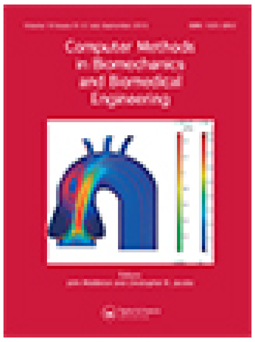

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An improved OpenSim gait model with multiple degrees of freedom knee joint and knee ligaments

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Musculoskeletal models are widely used to investigate joint kinematics and predict muscle force during gait. However, the knee is usually simplified as a one degree of freedom joint and knee ligaments are neglected. The aim of this study was to develop an OpenSim gait model with enhanced knee structures. The knee joint in this study included three rotations and three translations. The three knee rotations and mediolateral translation were independent, with proximodistal and anteroposterior translations occurring as a function of knee flexion/extension. Ten elastic elements described the geometrical and mechanical properties of the anterior and posterior cruciate ligaments (ACL and PCL), and the medial and lateral collateral ligaments (MCL and LCL). The three independent knee rotations were evaluated using OpenSim to observe ligament function. The results showed that the anterior and posterior bundles of ACL and PCL (aACL, pACL and aPCL, pPCL) intersected during knee flexion. The aACL and pACL mainly provided force during knee flexion and adduction, respectively. The aPCL was slack throughout the range of three knee rotations; however, the pPCL was utilised for knee abduction and internal rotation. The LCL was employed for knee adduction and rotation, but was slack beyond 20° of knee flexion. The MCL bundles were mainly used during knee adduction and external rotation. All these results suggest that the functions of knee ligaments in this model approximated the behaviour of the physical knee and the enhanced knee structures can improve the ability to investigate knee joint biomechanics during various gait activities.

Keywords: knee ligaments; knee joint; gait model

Introduction

Kinetic and kinematic characteristics of human gait have been investigated for many years and are well understood (Kadaba et al. 1990; Ounpuu 1994; Al-Zahrani and Bakheit 2008). In recent years, much research has focused on predicting muscle and joint contact forces during gait using musculoskeletal models (Lin et al. 2010; Pandy and Andriacchi 2010; Sasaki and Neptune 2010). However, two potential limitations commonly exist in musculoskeletal models, which may prevent accurate prediction of muscle and joint contact force during gait. First, muscles are considered to be the only force generators in the model and ligament forces are neglected during gait. However, some studies suggest that the predicted joint contact force may be underestimated due to the lack of ligament forces (Lin et al. 2010; Richards and Higginson 2010; Sasaki and Neptune 2010), especially the anterior cruciate ligament (ACL), for which reported peak forces ranged from 0.2 to 1.7 body weight during gait (Shelburne et al. 2004). Second, the knee is usually constrained as a one degree of freedom (DOF) hinge joint (flexion/extension) in most existing gait models (Anderson and Pandy 2001; Sasaki and Neptune 2010). However, some research indicates that knee motions in the frontal and transverse planes also exist during gait (Table 1) and are heavily affected by knee ligaments (Sutherland et al. 1980; Chao et al. 1983; Isacson et al. 1986; Kadaba et al. 1990). The lack of inclusion of knee motion in these two planes could result in inaccurate estimation of knee muscle forces and knee contact force due to an altered muscle excitation pattern (Glitsch and Baumann 1997; Xiao and Higginson 2008). Therefore, the purpose of this study was to develop an OpenSim gait model with multiple DOFs for the knee joint and knee ligament structures, and then verify the function of ligaments by evaluating independent knee rotations. This model will be available at www.simtk.org for public evaluation, refinement and application.

Methods

The gait model developed in this study is based on a lower extremity model with a torso and back joint (Delp et al. 1990; Anderson and Pandy 1999), which consists of 12 rigid segments, 23 DOFs (3 DOFs for hip joint, 1 DOF for knee joint and 1 DOF for ankle joint) and 92 muscle actuators.

The knee (tibiofemoral) joint

The reference frame of the tibia in the present model is based on the transverse axis, which passes through the

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Table 1. Mean range of total knee motion during gait cycle.

	Kadaba et al. $N = 40$	Sutherland $N = 15$	Isacson et al. $N = 20$	Chao et al. $N = 110$
Knee (°)				
Flexion	56.7	58	60.6	68
Varus	13.4	N/A	9	10
Rotation	16	12	12.9	13

centres of the medial and lateral posterior femoral condyles. The origin of the tibial reference frame lies on the transverse axis at the midpoint between these centres. The transverse axis points laterally and is the z axis of the tibia. The y axis is perpendicular to the transverse axis and points proximally. The x axis points anteriorly and is formed by taking the cross product of the y and z axes. The reference frame of the femur is fixed at the centre of the femoral head and has the same orientation as the reference frame of the tibia when the knee is fully extended. Six generalised coordinates described the rotations and translations of the tibia relative to the femur as knee abduction and adduction about the x axis; knee internal and external rotations about the y axis; knee flexion and extension about the z axis; knee anterior and posterior translations along the x axis; knee proximal and distal translations along the y axis; and knee medial and lateral shifts along the z axis. Three knee rotations and mediolateral translation are defined as independent variables. The knee proximodistal and anteroposterior translations are defined as a function of knee flexion based on previous research (Yamaguchi and Zajac 1989).

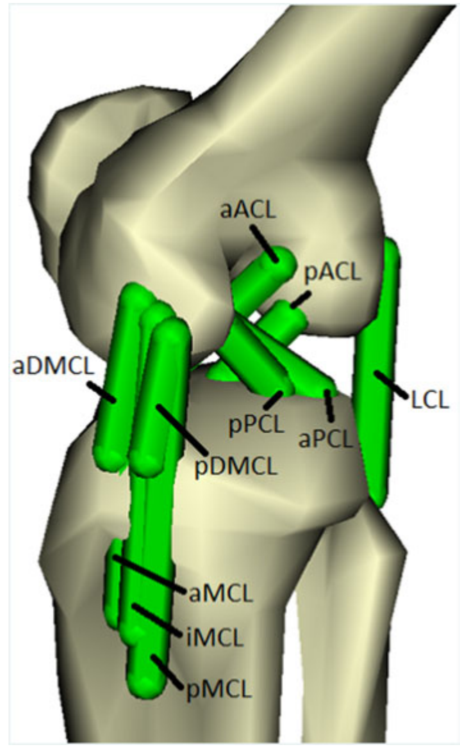

Figure 1. The attachment sites of knee ligament bundles. The abbreviation of ligament bundles are: aACL, anterior bundle of the ACL; pACL, posterior bundle of the ACL; aPCL, anterior bundle of the PCL; pPCL, posterior bundle of the PCL; aMCL, anterior bundle of the superficial layer of the MCL; iMCL, inferior bundle of the superficial layer of the MCL; pMCL, posterior bundle of the superficial layer of the MCL; aDMCL, anterior bundle of the deep layer of the MCL; pDMCL, posterior bundle of the deep layer of the MCL.

Knee ligaments

Ten separate bundles were used to model the geometrical and mechanical properties of knee cruciate and collateral ligaments (Figure 1). The ACL and posterior cruciate ligament (PCL) are each represented by an anterior bundle and a posterior bundle. The medial collateral ligament (MCL) is separated into two portions: a superficial layer, composed of anterior, inferior and posterior bundles; and a deep layer, represented by anterior and posterior bundles (Blankevoort et al. 1991). The lateral collateral ligament (LCL) is represented by one bundle (Pandy et al. 1998).

The path of each ligament bundle is considered as a straight line and the effect of ligament–bone contact is neglected. The ligament bundle properties are assumed to be nonlinear elastic, which means that ligament bundle tension is a function of its length L or strain ε (Equation (1)) (Blankevoort and Huiskes 1991).

$$\varepsilon = \frac{L - L_0}{L_0}, \tag{1}$$

where L_0 represents the zero-load length of a ligament, which is usually determined by the reference length and strain. The stiffness values and reference strains of the ligament bundles in this model are based on a previous study (Pandy et al. 1998).

The initial femoral and tibial attachment sites of ligament bundles in this model are based on the average values obtained from four knee data-sets (Blankevoort et al. 1991). Since different tibial coordinates exist between the present model and previous knee data-sets, the attachment sites of each ligament in this model are determined through a calibration strategy. First, the knee rotation, abduction/adduction and mediolateral translation are temporarily locked. Then, the attachment sites of each ligament bundle are adjusted in intervals of 2.5 mm in three axis directions based on the initial position. The reference length and orientations of each bundle are kept constant during the calibrating process. The attachment sites are determined when ligament bundles match the average ligament relative length change pattern during passive knee flexion from *in vitro* measurements (Blankevoort et al. 1991). Finally, the coordinate values of femoral attachment sites of each ligament are

Table 2. The attachment sites and parameters of the ligament bundles.

	Femur (cm)			Tibia (cm)			Reference position		
	X	*Y*	*Z*	*X*	*Y*	*Z*	Strain	Length (cm)	Stiffness (*N*)
aACL	− 0.718	− 40.037	0.407	1.657	− 3.009	− 0.074	0.02	3.23	1500
pACL	− 1.495	− 40.981	0.999	0.250	− 3.250	0.000	0.01	2.47	1600
aPCL	− 0.867	− 41.342	− 0.925	− 2.045	− 3.314	0.333	− 0.23	2.58	2600
pPCL	− 1.588	− 40.574	− 1.629	− 1.471	− 3.175	− 0.407	0.02	2.52	1900
LCL	− 0.978	− 40.056	3.450	− 0.710	− 6.056	3.725	0.02	5.59	2000
aMCL	− 0.741	− 40.435	− 3.500	0.768	− 7.960	− 2.762	0.02	7.22	2500
iMCL	− 1.274	− 40.620	− 3.351	0.249	− 8.256	− 2.910	0.04	7.31	3000
pMCL	− 1.777	− 40.361	− 3.351	0.249	− 9.403	− 2.640	0.02	8.80	2500
aDMCL	− 0.741	− 40.435	− 3.500	0.471	− 4.400	− 3.500	− 0.08	3.63	2000
pDMCL	− 1.777	− 40.361	− 3.351	− 0.500	− 4.400	− 3.500	0.03	3.72	4500

Notes: Abbreviations listed in this table are defined in Figure 1. The reference frame of the femur and tibia is defined in the Methods section.

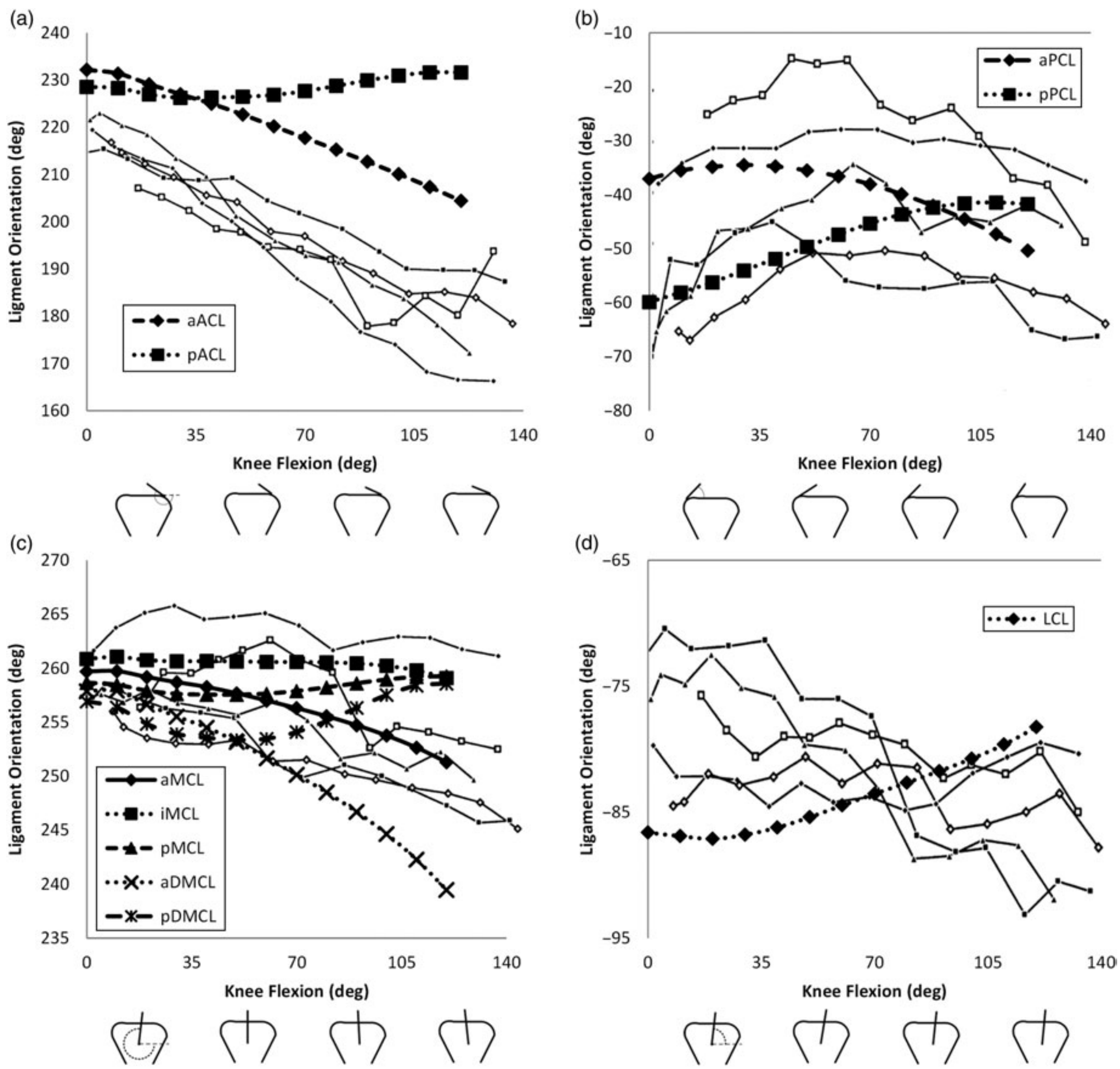

Figure 2. Orientation of knee ligaments in the sagittal plane. Orientation of a ligament bundle was defined as the angle formed between the ligament lines and the tibial plateau in the sagittal plane. Lines with small markers represent data obtained from five cadaver knees (Herzog and Read 1993).

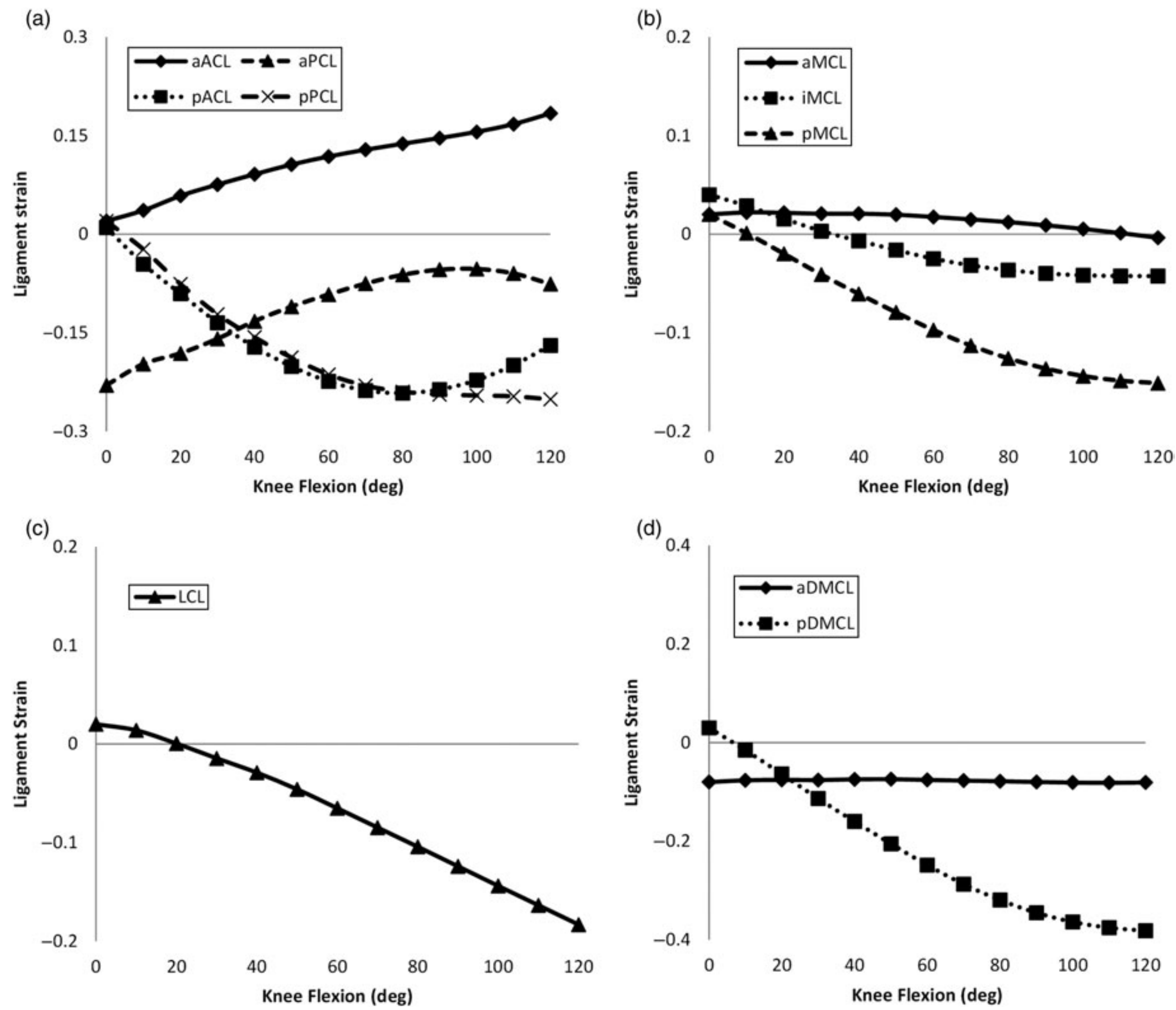

Figure 3. Ligament strain change by independent knee flexion. Ligament strain was defined as in Equation (1).

transferred from the tibial frame to the femoral frame in this model. The attachment sites and parameters of the model ligament bundles are given in Table 2.

Evaluation of knee rotations

Three independent conditions are evaluated for this model in OpenSim, which include (1) knee passive flexion from 0° to 120°, (2) knee passive internal rotation from − 30° to 30° and (3) knee passive adduction from − 15° to 15°. When the model is passively manipulated to represent knee rotation in one body plane, the knee rotations in the other two body planes are set to zero. For each ligament bundle, the ligament length is calculated by the distance between the femoral and tibial attachment sites and the ligament strain is determined by Equation (1).

Results

The orientations of the model ligaments are similar to the intact knee measurements (Figure 2) (Herzog and Read 1993). The anterior bundle and posterior bundle of knee cruciate ligaments, ACL and PCL, (aACL, pACL and aPCL, pPCL) twist during knee flexion, respectively. However, this intersection happens at approximately 40° of knee flexion for the ACL, but at nearly 90° for the PCL. The changes in orientation of the aACL and pPCL are both about 25° for the range of knee flexion. However, the angle decreases for the aACL, but increases for the pACL with increasing knee flexion (Figure 2(a),(b)). The aACL and pACL mainly provide forces during knee flexion and adduction, respectively. The aPCL is slack throughout the range of knee flexion, rotation and abduction/adduction. The pPCL is tight for knee abduction and internal rotation (Figures 3(a), 4(a) and 5(a)). The change in orientation of the LCL is less than 10° throughout the range of knee flexion (Figure 2(d)). The LCL is recruited for knee adduction and rotation, but is slack beyond 20° of knee flexion (Figures 3(c), 4(c) and 5(c)). The orientations of all the MCL bundles slightly change with knee flexion except for the anterior bundle of the deep layer of the MCL

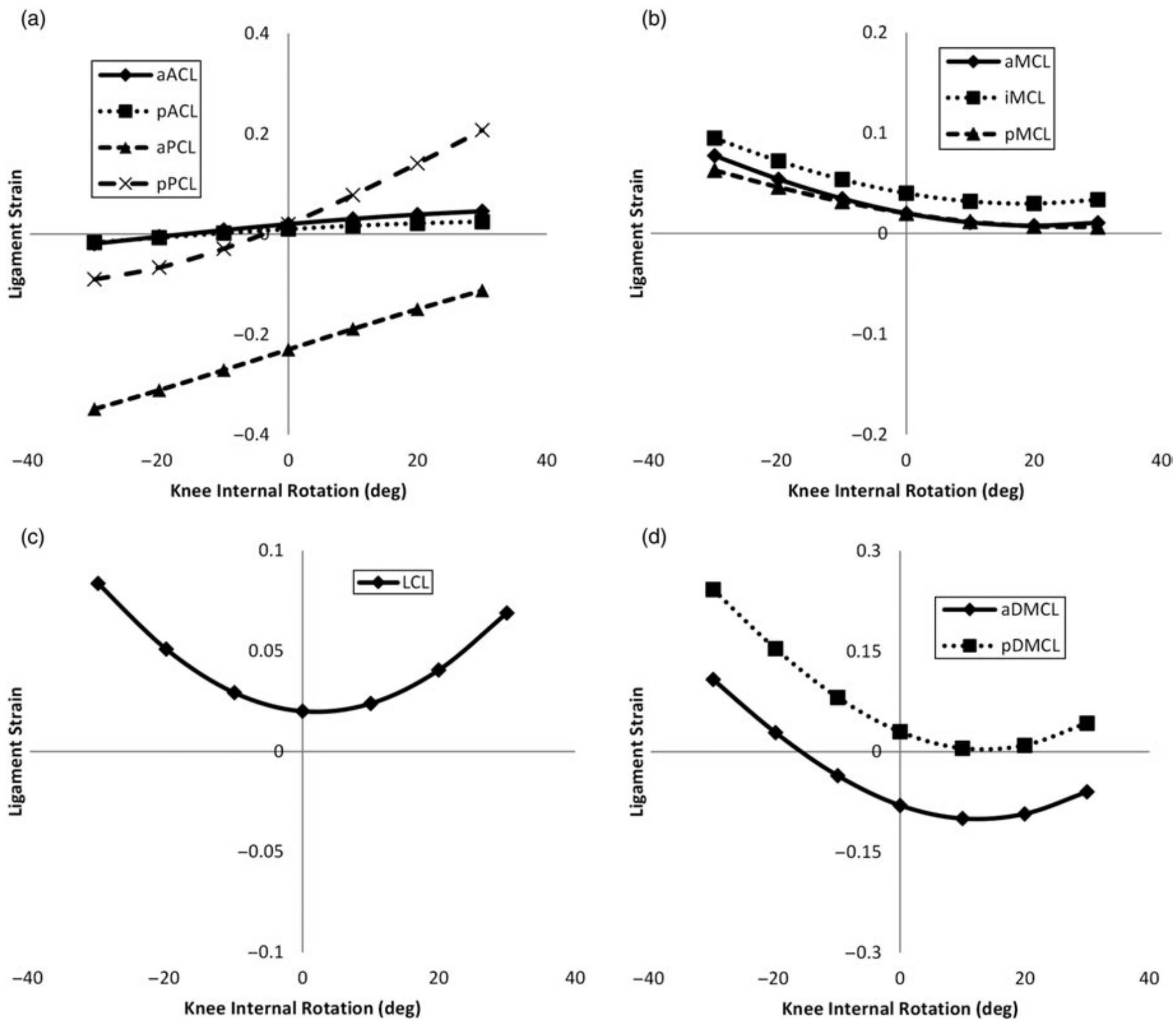

Figure 4. Ligament strain change by independent knee rotation. Ligament strain was defined as in Equation (1).

(aDMCL) (Figure 2(c)). All the bundles in the superficial and deep layer of the MCL are loaded for knee adduction and external rotation. The three MCL bundles in the superficial layer are also recruited for knee internal rotation though the strains are smaller than the condition of knee external rotation (Figures 3(b),(d), 4(b),(d) and 5(b),(d)).

Discussion

The zero-load length is a key parameter used to determine the status of the ligament bundle (recruited or slack). However, it is not directly known and for this study had to be indirectly calculated from reference length and strain. Some previous research indicated that anatomical differences existed for the attachment sites and reference lengths of the ligament (Hefzy et al. 1986; Edwards et al. 2008; Kopf et al. 2011). These variations could significantly affect the value of zero-load length and the judgment of whether the ligament bundle is recruited or not. In addition, this anatomical difference may be larger than the error from identifying the attachment sites of each ligament bundle and could explain the variations for ligament length and strain pattern in different research (Blankevoort et al. 1991; Pandy et al. 1998). A ligament strain sensitivity analysis was conducted. This indicated that the ligament strain error was dependent upon the zero-load length and ligament strain. The error was amplified with an increase of ligament strain and the variation of zero-load length, especially when the zero-length was underestimated. As little as 10% changes from zero-load length could cause an average strain error of about 0.1, as illustrated in Figure 6. Ligament strain can be measured with transducers directly attached to the ligaments in cadavers (Draganich and Vahey 1990; Bach et al. 1997), but data are limited for ligaments in the living body. Therefore, the reference strains of ligament bundles are usually adapted in an iterative fashion until model results resemble *in vitro* experimental data (Blankevoort and Huiskes 1996; Shelburne and Pandy 1997).

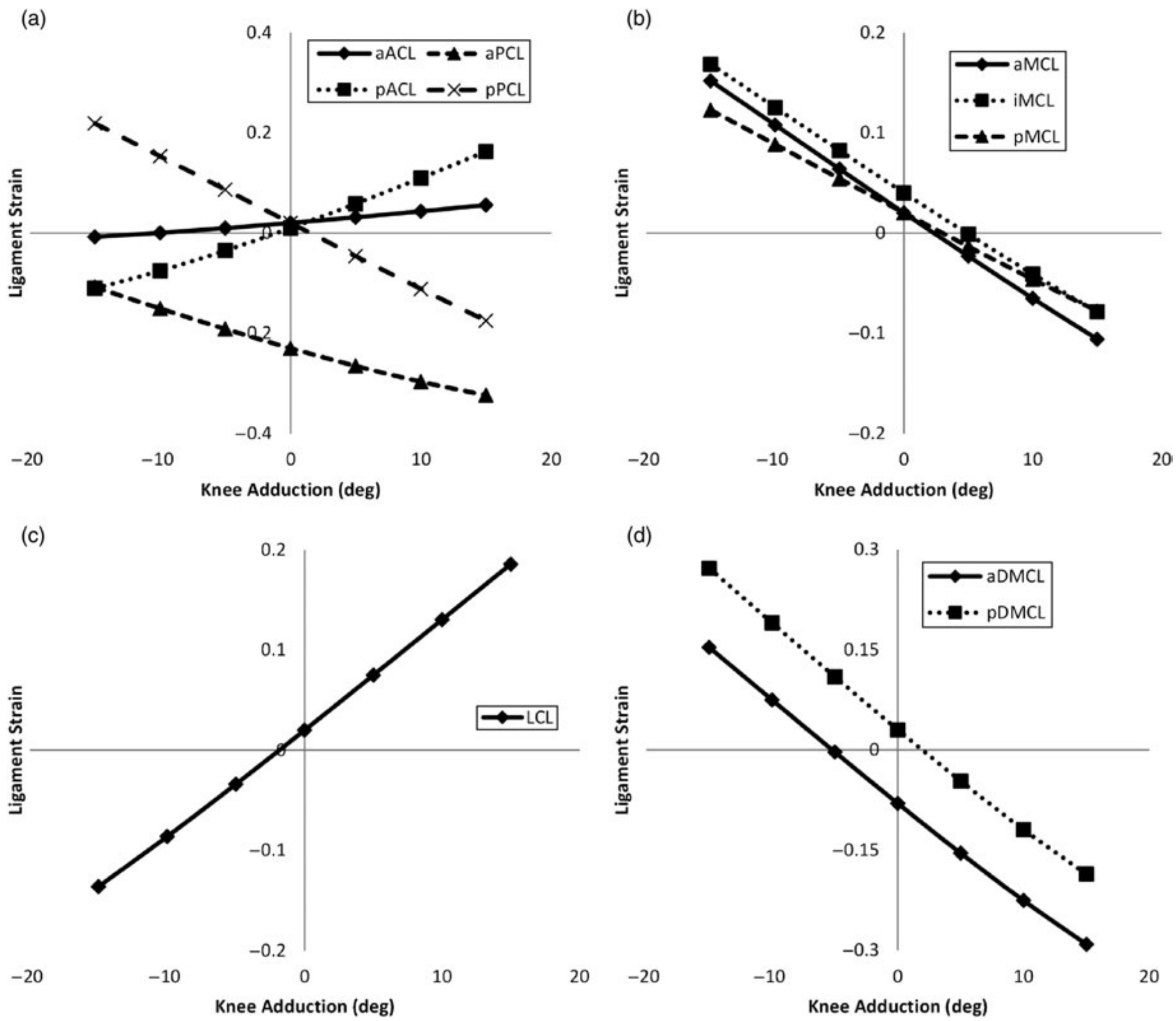

Figure 5. Ligament strain change by independent knee abduction and adduction. Ligament strain was defined as in Equation (1).

The length of each ligament bundle is considered as a straight line and is calculated between the femoral and tibial attachment sites in this model. However, real ligaments attach over a finite area of the bone and wrap around the bones. This approximation was reported and can be neglected in the sagittal plane during knee flexion (Blankevoort and Huiskes 1991). However, the observation that LCL and some bundles of MCL contact the tibia during extreme knee abduction/adduction and rotation indicate that ligament–bone contact may redirect the knee collateral ligaments and would, therefore, alter the calculated ligament force.

Although individual variation exists and the linear assumption of ligament bundles is a limitation, the present model still has strengths to represent real ligament functions. For the knee cruciate ligaments, the orientations of ACL followed by knee flexion and the intersection of aACL and pACL are consistent with a previous model (Pandy et al. 1998). The observation that there is strain in the aACL during knee flexion is in good agreement with experimental results (Li et al. 2004; Sonnery-Cottet and Chambat 2007). The model result indicating that the aACL and pACL are loaded for knee internal rotation and adduction is consistent with the ACL function to restrain tibial rotation and varus–valgus angulation (Zantop et al. 2006). It was not surprising that the aPCL is slack in any knee rotation condition due to the negative reference strain (-0.23). Both bundles of the PCL remain slack for passive knee flexion which agrees with previous research that reported the PCL did not generate force during knee flexion without external force (Nakagawa et al. 2004). The two bundles of the PCL show the same recruitment tendency for knee abduction and internal rotation, which are partly contradictory with the PCL function of resisting knee external rotation (Covey and Sapega 1994). This phenomenon can be explained by the different length change patterns of the pPCL among individuals and even between the left and right knees for the same individual

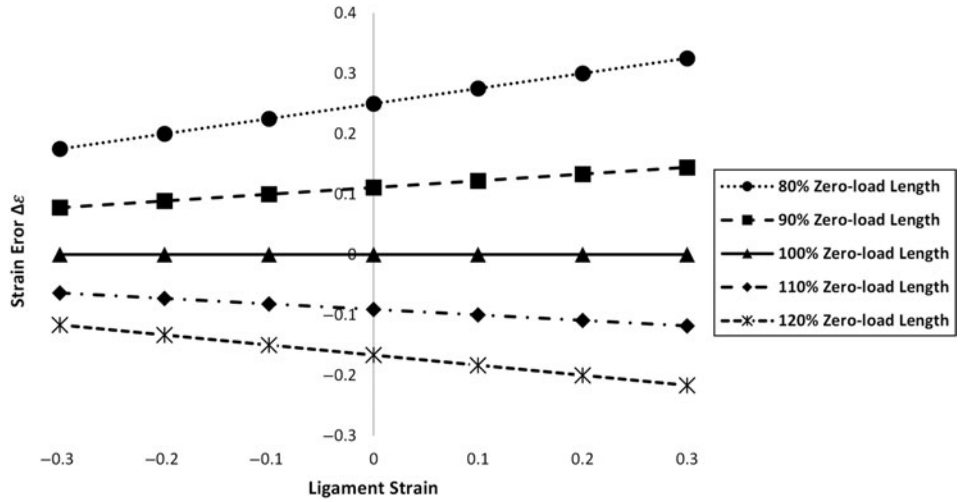

Figure 6. Ligament strain errors induced by a change in the zero-load length and ligament strain. The zero-load length was adjusted from 80% to 120% in 10% increments and the ligament strain was adjusted from −0.3 to 0.3 in 0.1 increments. The strain error $\Delta\epsilon$ was defined as the difference between estimated and actual ligament strains, and the positive strain error meant the estimated strain was overestimated.

(Blankevoort et al. 1991). For the knee collateral ligaments, there is no question that the main function of the MCL is to restrain valgus instability. This is reflected in our model by the loading of the MCL with knee abduction. However, in previous research, there was controversy relating to MCL function for knee internal or external rotation (Warren et al. 1974; Markolf et al. 1976; Seering et al. 1980; Jasty et al. 1982). The recruitment patterns of bundles of the MCL in this model mostly support the external rotation function. The result that there is considerable forces in the LCL for knee adduction and rotations and the LCL is slack for knee flexion coincides with the finding for the LCL in the physical knee (Sugita and Amis 2001; Lim et al. 2012).

In conclusion, the OpenSim model presented in this study is a 3D musculoskeletal model with enhanced knee structures. Two main contributions of this model are multiple DOFs for the knee joint, and knee cruciate and collateral ligaments, including geometrical and mechanical properties. The function of knee ligament bundles in this model was verified to be reasonable by evaluating independent knee motions in three body planes and comparing the results with the ligament functions in the physical knee and other existing knee models. This model provides the ability to investigate differences in knee joint biomechanics and ligament function during various types of gait. This was accomplished by improving the ecological validity of the model with a multiple DOFs knee joint and adding knee cruciate and collateral ligaments. By accounting for these functions and structures, the accuracy of the computed muscle forces and total joint load can be improved to obtain more realistic values.

Future work will include dynamic simulation of gait to characterise ligament function and total knee joint loads, which will allow the estimation of injury risks during dynamic activities. A comprehensive analysis of muscle force optimisation techniques with added ligament force contributions will also be performed. Finally, the muscle forces obtained with this knee model should be compared with those based on traditional knee models to compare and contrast the advantages of this improved knee model over traditional models, which have limited DOFs and no ligaments. This is especially important to better understand joint stabilisation and injury susceptibility.

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