach a variety in gait pattern and disease severity (from healthy to GMFCS II). The application of an ambulatory-based method for gait analysis in children with CP will be successful when it is able to identify abnormalities in the joint

kinematics which may be addressed with a variety of treatments.

When using the ambulatory-based protocol and system (Outwalk/IMMS) with respect to the laboratory-based protocol and system (CAST/Optoelectronic), mainly the transversal plane angles were affected by an offset (as defined via a mean difference over the entire gait cycle, and furthermore illustrated by the RMSE, the RMSE-offset and the CMC values). Furthermore, the previous study with Outwalk/IMMS in the healthy subject by Ferrari et al. [13] showed largest differences in the transversal plane angles. The differences in kinematics between the ambulatory and laboratory methods were higher in the children than in the healthy adult. The offset differences in the transversal plane angles (endo- and exorotation) were also found in the protocol comparison (Outwalk/Optoelectronic versus CAST/Optoelectronic), showing that these differences were due to the protocol (Outwalk versus CAST), and not due to the hardware system (IMMS versus Optoelectronic). In the sagittal hip angle, there was also an offset in both the overall (OI/CO) and the protocol (OO/CO) comparison, which indicates that this offset is the consequence of the different biomechanical hip models implemented in the protocols. However, differences seen in our study are less than those between different protocols as reported in a multi-protocol comparison using the same hardware [11].

**Table 1** The differences in ankle kinematics for Outwalk/IMMS versus CAST/Optoelectronic of seven subjects

Ankle OI versus CO	Mean	±SD	% SD CO	% ROM CO	P
<i>Dorsi/plantar flexion</i>					
Difference in ROM	-0.8	±1.5 (°)	29	3	0.18
Offset	-3.1	±4.3 (°)	116	13	0.09
Difference in peak dorsiflexion stance	-3.6	±4.2 (°)	135	15	0.04*
Difference in peak dorsiflexion swing	-3.1	±4.2 (°)	115	13	0.04*
RMSE	4.6	±3.4 (°)	172	19	
RMSE – Offset	1.1	±0.4 (°)	40	4	
CMC Kadaba within	0.90	±0.03			
CMC Ferrari (N = 7)	0.89	±0.14			
<i>Inversion/eversion</i>					
Difference in ROM	1.9	±1.3 (°)	108	24	0.03*
Offset	-4.3	±2.1 (°)	247	55	0.03*
RMSE	6.0	±1.1 (°)	345	77	
RMSE – Offset	1.8	±0.6 (°)	102	23	
CMC Kadaba within	0.76	±0.07			
CMC Ferrari (N = 6)	0.53	±0.35			
<i>Internal/external rotation</i>					
Difference in ROM	0.2	±1.7 (°)	13	3	0.87
Offset	-6.6	±8.7 (°)	382	88	0.18
RMSE	11.2	±2.5 (°)	642	148	
RMSE – Offset	1.0	±0.4 (°)	56	13	
CMC Kadaba within	0.63	±0.12			
CMC Ferrari (N = 1)	0.86				

A positive value in mean differences means OI > CO  
 OI Outwalk/IMMS, CO CAST/Optoelectronic, ROM range of motion, Offset mean difference, RMSE root mean square error, CMC coefficient of multiple correlation, N number of subjects included in CMC Ferrari calculation, % SD CO the mean difference as a percentage of the standard deviation (SD) of the CO trials, % ROM CO the mean difference as a percentage of the range of motion of the CO trials

\* P < 0.05

**Table 2** The differences in knee kinematics for Outwalk/IMMS versus CAST/Optoelectronic of seven subjects

Knee OI versus CO	Mean	±SD	% SD CO	% ROM CO	P
<i>Flexion/extension</i>					
Difference in ROM	-1.4	±2.2 (°)	30	2	0.20
Offset	-1.8	±4.3 (°)	37	3	0.24
Difference in flexion at initial contact	-1.2	±4.5 (°)	25	2	0.73
Difference in timing of peak flexion	0.1	±0.9 (s)			0.77
RMSE	6.2	±2.0 (°)	130	11	
RMSE – Offset	2.1	±0.6 (°)	44	4	
CMC Kadaba within	0.95	±0.03			
CMC Ferrari (N = 7)	0.96	±0.02			
<i>Varus/valgus</i>					
Difference in ROM	2.4	±3.4 (°)	152	23	0.18
Offset	-2.6	±4.8 (°)	168	25	0.18
RMSE	9.2	±6.0 (°)	595	88	
RMSE – Offset	3.6	±2.3 (°)	234	35	
CMC Kadaba within	0.79	±0.15			
CMC Ferrari (N = 5)	0.65	±0.23			
<i>Internal/external rotation</i>					
Difference in ROM	-0.1	±4.2 (°)	5	1	1.00
Offset	6.5	±13.5 (°)	241	47	0.24
RMSE	16.1	±9.8 (°)	593	116	
RMSE – Offset	3.5	±1.4 (°)	130	25	
CMC Kadaba within	0.64	±0.13			
CMC Ferrari (N = 4)	0.45	±0.30			

A positive value in mean differences means OI > CO  
 OI Outwalk/IMMS, CO CAST/Optoelectronic, ROM range of motion, Offset mean difference, RMSE root mean square error, CMC coefficient of multiple correlation, N number of subjects included in CMC Ferrari calculation, % SD CO the mean difference as a percentage of the standard deviation (SD) of the CO trials, % ROM CO the mean difference as a percentage of the range of motion of the CO trials

\* P < 0.05

**Table 3** The differences in hip kinematics for Outwalk/IMMS versus CAST/Optoelectronic of seven subjects

Hip OI versus CO	Mean	±SD	% SD CO	% ROM CO	P
<i>Flexion/extension</i>					
Difference in ROM	-0.4	±3.1(°)	14	1	0.87
Offset	-6.4	±7.2 (°)	211	16	0.06
Difference in minimal flexion	-5.5	±7.0 (°)	179	13	0.06
RMSE	8.8	±4.1 (°)	287	21	
RMSE – Offset	1.9	±0.9 (°)	64	5	
CMC Kadaba within	0.97	±0.01			
CMC Ferrari (N = 7)	0.88	±0.09			
<i>Ad/abduction</i>					
Difference in ROM	-0.9	±2.2 (°)	53	6	0.61
Offset	-1.8	±1.5 (°)	110	12	0.03*
Difference in peak abduction swing	-1.3	±1.9 (°)	79	8	0.09
RMSE	6.5	±3.5 (°)	399	42	
RMSE – Offset	1.3	±0.3 (°)	79	8	
CMC Kadaba within	0.93	±0.04			
CMC Ferrari (N = 6)	0.71	±0.13			
<i>Internal/external rotation</i>					
Difference in ROM	-0.1	±1.6 (°)	3	0.4	0.80
Offset	2.1	±6.0 (°)	97	14	0.40
Difference in mean rotation stance	2.0	±6.1 (°)	90	13	0.40
RMSE	13.8	±8.6 (°)	638	95	
RMSE – Offset	1.6	±0.7 (°)	76	11	
CMC Kadaba within	0.73	±0.18			
CMC Ferrari (N = 3)	0.66	±0.10			

A positive value in mean differences means OI > CO  
 OI Outwalk/IMMS, CO CAST/Optoelectronic, ROM range of motion, Offset mean difference, RMSE root mean square error, CMC coefficient of multiple correlation, N number of subjects included in CMC Ferrari calculation, % SD CO the mean difference as a percentage of the standard deviation (SD) of the CO trials, % ROM CO the mean difference as a percentage of the range of motion of the CO trials  
 \* P < 0.05

Furthermore, after removing the offset from the RMSE (RMSE-offset), the differences in Outwalk/IMMS versus CAST/Optoelectronic were less than 4°. Significant mean differences in other parameters, such as the ankle dorsiflexion, were also less than 4°. These differences are considered clinically not relevant with respect to the reliability of conventional 3D gait measurements [19].

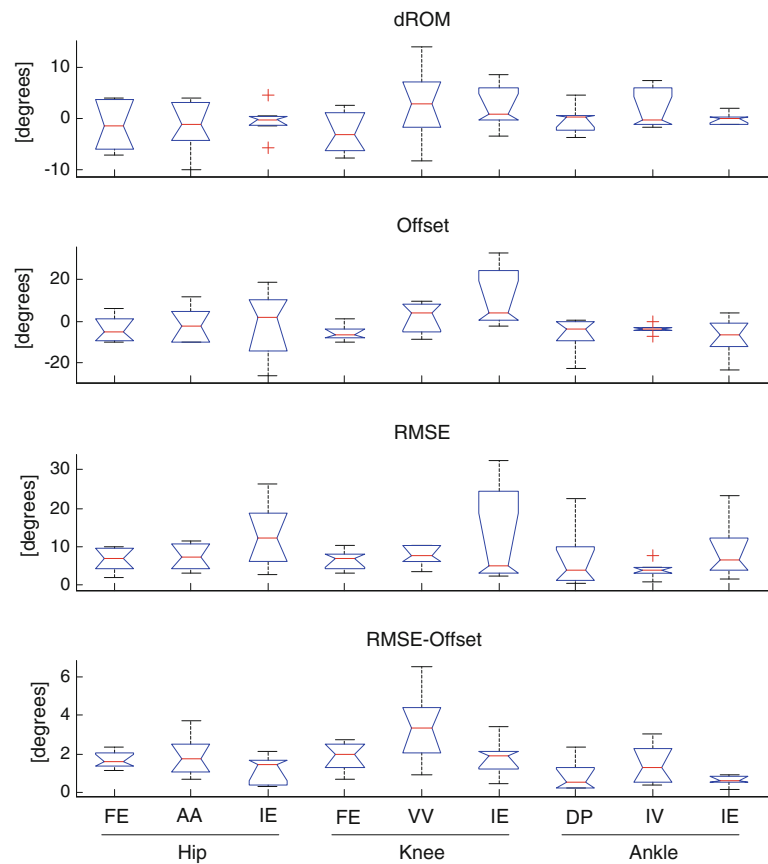
The calibration in the Outwalk protocol is based on the functional movement in the knee, static posture (upright or supine), and careful alignment of the pelvis and shank sensor with anatomical structures [7]. In contrast, the CAST protocol uses anatomical landmarks to define the CS of the segment. The differences in the frontal and transversal plane angles of the knee are probably caused by the differences in the CS of the thigh and the shank. It is well known that cross-talk is a primary concern for the knee joint [23, 30]. Brennan et al. [4] described the effect of anatomical frame variation on joint angles. They proposed equations as function of nominal joint angles to identify the coefficients that correspond to a variation in rotation of the anatomical CS and cause the cross-talk. The differences in joint angles that mainly appear as offsets in the frontal and transversal planes may be caused by multi-axis variations. Frontal and transversal plane angles are often regarded as unreliable [13, 23]. For example, Schache et al. [30]

described that when the knee flexion/extension axis is ill-defined, flexion could be transferred into ab/adduction or internal/external rotation.

For the Outwalk protocol, performance of a pure knee flexion/extension to define the distal thigh CS (the distal CS is the segment CS that is used for the distal joint of the segment [7]) may be more difficult for children with CP than for healthy subjects (particularly in a standing posture), since children with CP have less stability (with increasing GMFCS), bone deformities such as femoral anteversion and tibial torsion, and knee flexion contractures. However, the use of functional calibrations could also mitigate the error of cross-talk especially when bone misalignments are present. The functional axis might better represent the actual axis of rotation around which body segments are moving, in contrast to an axis defined by bony landmarks, especially in case of bone deformity. Since our study only included one typically developing child, we were not able to show any differences in kinematic errors between healthy subjects and CP due to performance of anatomical calibration.

Although the CAST protocol was used as reference, this protocol may also suffer from inaccuracies, due to the erroneous palpation of bony landmarks [34]. Studies reporting on the reliability of 3D gait measurements with

**Fig. 4** Box and whisker plots for Outwalk/Optoelectronic versus CAST/Optoelectronic; the dROM, Offset, RMSE and RMSE-Offset of the hip, knee and ankle joint angles of three children are shown. *FE* flexion/extension, *AA* ab/adduction, *IE* in/external rotation, *VV* varus/valgus, *DP* dorsal/plantar flexion, *IV* in/eversion



optoelectronic systems have found data errors of  $<5^\circ$ , with the exception of hip and knee rotations that show larger errors [19]. RMSE values of the OI versus CO were less than  $10^\circ$  in the sagittal and frontal planes, and  $<17^\circ$  in the transversal plane (that decrease to  $4^\circ$  when removing the offset). This means that both ambulatory-based and laboratory-based methods are less reliable for accurate identification of abnormalities in the joint kinematics in the transversal plane. For the CAST protocol, accurate palpation of bony landmarks, such as the femoral epicondyles, may be more difficult in children with CP when they have bone deformities. Furthermore, soft-tissue artefacts cause a relative movement of skin markers with respect to the underlying bone that may substantially affect knee joint kinematics [1]. It should also be noted that the functional knee axis, defined during a passive, non-weight-bearing, knee flexion/extension movement, does not have to be similar to the axis defined by the femoral epicondyles [22] or to the functional knee axis during gait [14, 17, 18].

The distal pelvis and proximal thigh CSs in the Outwalk protocol are defined from the upright static posture, assuming zero flexion, ab/adduction and rotation in the joints. In our study this posture was used, since upright calibration is easier and quicker to perform than a calibration in a supine position, and it has been used in the study of healthy adults [13], that we used for comparison.

Obviously, the advantage of upright calibration is a shortage of time needed for the calibration procedure.

However, the upright calibration may have affected the accuracy of the kinematics. Particularly, the offset that was observed in the sagittal hip kinematics of five children may have been caused by the use of the upright static posture. In an upright posture the pelvis is normally slightly anteriorly tilted, in the presence of hip flexion [35]. In pathological cases, this anterior tilt in upright stance might be even excessive [15]. Moreover, a neutral upright static posture is difficult to maintain for children with irreducible knee flexion, laxity or deformities.

Therefore, to correct for pelvis anterior tilt in an upright posture, the hip biomechanical model in the Outwalk protocol should be optimized. This may be achieved using the actual CS of the sensor on the pelvis, aligned with the posterior superior iliac spines (the proximal pelvis CS [7]), instead of using the assumption that the pelvis is not tilted (the distal pelvis CS, which is currently used for hip kinematics [7]).

Furthermore, the Outwalk protocol makes an alternative static trial possible in a supine position, with the hip and knee flexed at a certain known angle to account for flexion and joint deformities [7]. This could optimize the accuracy of the anatomical calibration. However, in unloaded supine posture positions of bones underlying the skin with respect



to the sensors attached to the skin might be different from a loaded upright posture. This may influence the accuracy of the anatomical calibration.

Moreover, also anatomical calibration in the frontal and transversal planes can be hard to achieve using a static posture in children with CP due to joint deformities or muscle contractures. Therefore, other anatomical calibrations that can be performed in children with CP should be investigated as well. Favre et al. [10] proposed two functional calibration movements for the knee joint when using IMMS, also including a rotation of the shank in the frontal plane (ab/adduction) in sitting posture to define the anterior–posterior axis of the knee, apart from the flexion/extension calibration. In this way, anatomical calibration might be optimized.

The KiC algorithm, used for the estimation of the ankle and knee kinematics, does not rely on the use of the IMMS magnetometers [26]. Therefore, these joint kinematics were not affected by a non-homogenous earth magnetic field which may be a main concern in gait laboratories [9]. However, for the hip kinematics, the KiC algorithm was not applied. Therefore, the observed difference in the hip kinematics might also be the result of a non-homogeneous earth magnetic field caused by the instrumentation (apart from the effect of the upright calibration, as discussed above). When also applying the KiC algorithm for the hip, the kinematics might be further improved.

In conclusion, the application of the IMMS is suitable for ambulatory gait analysis of the major joint angles in children with CP. However, further research is needed in a larger CP population to evaluate the accuracy and reliability of the IMMS and Outwalk protocol for specific gait rehabilitation parameters in CP. Moreover, a comparison of gait patterns in- versus outside the gait laboratory, as well as the role of 3D kinematic measurement via IMMS in clinical decision-making are topics of further study. Finally, to improve the accuracy of joint kinematic measurements, future studies should focus on the improvement of anatomical calibrations of the IMMS that can be performed in children with CP.

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