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Ambulatory measurement of the knee adduction moment in patients with osteoarthritis of the knee

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ABSTRACT

High knee joint-loading increases the risk and progression of knee osteoarthritis (OA). Mechanical loading on the knee is reflected in the external knee adduction moment (KAdM) that can be measured during gait with laboratory-based measurement systems. However, clinical application of these systems is limited. Ambulatory movement analysis systems, including instrumented force shoes (IFS) and an inertial and magnetic measurement system (IMMS), could potentially be used to determine the KAdM in a laboratory-free setting. Promising results have been reported concerning the use of the IFS in KAdM measurements; however its application in combination with IMMS has not been studied.

The objective of this study was to compare the KAdM measured with an ambulatory movement analysis system with a laboratory-based system in patients with knee OA. Gait analyses of 14 knee OA patients were performed in a gait laboratory. The KAdM was concurrently determined with two the systems: (i) Ambulatory: IFS and IMMS in combination with a linked-segment model (to obtain joint positions); (ii) Laboratory: force plate and optoelectronic marker system.

Mean differences in KAdM between the ambulatory and laboratory system were not significant (maximal difference 0.20 %BW$*$H in late stance, i.e. 5.6% of KAdM range, $P > 0.05$) and below clinical relevant and hypothesized differences, showing no systematic differences at group level. Absolute differences were on average 24% of KAdM range, i.e. 0.83 %BW$*$H, particularly in early and late stance. To achieve greater accuracy for clinical use, estimation of joint position via a more advanced calibrated linked-segment model should be investigated.

1. Introduction

High knee joint-loading due to e.g. malalignment, laxity, injury or obesity, increases the risk and progression of knee osteoarthritis (OA) (Englund, 2010; Hunter and Wilson, 2009). Knee OA is more common in women and elderly people. 1.5% of adults above 55 suffer from painful, severe knee OA. 10% has mild to moderate knee OA (Peat et al., 2001). Knee OA involves cartilage destruction, subchondral bone-thickening and new bone formation. It results in knee pain, instability, stiffness and swelling and could lead to knee arthroplasty. Patients frequently experience limitations in daily life activities and a decline in mobility (Andriacchi et al., 2004; van Dijk et al., 2006). To identify abnormal joint loading on the knee, the measure of the net external knee adduction moment (KAdM) during gait could be used. The KAdM reflects the internal loading on the medial compartment of the knee (tibio-femoral force) (Zhao et al., 2007). Increased KAdM peaks (20–40%) have been observed in patients with medial knee OA (Foroughi et al., 2009; Baliunas et al., 2002). Bracing, heel wedges, osteotomy, gait modifications, and weight management are used to minimize knee joint-loading in these patients (Brouwer et al., 2007; Simic et al., 2010; Glass, 2006; Hunter et al., 2012). To better direct and evaluate such

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treatments in knee OA, objective knee-load measurement via the KAdM may be important in clinical practice.

Currently, measurement of the KAdM is restricted to optoelectronic marker systems and force plates in gait laboratories. However, available and well-equipped laboratories in hospitals and rehabilitation centers are often lacking. Furthermore, optical markers have line of sight problems resulting in missing data, and targeted foot positioning on force plates causes an adaptation of the gait pattern (Schepers et al., 2007; Luinge and Veltink, 2005; Cutti et al., 2010; Best and Begg, 2006). Therefore, there is a need for feasible and validated measurements in clinical practice.

Recently, ambulatory movement analysis systems have been introduced, including instrumented force shoes (IFSs) for kinetic measurements, and inertial and magnetic measurement systems (IMMSs) for kinematic measurements. Application of these systems is not restricted to gait laboratories and could be used at any place and any time. IFSs have been applied and proven accurate in measuring ground reaction force (GRF) and center of pressure (CoP) in healthy subjects (Faber et al., 2010a; Schepers et al., 2007) and patients (Schepers et al., 2009; van den Noort et al., 2011; van den Noort et al., 2012). IMMSs with appropriate anatomical calibration procedures (i.e. sensor-to-segment calibration) were successfully evaluated to measure segmental orientations and joint angles (Luinge and Veltink, 2005; Cutti et al., 2010; van den Noort et al., 2009; Zhou et al., 2008).

The combination of IFSs and IMMSs could potentially be used to determine the KAdM of knee OA patients in laboratory-free setting. However, with IMMSs, it is difficult to obtain positions of segments or joints (Schepers et al., 2010), while for net joint-moment calculations joint positions are required, in addition to GRF and CoP measurements (Hof, 1992). Several methods have been suggested to obtain positions with IMMS, such as linked-segment models that represents skeletal geometry (Faber et al., 2010b), ambulatory position information using a magnetic source worn on the body (Schepers et al., 2010) or kinematic coupling algorithms (Roetenberg et al., 2010). Previously, (van den Noort et al., 2012) showed that segment orientations and fixed segment lengths could be used as input in a linked-segment model to obtain joint positions, that have been used in combination with IFS data to determine the KAdM. Estimation errors of the KAdM were found to be 0.78 %BW*H (22% of the KAdM range) in particularly late stance (BW is bodyweight, H is body height), while clinical relevant differences between medial knee OA patients and healthy controls are reported to vary about 1 %BW*H (20–40 % KAdM range) (Baliunas et al., 2002; Foroughi et al., 2009; Thorp et al., 2006).

As a proof of principle, van den Noort et al. used orientations from the optoelectronic reference system, evaluating only a *part* of the system. The objective of the present study was to compare the KAdM measured with the *entire* ambulatory movement analysis system (i.e. IFS *and* IMMS) with the KAdM measured with the laboratory system (optoelectronic marker system and force plate) as reference, in patients with knee OA. Based on results of the previous study and with the aim to show clinically relevant differences, we hypothesized a difference in the KAdM between the ambulatory and laboratory system of 0.90 %BW*H.

2. Methods

2.1. Patients

Fourteen patients, who all fulfilled the American College of Rheumatology (ACR) criteria for knee OA (Altman and Gold, 2007), participated in the study (3 males, 11 females, mean age 61.0 ± 9.2 years (mean ± standard deviation), body mass 83.7 ± 14.4 kg, and body height 1.66 ± 0.11 m), with dominant medial or lateral tibiofemoral radiographic OA (Kellgren/Lawrence grade > 1). The patients were recruited from the patient population of the Reade Centre for Rehabilitation and Rheumatology (Amsterdam, the Netherlands). The Medical Ethics Committee of the VU University Medical Center (Amsterdam, the Netherlands) approved the study. Full written informed consent was obtained from all participants.

2.2. Procedure

The patients walked in a gait laboratory on a 10 m walkway at comfortable self-selected speed. Kinematic and kinetic data were collected synchronously by means of an ambulatory movement analysis system and the standard laboratory system (as a reference). The ambulatory system consisted of IFSs and an IMMS. The IFS was based on an orthopedic sandal, with 6-degrees-of-freedom ATI mini45 SI-580-20 force/moment sensors (Schunk GmbH & Co. KG) (Schepers et al., 2007; van den Noort et al., 2011, 2012). The IMMS sensor units (MTx, Xsens Technologies, the Netherlands), were attached to each force/moment sensor of the IFS and to the shanks (Fig. 1). The IFS and IMMS were wirelessly connected to a computer, via two Xbus Master devices (Xsens Technologies, the Netherlands; sample frequency 50 Hz). The laboratory system consisted of a force plate (AMTI OR6-5-1000, Watertown, MA, USA) embedded in the floor of the laboratory (sample frequency 1000 Hz), and an optoelectronic marker system (OptoTrak 3020, Northern Digital Instruments, Waterloo, Canada) with marker clusters attached to the feet (IFS), shanks and thighs (sample frequency 50 Hz).

Prior to the gait measurements, an upright static measurement and a passive standardized flexion/extension movement of the patient's knee joint were performed by the examiner (non-weight bearing, sitting posture, maximal range of motion of 90°) for anatomical calibration of the IMMS coordinate system on the shank (Cutti et al., 2010; van den Noort et al., 2009). To determine anatomical coordinate systems with the optoelectronic marker system, anatomical landmarks were palpated according to Cappozzo et al. (1995) based on ISB standards (Wu et al., 2002).

Data on three successful trials were collected per leg, i.e. a step on the force plate during normal gait, and no missing marker data of the optoelectronic system. Prior to measurements, patients had time to practice the trials.

2.3. Data analysis

For the ambulatory system, the algorithms of (Schepers et al., 2007) were used to calculate the GRF and CoP, based on IMMS and IFS data. The orientations of the IMMS sensors on heel, forefoot and shank were calculated by integration of angular velocities from the gyroscopes (Bortz, 1971). At each stride, the orientations of the heel and forefoot sensors were corrected, using zero-velocity-update and assuming equal vertical position of the foot at each stride (Schepers et al., 2007). The inclination at each stride was estimated with the accelerometers. In this way integration time was limited to minimize integration drift. An orientation correction at each stride was not possible for the shank-sensor, since the shank is moving throughout the gait cycle. Inclination was corrected at the start of each trial using the accelerometers. The heading (direction) was corrected by the

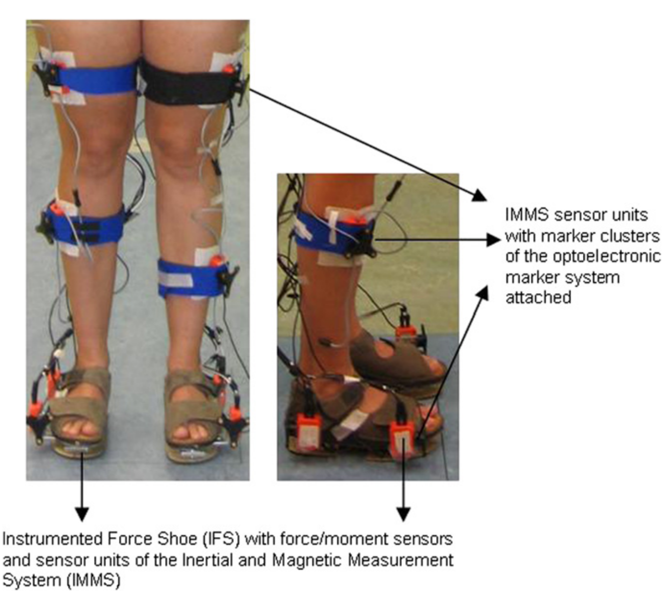

Fig. 1. Measurement set-up of instrumented force shoes (IFS), sensors of the inertial and magnetic measurement system (IMMS) and optoelectronic markers, positioned on the lower legs and feet.

known orientation of the feet at the start of the trials, assuming the shanks to be aligned with the feet and the patient in upright posture. Most trials included only three or four gait cycles, which limited the integration drift. A sensor-to-segment orientation matrix was first determined for the shank, using the gyroscope and accelerometer data from the passive knee flexion/extension calibration movement. The angular velocities and accelerations of the shank sensor were transformed to the anatomical segment coordinate system, using this matrix.

The ankle and knee positions with respect to the midpoint of the heel force/moment-sensor during gait were calculated with a linked-segment model (van den Noort et al., 2012), assuming the segments to be rigid bodies. Inputs of the model were heel and shank orientation during gait measured with the IMMS, and fixed segment lengths (heel–ankle and ankle–knee vectors), once calculated from optoelectronic data from the stance phase of a gait trial. The GRF, CoP, and joint positions from the ambulatory system were transformed to the global coordinate system of the laboratory for comparison, using a transformation matrix calculated via orientation difference between the GRF of the IFS (in gait direction) and the GRF of the force plate (Schepers et al., 2007; Faber et al., 2010a; van den Noort et al., 2012).

For the laboratory system, the GRF, CoP, and the segment and joint positions and orientations were calculated from optoelectronic marker and force plate data in BodyMech (www.bodymech.nl), custom-made software in MATLAB (R2009b, The Mathworks), based on the ISB anatomical frame definitions (Cappozzo et al., 1995). The ankle joint center and medio-lateral axis were defined as the midpoint and the line between the lateral and medial malleoli, the knee joint center and axis were defined as the midpoint and the line between the femoral epicondyles.

For each trial, two KAdMs (ambulatory and laboratory) were calculated using the GRF and its moment arm, defined by the CoP and the knee position that were synchronously measured with both systems (Hof, 1992). The joint moments were normalized to bodyweight (in N) and body height (in m), i.e. %BW*H.

2.4. Statistical analysis

In the statistical analysis (SPPS Software, Version 15.0), three trials of each leg with OA were included. Legs without radiographic knee OA were excluded from analyses (the unaffected leg from patients with unilateral knee OA). Prior to the analysis, normal distribution and sphericity of the data were statistically checked.

A linear-mixed model was used for calculation of statistical differences between the ambulatory and laboratory KAdM. This model was applied for the KAdM in early stance (ESP, at the moment of peak vertical GRF in the first 50% of the stance phase), in late stance (LSP, at the moment of peak vertical GRF in the last 50% of the stance phase), midstance (MS, at the moment of minimal vertical GRF between ESP and LSP), and impulse (the time-integral of the KAdM over the stance phase). OA legs were treated as random in the model. The statistical difference of system (ambulatory versus laboratory), accounting for the repeated measures (i.e. 3 trials per leg), was tested. A P-value of less than 0.05 was considered statistically significant.

Furthermore, as descriptive values, the mean and absolute difference (i.e. the absolute value of the difference between the systems) at ESP, MS, LSP and impulse, the root mean square error (RMSE) and the gain (non-dimensional) over the whole stance phase were calculated for the KAdM.

3. Results

Data of a total of 24 legs with knee OA (3 trials per leg) were included in the analyses. For one patient only one trial was included due to missing marker data.

The linear-mixed model showed no significant differences between the ambulatory KAdM and the laboratory KAdM at ESP, MS, LSP and impulse. Mean differences were less than 0.20 %BW*H (i.e. 5.6% of the KAdM range, with P-values of: ESP=0.604, MS=0.105, LSP=0.584, impulse=0.336, see Table 1). However, there was variability within the group in the differences between both measurement systems. This is supported by relatively high standard deviations of the mean differences (up to 0.80 %BW*H) and by absolute difference and RMSE values (up to 0.83 %BW*H, Table 1).

The mean KAdMs of both systems and their standard deviation for each of the 24 legs are illustrated in Fig. 2. Fig. 2 also includes an indication of the dominant affected compartment of the OA knee, either medial or lateral. Scatter plots of the ambulatory versus laboratory KAdM for all variables are presented in Fig. 3.

Table 1
Mean and absolute differences, RMSE and gain (averaged over all legs included) of the KAdM of the ambulatory movement analysis system versus the laboratory movement analysis system in patients with osteoarthritis of the knee.

KAdM		Descriptive values Difference		Linear mixed model results	
		mean ± SD	(% range)	SE	P
ESP [%BW*H]	Mean	−0.07 ± 0.80	(1.9%)	0.13	0.604
	Absolute	0.75 ± 0.49	(22%)		
MS [%BW*H]	Mean	−0.19 ± 0.42	(5.5%)	0.09	0.105
	Absolute	0.54 ± 0.27	(15%)		
LSP [%BW*H]	Mean	−0.20 ± 0.80	(5.6%)	0.15	0.584
	Absolute	0.83 ± 0.50	(24%)		
impulse [%BW*H*s]	Mean	0.06 ± 0.19	(1.7%)	0.06	0.336
	Absolute	0.19 ± 0.12	(5.5%)		
RMSE [%BW*H]		0.79 ± 0.32	(23%)		
Gain		0.94 ± 0.18			

KAdM = Knee Adduction Moment.
ESP = Early Stance Peak; MS = Midstance; LSP = Late Stance Peak.
RMSE = Root Mean Square error.
Gain > 1 = Ambulatory > Laboratory.
Mean difference > 0 = Ambulatory > Laboratory.
BW = body weight in Newton; H = body height in meters.
SD = standard deviation; SE = standard error of difference.

Highest absolute differences between the two measurement systems were found at LSP (0.83 %BW*H), which is 24% of the range of the laboratory KAdM.

Further analyses showed that in nine legs from eight individual patients there was a mean difference of more than 0.90 %BW*H (four legs at ESP, three legs at LSP, one leg at both ESP and LSP, and one leg at ESP, MS and LSP).

4. Discussion

We hypothesized a difference of 0.90 %BW*H between the KAdM measurements with the ambulatory and laboratory systems. Our results showed no significant mean differences between the two systems. Mean differences were below hypothesized and clinical relevant differences. This shows that, at group level, the ambulatory system does not systematically under- or over-estimate the KAdM with respect to the laboratory system as a reference. Furthermore, it might indicate that the ambulatory system is able to discriminate knee joint-loading of patients with medial knee OA from knee joint-loading of healthy subjects or patients with lateral knee OA (see also Fig. 2). However, this still needs to be tested.

Absolute differences were on average up to 0.83 %BW*H (24% of KAdM range), particularly at early and late stance. In eight participants we found differences higher than the hypothesized differences. The absolute differences were of similar magnitude as the significant mean differences at late stance shown previously (0.78 ± 0.43 %BW*H (van den Noort et al., 2012)). The use of IMMS did not result in higher KAdM errors, compared to segment orientation from an optoelectronic marker system, although direction of differences was not similar (i.e. no systematic difference with the IMMS). Differences between the ambulatory versus the laboratory KAdM may be caused by IFS inaccuracy, the linked-segment model used, or inaccuracy in IMMS orientation estimation. The accuracy of the IFS for GRF and CoP measurement was proven to be high (van den Noort et al., 2012; Schepers et al., 2007; Faber et al., 2010a). Therefore, the differences in KAdM are mainly due to joint position estimation via the linked-segment model with IMMS based orientations. Explanations of the found

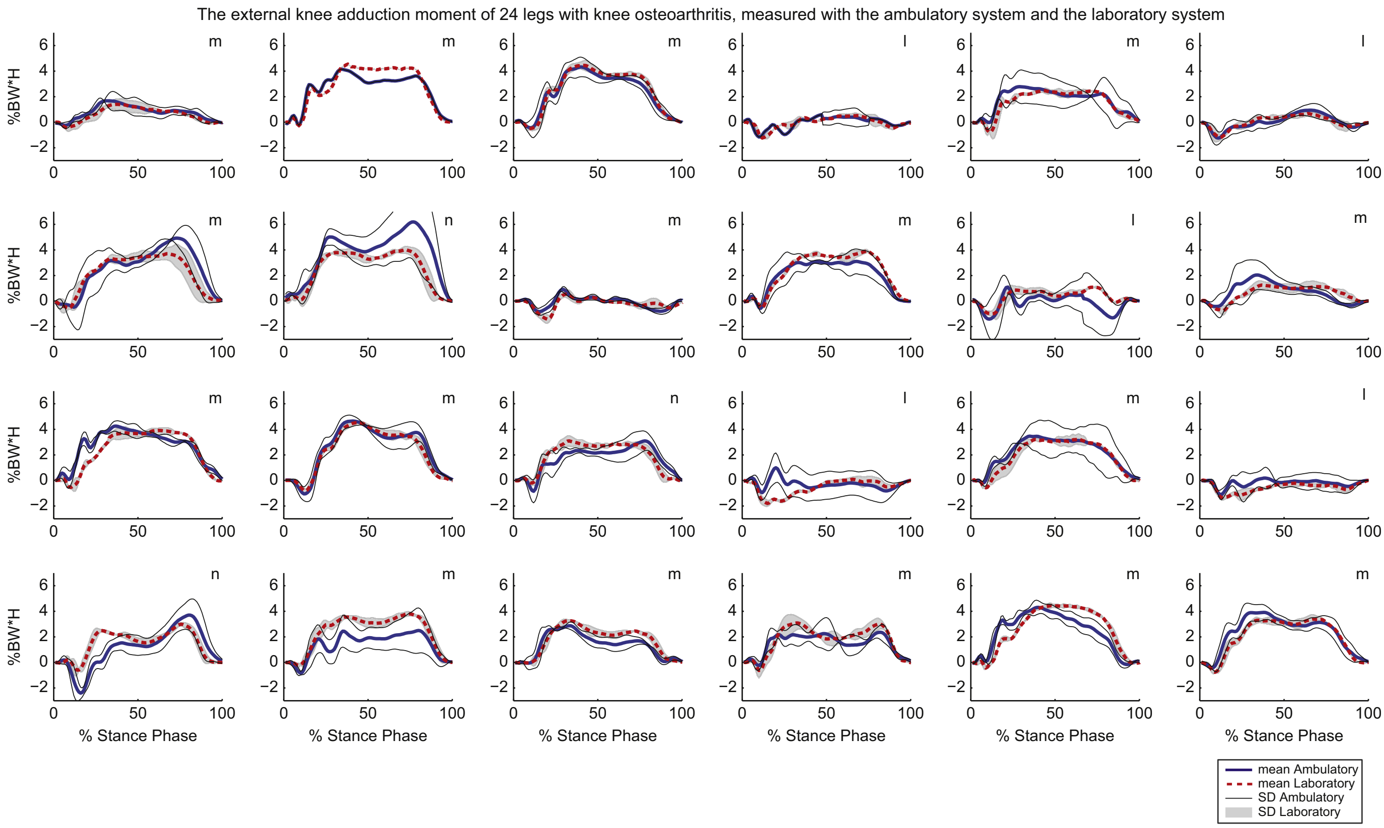

Fig. 2. The net external knee adduction moment (KAdM) during stance of 24 legs in patients with osteoarthritis of the knee (in %BodyWeight*Height), measured by the ambulatory movement analysis system (solid line) and the laboratory movement analysis system (dashed line). The ambulatory system consisted of an instrumented force shoe and sensors of an inertial and magnetic measurement system. The laboratory system consisted of a force plate and an optoelectronic marker system. An indication of dominantly affected compartment of the OA knee is provided: *m* is dominantly affected medial compartment, *l* is dominantly affected lateral compartment, *n* means that dominant compartment is not assessed.

differences and recommendations for optimization of the ambulatory method are described below.

Firstly, a limitation of the linked-segment model is its sensitivity to accurate fixed segment lengths as input. Differences of about 1 cm in knee position could cause differences of 0.64 %BW*H (18% of range) in the KAdM (van den Noort et al., 2012). This sensitivity of the KAdM to position can be explained by the relatively small GRF moment arm with respect to the knee joint center in the frontal plane (van den Noort et al., 2012).

Differences between ambulatory and laboratory estimated knee position were mainly caused by the use of a different model for joint center estimation (i.e. positions of bony landmarks related to segments, such as the ankle defined in the heel segment (ambulatory) or shank segment (laboratory), and the knee defined in the shank segment (ambulatory) or thigh segment (laboratory)). The particular higher difference in late stance may also be explained by movement of the foot in the shoe, causing more inaccurate ankle position estimation in the linked-segment model via the heel orientation and heel–ankle vector (van den Noort et al., 2012). Previously, similar errors in frontal ankle moments using plantar-pressure insoles and IMMS were reported (Rouhani et al., 2011), due to differences in CoP, ankle positions and foot model between ambulatory and laboratory system. In the present study we used an optoelectronic system to estimate fixed segment lengths. Using only a few markers this can easily be done prior to an ambulatory measurement. However, an optoelectronic system is still needed at hand, which limits the application. For a complete ambulatory assessment, a ruler might be used, however this could result in higher errors since slight deviations in position may cause high differences in KAdM. More accurate estimation of segment lengths might be reached using a kinematic coupling algorithm (Roetenberg et al., 2010), or adding ambulatory position information via a magnetic source worn on the body, which measures the relative position of IMMS sensors (Schepers et al., 2010). Optimization of joint position estimation might also be reached using a more advanced linked-segment model, including more segments for e.g. the foot.

Secondly, IMMS differs from an optoelectronic system in sensor orientation estimation. Correct segment orientations as measured via the IMMS are critical in the linked-segment model. For the feet sensors, integration time of angular velocity data was limited to one stride causing no integration drift (Schepers et al., 2007). For the shank sensor, integration was applied over the whole trial (to a maximum of about four strides). Visual inspection of shank angles from IMMS with respect to shank angles from optoelectronic markers showed no integration drift of the IMMS, therefore not expecting to influence the results. When only correction for initial orientation is not sufficient (in case of longer trials), correction using gravity at rest or nearly constant velocity of the segment (during or at the end of the trial) can remove effectively remaining integration drift (Favre et al., 2006). Also the kinematic coupling algorithm could be used to compensate for drift, assuming the movement of the proximal and distal segment to be equal in the joint (Roetenberg et al., 2010).

Thirdly, the anatomical calibration (sensor-to-segment) of the IMMS sensors of the ambulatory system is different from the anatomical calibration of the optoelectronic system that is based on anatomical landmarks. In the ambulatory system it is based on

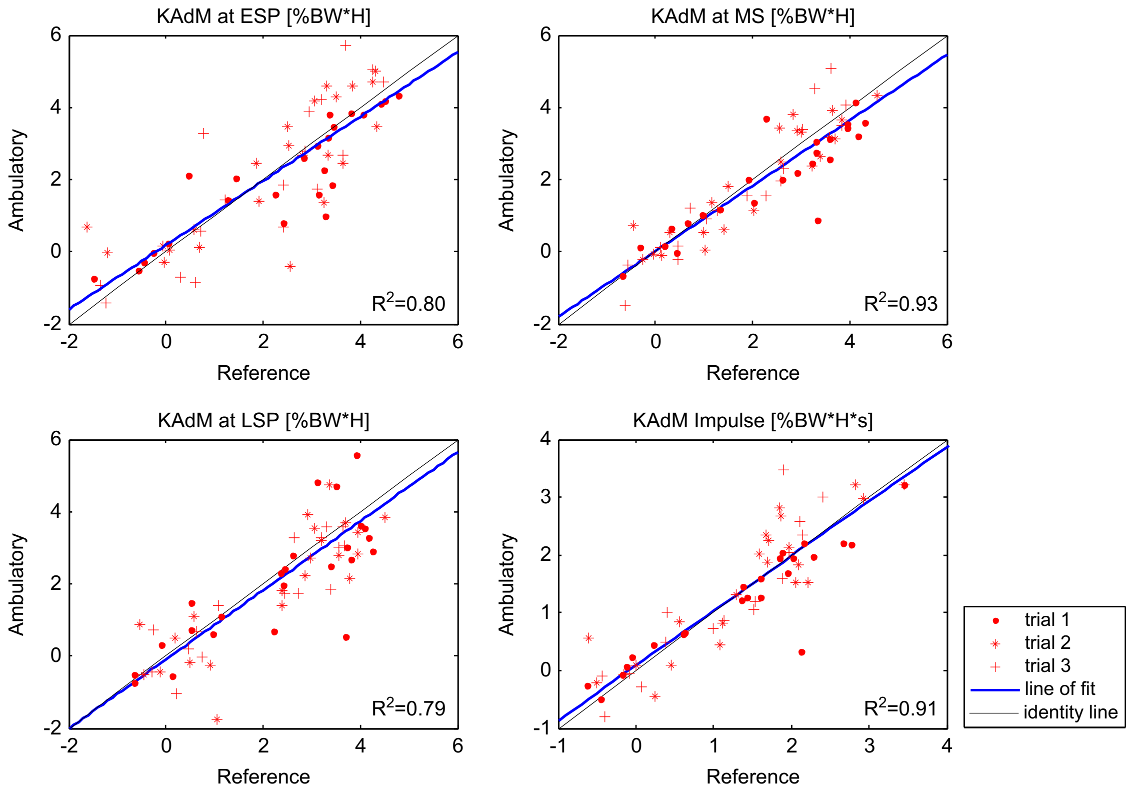

Fig. 3. Scatter plots of the external knee adduction moment (KAdM) of the ambulatory system versus the laboratory system of 24 legs of 14 patients with osteoarthritis of the knee, showing the difference of each individual trial included in the analysis (3 trials per leg). The best line of fit, *R*-square value and the identity line are shown as well. For each trial, data of both systems were measured synchronously. All three trials per leg were included in the statistical analysis. The ambulatory system consisted of the instrumented force shoe, an inertial and magnetic measurement system, and a linked-segment model. The laboratory system consisted of a force plate and an optoelectronic marker system. The KAdM values (in %BodyWeight*Height) are at ESP (early stance peak), MS (midstance), and LSP (late stance peak), and impulse (in %BodyWeight* Height*seconds).

functional movements, static postures with known orientation and/or precise alignment of sensors to anatomical structures (Cutti et al., 2010; van den Noort et al., 2009; O'Donovan et al., 2007). The shank sensor was anatomically calibrated with a standardised flexion/extension movement of the knee and an upright posture. (Favre et al., 2009) 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) to define the anterior-posterior axis. Instead, we used the gravity vector from the upright posture, assuming the longitudinal axis to be aligned with gravity. The IMMS shank coordinate system, based on this calibration, is not per definition equal to a coordinate system based on anatomical landmarks. The knee flexion/extension axis deviates from the axis defined by the malleoli, which are used for the anatomical coordinate system via the optoelectronic marker system (Stagni et al., 2006; Frigo et al., 1998; Ramakrishnan and Kadaba, 1991; Cappozzo et al., 1995). This may apply, in particular, to patients with a high body mass index (BMI) or malalignment (both frequently present in OA), that may cause difficulties to determine the bony landmarks, to perform a pure flexion/extension, to stand in upright posture with shanks and thighs aligned with the gravity, or movement artefacts of the sensors due to the fat percentage under the skin, influencing the anatomical calibration of the IMMS or the gait measurements itself. From the nine legs with a difference higher than 0.90 %BW*H, the errors could not be explained by a higher BMI, varus/valgus knee angle, dominant location of knee OA (medial/lateral), or KAdM value (although the only patient that showed high differences in both legs had the highest BMI (39,6 kg/m^2) of the patient population). Therefore, we cannot draw any conclusion about subpopulations for which the ambulatory technology may not be suitable. The small number of subjects furthermore limits the possibility of dividing the patient population in subgroups.

It should be realized that also the knee joint position measured with the optoelectronic marker system, based on palpation of the femoral epicondyles, is subject to variation, due to the large condylar surface and soft tissue artefacts (Kozanek et al., 2009; Akbarshahi et al., 2010). Therefore, the accuracy and reproducibility of the knee position measured with the laboratory system, and its effect on the KAdM, should be considered as well. To our knowledge, no reproducibility study has been performed on the KAdM yet, although some studies investigated reproducibility of movement analysis. Kinematic errors less than 5° with optoelectronic system are reported (McGinley et al., 2009), although hip and knee rotation angles show larger errors. (Schache et al., 2008) showed that joint moment expression (e.g. KAdM) is dependent of the reference frame that is chosen. A low reproducibility for frontal ankle moments has been observed (Rouhani et al., 2011).

An alternative method for anatomical calibration of the IMMS is palpation of bony landmarks with an IMMS-based calibration device (Picerno et al., 2008). A similar approach for definition of sensor-to-segment coordinate systems for IMMS is then used as for optoelectronic marker systems. It therefore may result in similar kinematic and position estimations.

Future studies should focus on the improvement of the linked-segment model, measurement of fixed segment lengths without using optoelectronic markers, improvement of sensor orientation estimation and of anatomical calibration of the shank. Furthermore, reproducibility of the KAdM needs to be investigated and the ambulatory system should be applied in a larger patient population. When these measures show a good reproducibility and validity, the ambulatory KAdM can be used in direction and evaluation of interventions for knee OA.

5. Conclusion

In clinical practise, the knee adduction moment (KAdM) is an important estimator of knee joint load in knee OA. The present study evaluates the measurement of the KAdM with an ambulatory measurement system consisting of an instrumented force shoe and sensors of an inertial and magnetic measurement system with a standard laboratory system in patients with knee OA. In conclusion, the KAdM measured with the ambulatory system did not show significant differences from the KAdM measured with a laboratory system, when evaluated at group level in patients with knee OA.

Conflict of interest

We certify that no party having a direct interest in the results of the research supporting this article has or will confer a benefit on us or on any organization with which we are associated.

We would like to disclose that one of the authors is employed at Xsens, the manufacturer of the instrumented force shoe and the inertial and magnetic measurement systems used.

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