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Real-Time Tracking of Knee Adduction Moment in Patients with Knee Osteoarthritis

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Abstract

Background—The external knee adduction moment (EKAM) is closely associated with the presence, progression, and severity of knee osteoarthritis (OA). However, there is a lack of convenient and practical method to estimate and track in real-time the EKAM of patients with knee OA for clinical evaluation and gait training, especially outside of gait laboratories.

New Method—A real-time EKAM estimation method was developed and applied to track and investigate the EKAM and other knee moments during stepping on an elliptical trainer in both healthy subjects and a patient with knee OA.

Results—Substantial changes were observed in the EKAM and other knee moments during stepping in the patient with knee OA.

Comparison with Existing Method(s)—This is the first study to develop and test feasibility of real-time tracking method of the EKAM on patients with knee OA using 3-D inverse dynamics.

Conclusions—The study provides us an accurate and practical method to evaluate in real-time the critical EKAM associated with knee OA, which is expected to help us to diagnose and evaluate patients with knee OA and provide the patients with real-time EKAM feedback rehabilitation training.

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Keywords

Knee adduction moment; knee osteoarthritis (OA); real-time estimation; biofeedback; diagnosis; outcome evaluation

1. Introduction

Knee osteoarthritis (OA) is an age-associated and chronic degenerative knee joint disease with disabling pain (Lawrence et al., 2008; Lawrence et al., 1998; Oliveria et al., 1995; Reeves and Bowling, 2011); affects more than 27 million Americans (Lawrence et al., 2008; Reeves and Bowling, 2011); is one of the most common cause of chronic disability in the elderly and prevalent with the increase of the human lifespan (Oliveria et al., 1995); and contributes to functional impairments and reduced independence in older adults (Lawrence et al., 1998; Peat et al., 2001; Reeves and Bowling, 2011). Aging is one of the major risk factor for the initiation and progression of degenerative joint diseases such as knee OA (Oliveria et al., 1995) possibly due to decreasing neuromuscular control of knee – especially in adduction-abduction DOF – that may be contributed by reduced muscle (e.g., quadriceps) strength and proprioception acuity with age (Landry et al., 2007; Lawrence et al., 1998; Reeves and Bowling, 2011; Roos et al., 2011), in contrast to healthy subjects' dynamic and static neuromuscular control ability with significant muscle strength in adduction-abduction of their knee within the limited frontal plane range of motion (Zhang and Wang, 2001; Zhang et al., 2001). Medial compartment knee OA is the dominant types of knee OA (McAlindon et al., 1992; Wise et al., 2012), and its pathomechanics is highly related to aberrant compressive load in the medial knee compartment (Englund, 2010; Maly, 2008; Reeves and Bowling, 2011; Shakoor and Moio, 2004).

The compressive load is difficult to measure *in vivo* and noninvasively (Fregly et al., 2007; Winby et al., 2009) and compute (Schipplein and Andriacchi, 1991; Winby et al., 2009). Thus, external knee adduction moment (EKAM) – which can be estimated *in vivo* and noninvasively – is commonly used as a surrogate measure of the compressive load (Foroughi et al., 2009; Miyazaki et al., 2002; Reeves and Bowling, 2011; Shull et al., 2013b; Simic et al., 2011), because increase in EKAM reflects increase in the medial tibiofemoral compartment compressive load (Schipplein and Andriacchi, 1991; Winby et al., 2009).

EKAM is correlated with the severity of medial compartment knee OA (Foroughi et al., 2009; Reeves and Bowling, 2011; Shull et al., 2013b; Simic et al., 2011); and an increased EKAM is a strong predictor of medial knee OA progression (Miyazaki et al., 2002). Especially, the peak EKAM during gait is strongly associated with the presence of medial compartment knee OA (Sharma et al., 1998) and of its symptoms (Thorp et al., 2007), radiographic disease severity (Sharma et al., 1998), and clinical outcomes following surgical realignment of the limb for varus gonarthrosis (Prodromos et al., 1985). Thus, large number of knee OA rehabilitation studies adopted the peak EKAM as a major *outcome measure* with a common goal: to *reduce* the peak EKAM (Barrios et al., 2010; Dowling et al., 2010; Foroughi et al., 2009; Fregly, 2012; Fregly et al., 2007; Hunt et al., 2011; Kang et al., 2013a, b, c; Kang et al., 2011; Miyazaki et al., 2002; Reeves and Bowling, 2011; Riskowski, 2010;

Shull et al., 2011; Shull et al., 2013a; Shull et al., 2013b; Simic et al., 2011). In a recent study, it was found that real-time peak EKAM feedback is helpful to reduce the peak EKAM on a treadmill in a motion analysis laboratory setting (Wheeler et al., 2011). The peak EKAM feedback was estimated as the maximum of the cross product of the moment arm from knee joint center to the ground reaction force (GRF) and the GRF in the frontal plane during the first 40% of stance (Wheeler et al., 2011), which, however, may give us moments of incorrect magnitude and/or wrong polarity (Kang et al., 2013c; Wells, 1981; Winter, 2009). Therefore, it is desirable to estimate the EKAM through inverse dynamics approach in *real-time* for biofeedback gait trainings, diagnosis and outcome evaluations for patients with knee OA.

However, implementation of the 3-D inverse dynamics – a widely used EKAM estimation method with inclusion of inertial contributions – is difficult, especially outside motion analysis laboratories (e.g., for large clinical trials, routine clinical evaluations, and trainings) due to the complex computation involved and the usual requirement of expensive motion capture systems that may occupy large space with extensive setup (Favre et al., 2012; Hunt and Bennell, 2011; Kang et al., 2013a, b, c; Mündermann et al., 2008; Shull et al., 2013a; Shull et al., 2013b; Wheeler et al., 2011). Some efforts to predict EKAM were made to reduce the difficulty using a regression model (Hunt and Bennell, 2011) and an artificial neural network (Favre et al., 2012). However, these methods are not suitable for real-time estimation, because the former only provides group un-normalized peak EKAM and the latter requires percentage of stance phase that cannot be determined in real-time. Therefore, a practical and reliable *real-time* EKAM estimation method is needed for clinical evaluations and real-time biofeedback gait trainings.

Recently, we developed a practical real-time EKAM estimation method on a custom ET (Figs. 1(a) and (b)) – equipped with a 6-axis force/torque (F/T) sensor and a six degree-of-freedom (DOF) goniometer at each footplate (Kang et al., 2013a, b, c) – based on a 3-D inverse dynamics calculation throughout the whole cycle, and verified its feasibility and reliability through experiments with healthy subjects with no previous history of musculoskeletal injuries (Kang et al., 2013a, b, c; Kang et al., 2011). EKAM estimate of healthy subjects during stepping on the ET with 3 different knee postures (regular, knee-adducted, and knee-abducted) from the real-time method well agreed with that from a conventional analysis performed at motion analysis laboratories using an optoelectric motion capture system (Optotrak™ 3020, Northern Digital, Waterloo, Canada).

ETs – which are widely used in many places for exercise and research purposes (Burnfield et al., 2010; Kang et al., 2013a, c; Kang et al., 2011; Lu et al., 2007) – allow closed-chain functional body weight-bearing stepping (Kang et al., 2013c; Lee et al., 2010) similar to over-ground walking (Burnfield et al., 2010; Damiano et al., 2011; Kang et al., 2013c) with kinematic similarity (Burnfield et al., 2010) and much reduced impulsive GRF that is harmful for knee OA (Lu et al., 2007). Moreover, ETs can provide repetitive and consistent training environment that does not require much assistance (Prosser et al., 2011). Further, training on multi-axis ETs, allowing additional pivoting and/or mediolateral sliding DOF movements of each footplate with precisely controlled motors, was found to be promising for knee injury prevention and rehabilitation accompanying improvements in functional

activities, and for the characterization of lower limb neuromuscular control with precise adjustment of footplate's apparent impedance to mimic various environments (Kang et al., 2013a, c; Kang et al., 2011; Lee et al., 2012; Lee et al., 2010, 2011; Lee et al., 2013; Ren et al., 2013; Ren et al., 2010; Ren et al., 2008). Thus, the developed real-time EKAM estimation method may be suitable for assisting diagnosis and outcome evaluations and providing real-time EKAM feedback training of patients with knee OA.

In addition to the peak EKAM alteration, external knee flexion moment (EKFM) of patients with knee OA may be reduced due to weakness of quadriceps (Landry et al., 2007; Roos et al., 2011), pain avoidance (i.e., a compensatory mechanism to reduce pain by reducing the joint articular loads) (Ebert et al., 2010; Kaufman et al., 2001), reduced quadriceps activation (Roos et al., 2011), or increased co-contraction of the quadriceps and hamstring muscle groups (Ebert et al., 2010). Further, alteration in transverse plane rotational moment (e.g., external knee internal rotation moment (EKIRM)) may occur in patients with knee OA (Landry et al., 2007), which may reflect alteration in rotational mechanics related to the knee screw home mechanism, as was observed in gait with quadriceps fatigue (Murdock and Hubley-Kozey, 2012), considering the quadriceps weakness observed in patients with knee OA (Landry et al., 2007; Roos et al., 2011), although association between alteration in EKIRM and pathomechanics and progression of knee OA remained unclear. Thus, real-time estimation of EKFM and EKIRM, which are included in the developed real-time EKAM estimation method (Kang et al., 2013a, b, c; Kang et al., 2011), may help us better understand the pathomechanics of knee OA and compensatory mechanisms of patients with knee OA involving multiple DOFs.

The goal of this study was to develop a real-time EKAM tracking method for the diagnosis and evaluations of patients with knee OA by comparing the EKAM (and EKFM and EKIRM) during stepping on the multi-axis ETs between subjects with and without knee OA. The EKAM (and the EKFM and the EKIRM) of patients with knee OA can potentially be used i) for diagnosis and outcome evaluations and ii) for real-time biofeedback gait training to reduce the EKAM (Kang et al., 2013a, b, c; Kang et al., 2011).

2. Materials and Methods

2.1. Subjects

Seven male controls (age: 25.3 ± 5.8 years; height: 1.81 ± 0.06 m; body mass: 77.6 ± 5.5 Kg; body mass index: 23.86 ± 1.99 Kg/m²), who have no previous history of musculoskeletal injury and neurological impairment, and a female patient with knee OA (age: 75 years; height: 1.60 m; body mass: 60.9 Kg; body mass index: 23.79 Kg/m²), who has suffered from bilateral knee OA ~20 years, were recruited for this study. The study was approved by the institutional review board of Northwestern University. An informed consent was obtained from each participant.

Body mass index (BMI) of the patient was similar to that of control group, and in a healthy weight range (Harding et al., 2012). The patient's normalized Knee Injury and Osteoarthritis Outcome Score (KOOS) (Collins et al., 2011; Roos et al., 1998) – lower represents worse status – was 56% for pain, 57% for symptoms, 68% for activities of daily living (ADL),

20% for sports and recreational activities, and 44% for knee-related quality of life (QOL); and Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) (Collins et al., 2011; McConnell et al., 2001) – higher indicates worse status – was 8 out of 20 for pain, 3 out of 8 for stiffness, and 22 out of 68 for physical function.

2.2. Experimental Procedure

Prior to stepping on the ET, the patient with knee OA filled out the KOOS (Collins et al., 2011; Roos et al., 1998) and WOMAC (Collins et al., 2011; McConnell et al., 2001), the measures of knee function (Collins et al., 2011). The flywheel resistance of the elliptical trainer was set at the lowest level.

With off-line inputs of each subject's kinematic and inertial properties measured or estimated from anthropometric data (Winter, 2009; Zatsiorsky, 2002), the real-time EKAM estimation program estimated not only EKAM but also EKFM and EKIRM in real-time based on the modified 3-D inverse dynamics developed and validated in (Kang et al., 2013a, c) and described in Section 2.3.3. All subjects wore their own gym shoes that are typically available in markets. As described in Subsection 2.3.2, the initial alignment of the 6-DOF goniometer was made after strapping the goniometer to the flat and bony anteromedial surface of subjects' tibia with alignment of one potentiometer of the goniometer with the lateral malleolus (Fig. 1(b)). Prior to data collection, subjects stepped on the ET for ~3 min for familiarization, and foot straps were adjusted to minimize relative motion between foot and the footplate. The subjects then stepped on the ET at their comfortable pace for ~1.5 min with adequate rest after familiarization. During the stepping on the ET, subjects did not firmly hold the ET's handle. All the kinematic and kinetic data – 3-D force ($\mathbf{F}_{F/T}$) and moment ($\mathbf{M}_{F/T}$) measured from the 6-axis F/T sensor, 3-D ankle angle from the 6-DOF goniometer, and the displacement (l_{ob}) of anterior end ($\mathbf{P}_r$) of the long beam ($\mathbf{P}_m$ to $\mathbf{P}_r$) measured from the front potentiometer – were sampled at 100 Hz.

2.3. Method for Estimating External Knee Adduction Moment in Real-time on An Elliptical Trainer

The developed real-time EKAM estimation method will be briefly presented with an introduction of the modified ET and the 6-DOF goniometer measuring ankle kinematics. Details can be found in (Kang et al., 2013a, c).

2.3.1. Modified Elliptical Trainer with the 6-DOF Goniometer—To measure 3-D force ($\mathbf{F}_{F/T}$) and moment ($\mathbf{M}_{F/T}$) exerted to the foot on the footplate, 6-axis F/T sensors (JR3, Woodland, CA) were installed on both sides underneath the footplates of an ET (Reebok® Spacesaver RL) as shown in Fig. 1(a) (Kang et al., 2013a, b, c). Footplate's gravitational and inertial forces and moments were modeled, estimated, and subtracted from the measured 3-D force ($\mathbf{F}_{F/T}$) and moment ($\mathbf{M}_{F/T}$). The displacement (l_{ob}) of anterior end ($\mathbf{P}_r$) of the long beam ($\mathbf{P}_m$ to $\mathbf{P}_r$) was measured by a precision potentiometer positioned at the ET's front end ($\mathbf{P}_{pot}$). 3-D ankle angles (i.e., dorsiflexion/planar-flexion, inversion/eversion, and internal/external rotation) were accurately measured with the 6-DOF goniometer: one part of the goniometer was securely strapped to the flat and bony anteromedial surface of tibia/shank using neoprene bands to prevent slip on skin, and the

other part was attached to the lateral side of footplate with no relative motion to the foot strapped to the footplate. (Fig. 1(b)) (Kang et al., 2013a, c; Lee et al., 2010, 2011; Ren et al., 2013; Ren et al., 2010; Ren et al., 2008).

2.3.2. Kinematics

Frames of Reference: To estimate EKAM conveniently, using palpable anatomic landmarks, four frames of references were defined: world frame, ET frame, ankle anatomical frame (AAF), and tibial anatomical frame (TAF) (Figs. 1(a) and (b)) (Kang et al., 2013a, c). The world frame was fixed to ground and its x , y , and z axes' positive directions were lateral, anterior, and upward directions, respectively, for right leg. ET frame, orientation of which was the same as that of world frame, has its origin at the center of rear driving wheel, $\mathbf{O}_e$ (Fig. 1(a)). AAF was attached to foot with origin at the midpoint between the tips of medial and lateral malleoli, $\mathbf{P}_a$ (Wu et al., 2002). Similar to (Wu et al., 2002), AAF's 3 orthogonal axes (x_a , y_a , and z_a axes) were defined as follows: the projection of a vector from $\mathbf{P}_a$ to the 2nd metatarsal head on to the footplate was defined as y_a -axis; the upward vector normal to the footplate top surface as z_a -axis; and the cross product of y_a - and z_a -axis vector as x_a -axis. TAF (x_k , y_k , z_k) was attached to tibia with its origin (tibial origin) at the midpoint between the peripheral margins of medial and lateral tibial plateaus, $\mathbf{P}_k$, instead of the two impalpable intercondylar eminences (Grood and Suntay, 1983). The vector from $\mathbf{P}_a$ to $\mathbf{P}_k$ was defined as z_k -axis (long axis); the cross product of the z_k -axis and a vector from the medial to lateral tibial plateaus as y_k -axis; and the cross product of y_k - and z_k -axis vectors as x_k -axis (Grood and Suntay, 1983). Hereafter, for all vectors and matrices, the left-leading superscripts denote the frame of reference of the vectors and matrices: o , e , a , and k denote the world frame, the ET frame, AAF, and TAF, respectively.

Kinematic Measurements: To utilize 3-D inverse dynamics for the real-time external knee moment estimation, the footplate angle $\beta (= \varphi_1 + \varphi_2)$, which represents foot orientation on the ET ${}^e\theta_{F/T} = [\beta \ 0 \ 0]^T$, and 3-D ankle angle (i.e., dorsiflexion/planar-flexion, inversion/eversion, internal/external rotation) shown in Fig. 1 were needed (Kang et al., 2013a, c).

Coordinates of $\mathbf{P}_r$, the anterior end of the long beam ($\mathbf{P}_m$ to $\mathbf{P}_r$), can be determined from the distance (l_{ob}) between $\mathbf{P}_r$ and the precision potentiometer at a known position ($\mathbf{P}_{pot}$) and the angle of the oblique linear guide, $\alpha = 20^\circ$ (Fig. 1(a)). Also, it can be determined from the ET's forward kinematics with fixed lengths of the two beams ($\mathbf{O}_e$ to $\mathbf{P}_m$ and $\mathbf{P}_m$ to $\mathbf{P}_r$), l_1 and l_2 , and the two angles, φ_1 and φ_2 . Thus, using this relationship, the two angles, φ_1 and φ_2 , were calculated in real-time by solving the ET's inverse kinematics with l_{ob} measured from the precision potentiometer at $\mathbf{P}_{pot}$, and consequently β . Inverse kinematics yielded two solutions, however, with given initial positions of the two beams, a correct solution could be selected if subjects did not change their stepping direction. With φ_1 and φ_2 thus obtained and l_1 and $l_{2F/T}$ given, the center of the F/T sensor position (${}^e\mathbf{P}_{F/T}$) could be calculated by solving the ET's forward kinematics (Fig. 1(a)) (Kang et al., 2013a, c).

The 3-D ankle angles were measured using the low-cost 6-DOF goniometer, having 5 rotational DOFs and 1 translational DOF (Fig. 1(b)), with two-step calibration as follows. An off-line calibration was performed using a calibration plate, on which all the rotational

DOF angles were zero and the translational DOF distance was known (Kang et al., 2013a, c; Shiavi et al., 1987; Zhang et al., 2003). Note that the zero angles calibrated were not the zero anatomical ankle angles. Prior to subjects' stepping on the ET, an on-line alignment was performed to obtain zero anatomical ankle angles using front and lateral alignment guides (long steel rods) attached to the footplates (Fig. 1(b)). For subjects without knee OA, the alignment was performed at the footplate's lowest position where the subjects bore ~100% bodyweight on that leg; and, for the patient with knee OA, at a position she could adjust her lower leg without pain. Note that the alignment is only for finding zero ankle anatomical angles. In the frontal plane of AAF, the center of the 6-axis F/T sensor, the second metatarsal head, the midpoint between the medial and lateral malleoli, and the tibial tuberosity were aligned. The midpoint between the peripheral margins of medial and lateral tibial plateaus in the frontal plane could be additionally considered, if needed. In the sagittal plane of AAF, the center of the 6-axis F/T sensor, lateral malleolus, and the peripheral margin of lateral tibial plateau were aligned. Dome-shaped rubber pads (diameter: 0.01 m), attached to the aforementioned points, helped alignment. Unlike the typical motion capture systems, this on-line alignment only took few minutes. In this way, a posture reaching the zero anatomical ankle angles was determined (Kang et al., 2013a, c).

Once zero ankle anatomical angles found, all 5 rotational DOF angles and the rotation matrix at the alignment were saved, and, based on that, the 3-D ankle anatomical angles – the x - y - z Euler angle set representing dorsiflexion/planar-flexion, inversion/eversion, and internal/external rotation – were obtained in real-time (Kang et al., 2013a, c). Note that any angle set chosen for the 3-D ankle angle description will yield the same 3-D knee moment theoretically.

2.3.3. Modified 3-D Inverse Dynamics—A modified 3-D inverse dynamics was developed to estimate EKAM (and also EKFM and EKIRM) in real-time without obtaining center of pressure (COP) location (Kang et al., 2013a, c). On the ET, considering that no relative motion between the foot and the footplate was allowed due to the foot straps, determining COP is difficult due to the possible existence of pure moments (also called couple) about x_a - and y_a -axes on the top surface of footplate, which can violate the usual assumption that there are no pure moments about those two axes (Cohen et al., 1980; Winter, 2009; Zatsiorsky, 2002), unlike over-ground walking (Kang et al., 2013a, c). Using the modified inverse dynamics developed, internal knee adduction moment was calculated, and the EKAM was then obtained as the negative of the calculated internal knee adduction moment (Fregly et al., 2007; Hurwitz et al., 1998; Kang et al., 2013c; Moisio et al., 2003) and similarly for EKFM and EKIRM. The EKAM was estimated with respect to the TAF (Dyrby and Andriacchi, 2004; Favre et al., 2012; Kang et al., 2013a, c; Shull et al., 2013a) among mathematically correct frames (Brandon and Deluzio, 2011), considering increase in medial tibiofemoral compartment compressive load is reflected in the increase in EKAM applied to tibia (Schipplein and Andriacchi, 1991). Also, EKFM and EKIRM were estimated with respect to the TAF.

In the process of derivation (Kang et al., 2013a, c), the foot and footplate was considered as one rigid body receiving the 3-D forces ($\mathbf{F}_{F/T}$) and moments ($\mathbf{M}_{F/T}$) measured with the 6-axis F/T sensor during stepping on the ET, because no relative motion between the foot and

footplate existed. Thus, one can draw the free body diagram for foot during stepping on the ET as shown in Fig. 2. From the free body diagram (Fig.2), the following 6-DOF dynamic equation of the foot with respect to the AAF was obtained:

$$m_{Ft} {}^a \mathbf{a}_{Ft} = {}^a \mathbf{F}_{F/T} + {}^a \mathbf{F}_a + m_{Ft} {}^a \mathbf{g} \quad (1)$$

$${}^a \mathbf{I}_{Ft} {}^a \ddot{\boldsymbol{\theta}}_{Ft} + {}^a \dot{\boldsymbol{\theta}}_{Ft} \times {}^a \mathbf{I}_{Ft} {}^a \dot{\boldsymbol{\theta}}_{Ft} = \left({}^a \mathbf{P}_{F/T} - {}^a \mathbf{P}_{Ft} \right) \times {}^a \mathbf{F}_{F/T} + \left({}^a \mathbf{P}_a - {}^a \mathbf{P}_{Ft} \right) \times {}^a \mathbf{F}_a + {}^a \mathbf{M}_{F/T} + {}^a \mathbf{M}_a \quad (2)$$

where m_{Ft} denotes foot mass; ${}^a \mathbf{I}_{Ft}$ foot inertia matrix; ${}^a \mathbf{a}_{Ft}$ the linear acceleration of foot center of mass (COM), $\mathbf{P}_{Ft}$; ${}^a \dot{\boldsymbol{\theta}}_{Ft}$ ${}^a \ddot{\boldsymbol{\theta}}_{Ft}$ the angular velocity and acceleration of foot, respectively; ${}^a \mathbf{g}$ gravitational acceleration; and ${}^a \mathbf{F}_a$ and ${}^a \mathbf{M}_a$ denote force and moment vectors acting on the ankle origin from the shank, respectively. Again, the left-leading superscript a means the vectors and matrices are written with respect to the AAF. Equations (1) and (2) represent translational and rotational 3-DOF dynamics about the foot COM ($\mathbf{P}_{Ft}$), respectively. The ankle force (${}^a \mathbf{F}_a$) and moment (${}^a \mathbf{M}_a$) can be obtained by solving Eqs. (1) and (2). The dynamic equations of foot given in Eqs. (1) and (2) do not require the COP. Instead, for each subject, the rotational dynamics of foot, Eq. (2), needed the fixed distances from foot COM to the center of the F/T sensor (${}^a \mathbf{P}_{F/T} - {}^a \mathbf{P}_{Ft}$) and from foot COM to the ankle origin (${}^a \mathbf{P}_a - {}^a \mathbf{P}_{Ft}$) that can be obtained off-line. The foot and shank masses and inertia matrices also are fixed values for each subject, and can be estimated from anthropometric data (Winter, 2009; Zatsiorsky, 2002). Therefore, the COP computation is no longer needed for the computation of the ankle forces (${}^a \mathbf{F}_a$) and moments (${}^a \mathbf{M}_a$). Thus, the COP computation is not required for the EKAM estimation (and also EKFM and EKIRM) that is, in turn, obtained with the computed ${}^a \mathbf{F}_a$ and ${}^a \mathbf{M}_a$ and the inertial forces and moments of shank. Thus, the EKAM (and EKFM and EKIRM) on the ET can be estimated using the modified 3-D inverse dynamics without computing COP.

This modified 3-D inverse dynamics was utilized for the real-time estimation of the EKAM (as well as EKFM and EKIRM) with kinematic data from the 6-DOF goniometer.

2.4. Data Analysis

Elliptical cycle (EC) was defined as starting at the time when the footplate reached the most anterior position and ending at the very next time when the footplate reached the same position (Kang et al., 2013a, c; Lu et al., 2007).

Ten cycles of data during stepping on the ET were selected for each subject. The 3-D knee moment (EKAM, EKFM, and EKIRM) estimates were normalized to the product of body weight and height and measured in percentage (noted as %BW×HT). These three external knee moments were time normalized using the EC (0-100%) and, for each subject, mean of 10 cycles was obtained. Similarly, 3-D ankle angles were time normalized and ensemble averaged. Alteration in 3-D ankle angles and 3-D external knee moments of the patient were determined to be significant, if the 10 cycles' ensemble average of the patient was outside of 95% confidence interval (CI) across subjects in the control group. Similarly, for any other

variables, ensemble average of the patient with knee OA was compared to 95% CI across subjects in the control group to determine significance.

3. Results

Although stepping speed of the patient with knee OA (27.8 rev/min) was slower than mean stepping speed of control group (36.3 rev/min), the speed of the patient with knee OA was larger than lower limit of 95% CI (27.5 rev/min) of the stepping speed across the subjects in the control group.

3.1. 3-D Ankle Angles

3-D ankle angles of control group and a patient with knee OA are shown in Fig. 3. The patient's mean dorsiflexion angle was always positive throughout the whole EC, similar to control group (Kang et al., 2013a, c). The patient less dorsiflexed her ankle compared to the control group in the interval of 0% to ~30% of EC and of ~44% to ~62% of EC. Substantial difference was found in inversion angle. Compared to the inversion angle of control group, the patient everted her ankle significantly. Maximum eversion angle (10.2°) was much larger than that of control group (0.7° ; 95% CI: $[-2.9, 1.6]^\circ$). Ankle internal rotation angle of the patient was within 95% CI of control group throughout the whole EC.

3.2. 3-D External Knee Moments

3-D external knee moments (EKFM, EKAM, and EKIRM) are shown in Fig. 4, and detailed each individual subject's values are provided in Fig. 5. First of all, peak ensemble average EKAM of the patient with knee OA ($1.72\% \text{BW} \times \text{HT}$) was substantially increased compared to near zero peak mean EKAM of control group ($-0.04\% \text{BW} \times \text{HT}$; 95% CI: $[-0.80, 0.72]\% \text{BW} \times \text{HT}$). EKAM of the patient increased in the interval of 26% to 59% of EC and decreased in the interval of 0% to ~15% and of ~74% to 100% of EC, compared to that of control group. EKFM of the patient was decreased in the interval of 0% to ~77% of EC and of ~95% to 100% of EC. Compared to the near zero minimum EKFM ($0.05\% \text{BW} \times \text{HT}$; 95% CI: $[-1.46, 1.56]\% \text{BW} \times \text{HT}$), that of the patient ($-3.59\% \text{BW} \times \text{HT}$) was substantially smaller. In other words, peak external knee extension moment of the patient with knee OA was larger than that of control group. EKIRM of the patient with knee OA was increased in the interval of ~25% to ~44% of EC and decreased in the interval of 0% to ~15% and of ~58% to ~100% of EC. Compared to the maximum EKIRM of control group ($0.21\% \text{BW} \times \text{HT}$; 95% CI: $[0.17, 0.26]\% \text{BW} \times \text{HT}$), that of the patient ($0.42\% \text{BW} \times \text{HT}$) was 2 times larger. Minimum EKIRM of the patient ($-0.20\% \text{BW} \times \text{HT}$) was smaller than that of control group ($0.05\% \text{BW} \times \text{HT}$; 95% CI: $[0.02, 0.10]\% \text{BW} \times \text{HT}$).

4. Discussion and Conclusion

In response to strong clinical and research needs to investigate and control the EKAM and other two direction external knee moments (EKFM and EKIRM), for the first time, a practical real-time EKAM estimation method on a custom ET based on a 3-D inverse dynamics calculation throughout the whole cycle was developed, and its capability of

capturing alteration in external knee moments – not only EKAM but also EKFM and EKIRM – in patients with knee OA was demonstrated and verified through experiments.

First of all, increase in peak EKAM of the patient with knee OA – which is highly related to the pathomechanics of knee OA, strongly associated with knee OA, and the major outcome of gait training of patients with knee OA – on the ET was first observed and reported. Moreover, reduction in EKFM of the patient with knee OA was detected. Similar reduction was observed in previous gait studies and may be due to the quadriceps weakness (Landry et al., 2007; Roos et al., 2011), a compensatory mechanism to reduce pain (Ebert et al., 2010; Kaufman et al., 2001), reduced quadriceps activation (Roos et al., 2011), or increased co-contraction of the quadriceps and hamstring muscle groups (Ebert et al., 2010). Thus, this reduction might be due to biomechanical changes (i.e., quadriceps weakness), neurological changes (i.e., the compensatory mechanism, reduced quadriceps activation, increased co-contraction of the quadriceps and hamstring muscle groups), or combination of the two. Further, larger variation in EKIRM of the patient with knee OA, compared to that of subjects without knee OA, was observed during stepping on the ET. Alteration in EKIRM was reported during over-ground walking of patients with knee OA (Landry et al., 2007). Alteration in EKIRM might reflect the alteration in rotational mechanics related to the knee screw home mechanism (Murdock and Hubley-Kozey, 2012), but it is not clear whether the alteration in EKIRM is related to the pathomechanics and progression of knee OA. Note that these results may only be minimally affected by stepping speed considering that the stepping speed difference between control group and the patient were not significant.

In addition to the 3-D external knee moments, changes in ankle dorsiflexion and inversion angles of the patient with knee OA were found. Ankle of the patient was less dorsiflexed and more everted compared to the control group. Since, on the ET, no relative motion between foot and footplate existed due to foot strap, increase in ankle eversion meant most probably increase in knee adduction angle (Kang et al., 2013a, c), which can increase EKAM by lengthening the moment arm from the knee joint to the GRF (Barrios et al., 2010; Reeves and Bowling, 2011; Simic et al., 2011), accompanied with possible body leaning (Kang et al., 2013a, c). EKAM was found to be strongly and positively associated with ankle eversion angle during stepping on the ET in authors' previous study (Kang et al., 2013a, c).

Overall, for the first time, the changes in 3-D knee moments and 3-D ankle angles of the patient with knee OA during stepping on the ET were found by using the developed real-time EKAM estimation method without typical long time preparation and post processing of motion capture data (Kang et al., 2013a, c). Indeed, right after the stepping trial, the patient was able to see this summarized results displayed.

The proposed real-time EKAM estimation method is promising in that the proposed method utilizes a compact and low-cost 6-DOF goniometer instead of an expensive motion capture system that usually occupies large designated space in motion analysis laboratories; that the proposed real-time method needs only a few minutes for initial alignment and no post processing for computing the EKAM (and EKFM and EKIRM) in contrast to other estimation methods that require cumbersome setup and long time post processing for computing EKAM (and EKFM and EKIRM); and that the proposed method can be used

virtually anywhere the modified ET is installed and suitable for clinical and gymnasium settings.

4.1. Measurement Methods of Kinematic Variables

3-D motion capture systems usually have their own difficulties in using them (Kang et al., 2013c; Zhou and Hu, 2008). For visual/optical motion capture systems, one need to consider the occlusion of the limb of interest by other body parts and/or devices including marker dropouts in optoelectric motion capture system. In using sensors/markers attached to skin, one need to take into account the error due to skin movement and preparation of attaching the sensors/markers. For accelerometer/gyroscope, drift in the integration of measured acceleration/velocity can degrade sensing performance. For magnetometer, distortion in sensed magnetic fields due to electromagnetic motors and/or ferrite structures around can occur.

The 6-DOF goniometer utilized in this study was free from most of the difficulties, except for the one that it might slip on subject's skin, because the goniometer was able to directly sense ankle 3-D rotation with firm and physical attachment to the lower leg using neoprene bands to reduce slip on skin.

Other practical advantages of the goniometer are that it only needs small space around the footplate; that it is a low-cost device compared to other typical motion capture systems; that outputs of the goniometer are all analog signals so that the goniometer can be connected to any analog to digital (A/D) data acquisition system conveniently with user selected sampling time. Thus, the goniometer can be an affordable alternative to other motion capture systems, especially for motion capturing outside motion analysis laboratories.

Since the modified 3-D inverse dynamics is not specific to the goniometer, wireless inertial measurement units (IMUs) – a combination of accelerometers, gyroscopes, and magnetometers – may be used to measure the 3-D kinematics instead of the goniometer for real-time EKAM estimation, provided that the IMU's aforementioned potential difficulties can be properly addressed for accurate measurement of kinematic variables.

4.2. Potential Clinical and Practical Benefits of Real-Time EKAM Estimation and Feedback

Gait training programs utilized real-time biofeedback of various locomotion variables (other than EKAM) to reduce EKAM of patients with knee OA. The locomotion variables fed-back to patients were mostly kinematic variables related to knee (Barrios et al., 2010; Reeves and Bowling, 2011; Riskowski, 2010; Shull et al., 2011; Shull et al., 2013a; Shull et al., 2013b; Simic et al., 2011) and trunk (Hunt et al., 2011; Shull et al., 2011; Shull et al., 2013b), and a kinetic variable, lateral foot pressure (Dowling et al., 2010). These biofeedback were used for gait training with gait modification strategies designed prior to the training (Fregly et al., 2007; Shull et al., 2011; Shull et al., 2013b; Simic et al., 2011), and the trainings succeeded in reducing the EKAM to a varying degree (Fregly, 2012; Simic et al., 2011). During these gait trainings, EKAM was not available at least until the end of a training session (including long post processing time to compute EKAM) and it might not be, thus, possible to check during the training how the EKAM is changing and whether the selected gait training

strategy is helpful for each subject or not. This might explain the varying degree of reduction of EKAM across studies (Fregly, 2012; Simic et al., 2011).

With the proposed real-time EKAM estimation method, patients can receive continuous and direct EKAM feedback in *real-time*. Thus, it is possible for the patients and clinicians to immediately check if the gait modification strategy each patient took is effective and what is an effective strategy for each patient during a training session (Kang et al., 2013a, c).

To authors' best knowledge, no study ever estimated all three components of external knee moments in real-time. However, as was demonstrated in this study and also in other gait studies reporting external knee moments (Ebert et al., 2010; Kaufman et al., 2001; Landry et al., 2007; Murdock and Hubley-Kozey, 2012), all these 3 components exhibited alterations, which may be related to the pathomechanics and progression of knee OA. Therefore, it is beneficial to estimate 3-D knee moments in real-time to provide immediate and comprehensive clinical evaluation, and to provide (subject-specific) targeted biofeedback gait training depending on the evaluation. Thus, the developed real-time EKAM estimation method – which can provide us not only EKAM but also EKFM and EKIRM – is expected to significantly contribute to evaluation and gait training of patients with knee OA, and may provide us a deeper understanding of underlying pathomechanics of knee OA.

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Abbreviations

AAF	Ankle Anatomical Frame
EC	Elliptical Cycle
EKAM	External Knee Adduction Moment
EKFM	External Knee Flexion Moment
EKIRM	External Knee Internal Rotation Moment
ET	Elliptical Trainer
TAF	Tibial Anatomical Frame

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Highlights

- Real-time knee moment tracking in knee osteoarthritis.
- Patient with knee OA showed substantial changes in knee moments during stepping.
- The method can be used for clinical evaluations and biofeedback gait training.

Fig.1.

A modified elliptical trainer (ET). (a) Sagittal plane view of the ET with a right leg (step length: 0.47m). (b) A compact and low-cost 6-DOF goniometer attached to the right leg of a subject on the ET. The ET was instrumented with a 6-axis F/T sensor (yellow box written F/T, $\mathbf{P}_{FT}$) underneath each footplate and a precision potentiometer ($\mathbf{P}_{pot}$) measuring distance (l_{ob}) along the $\alpha=20^\circ$ angled linear guide. $\mathbf{P}_{Sk}$ and $\mathbf{P}_{Ft}$ denote center of mass (COM) of shank and foot, respectively. l_1 , l_2 , and $l_{2F/T}$, the constant lengths, were 0.24m, 1.19m, and 0.77m, respectively. One end of the 6-DOF goniometer was securely strapped to the flat and bony anteromedial surface of tibia/shank to minimize slip of the goniometer on subject's skin, and the other end was attached to the lateral of the footplate while aligning one potentiometer of the goniometer with lateral malleolus. The easily removable alignment guides (long steel rods) at the lateral and front of the footplate helped initial alignment of the ankle and knee joints.

Fig. 2.

A free body diagram of foot. Ankle anatomical frame (AAF) is shown with its 3 orthogonal axes (x_a , y_a , and z_a). The left-leading superscript a means that the vectors and matrices are written with respect to the AAF. $\mathbf{P}_a$ denotes the origin of AAF; $\mathbf{P}_{Ft}$ the foot center of mass (COM); $\mathbf{P}_{F/T}$ the center of the 6-axis F/T sensor. ${}^a\mathbf{F}_{F/T}$ and ${}^a\mathbf{M}_{F/T}$ denote 3-D force and moment vectors measured by the 6-axis F/T sensor, respectively, with subtraction of the gravitational and inertial contributions of footplate. ${}^a\mathbf{F}_a$ and ${}^a\mathbf{M}_a$ denote 3-D force and moment vectors acting on the ankle origin from the shank, respectively. ${}^a\mathbf{g}$ denotes 3-D gravitational acceleration vector. ${}^a\mathbf{F}_{F/T}$ and ${}^a\mathbf{M}_{F/T}$ are measured values, and the two vectors $-({}^a\mathbf{P}_{F/T}-\mathbf{P}_{Ft})$ and $(\mathbf{P}_a-{}^a\mathbf{P}_{Ft})$ are constant vectors that can be obtained off-line for each subject.

Fig. 3.

3-D ankle angles during stepping on the elliptical trainer. From the top to the bottom plot, dorsiflexion angle, inversion angle, and internal rotation angle, respectively. Grey solid line represents mean angle across subjects in control group (10 cycles) and light grey shade stands for 95% confidence interval (CI) across subjects in control group. Black solid line represents ensemble average angle of a patient with knee OA (10 cycles), and dark grey shade stands for 95% CI of the ensemble average. Red circle markers over the angle of the patient mean that, at those % elliptical cycle, the ensemble average angle of the patient was significant different from mean angle across subjects in control group (i.e., the angle of the patient was outside of 95% CI across subjects in control group).

Fig. 4.

3-D external knee moments normalized to body weight $\times$ height (%BW $\times$ HT) during stepping on the elliptical trainer. From the top to the bottom plot, external knee flexion moment (EKF), external knee adduction moment (EKAM), and external knee internal rotation moment (EKIRM), respectively. Grey solid line represents mean moment across subjects in control group (10 cycles) and light grey shade stands for 95% confidence interval (CI) across subjects in control group. Black solid line represents ensemble average moment of a patient with knee OA (10 cycles), and dark grey shade stands for 95% CI of the ensemble average. Red circle markers over the moment of the patient mean that, at those % elliptical cycle, the ensemble average moment of the patient was significant different from mean moment across subjects in control group (i.e., the moment of the patient was outside of 95% CI across subjects in control group).

Fig. 5.

Individual subjects' normalized 3-D external knee moments during stepping on the elliptical trainer. From left to right, each column displays 3-D knee moments of each subject. From top to bottom, each row presents the external knee flexion moment (EKFM), external knee adduction moment (EKAM), and external knee internal rotation moment (EKIRM), respectively. The grey solid line represents ensemble average moment (10 cycles) of each control group subject, and light grey shadow stands for 95% confidence interval (CI) of the each subject's knee moment. For comparison, 3-D knee moments of the patient with knee OA, which are the same as those given Fig. 4, are also provided. Black solid line represents ensemble average of the patient's knee moment (10 cycles), and dark grey shade stands for 95% CI of the ensemble average. EKFM of the patient was decreased compared to that of each control group subject. Peak EKAM of the patient was increased compared to that of each control group subject. Larger variation in EKIRM of the patient, compared to that of each control group subject, was observed.