Chapter 7

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

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Submitted

7.1 Abstract

Biomechanical factors during gait are important determinants of knee-load in patients with osteoarthritis (OA) of the knee. Mechanical loading on the knee is reflected in the external knee adduction moment (KAdM) that can be measured during gait. However, clinical application of conventional laboratory-based measurement systems is limited. Ambulatory movement analysis systems, including an instrumented force shoe (IFS) and an inertial and magnetic measurement system (IMMS), could potentially be used to determine the KAdM in the gait of patients with knee OA 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 the IMMS has yet not been studied. The objective of this study was to evaluate the accuracy of ambulatory movement analysis systems (i.e. IFS and IMMS), to estimate the KAdM during gait of patients with knee OA. Gait analyses of 14 patients with knee OA were performed in a gait laboratory. The KAdM was concurrently determined with two methods: (i) IFS and IMMS in combination with a linked-segment model (to obtain joint positions); (ii) Laboratory: force plate and optoelectronic marker system. The KAdM measured with the ambulatory system did not differ systematically from the KAdM measured with the laboratory-based system, when evaluated at group level. However, on average, absolute differences up to 24% (i.e. 0.83 %BodyWeight*Height) were observed, particularly in early and late stance. For clinical use, the accuracy of the KAdM measurement should be improved by using a more advanced calibrated linked-segment model to estimate the joint position from the orientation of the IMMS.

7.2 Introduction

In recent years, there has been increasing evidence of the importance of biomechanical factors during gait in the initiation and progression of diseases such as osteoarthritis (OA) of the knee [1-4]. Knee OA causes cartilage destruction, subchondral bone-thickening, and new bone formation, and results in knee pain and limitations in daily physical functioning [5]. It occurs in a substantial percentage of the elderly population, and its incidence is likely to increase in the near future with increasing age and obesity. Abnormal joint-loading, due to joint malalignment, laxity, injury and obesity, increases the risk of knee OA, since cartilage has a limited ability to self-repair. [3,5-7]

The mechanical loading on the knee joint is a compressive force, resulting from bodyweight and internal forces, to counteract the external knee adduction moment (KAdM). Hence, the compressive load of the knee is reflected by the KAdM, that can be measured during gait, using kinematic and kinetic data obtained from optoelectronic marker systems and force plates [8,9]. However, clinical use of these conventional systems in gait laboratories is limited, due to the lack of availability of well-equipped laboratories, line of sight problems with markers resulting in missing data, and targeted foot positioning on the force plate causing an adaptation of the gait pattern [10,11]. New ambulatory movement analysis systems to measure kinematics and kinetics during gait have recently been introduced, and can be used in a laboratory-free setting. Instrumented force shoes (IFS) have been applied to measure ground reaction force (GRF) and centre of pressure (CoP) in healthy subjects [10,12] as well as patients [13]. It has been suggested that an inertial and magnetic measurement system (IMMS) should be used to measure the orientation of body segments when an optoelectronic marker system is either not available or not suitable for application [11,14-16]. An IMMS consists of small, lightweight sensor units, of which each sensor unit includes a 3D accelerometer, a 3D gyroscope and a 3D magnetometer. The sensors can easily be attached to a body segment of interest. Fusion filtering algorithms and anatomical calibration procedures can then be used to determine segment orientation and joint angles [11,14].

The combination of IFS and IMMS could potentially be used to determine the KAdM in the gait of patients with knee OA in a laboratory-free setting. However, since it is difficult to measure positions with IMMS, joint position needs to be determined from a linked-segment model that represents the skeletal geometry [13,17]. In a previous study, the IFS were used in combination with such a model to estimate the KAdM in patients with knee

OA [13]. Segment orientations and fixed segment lengths were used as input in the model. Although promising results were reported [13], the model was evaluated with orientations measured by the same optoelectronic marker system that was used for validation, and the application of IMMS has not yet been studied.

Therefore, the aim of the present study was to evaluate the accuracy of the *entire* ambulatory movement analysis system (i.e. IFS *and* IMMS) to estimate the KAdM during gait of patients with knee OA, compared with a laboratory system (optoelectronic marker system and force plate). In this study segment orientations recorded with the IMMS were applied in the linked-segment model, and combined with the information obtained from the IFS for ambulatory measurement of the KAdM. We hypothesized no difference between the ambulatory system and the laboratory system.

7.3 Methods

7.3.1 Subjects

A total of fourteen patients fulfilled the American College of Rheumatology (ACR) criteria for knee OA [18] and participated in the study. They had medial and/or lateral tibiofemoral radiographic OA, with a Kellgren/Lawrence grade of at least grade 1 [19,20]. The patients were recruited from the patient population of the Reade Centre for Rehabilitation and Rheumatology (Amsterdam, the Netherlands). The inclusion criteria were: between 40 and 75 years of age, diagnosed with OA of the knee, and consent to participation. 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.

7.3.2 Procedure

Gait analyses of the patients were performed in a gait laboratory. The patients walked on a 10 metre walkway at a comfortable self-selected speed. During the measurements, kinematic and kinetic data were collected by means of an ambulatory movement analysis system and a laboratory system (as a reference). Data on three successful trials were collected per leg.

The ambulatory system consisted of the IFS (Orthopaedic sandal, with 6-degrees-of-freedom ATI mini45 SI-580-20 force/moment sensors, Schunk GmbH & Co. KG) [10,21] and

the IMMS sensor units (MTx, Xsens Technologies, the Netherlands) attached to the force/moment sensors of the IFS and to the shanks of the patient. The IFS measured GRF and CoP with a sample frequency of 50 Hz. The IMMS sensor units were used to track segment kinematics with a sample frequency of 50 Hz. The IFS and IMMS were wirelessly connected to the computer, via two Xbus Master devices (Xsens Technologies, the Netherlands).

The laboratory system consisted of a force plate (AMTI OR6-5-1000, Watertown, MA, USA) embedded in the floor of the laboratory for GRF and CoP measurement with a sample frequency of 1000 Hz, and an optoelectronic marker system (OptoTrak 3020, Northern Digital Instruments, Waterloo, Canada) with marker clusters attached to the feet, shanks and thighs for tracking segment kinematics with a sample frequency of 50 Hz. The optoelectronic marker clusters were rigidly attached to the sensor units of the IMMS. To determine anatomical coordinate systems, anatomical landmarks were palpated according to Cappozzo et al. [22].

Time-synchronization of the IFS with the laboratory system was achieved with a synchronization pulse from the IFS (including the IMMS on the shoe), recorded in the laboratory system. Post-hoc, cross-correlation was applied to synchronize the IMMS on the shank with the optoelectronic marker system (in MATLAB, R2009b, The Mathworks).

Prior to the gait measurements, an upright static measurement and a passive (non-weight bearing) flexion/extension movement of the patient's knee joint was performed by the examiner for anatomical calibration of the IMMS coordinate system on the shank [16].

7.3.3 Data analysis

The algorithms of Schepers et al. [10] were used to calculate the GRF and CoP of the ambulatory movement analysis system, based on the IMMS and IFS data from the heel and forefoot. The data from the force/moment sensors were used to define the heel strike and toe-off.

The orientations of the IMMS sensors on the heel, forefoot and shank were calculated by means of signals from the gyroscopes and accelerometers in the IMMS. The information from the magnetometers was ignored, since their accuracy can be affected by a non-homogenous earth magnetic field, that is likely to be present near the floor [23]. The angular velocities measured with the gyroscopes were integrated [24], and the gravity vectors from the accelerometers were used to correct the inclination at the start of the trial. The heading (direction) was corrected by the known orientation of the feet at the start of the trials, assuming the shank to be aligned with the feet.

The orientations of the sensors on the heel and forefoot were corrected for each stride, using a zero velocity update and assuming an equal vertical position of the foot at each stride. This means that the inclination at each stride can be estimated with the accelerometers. The integration time was thus limited to minimize integration drift [10]. This type of orientation correction is not possible for the sensor on the shank, since the shank is moving throughout the gait cycle. However, most trials included only three or four gait cycles, which limited the integration drift. For the shank orientation, a sensor-to-segment orientation matrix was first determined, using the gyroscope and accelerometer data from the passive knee flexion/extension movement (i.e. the anatomical calibration of the shank that was available for each leg of each patient) [16,25]. The angular velocities and accelerations of the shank sensor in each entire gait trial were transformed to the anatomical segment coordinate system, using this sensor-to-segment coordinate matrix (for all trials of each leg separately).
The positions of the ankle and the knee with respect to the midpoint of the heel force/moment-sensor during gait were calculated with a linked-segment model [13], assuming the segments to be rigid bodies:

$$ {}^{g}\vec{P}_{knee} = {}^{g}R_{heel} \cdot {}^{heel}\vec{P}_{ankle} + {}^{g}R_{shank} \cdot {}^{ankle}\vec{P}_{knee} $$

(*equation 7.1)*

Inputs of the model were heel and shank orientation during gait (***R***) measured with the IMMS, and fixed segment lengths (***P***). The fixed segment lengths were calculated from optoelectronic bony landmark data from the stance phase of a gait trial, and transformed into the coordinate system of the IMMS. This included the vector from the midpoint of heel force/moment-sensor to the ankle, assuming the ankle to be fixed in the heel segment [10], and the vector from the centre of the ankle joint to centre of the knee joint, i.e. the shank segment. The ankle joint centre was defined as the midpoint between the medial and lateral malleoli. The knee joint centre was defined as the midpoint between the medial and lateral femur epicondyles.
The GRF, CoP, and joint position were transformed to the global coordinate system of the laboratory system. The transformation matrix was calculated as the orientation difference in the GRF of the IFS (in gait direction) and the GRF of the force plate [10,12,13]. The KAdM was calculated from the GRF and its moment arm, defined by the CoP and the knee joint position in the frontal plane of the global coordinate system.

For the laboratory system, the GRF, the CoP, and the segment and joint positions and orientations were calculated from optoelectronic marker data and force plate data in BodyMech (www.bodymech.nl), custom-made software in MATLAB (R2009b, The Mathworks), based on ISB anatomical frames [22,26]. The KAdM was calculated using the GRF from the force plate and its moment arm, defined by the CoP and the knee joint position from the optoelectronic marker data in the frontal plane. The contributions of the weight and acceleration of the centre of mass and moments of inertia of distal segments to the joint moment were not taken into account [27]. The joint moments were normalized to body weight (BW in N) and body height (H in m), i.e. %BW*H.

7.3.4 Statistical analysis

Statistical differences in the KAdM between the ambulatory movement analysis system and the laboratory system were calculated in early stance peak (ESP), midstance (MS), late stance peak (LSP), and impulse, by performing a linear mixed model analysis (SPPS Software, Version 15.0). The 3 trials of each leg with OA were included in the analysis. The ESP and LSP were respectively defined as the maximum values of the first and last 50% of the vertical GRF in the stance phase, and the MS was defined as the minimum value between ESP and LSP of the vertical GRF. The impulse is the time-integral of the KAdM over the stance phase [28]. Prior to the analysis, normal distribution of the data was checked. Statistical significance was determined as a *P*-value of less than 0.05.

Furthermore, the absolute difference (i.e. the absolute value of the difference for each subject) for each parameter was calculated. The root mean square error (RMSE) and the gain (non-dimensional) were calculated for the KAdM over the whole stance phase [13,29].

7.4 Results

Data on a total of 24 legs (three trials per leg) from 14 patients with OA of the knee (3 males, 11 females) were included in the analyses. The mean age of the patients was 61.0±9.2 years (mean ± standard deviation), their mean body mass was 83.7±14.4 kg and their mean height was 1.66±0.11 m. Four legs had to be excluded, due to unilateral OA, limited visibility of the optoelectronic markers, or technical problems with the IMMS or IFS.

Figure 7.1 shows the mean KAdM and its standard deviation (3 trials) for each of the 24 legs, measured with the ambulatory movement analysis system and the laboratory movement analysis system. The figure clearly shows significant inter-subject differences in the KAdM (*P*=0.001).

Compared to the KAdMs measured with the laboratory system, the KAdMs measured with the ambulatory system had greater standard deviations (Figure 7.1), indicating that the inter-trial variability within a patient (i.e. the intra-patient variability) is higher when using the ambulatory system. Table 7.1 shows the mean difference and its significance, the absolute differences, the RMSE and the gain, in the KAdM of the ambulatory system compared to the KAdM of the laboratory system, averaged for all legs included in the analysis. Figure 7.2 presents the scatter plots of the KAdM at ESP, MS and LSP, and from the KAdM impulse, of the ambulatory system versus the laboratory system, showing the difference of each individual trial included in the analysis.

Table 7.1. 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			**difference**		
		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		0.06	0.336
	absolute	0.19±0.12			
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 metres
SD = standard deviation; SE = standard error of difference

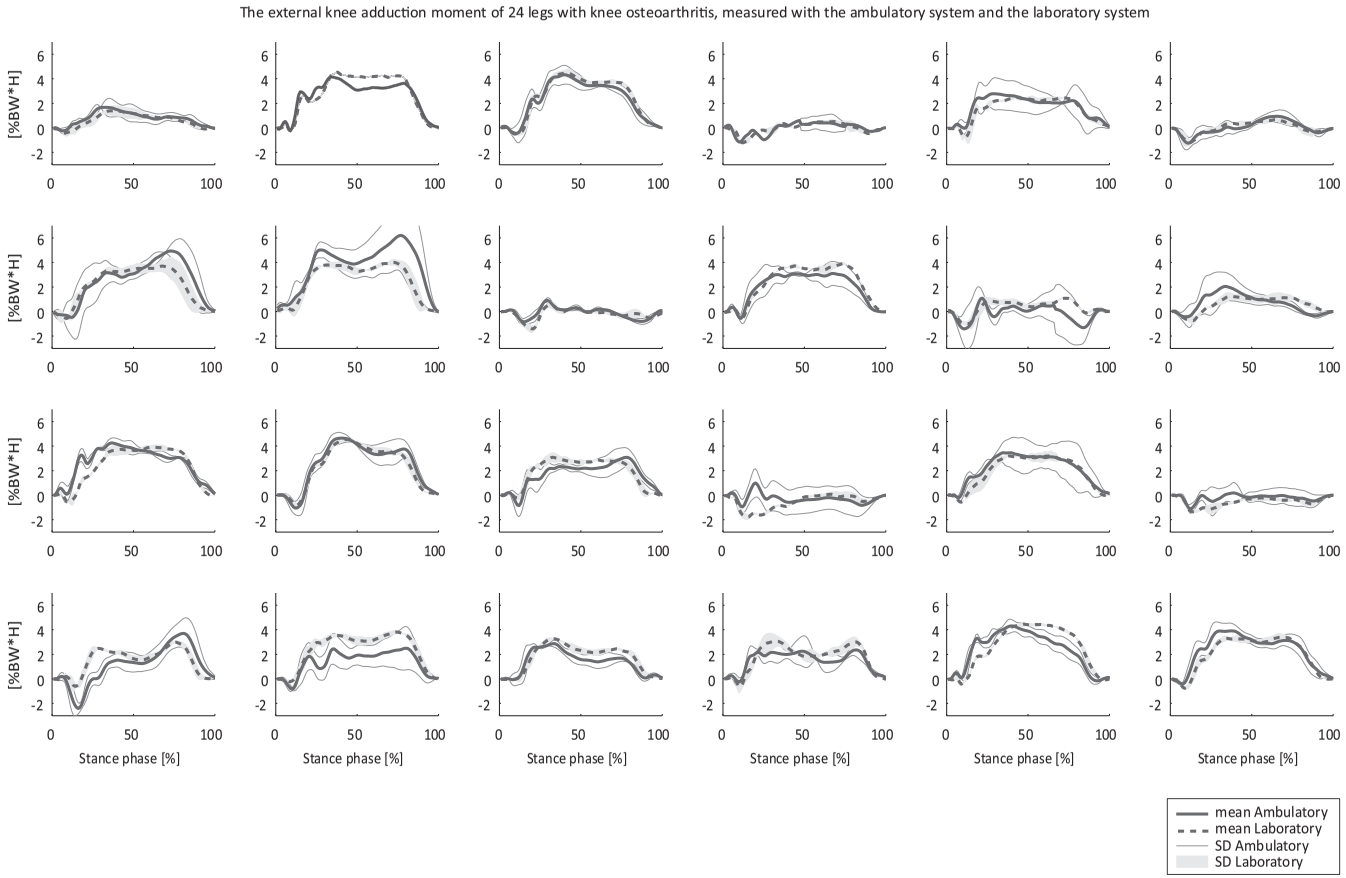

Figure 7.1. The net external knee adduction moment (KAdM) during stance of 24 legs in patients with osteoarthritis of the knee (in %BodyWeight*Height), measured by the ambulatory movement analysis system (solid line) and the laboratory movement analysis system (dashed line). The ambulatory system consisted 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.

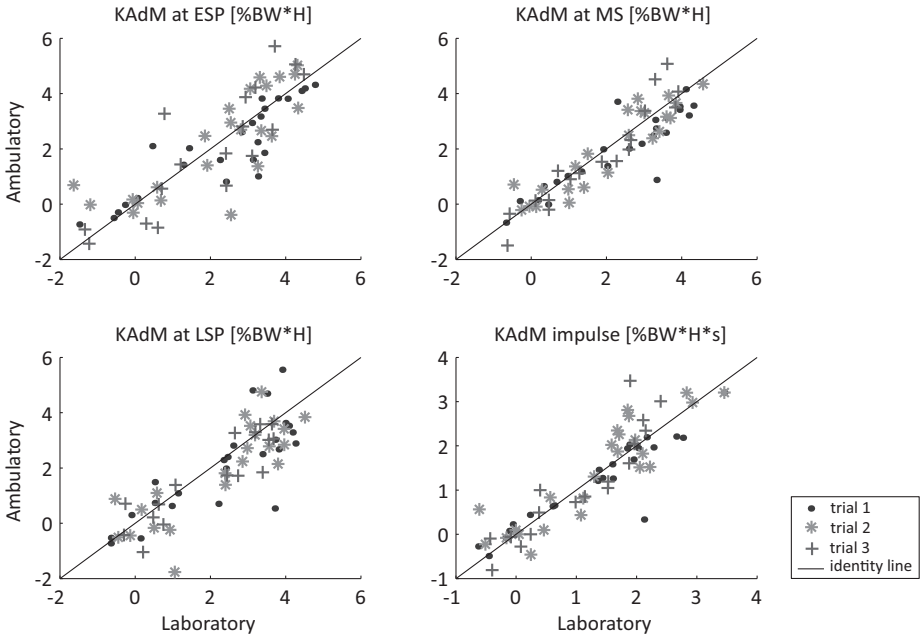

Figure 7.2. Scatter plots of the external knee adduction moment (KAdM) of the ambulatory system versus the laboratory system of 24 legs of 14 patients with osteoarthritis of the knee, showing the difference of each individual trial included in the analysis (3 trials per leg). The ambulatory system consisted 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 shown at ESP (early stance peak), MS (midstance), and LSP (late stance peak), and of the impulse (in %BodyWeight* Height*seconds).

The statistical analyses revealed no significant systematic differences at ESP, MS, LSP or in the impulse of the ambulatory system, compared to the laboratory system. Only the first of the three repeated trials showed a significant difference in the KAdM at MS (P=0.016). However, there was considerable variability within in the group in the differences between the KAdM measured with the ambulatory system and the KAdM measured with the laboratory system (see the absolute difference, the RMSE and the gain in Table 7.1 and Figure 7.2). The greatest differences were found at LSP, with an averaged absolute difference of 0.83 %BW*H, which is 24% of the range of the KAdM measured with the laboratory system over the entire stance phase (i.e. maximum value minus minimum value of the KAdM). Further analyses showed that in the data of nine legs (averaged over

the 3 trials) there was a difference of more than 1 %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).

7.5 Discussion

On average, the KAdM in early stance, midstance and late stance, and the KAdM impulse measured with the ambulatory system, did not significantly differ from the KAdM measured with the laboratory system. This indicates that, at group level, the ambulatory system does not systematically under-estimate or over-estimate the KAdM.

Nonetheless, absolute differences and RMSE values showed differences up to 24% (with respect to the KAdM measured with the laboratory system, i.e. approximately 0.83 %BW*H), particularly in the early and late stance data of nine legs (Table 7.1 and Figure 7.2). In contrast to the results of the present study, in a previous study a systematic under-estimation of the ambulatory KAdM was found in late stance [13]. These different results may be explained by the application of the IMMS in the present study in contrast to the application of the optoelectronic system in the previous study in the linked-segment model. These systems differ in their sensor orientation estimation and their anatomical calibration (sensor-to-segment). However, the mean systematic differences in the KAdM using an optoelectronic marker system for segment orientation measurement, found in the previous study [13], are in the same order of magnitude as the absolute differences and RMSE values presented here. This implies that the application of IMMS causes differences in ambulatory KAdM estimation (with respect to the KAdM measured with the laboratory system) which are of similar magnitude, but not of similar direction, as the application of an optoelectronic marker system for segment orientation (Figure 7.2).

Since the accuracy of the IFS in the measurement of GRF and CoP has been reported to be high [10,12,13], the differences in KAdM are mainly due to joint position estimation via the linked-segment model [13]. This might be the reason for the larger within-patient variability. The model is sensitive for accurate fixed segment lengths (as already reported by Van den Noort et al. [13]) and for sensor orientation as input. The fixed segment lengths have to be expressed with respect to the segment coordinate systems. When using IMMS, the sensor-to-segment calibration is essential.

An anatomical calibration of IMMS may be based on functional movements, static postures and/or precise alignment of the sensor units to anatomical structures [16,25,30].

The sensor on the shank was anatomically calibrated with a flexion/extension movement of the knee and an upright posture. The coordinate system of the shank that is defined via this calibration is not per definition equal to a coordinate system based on anatomical landmarks, since the flexion/extension axis of the knee deviates from the axis defined by malleolus medialis and lateralis, that is used for the optoelectronic marker system [31-34]. This may apply, in particular, to patients with malalignment or a high body mass index (BMI), both of which frequently are present in OA, in which case it may be difficult to determine the bony landmarks or to perform a pure flexion/extension. However, in our study, we found no relationship in the differences between the two measurement systems and BMI or knee malalignment (i.e. the knee varus/valgus). Furthermore, we also found no relationship between the variability within the ambulatory system and BMI or knee malalignment. Therefore, from our results we can not conclude that a high BMI or a large malalignment of the knee are responsible for a difficulty in performing the anatomical calibration.

The accuracy of the ambulatory system needs to be improved, when applying KAdM measurements to discriminate between OA patients and healthy controls. The KAdM differences between patients and healthy controls have been reported to vary about 1 %BW*H (or 20-40% of maximal KAdM) [8,9,35]. In particular the linked-segment model should be adjusted to achieve a more accurate knee joint position estimate (i.e. an error of less than 9mm in medial-lateral direction [13]). However, also the knee joint position measured with the optoelectronic marker system, based on palpation of the femoral epicondyles, may be subject to considerable variation, due to the large condylar surface and soft tissue artefacts [31,32,34,36]. This may affect the medio-lateral knee position, and subsequently the KAdM, estimated with the laboratory system. Therefore, the accuracy of the knee position measured with the laboratory system should be considered as well, and reproducibility of KAdM measurements needs to be further investigated before final conclusions can be drawn.

A few alternative methods that can be used to achieve greater accuracy of the orientation based ambulatory movement analysis system will be discussed below. Whether these methods would optimize joint position estimation, still needs to be investigated. First, an alternative method has been suggested for anatomical calibration of the IMMS: palpation of bony landmarks with an IMMS-based calibration device [37]. This method uses a similar approach to the definition of segment coordinate systems for the IMMS as for optoelectronic marker systems. This may improve the sensor-to-segment calibration of the IMMS.

Secondly, the Kinematic Coupling algorithm might be an alternative way to calculate a more accurate segment orientation [38]. Although integration time was limited, the algorithm used to estimate the orientation of the sensor on the shank still might have suffered from integration drift. The Kinematic Coupling algorithm compensates for this drift using the assumption that the movement of the proximal and distal body segment is equal in the joint. Moreover, it does also not rely on the magnetometers.

Thirdly, the model assumptions for joint positions of the ambulatory system differ from the laboratory system (i.e. the position of bony landmarks related to segments, such as the ankle joint with respect to the heel and shank and the knee joint with respect to the shank and thigh). A more advanced linked-segment model, including more segments, might improve joint position estimation. However, this will be difficult for the foot, due the size of the segments and the IMMS sensor units.

Another approach that may optimize joint position estimation with orientation based ambulatory systems is adding ambulatory position information. This can be done by the application of a magnetic source worn on the body [39], that measures the relative position of IMMS sensors with respect to the source and to each other.

Finally, another option is to indicate joint positions, using only a few markers and an optoelectronic system to track them. This method could be used in a calibration trial to define the joint position movement with respect to the heel during gait in an individual patient. Subsequently, this position could be adapted for measurement outside the laboratory. Although a laboratory optoelectronic marker system would still be required for the calibration, after the initial calibration, the feature of the ambulatory system to measure a large number of consecutive gait cycles can still be used, with improved joint position estimation.

The potential clinical use of KAdM measurement to evaluate disease severity and treatment effects strongly depends on the accuracy, precision and validity of the ambulatory movement analysis system in estimating the KAdM. Whether the KAdM is an important factor in clinical practice is still under discussion. In the literature, the KAdM is generally accepted as an indicator of load distribution on the knee compartments [8,9,35] and it has been associated with knee pain [40]. However, whether the KAdM is an important factor in clinical practice is still under discussion. In a recent study no correspondence of the KAdM first peak in early stance was found to medial knee contact forces measured with a knee implant [41]. Further research is needed to establish the clinical usefulness, as well as the reproducibility, of the KAdM, particularly when the KAdM is measured with an ambulatory system.

7.6 Conclusion

In conclusion, the KAdM measured with the ambulatory system did not differ systematically from the KAdM measured with the laboratory system, when evaluated at group level in patients with knee OA. However, the average individual differences between the two systems were up to 24% (0.83 %BW*H), particularly in early and late stance. For clinical use, the accuracy of the KAdM should be improved by using a more advanced calibrated linked-segment model to estimate joint position from segment orientation.

7.7 References

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