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Locomotor Adaptations During RaceRunning in People With Neurological Motor Disorders

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The aim of this study was to examine strategies to absorb impact shock during RaceRunning in participants with neurological motor disorders. For this purpose, eight RaceRunning athletes (four males and four females) voluntarily took part in this study. Each participant performed a series of 100-m sprints with a RaceRunning bike. Acceleration of the tibia and head was measured with two inertial measurement units and used to calculate foot impact shock measures. Results showed that RaceRunning pattern was characterized by a lack of impact peak in foot–ground contact time and the existence of an active peak after foot contact. Due to the ergonomic properties of the RaceRunning bike, shock is attenuated throughout the stance phase. In conclusion, the results revealed that RaceRunning athletes with neurological motor disorders are capable of absorbing impact shock during assisted RaceRunning using a strategy that mimics runners without disabilities.

Keywords: acceleration, running pattern, shock absorption

An estimated 6% of people in the United Kingdom have some form of neurological motor disorder (MacDonald, Cockerell, Sander, & Shorvon, 2000) that affects participation in health-related activities and poses a challenge for society to promote the health and well-being of all its members (Coates & Vickerman, 2010; Kiuppis, 2018). Indeed, it is reported that 70–75% of people with disabilities do not participate in any sport or physical activity (Sport England, 2013), and this increases the risk of developing cardiovascular and metabolic diseases (Ryan et al., 2014), muscle weakness (Wiley & Damiano, 1998), and

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reduced bone density (Cohen et al., 2009). Few sports currently exist for those with a severe neurological disability, and these are often limited to low levels of aerobic demand and weight-bearing demands (Van der Linden, Jahed, Tennant, & Verheul, 2018).

One activity that is fully customized for people with moderate to severe neurological motor disorders is RaceRunning. This activity enables people with motor disorders who are not independently ambulant or able to use a manual wheelchair to participate with the use of a customized RaceRunning bike (Van der Linden et al., 2018). A RaceRunning bike has three wheels (in a triangular orientation), a saddle, a chest plate, and two handlebars to regulate the user's postural control and balance while engaging in locomotor patterns of walking and running (see Figure 1). It is estimated that during a 6-min RaceRunning trial, heart rates can reach up to 55% of maximum heart rate, reflecting its valuable potential role in inducing cardiovascular benefits in people with motor disorders (Bolster, Dallmeijer, de Wolf, Versteegt, & van Schie, 2017), while encouraging regular participation in physical activity.

In addition to health-related benefits for people with neurological disorders, RaceRunning has become a competitive sport promoted internationally by the Cerebral Palsy (CP) International Sports and Recreation Association, formally structured in World and European competitions (World Para Athletics, 2016). However, few investigations currently exist on the mechanics of RaceRunning gait

Figure 1 —A RaceRunning bike.

patterns and the adaptive mechanisms responsible for regulating postural stability during performance. This information is key for the development of grassroot and elite coaching to enable safe and effective participation and training programs to be developed. Specifically, understanding foot strike patterns and related shock absorption mechanisms during RaceRunning may enhance performance and reduce the risk of potential injury.

Research on running-related injuries in those without a disability has shown how functional adaptations protect the body from the impact of specific environmental and task constraints (Derrick, Hamill, & Graham, 1998; Gruber, Boyer, Derrick, & Hamill, 2014; Mizrahi, Verbitsky, & Isakov, 2000a, 2000b). For example, impact shock emerges with each foot–ground collision during running (Derrick et al., 1998), leading to high ground reaction forces during the stance phase of running. The events surrounding these collisions are the main source of impact shock, which is transmitted through the leg and rest of the body (Gruber et al., 2014).

One important effect of impact shock is the rate (speed) of the shock impulse that is transmitted during the stance phase (Derrick et al., 1998). The shock impulse can be absorbed immediately after the point of foot–ground contact (high frequency shock absorption) or slightly after the point of contact (low frequency shock absorption). The frequency of this type of impact shock will depend on both the magnitude and timing of the vertical high-ground reaction forces, which has been shown to change depending on footfall pattern (Gruber et al., 2014). This may be a significant contributor to running-related injuries, since the capacity of certain tissues to transmit and attenuate shock may be frequency dependent (Smeathers, 1989).

The frequency content and signal power of impact shock and tibia acceleration during stance phase of normal running are thought to be governed primarily by movement of the leg and center of mass (Bobbert, Schamhardt, & Nigg, 1991). Specifically, tibia acceleration in those with a rearfoot strike pattern contains low frequency ranges, representing lower extremity motion and vertical acceleration of the center of mass during the stance phase (Gruber et al., 2014). Conversely, higher frequency ranges represent a rapid deceleration of the foot and leg at initial contact (Derrick et al., 1998). The low and high frequency ranges are representative of the active peak and impact peak of the vertical high-ground reaction forces, respectively. In the time domain, the shock impact can change at different moments of the stance phase, indicating the ability of the body to absorb shock as the center of mass is moving forward (Shorten & Winslow, 1992).

Impact shock must be attenuated during running to prevent disruption to the vestibular and visual systems, as a result of rapid head acceleration (Derrick et al., 1998; Edwards, Derrick, & Hamill, 2012). Attenuation occurs mainly through active and passive energy absorption mechanisms, such as muscle activation, changes in joint angle, and deformation of passive structures. In runners without disabilities, the body responds to greater impact by increasing attenuation through a combination of these active and passive mechanisms (Radin, 1972; Radin & Paul, 1970). However, changes to the neuromuscular system in people with motor disorders could impact their ability to attenuate shock during running for a number of reasons. For example, Van der Linden et al. (2018) showed how lower limb spasticity, weak leg strength, poor voluntary motor control, and reduced passive knee extension affected performance during 100-m RaceRunning. In particular, plantar flexor muscle weakness may change the spring-like action of the foot and ankle (Olney, MacPhail, & Hedden, 1990), while poor voluntary control may leave athletes vulnerable to excessive and uncontrolled impact shocks from the accelerating body during locomotion.

The emergence of functional movement adaptations to control posture and attenuate impact shock during running is an important aspect of coaching competitive and recreational athletes (Shorten & Winslow, 1992). One functional movement adaptation that may result from shock attenuation concerns changes to the kinematics and kinetics of movements by the athlete (Frederick, 1986), which may serve to reduce the impact shock on the musculoskeletal system. For example, kinematic change is exemplified by knee joint displacements (15–45°) during the entire impact phase, but it has a significant role (40–45°) at midstance (Derrick et al., 1998). The kinetic changes emerge in the temporal patterning of peak shock values (Derrick et al., 1998). The quality of such adaptive strategies defines skilled locomotion and underpins health and safety for athletes of all skill levels (Davids, Button, & Bennet, 2008). It is unclear whether individuals with neurological injury adapt the magnitude and frequency of running impact shock to their physical constraints in the same way as runners without disabilities attenuate impact shock during RaceRunning. Therefore, the purpose of the study was to determine the timing and frequency of impact shock and attenuation in people with motor disorders during 100-m RaceRunning.

Methods

Participants

Eight (four males and four females) competitive athletes (age: 18.6 ± 2.8 years; body mass: 50.2 ± 6.9 kg; and height: 168.8 ± 7.7 cm) at different levels of RaceRunning (RR2: n = 4 and RR3: n = 4), according to CP International Sports and Recreation Association classifications, volunteered to participate in this study. A classification of RR2 involves athletes with spasticity, athetosis, ataxia dystonia, or weakness, which limits the effective pushing movements of the lower extremities. Those athletes classified as RR3 have mild to moderate involvement in one or both upper extremities, fair to good trunk control, and moderate involvement of the lower extremities. Participants included were those with a diagnosis of a neuromuscular disorder, including six athletes with spastic CP and two athletes with acquired brain injuries. CP is defined as a nonprogressive motor disorder that affects posture and movement and often appears in the early years of life (Griffiths & Clegg, 1988). On the other hand, acquired brain injury is described as the outcome of a traumatic injury due to hemorrhage or a cerebral swelling (Campbell, 2004). All athletes were free from any musculoskeletal injury during data collection. Participants' level of ambulation was assessed using the Functional Mobility Scale (Graham, Harvey, Rodda, Nattrass, & Pirpiris, 2004). The scale required participants to attempt to walk 5, 50, or 500 m, while ratings of 1–6 were recorded according to if, and how (independent, walking aids, and wheelchair), these distances were completed (1 = *wheelchair use*, 6 = *independent walking*). All participants in this study were rated at 6 for the 5-m test, 6 for the 50-m test, and 1 for the 500-m test, representing their ability to walk independently over short distances. For inclusion to the study, participants were required to have experience of participation and practice in RaceRunning at a competitive level (mean experience: 3.0 ± 0.7 years) and with an international ranked classification, according to the CP International Sports and Recreation Association. Participants who fit the inclusion criteria were recruited from a local RaceRunning club, and all measurements were carried out at the 400-m athletics track at the RaceRunning club where participants trained. Participants provided written informed consent in the presence of their carers. The study was approved by an institutional University research ethics committee.

Experimental Setup

Two low-mass (<3 g) inertial measurement unit sensors (MetaMotion R; MBIENTLAB Co., San Francisco, CA) were used to measure impact shock in the magnitude and frequency domains during the trials. The sensors contained a three-axis accelerometer, three-axis gyroscope, and three-axis magnetometer. Each sensor was equipped with Bosch Sensortec (Stuttgart, Germany), which combines the measurements of the accelerometer, gyroscope, and magnetometer to provide a robust calculation of the orientation vector. One sensor was attached to the center of the forehead of the participant and another to the medial–distal part of the right tibia to reduce the effect of soft tissue vibration. The unobtrusive sensors were secured by double-sided tape and Velcro adjustable straps (Presco, Swindon, United Kingdom). The axis of each sensor was aligned with the vertical axis of the lower leg while the participant was standing (X: medio- lateral; Y: superior–inferior; Z: anterior–posterior). The sensors sampled movements at a frequency of 400 Hz. For detecting the stance phase, the gyroscope and accelerometer of the tibia sensor were synchronized. These sensors have previously been validated for use in different activities (McGrath, Green, O'Donovan, & Caulfield, 2012).

Procedure

Data collection took place at an indoor athletics track. Participants wore their usual training clothes and running shoes. Each participant used a RaceRunner bike, which was adjusted and scaled according to body size (Petra Cross Runner; Quest 88 Ltd., Shropshire, United Kingdom). Prior to the sprint trials, each participant performed a warm-up routine, which began with stretching exercises, followed by short walking and low-velocity running (10 m slow pace) with the bike. The coach supervised the warm-up phase and spent the time equally with all participants. The nature of stretching exercises was a combination of static stretches followed by partner-assisted stretches. For the experimental trials, participants were asked to replicate a 100-m race in two groups of four by initiating the run from the start line and sprinting the straight to the 100-m finish line. After a rest period of 7–8 min, experimental race trials were repeated until participants completed four trials each.

Data Analysis

Data from Trials 2 and 3 were averaged and used for further analysis to reduce possible fatigue effects. Because sprint speed was not constant throughout the 100-m race, accelerometers data for each participant were analyses over the middle part of the 100-m race, between 30 and 70 m, to exclude the acceleration and deceleration phases without participants' awareness. Accelerometers data during the stance phase of running over 20 successive strides were used for subsequent analysis.

Impact Shock Magnitude. The stance phase, defined from the point of initial contact to toe-off, was calculated by using a gyroscope in a sensor attached to the tibia, according to the methods of McGrath et al. (2012). Both initial contact and toe-off points were detected when the angular velocity of the tibia (in degrees per second) reached its minimum value in the *x* axis. These critical events in the stance phase were synchronized with accelerometers data from both the tibia and the forehead. Raw accelerometers data were filtered using a second-order Butterworth low-pass filter with a cutoff frequency of 40 Hz, after removing gravity (g = 9.81 m/s^2) in the raw signal. The stance phase of successive strides was normalized by using a spline interpolation method (0–100%).

The impact shock magnitude value was calculated in the stance phase by recording values of tibia acceleration (g), head acceleration (g), peak positive tibia acceleration, and peak positive head acceleration (Gruber et al., 2014). All analyses were performed using a custom-written MATLAB program (MathWorks, Inc.).

Frequency Domain Analysis. The power of acceleration value in the stance phase was calculated through Fast Fourier Transformation for the power spectral density (PSD) analysis (Derrick et al., 1998). The advantage of applying PSD to the raw acceleration signal is its sensitivity to detect the mechanisms of shock absorption in a frequency window. This is important because the magnitude of peak shock throughout the stance phase could differ as the result of the implementation of different shock absorption strategies by individual participants, based on the available time (Gruber et al., 2014).

The PSD analysis was performed on frequencies 0 to the Nyquist frequency and normalized to 1 Hz bins (Derrick et al., 1998). After binning, the PSD was normalized in order for the sum of the powers from 0 to Nyquist frequency to be equal to the mean squared amplitude of the data in the time domain. There were two frequency domains in this study: lower (3–8 Hz) and higher ranges (9–20 Hz). These frequencies were based on the data of forefoot runners, which broadly represent the typical footfall patterns of those with motor disorders (predominantly a lack of heel contact) in this study (Gruber et al., 2014). The frequency domain parameters were calculated for tibia power frequency (TPF), head power frequency (HPF), tibia power magnitude, and head power magnitude at both low and high ranges. The TPF and HPF represent the frequency at which peak acceleration emerged, and tibia power magnitude and head power magnitude represent the magnitude of shock in this frequency.

The shock transfer function (TF) was defined as the amount of shock transmitted from the tibia to head (Derrick et al., 1998; Gruber et al., 2014) and was calculated according to the following equation:

$$\mathrm{TF} = 10 \times \log_{10}(\mathrm{PSD}_{\mathrm{head}}/\mathrm{PSD}_{\mathrm{tibia}})$$

The TF value between the tibia and the head was calculated for both frequencies (in decibels), signaling either gain or attenuation. Positive values indicate gain, or increase in signal strength, from the tibia to the head, whereas negative values indicate attenuation, or decrease in signal strength. The time to complete the 100-m race was also recorded for each trial and divided by the distance to calculate average race speed. The average race speed from each trial was used as a performance outcome measure for each participant.

Results

Impact Shock Magnitude

Data on the magnitude of acceleration in the tibia and head during the stance phase are presented in Figure 2. Results showed that the peak positive tibia acceleration was highest at the initial 10% of stance (4.58 ± 3.33 m/s^2). The peak positive head acceleration value was highest (0.55 ± 0.3 m/s^2) during 15–20% of the stance phase.

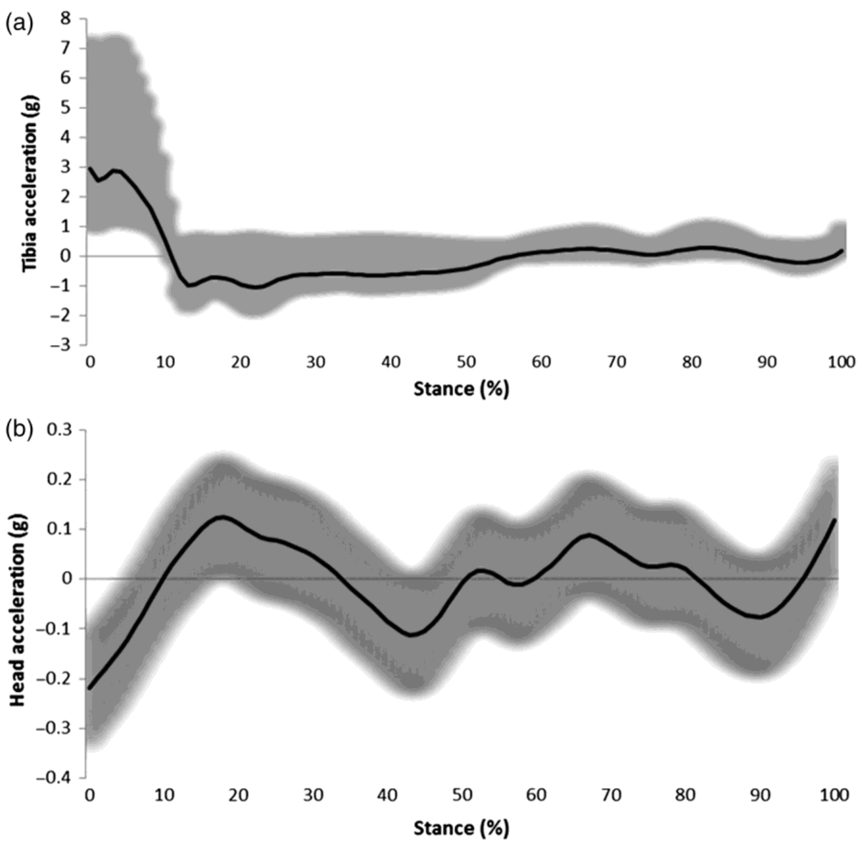

Figure 2 — Tibia (top) and head (bottom) acceleration during the stance phase.

Frequency Domain

The results of the PSD in the tibia and head attenuation ratio are presented in Figure 3. The peak acceleration value was mainly observed in the lower frequency ranges for both tibia ($TPF_{low} = 3.62 \pm 0.744$ Hz) and head (HPF_{low}: 3.5 ± 0.92 Hz).

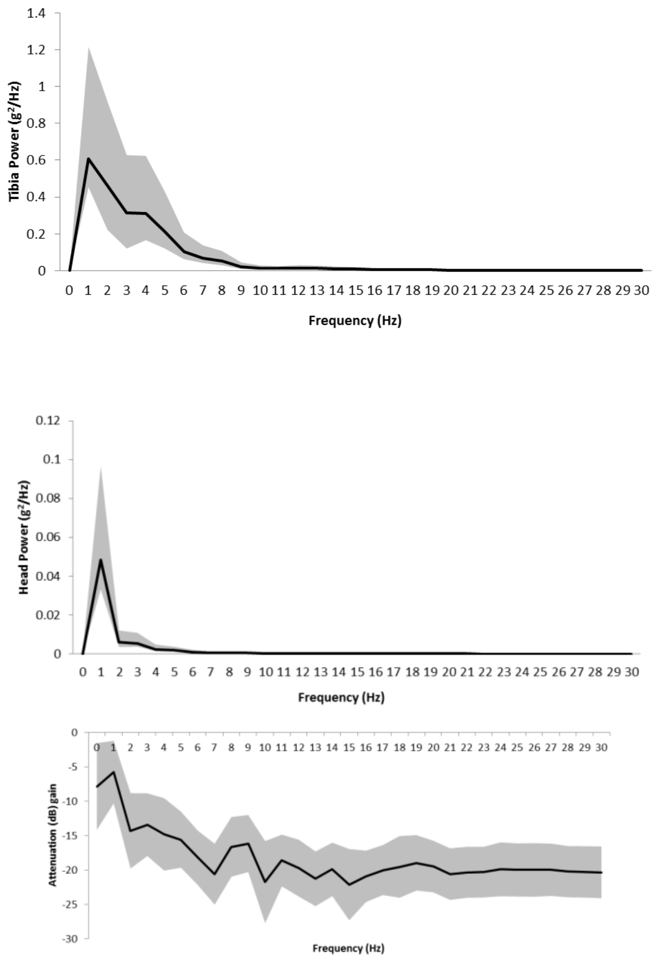

Figure 3 — The PSD in tibia (top), head (middle) and attenuation ratio (bottom) in the frequency domain.

The magnitude of tibia acceleration strength in the lower frequency ranges (0.177 ± 0.22 g^2/Hz) was greater than in the higher frequency ranges (0.0102 ± 0.10 g^2/Hz). The high frequency ranges in tibia acceleration emerged (TPF_{high} = 10.25 ± 1.28 Hz) slightly later than the acceleration of the head (HPF_{high} = 9.37 ± 0.74 Hz). The magnitude of head acceleration strength in the lower frequency ranges (0.002 ± 0.001 g^2/Hz) was greater than in higher frequency ranges (0.0001 ± 0.0001 g^2/Hz).

Results of TF analyses also showed that tibia shock was attenuated throughout the stance phase. More specifically, the TF value in low frequency (−16.56 ± 11.91 dB) and high frequency ranges (−19.88 ± 11.07 dB) did not differ. The mean frequency value of shock attenuation in the lower ranges was 6.37 ± 1.3 dB and in higher ranges was 14.62 ± 3.62 dB.

Discussion

The aim of this study was to examine the strategy by which athletes absorb impact shock during a 100-m RaceRunning sprint race in people with motor disorders. The findings of this study showed that the impact shock absorption pattern in RaceRunning is characterized by an active peak only at the initial stance phase. In addition, the participants were able to absorb impact shock throughout the entire stance phase. Tibial acceleration and impact power in the lower frequency ranges were similar to values observed in previous studies on forefoot runners without disabilities (see Table 1 for comparison between RaceRunning and forefoot runners without disabilities in a study by Gruber et al., 2014) demonstrating a similar amount of peak positive tibial acceleration relative to forefoot runners without disabilities (3.87 ± 1.36 m/s^2) at similar speeds.

This equivalent result of a forefoot running pattern is unsurprising given the toe-walking gait pattern observed in those with CP (Holt, Obusek, & Fonseca, 1996). The similarity of impact shock patterns recorded from the RaceRunning athletes compared with forefoot runners without disabilities is based on the observation that they create the same pattern of impact power in both the tibia and forehead parts of the body. Running patterns in both groups are characterized by an active peak at low frequency ranges, which indicates that foot placement and center of mass are adapted to accelerate the body forward, moments after the point of foot–ground contact. This movement organization strategy serves to reverse the downward velocity of the center of mass (Laughton, McClay Davis, & Hamill, 2003). Despite similarities in the existence of active shock, the two groups differed in the amount of frequency displayed in the lower ranges (RaceRunning: 3.62 vs. forefoot running: 7.2 Hz). This finding suggests a movement pattern that is constrained by characteristics such as spasticity and muscle weakness, resulting in an inability to absorb the impact shock in the whole area of the sole of the foot. This observation might indicate the role of body inclination through changes in the bike ergonomics and acquisition of a functional foot placement in this adapted sport. Further study is required to understand the underlying mechanisms for this difference.

In contrast to forefoot runners without disabilities, an impact peak at higher frequency ranges was not evident in the RaceRunning athletes in this study. This

Table 1-Time Domain and Frequency Domain Characteristics of RR Athletes Compared With Previous Data on Able-Bodied FF Runners[a]

Parameters	RR athletes	FF runners
PPA (g)	4.58 ± 3.33	3.87 ± 1.36
HP (g)	0.55 ± 0.30	0.47 ± 0.19
TPF_{low} (Hz)	3.62 ± 0.744	7.2 ± 1.5
TPF_{high} (Hz)	10.25 ± 1.28	10.7 ± 2.8
HPF_{low} (Hz)	3.5 ± 0.92	4.3 ± 0.5
HPF_{high} (Hz)	9.37 ± 0.74	11.8 ± 2.8
TSM_{low} (g^2/Hz)	0.177 ± 0.22	0.158 ± 0.101
TSM_{high} (g^2/Hz)	0.0102 ± 0.100	0.248 ± 0.253
HSM_{low} (g^2/Hz)	0.002 ± 0.001	0.235 ± 0.085
HSM_{high} (g^2/Hz)	0.0001 ± 0.0001	0.041 ± 0.018
TF_{low} (dB)	−16.56 ± 11.91	18 ± 21.3
TF_{high} (dB)	−19.88 ± 11.07	−88.7 ± 40.1
TFF_{low} (dB)	6.37 ± 1.3	6.9 ± 0.9
TFF_{high} (dB)	14.62 ± 3.62	28 ± 1

Note. Variables included: peak positive tibial acceleration (PPA); head acceleration peaks (HP); frequency of peak power of the tibial ($TPF_{low,high}$) and head ($HPF_{low,high}$) acceleration signal within the lower and higher frequency ranges; signal power magnitude of the tibial ($TSM_{low,high}$) and head ($HSM_{low,high}$) acceleration signal within the lower and higher frequency ranges; shock attenuation magnitude in the lower (TF_{low}) and higher (TF_{high}) frequency ranges; shock attenuation frequency in the lower (TFF_{low}) and higher (TFF_{high}) ranges. RR = RaceRunning; FF = forefoot; TF = transfer function.

[a]The results are extracted from Gruber et al. (2014) for comparison with data collected in this study.

adaptation in runners without disabilities could ensure that the shock is absorbed smoothly from the entire sole of the foot, but in RaceRunning athletes a different adaptive mechanism was employed, affected by their continuous interactions with the bike. Another responsible factor might be varied joint kinematics, such as ankle plantar flexion to place the foot flat on the ground and increased knee flexion angle during the entire impact phase (Derrick et al., 1998; Edwards et al., 2012), observed in people with motor disorders due to increased muscle stiffness and excessive muscle weakness (Van der Linden et al., 2018). The dominant frequency value in the tibia (TPF) was similar to that observed at the head (HPF), at both low and high frequency ranges, which differed slightly from observations of forefoot runners without disabilities. A key difference was that the dominant tibia frequency of forefoot runners without disabilities at lower ranges emerged later in the gait cycle than in the RaceRunning athletes (7.2 vs. 3.62 Hz, respectively). In contrast, the higher frequency ranges emerged at almost a similar point (10.7 vs. 10.25 Hz, respectively). For the head, the frequency ranges were similar between the RaceRunning athletes and those previously reported for forefoot runners without disabilities at both low (3.5 vs. 4.3 Hz) and high (9.37 vs. 11.8 Hz) ranges. The incident of peak impact in the tibia emerged faster than at the head in the forefoot runners without disabilities. The lack of temporal interval in the incident of peak impact between the tibia and head in the RaceRunning athletes could be associated with body adaptations to the ergonomic design of the bike, such as more stable balance provided by three wheels, maintained by using the saddle and chest plate. Another finding of this study indicated that the RaceRunning athletes were able to attenuate impact shock through an active peak as an active attenuation mechanism during the stance phase, contrasting with the pattern of forefoot runners without disabilities, and approximating the strategy of rearfoot runners (Derrick et al., 1998; Gruber et al., 2014). The RaceRunning athletes in this study did not show any gain in TF (positive shock power) from the tibia to head at any moment of the stance phase. Work on runners without disabilities by Gruber et al. (2014) showed that values of TF increased in forefoot and rearfoot runners at low frequency ranges and were attenuated at high frequency ranges. However, the mean TF was negative in the lower frequency ranges in rearfoot runners, but there was a high *SD* value in the group data, signifying a high level of inter-individual variability. The lack of shock attenuation observed in the forefoot running pattern is a result of vertical oscillation of the center of mass and joint flexion when generating power at the low frequency ranges, leading to a higher level of shock power in the head (Gruber et al., 2014). However, this was not observed in this study, despite the forefoot running pattern observed in the RaceRunning athletes.

This may be due to their interactions with the ergonomic design of the RaceRunning bike that is equipped with a saddle and a chest plate, which is beneficial for absorbing impact shock that would otherwise be transmitted to the trunk and head. It is important to note that the difference in values of TF observed at low frequency ranges between the RaceRunning and rearfoot and forefoot running patterns could not be related to variations in running task constraints, that is, differences in running speed (sprint running vs. distance running). For example, Mercer, Vance, Hreljac, and Hamill (2002) showed that, despite a positive linear trend between sprint running speed and shock attenuation, the magnitude of shock was positive (ranges 0.06–0.15) at all running intensities (50–100% max speed).

Another interesting difference in the TF data was related to the frequency domain. In fact, the dominant higher frequency in the RaceRunning pattern was half of the value typically observed in forefoot runners without disabilities (14.62 vs. 28 Hz). In the lower frequency ranges, the difference was trivial (6.37 vs. 6.9 Hz). The similarity of lower dominant frequency for shock attenuation between the RaceRunning and forefoot running patterns highlights the common mechanisms that control footfalls, such as a short stance time and a lack of heel contact (lack of impact peak). In contrast, the differences observed in the higher dominant frequency could be related to those parameters that affect active shock attenuation after foot–ground contact time, such as eccentric muscle contraction (Gruber et al., 2014; Radin, 1972), muscle stiffness (Boyer & Nigg, 2007), and joint kinematics (Edwards et al., 2012). The RaceRunning athletes displayed a lack of voluntary control, muscle stiffness, and spasticity (Van der Linden et al., 2018) that might negatively affect the temporal pattern of shock absorption during the stance phase, unless these athletes were able to adapt to these physical characteristics.

This study is the first to analyze body adaptations of RaceRunning athletes to external force loadings (e.g., foot impact contact). The findings have some implications for adapted sports and disability running, revealing a functional form of movement adaptation in people with neurological motor disorders during performance. The observed adaptations to body impact shock revealed the signatures of adapted running patterns in RaceRunning, compared with other types of locomotion, such as forefoot, midfoot, and rearfoot running patterns. In all types of locomotion, skilled performance is characterized by distinct performance features related to absorption and attenuation of impact shock. Skilled adaptations to organismic constraints in RaceRunners were observed in the active shock attenuation after the point of foot–ground contact time and the ability to attenuate impact shock throughout the stance phase. This observation implies that using a RaceRunner bike can provide a safe and effective locomotor activity for people with disabilities, leading to health and wellbeing benefits, or helping them reach new performance limits according to their organismic constraints. Coaches could encourage young people with disabilities to participate in this sport as a medium to enhance the ability to transport the body and increase their physical function and capacities through running different distances.

The study has some limitations that must be acknowledged. First, RaceRunning is a new sport, and few athletes currently compete in this sport. As a result, this study is necessarily based on a small sample size, which makes it challenging to apply normative profiling for the analysis of shock absorption patterns, shown in this study, to all RaceRunning competitors or those who are new to RaceRunning. Second, the results presented here are descriptive and do not provide a statistical comparison with a group of control athletes. The data of this study have revealed that it would be a relevant next step for future studies to compare different footstrike patterns of RaceRunning athletes as more athletes join the sport. This approach will provide more information on the running gait adaptations used by RaceRunning athletes.

Conclusions

The findings of this study showed that RaceRunning is a safe physical activity in terms of impact shock that could be undertaken in people who lack an ability to walk unaided in their daily lives. The ergonomic design of the RaceRunning bike provides effective affordances (opportunities for action) for individuals, facilitating adaptive strategies in people with neurological motor disorders to enable them to walk and run as they interact with their environment, without losing balance and without extra physical loading that could affect the function of the vestibular and visual systems.

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