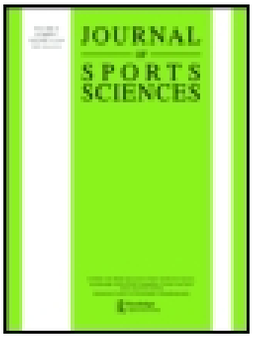

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Changes in tibiofemoral contact forces during running in response to in-field gait retraining

Richard W. Willy[a], Stacey A. Meardon[a], André Schmidt[b], Nathan R. Blaylock[a], Scott A. Hadding[a] and John D. Willson[a]

[a]Department of Physical Therapy, East Carolina University, Greenville, NC, USA; [b]Orthopedic University Hospital Friedrichsheim gGmbH, Frankfurt am Main, Germany

ABSTRACT

We evaluated the efficacy of an in-field gait retraining programme using mobile biofeedback to reduce cumulative and peak tibiofemoral loads during running. Thirty runners were randomised to either a retraining group or control group. Retrainers were asked to increase their step rate by 7.5% over preferred in response to real-time feedback provided by a wrist mounted running computer for 8 routine in-field runs. An inverse dynamics driven musculoskeletal model estimated total and medial tibiofemoral joint compartment contact forces. Peak and impulse per step total tibiofemoral contact forces were immediately reduced by 7.6% and 10.6%, respectively ($P < 0.001$). Similarly, medial tibiofemoral compartment peak and impulse per step tibiofemoral contact forces were reduced by 8.2% and 10.6%, respectively ($P < 0.001$). Interestingly, no changes were found in knee adduction moment measures. Post gait retraining, reductions in medial tibiofemoral compartment peak and impulse per step tibiofemoral contact force were still present ($P < 0.01$). At the 1-month post-retraining follow-up, these reductions remained ($P < 0.05$). With these per stance reductions in tibiofemoral contact forces in mind, cumulative tibiofemoral contact forces did not change due to the estimated increase in number of steps to run 1 km.

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KEYWORDS
Knee; gait retraining; running; biomechanics; osteoarthritis; biofeedback

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Introduction

Degenerative injuries to the tibiofemoral joint, including symptomatic meniscal lesions and osteoarthritis may account for as many as 16.8% of all knee injuries in runners (Taunton, 2002). Elevated tibiofemoral joint loads during gait are thought to relate to the onset, rate of progression and the severity of knee osteoarthritis (Frost, 1994; Miyazaki et al., 2002; Reimann, 1973; Sharma et al., 1998). High joint contact forces can result in articular cartilage degradation, particularly when applied in a repetitive manner such as seen during running (Kerin, Coleman, Wisnom, & Adams, 2003; Radin et al., 1973; Souza et al., 2014). Therefore, it would appear that both *peak and cumulative* estimates of joint contact forces are equally important to address in at-risk populations (Bennell et al., 2011; Chen, Burton-Wurster, Lust, Bank, & Tekoppele, 1999).

It has been suggested that individuals with osteoarthritis who exhibit a high loading rate of the vertical ground reaction force may experience elevated tibiofemoral joint articular loads (Mündermann, Dyrby, & Andriacchi, 2005). Indeed, articular cartilage may be most sensitive to high impact forces, rather than long sustained loads (Chen et al., 1999). Thus, individuals who run with high impact forces may benefit the greatest from running interventions that target a reduction in peak or cumulative knee joint contact forces. This may be particularly true if the runner is at increased risk for tibiofemoral joint osteoarthritis due to a previous knee injury or if a degenerative process has already begun (Lohmander, Ostenberg, Englund, & Roos, 2004; Noehren, Wilson, Miller, & Lattermann, 2013).

Increased tibiofemoral joint contact forces noted during running are disproportionately borne by the medial compartment of the tibiofemoral joint (Bergmann et al., 2014). As such, reducing tibiofemoral joint contact forces in the medial compartment may be especially beneficial to runners who are either at-risk for the development or have already developed medial compartmental osteoarthritis. While the knee external adduction moment is widely used as a surrogate for *in vivo* contact loads in the medial compartment (Zhao et al., 2007), considering the frontal and sagittal plane knee moments together likely provides a better estimate of medial compartment loading (Manal, Gardinier, Buchanan, & Snyder-Mackler, 2015; Meyer et al., 2013).

Increasing step rate at a given running velocity reduces loading rate of the vertical ground reaction force (Hobara, Sato, Sakaguchi, Sato, & Nakazawa, 2012; Willy et al., 2015), peak knee extensor moment and knee joint work (Heiderscheit, Chumanov, Michalski, Wille, & Ryan, 2011) and quadriceps forces (Lenhart, Thelen, Wille, Chumanov, & Heiderscheit, 2014). Based on these changes in lower extremity loading, reductions in tibiofemoral joint contact forces may also result. Increasing step rate at a given running velocity increases the number of gait cycles required to cover a given distance and, therefore, may have the unintended effect

CONTACT Richard W. Willy willyr@ecu.edu Department of Physical Therapy, College of Allied Health Sciences, East Carolina University, Greenville, NC 27834, USA.

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of increasing cumulative tibiofemoral joint contact forces. Ideal interventions for the prevention or treatment of tibiofemoral degenerative changes would reduce both *peak and cumulative* tibiofemoral joint contact forces.

In order to accomplish an acute increase in step rate (with a corresponding reduction in step length), runners tend to land with their foot closer to the body centre of mass, utilising a slightly flatter footstrike, greater knee flexion and less hip flexion (Clarke, Cooper, Hamill, & Clark, 1985; Heiderscheit et al., 2011). However, it is unknown if this kinematic pattern associated with a new step rate can be sustained beyond the initial kinematic pattern acquisition (Brashers-Krug, Shadmehr, & Bizzi, 1996; Overduin, Richardson, Lane, Bizzi, & Press, 2006).

Previous investigations have utilised gait retraining to reduce the knee external adduction moment during walking (Barrios, Crossley, & Davis, 2010; Shull et al., 2013; Wheeler, Shull, & Besier, 2011). With regards to gait retraining in runners, vertical loading rates have been shown to be reduced with real-time feedback on tibial shock in a population thought to be at high risk for the development of a tibial stress fracture (Crowell & Davis, 2011). Despite these foundational studies, no study has yet utilised a gait retraining intervention to reduce tibiofemoral contact forces in runners.

The purpose of this study was to determine (a) if a reduction in peak and cumulative tibiofemoral contact forces during running occur in response to an acute reduction in step length, (b) if a gait retraining programme using mobile biofeedback to cue a reduction in step length results in a reduction in peak and cumulative tibiofemoral contact forces and (c) if these changes can be maintained for at least 30-days post gait retraining. Knowledge of the eventual kinematic pattern that results from the step rate retraining program may assist clinicians in developing effective cueing techniques. Thus, a secondary purpose of this study was to identify the specific kinematic strategies used by participants during the gait retraining programme.

Methods

Initial gait screening session

Ethical approval was obtained from the Ohio University and East Carolina University ethics committees. Informed consent was obtained prior to the commencement of the study. Eighty-three runners attended a brief 15-min screening session to identify those who ran with excessive impact forces (Willy et al., 2015). In order to qualify, runners were required to be healthy runners of at least 18–35 years of age, running at least 10 km · week^{-1} and free of any injuries in the past 6 months. During this screening session, only ground reaction forces were sampled (1000 Hz, filtered with a 4th order, zero-lag, low pass, Butterworth filter at 50 Hz) during the 5th minute of running (3.3 m · s^{-1}) on an instrumented treadmill (Bertec, Worthington, OH, USA). High impact running mechanics were operationally defined as running with an instantaneous load rate of vertical ground reaction forces of ≥85 bw · s^{-1} in either limb. The impact criterion corresponds to the loading rate during running noted previously in individuals thought to be at high risk for the development of knee osteoarthritis (~5 years post-anterior cruciate ligament reconstruction) (Noehren et al., 2013). These screening sessions yielded 30 high impact runners. Through a blinded allocator, high impact runners were then randomly assigned to either a retraining or control group. All runners were blinded to the kinetic inclusion of criterion of high impact forces for the duration of their participation in this study. The limb with the highest instantaneous load rate of vertical ground reaction forces served as the analysed limb for the remainder of the study.

Baseline gait session

All high impact runners then underwent a fully instrumented gait analysis to assess running mechanics at a self-selected speed. First, 55 retroreflective markers were attached to the bilateral lower extremities and trunk of each high impact runners. After collection of static calibration and functional hip joint trials (Leardini et al., 1999), all participants completed a 5-min progressive warm-up run to allow for accommodation to the treadmill. 3-D treadmill running kinematics and kinetics were sampled at 200 Hz and 1000 Hz, respectively, using real-time data acquisition methods via a 10-camera motion capture system and a time synchronised instrumented treadmill (Qualysis Corporation™, Gothenburg, SWE and The MotionMonitor™, Chicago IL, USA). Running mechanics were sampled for 10 s of which 5 consecutive strides were utilised for subsequent analysis.

Acute modification trial measurement

Following the collection of this single baseline trial, a 3-min rest period was provided to all high impact runners. The retraining group was then cued to increase their step rate by 7.5% while running at their preferred speed. Real-time feedback on stride rate (stride rate = step rate × 0.5) was provided via a wrist-mounted running computer (Forerunner 70™, Garmin, Olathe, KS, USA) that calculated step rate via a shoe-mounted, wireless, tri-axial accelerometer. In pilot testing, we found the step rate measurements calculated by the Forerunner 70™ had a minimal 95% limit of agreement of −2.9 steps · min^{-1} and maximal 95% limit of agreement of 2.4 steps · min^{-1} compared with the step rate determined with our instrumented treadmill (Bland & Altman, 1986). A laboratory assistant, positioned immediately in front of the treadmill, held the running computer so that the retraining participants could view their real-time step rate (Figure 1). Importantly, participants were not provided any other feedback or cueing other than feedback on step rate e.g., neither kinematic feedback nor cueing were provided. The running computer uses a 5-s roaming average to calculate step rate to provide a stable feedback value that for the runner. Once step rate had reached the prescribed level, but not greater than 8.5%, for consecutive 30 s, a second motion trial was collected to assess the acute effects of the gait modification. The retraining participants continued to receive step rate feedback during the collection of the acute modification trial. Conversely, control runners were not instructed to make any gait modifications for their parallel acute modification trial. Importantly, control

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Figure 1. The Garmin FR70 (Olathe, KS, USA) that was used for all retrainers and controls: (a) a wireless accelerometer transmitted data to the wrist mounted running computer. (b) The wrist mounted running computer was configured such that running duration, real-time strides per minute (steps per minute/2) and running distance were visible to all retrainers during in-field retraining sessions 1–3, 5 and 7. (c) During in-field retraining session 4, 6 and 8, retrainers were only able to view feedback on run duration and distance run as step rate feedback was faded. All control participants only viewed the display in (c). Figure from Willy RW, http://www.ncbi.nlm.nih.gov/pubmed/25652871. Reprinted with permission from Wiley and Sons.

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participants were blinded to the existence of the retraining group and were merely told that we were interested in how their running mechanics may change over the course of the study.

Retraining phase

All high impact runners were then issued personal Forerunner 70™ running computers for in-field running trials and asked to return to their normal running routines outside the laboratory. The running computers issued to the control participants were configured such that only data on running distance and average pace were visible. Control participants were only told that we were interested in changes in their running mechanics over time. In response to real-time feedback on step rate provided by the running computer, retraining participants were instructed to maintain the 7.5% increase in step rate during the next 8 running sessions over the next several weeks. A self-controlled practice schedule was utilised such that the retrainers self-determined the frequency in which they would view the Forerunner 70™ watch to receive feedback on step rate. We opted for this type of feedback as self-controlled feedback has previously been suggested to increase learner engagement, enhance motivation ultimately resulting in greater motor learning performance (Wulf, Shea, & Lewthwaite, 2010). Retrainers were instructed to view the real-time feedback on step rate on the running computer on runs 1–3, 5 and 7. To encourage internalisation of the new step rate pattern, retrainers were not permitted to view feedback on step rate during runs 4, 6 and 8. All participants, regardless of group assignment, were not restricted in the mode of running i.e., treadmill or outdoors. The data logging function provided by the running computers was utilised to determine how each retraining participant performed with meeting their respective step rate target.

Follow-up measurements (post-retraining, 1-month post-retraining)

Following the retraining phase, all high impact runners immediately returned for follow-up gait assessments (post-retraining) in which data were captured in exactly the same manner as during the baseline trial, albeit with only a single 10-s data trial. To assess retention of the new running style, no feedback on step rate was given to the retrainers during the post-retraining data collection. All participants then returned to their normal running routines for an additional 30 calendar days. During this period, the data logging function provided by the running computers was used to monitor each runner's step rate performance. To assess retention of the new running style during this study phase, the retrainers were no longer able to view feedback on their step rate. Thus, both retrainers and control participants were only able to view feedback on running distance and average pace for the 30-day monitoring period. Immediately at the conclusion of the 30-day monitoring period, all high impact runners returned for a final gait assessment. At the conclusion of the 1-month post-retraining data collection, all participants were debriefed on inclusion criteria and the existence of the parallel group. Please see Appendix for CONSORT 2010 Diagram.

Data processing and musculoskeletal model

All motion data were processed utilising The MotionMonitor software and custom-written LabVIEW code. Marker and ground reaction force data were filtered with a 15-Hz cutoff frequency via a 4th order, zero-lag, low pass, Butterworth filter (Bezodis, Salo, & Trewartha, 2013; Kristianslund, Krosshaug, & van den Bogert, 2012). Segmental inertial parameters (Dempster, Gabel, & Felts, 1959) were used in inverse dynamic calculations to yield net internal joint moments and joint reaction forces of the ankle, knee and hip expressed in the distal segment coordinate system.

Tibiofemoral contact force estimates were based on a previously described knee joint model which uses inputs of hip, knee and ankle angles and corresponding net joint moments to derive hamstrings, quadriceps and gastrocnemius muscle forces (DeVita & Hortobagyi, 2001; Messier et al., 2011). Hamstrings muscle force was calculated based on the hip extensor moment, which was assumed to be produced by only the gluteus maximus and hamstrings muscles. The proportion of the hip extension moment attributed to the hamstring muscle was based on the physiological cross-sectional area of each muscle (Ward, Eng, Smallwood, & Lieber, 2009)

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and their moment arms to the hip joint as a function of hip flexion angle (Nemeth & Ohlsen, 1985). The direction of the hamstring muscle force was parallel to the thigh segment and was therefore applied to the tibia as a function of knee flexion angle.

Gastrocnemius muscle force was then calculated as the quotient of ankle joint plantar flexor moment and the Achilles tendon moment arm (Klein, Mattys, & Rooze, 1996). The ankle plantar flexion moment was assumed to be produced only by the gastrocnemius and soleus muscles. The proportion of the moment attributed to the gastrocnemius muscle was based on the physiological cross-sectional area of each muscle (Ward et al., 2009). The direction of the gastrocnemius force relative to the tibia was 3° posterior to the long axis of the shank segment (DeVita & Hortobagyi, 2001).

Next, quadriceps muscle force was determined using a two-stage approach. First, a knee flexor moment due to hamstring and gastrocnemius force acting through their moment arms to the knee joint as a function of knee flexion angle was calculated (Herzog & Read, 1993; Spoor & van Leeuwen, 1992; Spoor, van Leeuwen, Meskers, Titulaer, & Huson, 1990; Visser, Hoogkamer, Bobbert, & Huijing, 1990). The calculated knee flexor moment was then summed with the net knee extension moment to create a knee extension moment that accounts for co-contraction. Second, we divided this adjusted knee extension moment by the effective quadriceps moment arm to derive quadriceps muscle force (van Eijden, Kouwenhoven, & Weijs, 1987). This quadriceps muscle force was applied to the tibia though the patellar tendon at an angle as a function of knee flexion angle (Herzog & Read, 1993).

To derive tibiofemoral joint contact force, we summed the individual components of the muscle forces and the knee joint reaction forces parallel with the long axis of the shank segment coordinate system. Next, we summed the anterior–posterior components of the muscle forces and the knee joint reaction forces perpendicular to the shank segment coordinate system. The sum of the vertical and anterior–posterior components acting perpendicular to an 8.8° posterior tibial slope (Giffin, Vogrin, Zantop, Woo, & Harner, 2004) was the total tibiofemoral joint contact force. The proportion of the tibiofemoral joint contact force acting through the medial compartment was estimated based on the methods described by Schipplein and Andriacchi (1991). Using this approach, we determined the force necessary to balance the frontal plane knee joint moment and total knee compression forces around contact points at 25% and 75% of subject-specific knee joint width with the remaining moment balanced by the lateral knee soft tissue and ligamentous structures (Draganich, Andriacchi, & Andersson, 1987). Using this model, estimated peak tibiofemoral joint contact force and tibiofemoral joint contact force impulse using this model are within 3% and 7% of those recorded *in vivo* from an instrumented prosthesis during walking (Fregly et al., 2012). Medial peak tibiofemoral joint force and impulse estimated with this model are within 4% and 7%, respectively, of values recorded *in vivo* during walking (Fregly et al., 2012).

Statistical analysis

All statistical analyses were performed with SPSS Version 20 (IBM, Houston, TX, USA). Demographic data were analysed with independent *t*-tests and Fisher's exact tests ($\alpha = 0.05$) to detect baseline between group differences. Variables of interest from the musculoskeletal model were peak tibiofemoral joint contact force, tibiofemoral contact force × time impulse, peak medial compartment contact force, medial compartment force × time impulse, peak knee external adduction moment and impulse of the knee external adduction moment. Cumulative loads per kilometre of running for the tibiofemoral joint and the medial compartment of the tibiofemoral joint were estimated as the product of impulse per step and estimated number of strides per kilometre obtained during the respective data collections. Spatiotemporal and sagittal plane kinematic running variables of interest were step rate, step length, distance of the centre of pressure from the centre of mass at footstrike and sagittal plane angles of the foot, knee, hip and shank at footstrike. Segmental angles were referenced to the lab coordinate system. These sagittal plane variables are often used in clinical gait analysis and are often altered with acute changes in step rate. Separate two-way, mixed model, analyses of variance (ANOVA) (group (2) × time point (4)) were used for each dependent variable of interest. In the event that the assumptions of the ANOVA were violated, a Greenhouse–Geisser correction was made. When a significant interaction was detected ($\alpha = 0.05$), two-tailed post-hoc comparisons were made using the least significant difference approach. To assist with interpretation, effect sizes (*d*) were also calculated to assess the effect of any significant changes (Cohen, 1988). To assist with interpretation of the results, step rate performance during the retraining sessions in the retraining participants was analysed descriptively.

Results

We found no differences between groups regarding baseline demographic characteristics (Table 1). Retraining participants took an average of 13.1 days (SD = 3.8) to complete the 8-retraining sessions. Please see Figure 2 for a descriptive analysis of retrainers' performance during the retraining phase, with respect to the prescribed step rate. Retrainers

Table 1. Demographics for participants.

Variable	RT	CON	*P*-value
Age (months)	251.88 (16.3)	248.8 (15.0)	0.99
BMI ($kg \cdot m^{-2}$)	23.0 (2.6)	23.4 (3.3)	0.73
Reported running volume prior to study ($km \cdot week^{-1}$)	22.1 (10.8)	23.2 (17.9)	0.83
Self-selected running pace ($m \cdot s^{-1}$)	3.21 (0.27)	3.16 (0.28)	0.60
Males/females (*n*)	7/9	9/5	0.26

RT: Retrainers, CON: controls.

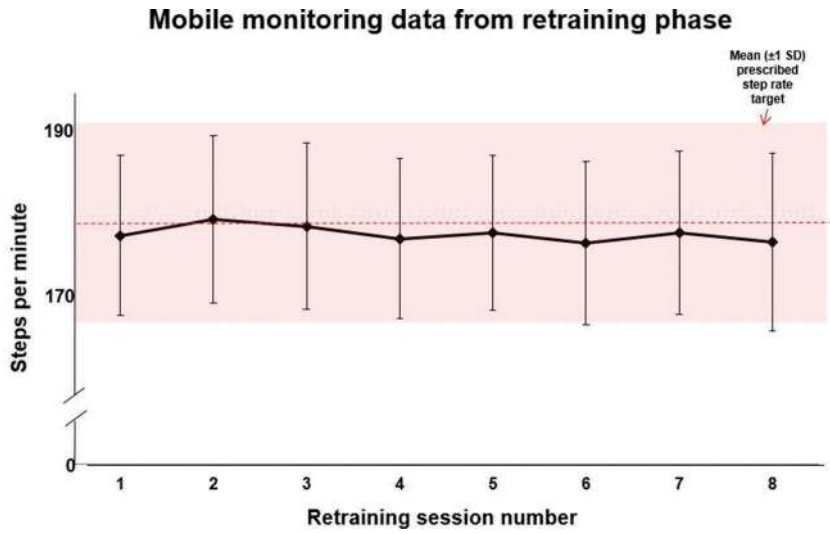

Figure 2. Mobile monitoring data (error bars corresponding to ±1 standard deviation) collected during the in-field retraining sessions from the wrist-mounted running computers for the retraining group only. The dotted line and shaded region represent the mean (standard deviation) prescribed step rate target for the retraining group. The data points represent the retrainers' mean (standard deviation) step rate during each retraining session.

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subsequently completed an additional 15.8 (SD = 3.2) running sessions during the 30-day monitoring period. One control participant was lost during the retraining phase due to a non-running related injury. Therefore, an intention to treat analysis was performed with the last observation carried forward. From the musculoskeletal model, there were significant group × time interactions for peak and impulse tibiofemoral joint contact force and peak and impulse medial compartment contact force (Table 2). Subsequent pairwise comparisons between the baseline and the acute gait trials indicated acute reductions in the retraining group from baseline in peak ($P < 0.001$, $d = -0.65$) and impulse ($P < 0.001$, $d = -0.77$) tibiofemoral joint contact force and as well as peak ($P < 0.001$, $d = -0.56$) and impulse medial tibiofemoral ($P < 0.001$, $d = -0.82$) compartment contact force. From baseline to post-retraining and from baseline to 1-month post-retraining, respectively, there were significant retraining group reductions in peak (post-retraining: $P < 0.001$, $d = -0.60$;1-month post-retraining: $P = 0.02$, $d = -0.30$) and impulse (post-retraining: $P < 0.001$, $d = -0.78$; 1-month post-retraining: $P = 0.005$, $d = 0.48$) tibiofemoral joint contact force and peak (post-retraining: $P = 0.007$, $d = -0.45$; 1-month post-retraining: $P = 0.02$, $d = -0.30$) and impulse (post-retraining: $P = 0.002$, $d = -0.63$; 1-month post-retraining: $P = 0.001$, $d = -0.54$) medial tibiofemoral compartment contact forces, respectively. We did not observe a change in cumulative impulse per kilometre for the tibiofemoral joint contact force nor for the medial compartment contact force at any time point, nor was the ANOVA significant for the peak and impulse of the knee external adduction moment. There were no significant post-hoc findings noted in the control group.

For the secondary analysis of the kinematic strategies, significant group × time interactions for step rate, step length, centre of pressure distance from the body centre of mass at footstrike, foot angle at footstrike, shank angle at footstrike and knee angle at footstrike were found (Table 3). With regards to the changes noted during the acute gait modification trial, only step rate ($P < 0.001$, $d = 1.07$), step length

Table 2. Results for the ANOVA (group (2) × time (4)) at each time point for the laboratory assessment of running biomechanics.

Variable	Group	Baseline	Acute	Post	1MO	*P*-value (group × time)
Peak knee external adduction moment ($N \cdot m \cdot (kg \cdot m)^{-1}$)	RT	0.509 (0.176)	0.4717 (0.155)	0.446 (0.174)	0.474 (0.168)	0.43
	CON	0.454 (0.113)	0.470 (0.192)	0.479 (0.198)	0.493 (0.175))	
Impulse knee external adduction moment ($N \cdot m \cdot (kg \cdot m)^{-1}$)	RT	0.062 (0.029)	0.055 (0.0277)	0.055 (0.030)	0.052 (0.030)	0.48
	CON	0.052 (0.025)	0.057 (0.031)	0.056 (0.035)	0.057 (0.031)	
Peak total tibiofemoral contact force (bw's)	RT	10.32 (1.57)	9.37 (1.35) **	9.53 (1.27)**	9.83(1.62)**,‡	0.004
	CON	9.85 (1.51)	9.75 (1.53)	9.75 (0.90)	9.66 (1.64)	
Impulse total tibiofemoral contact force (bw's)	RT	1.32 (0.20	1.18 (0.17)**	1.18 (0.19)**	1.23 (0.20)**	0.007
	CON	1.28 (0.17)	1.28 (0.18)	1.26 (0.12)	1.28 (0.16)	
Total tibiofemoral contact force per kilometre (bw's)	RT	570.43 (66.56)	545.76 (61.82)	554.04 (51.99)	571.85 (59.07)	0.40
	CON	558.03 (47.37)	556.95 (40.10)	566.76 (55.04)	566.94 (38.72)	
Peak medial compartment tibiofemoral contact force (bw's)	RT	6.45 (0.97)	5.92 (0.90)**	6.03 (0.97)**	6.16 (0.99)*	0.02
	CON	6.14 (0.92)	6.09 (0.92)	6.32 (0.97)	6.32 (0.85)	
Impulse medial compartment tibiofemoral contact force (bw's)	RT	0.85 (0.12)	0.76 (0.10)**	0.77 (0.15)**	0.78 (0.10)**	0.01
	CON	0.82 (0.14)	0.81 (0.15)	0.83 (0.12)	0.82 (0.0.17)	
Medial tibiofemoral contact force per kilometre (bw's)	RT	366.50 (37.38)	350.73 (33.5)	361.56 (60.92)	367.39 (32.62)	0.29
	CON	355.75 (45.80)	351.64 (45.69)	374.01 (50.95)	363.50 (58.11)	

Data represent group mean (standard deviation). Acute refers to the data collection that occurred on the same day as the baseline measurements at the beginning of the study. Post refers to the data collection that occurred immediately following the post-retraining phase. 1MO refers to the data collection that occurred at the conclusion of the 1-month monitoring period.

*Significantly different than baseline ($P < 0.05$), **significantly different than baseline ($P < 0.005$), ‡significantly different than acute ($P < 0.05$).

Table 3. Results for the ANOVA (group (2) × time (4)) at each time point for the laboratory assessment of running biomechanics.

Variable	Group	Baseline	Acute	Post	1MO	*P*-value (group × time)
Step rate (SPM) RT prescribed SPM = 179.0 (13.0)	RT	166.5 (12.1)	179.3 (11.8) **, ††	180.8 (10.9) **, ††	180.6 (12.2)**, ††	<0.001
	CON	166.7 (10.3)	167.1 (9.9)	169.7 (10.9)	168.6 (10.2)	
Step length (cm)	RT	116.0 (14.0)	108.1 (9.9)**	106.3 (11.2)**	106.5 (12.6)**	<0.001
	CON	114.6 (13.8)	114.6 (14.0)	111.9 (12.7)	112.5 (12.2)	
COP distance from body COM at footstrike (cm)	RT	20.9 (2.9)	19.9 (2.2)*	19.0 (3.0)**	19.6 (3.1)**	0.002
	CON	21.4 (3.3)	21.1 (3.5)	21.7 (3.7)	22.3 (3.5)	
Hip flexion at footstrike (°)	RT	24.8 (7.6)	23.5 (9.3)	22.5 (6.9)	21.5 (7.9)	0.43
	CON	25.1 (9.4)	24.7 (9.3)	26.2 (6.9)	22.6 (8.5)	
Knee flexion at footstrike (°)	RT	12.7 (6.5)	13.0 (8.3)	15.9 (4.7)**,‡, †	16.8 (5.4)**,‡,†	0.05
	CON	11.2 (5.2)	9.7 (5.1)**	11.6 (3.2)	10.3 (3.4)	
Shank sagittal angle at footstrike (°)	RT	6.8 (2.4)	6.8 (3.9)	4.7 (2.3)**,‡‡	4.4 (2.2)**,‡‡,†	<0.001
	CON	6.6 (3.9)	7.8 (2.5)	6.5 (2.9	8.1 (3.0)	
Foot sagittal angle at footstrike (°)	RT	11.9 (2.7)	11.4 (5.0)	9.3 (3.9)**,‡,†	9.6 (3.9)**,†	0.05
	CON	13.1 (3.3)	12. 3 (3.5)	12.3 (4.1)	13.2 (3.7)	

Data represent group mean (standard deviation). Acute refers to the data collection that occurred on the same day as the baseline measurements at the beginning of the study. Post refers to the data collection that occurred immediately following the post-retraining phase. 1MO refers to the data collection that occurred at the conclusion of the 1-month monitoring period. The prescribed step rate corresponds to the step rate target for the retraining group.

*Significantly different than baseline ($P < 0.05$), **significantly different than Baseline ($P < 0.005$), ‡significantly different than acute ($P < 0.05$), ‡‡significantly different than acute ($P < 0.005$), †significantly different than control for same time point ($P < 0.05$), ††significantly different than control for same time point ($P < 0.005$).

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($P < 0.001$, $d = 0.67$) and distance to the centre of pressure at footstrike ($P = 0.02$, $d = 0.38$) were significantly different. However, from baseline to post-retraining and baseline to 1-month post-retraining, respectively, there were significant increases in step rate (post-retraining: $P < 0.001$, $d = 1.00$; 1-month post-retraining: $P < 0.001$, $d = 1.21$) resulting in significant reductions step length (post-retraining: $P < 0.001$, $d = 0.77$; 1-month post-retraining: $P < 0.001$, $d = 0.71$), distance to the centre of pressure at footstrike (post-retraining: $P < 0.001$, $d = 0.74$; 1-month post-retraining: $P = 0.007$, $d = 0.44$), decreased sagittal foot angle at footstrike (post-retraining: $P = 0.001$, $d = 0.79$; 1-month post-retraining: $P = 0.003$, $d = 0.70$), decreased sagittal shank angle at footstrike (post-retraining: $P = 0.001$, $d = 0.94$; $P < 0.001$, $d = 1.02$) and increased knee flexion at footstrike (post-retraining: $P = 0.03$, $d = 0.57$; $P = 0.01$, $d = 0.69$) noted in the retrainers. No change in hip flexion at footstrike was observed. There were no significant changes for those allocated to the control group.

Discussion

Based on these data, a modest increase in step rate in response to a mobile biofeedback programme resulted in reduced peak and impulse contact forces per step for the total tibiofemoral joint and medial tibiofemoral compartment during running both acutely and after retraining. There was no change in the estimated cumulative impulse at the tibiofemoral joint or medial tibiofemoral compartment in response to the training. The results of this study provide the basis for future investigations utilising this retraining method in individuals who are at high risk for the development of knee osteoarthritis, such as those who are post-anterior cruciate ligament reconstruction or post-partial meniscectomy.

The reductions in tibiofemoral joint contact forces noted during the acute gait modification trial were consistent with the reductions at post-retraining and at 1-month post-retraining. The reduction in peak and impulse tibiofemoral joint loads in the medial compartment was perhaps the most noteworthy as this compartment is most likely to develop osteoarthritis (Felson et al., 2002). Similarly, at post-retraining, peak and medial compartment tibiofemoral joint contact forces were reduced by 9.1% and 8.1%, respectively. Even at the 1-month follow-up data collection, reductions in peak and medial compartment tibiofemoral joint contact forces were still present, albeit to a lesser magnitude. Indeed, peak tibiofemoral joint contact forces at 1-month post-retraining were significantly greater than those noted during the acute gait modification trial, although these values had not reached baseline levels. The drifting of peak tibiofemoral joint contact forces towards baseline values at 1-month post-retraining may be due to changes in other kinematic or temporospatial values that were not analysed in this current study, such as peak knee flexion or changes in vertical oscillation of the centre of mass.

To our knowledge, medial compartment tibiofemoral contact force loading characteristics have not been reported during running in a similar population. Based on previous studies (Shelburne, Torry, & Pandy, 2005; Zhao et al., 2007) and presence of an internal knee abduction moment during the stance phase of running, we expected the total tibiofemoral joint axial force would be predominantly transferred through the medial compartment. It is noteworthy that there were no changes in the knee external adduction moment as a result of the retraining programme, despite reductions in medial compartment tibiofemoral joint contact force. This finding underscores the limitation of using the knee external adduction moment in quantifying changes in medial tibiofemoral compartment loads as the knee external adduction moment fails to account for sagittal plane joint loads (Manal et al., 2015; Meyer et al., 2013). As such, our results indicate that approximately 64% of the tibiofemoral joint impulse during running was transferred through the medial compartment, which is similar to the distribution measured *in vivo* during running at $1.7\ m \cdot s^{-1}$ among three individuals willing to jog after a total knee arthroplasty implanted with 6° of freedom load cells (Bergmann et al., 2014). Taken together, it is likely that interventions such as the one used in this study to reduce

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tibiofemoral joint loads during running will disproportionally affect the medial compartment.

The relative importance of peak versus cumulative loads in the development and progression of knee osteoarthritis is worthy of debate. In walking, the impulse of the frontal plane knee loads, rather than the peak load, has been suggested to be a stronger predictor of the progression of loss of medial tibial cartilage volume in individuals with active tibiofemoral osteoarthritis (Bennell et al., 2011). Interestingly, total tibiofemoral contact force impulse per unit distance is not different between running and walking (Miller, Edwards, Brandon, Morton, & Deluzio, 2014). Thus, this lack of difference in cumulative tibiofemoral joint loads between the two modes of locomotion was postulated as a reason why runners do not seem to have an increased prevalence of tibiofemoral osteoarthritis (Miller et al., 2014). Despite these assertions, cartilage is also highly sensitive to peak loads, particularly when applied at a high rate as would be expected in runners with high impact forces (Chen et al., 1999). Whether peak or cumulative joint loads play a greater role in the development of tibiofemoral OA in certain runners is a matter for further study. However, interventions to reduce peak tibiofemoral loads will likely provide greater benefit if they do not adversely affect cumulative tibiofemoral impulse contact force per unit of distance. An intervention such as the present one would need a long-term follow-up with a much larger sample size to determine if a reduction in peak and impulse tibiofemoral contact forces per stance is relevant to tibiofemoral joint osteoarthritis prevention or treatment efforts.

Regardless of the phase of the study, the retrainers maintained consistent increases in step rate and subsequent reductions step length once the step rate change was prescribed. Even during the in-field retraining sessions where runners ran on a variety of terrains, the majority of the retrainers ran with step rate values that were within 5% of their respective prescribed step rate (Willy et al., 2015). While individual in-field retraining sessions were not analysed statistically, the retraining schedule appeared to have a consistent effect on in-field step rate performance in the retraining group. Therefore, it seems that in-field step rate performance generally met the prescribed step rate levels beginning on the first retraining session. The stability of the retrainers' step rate performance across the retraining sessions suggests that a shortened retraining period may be feasible. Further, once the feedback was removed during the 30-day monitoring period, a majority of retrainers continued to maintain adherence (within 5%) with the prescribed step rate (Willy et al., 2015). The mobile monitoring during the in-field sessions is a strength of this investigation in that it ensures that retrainers were not merely "performing" during laboratory gait analyses.

To provide greater insight as to how the retrainers adopted the increase in step rate, we conducted an analysis of the kinematics at footstrike. The retrainers consistently landed with the centre of pressure closer to the body centre of mass to meet the new step rate target after the increase in step rate was introduced. Despite this early temporospatial adaptation, no consistent kinematic strategy was initially observed. Specifically, we failed to find any significant change in kinematics at footstrike in retraining group after initial training on the first laboratory visit during the acute gait modification trial. At the post-retraining data session, the kinematic strategies utilised were considerably more uniform across the retrainers. Specifically, the retrainers landed with a significantly more knee flexion, a more vertical shank segment and a flatter footstrike. This kinematic strategy was still present at the 1-month follow-up data collection. A relatively unstructured initial, kinematic pattern followed by a more stable and consistent pattern in the later stages of motor learning is consistent with the concept of skill consolidation (Brashers-Krug et al., 1996; Overduin et al., 2006). During skill consolidation, spatial coding predominates in the early stages of motor learning followed by motor encoding of kinematic pattern in the consolidation stage of motor learning (Luft & Buitrago, 2005). This finding suggests that specific feedback on kinematics may not be necessary during the retraining process to increase step rate in runners.

Limitations and direction of future study

This investigation is not without its limitations. First, healthy individuals who run with high impact forces were targeted for this intervention. Future investigations are necessary to determine if similar reductions in tibiofemoral contact forces will reduce symptoms or delay the development of knee osteoarthritis in individuals with active knee pathology, such as those who are post-anterior cruciate ligament reconstruction or post-partial meniscectomy. As such, the effect of this intervention on symptomatic knee joint pain is presently unknown. Second, our tibiofemoral model is not entirely subject-specific and relies on a lumped muscle approach, i.e., considers the individual quadriceps muscles together when estimating quadriceps muscle force (Messier et al., 2011). However, due to the within-subject design, the impact of a non-subject specific model was likely minimal on the overall results on this study. Ideally, longer follow-up periods would be required to determine if this intervention results in long-term reductions in tibiofemoral contact forces sufficient to facilitate treatment efforts for runners with tibiofemoral joint symptoms or to decrease the risk of development of knee osteoarthritis in at-risk populations. Finally, the feedback provided to the retrainers on their real-time step rate was purely visual and the frequency of feedback was controlled by the runner. However, some runners may prefer bandwidth feedback i.e., an auditory alarm issued when step rate performance falls outside the acceptable range.

Individuals with certain pre-existing conditions that are thought to increase the risk of knee osteoarthritis may benefit the most from interventions that reduce tibiofemoral joint loads. For example, partial medial meniscectomy due to a traumatic tear is thought to increase the risk of medial tibiofemoral joint compartment osteoarthritis by approximately 50% within 5–16 years after the procedure (Englund & Lohmander, 2004; Mills et al., 2008). The majority of active individuals who undergo a partial meniscectomy will attempt a return to athletic activities that involve running (Kim, Nagao, Kamata, Maeda, & Nozawa, 2013) and, therefore, may benefit from this intervention.

Conclusion

In-field gait retraining cueing a 7.5% increase in step rate during running resulted in significant reductions in tibiofemoral and medial tibiofemoral joint contact forces per stance phase, despite no changes in the knee adduction moment. Despite the increase in number of gait cycles needed to cover a given distance, there was no change in cumulative tibiofemoral joint loads. It is unknown if similar results will be found in injured populations or in those who are at-risk for the development of tibiofemoral osteoarthritis.

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ORCID

John D. Willson http://orcid.org/0000-0001-5082-8330

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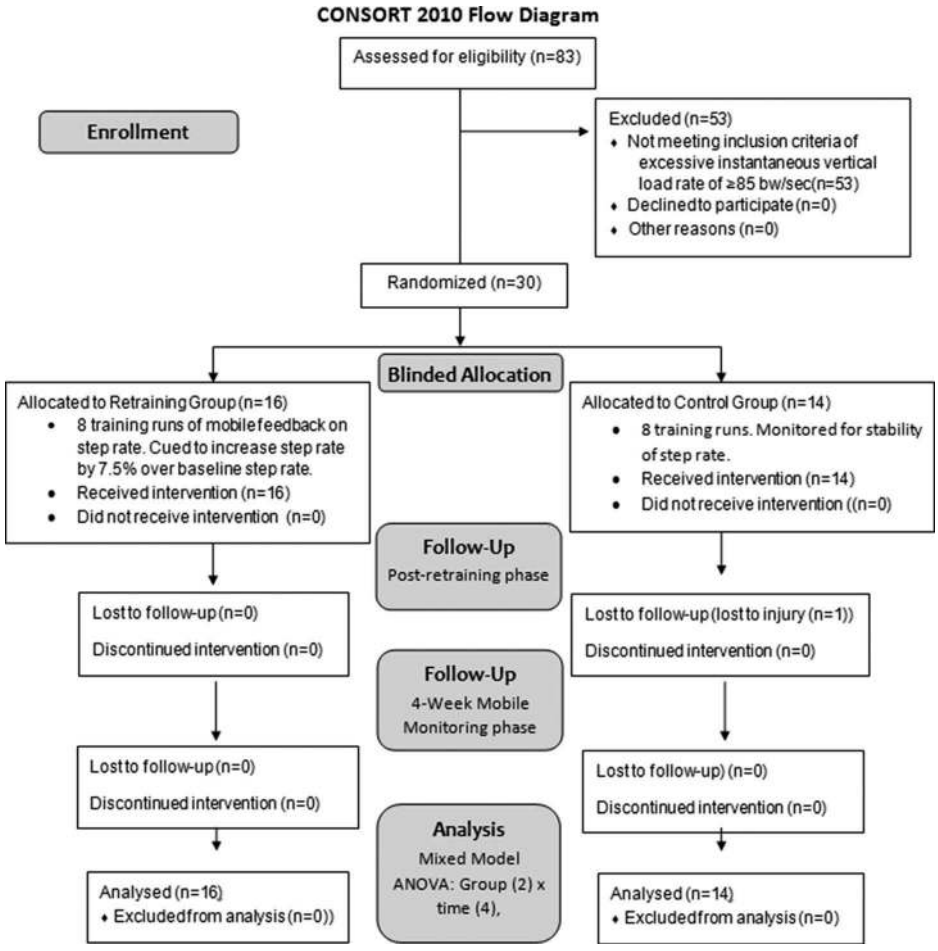
Appendix



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