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Variables during swing associated with decreased impact peak and loading rate in running

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ABSTRACT

When the foot impacts the ground in running, large forces and loading rates can arise that may contribute to the development of overuse injuries. Investigating which biomechanical factors contribute to these impact loads and loading rates in running could assist clinicians in developing strategies to reduce these loads. Therefore, the goals of our work were to determine variables that predict the magnitude of the impact peak and loading rate during running, as well as to investigate how modulation of knee and hip muscle activity affects these variables. Instrumented gait analysis was conducted on 48 healthy subjects running at 3.3 m/s on a treadmill. The top four predictors of loading rate and impact peak were determined using a stepwise multiple linear regression model. Forward dynamics was performed using a whole body musculoskeletal model to determine how increased muscle activity of the knee flexors, knee extensors, hip flexors, and hip extensors during swing altered the predictors of loading rate and impact peak. A smaller impact peak was associated with a larger downward acceleration of the foot, a higher positioned foot, and a decreased downward velocity of the shank at mid-swing while a lower loading rate was associated with a higher positioned thigh at mid-swing. Our results suggest that an alternative to forefoot striking may be increased hip flexor activity during swing to alter these mid-swing kinematics and ultimately decrease the leg's velocity at landing. The decreased velocity would decrease the downward momentum of the leg and hence require a smaller force at impact.

1. Introduction

The impact peak and loading rate of the ground reaction force during heel–toe running have been used to characterize soft tissue loading and are related to tissue injury (Hreljac, 2004). The first impact peak occurs shortly after the heel strikes the ground and may represent bone loading (Zadpoor and Nikooyan, 2011). Assessed together, the impact peak and loading rate give a measure of frequency and hence the fatigue load of the soft tissues (Hreljac, 2004). High loading (i.e. impact peak) applied at a high frequency (i.e. loading rate) can increase the potential for overuse injury (Hreljac, 2004).

A greater loading rate of the vertical ground reaction force in early stance has been associated with a history of overuse injuries such as tibial stress fractures and plantar fasciitis (Milner et al., 2006; Pohl et al., 2009; Zadpoor and Nikooyan, 2011). The impact peak has also been linked with overuse injuries in prospective studies; although, retrospective studies have not found a similar relationship (Zadpoor and Nikooyan, 2011). One potential strategy to reduce impact forces and loading rates that has been advocated is to make initial contact at the mid-foot or forefoot, as opposed to a traditional heel strike (Lieberman et al., 2010). However, shifting from a heel strike to forefoot strike pattern potentially places a greater demand on the gastrocnemius, soleus, and smaller mid-foot bones (Williams III et al., 2012). Therefore, this potential strategy may not be advisable for all runners, especially individuals with injuries such as tendonitis of the Achilles tendon and posterior tibialis (Almonroeder et al., 2013; Lieberman, 2012; Williams et al., 2000). Other potential gait modifications are needed to reduce initial impact loading. However, to better understand current gait modifications and develop new ones, it is important to understand which biomechanical factors contribute to impact forces and loading rates.

To date, there have been few reports as to what factors play a role in high impact forces and loading rates. Dynamic simulations have been used to demonstrate the influence of ankle angle, knee angle, and ankle velocity at touchdown (Gerritsen et al., 1995), as well as to highlight the importance of lower extremity touchdown velocity and mass over trunk velocity and mass (Zadpoor et al., 2007). In agreement with these simulation studies, reductions in the impact peak and loading rate that result from adopting

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a forefoot strike pattern during running have been associated with increased ankle plantar flexion at landing, increased leg compliance, and a decreased effective body mass that impacts the ground (which is dependent on the downward velocity of the foot) (Lieberman et al., 2010).

An understanding of how muscle activity affects these biomechanical variables is also important to help clinicians target muscle groups to modulate during running (Nigg and Wakeling, 2001). The downward motion of the leg in terminal-swing of gait is thought to be largely controlled by gravity and the activity of the hamstrings (Radin et al., 1991). A larger downward velocity of the ankle, along with a lack of counterbalancing quadriceps force, at contact may be associated with larger heel strike transients in walking (Jefferson et al., 1990; Radin et al., 1991). This occurs when the foot is positioned anteriorly to the knee prior to contact. Thus, increased quadriceps force would act to extend the knee and raise the foot up, decreasing the downward velocity of the lower leg and the impact force. However, it is unclear if this effect extends to running and how changing hip muscle activity directly affects the impact peak.

There are some gaps in the current literature on impact forces and loading rates during running. First, previous research has focused solely on biomechanical variables measured during the stance phase. The swing phase of gait is where the body positions the leg for impact and could yield insight into other biomechanical variables that might be related to the impact peak. Second, although muscle activation during swing has been studied (i.e. the muscle tuning paradigm) (Nigg and Wakeling, 2001), these reports considered muscle activation as a response to previous impact loading. Information on how muscle activity can be modulated proactively, rather than reactively, to alter impact loading is lacking. Thirdly, the association of segmental accelerations with the impact peak and loading rate has not been determined. Since force is related to acceleration via Newton's equations of motion, segmental velocities and accelerations may be important to consider. Finally, even though increased quadriceps activity may decrease the landing velocity and impact transients during walking, it remains unclear if this extends to running (Jefferson et al., 1990; Radin et al., 1991) or how other muscles, such as those at the hip, affect the landing velocity.

Given that the impact peak and loading rate occur early in stance, it is possible that the magnitudes are dependent on running mechanics prior to contact with the ground (Hreljac, 2004). Defining the biomechanical variables occurring prior to initial contact that are associated with the impact peak and loading rate may provide important information for the development of new gait modification strategies. Therefore, the goals of this study were twofold: (1) determine biomechanical variables during swing that predict the magnitude of the impact peak and loading rate during running; and (2) investigate how modulation of knee and hip muscle activity affects these variables. The variables of interest were the initial impact peak and loading rate, as well as superior–inferior segmental acceleration, velocity, and position at mid- and terminal-swing. We expected the impact peak to have the highest correlation coefficient with lower body segmental accelerations (i.e. Newton's equations of motion) and the loading rate to be highly correlated with the impact peak, segmental position, and velocity in terminal-swing (Lieberman et al., 2010). We also expected the knee extensors to have the greatest effect on modulating these variables such that the landing velocity is decreased (Jefferson et al., 1990; Radin et al., 1991).

2. Methods

2.1. Data collection

After written informed consent was obtained from forms approved by the university institutional review board, instrumented motion capture data was collected for 48 healthy, active subjects (16 male, 32 female, age 25 ± 6 yrs, height $1.70 \pm .08$ m, mass 65.6 ± 10.5 kg, mileage 16 ± 14 miles/week) consisting of trained and novice runners. Following a warm-up period of 5 min at a self-selected running speed, each subject ran on a treadmill at a speed of 3.3 m/s for 2 min. Forty-nine retroreflective markers were placed on each subject using a previously established procedure (Noehren et al., 2012). Twenty-seven of these markers were placed on anatomical landmarks: acromioclavicular joints, sternum, C7, L5/S1, iliac crests, anterior superior iliac spines, greater trochanters, femoral epicondyles, tibial plateaus, malleoli, and first and fifth metatarsal heads. Lower extremity motion was tracked using 22 markers placed as clusters on the thighs, shanks, and posterior heels. To minimize effects due to shoe design, all subjects wore a pair of New Balance WR662 running shoes (New Balance, Brighton, MA, USA). Three-dimensional coordinate data were measured with a 15 camera motion analysis system (Motion Analysis Corp, Santa Rosa, USA) using a sampling rate of 200 Hz. Force data was simultaneously collected at 1200 Hz using an instrumented Bertec treadmill (Bertec, Columbus, OH).

2.2. Segmental kinematics calculations

Visual 3D software (C-motion, Germantown, MD, USA) was used to filter the data, calculate a functional hip joint center (Schwartz and Rozumalski, 2005), and calculate segmental kinematics using Cardan angles. Raw marker coordinate data were filtered at 8 Hz and force data at 35 Hz using a fourth-order low-pass zero-lag Butterworth filter. The cutoff frequencies were chosen based on a residual analysis of the data (Winter, 2009).

2.3. Statistical analysis

Custom Matlab code (MathWorks Inc., Natick, MA) was first used to extract the variables of interest: the impact peak of the vertical ground reaction force and loading rate, as well as vertical position, velocity, and acceleration of the center of mass for the trunk, right thigh, right shank, and right foot at mid-swing and terminal-swing of running. Mid-swing was defined as the temporal mid-point between left foot toe-off and right foot heel-strike. Toe-off and heel-strike occurred when the vertical ground reaction force reached less than 20 N. Terminal-swing was defined as 10 ms prior to heel-strike. Loading rate was calculated as the slope of the vertical ground reaction force between 20% and 80% of the period between heel-strike and the impact peak. Data were collected from five trials for each subject and then averaged. Next, SPSS (SPSS Inc., Chicago, IL) was used to check for normality of the variables according to the Kolmogorov–Smirnov test with a Lilliefors significance correction and subsequently calculate Pearson's correlation coefficients between the kinematic variables, impact peak, and loading rate. *P*-values less that .05 were deemed statistically significant. To determine the level of variance in both impact peak and loading rate that was explained by the kinematic variables, forward stepwise multiple linear regressions were performed. The variables that were significantly correlated with the impact peak were input into one regression model. The variables that were significantly correlated with the loading rate were then input into a separate regression model. Only the variables with the four biggest effect sizes were used in each linear regression model to achieve a reasonable statistical power ($\beta \geq .8$) with the number of subjects gathered (i.e. 10 subjects needed per variable (Field, 2009)). For example, if five variables were significantly correlated with loading rate, only four were chosen, based on effect size, for the linear regression model. Pearson's correlation coefficients between variables input in the linear regression models were used to check for multicollinearity by ensuring Pearson's coefficients were less than .8.

2.4. Muscle analysis

A muscle-actuated forward dynamic simulation, developed by Hamner et al. (2010), was used to assess how muscle activity modulates those variables that significantly predicted the impact peak and loading rate. The model consisted of 92 muscles and 13 joints (Table 1): bilateral shoulder joints represented as a ball-and-socket, bilateral elbow joints modeled as a two revolute joints, trunk–pelvis junction as a ball-and-socket, bilateral hips as a ball-and-socket, bilateral knee joints as a custom revolute joint with a translating axis, and bilateral ankles modeled as a revolute joint. This model has been previously used to estimate muscle excitations and forces for a running cycle of a 65.9 kg male, with the muscle excitations validated against electromyography measurements (Hamner et al., 2010). This data was accessed from an open-source repository on SimTk. During the swing phase of running, the activation of all the muscles of a single muscle group were increased 2.5% from the nominal simulation of Hamner et al. (2010). The increased muscle activity was done one group at a time. This perturbation level was chosen to be high enough to elicit a response yet low enough to avoid physiologically unrealistic motion. The groups considered were hip extensors, hip flexors, knee flexors, and knee extensors. The ankle muscles were not considered as we were interested in alternatives to forefoot striking, which utilizes ankle muscles (Williams III et al., 2012). The muscles included in each group were the same as the categorization used in the model (Table 1) (Hamner et al., 2010). Using OpenSim (Delp et al., 2007), a Hill-type model was used to determine the force generated by