One-leg hop kinematics 20 years following anterior cruciate ligament rupture: Data revisited using functional data analysis

Kim Hébert-Losier [a,*], Alessia Pini [b], Simone Vantini [b], Johan Strandberg [c], Konrad Abramowicz [c], Lina Schelin [d], Charlotte K. Häger [d]

[a] National Sports Institute of Malaysia, National Sports Complex, Bukit Jalil, Peti Surat 7102, 57700 Kuala Lumpur, Malaysia
[b] MOX – Department of Mathematics, Politecnico di Milano, Piazza Leonardo da Vinci 32, 20133, Milan, Italy
[c] Department of Mathematics and Mathematical Statistics, Umeå University, 901 87, Umeå, Sweden
[d] Department of Community Medicine and Rehabilitation; Physiotherapy, Umeå University, 901 87, Umeå, Sweden

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1. Introduction

Anterior cruciate ligament (ACL) injuries are common worldwide, each year affecting approximately ¼ million Americans (Silvers and Mandelbaum, 2011) and up to 0.05% of several national populations (Røtterud et al., 2011). In the short-term, acute interventions may instigate high socio-economic costs (Brophy et al., 2009), while the in-juries themselves can cause deficits in knee function and strength still apparent 6–9 months post-injury (Xergia et al., 2013). In the middle-to-long term, ACL-injured individuals show changes in cartilage morphology up to 2 years post-treatment (Frobell, 2011); low rates of return to competitive sports despite promising functional outcome scores (Ardern et al., 2011); biomechanical differences compared to controls during functional tasks, such as walking (Patterson et al., 2014) and jogging (Kuenze et al., 2014); persisting knee-joint laxity 17 years post-reconstruction in ~70% of cases (Aït Si Selmi et al., 2006); and precursors of osteoarthritis or osteoarthritis itself ~25 years post-reconstruction in 50–90% of cases (Pernin et al., 2010; Tengman et al., 2014a; Yamaguchi et al., 2006).

Conservative treatment of ACL ruptures traditionally involves a physiotherapeutic (ACL_{PT}) approach including analgesic modalities, re-habilitative exercises, and physical activity modification. Alternatively, surgical reconstruction (ACL_R) is undertaken in addition to physiother-apy.

There is a longstanding controversy regarding whether ACL ruptures should undergo surgery and the long-term functional differences between ACL_{PT} and ACL_R (Delincé and Ghafil, 2012; Smith et al., 2014). Given the range of outcome measures and lack of consensus regarding their interpretation (Delincé and Ghafil, 2012), it becomes difficult to define differences between patient groups. Although some studies report more degeneration, greater knee instability, and poorer subjective outcomes with conservative treatment (Mihelic et al., 2011), others suggest relatively similar outcomes between ACL_{PT} and ACL_R (Ageberg et al., 2008; Frobell et al., 2010; Von Porat et al., 2006).

* Corresponding author.
E-mail address: hl.kim@isn.gov.my (K. Hébert-Losier).

Current evidence does not sufficiently address movement pattern in patients with ACL ruptures, which could be clinically useful to monitor rehabilitation and aid in determining risk of re-injury (Engelen-van Melick et al., 2013). Few studies have investigated the long-term effects (i.e., >15 years) of both ACL_{PT} and ACL_R treatment approaches on knee-joint kinematics during functionally demanding tasks, particularly in relation to matched knee-healthy controls (CTRL). Existing research provides mixed findings, with no consistent conclusions regarding the presence or extent of long-term biomechanical and functional deficits in ACL_{PT} and ACL_R (Roos et al., 2014; Stensdotter et al., 2013; Von Porat et al., 2006).

In most biomechanical reports, the use of traditional statistics is common and generally reduces continuous data series into discrete variables (e.g., peak knee flexion angle and time-to-peak force) for statistical comparisons. Yet, discrete variables cannot completely capture all the variability within continuous data. Over the years, functional data analysis (FDA) methods have been introduced to clinical biomechanics, including functional limits of agreement (Roislien et al., 2012), statistical parametric mapping (Pataky et al., 2013), bootstrap prediction bands (Lenhoff et al., 1999), point-wise functional ANOVA (Godwin et al., 2009), and functional principal components (Ryan et al., 2006). Such methods consider the time-dependent structure of the continuous biomechanical data (e.g., kinematic or kinetic curves) and may therefore discern differences overlooked by traditional analy-ses or provide further insight.

Many univariate statistical methods generalize to a multivariate framework. However, it is often challenging to perform inference with suitable local control of the family-wise error rate. Confidence bands derived from gait data through bootstrapping (Lenhoff et al., 1999) have been used to address the local inference problem via the duality of significance tests and confidence bands. Pataky et al. (2010; 2013) present an interesting approach that considers the time-dependent structure of biomechanical data, while accounting for multiple comparisons. This approach can be applied to various inference problems including functional ANOVA (Pataky et al., 2015). Gaussian smoothing and random field theories have also been used to adjust statistical threshold levels in biomechanics (Penny et al., 2011). This approach controls for the family-wise error rate using information about the smoothness of the random field, is less conservative than the Bonferroni correction, and permits calculations of P values for time clusters. Here, we propose using the interval testing procedure (ITP) (Pini and Vantini, 2013), a type of FDA that identifies time-intervals in which populations of interest differ over the *entire time-domain* of continuous data. The proposed method is non-parametric, does not depend on smoothness parameters, and provides an interval-wise control of the family-wise error rate.

Recently, we explored differences in one-leg hop kinematics between ACL_{PT}, ACL_R, and CTRL subjects involved in a long-term follow-up study (KACL20) using traditional statistics to compare peaks, ranges, and instantaneous values (Tengman et al., 2015). Here, we revisit the data and propose using an ITP-based ANOVA method to examine the *entire time-domain* relating to knee kinematics. Our aims were to employ ITP-based ANOVA on knee-joint kinematics of one-leg hops and compare kinematic curves between and within ACL_R, ACL_{PT}, and CTRL groups. Based on existing literature (Deneweth et al., 2010; Gokeler et al., 2010; Orishimo et al., 2010; Paterno et al., 2010) and the conventional analyses of these kinematic data (Tengman et al., 2015), we anticipated that FDA would identify time intervals where groups and legs differ, with the involved leg of ACL-injured subjects exhibiting lesser knee flexion during take-off and landing (Gokeler et al., 2010; Orishimo et al., 2010; Tengman et al., 2015), as well as greater knee abduction (Paterno et al., 2010; Tengman et al., 2015) and external rotation (Deneweth et al., 2010; Tengman et al., 2015) during landing. Isolating the time intervals where groups differ from this functionally demanding task is anticipated to provide information not available from conventional analyses that may promote our understanding of movement control in these groups, highlight possible compensation strategies after ACL injury, assist in distinguishing between successful and non-successful rehabilitation, and help guide clinical decision-making (e.g., readiness for sports participation and re-injury risk).

2. Method

The dataset used for this investigation comes from a long-term follow-up study (KACL20) involving ACL_R and ACL_{PT} subjects, and sex-and age-matched CTRL. Between-group differences in one-leg hop kinematics using traditional statistics have been published (Tengman et al., 2015), but here we employ FDA and consider the *entire time-domain* of the knee-kinematic curves.

2.1. Subjects

To meet inclusion, subjects had to be in good self-reported health with no contra-indication to complete the study protocol, such as an on-going injury or disease affecting their movement ability. Individuals were excluded when presenting with a current or prior traumatic mus-culoskeletal injury to the knee (other than the original injury in the ACL_R and ACL_{PT} subjects), inflammatory or rheumatic disease, neurological condition, or a bilateral ACL injury. Following radiological exams, the ACL subjects were examined by an orthopedist and physiotherapist to confirm eligibility. All CTRL subjects were screened by a physiotherapist to ensure appropriate inclusion and sought to match the age and sex of ACL-ruptured subjects. The study adhered to the *Declaration of Helsinki* and was approved by the Regional Ethical Review Board.

All subjects provided written informed consent to participate in this study, which was part of the KACL20 (Knee injury; ACL after more than 20 years) project addressing several aspects of knee function (Tengman et al., 2014a, 2014b; Tengman et al., 2015). The subjects' demographic characteristics and hopping distances are reported in Table 1. All ACL_R and ACL_{PT} subjects had been injured ~20 years ago (range: 17–28 years) and treated in two separate hospitals following different treatment protocols, with subjects' activity levels, graft types, and rehabilitation protocols detailed in full by Tengman et al. (2014a). The physical activity levels at the day of study participation of the three groups were similar (Tengman et al., 2014a) based on the International Physical Activity Questionnaire (Craig et al., 2003).

2.2. Experimental procedures

Subjects were familiarized with the experimental protocol and tested in one session. After recording height and body mass; subjects com-pleted a 6-min stationary bicycle warm-up at a fairly light intensity, i.e., 11 on the 20-point Borg scale (Borg, 1982). Subjects then practiced the one-leg hop sub-maximally under supervision. Familiarization was followed by a 2-min rest, after which testing began.

The one-leg hop for maximal distance was selected as experimental task since it is commonly employed to clinically assess functional capac-ities after ACL injuries (Ageberg et al., 2007), shows moderate correla-tions with patient-reported outcomes (Reinke et al., 2011; Sernert et al., 1999), and is reliable for assessing function post-ACL reconstruc-tion (Sernert et al., 1999). The one-leg hop was performed barefoot since footwear can influence the human–ground interaction (Bishop et al., 2006) and jump performance (LaPorta et al., 2013). Subjects began the test standing upright on one leg over a custom-made force-plate sampling at 1200 Hz (Department of Biomedical Engineering and Informatics, University Hospital of Umeå, Sweden). Force-plate data were time synchronized with the motion analysis system and used to determine hop take-off. Subjects were requested to hop forward maximally, landing on the same leg without losing balance. Arms were held across the chest to limit armswing contribution to performance and occlusion of lower-body markers. Hops were performed three times on each leg, starting on the non-injured leg for the ACL groups

Table 1
Demographic characteristics and maximal one-leg hop distances of subjects presented by group. Data are presented as means (standard deviations). Injured and non-injured legs are in ACL_R and ACL_{PT}, whereas non-dominant and dominant legs are in CTRL. Significant between-group differences were derived from non-parametric analyses (Kruskal–Wallis and Mann–Whitney *U* with Bonferonni corrections).

Subjects [$n = 95$]	Age [y]	Body mass index [kg/m^2]	Proportion of men [%]	One-leg hop distance [m]	
				Injured or non-dominant	Non-injured or dominant
ACL_R (n = 31)	46 (5)	27 (3)[a]	64.5	1.13 (0.27)	1.20 (0.26)
ACL_{PT} (n = 33)	48 (6)	28 (4)[a]	63.3	1.00 (0.22)	1.10 (0.26)
CTRL (n = 31)	47 (5)	25 (3)[b,c]	64.5	1.08 (0.23)	1.07 (0.25)

ACL_R, ACL-ruptured subjects treated with reconstructive surgery and physiotherapy; ACL_{PT}, ACL-ruptured subjects treated conservatively with physiotherapy only; CTRL, knee-healthy controls.
[a] Significantly different ($P < 0.05$) from CTRL.
[b] Significantly different ($P < 0.05$) from ACL_R.
[c] Significantly different ($P < 0.05$) from ACL_{PT}.

and dominant leg for controls (i.e., leg preferred to kick a ball), followed by the contra-lateral leg. When subjects failed to perform a hop properly (e.g., lost balance during landing), an additional trial was completed after rest.

2.3. Motion capture

Body motion was monitored during hops at 240 Hz using a calibrated 8-camera 3D motion analysis system (Oqus 300 +, Qualisys AB®, Gothenburg, Sweden) and QTM software v.2.7 (Qualisys AB®, Gothenburg, Sweden). Forty-two retro-reflective markers were taped onto the skin over anatomical landmarks following standardized guidelines (Tengman et al., 2015). From the reference markers, an 8-segment full-body biomechanical model with 6 degrees of freedom was constructed (Visual3D Professional™ Software v.4.96.7, C-Motion Inc., Germantown, Maryland, USA), with the local coordinates of all body segments derived from a static measurement captured prior to the hop-ping trials. Only data from the longest hop on each leg (determined from the horizontal displacement of the lateral malleolus marker) were analyzed. Hop distances are presented in Table 1.

2.4. Data processing

Marker data were routinely interpolated in QTM using a B-spline interpolation, allowing a maximum of 30 frames for gap filling, prior to exporting files to V3D. Both marker and force-plate data were then filtered with a 6-Hz bi-directional second order low-pass Butterworth filter.

Take-off event was defined from the kinetic data as the instance when the vertical ground reaction forces reached minimal values (i.e., 5 N). In the absence of a second force-plate to detect ground con-tact, touch-down was determined from kinematics as when the vertical velocity of the lateral malleolus marker reached a minimum value (Tengman et al., 2015). This method was chosen as it could be consistently applied across individuals, was not affected by missing segments caused by marker occlusion, and was used in our prior investigation (Tengman et al., 2015). All events were manually inspected to verify correct identification and adjusted when required. Based on these events, hops were divided into three phases: 1) take-off, before and including take-off; 2) flight, between take-off and touch-down; and
3) landing, following and including touch-down.

Kinematic parameters were calculated using rigid-body analysis and Euler angles obtained from the static calibration. Knee-joint angles (°) were computed using an x-y-z Cardan sequence equivalent to the Joint Coordinate System (Grood and Suntay, 1983). The knee-joint kinematic curves in the three planes of motion were extracted for FDA, wherein more positive values indicate greater knee flexion, adduction, and internal rotation in the sagittal, coronal, and transverse planes.

2.5. Statistical method

In this paper, we propose using FDA on knee-joint kinematics to compare curves between- and within-groups throughout the one-leg hop. We first applied an ITP-based MANOVA (Pini and Vantini, 2013) to the three-dimensional knee-kinematic curves to identify in which intervals of the movement the groups differed. The result of this analysis confirmed that there were significant differences (at the 5% level) between groups throughout the one-leg hop, except in vicinity of the take-off event. Since results from MANOVAs do not identify the dimensions in which the differences are present, an ITP-based ANOVA procedure (Pini and Vantini, 2013) was applied to the curves in the three planes independently from one another to identify which groups and where in the movement kinematics differences were present. Finally, to adjust for the multiplicity of variables analyzed (i.e., the three planes), a Bonferroni correction was applied.

To facilitate interpretation of the data, the three phases of the one-leg hop (i.e., take-off, flight, and landing) were delineated in the data. Data were then aligned using landmark registration to account for individual differences in the speed and duration of hops (e.g., flight times ranged from 150 to 350 ms). More specifically, the events of maximal knee flexion during the take-off and the landing phase and the specific events of take-off and touch-down were used as landmarks. Time was expressed as a percentage of the three phases, wherein the time from the first max-imal knee flexion to the take-off event represented 50% of the take-off phase, with the remaining 50% spanning the time prior to maximal knee flexion. Similarly, the time from the touch-down to the maximal knee flexion event represented 50% of the landing phase, with the remaining 50% spanning the time following maximal knee flexion. Finally, the time from the take-off to the touch-down event represented 100% of the flight phase. The instances of maximal knee flexion were selected for data alignment during the take-off and landing phases to facilitate between-study comparisons and clinical interpretation. According to such landmark registration, time was piece-wise linearly transformed to assure that the four designated landmarks (i.e., maximal knee flexion during take-off, take-off event, touch-down event, and maximal knee flexion during landing) occurred at the same relative time point for each subject. The latter adjustments allowed the continuous data series to be compared between subjects using identical relative time points.

Here, the functional data are represented through a B-spline expansion (Ramsay et al., 2009), implicitly splitting the time-domain into smaller parts associated with the coefficients of the basis expansion, thereby enabling identification of time-intervals wherein groups differed. In contrast to the non-adjusted *P* values, which only control the probability of wrongly detecting a coefficient component by component, the ITP-adjusted *P* values provide an extra control and ensure that the probability of wrongly rejecting any time-interval (i.e., false positive) is below the chosen significance level. In addition to the ITP adjustments, a Bonferroni correction was applied to the *P* values to account for the number of planes analyzed (i.e., sagittal, coronal, and transverse).

To compare motion curves between groups, the three groups were compared two at a time using Scheffé-based pairwise comparisons (Abramowicz et al., 2014). This procedure was first applied to data from the injured legs of ACL_R and ACL_{PT} and non-dominant legs of CTRL, and then to the data from the non-injured and dominant legs of these groups. An ITP-based *t*-test was then applied to the point-wise difference of curves between legs within each group, calculated as the injured minus non-injured leg in ACL_R and ACL_{PT} and non-dominant minus dominant leg in CTRL. The injured leg of the ACL groups was com-pared to the non-dominant leg of the CTRL group to be consistent with our previous work (Tengman et al., 2015) and to provide the most strin-gent comparison considering that the non-dominant leg tends to be more variable (Wang and Watanabe, 2012). Figures representing the mean knee-kinematic curves for each group and individual curves for each subject were generated, and the time-intervals wherein differ-ences were observed at the 5% and 1% level of significance were highlighted. Due to the interval-wise control of the family-wise error rate provided by the ITP, within each analysis (i.e., for each plane and leg), the probability of wrongly selecting any interval is below 5% and 1%, respectively, with Bonferroni correction. Given that the Bonferroni correction is known to be conservative and that previous clinical papers do not account for multiple comparisons (Ageberg et al., 2007; Gokeler et al., 2010; Roos et al., 2014), the uncorrected results are supplied as supplementary material. All computations were performed in R version 3.03.

3. Results

3.1. Between-group comparisons

Statistical comparisons of the knee-kinematic curves during the en-tire one-leg hop of the injured (ACL_R and ACL_{PT}) and non-dominant (CTRL) legs (Fig. 1) revealed no marked differences between ACL_R and CTRL. In contrast, flexion/extension in ACL_{PT} substantially differed from CTRL, with the difference spanning the maximal knee flexion during both take-off and landing. Specifically, ACL_{PT} exhibited less knee flexion than CTRL from 0 to 55% of the normalized take-off phase and 44–73% of the normalized landing phase. In contrast, comparisons of the kinematic curves involving non-injured and dominant legs (Fig. 2) revealed no significant between-group differences.

3.2. Within-group comparisons

No significant between-leg differences in the knee-angle curves were identified within the CTRL and ACL_R groups (Fig. 3). Conversely, ACL_{PT} showed significant between-leg differences, wherein the injured leg was more flexed during the first part (4–22%) of the normalized flight phase. Moreover, the injured leg of ACL_{PT} showed greater external rotation during the later part (57–85%) of the normalized landing phase.

4. Discussion

Using FDA, this paper confirms earlier results that long-term (~20 years) knee-joint movement discrepancies exist in ACL-ruptured individuals, notably when treated conservatively. This analysis, which statistically compared the entire knee-kinematic curves of one-leg hops between and within three different groups, also expands on previ-ous findings of altered knee kinematics post-ACL rupture (Oberländer et al., 2014; Tengman et al., 2015) by delineating the relative time-inter-vals during which movements differ. The FDA approach analyzes the entire knee-kinematic waveform and seems well suited to outline differences between patient groups, extending on previous reports targeting a few discrete variables (i.e., peaks, means, and ranges)(Augustsson et al., 2006; Gokeler et al., 2010; Orishimo et al., 2010; Paterno et al., 2010; Tengman et al., 2015). Such comprehensive analy-ses can be useful for hop evaluations and assist in detecting deviations in movements from anticipated norms. For example, our data inform practitioners that ACL_{PT} are likely to exhibit less knee flexion during one-leg hop take-off and landing compared to CTRL. The observed movement deficiencies are presumably due to a lack of strong stabiliz-ing structures around the knee, decreased single-leg balance abilities (Stensdotter et al., 2013), fear of movement (Hartigan et al., 2013) resulting in a potential protective strategy, or weak knee extensors (Schmitt et al., 2012; Tengman et al., 2014b), all of which the present study cannot disentangle.

As hypothesized, between-group and between-leg differences in ACL-ruptured groups were identified, although only in ACL_{PT}. In terms of between-group analyses, and consistent with previous investigations, the injured leg of ACL_{PT} was less flexed during take-off and landing (Gokeler et al., 2010; Orishimo et al., 2010) and more externally rotated during landing than CTRL (Tengman et al., 2015), which some studies have also seen in ACL_R (Deneweth et al., 2010). Our findings expand on previous work employing more traditional statistics and our previ-ous analysis on these data (Tengman et al., 2015) by defining that differ-ences in knee movement patterns occur not only at specific instances, such as at peak knee flexion during take-off, but rather throughout a larger part of take-off and landing. Here, our analysis spanned the entire one-leg hop movement, including the flight phase, which was not addressed previously. Our clinical ACL_{PT} population did exhibit minor differences during the flight phase between injured and non-injured legs. Whether flight-phase kinematics is clinically meaningful needs further investigation, in which FDA could be useful.

Our current analyses indicate no marked differences in knee kinematics between ACL_R and ACL_{PT}, which could be in part due to the conservative nature and properties of Bonferroni corrections (Moran, 2003). Indeed, adjusting for multiple comparisons can increase the probability of Type II error (Armstrong, 2014), and without adjusting, certain differences in movement patterns between ACL-ruptured groups become evident (see supplementary materials). However, we here present the Bonferroni-adjusted results to minimize the probabil-ity of Type I error. Furthermore, the kinematics of ACL_R did not signifi-cantly differ from those of CTRL at any time during the hop, which contradicts our hypotheses and prior conventional analyses performed on these data (i.e., ACL_R exhibited coronal and transverse plane differ-ences versus CTRL) (Tengman et al., 2015). The presented analysis did, however, not include covariates, which might explain some of the differing results with those reported by Tengman et al. (2015) where sex and severity of osteoarthritis were considered. Also, the temporal aspects of the jumps might differ between groups (e.g., time of maximal knee flexion), which we did not analyze here. Various factors may con-tribute to hop performance, including age and sex (Ageberg et al., 2001; Hewett et al., 2010), as well as take-off angle and technique (Wakeling, 2009). Accounting for hop distance might also further explain the performance of this task. We are currently further developing the FDA approach to incorporate covariates, i.e., extending its use to ANCOVA models.

The fact that the time interval of significant differences between ACL_{PT} and CTRL in the sagittal plane involved the event of maximal knee flexion during take-off and landing substantiates the use of the latter measures in more traditional statistical approaches. Clinically, it may be easier to quantify changes in peak flexion than relative intervals during which differences occur. However, the event of maximal knee flexion is not the only characteristic distinguishing one-leg hop perfor-mances between ACL-deficient individuals from matched controls. Our observations agree with findings from Godwin et al. (2009) where signs of movement deficiencies or compensations might be missed if focusing only on peaks or discrete biomechanical variables. Our results (Figs. 1 and 3) also indicate that using the uninjured ACL_{PT} leg as a ref-erence for comparisons (i.e., normative data) provides rather different results than when contrasted against the non-dominant leg of matched controls. For instance, the noted discrepancies during take-off and landing in flexion/extension between ACL_{PT} and CTRL (Fig. 1) were

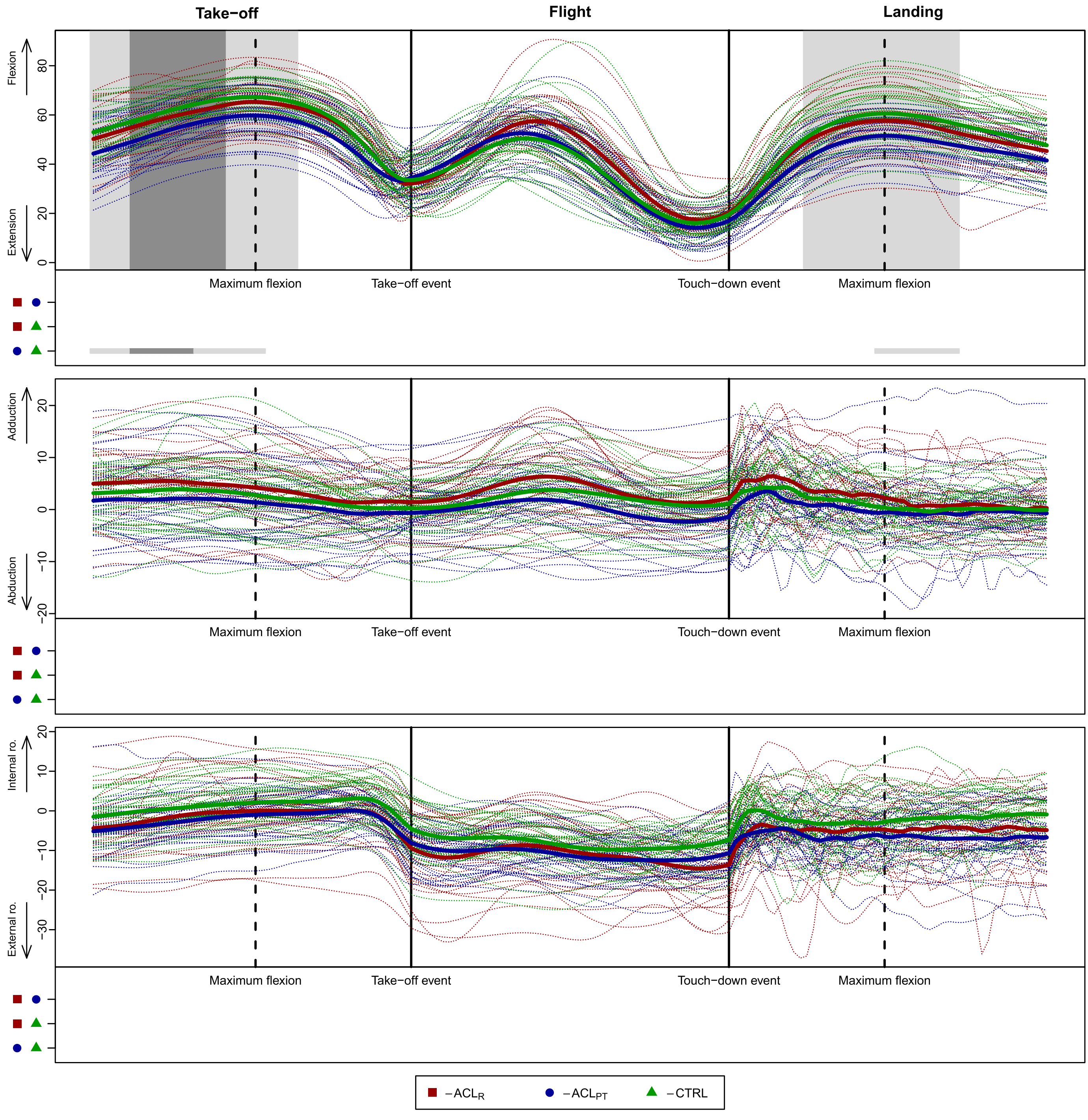

Fig. 1. Between-group comparisons of knee angles (°) in the sagittal (top row), coronal (middle row), and transverse (bottom row) planes of motion during the take-off (left), flight (center), and landing (right) phases of one-leg hops performed using the injured leg in ACL$_R$ and ACL$_{PT}$ and non-dominant leg in CTRL. The bold solid lines correspond to group means and the dashed lines represent individual curves (ACL$_R$ in red, ACL$_{PT}$ in blue, and CTRL in green). The vertical lines indicate the events used for landmark registration. The gray areas within the plots indicate significant between-group differences detected using ITP-based ANOVAs. Results from the pairwise group comparisons are underlined in gray in the panels below the plots, with the pairs indicated by the color-coded symbols (ACL$_R$ by red rectangles, ACL$_{PT}$ by blue circles, and CTRL by green triangles). Significant differences at $P < 0.05$ and $P < 0.01$ are represented in light and dark gray, respectively, with Bonferroni correction applied to account for the number of planes analyzed. ACL$_R$, ACL-ruptured subjects treated surgically with reconstruction and physiotherapy; ACL$_{PT}$, ACL-ruptured subjects treated conservatively with physiotherapy only; CTRL, knee-healthy controls. Please refer to the online version of this article to view this figure in color (recommended).

not present for between-leg comparisons in ACL$_{PT}$ (Fig. 3). Here, the time of maximal knee flexion was not able to detect differing movement patterns between injured and non-injured legs and was not sensitive in identifying movement disparities. Therefore, contrary to prior conclusions (Petschnig and Baron, 2009; van der Harst et al., 2007), our analyses caution against using the uninvolved leg as a normative reference in ACL-injured individuals as this approach may obscure functional impairments.

Considering the novelty of employing ITP on clinically relevant data, we piloted various approaches prior to selecting the most suitable for data reporting and interpretation. The ITP-based ANOVA was also applied during preliminary analyses to the angular velocity- and acceleration-

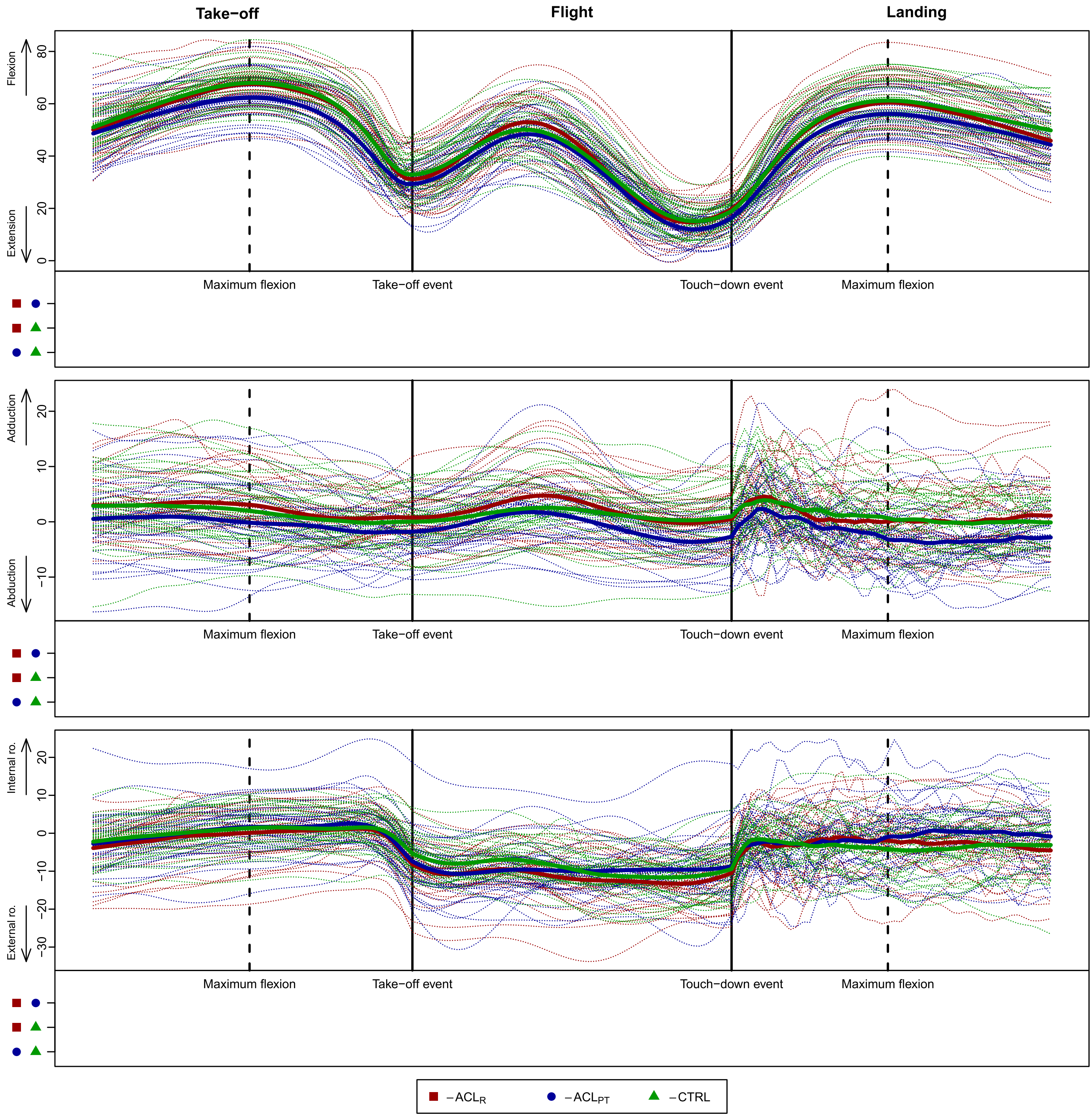

Fig. 2. Between-group comparisons of knee angles (°) in the sagittal (top row), coronal (middle row), and transverse (bottom row) planes of motion during the take-off (left), flight (center), and landing (right) phases of one-leg hops performed using the non-injured leg in ACL$_R$ and ACL$_{PT}$ and dominant leg in CTRL. The bold solid lines correspond to group means and the dashed lines represent individual curves (ACL$_R$ in red, ACL$_{PT}$ in blue, and CTRL in green). The vertical lines indicate the events used for landmark registration. The gray areas within the plots indicate significant between-group differences detected using ITP-based ANOVAs. Results from the pairwise group comparisons are underlined in gray in the panels below the plots, with the pairs indicated by the color-coded symbols (ACL$_R$ by red rectangles, ACL$_{PT}$ by blue circles, and CTRL by green triangles). Significant differences at $P < 0.05$ are represented in light gray (none present), with Bonferroni correction applied to account for the number of planes analyzed. ACL$_R$, ACL-ruptured subjects treated surgically with reconstruction and physiotherapy; ACL$_{PT}$, ACL-ruptured subjects treated conservatively with physiotherapy only; CTRL, knee-healthy controls. Please refer to the online version of this article to view this figure in color (recommended).

time curves. However, analysis of knee-joint position was the most effective at detecting deficits and provided the most practically relevant information. Inherently, observing angular displacements during functional activities is more viable in clinical environments than estimating velocities and accelerations. To complement our analyses, FDA was also applied on unaligned data. In such analysis, differences in movement speed are further emphasized, with the timing of peak knee flexion happening at different time points, for example. Conversely, analysis on aligned data focuses more on the amplitudes of the kinematic curves rather than their temporal characteristics. Again, comparable outcomes were obtained, emphasizing the presence of the between- and within-group differences detected herein, strengthening the rigor of the proposed statistical approach. However, it should be noted that our analysis did not take into account the BMI of individuals when

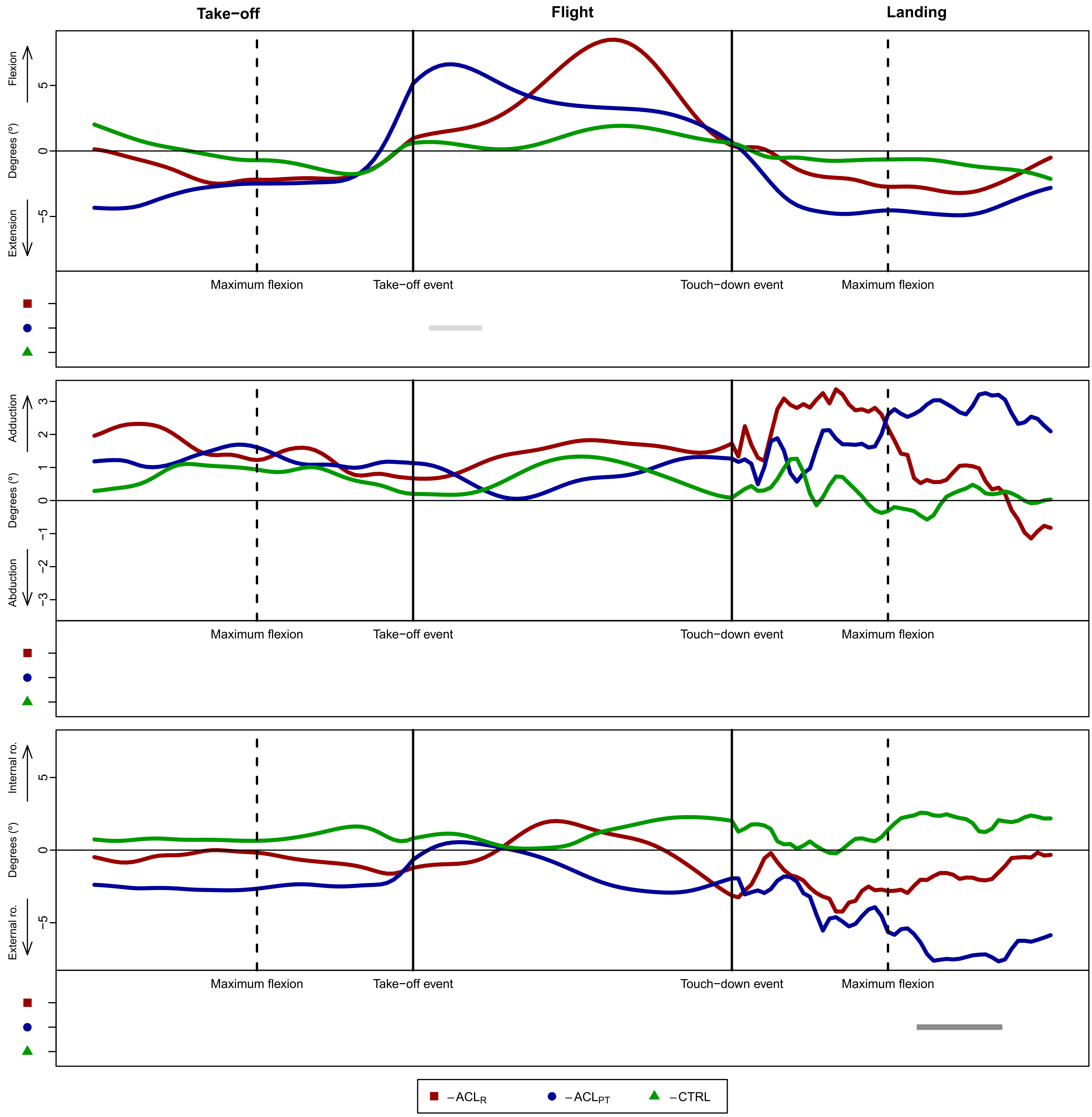

Fig. 3. Within-group comparisons of knee angles (°) in the sagittal (top row), coronal (middle row), and transverse (bottom row) planes of motion during the take-off (left), flight (center), and landing (right) phases of one-leg hops. The bold solid lines correspond to group means of between-leg differences (i.e., injured minus non-injured in ACL$_R$ and ACL$_{PT}$, and non-dominant minus dominant leg in CTRL). The vertical lines indicate the events used for landmark registration. For clarity, no individual curves are presented. Results from the ITP-based paired *t*-tests are underlined in gray in the panels below the plots, with the pairs indicated by the color-coded symbols (ACL$_R$ by red rectangles, ACL$_{PT}$ by blue circles, and CTRL by green triangles). Significant differences at $P < 0.05$ are represented in light gray, with Bonferroni correction applied to account for the number of planes analyzed. ACL$_R$, ACL-ruptured subjects treated surgically with reconstruction and physiotherapy; ACL$_{PT}$, ACL-ruptured subjects treated conservatively with physiotherapy only; CTRL, knee-healthy controls. Please refer to the online version of this article to view this figure in color (recommended).

investigating differences in knee kinematics between groups during the one-leg hop. Although there is conflicting evidence (Ballal et al., 2013; Hettrich et al., 2013; Park et al., 2013), high BMI has been associated with an increased risk for ACL injury (Myer et al., 2011; Uhorchak et al., 2003), as well as poorer outcomes and lower activity levels following ACL reconstruction (Dunn and Spindler, 2010; Griffith et al., 2013; Heijne et al., 2009; Kowalchuk et al., 2009). In non-injured populations, individuals with a higher percentage of body fat tend to hop shorter distances than leaner individuals (Gaunt and Curd, 2001; Yusof et al., 2013) and those with higher BMI have poorer postural control (Ku et al., 2012). Even though the significant difference in BMI between CTRL and ACL groups in our study did not lead to significant differences in one-leg hop distances (Table 1), the higher BMI values could increase knee loading and bring about injurious movement patterns (Myer et al., 2011). The significantly greater BMI in our ACL groups compared to controls is of general concern and should be addressed clinically. With future refinement of the presented statistical approach, it would be possible to account for relevant covariates, including BMI, leading to even more rigorous and comprehensive results.

It was a conscious choice to focus on knee-joint angles as a first step in using ITP-based ANOVA on kinematic data despite being a potential limitation. However, this statistical technique can be used on other continuous data series, including other joint kinematic, kinetic, and electromyographic data. Recently, Roos et al. (2014) reported lower center of mass velocities prior to one-leg hop landings in ACL_{PT} compared to ACL_R, with both significantly differing from CRTL. Applying FDA to center of mass displacements and velocities could complement their analyses, describing differences in movement not just at landing. Identifying the onset of such discrepancies might aid in further understanding the underlying mechanisms associated with impaired motion, and highlight the most challenging sections of one-leg hops, or other tasks.

5. Conclusions

Our implementation of FDA on knee-joint kinematic curves of one-leg hops highlights discrepancies in movement patterns persisting ~20 years post-ACL ruptures, only clearly identified herein when treated conservatively. Compared against traditional statistics, our analytical approach has the advantage of considering the *entire time-varying structure* of kinematic data, identifying relative time intervals in which compromised knee movement patterns are evident, exerting sound control of probabilities of accepting false-positive intervals, and providing a more comprehensive and detailed description of human motion. Applying FDA to movements like the one-leg hop is feasible and, if applied earlier in the rehabilitation process, could guide clinical decision-making by emphasizing deficiencies throughout the duration of any challenging task. With future refinement, it would be possible to account for numerous covariates, such as age, sex, and body mass index. Embracing such an approach could further explain the persistence of dysfunctions in previously injured individuals, assist in analyzing a range of functional movements, and guide rehabilitation programs while informing important clinical decisions, such as return to sport. The analytical approach presented provides a sound template for research in applied biomechanics.

Conflict of interest statement

The authors have no relevant conflict of interest to declare.

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Appendix A. Supplementary data

Supplementary data to this article can be found online.

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