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Contact forces in several TKA designs during squatting: A numerical sensitivity analysis

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ABSTRACT

Total knee arthroplasty (TKA) is a very successful procedure, but pain or difficulties during activities still persist in patients. Patient outcomes in TKA surgery can be affected by implant design, alignment or patient-related anatomical factors.

This paper presents a numerical sensitivity analysis of several TKA types: a fixed bearing, posterior stabilized prosthesis, a high flexion fixed bearing guided motion prosthesis, a mobile bearing prosthesis and a hinge prosthesis. Each prosthesis was virtually implanted on the same cadaver leg model and it underwent a loaded squat, in 10 s, between 0° and 120°, similar to several previous experimental tests performed on knee kinematics simulators. The aim of this examination was to investigate the sensitivity of the patello-femoral (PF) and tibio-femoral (TF) contact forces to patient-related anatomical factors, and component position in the different implant types. The following parameters were used for the sensitivity study: the proximo-distal patellar position, the patellar component tilting, the tibial component position and orientation, the locations of the medial and lateral collateral ligaments with respect to femur and tibia and the patellar tendon length.

The sensitivity analysis showed that PF contact forces are mostly affected by patella height (increases up to 67% for one TKA type in patella-alta configuration), by an anterior tibial component translation (increases up to 30%), and by patellar component tilting (increases up to 29%); TF contact forces are mostly affected by the anterior displacement of the insertion points of the medial collateral ligament with respect to the reference position (increases up to 48%).

1. Introduction

Total knee arthroplasty (TKA) is a highly successful and reproducible treatment for knee patients. Over 500,000 procedures are performed annually in the US, and that number is expected to increase exponentially over the next two decades (Rosen et al., 2002). Although the success of TKA is well documented, difficulties or pain during motion still persist in a limited number of patients. This might be explained by surgical errors or by excessive deviations from the standard knee anatomy which can lead to a different biomechanical behavior than what the prosthesis was designed for (Lewis et al., 1994; Wasielewski et al., 1994; Figgie et al., 1989). Most TKAs function as surface replacements within the soft tissue envelope that surrounds the knee. Consequently, positioning and sizing of the components will largely affect the post-operative result. Any misplacement or wrong sizing will affect loads on the interface and tension in the ligaments (Victor, 2009). This will lead to aberrant knee mechanics inducing stiffness, instability and early loosening (Tew and Waugh, 1985, Hsu et al., 1989, Ritter et al., 1994, Berger et al., 1998, Akagi et al., 1999, Matsuda et al., 2001, Green et al., 2002).

Also patellar position could change patellofemoral and tibiofemoral load distributions (Yamaguchi and Zajac, 1989, Hirokawa, 1991, Singerman et al., 1994, Luyckx et al., 2009).

But even if the implant components are perfectly positioned and sized, the kinematics and kinetics of the replaced joint could be different from the native knee. While most designs are developed with a reference anatomy in mind, specific patient anatomy will of course deviate from this reference. Thus, the interaction of the bony anatomy and the soft tissue morphology with the knee prosthesis during function might again lead to non-physiological loads and kinematics.

In any case, the resulting non-physiological load conditions and kinematics of the knee, might ultimately lead to pain, bone

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remodeling, high wear rates and, sometimes, implant failure or revision.

However, how implant mal-positioning or how deformity alters the patient's TKA output in terms of contact forces is not well documented. Our current research models have indeed been inadequate in providing us sufficient insight into these fundamental issues relevant in TKA.

Neither clinical nor in-vitro cadaver tests allow us the versatility to test the effect of these parameters in a systematic way. Recently, however, computational models replicating knee function have been developed and validated, which could serve this purpose (Innocenti et al., 2009b). Such analytical methods allow researchers to change certain parameters of potential influence, and investigate their effect under standardized conditions simulating knee function, and this in a non-destructive and repeatable manner.

In literature, several studies use computational models to investigate TKA contact mechanics. FEA or multibody dynamics were used to analyze polyethylene stress, tibio-femoral forces and contact area during walking (Morra and Greenwald, 2003, Bei et al., 2004, Soncini et al., 2004, Godest et al., 2002) or in high flexion activities (Morra and Greenwald, 2005). Sensitivity studies were also performed, using computational models, to investigate the effect of mal-positioning on knee and TKA performance (Shelbourne et al., 2010, Besier et al., 2008, Yao et al., 2006, Bendjaballah et al., 1997). However, these studies are mainly focused on only one TKA type or they investigate only the effect of one mal-position configuration, mainly during walking.

The aim of this work is to estimate and compare the contact forces in four different, commonly used TKA types during a loaded deep squat simulating surgical errors and patient-related anatomical factors. Although each prosthesis type is represented by a specific design, the purpose of this study is not to analyze the behavior of those specific TKA designs but rather to determine, in general, how surgical errors or anatomical factors can alter the PF and TF contact forces, for each type, compared to its own reference configuration.

2. Materials and methods

2.1. Physiological knee model

A CT scan of a cadaveric full leg of a Caucasian male (age=82 years, height=1.88 m and weight=72.6 kg) was made. CT image processing software, (Mimics 12.0, Materialise, Leuven, BE), was used to generate 3D models of all bones (Fig. 1).

The physiological knee model was built assuming physiological positions of the main soft tissue insertions described in literature (Victor et al., 2009a; LaPrade et al., 2007, 2003; Netter, 2006). Each insertion of ligaments and tendons is a single point (Fig. 1), due to limitations in the simulation software.

2.2. Replaced knee models

Four different TKA types were chosen for this sensitivity analysis (Fig. 2): a fixed bearing, posterior stabilized design, a high flexion fixed bearing guided motion design, a mobile bearing design and a hinge design. All prostheses were of the same size and replaced both cruciate ligaments and all resurfaced the patella. However, the contact geometries are different for different designs. In this study, we did not consider any cruciate retaining (CR) implants because adding this TKA type would add another level of complexity to the study. It is well known that CR implants kinematics and kinetics are governed to a large extent by correct PCL-balancing (Ritter et al., 1988; Pagnano et al., 1998) and this parameter is not considered in this study.

Following the surgical procedure of each TKA, the proper surgical cuts on the bone model were identified and performed. Each TKA was virtually implanted according to the cut bone geometries, thus defining the reference replaced knee model. Several derivative replaced knee models were then obtained by changing the values of one parameter of the reference model in a range which was based on literature and surgical experience (Hungerford et al., 1984; Eckhoff et al., 2003, 2001; Grelsamer et al., 2008). The derivative configurations are listed in Table 1.

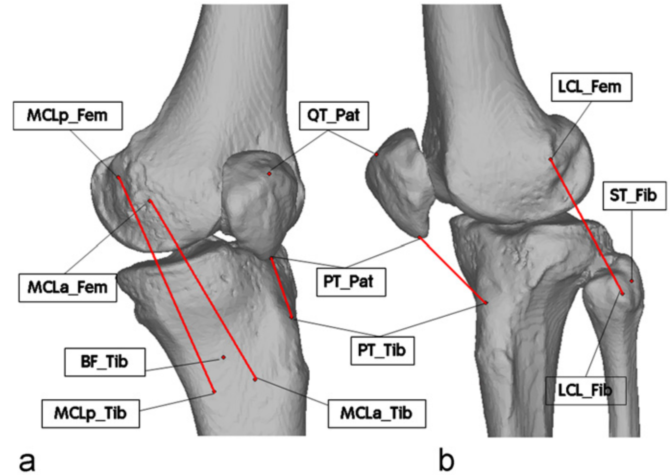

Fig. 1. Bone models and position of the soft tissues insertion points considered in this study (a—medial view and b—lateral view). Tibia and fibula were considered as a single rigid body; the centroid of the insertion area of the ligaments (shown as red line) was used as the best approximation of the position of the respective insertion point:

- LCL_Fib—lateral collateral ligament on the fibula;
- LCL_Fem—lateral collateral ligament on the femur;
- MCLa_Tib—anterior attachment of the medial collateral ligament on the tibia;
- MCLp_Tib—posterior attachment of the medial collateral ligament on the tibia.
- MCLa_Fem—anterior attachment of the medial collateral ligament on the femur;
- MCLp_Fem—posterior attachment of the medial collateral ligament on the femur;
- PT_Tib—patellar tendon on the tibia;
- PT_Pat—patellar tendon on the patella.
- QT_Pat—quadriceps tendon on the patella;
- BF_Tib—biceps femoris on the tibia;
- ST_Fib—semitendinosis on the fibula.

(For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

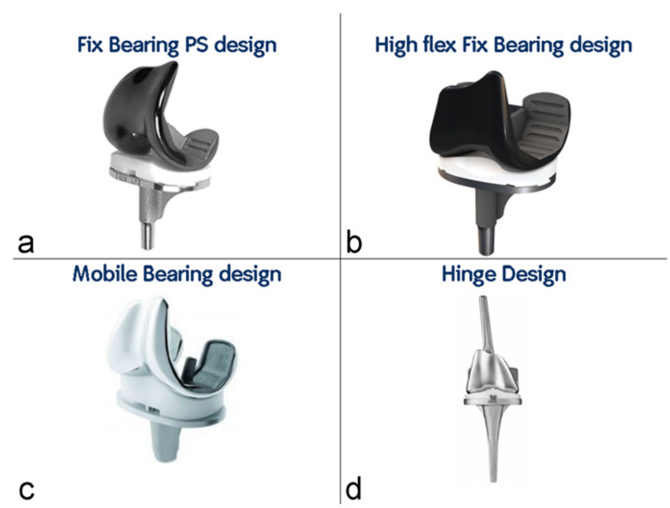

Fig. 2. The total knee arthroplasty designs analyzed in this study: (a) a fixed bearing, posterior stabilized (PS), design (Genesis II, Smith&Nephew, Memphis, TN); (b) a high flexion fixed bearing guided motion design (Journey BCS, Smith&Nephew, Memphis, TN); (c) a mobile bearing design (Solution EPP, Smith&Nephew, Memphis, TN) and (d) a hinge design (RT-Plus, Smith&Nephew, Memphis, TN).

2.3. Analyzed motor task

A 10 s loaded squat (one cycle), starting from 0° until a maximum flexion angle of 120°, was performed for each configuration, with a constant vertical hip load of 200 N and a sinusoidal vertical hip translation. These settings match the