2020

KNEE JOINT LOADING FOLLOWING ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION: LINK TO PATIENT REPORTED OUTCOMES AND A NOVEL METHOD TO MONITOR WITH WEARABLE SENSORS

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Digital Object Identifier: https://doi.org/10.13023/etd.2020.093

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KNEE JOINT LOADING FOLLOWING ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION: LINK TO PATIENT REPORTED OUTCOMES AND A NOVEL METHOD TO MONITOR WITH WEARABLE SENSORS

THESIS

A thesis submitted in partial fulfillment of the
requirements for the degree of Master of Science in the
College of Education
at the University of Kentucky

By

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2020

ABSTRACT OF THESIS

KNEE JOINT LOADING FOLLOWING ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION: LINK TO PATIENT REPORTED OUTCOMES AND A NOVEL METHOD TO MONITOR WITH WEARABLE SENSORS

Recovery from anterior cruciate ligament reconstruction (ACLR) commonly results in undesirable physical and patient-reported outcomes (PROs). Identification of modifiable factors such as knee contact force (KCF) early in rehabilitation that can improve these outcomes is important due to the rapid decrease in function, quality of life, and joint health in this population. Additionally, if noninvasive measurement of KCFs outside of a traditional laboratory were possible, clinicians could optimize patient treatment with personalized care. Therefore, there are two primary aims to this thesis: 1) quantify the link between KCF and PROs which measure pain, ability to perform activities of daily living, and quality of life 6 months after ACLR; and 2) develop a novel method to monitor KCF outside the laboratory using unobtrusive wearable sensors. To address the first aim, eighty subjects were enrolled six months following ACLR. Patient-reported quality of life, ability to perform activities of daily living, and pain were evaluated with the KOOS QOL, ADL, and Pain subscales, respectively. A musculoskeletal model was utilized to estimate peak KCF. Subjects with scores above the patient acceptable symptom state (PASS) threshold for the KOOS QOL and ADL demonstrated greater ACLR limb peak KCF ($p = 0.001$ and $p = 0.017$, respectively), which was not found with KOOS Pain-dichotomized groups ($p = 0.079$). To address the second aim, nine healthy subjects walked at a wide range of speeds on an instrumented treadmill. Thirteen insole force features were calculated as potential predictors of peak KCF and KCF impulse per step, estimated with musculoskeletal modeling. Prediction error was calculated as 10-fold cross validated median symmetric accuracy. Pearson product-moment correlation coefficients defined the relationship between variable pairs. Models developed per-limb demonstrated lower prediction error (KCF impulse: 2.19%; peak KCF: 3.50%) than those developed per-subject (KCF impulse: 3.40%; peak KCF: 6.47%). A number of insole features were associated with peak KCF (7 strong, 4 moderate), but not KCF impulse (all negligible). The findings from the first aim demonstrate that subjects with poor quality of life or ability to complete everyday activities underload their knee, possibly accelerating their path to osteoarthritis development. The findings from the second aim suggest that KCFs can be monitored with force-sensing insoles.

KEYWORDS: Knee Contact Force, Anterior Cruciate Ligament, Patient Reported Outcomes, Wearable Sensor, Gait.

Alex Spencer

04/09/2020

Date

KNEE JOINT LOADING FOLLOWING ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION: LINK TO PATIENT REPORTED OUTCOMES AND A NOVEL METHOD TO MONITOR WITH WEARABLE SENSORS

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DEDICATION

To my parents – All of my accomplishments, including this project, are fueled by your never-ending love and support. I have been astoundingly lucky in many ways through my life, but to be your son is everything to me and I do not take a day with you both for granted. To my brother Ryan – your strong moral values, sense of humor, unconditional loyalty to our family and respect to others, and feverish work ethic are just a few traits that I strive to uphold as I follow in your footsteps. I look forward to your weekly updates and cannot wait to see the positive impact you have on this world. To my wife Kylie – your willingness to love, challenge, and ground me over the past few years are just a few small reasons why we have accomplished so much, been so happy, and set ourselves up for a fantastic life ahead. You were there for every step of this project and much more. I can't wait to see what is next. Finally, to Leo – we all know you're really in charge. There is no way you could go without mention. Schnood up.

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ABBREVIATIONS

ACLR – Anterior cruciate ligament reconstruction

PRO – Patient-reported outcome

KOOS – Knee Injury and Osteoarthritis Outcome Score

QOL – Quality of life

ADL – Activities of daily living

PASS – Patient acceptable symptom state

KCF – Knee contact force

ACL – Anterior cruciate ligament

BPTB – Bone-patellar tendon-bone

IP – Impact peak

WAC – Weight acceptance

Prop. – Propulsive

DFT – Discrete Fourier transform

PF – Pseudo-Frequency

BW – Bodyweight

MSA – Median symmetric accuracy

LR – Loading rate

EMG – Electromyography

CHAPTER 1. INTRODUCTION

Over a quarter million anterior cruciate ligament (ACL) ruptures occur in the United States annually [1]. Anterior cruciate ligament reconstruction (ACLR) commonly results in undesirable physical and patient-reported outcomes (PROs). Ensuing quadriceps dysfunction following ACLR [2-12] results in altered biomechanics during walking, running, and stair ambulation [3, 5, 7, 8, 10, 13-18]. Further, as these mechanical changes outpace cartilage adaptation, the consequence is reduced overall knee function [19-22], decreased quality of life [23-26], and rapid osteoarthritis progression [27-30] for 51% of these patients [28]. Identification of modifiable factors early in the rehabilitation process which can improve these outcomes is vital.

One modifiable factor that may be associated with these outcomes is knee joint loading. Patients following ACLR commonly offload their injured limb during gait [31-33], which is known to have detrimental effects on cartilage health [13, 34]. Further, the cyclical loading frequency and overall volume from a common activity such as walking considerably alters the chronic mechanical loading environment of the cartilage, thus accelerating degeneration [35-37]. Consequently, those with lower knee contact force (KCF) demonstrate reduced cartilage thickness and increased risk of early osteoarthritis development [33, 38]. Patients may sense these negative changes in cartilage health, resulting in lower perceived knee function and overall quality of life.

The ability to monitor knee joint loading outside of a traditional laboratory could benefit ACLR patients. In addition to researchers identifying links between joint loading and tissue health, interventions have been implemented in rehabilitation protocols with the intent to modify knee loading through both gait retraining and external support devices

[39-42]. The results of these studies could be built upon through the measurement of their efficacy outside of the lab. The ability to measure and monitor knee contact force outside the lab could help both researchers and clinicians improve ACLR patient outcomes through enhanced understanding and targeted treatment.

The technological advancement of wearable sensors has allowed internal load estimations to transition into everyday life. Among the array of available sensor options, force-sensing insoles that estimate the normal component of foot-shoe contact force have received attention from researchers and clinicians due to their ease of use, unobtrusiveness, and potential to help answer important research questions. In particular, instrumented insoles have proven to be a valid and reliable tool to estimate foot contact forces during various activities including walking [43-45]. Researchers have used these sensors to predict knee moments [46] and further understand functional performance deficiency [47, 48]. However, no studies to date have used force-sensing insoles to estimate musculoskeletal-model generated KCF during walking.

Therefore, there are two primary aims to this thesis. The first aim is to quantify the link between KCF and PROs which measure pain, ability to perform activities of daily living, and quality of life 6 months after ACLR. Specifically, this study evaluates the difference in peak KCF and KCF symmetry during walking for subjects above and below the patient acceptable symptom state (PASS) threshold for these PROs. We hypothesize that those subjects with scores above the PASS threshold for each of the Knee Injury and Osteoarthritis Outcome Score (KOOS) subscales will walk with greater peak KCF on their ACLR limb and greater peak KCF symmetry than those with scores below the PASS threshold. The second aim is to develop a novel method to monitor KCF outside the laboratory using unobtrusive wearable sensors. Specifically, this study compares models with varying levels of specificity (per-limb and per-subject) in their ability to estimate peak KCF and KCF impulse per step with data from force-sensing insoles across a range of speeds. We hypothesized that models developed per-limb would result in the lowest error. The second purpose was to measure the relationship between individual foot contact force data features and KCF metrics (peak and impulse). We hypothesized that all metrics would be either moderately or strongly correlated with both KCF metrics.

CHAPTER 2. PATIENT-REPORTED OUTCOMES AND KNEE CONTACT FORCE 6 MONTHS AFTER ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION

2.1 Abstract

Background: Recovery from anterior cruciate ligament reconstruction (ACLR) commonly results in undesirable physical and patient-reported outcomes. Identification of modifiable factors such as knee contact force (KCF) early in the rehabilitation process that can improve these outcomes is important due to the rapid decrease in function, quality of life, and joint health in this population.

Purpose: The purpose of this study is to evaluate the differences in peak KCF and KCF symmetry during walking for subjects who score above versus below previously established Patient Acceptable Symptom State (PASS) thresholds for three Knee Injury and Osteoarthritis Outcome Score (KOOS) subscales that assess knee-related quality of life (QOL), ability to perform activities of daily living (ADL), and pain 6 months after ACLR. We hypothesized that subjects with scores above the PASS thresholds would demonstrate greater peak KCF and KCF symmetry than those with scores below the PASS thresholds.

Methods: Eighty subjects (37 F; age: 20.0 ± 5.9 years; height: 174.8 ± 9.7 cm; mass: 72.0 ± 12.4 kg; graft type: 65 bone-patellar tendon bone, 15 hamstring semitendinosus) participated in this study six months following ACLR. Patient-reported quality of life, ability to perform activities of daily living, and pain were evaluated with the KOOS QOL, ADL, and Pain subscales, respectively. A musculoskeletal model was utilized to estimate peak KCF per limb during walking (1.5 m/s) using OpenSim.

Results: Subjects with scores above the PASS threshold for the KOOS QOL and ADL demonstrated greater ACLR limb peak KCF ($p = 0.001$ and $p = 0.017$, respectively) than those below the threshold. However, subjects with KOOS Pain scores above and below the PASS threshold did not demonstrate a significant difference in peak KCF ($p = 0.079$). No differences existed in KCF symmetry for any subscale-dichotomized group pair.

Discussion: Similar knee joint loading between subjects with high and low pain scores suggests that the effect of pain on joint loading has subsided by 6 months post-ACLR. However, the decreased loading in subjects with low QOL and ADL suggests that those with poor quality of life or ability to complete everyday activities underload their knee, possibly accelerating their path to osteoarthritis development.

2.2 Introduction

While approximately 100,000 patients receive anterior cruciate ligament reconstruction (ACLR) annually in the United States with the intent to return to full activity without pain or other limitations [1], the recovery process from this procedure commonly results in undesirable physical and patient-reported outcomes (PROs). Acute and persistent deficits in muscle function surrounding the knee [2-12] result in altered biomechanics during common activities such as walking, running, and stair ambulation [3, 5, 7, 8, 10, 13-18]. Further, as cartilage degrades due to its inability to rapidly adapt to these mechanical changes, its response to load is simultaneously altered [29, 36, 49-53]. The consequences of these acute changes in applied mechanical loading and the tissue's response to load is a rapid progression of osteoarthritis [27-30] for 51% of these patients [28], reduced overall knee function [19-22], and decreased quality of life [23-26]. Identification of modifiable factors early in the rehabilitation process which can improve these outcomes is important due to the rapid decrease in function, joint health, and quality of life.

The Knee Injury and Osteoarthritis Outcome Score (KOOS) is an instrument which is commonly used to quantitatively assess ACLR patients' perception of their knee health and associated problems [54-58]. The KOOS contains subscales that evaluate patients' quality of life (QOL), ability to complete activities of daily living (ADL), and pain [59]. Patient acceptable symptom state (PASS) thresholds have been developed for each of these subscales which provide reference values for which patient outcomes are deemed acceptable [60]. PASS rates for each of these three subscales range from only 40-57% at a 1-year follow-up, and 45-69% at a 2-year follow-up [61, 62]. A challenge remains to identify objective measurements of modifiable factors that rehabilitation protocols could leverage to guide ACLR patients above the PASS threshold for these key PROs.

One modifiable factor that may be associated with PROs is knee joint loading. Patients following ACLR commonly offload their injured limb during gait [31-33], which is known to have detrimental effects on cartilage health due to the acutely altered mechanical loading environment [13, 34]. Further, the cyclical loading frequency and overall volume from a common activity such as walking would considerably alter the mechanical loading environment of the cartilage, thus accelerating degeneration [35-37]. Consequently, those with lower knee contact force (KCF) demonstrate reduced cartilage thickness and increased risk of early osteoarthritis development [33, 38]. Patients may sense these negative changes in cartilage health, resulting in lower perceived knee function and overall quality of life.

Therefore, the purpose of this study is to evaluate the difference in peak KCF and KCF symmetry during walking for subjects above and below the PASS threshold for PROs which measure pain, ability to perform activities of daily living, and quality of life 6 months after ACLR. We hypothesize that those subjects with scores above the PASS threshold for each of these KOOS subscales will walk with greater peak KCF on their ACLR limb and greater peak KCF symmetry than those with scores below the PASS threshold.

2.3 Methods

All recruitment methods and study procedures were approved by the University of Kentucky's Institutional Review Board. Eighty subjects who underwent primary ACLR provided their written informed consent and were enrolled in the study. One of two surgeons from the same orthopedic practice performed all anterior cruciate ligament (ACL) reconstructions. Subjects completed all testing 6 months following surgery at which time they were cleared to begin return to sport drills by an orthopedic surgeon and had completed a standardized rehabilitation protocol. Subjects were excluded if they were outside the age range of 16-40 years, reported less than a 5/10 Tegner activity level, injured their limb more than 6 months prior to surgery, had a previous knee injury on the involved limb beyond meniscectomy, underwent a previous surgery on the contralateral limb, or the injury to the involved limb included a total knee dislocation.

The KOOS is a knee-specific questionnaire that is comprised of 42 items divided into five subscales: Knee- Related Quality of Life (QOL; 4 items), Activities of Daily Living (ADL; 17 items), Pain (9 items), Sport and Recreational Function (Sport/Rec; 5 items), and Symptoms (7 items). The scores for each of these subscales range from 0 (worst) to 100 (best). The American-English version of this instrument has proven to be valid and reliable for ACLR patients ages 18-46 [63]. The KOOS was administered to all subjects, and the following subscales were analyzed: QOL, ADL, and Pain. PASS thresholds have been developed for each of these subscales (QOL: 62.5, ADL: 100.0, Pain: 88.9) and were used to dichotomize groups into PASS-Y (equal to or above PASS threshold) and PASS-N (below PASS threshold) [60].

The motion capture protocol was consistent with previously published methods [5]. Fifty-two retroreflective markers (25 for tracking clusters and 27 on anatomical landmarks) were adhered to each subject. All subjects wore New Balance WR662 shoes (New Balance, Brighton, MA). Marker locations were collected with a 10 camera motion capture system at 200 Hz (Motion Analysis, Santa Rosa, CA) while kinetic data was simultaneously collected at 1200 Hz from a dual-belt instrumented treadmill (Bertec Corporation, Columbus, OH) as subjects walked at 1.5 meters per second.

Data were post-processed in Visual 3D (C-motion, Germantown, MD). Marker position and force data were filtered with 4th order low-pass Butterworth filters at 8 and 35 Hz, respectively. The functional joint center was determined with a spherical fitting technique from a hip motion trial [64]. Joint angles and moments were calculated with inverse kinematics and inverse dynamics, respectively. These data were exported for use in OpenSim.

Subject-specific simulations for each subject were generated in OpenSim (version 4.0) [65]. The three-dimensional musculoskeletal model consisted of 92 muscles and 23 degrees of freedom (DOF). The knee was restricted as a 1 DOF hinge joint (flexion/extension). Models were scaled by height, weight, and segment lengths (from marker positions during static trial). Static optimization was used to estimate muscle forces [66]. KCFs were calculated using these muscle force estimations with the Joint Reaction Analysis tool and expressed in the tibial reference frame [67]. Peak KCF was computed as the maximum of the resultant KCF vector magnitude per stance phase. Peak KCF symmetry was calculated using the average of all peak KCF across stance phases from each limb according to Equation 2.1.

$$Symmetry\ [\%] = 100 \times \left(1 - \frac{Injured - Noninjured}{Noninjured}\right) \qquad \text{Eq. 2.1}$$

Statistical analyses were performed using IBM SPSS Statistics Version 24 (IBM Corp., Armonk, NY). To determine if either sex or graft type influenced patient-reported outcomes, chi-square and/or Fisher Exact tests were used to compare the number of patients above and below the KOOS Pain, ADL, and QOL scores. If a significant difference was found between either sex or graft type, an ANCOVA was then used to compare KCF between PASS groups with the significant variable (either sex or graft type) included as a covariate. If no differences were found in PASS results between sexes or graft types, KCF was compared between groups using one-way independent-samples t-tests ($\alpha = 0.025$).

2.4 Results

A total of 80 ACLR subjects took part in this study 6 months following surgery [Table 2.1]. Neither sex nor graft type were significantly associated with PRO status for either of the three subscales. KOOS subscale scores for the entire cohort were as follows (mean (95% CI)): KOOS QOL 62.6 (58.5-66.7); KOOS ADL 97.7 (96.9-98.5); KOOS Pain 91.6 (89.9-93.2). Mean KCF for the PASS-Y groups were greater for each of the subscale groups analyzed, but the only statistically significant differences present were for the KOOS QOL and KOOS ADL [Table 2.2]. Knee contact force symmetry was not different between the PASS-Y and PASS-N groups for any of the 3 subscales.

Table 2.1 Demographic Data

		N	Score	Age [years]	Mass [kg]	Graft
All		80 (37 F, 43 M)	-	20.0 (18.8-21.3)	72.0 (69.3-74.7)	65 BPTB 15 Hamstring
KOOS QOL	PASS-Y	46 (19 F, 27 M)	74.9 (72.3-77.4)	18.4 (17.4-19.4)	70.9 (68.4-73.4)	39 BPTB 7 Hamstring
	PASS-N	34 (18 F, 16 M)	46.0 (43.2-48.7)	22.0 (20.5-23.5)	76.1 (71.7-80.4)	26 BPTB 8 Hamstring
KOOS ADL	PASS-Y	39 (16 F, 23 M)	100	18.1 (17.3-19.0)	72.6 (70.0-75.1)	33 BPTB 6 Hamstring
	PASS-N	39 (20 F, 19 M)	95.4 (94.6-96.2)	21.7 (20.2-23.3)	74.3 (70.1-78.5)	31 BPTB 8 Hamstring
KOOS Pain	PASS-Y	48 (19 F, 29 M)	96.4 (95.8-97.0)	18.4 (17.5-19.3)	72.6 (69.8-75.4)	40 BPTB 8 Hamstring
	PASS-N	30 (17 F, 13 M)	83.8 (83.7-84.9)	22.3 (20.7-23.9)	74.7 (70.4-79.0)	24 BPTB 6 Hamstring

Note: Two subjects did not complete the KOOS ADL and KOOS Pain subscales. Ranges are 95% confidence intervals. BPTB = bone-patellar tendon-bone

Table 2.2 Peak KCF of PASS-Y and PASS-N groups

		PASS-Y	PASS-N	p-value
KOOS QOL	**Injured [BW]**	2.28 (2.17-2.40)	1.99 (1.93-2.05)	**0.001****
	Symmetry [%]	94.8 (91.6-98.1)	94.9 (92.7-97.0)	0.490
KOOS ADL	**Injured [BW]**	2.27 (2.15-2.39)	2.05 (1.98-2.12)	**0.017***
	Symmetry [%]	95.6 (92.6-98.7)	94.0 (91.3-96.6)	0.291
KOOS Pain	**Injured [BW]**	2.21 (2.11-2.32)	2.07 (1.98-2.16)	0.079
	Symmetry [%]	94.7 (91.8-97.5)	95.0 (92.1-97.9)	0.460

Note: Ranges are 95% confidence intervals.

2.5 Discussion

The purpose of this study was to evaluate the differences in peak KCF and KCF symmetry during walking for subjects with scores above and below PASS thresholds for the KOOS Pain, QOL, and ADL subscales. We hypothesized that subjects with scores above the PASS thresholds for each subscale would walk with greater ACLR limb peak KCF and greater KCF symmetry than those below the threshold. We found that subjects with KOOS QOL and ADL scores above the PASS threshold demonstrated greater ACLR limb peak KCF than those below the PASS threshold, but KCF symmetry was similar between groups. There were no differences in peak KCF between the KOOS Pain PASS-Y and PASS-N groups for the ACLR limb, or KCF symmetry for any subscale. These results indicate an offloading strategy that is linked with poor knee-related quality of life and ability to perform activities of daily living.

The results suggest that those with low knee-related quality of life and diminished ability to perform activities of daily living are prone to offload their limb(s). This connection between PROs and joint loading is critical because underloading has been found to lead to significant joint degeneration [33, 35, 68-70]. Chaudhari et al. speculates that a detrimental positive feedback loop is initiated by acute kinematic changes which alter the mechanical environment of the cartilage, leading to biological changes if the tissue is unable to adapt [36]. Further, the transition from osteoarthritis initiation to progression is marked by an increased cartilage sensitivity to compressive loading which accelerates joint degeneration [71, 72]. Therefore, we speculate that those with low quality of life and ability to complete activities of daily living may be prone to enter this positive feedback loop of cartilage degeneration due to their tendency to underload.

The lack of difference in peak KCF between subjects in the KOOS Pain PASS-Y and PASS-N groups indicates that pain may not be a significant factor in joint loading alterations 6 months following ACLR. Work by Azus et al. supports this idea, as they identified significant correlations between walking kinetics and KOOS Pain scores at baseline, but not 6 months following surgery [73]. Additionally, they found substantial improvement in pain through the first 6 months post-surgery and relative stability from 6 months to 1 year post-surgery [73]. Alternatively, we speculate that acute kinematic alterations soon after surgery due to pain may accelerate cartilage degeneration, leading to poor QOL and ADL 6 months post-surgery. Combined, these findings indicate that pain early in the rehabilitation process is important to manage, but its effect on joint loading and associated consequences decreases by 6 months post-surgery.

The findings from this study suggest that joint loading symmetry is not indicative of quality of life or ability to complete activities of daily living. Metrics that quantify loading symmetry between ACLR and contralateral limbs are often used as a measure of movement quality [47, 48, 74-76], but these data suggest a bilateral compensation strategy which maintains joint loading symmetry. We speculate that a proximal shift of muscle contributions within the PASS-N group may be driving the decreased KCF during the propulsive phase of gait (Figure 2.1), although this would need to be supported with further analyses. These results carry implications for researchers and clinicians, as the use of biofeedback or load monitoring via wearable sensors is becoming more popular [46, 77, 78]. We suggest that these results be considered in the interpretation of these device outputs or biofeedback metrics. While joint loading symmetry may be a valid marker of movement quality, it is not a direct measure of patient well-being or overall rehabilitation success.

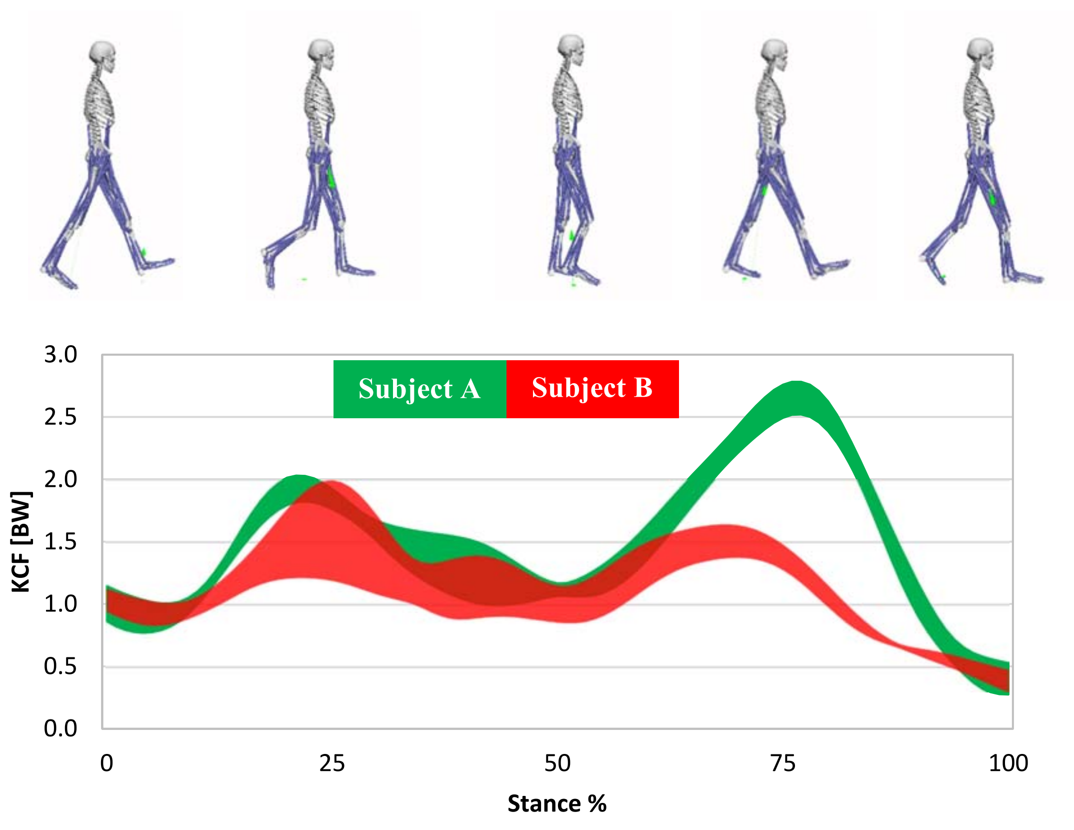


Figure 2.1 Knee contact force curves (Mean +/- SD from 5 trials) for the ACLR limb of two representative subjects. Subject A (green) achieved PASS-Y scores for the KOOS QOL, ADL, and Pain subscales, while Subject B (red) had PASS-N scores for KOOS QOL and ADL, and a PASS-Y score for KOOS Pain.

This study has a number of limitations. First, we only assessed peak loading during a data collection on a single day. While this peak load per step may define the cyclic loading environment of the tissue, it does not indicate the cumulative load or damage experienced by the tissue [79, 80]. A longitudinal study outside the laboratory using wearable sensors to monitor knee loading throughout rehabilitation could provide further clarity. A generic musculoskeletal model with muscle forces estimated via static optimization was used to estimate KCF per step. While static optimization has been validated during walking [66], ACLR patients have demonstrated abnormal muscle activation and co-contraction [81-83] which may have been better estimated with a different method of muscle force estimation. While other methods may achieve contact forces more similar to those measured directly (such as with instrumented prostheses), our intent was to establish the relationship with the simplest model possible which would answer the question.

In conclusion, this study evaluated the difference in peak KCF during walking for ACLR patients with scores above and below PASS thresholds for 3 KOOS subscales that evaluate knee-related quality of life, ability to complete activities of daily living, and pain. We found that subjects with scores above the KOOS QOL and ADL PASS thresholds demonstrated greater peak KCF than those below the PASS thresholds. However, there were no differences in KCF between the KOOS Pain PASS-Y and PASS-N groups, or KCF symmetry in any of the subscales. These results indicate a connection between how ACLR patients perceive their rehabilitation success and how they load their joints, providing additional information for clinical intervention to mitigate the risk of early-onset initiation and progression of OA.

CHAPTER 3. MONITORING KNEE CONTACT FORCE WITH FORCE-SENSING INSOLES

3.1 Abstract

Background: There are numerous applications for monitoring knee contact force (KCF) in clinical environments and activities of daily living. However, the ability to estimate these forces is currently restricted to a traditional laboratory setting. A need remains to develop wearable sensor algorithms which can be used to estimate KCF outside of the lab.

Purpose: The first purpose of this study was to develop models with varying levels of specificity (per-limb and per-subject) that estimate peak KCF and KCF impulse per step with data from force-sensing insoles. We hypothesized that models developed per-limb would result in the lowest error. The second purpose was to quantify the relationship between individual foot contact force data features and KCF metrics (peak and impulse). We hypothesized that all insole force features would be either moderately or strongly correlated with both KCF metrics.

Methods: Nine healthy subjects (3F, age 27 ± 5 years, mass 74.8 ± 11.8 kg, height 1.74 ± 0.084 m) walked at a wide range of speeds (0.8-1.6 m/s) on an instrumented treadmill. Thirteen insole force features were calculated as potential predictors of peak KCF and KCF impulse per step, estimated with musculoskeletal modeling. Prediction error was calculated as 10-fold cross validated median symmetric accuracy. Pearson product-moment correlation coefficients defined the relationship between variable pairs.

Findings: Models developed per-limb demonstrated lower prediction error (KCF impulse: 2.19%; peak KCF: 3.50%) than those developed per-subject (KCF impulse: 3.40%; peak KCF: 6.47%). A number of insole features were moderately to strongly associated with

peak KCF (7 strong, 4 moderate, and 2 negligible), but not KCF impulse (all negligible) at a group level.

Discussion: These methods can be used to monitor KCFs with force-sensing insoles during flat surface walking outside of the lab. While models developed per-limb performed only moderately better within this healthy group, the error of per-subject models would likely increase as these methods extend to post-surgical patients with greater between-limb KCF differences. There are a number of insole features which can be used as surrogate measures to monitor peak KCF. These results carry promising implications for the estimation and monitoring of KCFs outside of a traditional laboratory with wearable sensors.

3.2 Introduction

Applications for internal tissue load measurement include the treatment of orthopedic injuries, management or avoidance of overuse injuries, and furthering the understanding of how chronic loading influences tissue health. Currently, these load measurements are limited to either invasive approaches (e.g. instrumented prostheses) or noninvasive estimations requiring significant instrumentation inside of a traditional laboratory setting (e.g. musculoskeletal modeling). The restriction of noninvasive methods to a laboratory setting limits their clinical applicability. If noninvasive measurement of internal tissue loads outside of a traditional laboratory were possible, rehabilitation specialists could optimize patient treatment with a precision medicine approach.

Numerous methods have been developed to measure or estimate KCF in vivo. The gold standard for these measurements is through the use of instrumented prostheses or other implantable devices [84-90]. While these methods provide direct load measurement, their invasive nature limits their applicability. Laboratory-based noninvasive methods include the estimation of intersegmental moments and forces with inverse dynamics [91-95], joint contact forces with musculoskeletal modeling [31, 96-115], or tissue stress and strain with finite element analysis [116-126]. While these methods provide substantial analytical depth, the laboratory-based instrumentation required limits their applicability in a clinic or in patients' activities of daily living.

Patients recovering from knee injury or surgery would benefit from the ability to monitor knee contact force (KCF) outside the laboratory. One potential avenue to accomplish this task is through the use of wearable sensors. In particular, force-sensing insoles that estimate the normal component of foot-shoe contact force have received attention due to their ease of use, unobtrusiveness, and potential to help answer important research questions [44, 46]. These sensors have proven to be a valid and reliable tool to estimate foot contact forces during common activities including walking [43-45]. While these sensors have been used to monitor rehabilitation progress in various ways [46-48], no studies to date have used force-sensing insoles to estimate musculoskeletal-model generated KCF during walking.

Two possible ways to monitor KCF with wearable sensors are through direct estimation or utilization of surrogate measures. Predictive models must be applicable to both healthy and injured populations. Post-surgical patients that demonstrate significant limb asymmetry may require different KCF prediction models for each limb. Alternatively, KCF prediction models developed per-subject may be sufficient for a healthy population. Utilizing surrogate measures of KCF metrics would provide a simpler but potentially less precise solution. The associations between insole force features and KCF metrics should be consistent among the population of interest so that any patient's KCF metrics could be monitored without calibration to patient-specific movement characteristics. Each of these approaches provides different advantages and must be considered within the population of interest.

The first purpose of this study was to develop models with varying levels of specificity (per-limb and per-subject) that estimate peak KCF and KCF impulse per step with data from force-sensing insoles across a range of speeds from a healthy population. We hypothesized that models developed per-limb would result in the lowest error. The second purpose was to measure the relationship between individual foot contact force data features and KCF metrics (peak and impulse). We hypothesized that all insole features would be either moderately or strongly correlated with both KCF metrics.

3.3 Methods

This study was approved by the University of Kentucky Institutional Review Board. Nine subjects (3F, age 27 ± 5 years, mass 74.8 ± 11.8 kg, height 1.74 ± 0.084 m) provided their written informed consent and were enrolled in the study. Only subjects meeting the following inclusion criteria were considered for the study: Tegner score of or above 4; 15-40 years of age; Body Mass Index 18-25 kg/m^2; participate in competitive sport or run at least 10 miles per week; and no history of movement impairment or lower extremity injury. Subjects were excluded if they had a history of previous surgeries or other conditions that may affect physical performance.

Fifty-two retroreflective markers were placed on each subject (25 as tracking clusters and 27 on anatomical landmarks). Anatomical markers included: sternum, left and right superior acromion processes, C7, left and right iliac crests, left and right greater trochanters, L5/S1, left and right medial and lateral femoral epicondyles, left and right medial and lateral tibial condyles, left and right medial and lateral malleoli, left and right first and fifth metatarsal heads, and left and right distal foot. The tracking markers included: 4-marker clusters attached to rigid plates on the left and right shanks and thighs, 3-marker clusters on each posterior shoe (lateral, distal, and proximal heel), left and right iliac crests, L5/S1, sternum, left and right superior acromion processes, C7, and markers on the anterior right thigh, shank, and foot to differentiate the right side from the left. Marker locations were collected at 200 Hz with a 12-camera motion capture system (Motion Analysis, Santa Rosa, CA) simultaneously with force plate data at 1200 Hz from a dual-belt instrumented treadmill (Bertec Corporation, Columbus, OH) as subjects walked at five different speeds (0.8, 1.0, 1.2, 1.4, and 1.6 m/s) for 60 seconds each. Marker position and force plate data were filtered with 4th order low-pass Butterworth filters at 8 and 35 Hz, respectively. Foot contact force data was collected from each condition using single sensor loadsol® insoles (Novel Electronics, St. Paul, MN, USA) at 100 Hz. All subjects wore New Balance WR662 running shoes (New Balance, Brighton, MA).

In order to sync the motion capture and foot contact force data, each trial began with a right-foot stomp while the treadmill was stopped followed by a controlled increase in speed until the condition speed was achieved. Force plate and insole force peaks corresponding to the stomp were semi-automatically identified and matched. The data was then scanned across a +/- 50ms window of these peaks to optimize the synchronization by maximizing the cross-correlation of the foot contact force (force-sensing insole) and ground reaction force (force plate) data. Stance intervals were defined with 20 N thresholds from the insole data.

Knee contact forces were estimated using OpenSim (version 4.0) [65]. The Gait2392 musculoskeletal model consists of 92 muscles and 23 degrees of freedom (DOF), with the knee restricted as a 1 DOF hinge joint (flexion/extension). Model weight, height, and segment lengths were scaled per subject from static trial marker positions. Muscle forces were estimated via static optimization [66], then used within the Joint Reaction Analysis tool to estimate KCF expressed in the tibial reference frame [67]. Peak KCF and KCF impulse were computed as the maximum and time integral of the resultant KCF vector magnitude per stance phase, respectively.

A total of 13 foot contact force features were extracted per step [Table 3.1]. Traditional features such as stance time, peak force, and loading rate were supplemented with non-traditional metrics that may generate deeper insight into subtle gait mechanics. Equations, illustrations, and descriptions for each of these features are provided in Appendix A.

Table 3.1 Features extracted from insole data per step

Time Domain – Traditional	**Time Domain – Other**	**(Pseudo) Frequency Domain**
Stance Time [s]	Skewness	DFT Max [Hz]
Peak Magnitude [BW]	Mid-Drop Peak Magnitude	Mean PF
Impulse [BW*s]	WAC Impulse [BW*s]	
Loading Rate [BW/s]	Prop. Impulse [BW*s]	
IP Magnitude [BW]	WAC/Prop. Impulse Symmetry	

Note: IP = impact peak; WAC = weight acceptance; Prop. = Propulsive; DFT = Discrete Fourier Transform; PF = Pseudo-Frequency; BW = bodyweight

Knee contact force prediction models were created per-limb and per-subject for comparison. All features were first z-score normalized to ensure zero-mean and unit variance. Individual predictors were chosen through a best subset selection method (test all possible predictor combinations, then select the one with the lowest error). To ensure generalizability of the predictive models, we utilized 10-fold cross validated linear regression stratified by walking speed. Median symmetric accuracy (MSA) was chosen as the evaluation metric, as it has been shown to produce unbiased and robust models while maintaining a translatable output (percent error) (Eq. 3.1) [127]. The overall method of KCF prediction is illustrated in Figure 3.1.

$$MSA = 100\left(e^{(M(|\ln(Q)|))} - 1\right) \quad \text{(Eq. 3.1)}$$

$$\text{where } Q = \frac{prediction}{observation} \text{ and } M \text{ is the median operator}$$

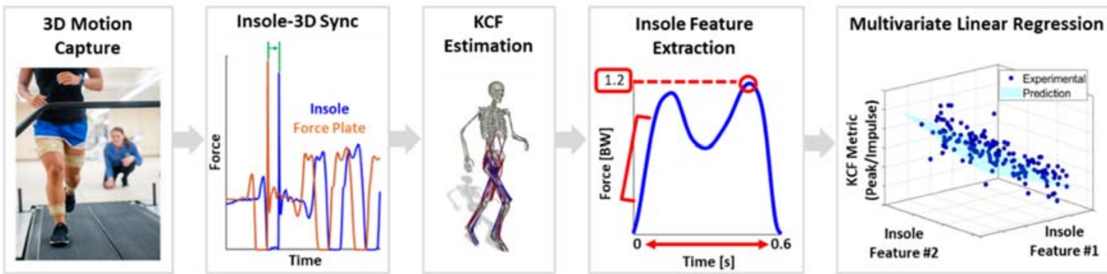


Figure 3.1 Overall method of KCF prediction with force-sensing insole data

Pearson product-moment correlation coefficients defined the relationship between variable pairs. The following ranges defined the strength of correlations: strong → $|r| \geq 0.7$; moderate → $0.5 \leq |r| < 0.7$; weak → $0.3 \leq |r| < 0.5$; negligible → $|r| < 0.3$. These correlation coefficients were calculated per subject per limb, then averaged across the group.

3.4 Results

The KCF prediction error was lower in models developed per-limb (2.19% for KCF Impulse and 3.50% for peak KCF) than those developed per-subject (3.40% for KCF Impulse and 6.47% for peak KCF) [Table 3.2]. Error was consistently lower for KCF Impulse than peak KCF. Finally, the models utilized a relatively low number of features, indicating good computational feasibility for these methods [Table 3.2].

Table 3.2 Performance of KCF prediction models

	KCF Impulse		Peak KCF	
	MSA [%]	**# Predictors**	**MSA [%]**	**# Predictors**
Per Limb	2.19 (1.65-2.54)	8.0 (7.0-8.8)	3.50 (2.78-5.09)	7.0 (5.3-8.0)
Per Subject	3.40 (2.87-4.24)	9.0 (7.0-9.0)	6.47 (5.07-11.06)	8.0 (7.0-9.0)

Note: MSA = median symmetric accuracy;
All values are reported as median (interquartile range)

The correlation analysis identified a number of insole features which are moderately to strongly associated with peak KCF (7 strong, 4 moderate, and 2 negligible) [Table 3.3]. All insole features were negligibly correlated with KCF impulse at a group level. However, a number of correlations on a per-limb basis were moderate to strong, as illustrated in Appendix B.

Table 3.3 Correlation coefficients between insole features and KCF metrics

		Peak KCF	**KCF Impulse**
Time Domain – Traditional	Stance Time [s]	-0.84 ± 0.05***	0.09 ± 0.48
	Peak [BW]	0.87 ± 0.09***	0.14 ± 0.40
	Impulse [BW*s]	-0.76 ± 0.07***	0.11 ± 0.48
	Loading Rate [BW/s]	0.80 ± 0.24***	0.11 ± 0.44
	IP Magnitude [BW]	0.62 ± 0.16**	0.09 ± 0.34
Time Domain – Other	Skewness	0.81 ± 0.12***	0.20 ± 0.41
	Mid-Drop Peak	-0.82 ± 0.12***	-0.10 ± 0.46
	WAC Imp [BW*s]	-0.05 ± 0.34	0.08 ± 0.19
	Prop Imp [BW*s]	-0.66 ± 0.14**	0.09 ± 0.42
	WAC/Prop Impulse	0.56 ± 0.17**	-0.02 ± 0.33
	Symmetry	-0.24 ± 0.44	-0.03 ± 0.25
(Pseudo) Frequency Domain	DFT Max [Hz]	0.63 ± 0.09**	-0.14 ± 0.42
	Mean PF	0.73 ± 0.12***	-0.14 ± 0.45

Note: Values are presented as mean ± standard deviation of all speed conditions per subject per limb.
*** = strong correlation (|r|≥0.7)
** = moderate correlation (0.5≤|r|<0.7)
* = weak correlation (0.3≤|r|<0.7)

3.5 Discussion

This study had two primary purposes: 1) develop models of varying specificity to estimate peak KCF and KCF impulse per step; and 2) quantify the relationship between KCF metrics and individual foot contact force data features. We found that models developed per-limb produced lower error than those developed per-subject. Additionally, we identified a number of insole features which are moderately to strongly associated with peak KCF (7 strong, 4 moderate, and 2 negligible). However, while all insole features were negligibly correlated with KCF impulse at a group level, there were individual strong and moderate correlations on a per-limb basis. These results demonstrate viability in monitoring KCFs outside of a traditional laboratory with wearable sensors.

The prediction error from the models developed in this study demonstrates that a single wearable sensor can produce accurate KCF estimates. These models typically utilized between five and nine predictors [Table 3.2], which demonstrates both computational efficiency and robust biomechanical representation. The collection of insole force features [Appendix A] was developed to capture subtle gait mechanics which are not readily captured by individual metrics. For example, traditional ground reaction force metrics measured through different phases of stance (weight acceptance and propulsive) have been previously found to correlate with walking kinematics such as knee flexion excursion [75]. Additionally, mean pseudo-frequency has been shown to discriminate rearfoot and non-rearfoot patterns [128]. Combining features that capture movement characteristics with only foot contact force data results in robust biomechanical representation and accurate KCF estimation.

The moderate to strong correlations identified suggest that individual insole force features can be used as surrogate measures to monitor peak KCF. For example, given the consistently strong relationship between insole peak force and peak KCF, this metric could be used to monitor peak KCF, although it is not a direct estimate. Alternatively, the relationships between KCF impulse and all insole features were negligible at a group level. The group-wide negligible relationships stem from the variability in the relationship directions between subjects [See Appendix B]. For example, the average correlation between KCF impulse and insole loading rate was 0.11, but with a range of -0.75 to 0.71. We speculate that the variability in correlation direction stems from differences in how subjects' gait changes with speed. As illustrated in Figure 3.2, Subject A developed a significant peak in their KCF curve during the first half of stance as the speed increased, while Subject B did not. These individual gait differences require relationships between force-sensing insole features and KCF impulse to be evaluated per-limb or per-subject. Future research and clinical use of these methods should perform a set of walking trials to establish relationships between insole force features and KCF impulse for each patient. Conversely, the results from this study suggest that these trials would not be necessary to monitor peak KCF with select insole force-based surrogate features.

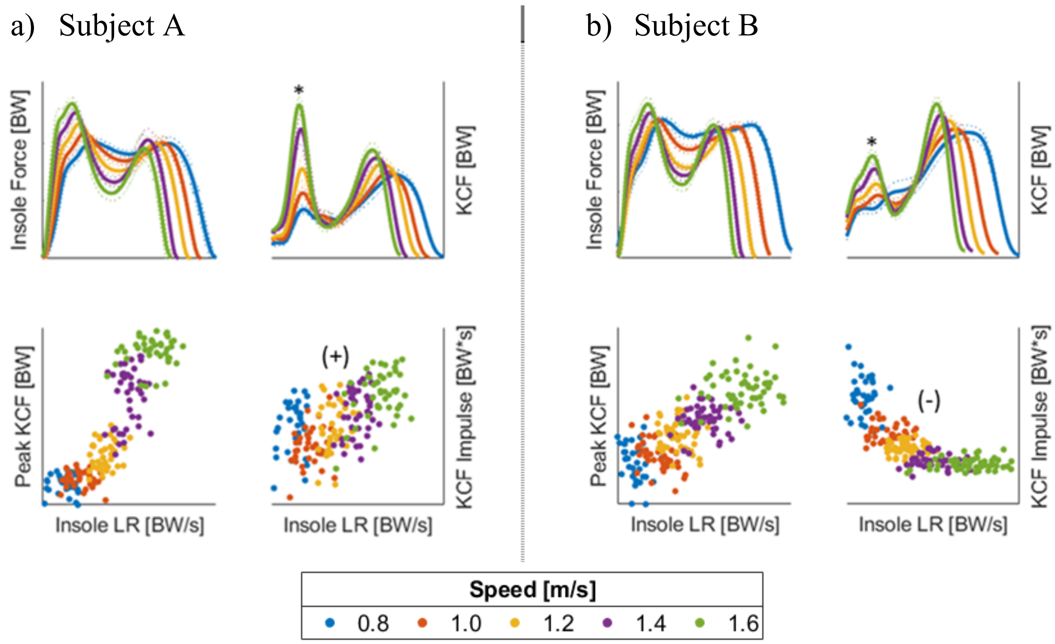


Figure 3.2 Comparison of two subjects' KCF evolution as speed increased. Subject A developed a much larger first peak (*) in KCF than Subject B as speed increased, possibly driving the opposite direction of the association between insole & KCF metrics. LR = loading rate

While the models from this study used data from a healthy population, the methods could be extended to patients following surgery or with movement pathology. For example, clinicians could mitigate the risk of premature osteoarthritis development in anterior cruciate ligament reconstruction patients through the restoration and monitoring of normal knee loads during activities of daily living. Additionally, gait retraining and subsequent load monitoring for total knee arthroplasty patients could optimize the long-term health of their own tissues and the artificial joint. One future consideration is that while the models developed per-limb from this healthy group performed only moderately better than per-subject models, the error of models developed per-subject would likely increase as the methods extend to post-surgical patients with greater between-limb KCF differences.

Consequently, the population of interest must be considered when developing KCF prediction models.

There are a number of limitations to this study. First, because the application of the model is limited to the training data from which it is developed, the scope of KCF prediction from this study is limited to flat surface, straight line walking. Future studies could implement various inclines, steps, and turns on multiple surfaces to increase model generalizability. Additionally, these models were developed from a single data collection. Collecting data on multiple days would provide further generalizability through the introduction of controlled variability in the training data. Finally, although EMG-informed models have been shown to be the most effective option to estimate KCFs with musculoskeletal modeling [83, 98, 115], the KCF results in this study are consistent with those previously reported [66, 96] and were generated with a validated musculoskeletal model and muscle force estimation method.

In conclusion, we present methods which can be used to monitor KCF metrics during activities of daily living using force-sensing insoles. The performance of both the per-limb and per-subject based models developed indicates that accurately estimating KCF metrics can be done with a single wearable sensor. Further, we identified a number of insole features which are strongly or moderately associated with peak KCF, but not KCF impulse. These results carry promising implications for the estimation of KCFs outside of a traditional laboratory with wearable sensors.

CHAPTER 4. CONCLUSIONS

There were two primary aims to this thesis. The first aim was to quantify the link between KCF and PROs which measure pain, ability to perform activities of daily living, and quality of life 6 months after ACLR. We found that subjects with KOOS QOL and ADL scores above the PASS threshold demonstrated greater ACLR limb peak KCF than those below the PASS threshold, but KCF symmetry was similar between groups. There were no differences in peak KCF between the KOOS Pain PASS-Y and PASS-N groups for the ACLR limb, or KCF symmetry for any subscale. These results indicate an offloading strategy that is linked with poor knee-related quality of life and ability to perform activities of daily living.

The second aim was to develop a novel method to monitor KCF outside the laboratory using unobtrusive wearable sensors. We found that models developed per-limb resulted in lower error (2.19% for KCF Impulse and 3.50% for peak KCF) than those developed per-subject (3.40% for KCF Impulse and 6.47% for peak KCF). Additionally, we identified a number of insole features which are, on average, associated with peak KCF. However, while all insole features were, on average, negligibly correlated with KCF impulse, there were individual strong and moderate correlations on a per-limb basis. These results carry promising implications for the estimation of KCFs outside of a traditional laboratory with wearable sensors.

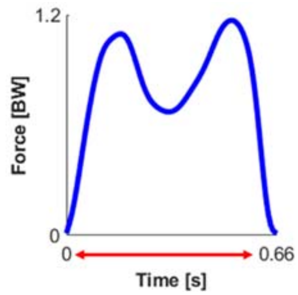
Future researchers could build on these findings by performing longitudinal studies outside the laboratory using wearable sensors to monitor knee loading throughout rehabilitation. These studies would allow for researchers and clinicians to better understand how knee joint loading in everyday life influences important physical and patient reported outcomes (cartilage biology & morphology, PROs, functional performance, etc.). Monitoring measures of interest throughout a gait intervention protocol would allow the efficacy of these methods to be directly evaluated. These findings would allow for future rehabilitation protocol optimization through evidence-based design.

The ability to monitor knee joint loading with wearable sensors could provide clinicians an additional biofeedback tool for their patients. In particular, ACLR patients could be trained to walk with peak KCF in the same range as the PASS-Y group for the KOOS QOL and ADL. Through the use of periodic biofeedback to ensure gait modification compliance, these patients would hopefully demonstrate improved quality of life, ability to complete activities of daily living, and tissue health. Together, these studies provide both justification and methods to monitor KCF outside of a traditional laboratory for ACLR patients.

APPENDIX A – FEATURE DESCIRPTIONS & ILLUSTRATIONS

Table A.1 Illustrations of features extracted from insole data

1- Stance Time

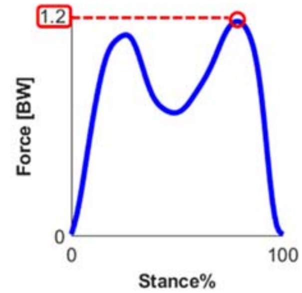


Description
Total time with force > 20 N

Equation
$Time_{end} - Time_1$

Units
second

2- Peak Magnitude

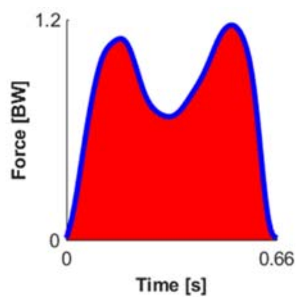


Description
Maximum force during stance

Equation
$\max(Force_{1\to end})$

Units
Bodyweight

3- Impulse



Description
Total area under force curve during stance

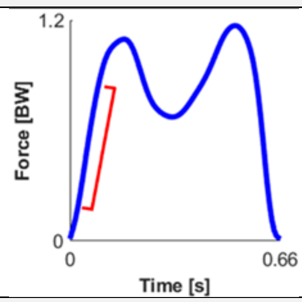
Equation

$$\frac{Time_{end} - Time_1}{2N} \sum_{n=1}^{N} Force(Time_n) + Force(Time_{n+1}))$$

where $N = \#\, points\ during\ stance$

Units
Bodyweight*second

4- Loading Rate

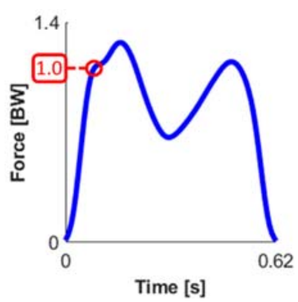


Description
Slope of the foot-contact-force curve from 3-12% of stance

Equation
$(Force_{12\%} - Force_{3\%})/(Time_{12\%} - Time_{3\%})$

Units
Bodyweight/second

5- Impact Peak Magnitude

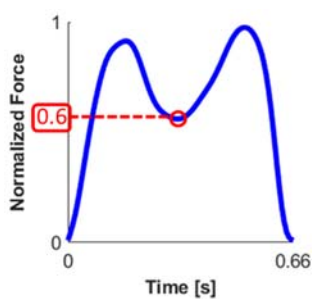


Description
Force at impact peak location (if no peak, then 15% of stance)

Equation
$Force_{IPLocation}$

Units
Bodyweight

6- Mid-Drop Magnitude



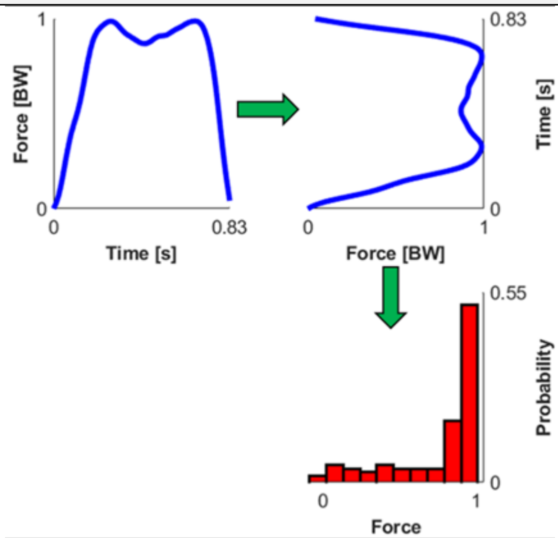
Description
Max-normalized force at mid-drop location (local min from 40-60% of stance)

Equation
$Force_{MDLocation}$

Units
None (0-1)

Table A.2 Illustrations of features extracted from insole data

7- Skewness



Description

Sample skewness of the force data

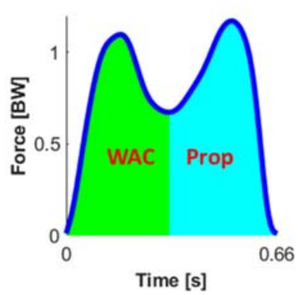
Equation

$$\frac{\frac{1}{N}\sum_{n=1}^{N}(x_n-\bar{x})^3}{\left(\sqrt{\frac{1}{N}\sum_{n=1}^{N}(x_N-\bar{x})^2}\right)^3}$$

where $N = \#\, points\ during\ stance$

Units

none



8- Weight Acceptance (WAC) Impulse

Description

Impulse in weight acceptance phase of gait

Equation

$$J_{WAC} = \frac{Time_{MDLoc} - Time_1}{2N} \sum_{n=1}^{N} Force(Time_n) + Force(Time_{n+1}))$$

where $N = \#\, points\ during\ WAC\ phase$

Units

Bodyweight*second

9- Propulsive (Prop) Impulse

Description

Impulse in propulsive phase of gait

Equation

$$J_{Prop} = \frac{Time_{end} - Time_{MDLoc}}{2N} \sum_{n=1}^{N} Force(Time_n) + Force(Time_{n+1}))$$

where $N = \#\, points\ during\ Prop\ phase$

Units

Bodyweight*second

10- WAC/Prop Impulse Ratio

Description

Ratio of impulse from WAC and Prop phases of gait

Equation

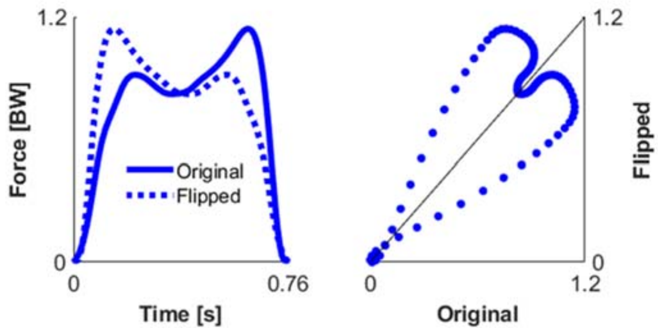
$WAC{:}\,Prop\ Impulse\ Ratio = J_{WAC}/J_{Prop}$

Units

none

Table A.3 Illustrations of features extracted from insole data

11- Symmetry



Description

Symmetry of insole force signal

Equation

$$\frac{1}{N-1}\sum_{n=1}^{N}\left(\frac{A_n-\mu_A}{\sigma_A}\right)\left(\frac{B_n-\mu_B}{\sigma_B}\right)$$

where

A = original signal

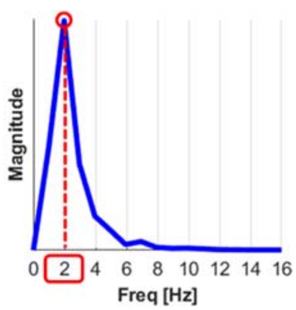
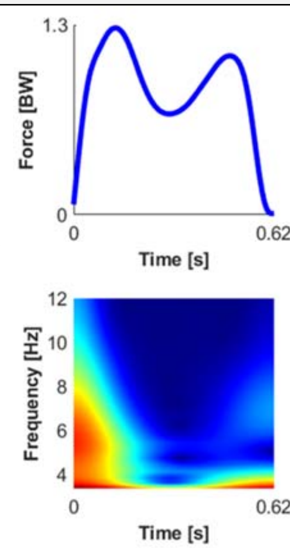
B = flipped signal

μ = mean

σ = standard deviation

Units

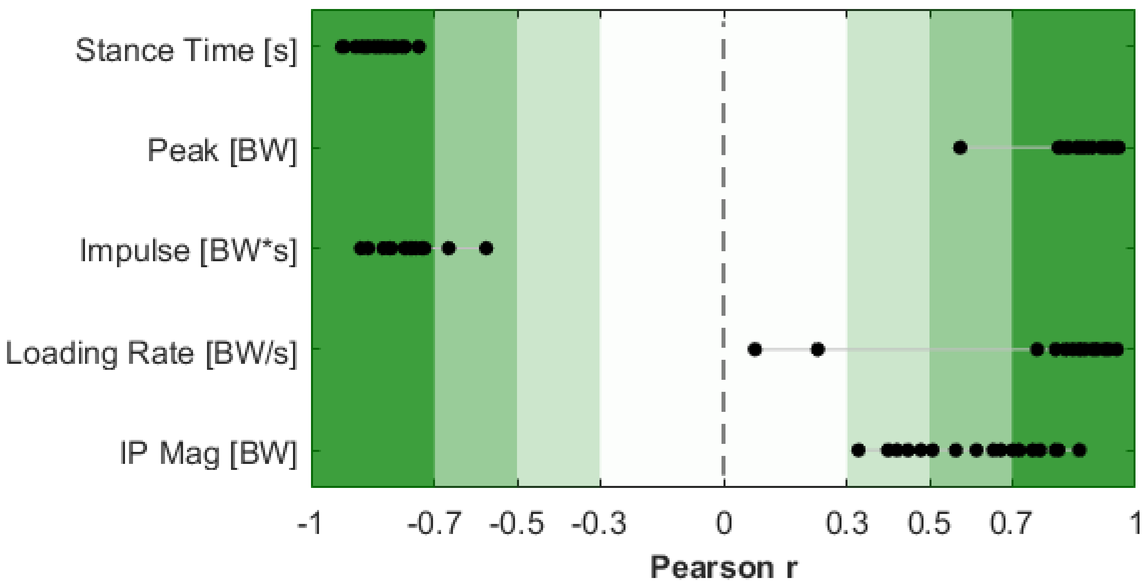
none

12- Frequency @ Max DFT Magnitude	**13- Mean Pseudo-Frequency**
 **Description** Frequency with greatest DFT magnitude. **Equation** $Freq_{\max(DFT)}$ **Units** Hz	 **Description** Average pseudo-frequency using Mexican hat wavelet **Equation** See (Gruber, 2017) [128] **Units** ~Hz

APPENDIX B – INDIVIDUAL CORRELATION COEFFICIENTS

The following are illustrations are plots of individual (per-limb) correlation coefficients between insole features and KCF metrics. Each black scatter point denotes the Pearson r value computed per-limb between the corresponding insole feature on the y-axis and the KCF metric denoted in the section header. The important takeaway from these plots is that individual correlations between many insole features and peak KCF are consistent in their strength and direction, while those with KCF impulse alter direction between individual limbs even though there are a number of moderate to strong correlations.

a) Peak KCF & Traditional Metrics



b) KCF Impulse & Traditional Metrics

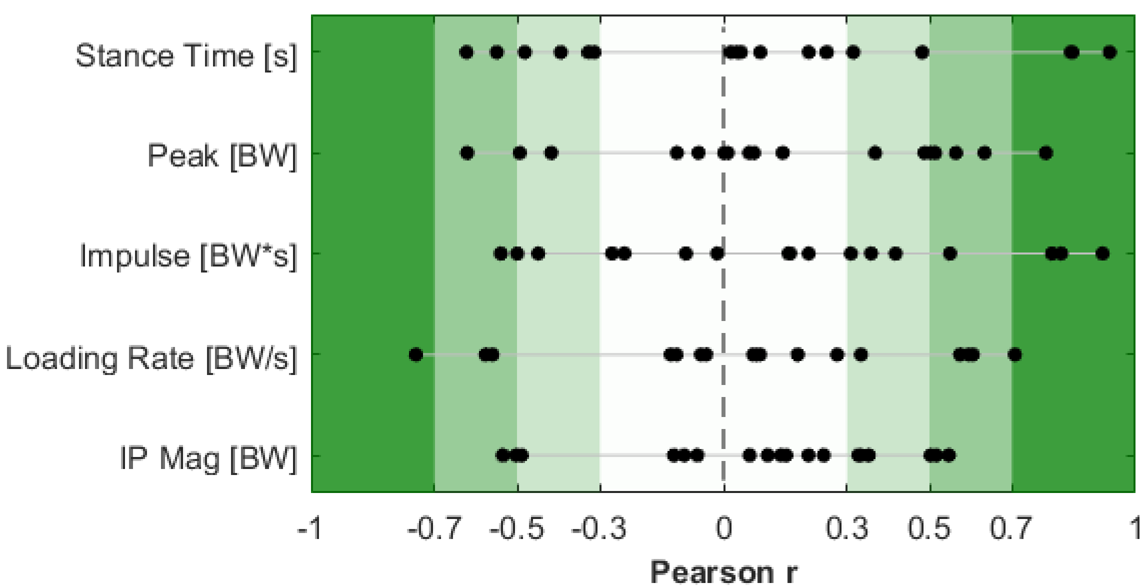
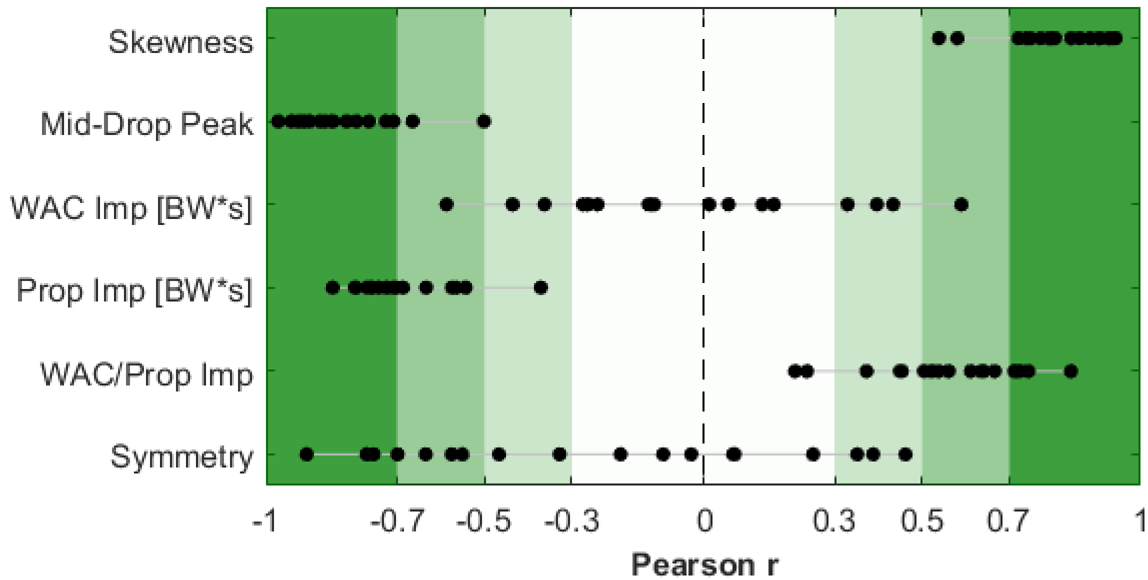


Figure B.1 Correlation coefficients of traditional time domain features with a) peak KCF and b) KCF impulse per subject per limb.

a) Peak KCF & Non-Traditional Metrics



b) KCF Impulse & Non-Traditional Metrics

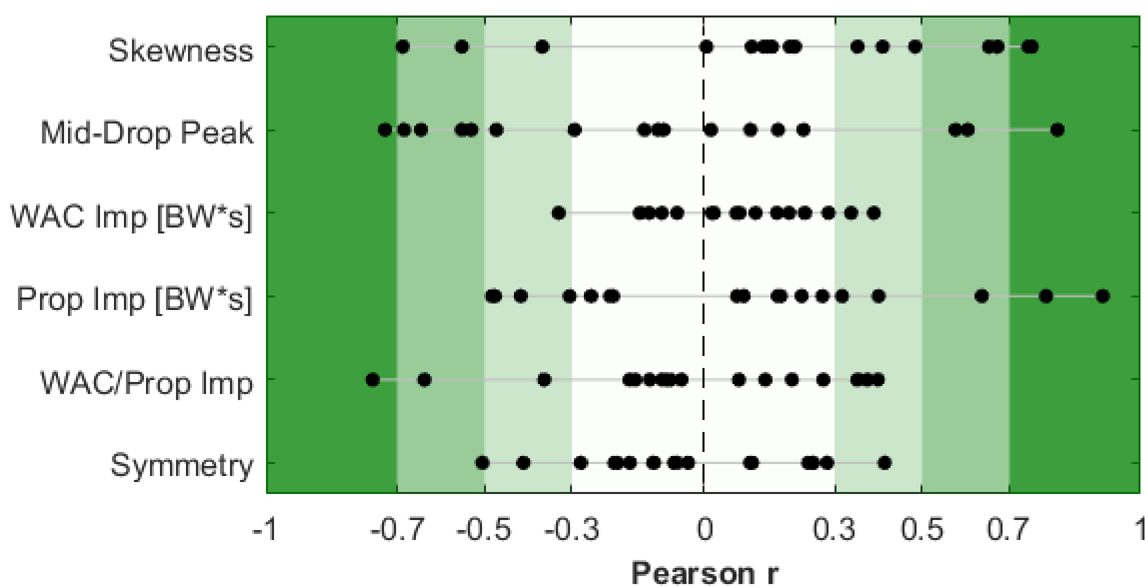
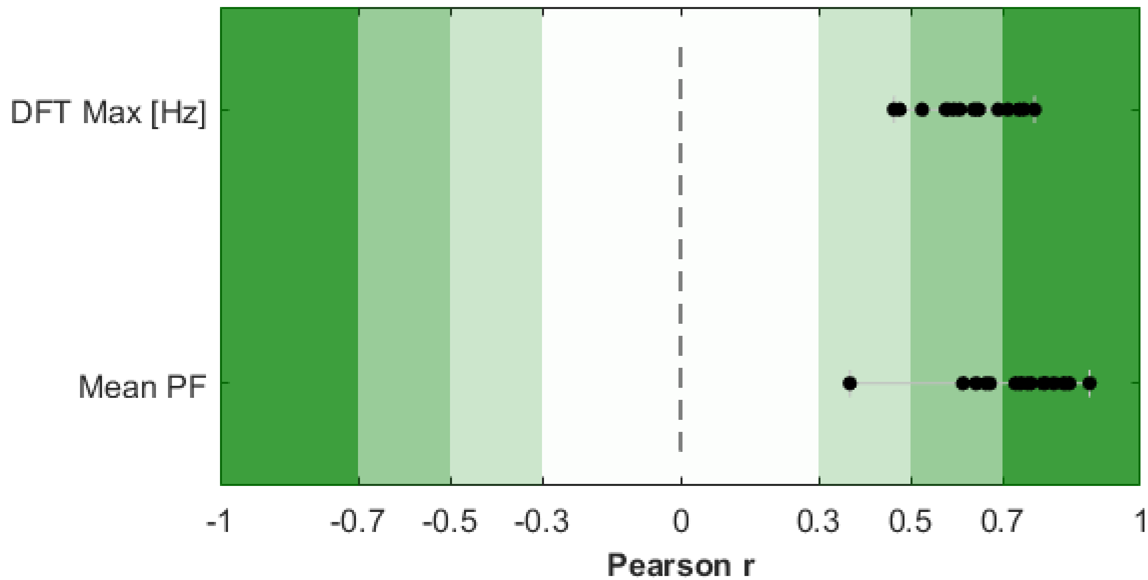


Figure B.2 Correlation coefficients of other time domain features with a) Peak KCF and b) KCF Impulse per subject per limb.

a) Peak KCF & (Pseudo-)Frequency Domain Features



b) KCF Impulse & (Pseudo-)Frequency Domain Features

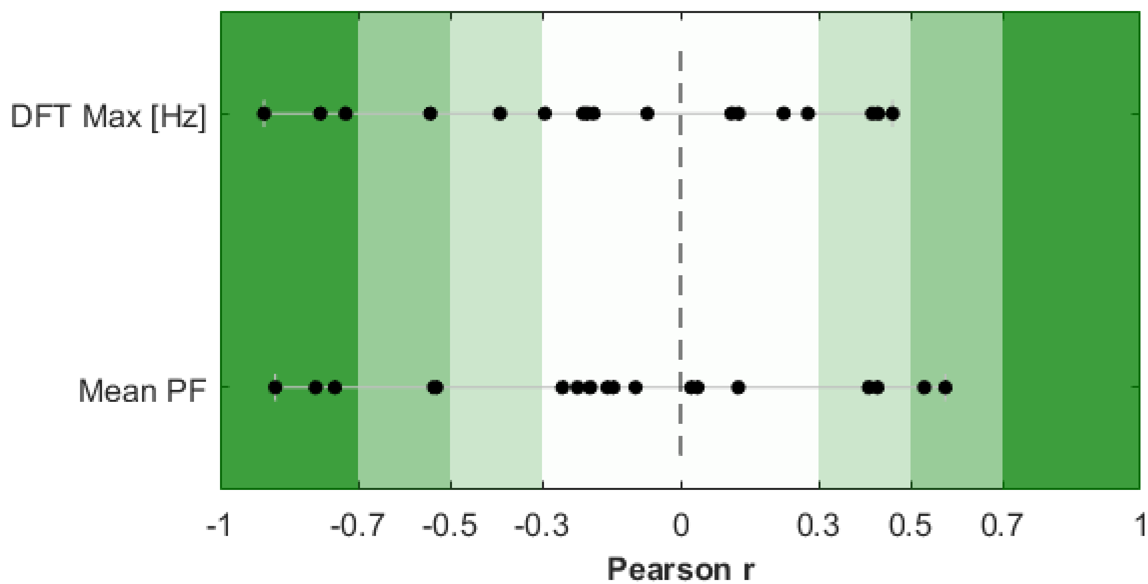


Figure B.3 Correlation coefficients of frequency domain features with a) Peak KCF and b) KCF Impulse per subject per limb.

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Spencer A., Jacobs C., Davis K., Johnson D., Ireland M.L., Noehren B. (2018). Quadriceps Force Steadiness following Anterior Cruciate Ligament Reconstruction during a Maximum Voluntary Isometric Contraction. *Mid-South Biomechanics Conference*, Memphis, TN, February 22-23, 2018 ***[MSBC Student Research Award]***.

Spencer A., Jacobs C., Davis K., Johnson D., Ireland M.L., Noehren B. (2018). Quadriceps Force Steadiness and Functional Implications following Anterior Cruciate Ligament Reconstruction. *American College of Sports Medicine 66th Annual Meeting*, Minneapolis, MN, May 28-June 1, 2018 ***[ACSM Biomechanics Interest Group Student Research Award]***.