New Accelerometric Method to Discriminate Between Asymptomatic Subjects and Patients With Medial Knee Osteoarthritis During 3-D Gait

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***Abstract*—This study presents a new method to estimate 3-D linear accelerations at tibial and femoral functional coordinate systems. The method combines the use of 3-D accelerometers, 3-D gyroscopes and reflective markers rigidly fixed on an exoskeleton and, a functional postural calibration method. Marker positions were tracked by a six-camera optoelectronic system (VICON 460, Oxford Metrics). The purpose of this study was to determine if this method could discriminate between medial osteoarthritic and asymptomatic knees during gait. Nine patients with osteoarthritic knees and nine asymptomatic control subjects were included in this study. Eighteen parameters representing maximal, minimal, and range of acceleration values were extracted during the loading and preswing to mid-swing phase periods, and were compared in both groups. Results show good discriminative capacity of the new method. Eight parameters were significantly different between both groups. The proposed method has the potential to be used in comprehending and monitoring gait strategy in patients with osteoarthritic knee.**

***Index Terms*—Accelerometer and gyroscope, biomechanics, exoskeleton, gait analysis, internal and external accelerations, knee osteoarthritis (OA).**

I. Introduction

OSTEOARTHRITIS (OA) is the most common type of musculoskeletal disorder, and the knee remains one of the most affected joints [1]. The physiopathology of knee OA is complex and involves interrelated biological, structural, and mechanical factors that are still not yet clearly understood [2].

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Current treatments are unable to prevent the progression of the pathology but only manage the pain relative to the joint [1], [3]. Exercise and physical therapies have shown good results on the reduction of pain and disability [4]. However, the treatment consequences on knee biomechanics during functional activities remain unclear. To establish the effectiveness of OA treatments on knee biomechanics, effective tools are needed and must be able to quantify biomechanical factors associated with knee OA.

Impulsive loading has been closely linked to the onset and progression of knee OA. In an *in vitro* accelerometric study, Chu and Yazdani-Ardakani [5] reported a reduction of 5% of load attenuation capacity in a degenerative knee compared to healthy one. Hoshino and Wallace [6] investigated the absorbing properties of knee joint during longitudinal impulsive loads and found a significant decrease in the absorbing capacity in a degenerative knee. In physiological situation, studies have also been conducted to understand the capacity of the human body to deal with impulsive loading. In locomotor activities, every foot-ground contact induces an internal joint loading and a shock wave traveling from the foot to the head [7]. The transmission of shock waves during gait has been studied using accelerometers [7]–[16] and optoelectric systems [17], [18]. Radin *et al.* [13] used accelerometers fixed on the lateral side of the shank and the thigh (5-cm below and above the knee joint line) and showed a significant difference in longitudinal tibial and femoral accelerations between painful and asymptomatic knees at initial foot contact. Ogata *et al.* [14] and Yoshimura *et al.* [15] assessed medial lateral (ML) accelerations at the tibial tubercle during gait. Ogata *et al.* [14] evaluated the effect of wedge insoles in patients with knee OA and found a decrease of 33% in medial acceleration at the initial contact with the use of insoles. Yoshimura *et al.* [15] compared ML tibial accelerations between anterior cruciate ligament (ACL) deficiency and normal knees and showed significant higher ML peak accelerations on ACL deficiency. Recently, Henriksen *et al.* [16] showed that longitudinal acceleration measured at tibial tubercle level does not differ in either healthy subjects or patients with painful knee OA. On the basis of this literature review, it is difficult to draw a conclusion about the use of accelerometric data to distinguish between OA and healthy knee. In fact, two main limitations were identified here: the fixation and the location of accelerometers onto the body segment.

Skin-mounted fixation of accelerometers induces important artefacts during locomotors activities [8], [19]. Light *et al.* [8] investigated the influence of the accelerometer fixation

TABLE I
INFORMATIONS RELATIVE TO PARTICIPANTS

Groups	Age (years)	Sex	Weight (kg)	Height (m)	Gait velocity (m/s)	OA grade (KL)
	60	F	56.93	1.60	1.03	n/a
	67	F	52.3	1.49	1.03	n/a
	81	M	80.97	1.69	0.53	n/a
	64	M	83.35	1.63	0.69	n/a
Asymptomatic	70	F	61.68	1.59	1.11	n/a
(n = 9)	62	M	84.05	1.82	1.17	n/a
	67	F	48.24	1.59	0.61	n/a
	55	F	51.8	1.62	0.94	n/a
	68	M	77.34	1.73	0.53	n/a
	66 (7.3)		66.3 (14.9)	1.64 (0.09)	0.85 (0.26)	
	60	M	128.79	1.78	0.83	4
	66	M	92.28	1.68	1.08	1
	64	F	62.83	1.45	0.95	3
	62	F	90.81	1.50	1.14	4
Osteoarthritis	71	F	62.23	1.50	0.64	3
(n = 9)	67	F	65.94	1.53	0.92	4
	56	M	98.91	1.70	0.97	1
	59	F	60.79	1.59	0.83	2
	66	F	60.56	1.47	0.61	4
	63.4 (4.6)		80.3 (23.9)	1.58 (0.17)	0.89 (0.18)	

n/a: not applicable

by comparing bone-mounted and skin-mounted techniques with two accelerometers fixed below the tibial tubercle. The authors reported that signals collected with both techniques were approximately of the same magnitude. However, a loss of high frequency and vibration were qualitatively observed from the skin mounted sensor signal. Lafortune *et al.* [19] had also quantified the difference between both techniques during a running task. They reported a substantial increase in magnitude for skin-mounted accelerometer at the tibial level. Although bone-mounted techniques reduce skin movement artefacts, they are still too invasive for clinical use. Therefore, different knee exoskeletons have been developed to reduce artefacts induced by skin-mounted techniques [20]–[23]. Recently, Sudhoff *et al.* [23] compared the displacement of three exoskeletons after fifty gait cycles using an EOS low dose biplanar X-ray system and reported that the exoskeleton proposed in [20] was the most stable.

Although novel fixation techniques are used to limit skin movement artefacts, the location of sensor on the human body segment (e.g., tibial tubercle, lateral side of the shank) is another important limitation in accelerometric studies. In fact, the acceleration magnitude is closely related to the location of the sensor along the segment and its angular velocity [5], [24], [25]. The human musculoskeletal system has natural shock absorbers such as bones and soft tissues that influence the shock wave transmission along body segments. Consequently, the magnitude of the acceleration signal depends on the location of the sensor along the segment [5]. Moreover, if the accelerometer is positioned at a distal location from the center of rotation of the joint, the component of linear acceleration will include a greater angular component than the one measured proximally [24].

These two limitations (i.e., fixation and location of accelerometers) induce difficulty when comparing results between past accelerometric studies because large variations exist between data results [7], [8], [13], [16], [26]. It is believed that these variations can be reduced by the measurement of the linear accelerations at the same location onto the segment. The authors also hypothesize that a significant difference in the pattern of linear acceleration exists between OA and the healthy knee when measurements are taken close to the joint contact surfaces instead of at an arbitrary location onto the segment.

Recently, Dejnabadi *et al.* [27] developed a virtual accelerometer sensor positioned at the knee joint center to measure 2-D knee flexion angle on the basis of an external skin accelerometer. This study proposed to develop 3-D tibial and femoral virtual accelerometers located close to knee joint contact surfaces and on the basis of external accelerometers and gyroscopes fixed rigidly on the knee exoskeleton system proposed in [20].

The purpose of this study was to show that the new accelerometric method was able to discriminate between a group of asymptomatic subjects and patients with knee osteoarthritis during a 3-D gait analysis.

II. METHOD

A. Subjects

Nine patients with knee OA were included in this study. All patients had predominant medial knee OA diagnosed by a physician, confirmed radiographically with the criteria developed by Altman *et al.* [28], and graded with the Kellgren–Lawrence scale (1 to 4). The patients were excluded if they had vestibular, neurological, or musculoskeletal disorders, fracture of the lower extremity, rheumatoid arthritis, or generalized osteoarthritis, limping gait or any condition that could affect a treadmill walking evaluation. The mean age, weight, and height were respectively: 63.4 (4.6) years, 80.4 (23.9) kg, 1.58 (0.1) m (Table I).

Nine asymptomatic subjects were included as a control group. The asymptomatic subjects were evaluated by a physician and were excluded if they had orthopedic (joint fracture, joint laxity,

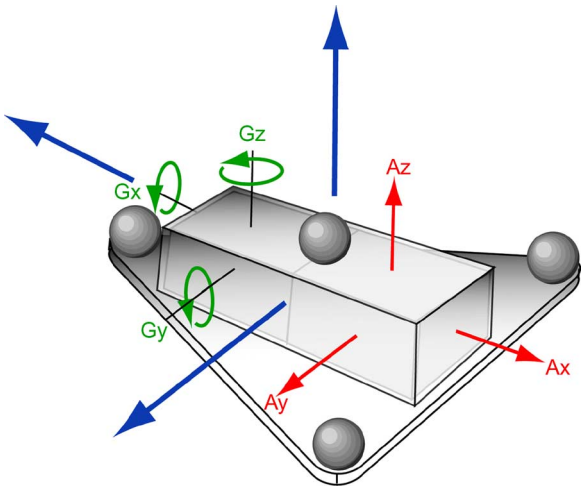

Fig. 1. Design of the rigid body including four reflective markers, one triaxial accelerometer, and one triaxial gyroscope. Coordinate systems axes of the accelerometer (Ax, Ay, and Az), gyroscope (Gx, Gy, and Gz), and rigid body are illustrated by arrows.

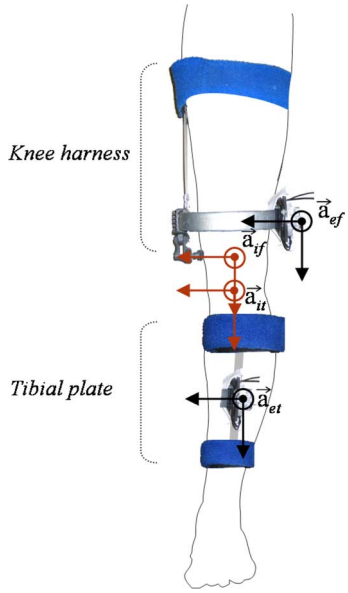

Fig. 2. Frontal view of the exoskeleton consisted of femoral and tibial parts. Coordinate systems axes of the external and internal accelerations of the tibia $(\vec{a}_{et}, \vec{a}_{it})$ and the femur $(\vec{a}_{ef}, \vec{a}_{if})$ are identified.

osteoarthritis, arthritis) or neurological problems that could affect their gait pattern. The mean and standard deviation (SD) of age, weight, and height were 66 (7.3) years, 66.3 (14.9) kg, and 1.64 (0.1) m, respectively (Table I).

Both patients and asymptomatic subjects gave their written consent to participate in this study, which was approved by institutional ethics committees.

B. Instrumentation

Kinematics data were collected using a six-camera optoelectronic system (VICON 460, Oxford Metrics). Tibial and femoral linear accelerations and angular velocities were collected with two triaxial accelerometers (ADXL320, ± 5 g) and two triaxial gyroscopes (Murata, ENC-03J, $\pm 400°/s$), respectively. The signals from the sensors were recorded on a portable data logger (Physilog, BioAGM, CH). Rigid bodies were designed to fix and to align the triaxial accelerometer and gyroscope reference system with the body coordinate system. Four reflective markers enabled the determination of the rigid body coordinate system (Fig. 1). Two rigid bodies were fixed onto an exoskeleton which has been previously validated during gait [23], [29], [30]. The exoskeleton included femoral and tibial parts (Fig. 2). Additional reflective markers were respectively fixed onto lateral and medial malleoli and onto the sacrum using a sacral belt.

C. Defining Femoral and Tibial Coordinate Systems

Femoral and tibial coordinate systems were defined using a functional and postural approach (FP method) [31]. The hip joint centre (HJC) was defined by an optimization method during a leg circumduction movement. The knee joint centre (KJC) was defined by projecting the midpoint from the lateral and medial femoral epicondyles on a mean helical knee flexion-extension axis. The ankle joint centre (AJC) corresponded to the midpoint between lateral and medial malleoli. The longitudinal axis of the femur corresponded to the vector from the HJC and to the KJC, whereas the longitudinal axis of the tibia corresponded to the vector from the KJC and to the AJC. The frontal, sagittal, and transverse planes were defined while the subject performed a small knee flexion movement near full extension. During this movement, the subject was leaning back against a vertical plane surface with his feet parallel as fixed by a guiding frame. The surface's normal, combined with the vector joining HJC and AJC were used to define the leg's sagittal plane. Zero-knee flexion was defined when the femur and tibia's longitudinal axis projections onto this sagittal plane were best aligned. Anterior posterior axes of the femur and the tibia were defined as lying in this sagittal plane, as perpendicular to the longitudinal axes. Medial lateral axes were finally defined by completing right-handed coordinate systems. At zero-knee flexion, femur and tibia coordinate system origins are positioned at the functional KJC [31].

To estimate tibial accelerations close to bone contact surfaces (i.e., tibial plateaus), the tibial coordinate system origin was translated distally, by a distance d, along the longitudinal axis. For each OA patient, the distance d between the KJC and tibial plateaus was calculated using weight-bearing radiography of the knee and used to translate the tibial coordinate system (Fig. 3). The mean distal translation was about 30.6 ± 4.3 mm for the OA group. For ethic considerations, no radiography was taken for the asymptomatic group; thus, the mean distal translation calculated for the OA group (i.e., 31 mm) was used.

D. Experimental Design and Data Acquisition

After the calibration process, the participant was instructed to walk on a treadmill at its self-determined comfortable speed (Table I). When a steady-state gait was reached, 25 s of gait data were collected. The same neutral sandals were used during the evaluation to avoid differences in absorption effect.

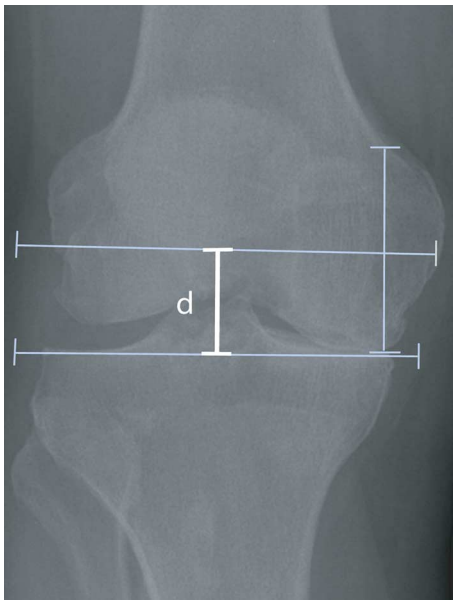

Fig. 3. Method used to estimate the distance d between the KJC and the tibial plateaus. First, two parallel lines were drawn using numerical X-rays: one passing through the tibial plateaus and another passing through the mid-point of the medial condyle. Then, the distance between both parallel lines, at the point passing through intercondylar eminence of the tibia and intercondylar fossa of the femur, was identified as the distance d and used to translate the tibia coordinate system axes along its longitudinal axis.

E. Data Processing

During the gait trial, all data were collected in a synchronised way using an external trigger device at a frequency of 120 Hz. Markers' positions were filtered with an automatic singular spectrum analysis (SSA) using a window length of 10 [32]. The SSA is an accurate nonparametric approach applied to time series analysis [32], [33]. Gait cycles events were identified using ground reaction forces collected with two Kistler forces plates integrated into the treadmill (Adal, TECMACHINE, Medical development). Kinetics data were filtered using a fourth order zero-lag Butterworth filter with a cutoff frequency of 30 Hz. The gait cycles were normalized (0%–100%) between two successive foot contacts, which correspond to instants when the magnitude of vertical ground reaction forces exceeded 2% of the participant's body weight.

F. Estimation of Linear Accelerations

Linear accelerations measured at rigid bodies were expressed on tibial and femoral coordinate systems by the mean of FP method [31]. At each instant, the gravitational component was removed from the accelerometric signal. Accelerations were then referred as external linear accelerations of the tibia $\vec{a}_{et}$ and the femur $\vec{a}_{ef}$ (Fig. 2). To evaluate the linear accelerations close to joint contact surfaces, tibial and femoral accelerations at their functional coordinate system origins were estimated. These accelerations are referred to as internal linear accelerations of the tibia $\vec{a}_{it}$ and the femur $\vec{a}_{if}$ (Fig. 2). The relation between external and virtual internal acceleration is given by (1)

$$\vec{a}_e = \vec{a}_i + (\vec{\alpha} \times \vec{r}_{i\to e}) + (\vec{\omega} \times \vec{\omega} \times \vec{r}_{i\to e}) \quad (1)$$

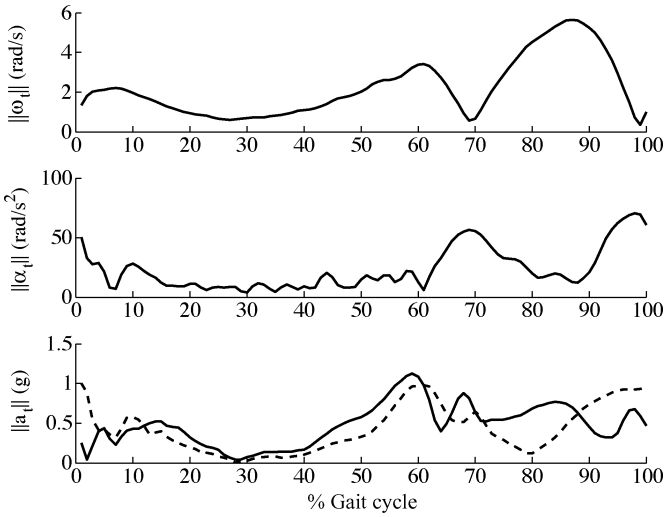

Fig. 4. Magnitude of tibial angular velocity ($\|\vec{\omega}\|$), angular acceleration ($\|\vec{\alpha}\|$), and linear accelerations ($\|\vec{a}_{it}\|, \|\vec{a}_{et}\|$). Internal acceleration (g) is in solid line whereas external acceleration (g) is in dashed line.

where $\vec{a}_e$ corresponds to the external acceleration vector, $\vec{a}_i$ to the virtual internal acceleration vector, $\vec{\alpha}$ and $\vec{\omega}$ represent the segment's angular acceleration and angular velocity vectors, respectively, $\vec{r}_{i\to e}$ represents the constant vector joining the segment's functional coordinate system origin to the accelerometer sensor origin expressed in the segment coordinate system. Positive accelerations were directed in medial, anterior, and distal directions respectively. Negative accelerations were directed in lateral, posterior, and proximal directions, respectively.

A representation of the norm of tibial angular velocity ($\|\vec{\omega}\|$), tibial angular acceleration ($\|\vec{\alpha}\|$), and tibial linear accelerations ($\|\vec{a}_{it}\|, \|\vec{a}_{et}\|$), for a typical gait evaluation, are shown in Fig. 4 to illustrate the influence of angular components in measurement of linear acceleration.

G. Data Analysis

The mean medial lateral (ML), anterior posterior (AP), and proximal distal (PD) accelerometric patterns were calculated along 15 gait cycles. Maximal (Max1, Max2), minimal (Min1, Min2), and range (R1, R2) values occurring during the loading and between preswing to mid-swing phase periods were extracted for statistical analysis (Fig. 5). Therefore, six parameters in three directions (ML, AP, and PD) were analyzed, which represent 18 parameters for each of the tibial and femoral segments (Table II). The mean knee flexion extension pattern was also calculated to ensure that this latter was comparable to literature and not affected by the knee exoskeleton. To determine the capacity of the accelerometric method to discriminate between osteoarthritic and asymptomatic knees, a one-way ANOVA using independent testing on each parameter was used. To see the impact of body weight on accelerometric parameters a one-way ANOVA was done also with subjects' body weight as covariate. A significant P value was set to 0.05.

III. Results

Fig. 6 shows mean flexion extension patterns of OA and asymptomatic groups. Maximal knee flexion of both groups

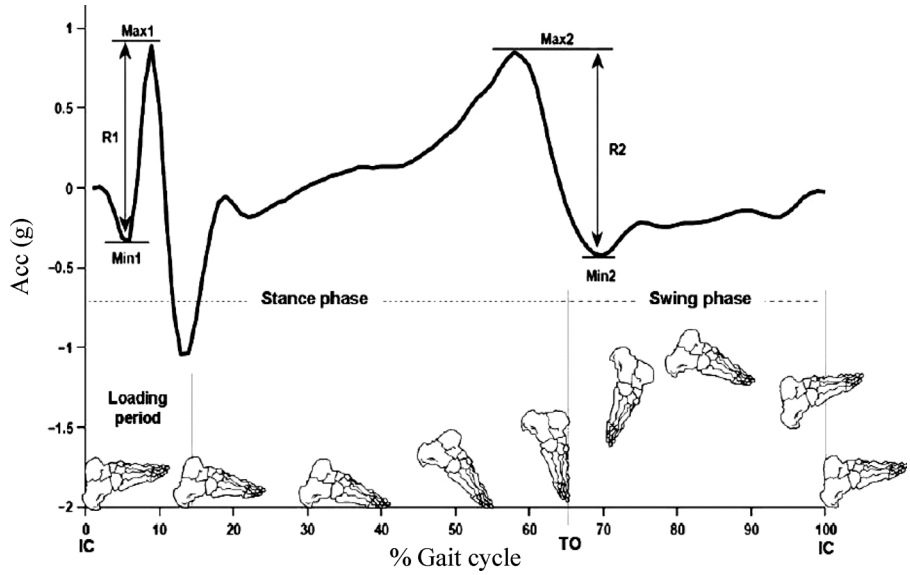

Fig. 5. Parameters extracted from a typical mean accelerometric curve. Maximal, minimal, and range values on loading phase (2%–12%) and between preswing to mid-swing phase (55%–90%) are identified. Initial contact (IC) and toe-off (TO) events are identified.

TABLE II
TIBIAL AND FEMORAL ACCELERATION PARAMETERS

Acceleration	Loading phase	Preswing to mid-swing phase
Medial lateral	Max (Max1)	Max (Max2)
(ML)	Min (Min1)	Min (Min2)
	Range (R1)	Range (R2)
Anterior posterior	Max (Max1)	Max (Max2)
(AP)	Min (Min1)	Min (Min2)
	Range (R1)	Range (R2)
Proximal distal	Max (Max1)	Max (Max2)
(PD)	Min (Min1)	Min (Min2)
	Range (R1)	Range (R2)

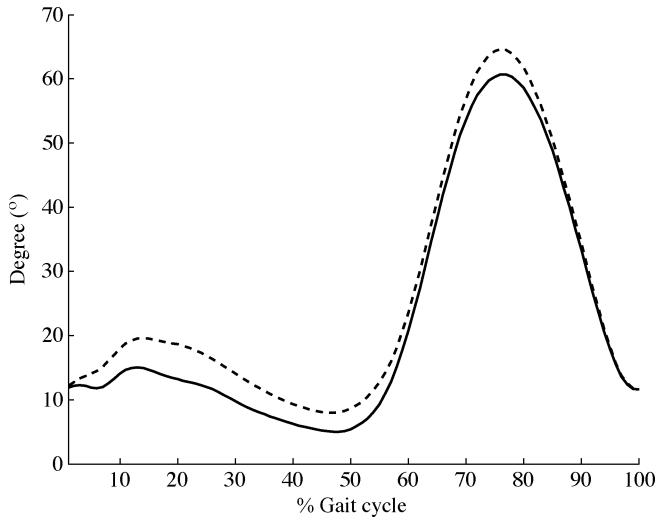

Fig. 6. Mean knee flexion extension patterns in degree (°) of OA (solid line, $\mathrm{N} = 9$) and asymptomatic (dashed line, $\mathrm{N} = 9$) groups during gait.

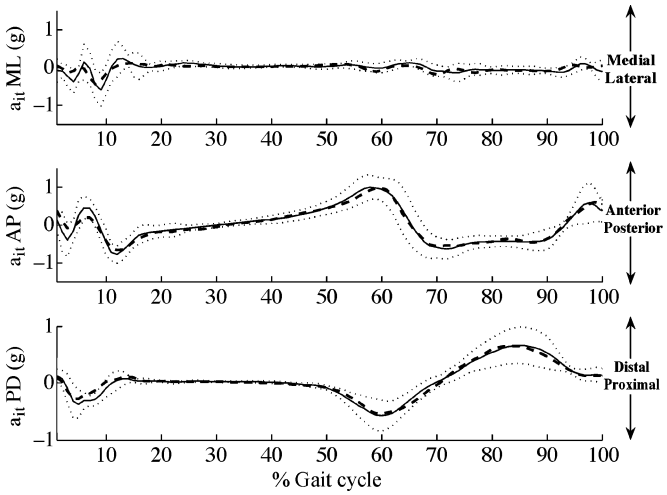

Fig. 7. Internal tibial acceleration in ML, AP, and PD directions for knee OA (mean in solid line, standard deviation in dotted line, $\mathrm{N} = 9$) and asymptomatic (mean in dashed line, $\mathrm{N} = 9$) groups.

was comparable to the results presented in Kaufmann *et al.* [7], [34] in which 139 OA patients and 20 healthy subjects were evaluated during level walking.

Figs. 7 and 8 show the mean accelerometric gait patterns in ML, AP, and PD directions from internal tibial and femoral accelerations, respectively. In both tibial and femoral accelerations, the differences between asymptomatic and OA groups were limited exclusively to the loading phase period. Eight among 36 parameters were statistically different between groups, and six were obtained in ML and AP directions.

The parameters related to the loading phase period (Max1, Min1, and R1) are summarized in Table III. For internal tibial accelerations $\vec{a}_{it}$, in ML direction, Max1 and R1 showed significant differences between groups. Max1 and R1 were greater in the OA group: 182% (0.48 g versus 0.17 g) and 88% (1.20 g versus 0.64 g), respectively. No significant difference was found in AP and PD directions.

For internal femoral acceleration $\vec{a}_{if}$, in ML direction, Min1 showed significant difference between groups and was 55% greater (0.51 g versus 0.33 g) in OA group. In AP direction, Max1, Min1, and R1 parameters showed significant differences

TABLE III
TIBIAL AND FEMORAL ACCELERATION PARAMETERS RELATED TO THE LOADING PHASE PERIOD

	Max1 (g)		Min1 (g)		R1 (g)	
	Asymptomatic	Osteoarthritis	Asymptomatic	Osteoarthritis	Asymptomatic	Osteoarthritis
Tibia						
ML	0.17 (0.06)	0.48 (0.37) *	- 0.48 (0.18)	- 0.71 (0.35)	0.64 (0.20)	1.20 (0.68) *
AP	0.46 (0.19)	0.55 (0.26)	- 0.72 (0.44)	- 0.90 (0.22)	1.17 (0.49)	1.45 (0.32)
PD	0.12 (0.09)	0.15 (0.15)	- 0.40 (0.20)	- 0.49 (0.17)	0.52 (0.25)	0.63 (0.24)
Femur						
ML	0.18 (0.11)	0.20 (0.17)	- 0.33 (0.07)	- 0.51 (0.21) *	0.51 (0.17)	0.72 (0.33)
AP	0.38 (0.21)	1.00 (0.27) ***§	- 0.65 (0.26)	- 1.03 (0.37) * §	1.03 (0.31)	2.03 (0.53) ***
PD	0.00 (0.11)	0.08 (0.08)	- 0.32 (0.14)	- 0.47 (0.13) *	0.32 (0.13)	0.55 (0.16) **§

Significant P value $< 0.05(^{*}), < 0.01(^{**}), < 0.001(^{***})$.
Parameters that are still significant when the weight was included as covariate (§).

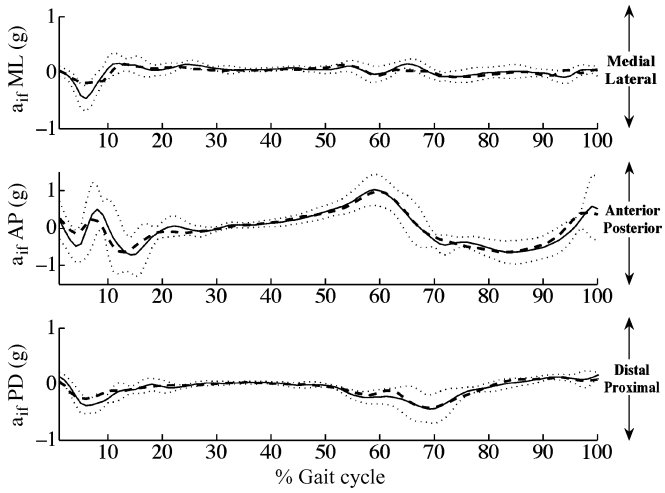

Fig. 8. Internal femoral acceleration in ML, AP, and PD directions for knee OA (mean in solid line, standard deviation in dotted line, $\mathrm{N} = 9$) and asymptomatic (mean in dashed line, $\mathrm{N} = 9$) groups.

between groups and were greater in OA group: 163% (1.00 g versus 0.38 g), 58% (−1.03 g versus −0.65 g), and 97% (2.03 g versus 1.03 g), respectively. In PD direction, a significant increase of 47% was noted for Min1 (−0.47 g versus −0.32 g) and of 72% for R1 (0.55 g versus 0.32 g) both in OA group.

External tibial $\vec{a}_{et}$ and femoral $\vec{a}_{ef}$ accelerations have also been analyzed for both groups. Only two parameters (Max1 and R1) showed significant differences for external tibial acceleration in AP direction. The latter were greater for OA group: Max1 was 260% greater with 0.36 g versus 0.10 g, and R1 was 49% greater with 1.28 g versus 0.86 g. No significant difference was found for the external femoral acceleration.

When analyses were done including participants' weight as covariate, internal accelerations showed three significant parameters (Table III) whereas external accelerations were no more significant between groups.

IV. DISCUSSION

A new accelerometric method has been developed to estimate 3-D internal tibial and femoral linear accelerations during gait. This method counteracts the major limitations of traditional accelerometric methods (i.e., sensor fixation, sensor location) by the use of an exoskeleton and the calculation of the acceleration close to knee-joint contact surfaces. The impact of wearing the exoskeleton on knee pain and OA gait patterns has been previously evaluated and revealed that its use does not increase pain and have no effect on spatiotemporal parameters [29]. Nevertheless, its impact on the thigh and shank segments kinematics during gait has never been evaluated.

The objective of this method was to explore accelerometric parameters that have the potential to be different between osteoarthritic and asymptomatic knee during gait. The new method shows a good discriminative capacity of eight parameters that exhibit significant difference between OA and asymptomatic group. When using external accelerations, only two parameters were found significantly different. Moreover, it is interesting to note that these two parameters were different from those found in internal acceleration patterns. This confirms the authors' concerns about the lack of standardization in the location of accelerometers and can explain variability on the results found in previous studies. As noted before, Henriksen *et al.* [16] found no difference in longitudinal tibial peak acceleration between asymptomatic and osteoarthritic knee. The authors [16] suggested that the lack of difference between groups could be because of the presence of pain in the OA group. In the present study, although all patients had a painful knee during gait evaluation, a significant difference was found between groups in eight parameters. Interestingly, the main differences were not obtained in PD direction but rather in ML and AP directions. This was in agreement with the study of Lafortune [7], who found high tibial acceleration in ML (0.90 g) and AP (1.26 g) directions in a healthy subject during a treadmill gait evaluation. The author [7] recommended ML and AP accelerations for evaluation of tibial loading be taken into consideration. The results obtained from this study in ML and AP directions were lower in magnitude from those reported by Lafortune [7]. The difference between results (i.e., [7] versus the present study) were possibly because of several aspects: gait velocity (1.5 m/s versus 0.9 m/s), population characteristics (young versus elderly), sensor fixation (bone-mounted sensor versus exoskeleton), and location of the accelerometer (3-cm below the tibial plateaus versus tibial plateaus).

We believe that the high accelerations estimated in ML and AP directions could be a consequence of both varus lower limb alignment and joint instability present in medial knee OA. Ogata *et al.* [14] reported an increase in ML acceleration in

knee OA during gait. The authors [14] defined the lateral acceleration generated at initial foot contact as acceleration caused by the varus deformity noted in medial knee of osteoarthritic patients. Their results are in agreement with the present study.

When compared to previous studies, the linear accelerations calculated in this study were estimated at the same functional location for each participant (i.e., tibial plateaus and knee joint center) instead of at an arbitrary location on the segment [13]–[16]. By transposing accelerations close to joint contact surfaces we believe that the estimation of linear accelerations is less affected by angular components induced by the movement of segments during gait. However, the estimation of internal acceleration could be affected by the vector joining sensor to bone. Hence, to verify the sensitivity of the method, we introduced a variation of 1%–10% in the $\vec{r}_{i\rightarrow e}$ vectors. The results show a difference in 3-D acceleration peak magnitude less than 3.6% with a $\vec{r}_{i\rightarrow e}$ variation of 5% and less than 8% with a $\vec{r}_{i\rightarrow e}$ variation of 10%. No modification in 3-D accelerometric patterns has been observed.

The present study is also the first to consider the anatomical aspect of OA patients by the use of both a functional calibration method [31] and 2-D weight-bearing knee radiography. Nevertheless, the use of 2-D knee radiography to translate 3-D tibial coordinate system origin from the knee joint centre to tibial plateaus may have induced some misplacement. Moreover, the use of the mean distal translation of the OA group to translate the tibial coordinate system of the AS group may have under estimate differences found in this study. Hence, we believe that the definition of tibial coordinate system origin for both groups could be improved by the use of a 3-D imaging technique [35].

The results presented in this study were also affected by the difference in weight between both groups. Since obesity is one of the main factors associated to knee OA, it was difficult to have an equivalent weight between asymptomatic and OA groups. However, even when statistical analyses were done including weight as covariate, linear internal accelerations continued to show significant differences between both groups.

V. Conclusion

Although the proposed method still has some limitation, it is very promising in providing new parameters that could be used in the comprehension of knee instability (medial lateral and anterior posterior accelerations) and transmission of shock (proximal distal acceleration) between tibia and femur during gait in knee OA. The accelerometric parameters that are identified here as significantly different from an asymptomatic group have a great potential to be used as follow-up parameters for patients having knee OA in a rehabilitation context.

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