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Author(s)	Otani, Tomohiro; Kobayashi, Yo; Tanaka, Masao
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Computational Study of Kinematics of the Anterior Cruciate Ligament Double-Bundle Structure during Passive Knee Flexion–Extension

Tomohiro Otani^{a,*}, Yo Kobayashi^a, Masao Tanaka^a

^aDepartment of Mechanical Science and Bioengineering, Graduate School of Engineering Science, Osaka University, Toyonaka, Osaka, 560-8531, Japan

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*Corresponding author

Email address: `otani@me.es.osaka-u.ac.jp` (Tomohiro Otani)

1 Abstract

2 The anterior cruciate ligament (ACL) comprises an anteromedial bun-
3 dle (AMB) and posterolateral bundle (PLB). Cadaver studies showed that
4 this double-bundle structure exhibits reciprocal function during passive knee
5 flexion–extension, with the PLB taut in knee extension and the AMB taut
6 in knee flexion. *In vivo* measurements indicated that straight-line lengths of
7 both bundles decrease with increasing knee-flexion angle (KFA). To inter-
8 pret these seemingly conflicting facts, we developed a computational ACL
9 model simulating the kinematics of the double-bundle structure during pas-
10 sive knee flexion–extension. Tibial and femoral shapes were reconstructed
11 from computed-tomography images of a cadaver knee and used to construct
12 an idealized model of an ACL including its bundles at the tibiofemoral joint.
13 The ACL deformations at various KFAs were computed by finite element
14 analysis. Results showed that the PLB was stretched in knee extension (KFA
15 = 0°) and slackened with increasing KFA. The AMB was stretched in knee
16 extension (KFA = 0°) and remained stretched on the medial side when the
17 knee flexed (KFA = 90°), but its straight-line length decreased with increas-
18 ing KFA. These findings are consistent with cadaver and *in vivo* experimental
19 results and highlight the usefulness of a computational approach for under-
20 standing ACL functional anatomy.

²¹ **keyword**

²² Anterior Cruciate Ligament, Functional anatomy, Passive knee flexion–
²³ extension, Double-bundle structure, Finite element method.

24 1. Introduction

25 The anterior cruciate ligament (ACL) is one of four essential knee-joint
26 ligaments that stabilize the joint, especially in the anterior drawer. ACL in-
27 jury is one of the most common knee-joint injuries and often occurs without
28 contact [1]. Many biomechanical studies of the ACL from various viewpoints
29 (*e.g.*, functional anatomy, mechanical properties) have been conducted to
30 better understand the mechanism of ACL injuries and to improve ACL re-
31 construction techniques, as shown in recent review articles [2, 3].

32 According to existing cadaver studies about ACL functional anatomy
33 [4, 5, 6], the ACL structure can be divided into two fiber bundles located
34 on the anteromedial side (anteromedial bundle; AMB) and the posterolat-
35 eral side (posterolateral bundle; PLB). It is commonly believed that this
36 fiber double-bundle structure has a reciprocal relation in that the PLB is
37 tensed in knee extension, and the AMB is tensed in knee flexion. Recent
38 *in vivo* three-dimensional imaging measurements revealed that the straight-
39 line lengths of both bundles were longest at low knee flexion angles (KFAs)
40 and shortened significantly with an increasing KFA. Based on their *in vivo*
41 experimental results [7, 8, 9, 10], Jordan *et al.* [8] noted that the function
42 of these two bundles may be better characterized as complementary, rather
43 than reciprocal.

44 To interpret these seemingly conflicting facts and deepen our understand-
45 ing of the function of the double-bundle structure, we hypothesized that a
46 computational approach to expressing the ACL deformation and kinematics

47 of the double-bundle structure in the mechanical sense might be a pow-
 48 erful approach. Various computational ACL models have been proposed
 49 [11, 12] following earlier computational three-dimensional (3D) ACL models
 50 with anisotropic properties exhibited by the fiber orientation to represent
 51 the stress field in the ACL during passive knee flexion–extension [13, 14]. In
 52 the past decade, greatly advanced simulations were used in studies that con-
 53 sidered other ligaments and cartilage as well [15, 16]. However, knowledge
 54 about the function of the ACL double-bundle structure is still limited, and
 55 there is still great interest in improving the means to reconstruct the ACL
 56 after injury [17]. Computational simulation based on existing computational
 57 mechanical ACL models and recently gained anatomical knowledge may help
 58 us interpret experimental facts and update our understanding of the function
 59 of the ACL double-bundle structure.

60 This study therefore aimed to develop a computational ACL model to ex-
 61 press the kinematics of the ACL double-bundle structure during passive knee
 62 flexion–extension. Tibial and femoral shapes were extracted from computed-
 63 tomography (CT) images, and their postures at various KFAs were repre-
 64 sented based on recent *in vivo* measurements. A computational ACL model
 65 with its double-bundle structure was then set up as a tibiofemoral joint. Its
 66 deformation under prescribed postures at various KFAs was then calculated
 67 by a finite element method. The fiber orientation of the double-bundle struc-
 68 ture and its deformations were then evaluated based on the computational
 69 results.

2. Materials and Methods

2.1. Tibiofemoral joint kinematics

To represent the tibiofemoral joint geometry and its postures during passive knee flexion–extension, the right femur and tibia geometries (supine position) were extracted from representative CT images of a female cadaver that are archived in the Visible Human Project [18] using ScanIP version 7.0 (Synopsys; Mountain View, CA, USA) (Figure 1 (a)). Insertion sites of the ACL at each bone— Γ_t on the tibia and Γ_f on the femur—were modeled as ovals [19], in which the sizes of the major axis [mm] and minor axis [mm] were set at 8.0:5.5 in Γ_t and 7.5:5.0 in Γ_f with consideration of the AMB and PLB positions [20, 21].

The femoral and tibial postures at various KFAs were expressed based on the functional flexion axis (FFA), in which the knee flexion–extension is expressed as rotations around two body-embedded axes (the flexion–extension and internal–external (IE) rotation axes) [22, 23]. The surgical trans-epicondylar axis in the femur was set based on *in vivo* measurements of knee-joint kinematics conducted by Asano *et al.* [24] and used as the flexion–extension axis (Fig. 1(b) top). The longitudinal rotation axis was set based on previous cadaver studies [23, 25] and used as the IE rotation axis (Fig. 1(b) bottom) at low KFAs (screw-home movement). The IE rotation was considered at low KFAs ($\leq 30^\circ$) [23] and expressed as the internal rotation of the tibia by a constant fraction with increasing KFA. The range of the IE rotation θ_{IE} [$^\circ$] was 30° based on previous experimental measurements [26, 23]. Addi-

tionally, the anterior–posterior (AP) displacement of the femur was used to describe its sliding motion on the tibial plateau at high KFAs (*e.g.*, as shown in [24, 27]). The AP displacement was considered at high KFAs ($> 45^\circ$) and expressed as femoral displacement to the posterior side of the tibial plateau proportionally with increasing KFA. The total AP displacement, l_{AP} [mm], was set to 10 mm based on previous experimental observations [24].

To evaluate the effects of the degrees of IE rotation and AP displacement on the femoral and tibial postures associated with the ACL kinematics, parametric studies of the AP displacement and IE rotation were conducted, and the results were compared with the straight-line lengths of the AMB and PLB. First, the degrees of IE rotation in three cases, $\theta_{IE} = 30^\circ$, 15° , and 0° , with a constant l_{AP} of 10 mm were compared. Next, the degrees of AP displacement in three cases, $l_{AP} = 10$, 5, and 0 mm with a constant θ_{IE} of 30° were compared.

2.2. ACL geometry construction

An ACL model with the double-bundle structure was constructed (Fig.1(d)). The initial geometry of the ACL was constructed to interpolate insertion sites on the tibia, Γ_t , and femur, Γ_f , at a KFA of 90° , when the straight-line length of the ACL was shortest. ACL reference geometry was constructed by in-plane rotation of both insertion sites of the initial ACL geometry to parallel the double-bundle structure. The ACL reference geometry was discretized by a set of eight-node hexahedral elements, consisting of 18,880 elements with

115 20,459 nodes.

116 Fiber orientations of each bundle were defined by the method of Otani
117 and Tanaka [28] to assign unit direction vectors of fibers. In each bundle,
118 the fiber direction vector, \mathbf{a} , was defined as the unit vector of the spatial
119 gradient of the scalar variable, ϕ , which is given by

$$\mathbf{a} = \frac{\nabla\phi}{\|\nabla\phi\|}. \quad (1)$$

120 The spatial distribution of ϕ was modeled to be smooth and was expressed
121 by solving the Laplace equation in each bundle while following the Dirichlet
122 boundary condition ($\phi = 1$ on Γ_t and $\phi = 0$ on Γ_f).

123 2.3. Computation of ACL deformation

124 We simulated ACL deformation at various KFAs using the ACL model
125 of Weiss *et al.* [29], which can express nearly incompressible, transversely
126 isotropic properties of the ACL. The strain energy density function W was
127 described by the following three terms:

$$W = W_{\text{iso}}(\hat{I}_1) + W_{\text{aniso}}(\lambda) + W_{\text{vol}}(J) \quad (2)$$

128 where W_{iso} and W_{aniso} are the chronic components that express the isotropic
129 and anisotropic characteristics, respectively. W_{iso} and W_{aniso} are given by

$$W_{\text{iso}}(\hat{I}_1) = c_1(\hat{I}_1 - 3) \quad (3)$$

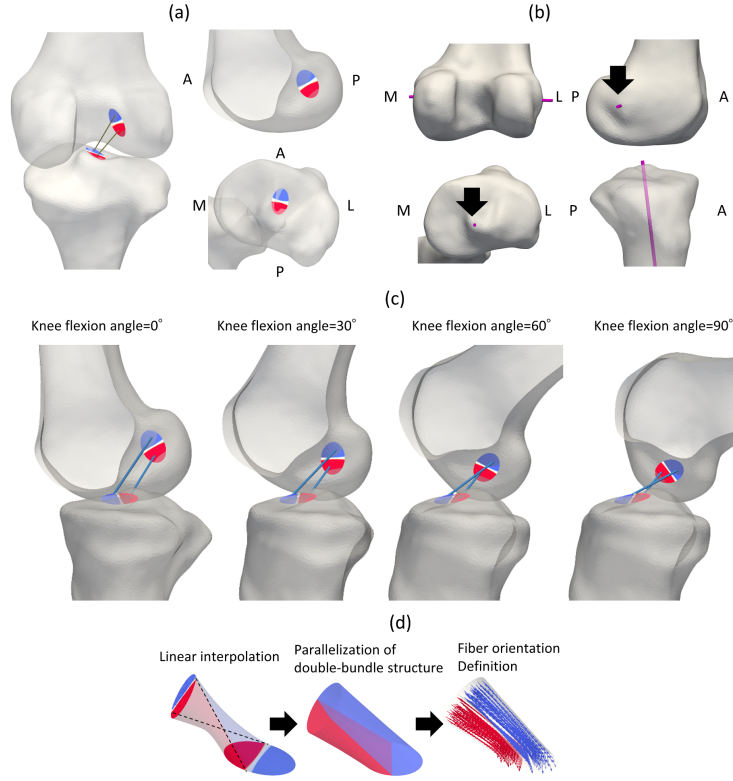


Figure 1: Workflow representing the geometries and postures of a tibiofemoral joint with the anterior cruciate ligament (ACL). (a) Femoral and tibial shapes extracted from computed-tomography images with the ACL insertion sites, including the anteromedial bundle (AMB, blue) and posterolateral bundle (PLB, red). (b) Axes of knee flexion-extension for the femur and interior-exterior rotation of the tibia. (c) Postures of the tibiofemoral joint at knee flexion angles of 0° , 30° , 60° , and 90° (viewed from the medial side). (d) Reference geometry construction of the ACL and definition of fiber orientation in the double bundle. Initially, the ACL geometry was represented by linear interpolation between insertion sites at a knee-flexion angle of 90° , when the straight-line length of the ACL was shortest (left). The ACL reference geometry was constructed by in-plane rotation of both insertion sites in the initial ACL geometry to parallel the double-bundle structure (center). The fiber orientations of each bundle were defined by the method presented by Otani and Tanaka [28] (right).

$$\lambda \frac{\partial W_{\text{aniso}}}{\partial \lambda} = \begin{cases} 0 & \text{if } \lambda < 1 \\ c_3[\exp[c_4(\lambda - 1)]] & \text{if } 1 \leq \lambda \leq \lambda^* \\ c_5\lambda + c_6 & \text{Otherwise} \end{cases} \quad (4)$$

131 where c_1 , c_3 , c_4 , c_5 , and c_6 are constants, and λ^* is the threshold used to de-
 132 termine the phenomenological properties of the bundles. These terms were
 133 established by the modified first invariants of the right Cauchy–Green defor-
 134 mation tensor, $\hat{I}_1 = \text{tr}(\hat{\mathbf{C}})$, and the stretch ratio expressing the deformation
 135 along the orientations of the fiber bundles, $\lambda = \sqrt{\mathbf{a}^T \cdot \mathbf{C} \cdot \mathbf{a}}$. Here, $\hat{\mathbf{C}}$ repre-
 136 sents the modified right Cauchy–Green deformation tensor and is expressed
 137 as $J^{-\frac{2}{3}}\mathbf{C}$ based on the volume ratio, J , and the right Cauchy–Green defor-
 138 mation tensor, \mathbf{C} . The volumetric component, W_{vol} , is expressed in terms of
 139 J as follows:

$$W_{\text{vol}}(J) = k_v \ln J^2 \quad (5)$$

140 where k_v is the penalty parameter. The values of these parameters were set
 141 as described by Pena *et al.* [30] and calculated by fitting the experimental
 142 measurements reported by Butler *et al.* [31].

143 The ACL deformations at various KFAs were calculated by solving the
 144 weak form of the equilibrium equation using the Galerkin finite element
 145 method with the Newton–Raphson scheme (*cf.* [32]). Selective reduced inte-
 146 gration was applied to alleviate volume locking. The linearized equation was
 147 calculated by PARDISO implemented in the Intel Math Kernel Library.

148 The initial stretch ratio was set for each bundle using the method of
 149 Limbert *et al.* [14] to avoid non-physiological fiber extension. In this study,
 150 the initial stretch ratio was assumed to be uniform for each bundle and was
 151 set at 0.75 for the AMB and at 0.7 for the PLB, not to exceed the failure
 152 limit of the ACL— that is, 1.2, based on Butler *et al.* [33]—regardless of the
 153 KFA.

154 3. Results

155 3.1. ACL kinematics during passive knee flexion-extension

156 Figure 2(a) and (b) show the motion of the femur relative to the tibia and
 157 the surgical trans-epicondylar axis for KFAs from 0° to 90° . The tibiofemoral
 158 joint motion indicates that the posterior displacement of the lateral condyle
 159 was higher than that of the medial condyle, which is well-known behavior
 160 based on *in vivo* and *in vitro* measurements (*e.g.*, [34, 35]). The changes in
 161 the ACL bundle lengths in the above case for KFAs from 0° to 90° are shown
 162 as black lines in Fig. 2(c). Both the AMB and PLB lengths were within the
 163 ranges observed thorough *in vivo* measurements by Yoo *et al.* [9] at KFAs of
 164 45° and 90° .

165 Furthermore, the effects of the degrees of AP displacement and IE ro-
 166 tation were evaluated by parametric studies about l_{AP} and θ_{IE} . As l_{AP} de-
 167 creased to 5 and 0 mm, the AMB was shortened, and the AP displacement
 168 decreased; these lengths were outside the range of the *in vivo* measurements
 169 reported by Yoo *et al.* [9] at a KFA of 90° . However, the decreases in the

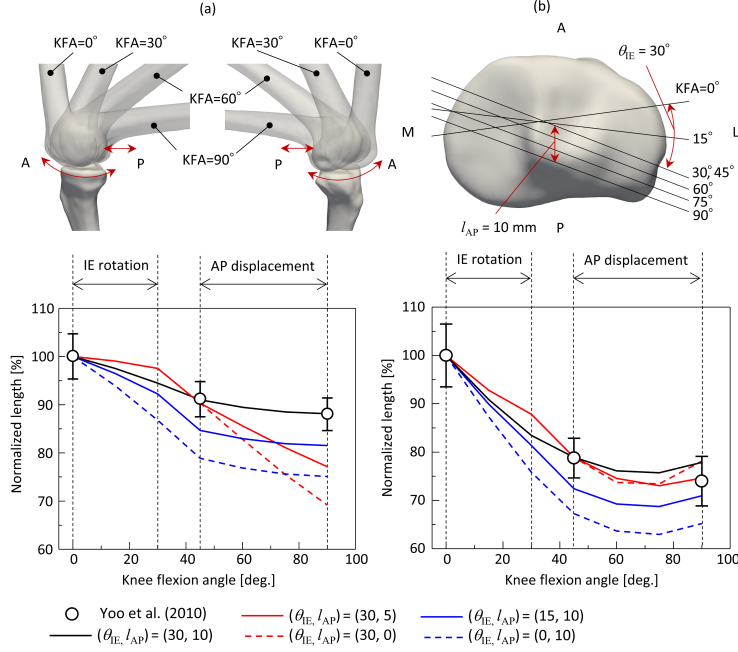


Figure 2: Tibial and femoral postures during knee flexion angle (KFA) from 0° to 90°. Representative postures of the tibiofemoral joint at KFAs of 0°, 30°, 60°, and 90° (a) and movement of the surgical epicondylar axis for KFA from 0° to 90° (b) for a total IE rotation angle of 30° and AP displacement of 10 mm. (c) Normalized lengths of the anteromedial bundle (AMB, left) and posterolateral bundle (PLB, right) at KFAs from 0° to 90° with various degrees of the internal-external rotation, θ_{IE} , and AP displacement, l_{AP} . *In vivo* experimental data (open circles) [9] are shown for comparison.

170 PLB length were consistent regardless of the AP displacement and remained
 171 within the range of the *in vivo* measurements reported by Yoo *et al.* [9] in
 172 all cases. When θ_{IE} was decreased to 15° and 0°, although both the AMB
 173 and PLB lengths reduced similarly in all cases, the reduction ratios increased
 174 with decreasing θ_{IE} . These values were out of the range of the *in vivo* mea-
 175 surements reported by Yoo *et al.* [9] except for the case in which $\theta_{IE} = 15^\circ$
 176 and KFA = 90°.

177 3.2. ACL fiber orientations

178 Snapshots of the ACL geometry, fiber orientation of each bundle, and
179 stretch ratios at KFAs of 0° , 30° , 60° , and 90° are shown in Fig. 3. The
180 fiber orientation was visually represented as tangential lines showing the
181 direction vectors of fibers. The double-bundle structures in the ACL were
182 almost parallel at a KFA of 0° and became twisted and curved when the knee
183 flexed. The stretch ratio of the PLB in the posteromedial side was locally $\lambda >$
184 100% at a KFA of 0° and decreased with increasing KFA. Almost all domains
185 in the PLB slackened ($\lambda < 100\%$) even at a KFA of 30° . The stretch ratios
186 of the AMB on the medial side were also $\lambda > 100\%$ regardless of the KFA,
187 whereas those on the lateral side were relatively low and slackened when the
188 knee flexed.

189 To evaluate the extents of the two bundle stretches quantitatively, we
190 calculated the volume fraction of the stretch ratio in each bundle domain
191 (Fig. 4). Approximately 50% of each bundle had stretched ($\lambda > 100\%$) when
192 the KFA was 0° . In the PLB, the slackened domain ($\lambda < 100\%$) consistently
193 increased with increasing KFA, and almost all domains had slackened at a
194 KFA of 90° . In the AMB, the degree of stretching became relatively mild
195 at a KFA of 30° , and approximately 38% of the AMB domain was stretched
196 ($\lambda > 100\%$); the domain remained stretched regardless of the KFA. The
197 AMB had the least stretched domain (approximately 23%) at a KFA of 60° ,
198 and it was approximately 29% at a KFA of 90° .

199 Finally, we considered the influences of the AP displacement and IE ro-

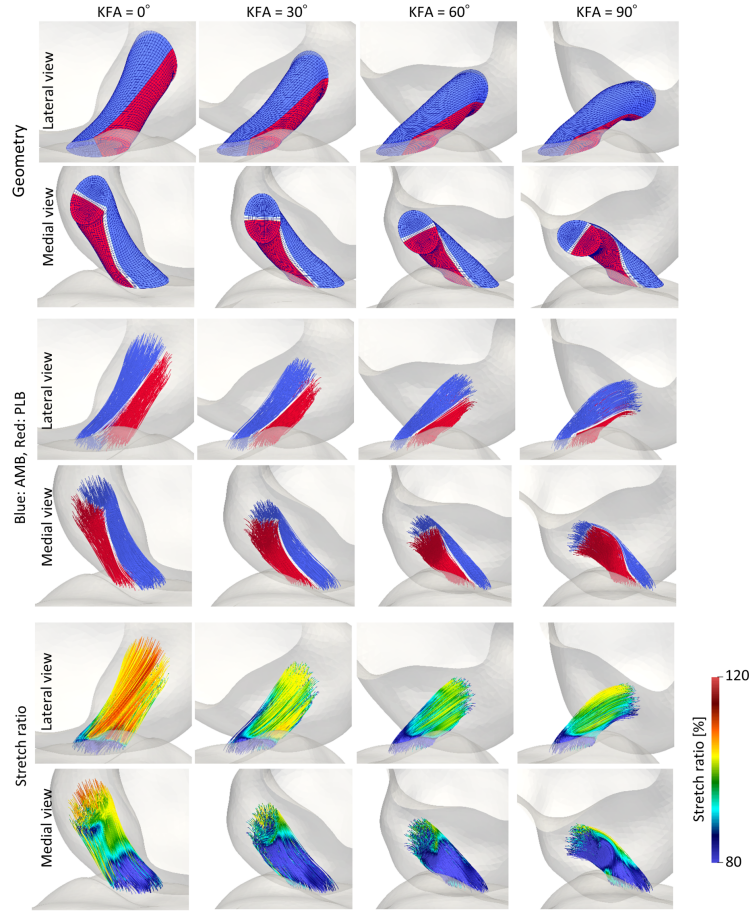


Figure 3: Geometry of the anterior cruciate ligament (ACL, top), fiber orientations of the anteromedial bundle (AMB) and posterolateral bundle (PLB) (middle), and fiber stretch ratios, λ (bottom), at knee flexion angles of 0°, 30°, 60°, and 90° (viewed from the medial and lateral sides). The fiber orientation is represented as tangential lines along the fiber direction vectors, and the fiber stretch ratios are visualized on these tangential lines.

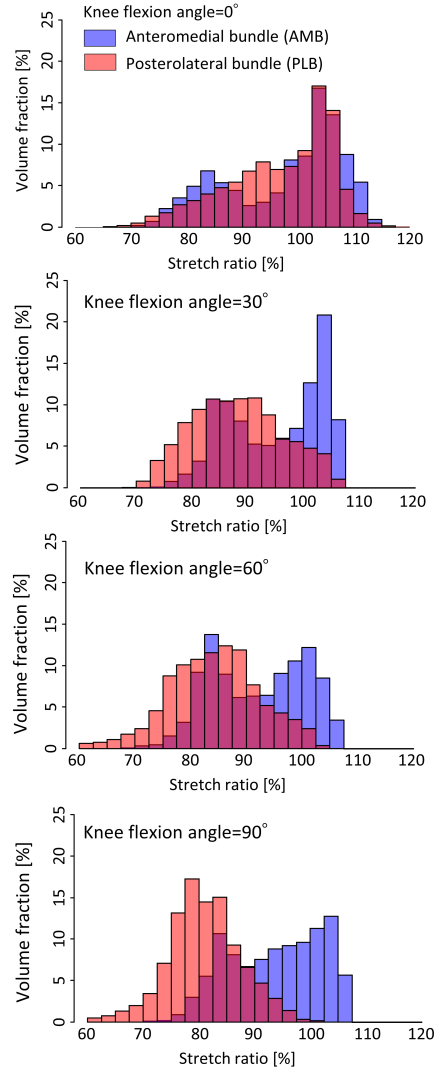


Figure 4: Volume fractions of the stretch ratios in the anteromedial bundle (AMB, blue) and posterolateral bundle (PLB, red) at knee flexion angles of 0° (top), 30° (top middle), 60° (bottom middle), and 90° (bottom).

200 tation on the ACL deformation. The ACL deformations in the flexion states
 201 relative to the fully extended state ($\text{KFA} = 0^\circ$) were computed for two cases:
 202 small $l_{\text{AP}} = 5$ mm with $\theta_{\text{IE}} = 30^\circ$ and small $\theta_{\text{IE}} = 15^\circ$ with $l_{\text{AP}} = 10$ mm.
 203 Figure 5 (a) shows the volume fraction of the stretch ratio in each bundle
 204 domain at a KFA of 90° in each case. Although the PLB domain was almost
 205 slack (stretch ratio $< 100\%$) in the case of $(\theta_{\text{IE}}, l_{\text{AP}}) = (30, 10)$ (Fig. 4 (bot-
 206 tom)), the PLB domain was still partially stretched (stretch ratio $> 100\%$)
 207 in in the cases of $(\theta_{\text{IE}}, l_{\text{AP}}) = (30, 5)$ and $(15, 10)$.

208 To evaluate the differences between the ACL deformation states in the
 209 above computations, we evaluated the changes in the AMB and PLB ridge
 210 lengths, which were defined as connections between the major vertexes of
 211 the insertion sites along the ACL (black lines in Fig. 5 (b)). Figure 5 (c)
 212 shows the changes in the AMB and PLB ridge lengths from KFAs from 0°
 213 to 90° in each case. The length of the AMB ridge line was almost constant
 214 during knee flexion in all cases. However, the length of the PLB ridge line
 215 monotonically decreased with increasing KFA and fell below 80% when KFA
 216 $= 90^\circ$ in the case of $(\theta_{\text{IE}}, l_{\text{AP}}) = (30, 10)$, but increased with increasing KFA
 217 from 45° , approaching 90% when $\text{KFA} = 90^\circ$ in other cases.

218 4. Discussion

219 The function of the ACL double-bundle structure has been of great in-
 220 terest, with various experimental studies having evaluated its functional
 221 anatomy [36]. Results of earlier cadaveric studies have been interpreted as

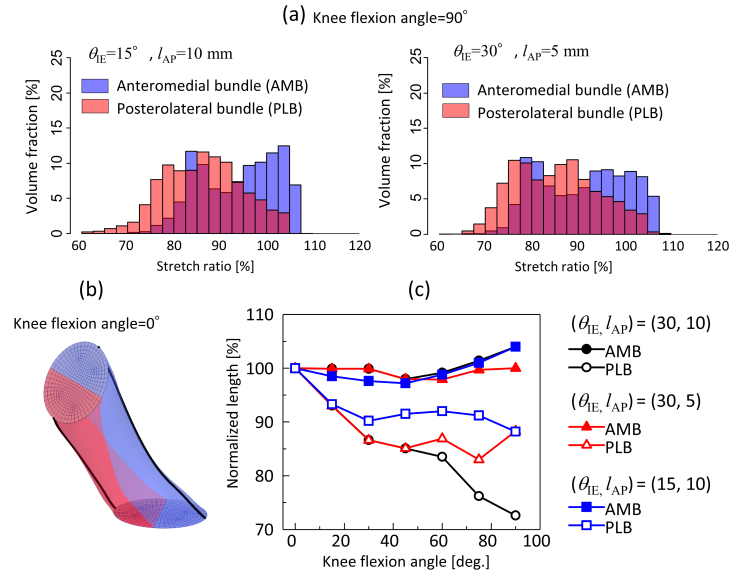


Figure 5: (a) Volume fraction of the stretch ratio in the anteromedial bundle (AMB, blue) and the posterolateral bundle (PLB, red) at knee flexion angles of 90° in the case that $(\theta_{IE}, l_{AP}) = (30, 5)$ (left) and $(15, 10)$ (right). (b) The AMB and PLB domains are shown in blue and red, respectively. (c) Normalized lengths of the AMB and PLB ridges (black lines in (b)) as the KFA increased from 0° to 75°.

222 showing that the double-bundle structure functions reciprocally, such that
223 the PLB is taut in knee extension, and AMB is taut in knee flexion [4, 5, 6].
224 Recent *in vivo* measurements showed that both bundles are longest at full
225 knee extension [7, 8, 9, 10]. Amis [36] summarized these facts and noted that
226 AMB is close to isometric when the knee flexes, whereas the PLB slackens.

227 The present study developed a computational ACL model to represent the
228 kinematics of the ACL double-bundle structure during passive knee flexion–
229 extension based on recent *in vivo* measurements. The results showed that
230 the PLB was stretched at full knee extension and slackened when the knee
231 flexed. In contrast, the AMB was continuously taut, especially on the me-
232 dial side, regardless of the KFA, although the straight-line length of the AMB
233 monotonically decreased with increasing KFA. This apparently isometric be-
234 havior of the AMB can be explained by a torsional deformation of the ACL
235 determined by the relative positional relation between the insertion sites on
236 the tibiofemoral joint (Fig. 3). This result is consistent with those of both
237 cadaver and *in vivo* experimental results. This finding also shows that the
238 function of the double-bundle structure of the ACL is associated with the 3D
239 kinematics of the insertion sites on the femur and tibia.

240 Furthermore, the effects of the IE rotation and AP displacement on the
241 kinematics of the ACL bundles during knee flexion were evaluated. Results
242 showed that small degrees of IE rotation and AP displacement made the
243 reduction in the bundle ridge lengths small (Fig 5) compared with the reduc-
244 tion in the straight-line lengths (Fig. 2) during knee flexion, which highlights

its torsional deformation. This finding suggests that small degrees of IE rotation and AP displacement enhance the ACL torsional deformation, which may reduce the slack in the PLB when the knee is flexed (Fig. 5). The PLB slackness at high KFAs is well acknowledged from clinical measurements [36]. Thus, under normal conditions, these complex motions may act to reduce the load on the PLB when the knee is flexed against the torsional deformation of the ACL. However, the PLB might not be slackened, even at high KFAs, when the IE rotation and AP displacement are small. Therefore, the 3D ACL kinematics and deformation during knee flexion–extension should be considered in terms of the ACL length as well as its torsional deformation.

4.1. Limitations

In this study, we modeled the ACL kinematics using several simplifying assumptions based on previous studies. Therefore, the obtained results may have inconsistencies in terms of the subject-specific ACL kinematics. From this viewpoint, this study has two primary limitations. First, the subject-specific posture of the tibiofemoral joint at various KFAs was not considered in this study. Instead, we represented the femoral and tibial postures based on the FFA with consideration to the IE rotation and AP displacement. We conducted parametric studies of the effects of IE rotation and AP displacement on the tibiofemoral posture and confirmed anatomical consistency of represented postures in terms of the ACL kinematics. However, knee joint motions vary widely due to subject-specific differences, measurement

267 approaches, and FFA selection (*e.g.*, [37, 38]). Thus, it is still challenging
 268 to uniquely determine the knee-joint motion from tibiofemoral bone geome-
 269 tries. To overcome this issue and evaluate actual ACL kinematics during
 270 knee flexion--extension, *in vivo* measurements of the kinematic relationship
 271 between the two ACL insertion sites would be useful. Second, we simplified
 272 the mechanical properties of the ACL, such as the reference geometry and
 273 fiber orientation of each bundle, due to incomplete anatomical knowledge.
 274 However, recent experimental studies showed that the microstructural prop-
 275 erties and mechanics of each bundle are different [39, 40, 41, 42]. Thus, lack
 276 of knowledge about these mechanical properties makes computational evalu-
 277 ations of the mechanical stress field of the ACL challenging. To address this
 278 issue, further development of the ACL model that reflects the mechanical
 279 characteristics is strongly required.

280 The anatomical characteristics of the ACL double-bundle structure are
 281 still under debate in the medical field, and the ACL torsional deformation
 282 is central in the discussion of this issue. Śmigielski *et al.* [43] conducted a
 283 cadaveric study including 111 knees and concluded that the ACL forms a flat
 284 ribbon without a clear separation between the AMB and PLB . Following
 285 this finding, Noailles *et al.* [44] studied the geometric characteristics of the
 286 ACL using 60 cadaver knees and concluded that the torsion in the ACL fibers
 287 because of the relative position of bone insertions makes the ACL appear to
 288 have a double-bundle structure. Furthermore, Skelley *et al.* [41] reported
 289 that most of the microstructural and material properties of the ACL appear

290 to follow a linear gradient across the ligament, rather than varying between
291 bundles. Regarding this issue, our results show that torsional deformation
292 can also occur when the AMB and PLB have different deformation charac-
293 teristics. To bear this specific deformation, it may be reasonable that the
294 AMB and PLB have different material properties. In future research, we plan
295 to investigate the effects of spatial differences in the material properties of
296 the ACL on its mechanical state during knee flexion—extension, which may
297 contribute to the interpretation of recent ACL anatomical findings from a
298 mechanical viewpoint.

299 5. Conclusions

300 This study computationally modeled ACL kinematics, focusing on its
301 double-bundle structure, during passive knee flexion—extension. The results
302 showed that the PLB was taut in knee extension, whereas the AMB was
303 consistently taut, regardless of the KFA, although the straight-line length of
304 AMB consistently decreased with increasing KFA. This apparently isometric
305 behavior of the AMB can be explained by torsional deformation of the ACL
306 determined by the relative positional relation between the insertion sites on
307 the tibiofemoral joint. These results are consistent with existing experimental
308 facts, and thus highlight the capabilities of the computational modeling to
309 deepen our understanding of ACL kinematics and the functional anatomy of
310 its double-bundle structure.

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317 **Conflict of interest**

318 The authors have no financial or personal interests in the work reported
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322 **Ethical Approval**

323 None.

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