

Increasing Running Step Rate Reduces Patellofemoral Joint Forces

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ABSTRACT

LENHART, R. L., D. G. THELEN, C. M. WILLE, E. S. CHUMANOV, and B. C. HEIDERSCHEIT. Increasing Running Step Rate Reduces Patellofemoral Joint Forces. *Med. Sci. Sports Exerc.*, Vol. 46, No. 3, pp. 557–564, 2014. **Purpose:** Increasing step rate has been shown to elicit changes in joint kinematics and kinetics during running, and it has been suggested as a possible rehabilitation strategy for runners with patellofemoral pain. The purpose of this study was to determine how altering step rate affects internal muscle forces and patellofemoral joint loads, and then to determine what kinematic and kinetic factors best predict changes in joint loading. **Methods:** We recorded whole body kinematics of 30 healthy adults running on an instrumented treadmill at three step rate conditions (90%, 100%, and 110% of preferred step rate). We then used a 3-D lower extremity musculoskeletal model to estimate muscle, patellar tendon, and patellofemoral joint forces throughout the running gait cycles. In addition, linear regression analysis allowed us to ascertain the relative influence of limb posture and external loads on patellofemoral joint force. **Results:** Increasing step rate to 110% of the preferred reduced peak patellofemoral joint force by 14%. Peak muscle forces were also altered as a result of the increased step rate with hip, knee, and ankle extensor forces, and hip abductor forces all reduced in midstance. Compared with the 90% step rate condition, there was a concomitant increase in peak rectus femoris and hamstring loads during early and late swing, respectively, at higher step rates. Peak stance phase knee flexion decreased with increasing step rate and was found to be the most important predictor of the reduction in patellofemoral joint loading. **Conclusion:** Increasing step rate is an effective strategy to reduce patellofemoral joint forces and could be effective in modulating biomechanical factors that can contribute to patellofemoral pain. **Key Words:** PATELLOFEMORAL PAIN, CADENCE, KNEE, STRIDE LENGTH, REHABILITATION

Running is a very popular mode of exercise around the world, with over 13.9 million people participating in the United States alone (28). Despite its health benefits, injury due to long-distance running is frequent, with studies reporting injury incidence rates as high as 79% within a 6-month period (18,34). The most common site of injury is the knee, with patellofemoral pain being the most frequent complaint (32). Prevention and treatment of this type of pain are essential for keeping runners active.

The cause of patellofemoral pain has been described as multifactorial, and many biomechanical risk factors have been identified as possible contributors. These factors include kinematic abnormalities, patellar maltracking, overuse, and excessive compressive stresses on the patellofemoral joint

cartilage (8,11,15,19,20). Net biomechanical loading at the patellofemoral joint, a major determinant of cartilage stress, is estimated to reach 4.5–7.6 times body weight (BW) during running (2,10,27,29), which is higher than most other everyday activities (2,25). Hence, finding a method to reduce the magnitude of the patellofemoral joint force during running may be effective in mitigating patellofemoral pain for runners.

Prior work has shown that increasing running step rate, while keeping a constant forward velocity, significantly alters lower extremity joint kinetics and kinematics. For example, increasing step rate by 5%–10% above the preferred will reduce stance phase knee flexion angle and knee extension torque (14). Conversely, reducing step rate, i.e., overstriding, tends to increase these variables. In addition to the biomechanical changes, this strategy has been found beneficial in reducing pain and increasing training ability in runners with patellofemoral pain (3,37). As both knee flexion and quadriceps muscle forces are recognized as primary factors affecting patellofemoral compression (39), an increase in step rate may be a simple means of modulating internal joint loading.

The objective of this study was to assess changes in muscle and patellofemoral loading with systematic variations in step rate in healthy runners. We hypothesized that

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increasing step rate would decrease patellofemoral joint force, with the net reduction arising from changes in external loading and limb posture. This hypothesis was tested by using dynamic musculoskeletal models to estimate internal muscle and joint contact loads from kinematic and kinetic measures collected during treadmill running at step rates of 90%, 100%, and 110% of the preferred.

METHODS

Participants. Thirty healthy subjects (15 males, 33 ± 14 yr, 68.6 ± 10.9 kg, and 1.75 ± 0.11 m (mean \pm SD)) agreed to participate in the study. All subjects were recreational runners (running at least $24 \text{ km}\cdot\text{wk}^{-1}$ for 3 months, $44 \pm 21 \text{ km}\cdot\text{wk}^{-1}$) who were currently pain free while running and had not previously undergone lower extremity surgery or sustained a leg injury in the past 3 months. The protocol was approved by the University of Wisconsin–Madison’s Health Sciences Institutional Review Board, and all volunteers gave written informed consent before participation.

Experimental protocol. Each participant’s preferred step rate was first determined during the final minute of a 5-min treadmill run at his/her preferred speed. Participants were then asked to perform a series of randomly ordered running trials at their preferred speed at three specified step rates: preferred step rate (100%), 10% greater than the preferred (110%), and 10% less than the preferred (90%). All step rates were maintained using an audible metronome. Whole body kinematics were recorded for 15 s during each of the running trials using a passive eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA). A total of 40 markers were tracked at 200 Hz, including 21 markers on anatomical landmarks (eight upper extremity, five pelvis, and eight lower extremity), and 14 tracking markers adhered to rigid plates that were strapped to the thigh and shank segments. Marker data were subsequently low-pass filtered at 12 Hz using generalized cross-validation splines (38). Ground reaction forces were simultaneously recorded at 2000 Hz using an instrumented treadmill (Berotec Corporation, Columbus, OH) and then low-pass filtered at 50 Hz, also using generalized cross-validation splines (38).

Computational modeling. A 3-D 29 degree-of-freedom (DOF) whole body model was used to analyze joint kinematics and kinetics during running. The pelvis was the base segment with 6 DOF. Each lower limb included a 3 DOF ball-and-socket representation of the hip, a 1 DOF ankle, a 1 DOF tibiofemoral joint, and a 1 DOF patellofemoral joint. Rolling and gliding at the tibiofemoral joint were accounted for by specifying the tibiofemoral translations and nonsagittal tibiofemoral rotations as constrained functions of the knee flexion angle (1). The patella was assumed to translate within a fixed path relative to the femur with the patella position determined assuming a constant patella tendon length (1). The hip joint center in the pelvic reference frame was then calibrated using a hip circumduction task and a functional joint center identification routine (16). All

other segment dimensions in the model were scaled to each subject using anatomical marker positions measured in a standing upright trial. These calibration trials included use of 10 additional markers, with 8 of 10 in the lower extremity. We analyzed lower extremity muscle and joint loading using a musculoskeletal model that included geometric descriptions of the patellar tendon and 92 additional musculotendon units acting about the low back, hip, knee, and ankle joints (1). Musculotendon attachment points were scaled in proportion to the factors used to scale the segments to which they were attached.

The generalized coordinates of the model were first calculated at each frame of a running trial using a global optimization inverse kinematics routine, which minimized the weighted sum of squared differences between measured and model marker positions (17). Patella translation was computed assuming that the patella tendon length was constant (5.5 cm in the nominal model) (5). Generalized coordinates were then fit with fifth-order generalized cross-validation splines (38) and then differentiated to ascertain the generalized speeds and accelerations. The equations of motion describing whole body linked segment dynamics were derived using SIMM/Pipeline (Musculographics Inc., Santa Rosa, CA) and SDFast (Parametric Technology Corporation, Needham, MA). Muscle forces were assumed to vary linearly with muscle activation from zero to the maximum isometric force for that muscle, i.e., $F = aF_o$, where a is a muscle’s activation level and F_o is the maximum isometric force (1). At each frame of motion, numerical optimization was used to estimate lower extremity muscle and patellar tendon forces by determining the activations that generated the measured accelerations at the hip, tibiofemoral, patellofemoral, and ankle joints while minimizing a weighted sum of squared muscle activations (13). The weighting factor for each muscle was taken as that muscle’s volume, which was the product of the muscle’s optimal fiber length and physiological cross-sectional area. Accelerations at the lower back and upper extremity joints were prescribed to measured values. To assess the veracity of the model estimates, the average muscle force patterns over a gait cycle were cross-correlated with ensemble EMG data collected from the vastus lateralis, rectus femoris, tibialis anterior, medial gastrocnemius, biceps femoris, semimembranosus, gluteus medius, and gluteus maximus. These EMG data were previously presented in Chumanov et al. (4) and included the subjects analyzed in this study. Cross-correlations were computed at varying lags between EMG and force to account for electromechanical delays, and the lag resulting in the peak correlation was used.

Upon computation of the muscle forces, the multibody dynamics model was queried for the net reaction force vector acting across the patellofemoral joint. The magnitude of the patellofemoral loading per unit BW was computed at each time frame. The patellofemoral loading rate was then determined by numerically differentiating (central difference) the force magnitude–time curve. The patellofemoral impulse was determined by numerically integrating (trapezoidal integration) the force magnitude–time curve. The

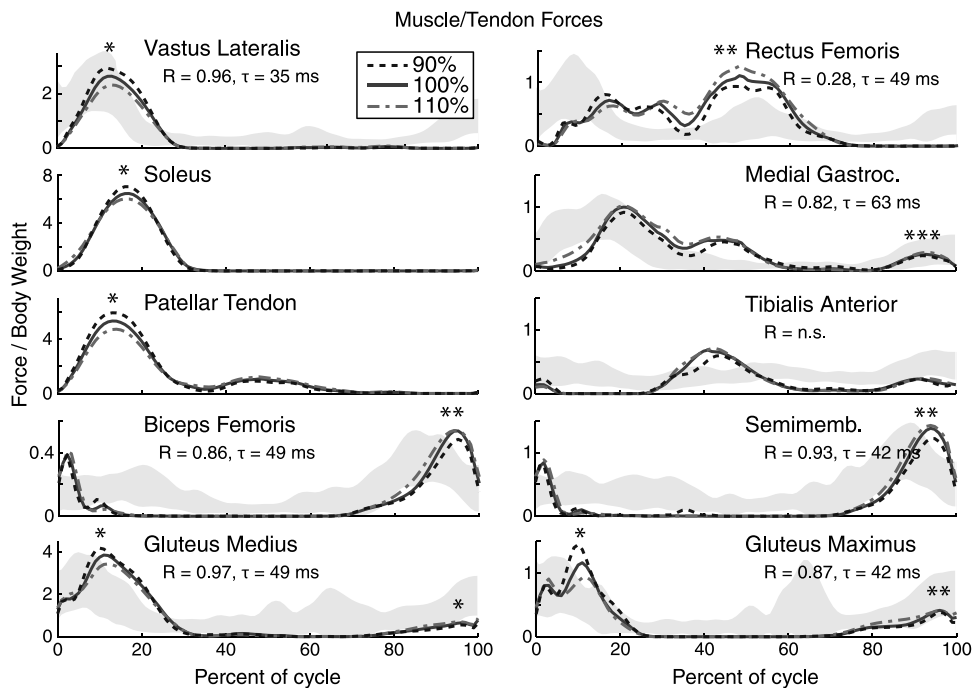


FIGURE 1—Average muscle force trajectories per unit BW across a gait cycle for each of the step rate conditions. The gray shaded regions reflect the EMG (mean \pm 1 SD) patterns measured by Chumanov et al. (4) at a preferred step rate. The peak correlations (R) and associated lag (τ) between EMG and force data are given for the average preferred step rate. All listed correlations are significant to $P \leq 0.005$ (n.s., not significant). Significant step rate effects on peak muscle forces are denoted. *All conditions are significantly different from one another, $P < 0.05$. **The 90% vs 100% and 90% vs 110% are significantly different, $P < 0.05$. ***The 90% vs 110% are significantly different from one another, $P < 0.05$.

specific metrics extracted from these curves were peak patellofemoral force, peak patellofemoral loading rate, and the patellofemoral impulse during stance. Peak muscle forces during stance and swing were also extracted from each stride and subsequently normalized to BW. Five right-footed strides were analyzed for each subject at each step rate condition.

Repeated-measures ANOVA, with both step rate and stride number being repeated factors, was used to compare peak muscle forces and patellofemoral loading metrics between the different step rate conditions. *Post hoc* analyses were performed using Tukey honest significance tests. The criterion significance level was set to $P = 0.05$. Finally, univariate regression analysis determined whether patellofemoral force was more closely associated with posture or load characteristics. Lines of best fit and coefficients of determination were computed with peak patellofemoral force magnitude as the dependent variable. Maximum knee flexion angle during stance and peak vertical ground reaction force were used as independent variables in separate analyses. All statistical analyses were performed using STATISTICA 6.1 (Statsoft, Inc., Tulsa, OK).

RESULTS

Subjects' preferred running speeds ranged from 2.4 to 3.8 $\text{m}\cdot\text{s}^{-1}$ ($2.81 \pm 0.38 \text{ m}\cdot\text{s}^{-1}$ (mean \pm SD)), and preferred step rates ranged from 156 to 192 steps per minute (174 ± 9).

Model-based estimates of muscle force patterns generally agreed well with EMG data over the gait cycle (4) (Fig. 1). Specifically, there was good temporal agreement between EMG bursts and the phasing of peak muscle forces in the vastus lateralis, gastrocnemius, and gluteal muscles during stance, with the rectus femoris during early swing and with the hamstring muscles during late swing. After accounting for electromechanical delays, the correlation between average EMG and force patterns were greatest for the vastus lateralis, medial gastrocnemius, biceps femoris, semimembranosus, gluteus medius, and gluteus maximus muscles ($R = 0.82$ – 0.97 at a preferred step rate, $P < 0.0001$). A lower, but significant, correlation was seen for the rectus femoris ($R = 0.28$), where the model predicted greatest loading in early swing, whereas the EMG data indicate greater activation during early stance and midstance. Correlations between estimated force and EMG patterns were not significant for the tibialis anterior muscle.

Step rate did not substantially alter the temporal patterns of lower extremity muscle forces, but it did modulate muscle force magnitudes (Fig. 1 and Table 1). Notably, increasing step rate led to significant reductions in vasti, gluteal, soleus, and patellar tendon forces during stance (all conditions, $P < 0.01$). In late stance/early swing, increasing step rate (from the 90% condition) led to higher rectus femoris forces (90% vs 100% and 90% vs 110%, $P < 0.05$). In late swing, increasing step rate resulted in higher peak gastrocnemius (90% vs 110%, $P < 0.001$), hamstring (90% vs 100% and 90% vs 110%, $P < 0.0005$), and gluteal muscle forces (medius: all comparisons,

TABLE 1. Peak muscle forces per unit BW, mean (SD) across all subjects

Muscle	90%	100%	110%
Vastus lateralis	3.03 (0.38)*	2.71 (0.40)	2.39 (0.43)***
Rectus femoris	1.14 (0.27)*	1.32 (0.51)	1.41 (0.32)**
Soleus	7.30 (1.26)*	6.65 (1.24)	6.18 (1.28)***
Medial gastrocnemius	0%–40%	1.05 (0.34)	1.16 (0.40)
	80%–99%	0.28 (0.09)	0.30 (0.06)
Patellar tendon	6.13 (0.88)*	5.49 (0.88)	4.87 (0.91)***
Tibialis anterior	0.65 (0.18)	0.80 (0.60)	0.75 (0.20)
Biceps femoris	0.51 (0.10)*	0.57 (0.11)	0.57 (0.10)**
Semimembranosus	1.29 (0.25)*	1.44 (0.25)	1.49 (0.24)**
Gluteus medius	0%–40%	4.31 (0.69)*	3.99 (0.47)
	80%–99%	0.65 (0.20)*	0.76 (0.24)
Gluteus maximus	0%–40%	1.52 (0.29)*	1.30 (0.27)
	80%–99%	0.43 (0.12)*	0.48 (0.13)

Percentages denoted for medial gastrocnemius and gluteal muscles represent the windows in the gait cycle where the peak was analyzed.

*Condition significantly different than the preferred (100%) with $P < 0.05$.

**The 110% condition is significantly different than the 90% condition with $P < 0.05$.

$P < 0.005$; maximus: 90% vs 100% and 90% vs 110%, $P < 0.001$). Patellar tendon force also changed with step rate. The increased (110%) step rate induced an 11% lower peak patellar tendon force in midstance than the preferred (100%) condition (4.87 vs 5.49 BW, $P < 0.0005$), whereas the decreased step rate (90%) resulted in a 12% higher peak patellar tendon force than the preferred (6.13 vs 5.49 BW, $P < 0.0005$).

Peak patellofemoral force occurred in midstance, at 14.5% ($\pm 1.9\%$) of the overall gait cycle (Fig. 2), which is well aligned with the time of peak stance knee flexion (14.8% $\pm 1.6\%$ of the gait cycle). A second smaller peak in patellofemoral loading occurred just after toe-off and corresponded with rectus femoris loading during initial swing. Peak patellofemoral force magnitude was inversely proportional to step rate, with the highest step rate (110% of the preferred) having the lowest patellofemoral force (Fig. 3). The predicted net joint force at the preferred step rate condition averaged 5.76 ± 1.02 times BW. The 110% condition has a peak force that was 14% lower (4.96 ± 1.05 BW, $P < 0.0005$), whereas the 90% step rate condition has a peak force that was 15% higher (6.64 ± 1.06 BW, $P < 0.0005$). Patellofemoral loading rate and impulse also decreased with a higher step rate. At the 110% condition, peak loading rate was 11% lower than the preferred condition (0.96 vs 1.08 BW·s⁻¹, $P < 0.005$) and stance phase impulse was 20% lower (0.51 BW·s vs 0.63 BW·s, $P < 0.0005$). Conversely, the 90% condition led to a 6.4% increase in peak loading rate compared with the preferred condition (1.15 BW·s⁻¹ vs 1.08 BW·s⁻¹, $P < 0.01$) and a 27% increase in stance phase impulse (0.81 vs 0.63 BW·s, $P < 0.0005$).

Kinematic changes were also observed, with an increase in step rate from 100% to 110% of the preferred, leading to an average of 3.3° decrease in stance phase peak knee flexion, a 2.5° decrease in ankle dorsiflexion at midstance, and a 1.8° increase in knee flexion at initial contact (all variables, $P < 0.005$; data not shown). Peak vertical ground reaction forces decreased by 2.6% when increasing step rate from 100% to 110% of the preferred ($P < 0.005$). Anterior ground reaction force also changed with step rate, with the

increased step rate leading to a 5.5% decrease in the maximum magnitude ($P < 0.005$).

Linear regression revealed that knee flexion angle was the univariate predictor most closely associated with patellofemoral force, with 68% of the variance in peak patellofemoral force being explained by the maximum knee flexion angle during stance (Fig. 4, $P < 0.001$). A 1° increase in peak knee flexion led to a 0.21 BW increase in patellofemoral force. Peak vertical ground reaction forces were less associated with the peak patellofemoral force ($R^2 = 0.28$, $P < 0.001$).

DISCUSSION

This study used a 3-D modeling-based approach to predict how muscle and patellofemoral joint forces change with step rate during running. The results support our hypothesis that increasing step rate, while maintaining speed, can

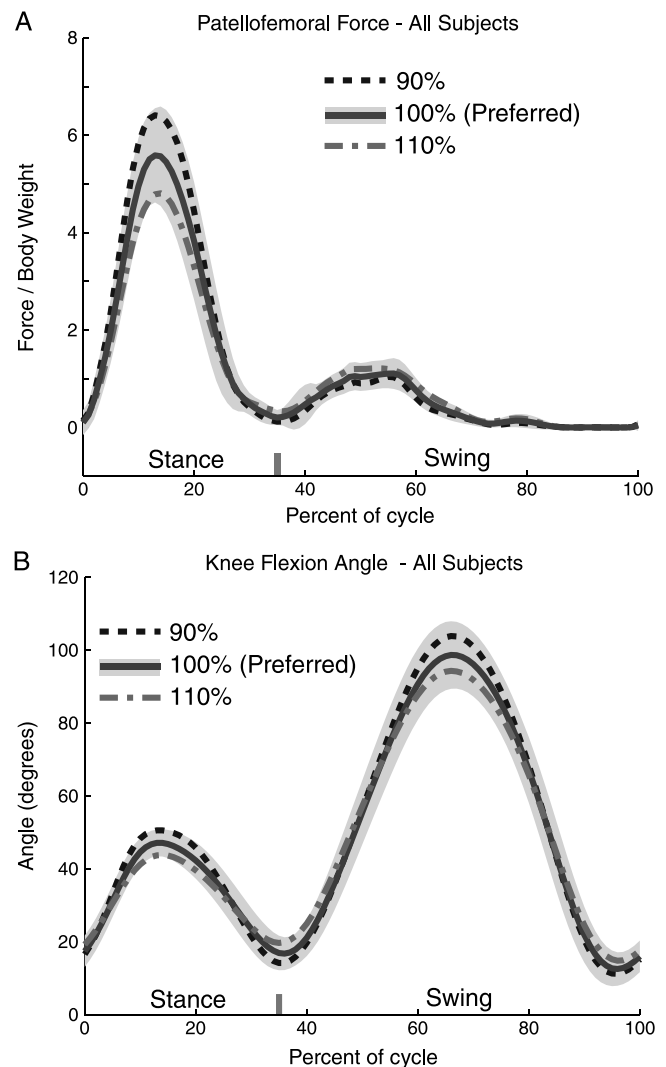


FIGURE 2—Patellofemoral force magnitude (A) and knee flexion angle across one running stride (B) averaged across all subjects. Lines represent the mean of all strides, and the shaded region represents the SD of all strides for the preferred (100%) condition. SD magnitude and profile were similar among conditions (not shown).

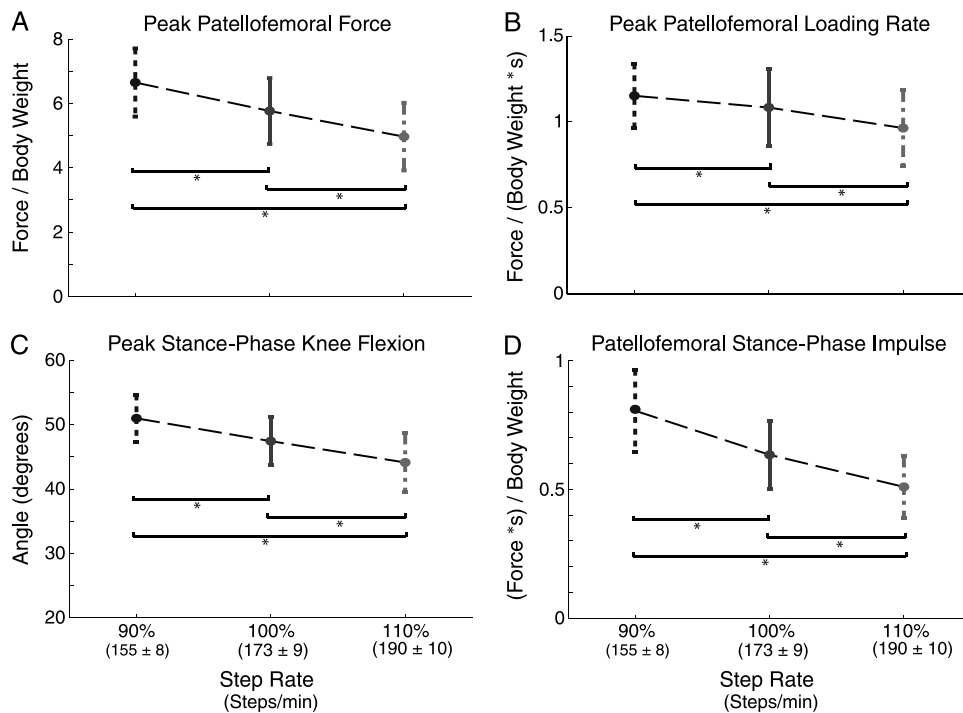


FIGURE 3—Peak patellofemoral force (A), loading rate (B), stance-phase knee flexion angle (C), and patellofemoral stance-phase impulse (D). Mean, connected dots; SD, error bars. * $P < 0.01$.

substantially diminish patellofemoral joint loading (Fig. 5). The reduction in joint compressive load primarily arises from altered muscular coordination, which places the knee in a more extended posture during midstance. The strong relationship between patellofemoral load and knee flexion (Fig. 4) indicates that posture is an important feature associated with increasing or decreasing patellofemoral force. Hence, step rate manipulation could be an effective way of addressing patellofemoral pain that arises from excessive joint loading.

Our estimates of patellofemoral joint loading during running are comparable with values reported by others. Several studies have used sagittal plane models to estimate patellofemoral forces during running, with peak forces ranging anywhere from 4.3 to 7.6 BW (10,27,29). Our average patellofemoral force estimates ranged from 5.0 (110%) to 6.6 (90%) BW across a fairly broad range of running speeds (2.4–3.8 m·s⁻¹) and step rates (140–211 steps per minute). A recent 3-D modeling study estimated peak patellofemoral forces of 5.9 BW in runners at a preferred cadence and speed (2), which is very close to our estimate of the mean force at the preferred step rate of 5.8 BW. Notably, these patellofemoral loads are much higher than that seen in other locomotor activities such as walking (0.5–1.0 BW (2,25)), stair climbing (3.3–3.5 BW (2,25)), and backward running (3.0–3.4 BW (10,27)).

In addition to these patellofemoral changes, our results suggest that increasing step rate leads to decreased extensor muscle forces during stance, an increase in rectus femoris forces in early swing, and an increase in gastrocnemius, hamstring, and gluteus maximus forces in late swing. The

decrease extensor loading in stance is likely related to the more extended limb posture adopted at higher step rates. The increase in rectus femoris and hamstring forces presumably result from the greater inertial forces involved with initiating and braking swing limb motion at higher step rates (23). The step rate influences on muscle force patterns generally mirror the EMG results reported by Chumanov et al. (4) (Fig. 1, same running speed) and Swanson and Caldwell (31) (faster running speeds), which increase our confidence that the step rate modulation effects are an accurate representation of changes in the runners' muscle coordination. Also, patellar tendon forces paralleled the patellofemoral forces, with the maximum force occurring in midstance and at the lowest step rate condition. Moreover, the pattern and magnitude of the patellar tendon force demonstrated here is similar to what has been predicted by others (29), further supporting our results. Although these findings give insight into muscular activation/force changes at different step rates, future work is needed to determine how individual muscles contribute to patellofemoral force (21).

This study focused on step rate as a method of modifying running form. We recognize that varying step rate has an inverse effect on step length when speed is kept constant, and thus, step length could alternatively be considered predictive of the biomechanical changes seen in this study. However, it is challenging to train and subsequently maintain a desired step length, especially when subjects are fatigued. Alternatively, a simple audible metronome can be used to quickly and easily vary step rate (or cadence) by small increments. This metronome, in combination with a watch or GPS to monitor overall

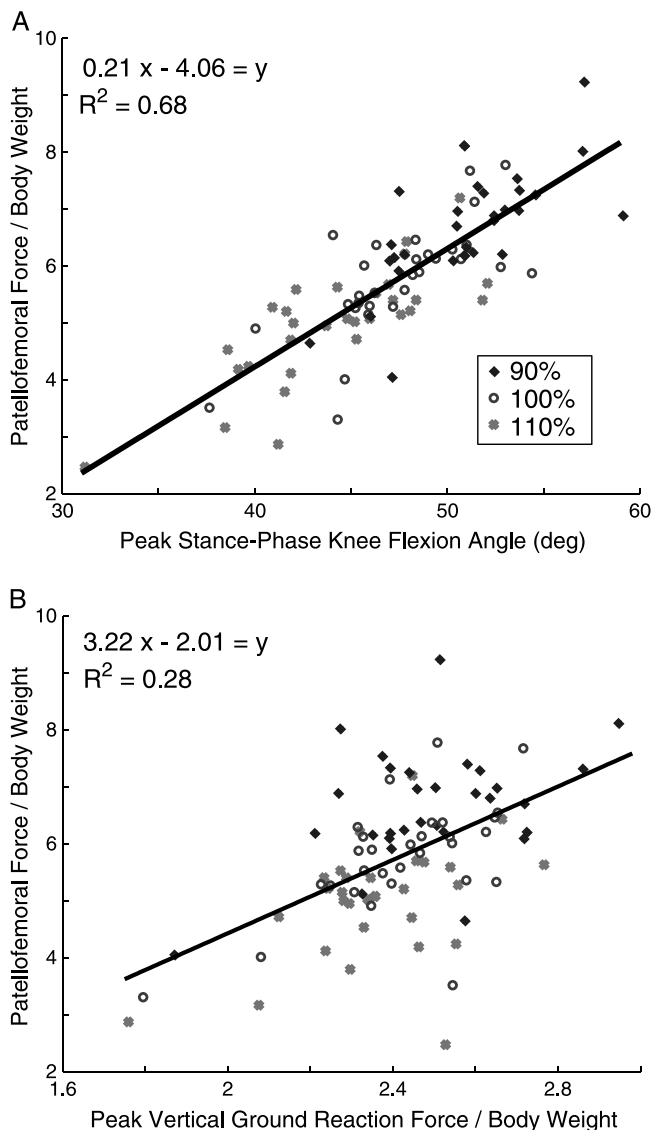


FIGURE 4—Univariate regression analyses. Patellofemoral force is plotted versus peak knee flexion angle (A) and peak vertical ground reaction force (B). Each dot represents the average for each subject for a given condition. Line of best fit is also shown along with its equation and R^2 value.

running pace, could be easily implemented to reproduce the experimental conditions.

A byproduct of step rate manipulation is a decrease in stance duration and increase in loading cycles, if running speed and distance are both maintained constant. The decrease in stance duration would act to increase patellofemoral loading rate. However, we found that increasing step rate by 10% resulted in an 11% reduction in peak patellofemoral joint loading rate, primarily because of the 14% decrease in force magnitude. To assess the influence of increased number of cycles, one can consider the net impulse accumulated over time. When we did such analyses, we found that the cumulative patellofemoral load impulse would be 5.5% lower in the 110% step rate condition, if the same distance was traversed.

Hence, we conclude that patellofemoral force magnitude, loading rate, and impulse are all diminished when increasing running step rate by 10%.

Although we have shown that we can use musculoskeletal models to estimate patellofemoral forces during running, practically, it would be beneficial to have a simpler way of doing so. Our regression analyses showed that, in fact, peak knee flexion angle at midstance is a good predictor of patellofemoral force. This implies that knee flexion could be a simple metric to monitor clinically to indirectly assess patellofemoral joint loading that exists in a runner. An increase in patellofemoral force with increasing knee flexion during activity has been described previously (7,30,33); however, the implications in running analysis were unknown. Future work will be done to determine whether this result extends beyond healthy runners to those with patellofemoral pain. Interestingly, it has been noted that those with patellofemoral pain do exhibit lower peak knee flexion during stance and is thought to represent a compensation technique (6).

Results of this study have potential clinical relevance in treating runners with patellofemoral pain. High loads have long been suggested as a possible cause of anterior knee pain (11,15), with the belief that high net loads cause large cartilage stress. Indeed, a recent biomechanical modeling study found that patellofemoral stress estimates were elevated in individuals reporting patellofemoral pain (8). Therefore, reducing patellofemoral load by modulating step rate might be an effective way to mitigate cartilage compressive stress and hence diminish pain symptoms although maintaining the ability to run. Future work will determine whether increasing step rate has the same affect on step rate in those with patellofemoral pain and if it is successful at reducing symptoms.

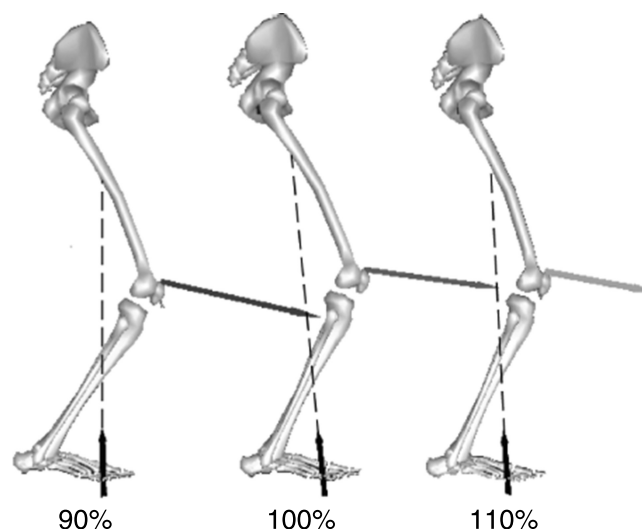


FIGURE 5—Representative subject with depiction of limb position, ground reactions, and patellofemoral joint force at the 90% (left), preferred (middle), and 110% (right) step rate conditions, shown at the time of peak patellofemoral force.

Although there are novel insights that can be gained from this study, we realize that there are some limitations to the work. One limitation of this study is the use of a 1 DOF patellofemoral joint. Some theories of anterior knee pain include a maltracking patella that would require a more sophisticated patellofemoral contact model to address. Furthermore, the lack of a cartilage contact model inhibits us from commenting on variations in cartilage tissue stress and contact area through the running cycle. Changes in contact area may be important in stress alteration (36) and possibly pain reduction. Also, we assumed that the patellar tendon was inextensible, which ignores the variations in patellar tendon line of action that can occur with tendon stretch. The patellar tendon is estimated to stretch 5%–10% with maximal quadriceps loading (12,22,24), such that small changes in quadriceps loading with step rate (12%) would likely have a small effect on patella position between step rate conditions. We performed a sensitivity study of the influence of patellar tendon length on results for a representative subject. These analyses revealed that absolute patellofemoral loads vary with patellar tendon length, but the percent difference in patellofemoral force between step rate conditions varied less than 2% for assumed patellar tendon lengths ranging from 4.5 to 6.5 cm. This result suggests that step rate variations in patellofemoral loading are relatively insensitive to the presumed patellar tendon length. Results may also have been affected by data collection methods. For convenience, all running analyses were performed on a treadmill rather than overground (9,26,35). Finally, it should be made explicit that these analyses were completed using healthy individuals, and further work is needed to ascertain if similar effects are seen in individuals with existing patellofemoral pain.

It is pertinent to note that the patellofemoral force estimates cannot be directly validated because of a lack of direct

empirical measures of this variable. However, the numerical optimization approach did generate muscle force trajectory estimates that generally agreed well with EMG patterns of the major hip and knee extensors, including the vastus lateralis, gluteus maximus, and hamstring muscles (Fig. 1). EMG–force correlation for the rectus femoris was lower ($R = 0.28$) but still significant ($P = 0.005$). The lower correlation was a result of the rectus femoris EMG data exhibiting a greater burst in early stance, whereas the model predicted relatively greater rectus femoris force at toe-off. There is potential for EMG cross-talk from the vastii onto the rectus femoris EMG recording when using surface electrodes, which could have contributed to this result. There was also no significant agreement between the tibialis anterior EMG pattern and its estimated force trajectory, but the tibialis anterior is not a major contributor to knee extensor loading such that this likely did not have a major effect on the patellofemoral loading results.

In conclusion, we have shown that increasing step rate alters running form in a way that that can significantly reduce the magnitude and rate of patellofemoral loading. This implies that a prescribed increase in step rate may be a simple strategy to attempt to mitigate patellofemoral pain that arises from excessive force.

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The authors have no conflicts of interest to disclose.

The results of the present study do not constitute endorsement by the American College of Sports Medicine.

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