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Image based weighted center of proximity versus directly measured knee contact location during simulated gait

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ABSTRACT

To understand the mechanical consequences of knee injury requires a detailed analysis of the effect of that injury on joint contact mechanics during activities of daily living. Three-dimensional (3D) knee joint geometric models have been combined with knee joint kinematics to dynamically estimate the location of joint contact during physiological activities—using a weighted center of proximity (WCoP) method. However, the relationship between the estimated WCoP and the actual location of contact has not been defined. The objective of this study was to assess the relationship between knee joint contact location as estimated using the image-based WCoP method, and a directly measured weighted center of contact (WCoC) method during simulated walking. To achieve this goal, we created knee specific models of six human cadaveric knees from magnetic resonance imaging. All knees were then subjected to physiological loads on a knee simulator intended to mimic gait. Knee joint motion was captured using a motion capture system. Knee joint contact stresses were synchronously recorded using a thin electronic sensor throughout gait, and used to compute WCoC for the medial and lateral plateaus of each knee. WCoP was calculated by combining knee kinematics with the MRI-based knee specific model. Both metrics were compared throughout gait using linear regression. The anteroposterior (AP) location of WCoP was significantly correlated with that of WCoC on both tibial plateaus in all specimens ($p<0.01$, 95% confidence interval of Pearson's coefficient $r>0$), but the correlation was not significant in the mediolateral (ML) direction for 4/6 knees ($p>0.05$). Our study demonstrates that while the location of joint contact obtained from 3D knee joint contact model, using the WCoP method, is significantly correlated with the location of actual contact stresses in the AP direction, that relationship is less certain in the ML direction.

1. Introduction

It has been postulated that alterations in knee kinematics lead to an increased risk for the development of posttraumatic osteoarthritis (Andriacchi and Mundermann, 2006; Chaudhari et al., 2008). Thus, to understand the mechanical consequences of knee injury requires a detailed analysis of the effect of that injury on joint contact mechanics during activities of daily living. While gait analysis provides information about the tibiofemoral motion (Andriacchi and Alexander, 2000), it was not until the advent of image-based knee specific models that the location of contact between the tibial and femoral articular surfaces could be assessed.

Several knee joint contact models have been developed for quantifying dynamic joint contact location based on three-dimensional (3D) bone geometry (Anderst and Tashman, 2003; Asano et al., 2001; Li et al., 2004), and cartilage geometry (DeFrate et al., 2004; Li et al., 2005). Briefly, Asano et al. (2001), Li et al. (2004) estimated the tibiofemoral contact location by finding the shortest distance between the tibia and femur in the superior–inferior direction (so called *shortest distance method*). Subsequently, DeFrate et al. reported that during the weight-bearing phase of a forward lunge, the shortest distance method overestimated the translation of the contact point. So, they developed a *cartilage-overlap method*, in which the location of contact was determined as the geometric centroid of the cartilage overlapping area (DeFrate et al., 2004). Instead of depending solely on a single measurement of the shortest distance, Anderst and Tashman proposed a method of using *weighted center of proximity* between the tibial and femoral bone surfaces to define the contact point. In this method, vertices on the tibial plateau with shorter

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tibia-femur distances were assigned higher weights and therefore considered more important for determining the location of contact (Anderst and Tashman, 2003). Beveridge et al. (2013b) found that the *weighted center of proximity* can detect subtle changes in tibiofemoral contact resulting from combined ligament transection, and a connection between the altered tibiofemoral contact and the extent of cartilage degeneration at the site of contact has also emerged (Anderst and Tashman, 2009; Beveridge et al., 2013a). Despite this connection, the relationship between the estimated location of contact using the *weighted center of proximity* method and the actual location of contact experienced by the knee during daily activities has not been quantified.

The purpose of this study was to assess the relationship between the tibiofemoral contact location as estimated using the *image-based weighted center of proximity* and a *directly measured weighted center of contact* for the human knee during the stance phase of simulated walking. Our hypothesis is that the location of contact as quantified using both methods on each plateau would be significantly correlated throughout the stance phase of gait.

2. Material and methods

2.1. Overview

To test the hypothesis, we created knee specific models for six cadaveric knees, which were then subjected to physiological loads intended to mimic gait. The weighted center of contact was directly measured throughout gait using a thin electronic sensor placed on the tibial plateau as reported in our previous studies (Gilbert et al., 2013; Wang et al., 2014). Marker-based kinematic analysis of the physical experiment was used as input to the knee-specific in silica models to enable the weighted center of proximity to be computed.

2.2. Magnetic resonance imaging

Six human cadaveric knees with no history of surgery or trauma were acquired and stored at −20 °C (Anatomy Gifts Registry), the demographics of which are shown in Table 1. The knees were thawed for 12 h at room temperature and were then scanned using Magnetic Resonance Imaging (MRI). All scanning was performed on a clinical 3 T scanner (GE Healthcare, Waukesha, WI) using an 8 channel phased array knee coil (Invivo, Gainesville, FL). A 3D CUBE (Gold et al., 2007) series was acquired to generate an image dataset for segmentation of the menisci: echo time (TE)=31 ms, repetition time (TR)=2500 ms, echo train length=35–40, receiver bandwidth (RBW)= ±41.7 kHz, number of excitations (NEX)=0.5 with voxel dimensions: $0.3 \times 0.3 \times 0.6$ mm^3. A 3D SPGR with frequency selective fat suppression image series was acquired to segment cartilage and osseous geometries: TE=3.2 ms, TR=13.9 ms, RBW= ±41.7 kHz, NEX=2, voxel dimensions=$0.3 \times 0.3 \times 0.7$ mm^3. Images were manually segmented using custom software (Fig. 1a). Note: the articular cartilage surfaces were extracted so that the knee model could be appropriately aligned with the physical digitization of the articular surfaces (see *Cadaveric Model and Physical Experiments* section).

Table 1
Demographics of the knee joint donors.

Specimen	Side	Gender	Age	Weight (kg)	Height (m)
1	R	Female	39	64	1.68
2	L	Male	53	91	1.73
3	R	Female	56	104	1.7
4	R	Male	58	90	1.78
5	R	Female	62	41	1.63
6	L	Female	27	59	1.57
Mean ± SD			49.2 ± 13.4	74.8 ± 23.9	1.68 ± 0.07

2.3. Cadaveric model and physical experiments

After stripping the surrounding soft tissue (fat, musculature), the specimens were fixed to a modified load-controlled Stanmore Knee Simulator (University College London, Middlesex, UK) (Fig. 2a) (Bedi et al., 2010; Gilbert et al., 2013; Wang et al., 2014). The normal contact stresses across the tibial plateaus were measured using a thin electronic sensor (4010N, Tekscan Inc., MA) which is a matrix of 23 by 34 sensing elements (Fig. 2b). The sensor was inserted underneath the meniscus and the placement was adjusted to capture loads across the entire plateau under a static 1000 N axial load. The sensor was fixed to the tibial plateau by suturing the augment tabs to the tibial insertion of the anterior cruciate ligament (ACL) and the posteroinferior capsule, as detailed in our previous studies (Gilbert et al., 2013; Wang et al., 2014). The knee was then driven under prescribed femur flexion/extension angles and cyclic multidirectional loads (period=2 s) including axial force, anterior/posterior force and internal/external torque applied on the tibia to mimic the activity of gait (ISO 14243-1) (Bedi et al., 2013; Gilbert et al., 2013). The medial/lateral translation and varus/valgus rotation were unconstrained. Reflective markers were rigidly attached to the femoral and tibial fixtures (Fig. 2a).

Anatomical bony landmarks and fiducial landmarks were digitized at neutral position (0° flexion and under an axial force of 1000 N) using a digitizing pointer (Fig. 1b). Specifically, the medial and lateral femoral epicondyles, medial and lateral edges of tibial plateau and tibial spine were identified. Reference frames of the femur and the tibia were defined based on the bony landmarks to describe the motion of the femur relative to the tibia (Wang et al., 2012; Wang and Zheng, 2010). Meanwhile, the fiducial landmarks rigidly attached to the bones were also traced using the digitizing pointer. For each knee, we also digitized a set of point clouds from the following surface areas: the anterior portion of femoral articular cartilage, central region of tibial articular cartilage, and medial/lateral portion of the femur and tibia (Fig. 1b). The surfaces were digitized using a commercial 3D digitizer (accuracy: 0.23 mm) (MicroScribe, Immersion, San Jose, California). Again, the fiducial landmarks were traced using the MicroScribe digitizer, which would be used to transform the digitized point clouds into the Laboratory coordinate frame of the motion capture system. After collecting all the data at neutral position, the motion data (MotionAnalysis Inc., CA) and the contact stress data (Tekscan Inc., MA) were synchronously recorded at 50 Hz and 100 Hz respectively during simulated walking. Data from twenty continuous gait cycles were collected and the average of the last three cycles was used for analysis to ensure the electronic sensor and the knee simulator reached steady-state (Cottrell et al., 2008).

2.4. Aligning the physical and in silica models

To align the MRI models with the physical location of the femur and tibia on the knee simulator, first, the digitized joint surfaces (point clouds) were reconstructed at the neutral position in the Laboratory coordinate frame of the motion capture system by coordinate transformation based on the fiducial landmarks using a custom program (MATLAB R2012b, MathWorks, Natick, MA). The MRI models were then aligned to the point clouds which included a minimum of 6000 points using an iterative closet-point (ICP) technique (Abebe et al., 2009; Besl and McKay, 1992; Defrate et al., 2006) (Fig. 1b). After alignment, the root mean square errors were calculated by measuring the minimum distances from the point clouds to the respective MRI model surfaces.

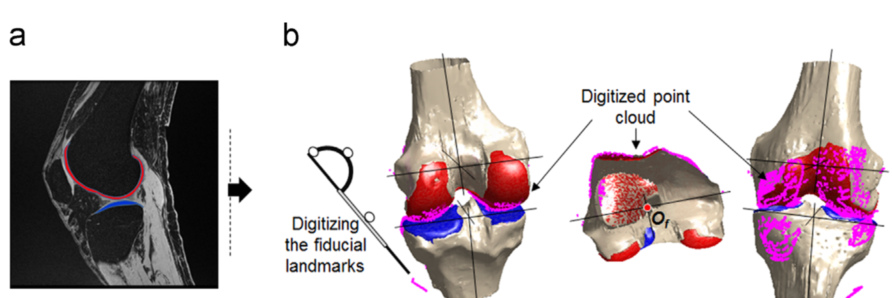

Fig. 1. (a) Segmentation of bone and cartilage from knee joint sagittal MR images. (b) The reconstructed 3D knee joint models were aligned with the digitized point clouds using an iterative closet-point (ICP) technique. Note: O_f represents the mid-point of the distance bewteen the femoral eipcondyles.

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