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Joint loading asymmetries in knee replacement patients observed both pre- and six months post-operation

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ABSTRACT

Background: Studies have highlighted asymmetries in knee joint moments in individuals with osteoarthritis and joint replacements. However, there is a need to investigate the forces at the knee joints to establish the extent of loading asymmetries.

Methods: Twenty healthy (mean age, 62; range, 55–79 years) and 34 pre- to post-knee arthroplasty (mean age, 64; range, 39–79 years) participants performed gait and sit–stand activities in a motion capture laboratory. Knee joint forces and moments were predicted using inverse dynamics and used to calculate peak loading and impulse data which were normalized to body weight. Comparisons were made between affected and contralateral limbs, and changes from pre- to post-knee arthroplasty.

Findings: Pre-knee arthroplasty peak vertical knee forces were greater in the contralateral limb compared to the affected limb during both gait 3.5 vs. 3.2 $*$ body weight and sit–stand 1.8 vs. 1.5 $*$ body weight. During gait, peak knee adduction moment asymmetries significantly changed from pre- to post-knee arthroplasty (-0.3 to 0.8 $*$ % body weight $*$ m $*$ height), although differences in vertical knee forces remained. There were no significant changes in loading during sit-stand from pre- to post-knee arthroplasty. The healthy participants showed no noteworthy asymmetries.

Interpretation: This study showed loading asymmetries in knee forces between affected and contralateral limbs both pre- and post-knee arthroplasty. Continued over reliance of the contralateral limb could lead to pathology.

1. Introduction

Knee osteoarthritis (OA) is a common age-related pathology causing pain and loss of function (Fitzgerald et al., 2004). The prevalence of knee joint OA has increased in recent years and now comprises one of the greatest sources of expenditure for modern society (NJR, 2007). When OA causes significant pain and functional decline for an individual, joint surgery is used to replace degenerated articular surfaces, with knee arthroplasty (KA) being the most common procedure for advanced OA (NJR, 2010). Evidence suggests that KA patients experience more difficulty performing daily tasks than the healthy age, matched population (Noble et al., 2005), and they often use compensatory strategies during gait and sit-to-stand (Farquhar et al., 2009; McClelland et al., 2007). The symmetry of joint movement (kinematics) and loading (kinetics) has been described to vary in the health population during activities such as gait, although relatively small differences are commonly observed (Sadeghi et al., 2000). When an individual has joint pain and pathology significant asymmetries can develop between the affected and contralateral limb, commonly to reduce loading in pathological joints, which can increase loading in the contralateral limb.

Asymmetries between limbs have been reported during several activities of daily living in patients with OA or with joint replacement. Studies have combined motion capture and inverse dynamic techniques to show asymmetries of joint moments during sit to stand (Christiansen and Stevens-Lapsley, 2010; Farquhar et al., 2009), stair ascent (Lamontagne et al., 2011), and gait (Alnahdi et al., 2011). Difference in affected and contralateral joint loading has been assessed in patient pre- and post-total hip arthroplasty (post-THA) using motion capture and inverse dynamic modeling techniques (Shakoor et al., 2003). These studies found that the contralateral knee was subjected to higher dynamic loading during gait pre-operatively, which was retained post-THA (range 10–23 months), with three of the five knee force and moment variables being significantly higher in the contralateral limb (knee adduction and extension moment, medial knee contact force). This was despite improvements in pain and function scores. Metcalfe et al. recently showed that OA patients experienced increased

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joint moments in the affected knees compared to age matched healthy individuals. One year post-KA (patients received either unilateral or total replacement), the affected limbs had returned to normal, with slightly higher moments in the contralateral limb (Metcalfe et al., 2012). These previous studies, however, have relied on inverse dynamics techniques that either include basic or no muscle forces to calculate joint reactions. Research has shown that muscles and soft tissues have a significant contribution to forces and moments acting across a joint (Shelburne et al., 2006; Winby et al., 2009).

Recent evidence has shown a significant association between elevated joint loading and OA progression (Bennell et al., 2011). In addition, the contralateral limb has been shown to predict long term function post-KA (Farquhar and Snyder-Mackler, 2010) and a large proportion of primary TKA patients will have their contralateral joints replaced within 10 years (Sayeed et al., 2011). There is a need to expand research surrounding joint loading pre- and post-KA and include predicted muscle forces that estimate the full extent of joint loading asymmetries. We therefore investigated whether (1) pre-KA patients would have greater asymmetry in joint loading between limbs (larger loading through the contralateral knees) compared to healthy participants and (2) if this asymmetry would be retained post-KA.

2. Methods

We recruited 20 healthy participants between the ages of 50 to 80 years (nine men, 11 women) from the local community who had no pain in the lower limbs, no previous pathologies in the last 2 years, and no known musculoskeletal or neurological diseases. We also recruited the patients using the following criteria: (1) primary knee arthroplasty, (2) no other comorbidities which significantly affect pain and function, and (3) able to walk 50 m. The mean age of the patients was 64 years (range, 40–82 years); there were 14 men and 20 women. These patients were all diagnosed with OA after radiographs and clinical assessments were performed by their consultant; the patients were tested approximately 4 weeks prior to their KAs (14 unicompartmental KA and 20 total KA) and 6 months after their KA (Table 1). Operations were performed using a conventional medial parapatellar approach, the same implant was used for patients undergoing unilateral (The Oxford, Biomet Orthopaedics) and total KA (P.F.C. Sigma, Depuy, Johnson and Johnson). Despite there being some demographic differences between unicompartmental and total KA patients, they were grouped together to form one KA group in order to achieve statistical power in the analysis. Institutional and National Health Service (NHS) ethics approval were attained prior to the study, and written informed consent was obtained from each participant.

The participant demographics showed that patients scheduled to undergo KAs were slightly older and had higher BMI compared to the healthy participants although none of these variables were significantly different between the groups (Table 1). All pre-KA patients had higher perceived pain and instability scores, as well as lower perceived function, measured by the Western Ontario and McMaster Universities Arthritis Index (WOMAC) (Bellamy et al., 1988) and the Oxford Knee Score (OKS) (Dawson et al., 1998). Significant gains in perceived pain and function scores were observed from pre- to post-KA ($p < 0.001$).

Table 1
Demographics of 20 healthy and 34 pre- and post-knee arthroplasty (KA) participants.

Variable	Healthy	Pre-KA	Post-KA	Healthy vs. pre-KA	Pre-KA vs. post-KA
Age (years)	62(6)	64(10)	65(9)	$P = 0.43$	$P = 0.71$
Weight (kg)	78(13)	85(18)	86(17)	$P = 0.18$	$P = 0.57$
Height (cm)	166(11)	167(10)	167(10)	$P = 0.97$	$P = 0.97$
BMI	28(4)	31(6)	31(5)	$P = 0.23$	$P = 0.93$
WOMAC	1(3)	46(5)	17(13)	$P < 0.001$	$P < 0.001$
OKS	47(2)	24(9)	38(8)	$P < 0.001$	$P < 0.001$

Mean and standard deviation (in brackets); OKS = Oxford Knee Score; WOMAC = Western Ontario and McMaster Universities Arthritis Index.

Gait and sit to stand activities were assessed in all participants using a Vicon motion analysis system (Combination of 460 and T series, Vicon Motion Systems, Oxford, UK) and two Kistler force plates (Kistler Instrument AG, Kistler Group, Winterthur, Switzerland). Marker data were collected at 120 Hz, and analogue data from the force platforms were collected at 1080 Hz (Worsley et al., 2011). Marker and force plate data were low-pass filtered at 5 Hz during post-processing. Twenty-four retroreflective markers (9 mm) were placed directly on the skin of each participant using double-sided adhesive tape. Markers were placed in a modified Helen Hayes (Kadaba et al., 1990) marker set-up with anatomical landmarks established by a physiotherapist (PW). Additional markers were placed on the superior surface of the iliac crests to reconstruct the pelvis if other markers were occluded. Further markers were also added to the foot (the fifth metatarsal head, and cuboid and navicular bones) in order to model ankle inversion and eversion articulations more accurately (Fig. 1A). Participants were asked to perform gait and sit–stand activities three times. Gait trials were performed along a 10-meter walkway and were normalized from heel strike to heel strike. The sit–stand motion was normalized from full-sitting to standing with the knees and trunk extended. The chair used for the sit–stand activity was of a standard 45-cm height, and the back of the chair was removed to ensure all pelvic markers were visible to the motion capture cameras. Participants were encouraged to perform the activities as they normally would in their home environments.

We then used a previously published musculoskeletal modeling process (Worsley et al., 2011). Briefly, inverse dynamics were calculated from the motion capture and force plate data using musculoskeletal modeling software (The AnyBody Modeling System™, AnyBody Technology, Aalborg, Denmark) (Damsgaard et al., 2006). From these models we obtained the following parameters: knee joint kinematics (angles) and kinetics (resultant joint moments and forces). Key parameters were; 1) vertical force plate reaction, 2) vertical TFJ force, 3) posterior–anterior TFJ force, 4) TFJ flexion moment and 5) TFJ adduction moment. Patient-specific musculoskeletal models were derived from static standing postures (soft-tissue artifact is assumed to be minimal during quiet standing) and used to create the subject-specific models. Models were scaled from a single anthropometric data set (Klein Horsman et al., 2007) using criteria that takes individuals BMI into account. A 13-segment, rigid body model, with 16° of freedom, was orientated to match participants posture including the lower limb structures, the trunk, and the head. During the dynamic modeling process joint kinematics were established using a global optimization method, which utilized a set of Karush–Kuhn–Tucker optimality conditions. This approach calculates the position of each segment in relation to the measured markers, subject to the degrees of freedom within the model (Andersen et al., 2009). Once optimized kinematics were derived, inverse dynamics were performed. In order to solve the known moments about each joint muscles were recruited using a MinMax solver where the load is distributed across muscle elements so that fatigue of a given muscle is postponed as long as possible (larger muscles provide most of the force) (Rasmussen et al., 2001). The model had over 300 Hill-type muscle elements, these were established based on anthropometric data and International Society of Biomechanics (ISB) standards (Klein Horsman et al., 2007; Wu et al., 2002). Final joint forces and moments were derived from the combination of applied (force plate), known (segment mass), and optimized muscle forces acting about each joint.

The knee was simplified to a hinge joint because of the known soft tissue artifact errors in motion capture techniques. This constraint on the model was placed because evidence surrounding estimations of secondary motions of the knee (e.g. internal external rotation) from motion capture data show significant errors (>4°), despite optimization techniques (Andersen et al., 2010). With the known adaptations in knee alignment in the OA population, the coronal plane angle of the knee was