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ORIGINAL PAPER

Measuring biomechanical loads and asymmetries in junior elite long-distance runners through triaxial inertial sensors

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Summary

Background: Despite its various positive health effects, long-distance running implies substantial biomechanical loads on the musculoskeletal system, especially in a competitive context. The incidence of running-related overuse injuries is high, especially for the tibia being one of the predominant injury localisations. Tibial impact accelerations are an established valid proxy measure for impact forces and, as such, considered a more direct estimate of tibial bone load than ground reaction forces. Assessing individual tibial and upper-body impact accelerations may help balancing biomechanical loads during training for long-distance running and triathlon.

Material and methods: We employed inexpensive, wireless, lightweight, triaxial inertial measurement units (IMUs) as wearables for measuring tibial, sacral and scapular impact loads and asymmetries in 45 healthy junior-elite long-distance runners at submaximal running speeds. Moreover, we investigated the effects of ground surface vs. running speed on tibial load in 8 well-trained, healthy runners and triathletes.

Results: Mean peak tibial accelerations in junior elite long-distance runners ranged between 14 ± 3 and 16 ± 3 g ($g \approx 9.81$ m s^{-1}) for running speeds of 14–16 km h^{-1}. Corresponding mean peak sacral and scapular accelerations amounted to 4 ± 1 g to 5 ± 1 g (32 $\pm$ 8% of tibial load) and 4 ± 1 g (27 $\pm$ 6%), respectively. Observed lateral asymmetries in tibial and scapular accelerations yielded mean absolute values of 9 ± 8% (95th percentile of 24%) and 9 $\pm$ 10% (32%). The effect of running speed on tibial accelerations was approximately twice as large as the effect of ground surface. Among asphalt, tartan, treadmill and turf, only turf induced a significantly lower tibial load in well-trained runners.

Conclusions: IMU sensors represent a promising tool for assessing biomechanical loads and asymmetries in running under field conditions. With the orientation values derived in this study, they may be used to reduce individual

Introduction

Belonging to the core disciplines of the modern Olympic Games, long-distance running represents one of the most popular recreational sports, probably because of easy access, its fitness and health benefits as well as its economical nature [21,29,34]. From a medical point of view, long-distance running favourably contributes to individual and national healthcare by offering a moderate to high internal stimulus and a whole-body workout, whereas the common belief of a rather low injury risk is not confirmed by statistics [11,21,29]. The success of running events, like the Berlin Marathon, reflects a strong interrelation between elite and mass sports. For instance, two new records were set at the Berlin Marathon in 2018: Since its first edition in 1972, when only 244 athletes finished, the number of successful runners has now reached a new all-time best of 40,781. In parallel, the Kenyan Eliud Kipchoge won the race with a new world record of 2:01:39 h, the seventh improvement over the marathon distance within the past 15 years.

Despite its positive internal effects (e.g., on the cardiovascular, pulmonary and metabolic system), long-distance running implies substantial biomechanical loads on the musculoskeletal system, as also suggested by injury epidemiologic data [11,21,29,33]. Especially during high-volume phases, the lower limbs are prone to orthopaedic overuse injuries; particularly, in a competitive context, where top-level runners complete up to 150–200 km per week [31,33]. Beyond a "critical" (individual) mileage of roughly 60 km per week, the risk of sustaining an overuse injury increases [33]. In fact, 27–70% of all runners report to incur at least one running-related injury per year, the nominal incidence amounting to

risks of overuse injuries and, in future, possibly to assess running efficiency.
Level of evidence: Ib.

Keywords
Acceleration – Inertial measurement unit (IMU) – Sacrum – Scapula – Tibia

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Erfassung von biomechanischen Belastungen und Asymmetrien bei Nachwuchsleistungssport-Langstreckenläufern mithilfe mehrachsiger Inertialsensoren

Zusammenfassung

Hintergrund: Die Sportart Laufen ist einerseits gesundheitsförderlich, bedingt andererseits jedoch eine erhebliche biomechanische Beanspruchung des muskuloskelettalen Systems, insbesondere im Leistungssportkontext des Langstreckenlaufs. Statistisch ist u. a. die Tibia häufig von Über- und Fehlbelastungen betroffen. Die Erfassung tibialer Beschleunigungen gilt als etabliertes indirektes Messverfahren zur Ermittlung von Stauchungskräften und ist zur Abschätzung der ossären Beanspruchung dem weniger genauen Ansatz über Bodenreaktionskräfte vorzuziehen. Unter Kenntnis der individuell auftretenden Stauchungsbelastungen an Schienbeinen und Oberkörper kann die biomechanische Trainingsbelastung in den Sportarten Langstreckenlauf und Triathlon besser berücksichtigt und auf den jeweiligen Sportler abgestimmt werden.
Material und Methoden: Es wurden kostengünstige, funkbasierte dreidimensionale Inertialsensoren als leichte Wearables eingesetzt, um die tibialen, sakralen und skapularen Stauchungsbelastungen sowie diesbezügliche Asymmetrien in 45 gesunden Nachwuchsleistungssport-Langstreckenläufern bei submaximalen Laufgeschwindigkeiten zu messen. Darüber hinaus wurden die Einflüsse von Bodenbelag und Laufgeschwindigkeit auf die

approximately 2.5–12.1 injuries per 1000 h of training [5,33]. Among running-related overuse injuries, the knee joint, Achilles tendon and tibia belong to the predominant injury sites [5,29,33]. Their etiopathology is often embedded in a multifactorial context, especially for stress fractures [5,11,27,29,32].
In particular, the relationship between tibial stress fractures and lower limb accelerations is affected by various individual factors, including anatomical and biomechanical conditions, bone mineral density, applied muscular shock attenuation strategy, etc [2,9,4,16,18,21–23,26]. Nonetheless, there is evidence that an accumulated tibial acceleration load is positively linked to an increased incidence of tibial stress fractures [2,4,16,18,26]. Tibial accelerations are widely accepted by clinicians and researchers as a proxy measure for impact forces experienced at the tibia [26]. Although tibial impact is a surrogate measure of bone loading, it is considered a more direct estimate of tibial load than ground reaction forces [16]. Depending on the context, either the magnitude of the resultant triaxial vector (*acceleration magnitude*) $|\vec{\mathbf{a}}| = \sqrt{a_x{}^2 + a_{\mathrm{ML}}{}^2 + a_{\mathrm{AP}}{}^2}$ or its three uniaxial components along the anatomical axes, i.e. the axial (a_x), medio-lateral (a_{ML}) and anterior-posterior (a_{AP}) direction, are considered [26]. In the context of running, tibial accelerations exhibit a high inter-study variance. For example, a pioneer study from 1991 reported peak tibial acceleration magnitudes of 11 g (with $g \approx 9.81\ \mathrm{m\ s^{-2}}$) for running at 19 km $\mathrm{h^{-1}}$ on a motorised treadmill as measured by an invasive Steinmann-pin-fixed accelerometer [14]. In contrast, a recent non-invasive study yielded higher axial tibial accelerations of $15 \pm 7\ g$ and $25 \pm 11\ g$ for running on a motorised treadmill and on a tartan track, respectively, at a lower speed of 14 km $\mathrm{h^{-1}}$ [7]. Interestingly, that discrepancy proves substantial even though triaxial magnitudes from [14] are compared to uniaxial values from [7] (the latter, by rules of mathematics, being smaller than or at maximum equal to their corresponding 3D magnitudes). The deviation is presumably explained by differences in research design, e.g. positioning of the accelerometers, participants, footwear and ground surfaces) [14,15]. In relative terms, the rate of transfer of tibial acceleration shock to the pelvis and shoulders, named *shock attenuation*, provides valuable insights into limb stiffness, core stability, running efficiency and status of fatigue. Obviously, shock attenuation is thus also discussed for its role in running-related injuries [16,19]. Furthermore, first investigations report that shoulder accelerations are negatively related to performance and, as such, increase during long-distance running, presumably as a sign of fatigue [28]. Scapula rotation may equally be an indicator of running efficiency, since beginners were shown to exhibit a higher degree of shoulder rotation than advanced runners [28].
To date, the state of research is still limited regarding the question whether commercially available inexpensive inertial measurement units (IMUs), consisting of (triaxial) accelerometers, gyroscopes and magnetometer, are practically suitable for assessing biomechanical loads and asymmetries during long-distance running. For sport science and medicine, such knowledge is of high practical relevance when aiming at balancing internal and external training loads under outdoor conditions on a daily basis, where laboratory technologies, like force plates or dynamometers, are neither applicable nor useful. Ultimately, employing IMUs not

Tibiabelastung bei acht gesunden, trainierten Läufern und Triathleten verglichen.
Ergebnisse: Bei den Nachwuchsleistungssport-Langstreckenläufern wurden mittlere Spitzenbeschleunigungen an den Schienbeinen von 14 ± 3 bis 16 ± 3 *g* ($g \approx 9{,}81$ m s^{-2}) für Laufgeschwindigkeiten von 14 bis 16 km h^{-1} gemessen. Die zugehörigen mittleren Spitzenbeschleunigungen am Kreuzbein und den Schultern betrugen 4 ± 1 *g* bis 5 ± 1 *g* (d. h. 32 ± 8% der Schienbeinbelastung) bzw. 4 ± 1 *g* (27 ± 6%). Die beobachteten lateralen Asymmetrien der tibialen und skapularen Beschleunigungen beliefen sich im Absolutwert-Mittel auf 9 ± 8% (95. Perzentile: 24%) bzw. 9 ± 10% (32%). Weiterhin fiel der Einfluss der Laufgeschwindigkeit auf die tibiale Beschleunigung mehr als doppelt so stark als jener der Bodenbeschaffenheit aus. Im direkten Vergleich von Asphalt, Tartan, Laufband und Rasen führte lediglich der Rasenuntergrund zu einer signifikanten Verringerung der Schienbeinbelastung bei trainierten Läufern.
Schlussfolgerungen: Inertialsensoren stellen einen vielversprechenden Ansatz für die Erfassung biomechanischer Belastungen und Asymmetrien unter Feldbedingungen dar. Mithilfe der Orientierungswerte, wie sie in der vorliegenden Arbeit abgeleitet werden, könnten Inertialsensoren einerseits dazu beitragen, das individuelle Risiko für Über- und Fehlbelastungen zu verringern und andererseits zukünftig zusätzlich zur Bewertung der Laufeffizienz eingesetzt werden.
Evidenzklasse: Ib.

Schlüsselwörter
Beschleunigung – Inertialsensor – Kreuzbein – Schienbein – Schulterblatt

only for performance increase, but also for rehabilitative and preventive purposes may evolve the field of load monitoring in long-distance running to a next level. This study is intended to contribute to the evaluation of (triaxial) tibial, sacral and scapular accelerations, shock attenuation and asymmetries in junior elite long-distance runners based on inexpensive, commercially available IMU devices. Special attention will be paid to the effects of running speed, ground surfaces and terrain inclination, seeking to provide helpful and readily transferable information for scientists and practitioners. Our study is divided into two parts: First, we present data on shock attenuation, tibial and scapular accelerations and asymmetries in junior elite long-distance runners. Second, we investigate the effects of ground surface and running speed in well-trained runners. For the sake of clarity, we first present the result of both parts separately, and provide a combined discussion along with practical conclusions at the end of this article.

Methods

Inertial measurement units (IMUs)

Nowadays, several manufacturers offer integrated, wireless multi-axis IMUs for non-invasive three-dimensional measurements of accelerations, angular speeds and (partially) magnetic field strengths for sports and orthopaedic purposes. In our study, we use the validated IMUs of a leading supplier (Xsens Technologies B.V., MTw Awinda, Enschede, Netherlands) for allowing comparisons to previous studies with those devices [10,12,13,20,35]. The employed IMUs are small (4.7 cm × 3.0 cm × 1.3 cm), lightweight (16 g) and operate wirelessly for approximately 4 h. The output data sampling rate was set to 120 Hz, while the internal sampling rate of each acceleration axis is 1000 Hz (i.e. before sensor fusion calculations). Each of the three acceleration axes has a measurement range of ±160 m s^{-2} and a noise uncertainty of ±0.02 m s^{-2}. Based on the measured 3D components a_x, a_y and a_z of the acceleration vector $\vec{\mathbf{a}} = \left(a_x, a_y, a_z\right)^{\mathrm{T}}$, its magnitude is calculated by $a \equiv |\vec{\mathbf{a}}| = \sqrt{{a_x}^2 + {a_y}^2 + {a_z}^3}$. Notably, because the direction of each 3D vector changes during running with respect to the global frame of reference, the axial limitation of a_x, a_y or a_z to ±16 *g* actually does not apply to the magnitude measurement range. Ideally, the latter is increased to $\sqrt{3}\cdot 16\ g \approx 28\ g$, and practically to approximately 18–22 *g* ($\approx \sqrt{2}\cdot 16\ g$), according to our experience. Generally, current microelectronic accelerometers (and gyroscopes) are subject to inertial sensor drift, i.e. a non-predictable, systematic, long-term measurement error for small, quasi-constant accelerations $a \ll g$ (or small angular velocities, respectively). For the accelerometers used in this study, the reported bias stability is high, amounting to a drift bias of only 0.0001 *g* [24]. Moreover, accelerometer measurements within a running cycle are characterised by multiple intra-cyclic, short-term sign inversions (e.g., a change from axial +7 *g* to −3 *g* within 20 ms, see Fig. 2 (b)) and by acceleration magnitudes $a \gtrsim g$ greater than gravity. Both aspects minimise the distorting effects of inertial sensor drift in the context of running applications, especially for peak value measurements. In particular, sensor drift proves negligible for this study. Regarding mechanical alignment deviations, the axes of the accelerometer used are statically fixed to the

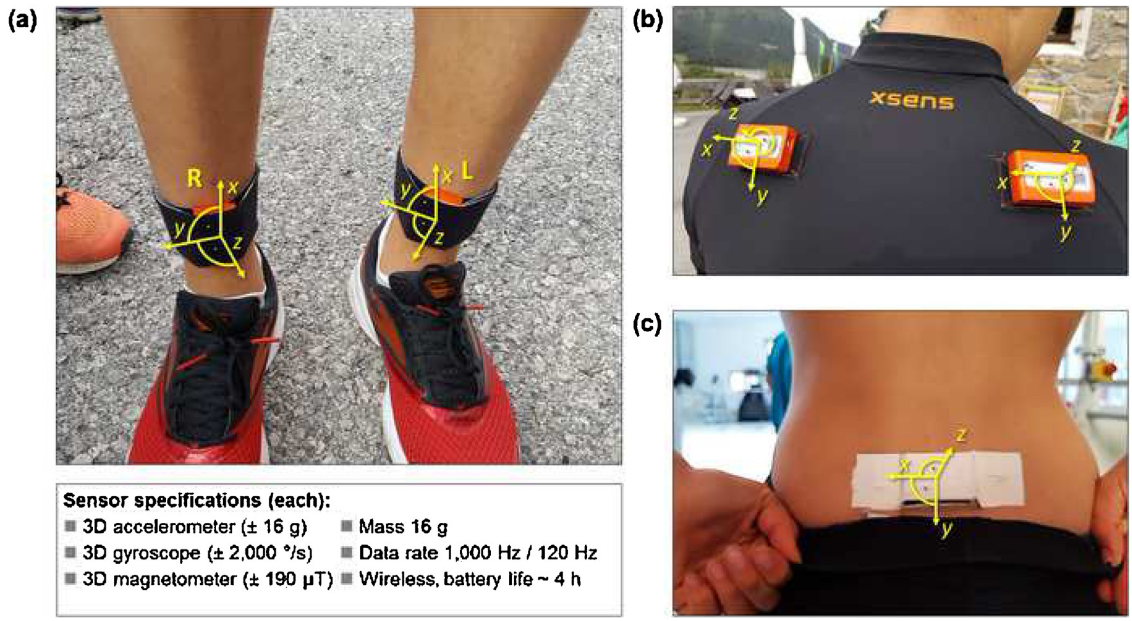

Figure 1
Sensor placement at the tibiae, scapulae and sacrum. (a) Tibial sensors. (b) Scapular sensors. (c) Sacral sensor. Yellow sketches depict the alignment of the coordinate systems for 3D acceleration.

IMU housing with a maximum alignment error of 0.1° [24]. Therefore, in summary, no alignment reset or recalibration is required in this study even for long-term measurements. Because only accelerometer data are used, also gyroscope drift is irrelevant.

Sensor placement

In each study, two IMUs were fixed noninvasively at the distal anteromedial section of the right and left tibia of each athlete using tightly fitting Velcro™ straps. The lower edge of each IMU was placed 8 cm above the transverse plane defined by the centres of medial and lateral malleolus (Fig. 1(a)). At that particular position, the relative motion between the tibiae and sensor is minimal during running due to the absence of voluminous muscle or connecting tissue in between. Moreover, if required for the respective study, additional IMUs were placed on each scapular and the sacrum either by means of dedicated functional shirts (Fig. 1(b)) or by a combination of standard medical Leukotape™ and adhesive Velcro™ tape. Due to the small stack height of the IMU sensor chip within its housing, any acceleration amplification bias due to an additional mechanical lever is negligible for the considered tibial, sacral and scapular application sites. The additional IMU lever, given by the distance from the actual sensor microchip to the skin, is approximately 6 mm and thus at least one order of magnitude smaller than the distance of the IMU from the relevant anatomical fulcra of $\gtrsim$150 mm. The resulting maximum measurement bias is thus below 5%. Nonetheless, a certain damping or distortion effect due to soft tissue and/or skin movements cannot be fully excluded, yet is expected to be sufficiently small for the anatomical sites and slim somatotypes relevant to this study. Finally, to exclude any potential sensor uncertainty-based bias with respect to lateral asymmetries of the limbs, sensors were swapped in randomised order between the anatomical sites after a test series. Before each test, all sensors were synchronised and checked for apparent functionality.

Data analysis

For all three parts of the study, raw acceleration data were first processed by a proprietary software (MT Software Suite MTw™ AwindaTM 4.8, Xsens Technologies B.V., Enschede, Netherlands), converting binary to ASCII data. Subsequent analyses were accomplished by a self-programmed

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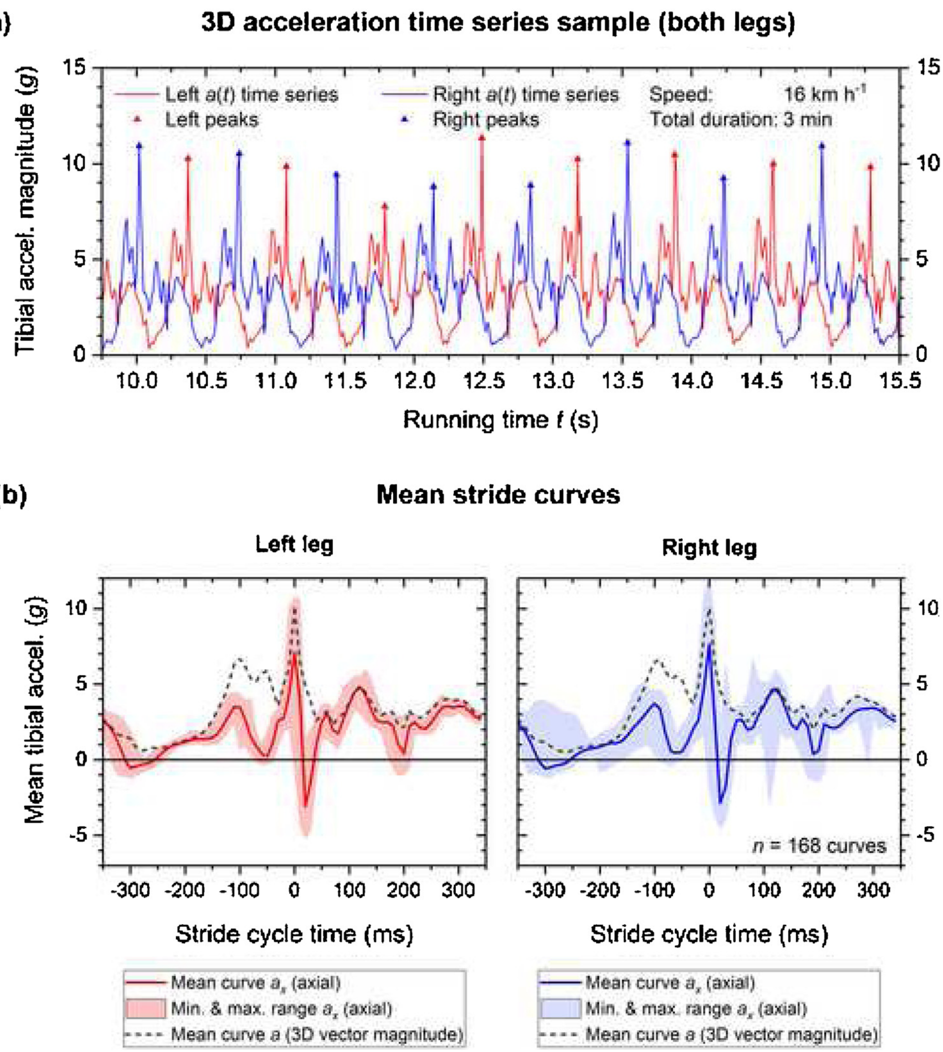

Figure 2
Sample of tibial accelerations and mean peak curves calculated therefrom. (a) Sample time series of tibial acceleration vector magnitudes $a \equiv |\vec{\mathbf{a}}|$ at a running speed of 16 km h^{-1} ($\hat{=}$3:45 min km^{-1}) for the left (red) and right leg (blue), showing eight impact events for each leg corresponding to eight complete strides. Peak tibial impact accelerations as detected by the data analysis software are marked with small triangles. (b) Mean stride curves as calculated from 168 single impact curves (i.e. 120 s) partly presented in (a).

routines (LabVIEW 2016, National Instruments, Austin, USA). First, the peak impact events of each step and complete stride were detected in the acquired triaxial magnitude acceleration time series (Fig. 2(a)). Then, descriptive statistics for selected time frames were computed, yielding mean peak accelerations and standard deviations (SD) for the left and right tibiae and scapulae and, where applicable, for the sacrum (Fig. 2(b)). In addition to those data for the triaxial vector magnitude $a \equiv |\vec{\mathbf{a}}|$, the uniaxial acceleration component a_x along the tibia ("axial component" in the following) was analogously processed for each leg. Intra- and inter-individual statistics were conducted with Microsoft Excel 2016 (Microsoft Corporation, Redmond, USA) and IBM SPSS Statistics 23 (IBM, Armonk, USA). For the error bars, SDs were used unless stated otherwise.

For examining asymmetries between left and right limbs, the asymmetry

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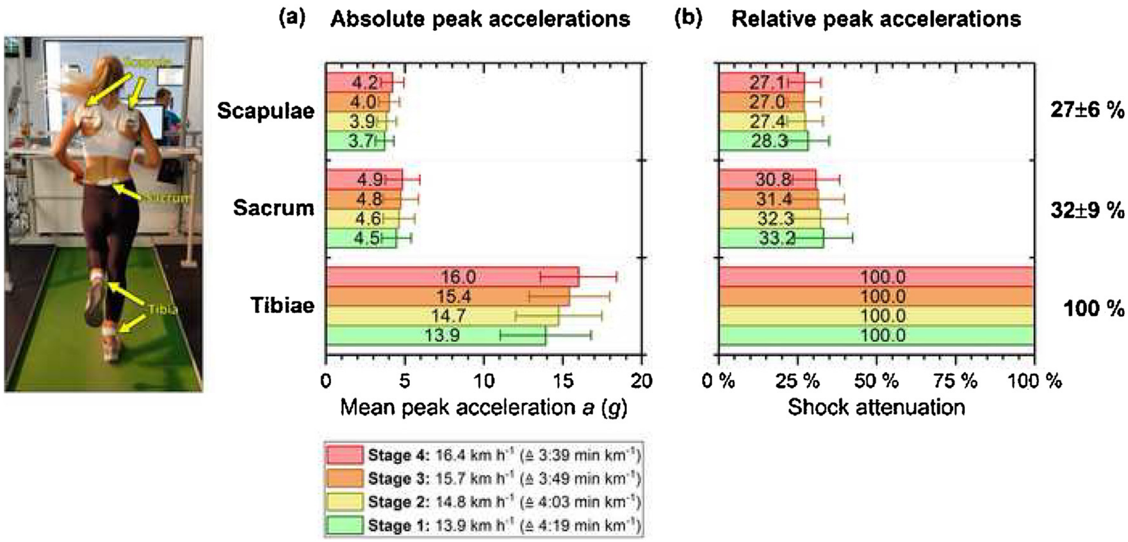

Figure 3
Shock attenuation from tibiae via sacrum to scapulae at different running speeds in junior elite long-distance runners. (a) Mean peak accelerations ± SD for the tibiae (mean of left and right leg), sacrum and scapulae (mean of left and right shoulder) as a function of running speed (b) Shock attenuation ratio as mean ± SD.

indexes Δ_a for triaxial acceleration magnitude (tibiae and scapulae) and Δ_{a_x} for axial acceleration (tibiae only) were defined as follows:

$$\Delta_a \equiv \frac{a^{(\mathrm{L})} - a^{(\mathrm{R})}}{\frac{1}{2}\left(a^{(\mathrm{L})} + a^{(\mathrm{R})}\right)}; \qquad (1)$$

$$\Delta_{a_x} \equiv \frac{a_x^{(\mathrm{L})} - a_x^{(\mathrm{R})}}{\frac{1}{2}\left(a_x^{(\mathrm{L})} + a_x^{(\mathrm{R})}\right)}$$

where $a^{(\mathrm{L})}$, $a_x^{(\mathrm{L})}$ and $a^{(\mathrm{R})}$, $a_x^{(\mathrm{R})}$ denote the corresponding mean peak magnitude and axial accelerations of the left (L) and the right (R) tibia or scapula, respectively. In theory, an entirely balanced acceleration load would result in $\Delta_a = 0$ and $\Delta_{a_x} = 0$. For higher accelerations at the left leg, however, Δ_a and/or Δ_{a_x} would become positive, for higher accelerations at the right leg negative instead.

Part 1: Shock attenuation and asymmetries in tibial and scapular accelerations at different running speeds in junior elite long-distance runners

The aim of this part was to obtain quantitative figures for shock attenuation from the tibiae via the sacrum to the scapulae and to detect asymmetries in tibial and scapular accelerations at different running speeds in junior elite long-distance runners.

Subjects

Forty-five junior elite long-distance runners of the German National Team participated (27 male: 18.5 ± 1.8 years, 182 ± 5 cm, 67.8 ± 6.9 kg, BMI 20.3 ± 1.5 kg m^{-2}; 18 female: 18.2 ± 2.7 years, 170 ± 5 cm, 52.8 ± 6.1 kg, BMI 18.3 ± 1.6 kg m^{-2}). Their mean maximum oxygen uptake was 67.0±4.8 ml min^{-1}kg^{-1} and 57.7±2.7 ml min^{-1}kg^{-1}, respectively, while the weekly training mileage amounted to 50.2 ± 19.5 km at the period of testing.

Study design

The athletes completed their regular performance diagnostic programme at the Institute of Applied Training Science in Leipzig (Germany), including a submaximal treadmill test. That test consisted of four stages of 2000–3000 m with zero inclination, while the exact distance and speed were individually set according to the competition distance and performance level of the athletes. Between each stage, a 1 min rest period was allowed. After each break, the speed was increased by 0.9 km h^{-1}. The initial speed was chosen by a professional coach based on the current

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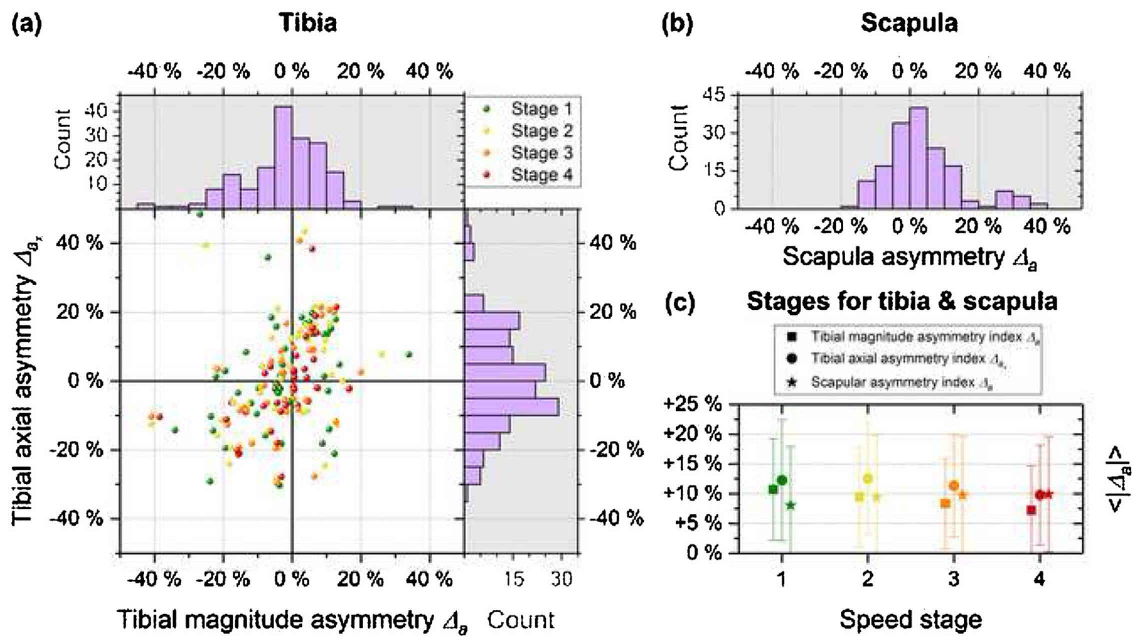

Figure 4

Asymmetries of tibial and scapular accelerations in junior elite long-distance runners. (a) Scatter plot of asymmetry indexes of Δ_a (mean peak tibial 3D acceleration, abscissa) and Δ_{a_x} (mean peak axial tibial acceleration, ordinate) for all subjects and speed stages (colours) and related histograms (top and left). (b) Histogram of scapular asymmetry index Δ_a for all subjects and stages. Each stage for each subject is counted separately, as in panel (a). (c) Mean absolute values $\langle|\Delta_a|\rangle$ and $\langle|\Delta_{a_x}|\rangle$ for tibial and $\langle|\Delta_a|\rangle$ for scapular acceleration asymmetry as function of speed stage ($\pm$ SD).

performance level of the athlete, resulting in a mean initial speed of 13.9 ± 1.4 km h^{-1}. Of the 45 athletes, 45 completed stage 1 and 2, while stages 3 and 4 were finished by 44 and 37 athletes, respectively. The average time for completing all four stages, including breaks, was 34 $\pm$ 4 min. All athletes wore their own footwear. During the runs, the accelerations at the tibiae, sacrum and scapulae were measured as described above and illustrated in Fig. 3(a). The employed microelectronic accelerometers did not require calibration or re-calibration, as explained above.

Results

The results are presented in Fig. 3. Panel (a) shows the peak tibial, sacral and scapular accelerations in terms of mean $\pm$ SD at different running speeds, while panel (b) depicts the relative shock attenuation cascade as derived therefrom. For the two tibial and scapular sensors, the arithmetic mean of both sensors (i.e., left and right leg; left and right shoulder) is used. The accelerations significantly increase with speed ($p < 0.004$) and decrease with a more cranial position ($p < 0.001$, two-way with ANOVA with repeated measures). The speed-related relative increase from stages 1 to 4 amounts to 15% at the tibiae, 9% at the sacrum and 13% at the scapulae. Altogether, 32 $\pm$ 9% of the peak impact accelerations at the tibiae are still present at the sacrum. At the scapulae, that fraction of remaining peak acceleration is further reduced to 27 $\pm$ 6%. Notably, acceleration magnitudes measured at the sacrum and the scapulae generally reflect a superposition of the transferred tibial impacts and any additional co- and/or counter-movement of the pelvis and the shoulders. That is particularly the case for the typical contralateral balancing coordination of legs and arms in running. However, as the acceleration peaks at the sacrum and the shoulders are considered in direct temporal relation to their preceding tibial impact peaks in this study, that measurement bias is reasonably minimised. Well-controlled, balancing co- and counter-movements of the upper body typically do not coincide with tibial impact peaks.
Concerning lateral symmetry, the tibial asymmetry indexes for 3D acceleration Δ_a and axial acceleration Δ_{a_x} do not show any

pronounced overall lateral preference (Fig. 4(a)). The total arithmetic mean for all subjects and all stages amounts to[1] $\langle\Delta_a\rangle = -2.5 \pm 11.8\%$ (i.e., a minor dominance of right leg) and to $\langle\Delta_{a_x}\rangle = +0.4 \pm 14.8\%$ (i.e., virtually no lateral dominance), respectively. For the scapulae, the total mean asymmetry index is $\langle\Delta_a\rangle = +4.7 \pm 12.8\%$, meaning a slight "dominance" of the left shoulder (Fig. 4(b)), and confirms a general contralateral correlation of legs and arms. Notably, despite all three mean values of signed asymmetry being close to zero, SDs are between 11% and 15%, suggesting a substantial range of individual lateral asymmetry.

As regards absolute values, mean absolute asymmetry indexes amount to $\langle|\Delta_a|\rangle = 9.0 \pm 8.1\%$ (95th percentile: 24.3%), $\langle|\Delta_{a_x}|\rangle = 11.6 \pm 9.2\%$ (29.3%) and $\langle|\Delta_a|\rangle = 9.2 \pm 9.9\%$ (31.7%) for 3D tibial, axial tibial and 3D scapular asymmetries, respectively, and are practically independent of running speed (Fig. 4(c)). Comparing the more highly loaded sides of tibial and scapular asymmetries in the individuals, the pair-wise total Spearman correlation coefficient amounts to ≈0.14, suggesting an only small positive correlation between the more highly sides in peak tibial and peak scapular acceleration.

Part 2: Effect of ground surfaces on tibial accelerations at different running speeds in well-trained athletes

This part focuses on the effect of different ground surfaces on tibial accelerations for various running speeds in well-trained runners. Commonly, changing the ground surface from rigid to more elastic conditions (e.g. from asphalt to tartan) is believed to be an effective way of reducing the musculoskeletal loading. The outcome of this study is to challenge this common belief among practitioners.

Subjects

Eight well-trained male amateur runners and triathletes (29 ± 6 years, 179 ± 6 cm, 73 ± 6 kg, BMI = 22.5 ± 1.3 kg m^{-2}) took part, possessing a personal best over 5 km road or track running of less than 19 min (mean of 16:47 $\pm$ 0:40 min).

Study design

Each athlete completed four series of four runs of 3 min at 12, 14, 16 and 18 km h^{-1} (i.e., 5:00, 4:17, 3:45 and 3:20 min km^{-1}). Each series was conducted in randomised order on a different ground surface. Four surfaces were chosen based on their common usage in daily exercise routine:

1. Turf (well-kept turf field within a track and field stadium)
2. Tartan (400 m with eight competition lanes in a track and field stadium built 2009)
3. Asphalt
4. Motorised treadmill (custom-built, comparable to h/p cosmos saturn).

Each athlete wore the same individual pair of shoes during all runs. The four series were split in two times two over two days with the same moderate to low level of prior training load to minimise effects of fatigue. Weather conditions were the same during all test days, ensuring dry surface conditions and moderate running temperatures between 18 and 25 °C. For ensuring correct running speed, the subjects controlled their pace with global positioning system (GPS)-enabled sports watches (Garmin Forerunner 910XT or comparable) and performed the runs in couples side-by-side. The GPS-recorded speeds were verified prior to further data analysis, confirming intended paces. For acquiring 3D distal tibial accelerations, each athlete was equipped with two IMUs as described above (Fig. 1). For simplicity, the mean value of the peak accelerations of the left and right leg was used for subsequent data analysis. Accounting for any settling effects in speed or locomotion pattern, the first and last 30 s of each 3 min bout were disregarded.

Results

The results are summarised in Fig. 5. Apparently, both ground surface ($\eta^2 \approx 0.165$) and running speed ($\eta^2 \approx 0.428$) exhibit a significant effect on tibial accelerations ($p < 0.001$, two-way ANOVA with repeated measures). As for running speed, Bonferroni-corrected post-hoc tests confirm a significant intensification of tibial accelerations for each speed step from 12 to 16 km h^{-1} ($p < 0.05$). The difference between 16 and 18 km h^{-1} was not significant. Regarding surface, tibial accelerations increased from turf (overall mean for all subjects and speeds of 12.6 ± 2.5 g) via tartan (14.1 ± 2.5 g) to asphalt (14.7 ± 2.2 g) and treadmill (14.8 ± 2.3 g). However, only turf proved to be produce significantly lower accelerations vs. treadmill ($p < 0.001$), asphalt ($p < 0.002$), and tartan ($p < 0.05$) by pairwise Bonferroni-corrected post-hoc tests. Notably, all other differences were not significant, i.e. between tartan, asphalt and treadmill. In summary, as indicated by the effect sizes in terms of η^2, running speed

[1] Angle brackets $\langle\ldots\rangle$ denote the arithmetic mean throughout this article.

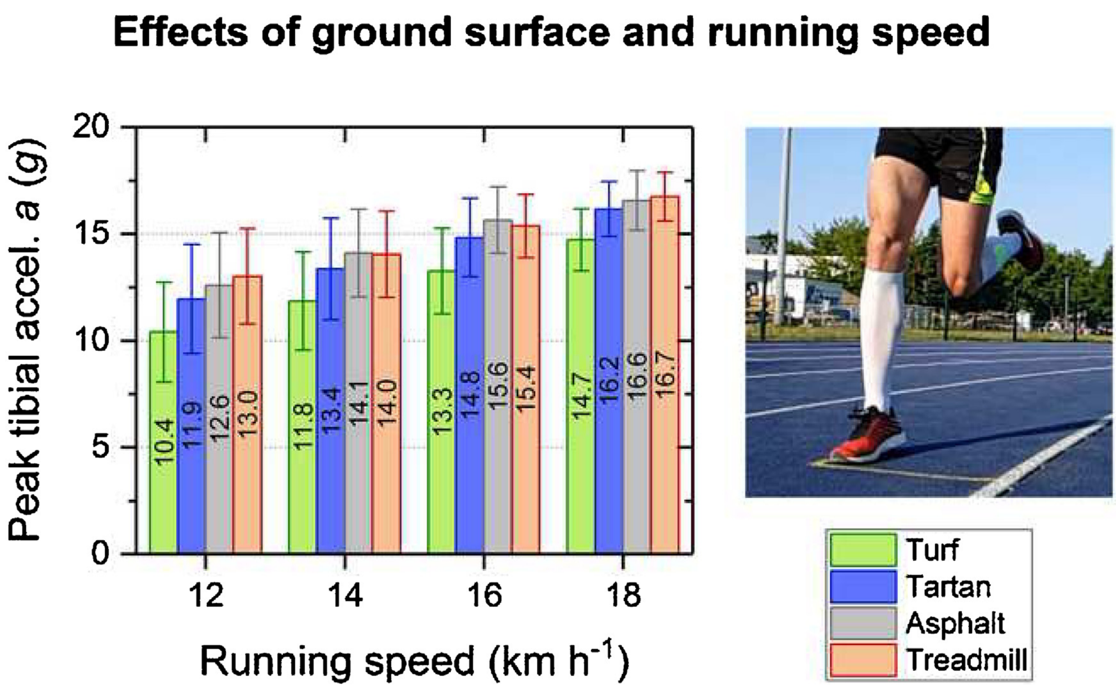

Figure 5
Effect of different ground surfaces on tibial accelerations at different running speeds in well-trained runners. The left plot shows mean peak tibial acceleration ± SD as a function of surface (colours) and running speed (abscissa).

induces a more than twice larger effect on tibial accelerations than ground surface (η^2 ratio of ≈2.6).

Discussion

This study has yielded three major results on assessing biomechanical loads in tibiae, sacrum and scapulae and their asymmetries therein by wearable IMUs: (1) shock attenuation from tibiae via sacrum to scapulae, (2) lateral asymmetries in peak tibial and peak scapular accelerations and (3) the effect of ground surfaces vs. running speed. These outcomes will be discussed in the following.

Shock attenuation

For junior elite long-distance runners, it was found that tibial impact acceleration shock is reduced on average to 32 ± 9% at the sacrum and to 27 ± 6% at the scapulae. That shock attenuation cascade is qualitatively known from biomechanical and orthopaedic literature and reflects the site-dependent active and passive damping effect of the lower limbs and upper body including multiple joints and tissues [18,26]. According to our data, about two thirds (68 ± 9%) of the tibial shock are absorbed by the knee and hip joints and the articulated musculoskeletal segments, while further shock attenuation within the trunk from the sacrum to the scapulae amounts to approximately one seventh (14 ± 7%). That substantially greater degree of shock attenuation observed in the lower limbs is biomechanically expected because of the greater number of degrees of freedom in the legs, including knee and hip joints, their higher mobility and their wider range of motion as compared to the trunk. In the trunk, it is basically the less mobile vertebral joints and the adjacent small muscles basically to provide shock absorption, whereby the overall effectiveness is lower than in the legs. Moreover, an active, balancing co- or counter-movement of the shoulders could principally also be taking place at the moment of tibial impact, which might superimpose residual accelerations from the lower limbs and thus reduce apparent shock attenuation from sacrum to scapulae. In addition, fatigue-induced hip instability, manifesting itself in a lateral pelvic drop especially at prolonged exercise duration, could influence observed shock attenuation figures because of the large levers and effective masses of the upper body occurring in single-leg stance. In fact, weak/fatigued hip abductors in the stance leg could result in an amplification of the residual tibial acceleration at the contra-lateral shoulder due to a lateral pelvic drop to that side (thereby resembling an (initial) Trendelenburg's sign). However, there is no simple, IMU-based methodology compatible to our measurement setup to distinguish between those different potential contributions to resultant shoulder accelerations. Hence, the shock attenuation figures derived in this study systematically comprise both effects, i.e. ipsilateral and contralateral shock attenuation as well as contra-lateral shock amplification resulting from insufficient hip stability.

In absolute terms, our data are in agreement with values previously reported for the upper body (e.g., 2.5–5.5 g measured at the shoulders [28]). To the best of our knowledge, our findings firstly report quantitative, anatomical-site specific data for junior elite long-distance runners based on inexpensive, commercially available IMU sensors.

Lateral asymmetries in peak accelerations

For junior elite long-distance runners, the total mean of the 3D tibial, axial tibial and scapular asymmetry indexes for all subjects and running speeds was found to lie within the range of $\pm5\%$, showing that there is virtually no general lateral dominance in healthy subjects from that group. The degree of asymmetry does not alter with running speed, as other current studies confirm [8].

In view of the mean absolute asymmetry indexes and their 95th percentiles, it can be concluded that tibial asymmetries of less than approximately 13% in 3D magnitude and of less than 22% in axial acceleration may be considered "normal" for junior-elite long-distance runners. Equally, scapular peak acceleration asymmetries of less than about 32% may still be "normal" in statistical terms. Because significant asymmetries may point at unfavourable individual orthopaedic conditions, however, we generally recommend a prophylactic assessment of the athlete by an experienced (sports) orthopaedist and/or physiotherapist when asymmetries exceed 20–30%.

The observed only small positive correlation between the more highly loaded sides in peak tibial and peak scapular acceleration suggests that active shock attenuation in running can diminish or even reverse lateral asymmetries from the lower to the upper limbs. In addition to the conceivable contribution of pelvic drop already mentioned, further physiological and unphysiological individual conditions may play a role, including practical compensatory strategies. For instance, a leg length difference, scoliosis or a sacroiliac joint dysfunction could be (subconsciously) compensated for by the runner, or possibly a (unilateral) cartilage damage in a knee or hip joint may be relieved thereby, to name just a few possible causes for a lateral shift in tibial vs. scapular asymmetries.

Regarding 3D and axial peak tibial accelerations, the degree of correlation between 3D and axial asymmetry indexes increases with increasing running speed, i.e. with increasing physiological demand: The Spearman correlation coefficient of Δ_a and Δ_{a_x} grows with running speed from $\approx$0.28 via 0.37 and 0.51–0.57 for stages 1–4, respectively, as already suggested by the "increasingly dense" scatter plots in Fig. 4(a). From a biomechanical perspective, a difference in axial tibial and 3D tibial asymmetry index implies that transversal accelerations (i.e. the anteroposterior and/or the mediolateral acceleration components) differ from left to right leg. Such a difference must be counter-balanced by the rest of the body for every step, thereby decreasing running efficiency. Since those differences are found to diminish with running speed for our subjects, their motor control apparently becomes more efficient with increasing physiological demand.

When discussing asymmetries, it is widely acknowledged that human movement generally does not fulfil the assumption of perfect symmetry. In fact, many athletes possess a dominant leg for specific tasks, manifesting itself, e.g., in a preferred leg for kicking a ball or in a favoured take-off leg for jumping tasks. Notably, those preferences may differ also intra-individually from task to task [1,17]. Although the degree of lateral dominance is generally less pronounced for the lower then for the upper limbs, about two thirds of people, e.g., choose their left leg for take-off in jumping tasks, whereas for kicking tasks, in contrast, the majority prefers to use their right leg [17]. While in such acyclic tasks, the habitual preference of one leg is often apparent to the observer, for highly repetitive cyclic motion tasks like running both the detection and the consequences of a laterality are often less obvious and thus require objectification. The tibial IMU approach of this study offers a practical means of such assessment under virtually any running exercise conditions. As for orthopaedic implications, a pronounced asymmetry is generally considered a risk factor for running-related overuse injuries [25,36]. For instance, Zifchock and co-workers found higher peak tibial accelerations in the injured than in the non-injured leg of injured runners ($5.2 \pm 2.2\ g$ vs. $4.8 \pm 1.6\ g$), while

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their overall peak accelerations (i.e., mean of both legs) did not differ from those of healthy runners [37]. Those researchers concluded that elevated values of tibial acceleration on the injured side possibly indicated an insufficiency of the musculoskeletal system in attenuating mechanical load [37].

Notably, the upper limits of statistically "normal" asymmetries found in this study, e.g. $\Delta_{a_x} = +22\%$ for axial tibial and $\Delta_a = +32\%$ for scapular accelerations, correspond to ratios of the left vs. the right limb of $(1+0.23)/(1-0.23) \approx 1.60$ and 1.94, respectively (cf. Eq. (1)). In other words, one limb may be subject to loads by 60% or 94% higher than the other. It appears at least questionable if those ranges of "normal" asymmetries can still be considered generally "physiological", given the high training loads in junior elite running with extensive mileages and elevated exercise intensities. However, as discussed above, lateral asymmetries might also reflect a sensible and effective measure of the runner to compensate for individual anatomic and/or orthopaedic conditions, such as, e.g., scoliosis or a congenital/acquired articular malalignment. In such cases, the observed asymmetry is to be regarded functional und physiological and should not be "corrected". The fact that the highly trained subjects of this study were all self-reportedly healthy despite their partially substantial asymmetries, supports this perception. Nonetheless, there are no data at hand yet if substantial lateral asymmetries, irrespective of their cause, might result in harmful imbalances and eventually lead to higher incidences of overuse injuries in the long term. Therefore, an individual, thorough orthopaedic examination is generally recommended in the case of significant lateral asymmetries.

In essence, future research is needed, especially with respect to long-term longitudinal aspects, to eventually enable the practitioner to validly differentiate between physiological and unphysiological/critical musculoskeletal loading asymmetries in running.

Effect of ground surface vs. running speed

For well-trained runners and triathletes, the effect of running speed on tibial acceleration turned out to be more than twice the effect of ground surface. In absolute terms, our data show that reducing the tibial impact from, e.g., 17 g to 15 g can be achieved either by running on turf instead of asphalt or simplify by reducing speed from 18 km h^{-1} to 16 km h^{-1} (Fig. 5), the latter option probably being easier to realise in everyday training than the first. However, reducing running speed may imply a drop in training intensity, and thus might be counterproductively "compensated for" by the athlete/coach in terms of a prolonged exercise duration. Nonetheless, the observed subordinate role of surface vs. running speed is an important result that should be kept in mind by practitioners when planning training sessions, especially for compensatory, preventive and rehabilitative purposes.

In particular, turf proved to be the only surface with a statistically significant decrease of tibial accelerations for a given running speed. Tartan did not significantly reduce tibial impact as compared to asphalt. These findings are in accordance with previous studies that measured ground reaction forces and plantar pressure for various running surfaces, where the surface effects were rather small for asphalt, concrete and rubber (i.e. tartan) with the only exception for turf [3,6,30]. It remains to be elucidated in future studies if the difference in surface elasticity and shock absorption is the primary and only cause for altered tibial accelerations, especially with respect to turf, or if there is perhaps a secondary contribution of an implicit change in running pattern, i.e. a change from mid-foot to forefoot strike. Moreover, future research should investigate if there are surface-related differences between well-trained and highly trained runners. Finally, also the damping effects of different running footwear, i.e. provided by different sole elasticity or thickness, should be thoroughly examined in this context in future research.

Outlook

As suggested by this study, IMU sensors are a promising measurement option for movement science and orthopaedics under field conditions. As such, integrated, smart IMU running sensors might eventually contribute to dynamic load control in running and triathlon in a similar fashion as wearable heart rate monitors and GPS-enabled sports watches did in the 1980s and 2000s, respectively. In either of those two cases, affordable portable devices replaced expensive stationary laboratory equipment, allowing the athlete to freely train outdoors while having precise heart rate, distance, speed and altitude information at hand, as formerly only known form the laboratory. In that sense, IMU sensors may complement established overall distance and speed information for the runner as a whole by highly time-resolved acceleration data for all involved body segments. By that, IMUs might possibly be used in the future to reduce individual

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overuse injury risks and to determine energetic cost and efficiency of running. Altogether, in view of powerful state-of-the-art IMU sensors (and their continuously rapid rate of improvement driven by smart phones and mobile devices), practitioners in sports science and medicine are likely to benefit from IMU technology in their daily work and are thus encouraged to take advantage of it for improving exercise control and efficiency.

Conclusions for practitioners

- Modern wearable IMUs are lightweight, small and operate wirelessly, rendering them a promising tool for assessing biomechanical loads and asymmetries in running under field conditions.
- A certain degree of asymmetry in peak tibial and peak shoulder accelerations (as measured with IMUs) proves statistically "normal". For healthy junior elite long-distance runners, mean absolute asymmetries of $9 \pm 8\%$ (95^{th} percentile of 24%) and $9 \pm 10\%$ (32%) are observed in tibial and scapular 3D acceleration, respectively. Asymmetries are found to be rather independent of running speed. In case of peak acceleration asymmetries exceeding 20–30% in a healthy runner, we recommend an individual, integral orthopaedic examination by an experienced (sports) orthopaedist.
- Lateral asymmetries in running may either be habitual and thus unnecessary and unfavourable, or reflect sensible (subconscious) compensatory strategies for anatomic/orthopaedic conditions. In the latter case, a "correction" would be disadvantageous.
- In junior elite long-distance runners, impact shock is on average reduced to $32 \pm 8\%$ from distal tibiae to sacrum and to $27 \pm 6\%$ from tibiae to scapulae (i.e. $-14 \pm 7\%$ from sacrum to scapulae).
- Briefly summing up the findings on tibial accelerations, load may effectively be diminished by (1) reducing running speed, (2) running on turf and (3) running uphill. In particular, reducing running speed exhibits an approximately twice larger effect on tibial load than changing the ground surface from rigid to more elastic conditions, e.g. from asphalt to tartan.
- For a valid assessment of acceleration load in running, 3D (i.e. triaxial) IMUs should be used. In 1D and magnitude-only measurements, some biomechanical aspects may remain undiscovered.

Conflict of interest

There is no conflict of interest.

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