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Soft tissue artifact compensation in knee kinematics by multi-body optimization: Performance of subject-specific knee joint models

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ABSTRACT

Soft tissue artifact (STA) distort marker-based knee kinematics measures and make them difficult to use in clinical practice. None of the current methods designed to compensate for STA is suitable, but multi-body optimization (MBO) has demonstrated encouraging results and can be improved. The goal of this study was to develop and validate the performance of knee joint models, with anatomical and subject-specific kinematic constraints, used in MBO to reduce STA errors. Twenty subjects were recruited: 10 healthy and 10 osteoarthritis (OA) subjects. Subject-specific knee joint models were evaluated by comparing dynamic knee kinematics recorded by a motion capture system (KneeKG™) and optimized with MBO to quasi-static knee kinematics measured by a low-dose, upright, biplanar radiographic imaging system (EOS®). Errors due to STA ranged from 1.6° to 22.4° for knee rotations and from 0.8 mm to 14.9 mm for knee displacements in healthy and OA subjects. Subject-specific knee joint models were most effective in compensating for STA in terms of abduction–adduction, inter–external rotation and antero–posterior displacement. Root mean square errors with subject-specific knee joint models ranged from $2.2 \pm 1.2°$ to $6.0 \pm 3.9°$ for knee rotations and from 2.4 ± 1.1 mm to 4.3 ± 2.4 mm for knee displacements in healthy and OA subjects, respectively. Our study shows that MBO can be improved with subject-specific knee joint models, and that the quality of the motion capture calibration is critical. Future investigations should focus on more refined knee joint models to reproduce specific OA knee geometry and physiology.

1. Introduction

Stereophotogrammetry is the most widespread technique used to analyze the 3D kinematics of knee joint (Cappozzo et al., 2005). This technique consists of measuring the trajectories of markers glued on skin of the lower limb, or mounted on special instruments fastened to the thigh and shank, and deducing the movements of tibial and femoral bones. Although stereophotogrammetry assesses knee movements in a simple and non-invasive way, the resulting 3D kinematics remains highly inaccurate. Indeed, the markers move relative to the underlying bones because of inertial effects, skin deformations and muscle contractions (Cappozzo et al., 1996), which can lead to kinematic errors exceeding 20° and 30 mm for knee rotations and displacements respectively (Peters et al., 2010). These errors, known as soft tissue artifact (STA), still remain the main limitation of stereophotogrammetry in clinical practice (Stagni et al., 2009).

Many mathematical methods designed to compensate for STA have been developed in the past two decades, e.g., dynamic calibration (Lucchetti et al., 1998), multi-body optimization (MBO) (Lu and O'Connor, 1999), point cluster technique (Alexander and Andriacchi, 2001) and double anatomical landmark calibration (Cappello et al., 2005). To date, however, none of these methods is universally accepted by the scientific community, since they cannot accurately estimate 3D knee kinematics (Leardini et al., 2005) or are too restrictive for clinical application (Cappello et al., 2005).

MBO optimizes the position and orientation of a lower limb model by minimizing, under kinematic constraints, the sum of squared differences between measured and model-predicted marker coordinates. MBO is easy to implement and requires only one calibration procedure. Moreover, although simple kinematic

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constraints (e.g. hinge or spherical joint) showed limitations in reducing errors due to STA (Andersen et al., 2010; Clément et al., 2014b; Li et al., 2012; Stagni et al., 2009), recently proposed anatomical constraints provided encouraging results (Duprey et al., 2010; Gasparutto et al., 2015). Duprey et al. (2010) compared the performance of 8 sets of kinematic constraints used in MBO to compensate for STA. Results showed that model-based knee kinematics greatly depended on the set of kinematic constraints imposed by the lower limb model, and that anatomical constraints were the only to offer physiological knee kinematics, i.e. limited abduction–adduction and femoral rollback, according to the literature (Reinschmidt, 1996). These anatomical constraints were defined by modeling the knee with a parallel mechanism composed of 2 sphere-on-plane contacts and 3 isometric ligaments: anterior cruciate ligament (ACL), posterior cruciate ligament (PCL) and medial collateral ligament (MCL). Gasparutto et al. (2015) later improved this knee modeling by introducing 4 deformable ligaments with minimal length variation: ACL, PCL, MCL and lateral collateral ligament (MCL). This study concluded that anatomical constraints improved the compensation for STA using MBO, compared to no kinematic constraint or degree-of-freedom coupling curves (Walker et al., 1988). However, the knee kinematic errors obtained with anatomical constraints, below 2.5° and 4.1 mm, were similar to those obtained with spherical joint constraints when compared to kinematics measured with intracortical pins (Gasparutto et al., 2015).

It is important to note that the geometry of the parallel mechanisms used in the two previous studies derived from in vitro measurements (Feikes et al. (2003) for Duprey et al. (2010) and Parenti-Castelli and Sancisi (2013) for Gasparutto et al. (2015)) and was not personalized to the geometry of the subjects' knees. Our recent works conducted to validate the 8 sets of kinematic constraints used in Duprey et al. (2010) showed that none of these unpersonalized models seemed perfect for STA compensation, with knee kinematic errors around 5.3–11.2° and 1.4–6.2 mm (Clément et al., 2014b). Furthermore, the model-based kinematics showed reduced inter-subject variability and a loss of specific movements on pathological subjects (Clément et al., 2014b). It has therefore been suggested that subject-specific knee models could improve MBO, as stated in previous studies (Duprey et al., 2010; Gasparutto et al., 2015).

To our knowledge, very few studies have used personalized knee model in order to improve gait analysis (Scheys et al., 2011; Zheng et al., 2014). Moreover, MBO has been validated in patients with ligament deficiency (Li et al., 2012) or total knee arthroplasty (Stagni et al., 2009) but never in patients with knee OA. Yet it would be necessary to develop a fairly-accurate motion capture method to analyze and better understand this extremely prevalent disease in a clinical context (Hunter, 2009; Zhang et al., 2008). The aim of the present study was therefore to develop and validate the performance of knee joint models, with anatomical and subject-specific kinematic constraints, used in MBO to correct kinematic data recorded during weight-bearing squatting activities carried out by healthy and OA subjects.

2. Materials and methods

2.1. Subjects

Twenty subjects volunteered to participate and gave informed consent after study approval by the CRCHUM and ÉTS ethics committees: 10 healthy subjects (age 54.9 ± 9.3 years, height 166.7 ± 9.4 cm, weight 70.9 ± 13.0 kg) and 10 OA subjects (age 60.8 ± 5.9 years, height 161.4 ± 7.7 cm, weight 85.7 ± 9.9 kg). All subjects, aged between 38 and 70 years, had no neurological, heart or balance problems. The healthy subjects had no previous knee injury or any evidence of knee attrition, while OA subjects were waiting for total knee replacement surgery.

2.2. Experimental protocol

All of our protocol is summarized in Fig. 1. Each subject performed dynamic squatting movements (0–60–0° of knee flexion) standardized with a positioning jig. As described in Clément et al. (2014a), this positioning jig was composed of feet wedges that maintained the feet in neutral internal–external rotation, and a proprioceptive reminder (PR) that indicated to the subjects when they reached 60° of knee flexion (Fig. 1A). The PR was adjusted in height according to the length of the tibia and femur (Clément et al., 2014a). 3D knee kinematics during the dynamic squatting movements was recorded by a motion capture device designed to limit STA: the KneeKG™ (Emovi Inc., Laval, QC, Canada). Its markers were placed on rigid devices attached onto the iliac crest, the femoral condyles and the tibial crest, and were measured by a Polaris Spectra® camera (60-Hz, NDI, Waterloo, ON, Canada) (Fig. 1A). The KneeKG™ provides repeatable (0.4–0.8° and 0.8–2.2 mm) and reliable (intra class coefficient of 0.88–0.94) measurements (Lustig et al., 2012), but they are still influenced by STA. Kinematic errors can reach 7° and 11 mm for knee bone positions and orientations, but remain between 0.6° and 1.4° in the sagittal plane (Südhoff et al., 2007). The KneeKG™ was calibrated (i.e., the anatomical segment axes were defined with respect to markers placed on the rigid devices) using the functional approach described by Hagemeister et al. (2005), or using medical imaging as detailed below.

Each subject then performed a quasi-static squatting movement consisting of 5 positions of knee flexion (0°, 30°, 40°, 50° and 60°). This movement was also standardized with the positioning jig by adjusting the height of the PR for the 5 positions. 3D knee kinematics during the quasi-static squatting movement was recorded with the EOS® system (EOS Imaging Inc., Paris, France) (Fig. 1C). This upright biplane radiographic imaging system allows 3D bone reconstruction with an accuracy about 2 mm (Chaibi et al., 2012) and radiation doses 800–1000 times lower than that of a CT-scan (Deschênes et al., 2010). Each position was maintained for 5 s, which correspond to the acquisition time of EOS®. Subjects were instructed not to sit on the PR and to keep their trunk straight during the dynamic and quasi-static squatting movements.

2.3. Multi-body optimization: dynamic knee kinematics (see Appendix for more details)

The lower limb models used in MBO were defined using the generalized coordinates $\mathbf{Q}_i$ introduced by Dumas and Chèze (2007) (Fig. 1B). The models consisted of 4 segments (pelvis, femur, tibia and foot) and imposed the following kinematic constraints at the ankle, knee and hip joints: NNN, SSS, SPS (N, S, and P stand for no kinematic constraint, spherical joint constraints, and parallel mechanism constraints). Different levels of personalization were tested for the knee joint models. The generalized coordinates $\mathbf{Q}_2$ and $\mathbf{Q}_3$ used as initial guess to impose the constraints N, S, and P at the knee were defined either from the functional calibration of the KneeKG™ (Hagemeister et al., 2005) or from a calibration using the subject-specific knee bone models (Fig. 1F). Similarly, the geometry of the parallel mechanism (the size and center of spheres modelling the femoral condyles, the normal and point of planes modelling the tibial plateaus, and the origin and insertion of the 4 ligaments) was defined either from the literature (Parenti-Castelli and Sancisi, 2013) as in Gasparutto et al. (2015) or from the subject-specific knee bone models (Fig. 1F).

A total of 7 lower limb models, with and without subject-specific knee joint model, were evaluated in this study (Table 1). The initial guess used in MBO was the $\mathbf{Q}_i$ computed from KneeKG™ markers at each frame of dynamic squatting movements. Model-based dynamic knee kinematics was directly deduced from $\mathbf{Q}_2$ and $\mathbf{Q}_3$ (Dumas et al., 2012; Duprey et al., 2010). Knee rotations were calculated from motion of the tibia relative to the femur according to Wu et al. (2002), and knee displacements were defined as non-orthonormal projection (Desroches et al., 2010) of the vector connecting the distal endpoint of the femur (D_3) and the proximal endpoint of tibia (P_2) on the knee joint coordinate system (see Appendix).

2.4. Image processing: quasi-static knee kinematics

Subject-specific bone models and quasi-static knee kinematics were computed from the 5 biplane radiographs recorded with EOS® (Fig. 1D). As detailed in Kanhonou et al. (2014), radiographs taken at 0° of knee flexion served to create subject-specific bone models by deforming generic models until their projected silhouettes were adjusted to the contours of the study subjects' bones (Chaibi et al., 2012). Thereafter, the bones were segmented on the 4 remaining biplane radiographs, and subject-specific knee bone models were positioned and oriented by a rigid 2D/3D registration method and an iterative closest point-based algorithm (Fig. 1E). The accuracy of the positions and orientations of the femur and tibia is less than $0.3 \pm 0.3°$ and 0.3 ± 0.2 mm respectively, while the precision is less than $0.3 \pm 0.3°$ and 0.3 ± 0.2 mm respectively (Kanhonou et al., 2014). Quasi-static knee kinematics was finally calculated with the generalized coordinates $\mathbf{Q}_2$ and $\mathbf{Q}_3$ determined from the subject-specific bone models and constituted the gold standard of this study.