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ORIGINAL PAPER

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Summary

Background: Hypogravity treadmills like the AlterG® have become a popular exercise tool especially in return-to-sports and return-to-competition rehabilitation programmes for distance runners, triathletes, footballers and further sports disciplines. By applying a slight positive air pressure in a chamber around the lower limbs, the effective bodyweight force is reduced, thereby reducing gravity load to the musculoskeletal system. So far it was unclear in how far that hypogravity unloading also affects peak accelerations at the tibiae, a common site of running-related overuse injuries.

Material and Methods: Fifteen well-trained male runners (peak oxygen consumption of 60 ± 4 ml min^{-1} kg^{-1}) completed three incremental treadmill tests until volitional exhaustion in randomised order, two of which on the hypogravity treadmill AlterG® at 80% and 60% effective bodyweight, respectively, and the third on a conventional treadmill, i.e. at 100% gravity. Triaxial accelerations at the distal portions of both tibiae were captured throughout the three (hypo-)gravity conditions and all running speeds, along with the physiological cost of running in terms of heart rate, oxygen consumption and rating of perceived exertion.

Results: Mean peak tibial accelerations at impact and active push-off amounted to 12.9 ± 2.3 g_0, 12.6 ± 1.9 g_0 and 12.5 ± 2.3 g_0 for 12–18 km h^{-1} at 100%, 80% and 60% effective bodyweight, respectively ($g_0 \approx 9.81$ m s^{-2}). They were not reduced by hypogravity unloading within measurement uncertainty ($p = 0.668$). However, they exhibited a clear dependence on running speed ($p < 0.001$), with mean values ranging from 10.0 ± 2.0 g_0 at 12 km h^{-1} to 15.4 ± 2.2 g_0 at 18 km h^{-1}. Stride period was prolonged under hypogravity (+3% and + 7% at 80 % and 60 %, respectively), while maximal running speed was higher (+20% and + 30%). Because of those higher maximal running speeds under hypogravity, peak tibial accelerations at a

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Running on the hypogravity treadmill AlterG® does not reduce the magnitude of peak tibial impact accelerations

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Introduction

AlterG® running has recently become a popular measure in return-to-sports and return-to-competition programmes for runners, triathletes, footballers etc. as part of their rehabilitation programme after injuries in the lower limbs, e. g. the hip, pelvis, knee, ankle, Achilles tendon, thigh or shank. The AlterG® treadmills (AlterG®, Fremont, California, USA) are class of motorised treadmills that support the athlete by so-called "hypogravity", i.e. a lower effective gravitational force (= weight force) of their body as compared to normal conditions. To accomplish that bodyweight support, the AlterG® treadmills employ the concept of a so-called "lower body positive pressure treadmill": By a positive air pressure $\Delta p = p_{in} - p_{out} > 0$ inside a chamber around the lower limbs (Fig. 1), with p_{in} denoting the chamber pressure and p_{out} the ambient room pressure, the athlete is statically lifted upwards. Thereby, the acting gravitational force $F_G = mg_0$ (where m is the athlete's mass and $g_0 \approx 9.81$ m s^{-2} standard acceleration of free fall) is counteracted by the positive pressure gradient force $F_{\Delta p}$ (Fig. 1), yielding an effective net bodyweight force $F_{G,eff} = F_G - F_{\Delta p}$. In relative terms, this can be written as

$$\text{Effective bodyweight} := \frac{F_{G,eff}}{F_G} = \frac{F_G - F_{\Delta p}}{F_G} = \frac{m\,g_{eff}}{m\,g_0} = \frac{g_{eff}}{g_0} \leq 100\%$$

using the definition of the net effective gravitational acceleration $g_{eff} := \left(F_G - F_{\Delta p}\right)/m \leq g_0$.

On the AlterG®, effective bodyweight can be freely adjusted in steps of 1 % from 100 % down to 20 %, the latter roughly corresponding to effective bodyweight on the Moon (there, however, due to the lower mass of the Moon as compared to Earth and not due to (absent) air pressure). The exact value of

comparable physiological cost were significantly elevated on the AlterG® ($p < 0.001$). Increases amounted to + 1.8 g_0 and + 2.5 g_0 at 80% and 60% as compared to 100% effective bodyweight, respectively.
Conclusions: Distal tibial accelerations are not reduced in hypogravity running on the AlterG® treadmill in terms of absolute speeds, and are elevated in terms of relative speeds, i.e. at running speeds demanding the same physiological effort. Thus, precaution should be taken by clinicians and coaches when planning rehabilitation programmes for athletes with a recent injury background at the distal tibiae. The cause for the ineffectiveness of hypogravity running in reducing tibial acceleration load is most probably a subconscious energetic optimisation in human running motor control: For energetic reasons, the leaps of each step become flatter under hypogravity instead of steeper, whereas the latter would have intuitively been expected because of the lower effective acceleration of free fall.
Level of Evidence: Ib

Keywords
Acceleration – Anti-gravity – Bodyweight support treadmill – Inertial measurement unit (IMU) – Tibia

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Laufen auf dem Teilschwerelosigkeitslaufband AlterG® verringert nicht tibiale Impact-Beschleunigungsspitzen

Zusammenfassung

Hintergrund: Teilschwerelosigkeitslaufbänder wie das AlterG® stellen ein beliebtes Trainingsmittel in der posttraumatischen Rehabilitation von Läufern, Triathleten, Fußballer sowie Vertretern weiteren Sportarten dar. Durch einen geringen Überdruck in einer die unteren Gliedmaßen umschließenden Kammer wird die effektiv einwirkende Gewichtskraft des Körpers beim Laufen auf dem AlterG® reduziert,

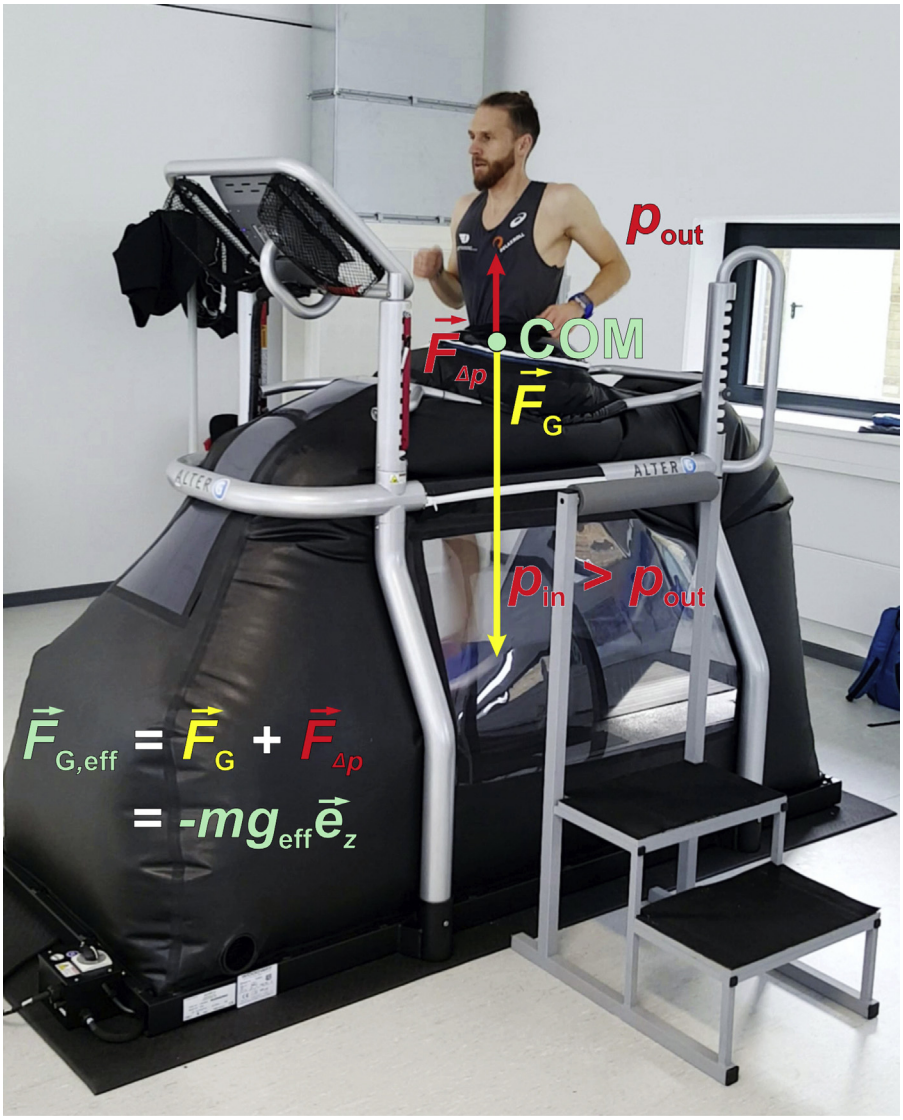

Fig. 1
Working principle of the employed hypogravity treadmill AlterG®. As illustrated, the effective bodyweight force $A_{80\ \%}/A_{100\ \%}$ is the difference of the standard weight force $A_{60\ \%}/A_{100\ \%}$ pointing downwards (i.e. in negative z axis direction $-\vec{e}_z$) and the positive air pressure gradient force F_p that points upwards and arises from different air pressures inside (p_{in}) and outside (p_{out}) the chamber. While the weight force acts directly on the athlete's centre of mass (COM), the application point of the pressure gradient force is situated slightly displaced in the centre of buoyancy of the body parts covered by the AlterG® chamber (not shown in sketch). To good approximation, however, that offset is negligible, so that resulting additional torques on the runner are minimal.

positive pressure v/v_{max} required to reduce effective bodyweight to a certain degree, e.g. to 80 %, depends on the athlete's anthropometry, and usually ranges between 10 to 100 mbar. Such small pressure differences are known from everyday life, e.g. when due to weather conditions, a low pressure area is replaced by a high pressure area. The positive pressure applied in the AlterG® treadmills can thus be

wodurch sich die Gewichtsbelastung des muskuloskelettalen Systems insgesamt verringert. Bislang war nicht bekannt, inwiefern sich diese Gewichtsentlastung auf die Beschleunigungsspitzen an den Schienbeinen, eine der häufigsten Lokalisationen von laufbedingten Überlastungsverletzungen, übertragen lässt.

Material und Methoden: Fünfzehn trainierte männliche Läufer (Spitzensauerstoffaufnahme 60 ± 4 ml min^{-1} kg^{-1}) absolvierten drei Stufentests in randomisierter Reihenfolge bis zur individuellen Ausbelastung. Zwei Tests fanden auf dem Teilschwerelosigkeitslaufband AlterG® bei 80% bzw. 60% effektivem Körpergewicht statt, während ein dritter Test auf einem konventionellen Laufband, d. h. bei 100% effektiver Gewichtskraft, durchgeführt wurde. Während der Tests wurden die auf die distalen Schienbeine wirkenden dreidimensionalen Beschleunigen durchgehend mikroelektronisch gemessen. Zusätzlich wurde die physiologische Beanspruchung in Bezug auf Herzfrequenz, Sauerstoffaufnahme und subjektives Belastungsempfinden erfasst.

Ergebnisse: Die gemittelten Spitzenbeschleunigungen in Impact- und Abdruckphase beliefen sich auf 12,9 ± 2,3 g_0, 12,6 ± 1,9 g_0 bzw. 12,5 ± 2,3 g_0 bei 100%, 80% bzw. 60% effektivem Körpergewicht ($g_0 \approx 9{,}81$ m s^{-2}) und einer Laufgeschwindigkeit von 12–18 km h^{-1}. Es konnte im Rahmen der Messgenauigkeit keine Reduktion durch die Teilschwerelosigkeit festgestellt werden ($p = 0.668$). Die Beschleunigungsspitzen zeigten lediglich eine klare Abhängigkeit von der Laufgeschwindigkeit ($p < 0.001$), wobei die Mittelwerte im Bereich von 10,0 ± 2,0 g_0 bei 12 km h^{-1} bis 15,4 ± 2,2 g_0 bei 18 km h^{-1} lagen. Die Schrittdauer war unter Teilschwerelosigkeit erhöht (+3% bzw. + 7% bei 80 % bzw. 60 %), wie auch die maximal erreichte Laufgeschwindigkeit (+20% bzw. + 30%). Infolge dieser höheren Maximalgeschwindigkeit unter Teilschwerelosigkeit fielen die Impact-/Abdruckbeschleunigungen auf dem AlterG® bei einer vergleichbaren

regarded as uncritical to the human organism even in the context of most cardiovascular disorders.

For correctly applying the AlterG® treadmill in sports and rehabilitation sciences, it is important to see that the effective bodyweight force $F_{G,eff} = F_G - F_p = mg_{eff} \leq F_G = mg_0$ is altered, but not the athlete's mass m, as required by the basic physical principle of mass conservation. Therefore, inertial forces as given by Newton's Second Law $F = ma$ stay entirely unaffected by hypogravity.

The tibia belongs to the common sites of overuse injuries in distance running and related sports disciplines [1–8]. Tibial accelerations are commonly considered a valid proxy measure for the impact forces experienced at the tibia by clinicians and researchers [1]. Previous studies have shown that hypogravity running implies lower vertical ground reaction forces [9–11]. However, the effect of hypogravity on triaxial (i.e. 3D) tibial impact and active push-off accelerations, i.e. during the so-called impact and active peak [12], has not been elucidated yet. So far, only selective results about uniaxial (1D) accelerations have been reported [13,14]. In uniaxial measurements of tibial accelerations, however, important information, e.g. on shear components in the acceleration vector, may stay concealed. Hence, there is a research deficit, given the growing number of AlterG® training sessions run in the context of injury rehabilitation programmes around the world on the one hand and the proven positive correlation between sustained tibial accelerations and the incidence of running-related overuse injuries on the other. The present study intends to close this highly topical knowledge gap by providing new experimental data and a biomechanical explanation for the observed effects.

Methods

Study design

All subjects completed three incremental treadmill tests in randomised order. Two treadmill tests were performed on the hypogravity treadmill AlterG® Anti-Gravity Treadmill® Pro 200 Plus (i.e., the training device with a maximum forward belt speed of 28 km h^{-1}; AlterG®, Fremont, California, USA) with an effective bodyweight setting of 80% and 60%, respectively. The third test was completed on a conventional treadmill h/p/cosmos saturn® 250/100 without hypogravity support, i.e. at 100 % effective bodyweight (h/p/cosmos sports & medical GmbH, Nußdorf-Traunstein, Germany). A conventional treadmill was chosen rather than the AlterG® at a setting of 100 % effective bodyweight to ensure that any residual vertical and/or horizontal support possibly given by the AlterG® belt was absent, so that running dynamics were entirely "natural" and closer to daily training routine.

Each of the three incremental tests started with a running speed of 6 km h^{-1}, which was increased in steps of + 2 km h^{-1} every 3 min. Between the speed stages, a break of 30 s with resting treadmill belt was made to facilitate gathering rating of perceived exertion (RPE) according to the Borg scale [15]. Each incremental test was performed until individual volitional exhaustion. All subjects successfully completed the stages until (including) 18 km h^{-1} with whole stage duration. From 20 km h^{-1} on, some athletes had to abort the test before the end of the current stage. To ensure the same number of data sets for each speed stage, those bouts beyond 20 km h^{-1} were thus disregarded in later data analysis. The speed stages of 6, 8 and 10 km h^{-1} were used as a

physiologischen Beanspruchung höher aus ($p < 0.001$). Dieser relative Zuwachs lag bei + 1,8 g_0 bzw. + 2,5 g_0 für 80% bzw. 60% effektives Körpergewicht im Vergleich zur Referenz von 100 %.
Schlussfolgerungen: Der Betrag der an den distalen Tibiae wirkenden Beschleunigungsspitzen wird durch die Teilschwerelosigkeit auf dem AlterG ® bei vorgegebener absoluter Laufgeschwindigkeit nicht verringert. In Bezug auf eine Laufgeschwindigkeit, die eine vergleichbare physiologische Beanspruchung erfordert, fallen die wirkenden Beschleunigungsspitzen sogar größer aus. Sportärzte und Trainer sollten daher äußerst vorsichtig sein, wenn sie die Nutzung eines Teilschwerelosigkeitslaufbandes für die Return-to-Sports- bzw. Return-to-Competition-Rehabilitation von tibialen Verletzungen in Erwägung ziehen. Der Grund, weswegen die Teilgewichtsentlastung nicht zu einer Verringerung der tibialen Beschleunigungsspitzen führt, liegt höchstwahrscheinlich in einer unterbewussten Optimierung der menschlichen Laufökonomie begründet: Aus energetischen Gründen wählt der gesunde Mensch eine geringere Sprunghöhe der Schritte unter Teilschwerelosigkeit anstatt einer größeren zu folgen, wobei Letzteres angesichts der verminderten effektiven naiv zu erwarten wäre.
Evidenzlevel: Ib

Schlüsselwörter
Beschleunigung – Inertialsensor – Teilschwerelosigkeitslaufband – Teilgewichtsentlastung – Tibia

standardised warm-up. Directly after test abortion, final RPE was gathered. Throughout each test, tibial accelerations, heart rate and oxygen were continuously acquired (see next section for details).
The subjects completed all tests with the same, individual pair of running shoes to exclude any footwear-related bias. All tests were scheduled at the same time of day (± 30 min) with a seven-day break in between. The runners agreed to accomplish similar patterns of training before the trials, including a tapering period without exercises of 24 h and a similar nutrition pattern of at least 48 h in advance.

Subjects

Fifteen well-trained male athletes (30 ± 7 years, 181 ± 6 cm, 73 ± 6 kg, peak oxygen consumption 60.2 ± 3.8 ml min^{-1} kg^{-1}, personal best in 10 km road running 34:06 ± 2:26 min) successfully completed three incremental treadmill tests. Criterion for inclusion was a personal best in 10 km road or track running better than 40:00 min. All participants confirmed to be familiar with treadmill running, usually completing several treadmill sessions per year. The study was approved by the ethics committee of the Humboldt-Universität of Berlin and is in accordance with the Helsinki declaration. All subjects gave written consent to participate in the study.

Measurement setup

To measure tibial accelerations, the established and validated inertial measurement unit (IMU) system MTw Awinda (Xsens Technologies B.V., Enschede, Netherlands) was used. Those triaxial sensors are small in dimensions (4.7 3.0 1.3 cm^3), lightweight (16 grams) and operate wirelessly. The data output rate after sensor fusion was set to 120 Hz, while accelerations were internally acquired at 1,000 Hz for each spatial axis and strapped down for transmission. Peak value amplitudes are preserved by the employed algorithms [16–19]. Each axis had a sufficient measurement range of ± 16.3 g_0 and a maximal measurement uncertainty of ± 0.002 g_0. Accelerations were processed in raw-data form as returned by the microelectronic accelerometers within the co-moving reference frame of the IMU housing (inset in Fig. 2) by employing the proof mass method. Gyroscope and magnetometer data were not used. The two triaxial IMUs were fixed noninvasively at the anterior distal portion of each tibia by means of tightly and comfortably fitting Velcro™ straps, and additionally by Velcro™ a second layer straps on top (inset of Fig. 2). By this two-fold fixation, relative motion between IMU and skin was minimal even for repetitive, high-impact accelerations of up to 16 g_0, as confirmed by high-speed video camera footage (taken at 300 fps, data not shown). Further technical details are described in [20,21].
For assessing physiological load, heart rate was monitored using a wireless chest belt (HRM Run™, Garmin Ltd., Canton Schaffhausen, Switzerland). Oxygen consumption was measured by a stationary breath-by-breath system (Quark CPET, COSMED, Pavona di Albano, Italy) and evaluated using proprietary software (OMNIA 1.6, COSMED, Pavona di Albano, Italy). Oxygen uptake values were averaged over 30 s [22]. To account for the incremental nature of the tests, the oxygen consumption during the last 30 s of each 3 min running bout were regarded as most representative for the pertinent speed stage in each test.

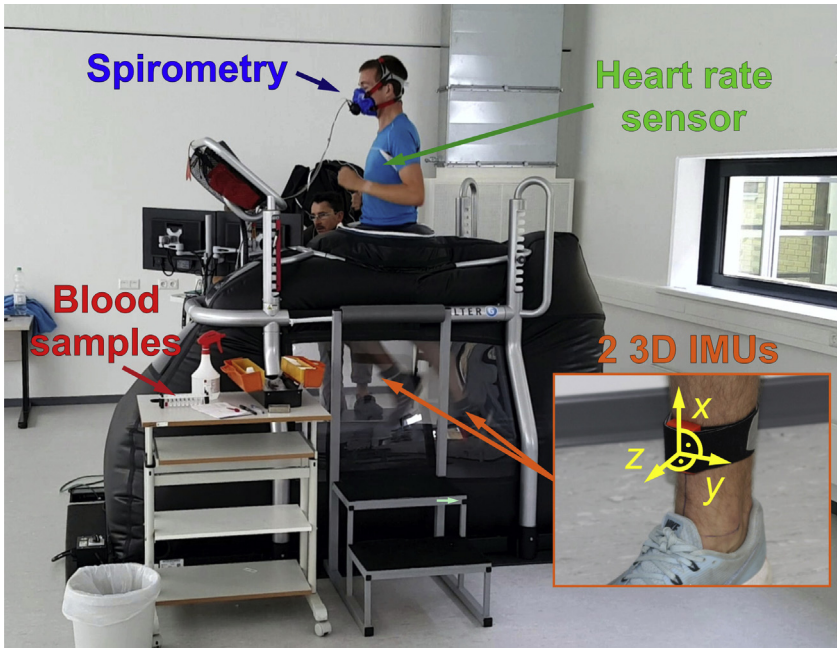

Fig. 2
Measurement setup. For tibial acceleration measurements, two triaxial inertial measurement units (IMUs) were worn at the distal tibiae. The physiological state of the athlete was continuously monitored by a chest-worn heart rate sensor and a stationary breath-by-breath spirometry system. In addition, capillary blood samples from the ear lobe were gathered between the speed stages for lactate measurements along with the rating of perceived exertion (see text for details).

Acceleration data analysis

Acceleration data were pre-processed using proprietary software of the manufacturer (MT Manager 4.8, Xsens Technologies B.V., Enschede, Netherlands). Of the 3-min periods of each running bout, the first and last 30 s were disregarded to exclude any initial/final distortion effects. The remaining 2 min of each bout were analysed for impact and active push-off events in 3D acceleration magnitude, the latter given by

$$a(t) := |\vec{a}(t)| = \sqrt{(a_x(t))^2 + (a_y(t))^2 + (a_z(t))^2} \quad (2)$$

using a custom-made automatic LabView 2016 routine (National Instruments, Austin, Texas, USA). Roughly 160 impact and active push-off events were detected for each leg and each 2 min period. The detected peak amplitudes were averaged for each speed stage, yielding the mean 3D peak acceleration amplitude $a_{peak}\left(g_{eff}, v\right)$ as a function of hypogravity condition g_{eff} and running speed v. Based on those impact and active push-off peaks, mean stride period $T_{stride}\left(g_{eff}, v\right)$ was calculated as the mean time lag between two neighbouring peak events.

In addition to the figures for absolute running speed in terms of kilometres per hour, also an individual relative running speed v_{rel} (in %) was calculated, as defined by the ratio of current running speed v over individually achieved maximal running speed $v_{\max}\left(g_{eff}\right)$ for a given hypogravity condition g_{eff}, i.e.

$$v_{rel}\left(g_{eff}\right) := \frac{v}{v_{\max}(g_{eff})} \quad (3)$$

To facilitate a convenient, direct numerical comparison of acceleration data in terms of those subject-dependent, individual relative speeds, equidistant data points were computed as group means for relative speeds of 50 % » v_{rel} » 100 % π by applying linear interpolation (while ensuring sufficiently high coefficients of determination of $0.92 \leq R^2 \leq 0.99$). Finally, a cumulative training session load A as a function of relative running speed

$$A\left(g_{eff}, v_{rel}\right) = \frac{60\ \mathrm{min}}{T_{stride}(g_{eff}, v_{rel})}\, a_{peak}\left(g_{eff}, v_{rel}\right) \quad (4)$$

was calculated for a hypothetical training session of 60 min for each treadmill setting, i.e. an effective bodyweight g_{eff}/g_0 of 100%, 80% and 60%, respectively, and compared to each other.

Statistical analysis was performed using IBM SPSS Statistics 22 (IBM, Armonk, USA). To exclude any systematic physiological bias among the different treadmill settings, a one-way analysis of variance (ANOVA) was performed for the dependent variables *maximal running velocity* and *rating of perceived exertion* (RPE) and confirmed comparability of the treadmill settings. Moreover, a two-way ANOVA was conducted for the dependent variables *stride period* and *peak tibial acceleration* based on the two independent variables *effective bodyweight* (i.e., 100%, 80%, 60%) and *absolute running speed* (12–18 km h^{-1}). A second separate ANOVA was done for *stride period* and *peak tibial acceleration* based on the two independent variables *effective bodyweight* (100%, 80%, 60%) and *relative running speed* (50 % – 100 %). Additionally, a one-way ANOVA for

the ratio of training session loads $A_{80\ \%}/A_{100\ \%}$ and $A_{60\ \%}/A_{100\ \%}$ between hypogravity settings was conducted in terms of the independent variable *relative running speed* (50 % – 100 %). Variance homogeneity was confirmed using Levene's test ($p > 0.05$). Level of significance was defined with $p < 0.05$. Bonferroni correction was used for post-hoc multiple comparison of means for main effects and significant interactions. Standard deviations (SD) are used for error bars in the graphs and confidence levels in the text.

Results

As regards physiological effort, heart rate ($p = 0.300$), maximal oxygen consumption $V_{O_2,\ peak}$ ($p = 0.078$) and Borg scale RPE ($p = 0.446$) did not differ between the three tests, see Fig. 3. However, the maximal running speed at test abort was significantly higher under hypogravity ($p < 0.001$), yielding 19.2 $\pm$ 1.2 km h^{-1}, 22.6 $\pm$ 1.4 km h^{-1} and 24.9 $\pm$ 1.7 km h^{-1} at 100%, 80 % and 60 % effective bodyweight, respectively (green columns in Fig. 3). Relative increases amounted to + 20 $\pm$ 5% and + 30 $\pm$ 7% for 80 % and 60% as compared to 100% effective bodyweight.

As for stride period, a significant dependence on effective bodyweight, i.e. 100%, 80% or 60%, ($p < 0.001$, $\eta^2 = 0.228$) and absolute running speed ($p < 0.001$, $\eta^2 = 0.329$) was found. In particular, stride period was on average by 20 ms and 48 ms ($\triangleq$ + 3% and + 7%) higher at 80% and 60% as compared to 100% effective bodyweight (Fig. 4 a). In terms of relative running speed, however, the hypogravity setting did not alter stride period ($p = 0.118$, $\eta^2 = 0.017$; Fig. 4 b).

Concerning tibial accelerations, peak tibial impact and active push-off accelerations exhibited a significant dependence on absolute running speed ($p < 0.001$, $\eta^2 = 0.483$). However, they were not different at 80 % and 60 % effective bodyweight as compared to 100 % ($p = 0.668$, $\eta^2 = 0.005$), see Fig. 5 (a) and (b). In addition, when looked at in terms of relative running speed, significantly higher peak tibial accelerations were observed under hypogravity of 80% and 60% than at standard gravity of 100% ($p < 0.001$, $\eta^2 = 0.204$). In particular, peak tibial accelerations at 80 % and 60% were on average (mean all stages of all individuals) by + 1.8 g_0 and + 2.5 g_0 higher than at 100%.

As regards vector components, both peak acceleration in axial direction, i.e. along the tibia, and in the horizontal plane, i.e., in the combined anteroposterior and mediolateral direction, possess comparable values that depend on running speed but are independent from hypogravity condition (Fig. 6). Their overall means amount to 68 $\pm$ 1 % and 73 $\pm$ 2% for the axial and shear component, respectively.

Finally, cumulative session load ratios $A_{80\ \%}/A_{100\ \%}$ and $A_{60\ \%}/A_{100\ \%}$ were significantly greater than 1.00, indicating that cumulative sessions were greater under hypogravity than under normal gravity ($p < 0.001$). They proved independent of relative running speed ($p = 0.811$ and $p = 0.987$, respectively, Table 1). Hence, cumulative tibial acceleration load was

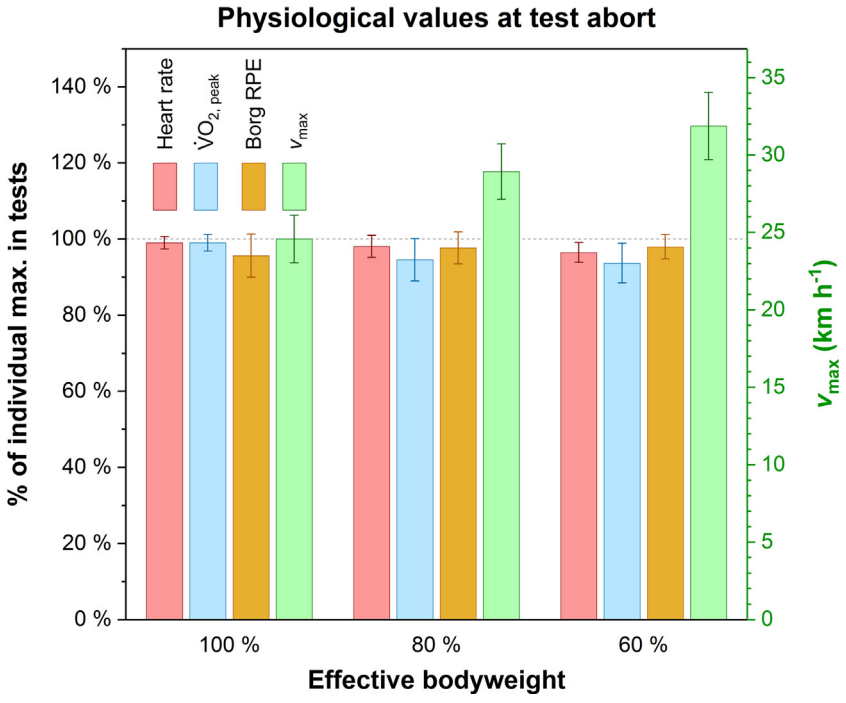

Fig. 3
Physiological values at volitional test abort. The four columns (heart rate, peak oxygen consumption, Borg scale of rating of perceive exertion and maximal running speed) depict the mean ratio $\pm$ SD as percentage of the individually achieved value at test abort for each (hypo-)gravity condition and the corresponding maximal value individually achieved throughout all three tests (left scale). The right scale shows the mean running speed at test abort (maximal speed) in absolute terms, i.e. km h^{-1}.

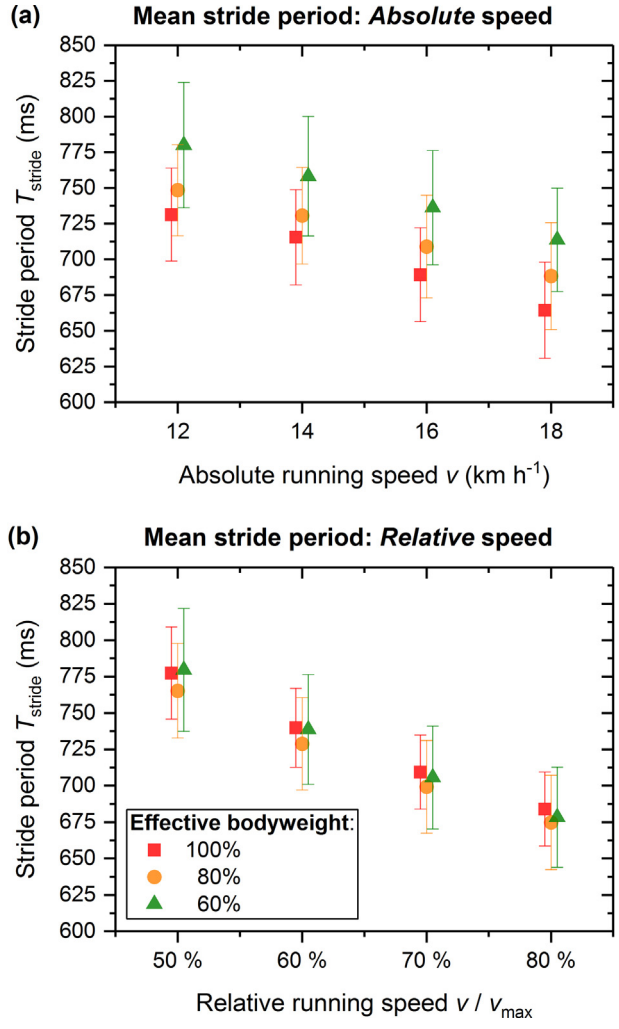

Fig. 4
Stride period as a function of effective bodyweight (colours) and (a) absolute or (b) relative running speed, respectively. Please note that the data points (mean ± SD) have been plotted with a small artificial horizontal offset to avoid visual convergence.

found to be higher under hypogravity for a given relative running speed, despite the prolonged stride periods under hypogravity. No significant difference was found for 80% vs. 60% effective bodyweight.

Discussion

This study has yielded three main findings on hypogravity running on the hypogravity treadmill AlterG®: First, when performing an incremental treadmill test until volitional exhaustion, the physiological parameters heart rate, maximum oxygen consumption and rating of perceived exertion at test abort do not differ under hypogravity of 80% and 60% effective bodyweight as compared to standard gravity. However, maximum running speed is by 20 ± 5% and 30 ± 7% higher at 80% and 60%, respectively, than at 100%.

Second, stride period increases under hypogravity by 20 ms and 48 ms at 80% and 60% effective bodyweight, respectively, as compared to standard gravity. This finding is in accordance with current results from other authors, reporting that time of flight on hypogravity treadmills increases while stance time is practically unchanged [23,24]. Interestingly, we can extend those results by our finding that time of flight has the same value under hypogravity as under normal gravity when relative running speeds are considered, i.e. when it is accounted for that the athlete can run faster at the same physiologic cost under hypogravity. The third major result, and in our option the most important of this study, is that the magnitude of 3D peak tibial accelerations at impact and active push-off during stance phase is not reduced under hypogravity for a given absolute running speed (i.e., a running speed in terms of kilometres per hour), as mistakenly – yet commonly – presumed by many athletes and coaches. Taking into account the above result that the athletes can run faster under hypogravity at the same physiologic cost, this means that peak tibial accelerations at a comparable physiological effort are even higher for AlterG® running than on a standard treadmill, as shown by our data on relative running speeds. This overall finding is in accordance with the results of two previous studies that investigated 1D (axial) tibial accelerations on a hypogravity treadmill and found that peak tibial accelerations were unchanged [14] or at least independent of the level of hypogravity [13]. The results of our present study extend those findings by confirming and further qualifying them for triaxial IMU measurements. Moreover, some researchers [14] speculated that the observed

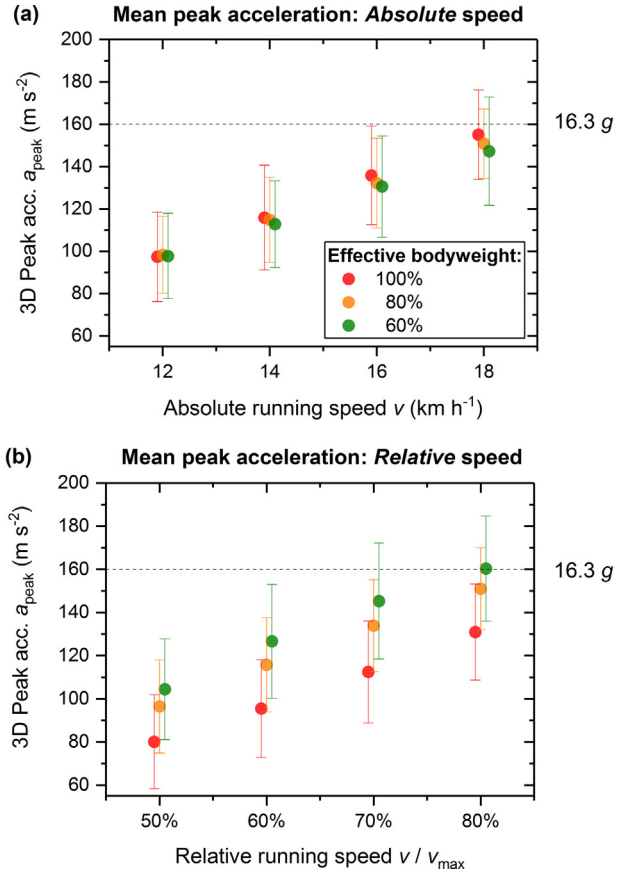

Fig. 5
Tibial peak impact and active push-off acceleration magnitude (3D) as a function of effective bodyweight (colours) and (a) absolute or (b) relative running speed, respectively. Please note that the data points (mean ± SD) have been plotted with a small artificial horizontal offset to avoid visual convergence.

reduction in step rate under hypogravity (i.e. the increase in stride periods) may be the reason nullifying any hypogravity unloading effect at the tibia and thus result in unchanged tibial accelerations under hypogravity. In contrast, our results on cumulative session loads prove that the effect of higher tibial accelerations (in terms of relative running speeds) is not compensated for by the longer stride periods under hypogravity, even despite the lower number of impact and active push-off events per leg for a given running time. Instead ratios well beyond 1.00 are observed, indicating that the hypogravity effect on peak tibial accelerations is higher than on stride period, so that no nullification is achieved. Moreover, our finding that cumulative session loads for 80% and 60% effective bodyweight are practically the same suggests that further increasing the level of hypogravity unloading is equally unlikely to result in a nullification. Thus, we recommend running at the higher effective bodyweight in this respect, i.e. at 80% instead of 60%, to stay closer to the normal spatiotemporal stride structure.

In our perception, the implications of our present findings on tibial impact and active push-off accelerations in hypogravity running may be crucial to practitioners in sports rehabilitation employing treadmills like the AlterG®: As stated in the Introduction, tibial accelerations are commonly regarded as a valid surrogate measure for the mechanical forces experienced at the tibia [1]. As such, our results suggest that impact and active push-off compression forces in hypogravity running on the AlterG® treadmill (1) are not reduced for a given absolute running speed and (2) are even increased for a given relative speed, i.e. a running speed demanding the same physiological cost as under standard gravity. In this respect it is imperative to see that the experienced tibial accelerations can readily be transformed into acting inertial forces on the tibiae by applying Newton's Second Law $\vec{\mathbf{F}} = m\vec{\mathbf{a}}$, irrespectively of hypogravity unloading. The reason is, as already stated in the Introduction section above, that only the weight force $\vec{\mathbf{F}}_G = m\vec{\mathbf{g}}_0$ pointing downwards is counteracted by a positive pressure force $\vec{\mathbf{F}}_{\Delta p}$ pointing upwards, whereas the athlete's mass m (and along with it the masses of the thighs and shanks in particular) are not changed according to the basic principle of mass conservation in physics (cf. also [25]). Therefore, a higher tibial acceleration magnitude $a = |\vec{\mathbf{a}}|$ (cf. Eq. (2)) inevitably corresponds to a higher inertial force magnitude $|\vec{F}| = m|\vec{a}| = ma$ and vice versa –a point that is unfortunately sometimes overseen by practitioners. Given the direction of tibial impact and active push-off accelerations mainly pointing proximal along the tibiae, those forces can be regarded as proxy measures for an osseous compression load.

Nonetheless, despite that non-reduction in compression load, it needs to be noted that tensile

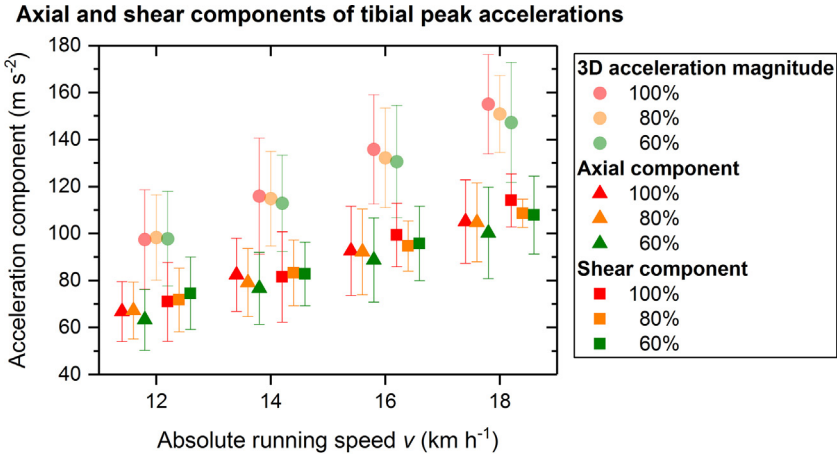

Fig. 6
Tibial peak impact and active push-off acceleration vector components as a function of effective bodyweight (colours). Triangles represent axial components, i.e., acting along the tibia, while squares depict corresponding shear components, i.e., in antero-posterior and mediolateral direction. Please note that the data points (mean ± SD) have been plotted with a small artificial horizontal offset to avoid visual convergence.

Table 1. Calculated cumulative session loads (cf. Eq. (4)).

Relative speed v/v_{max}	Ratio $A_{80\ \%}/A_{100\ \%}$	Ratio $A_{60\ \%}/A_{100\ \%}$
50%	1.29 ± 0.40	1.31 ± 0.61
60%	1.27 ± 0.32	1.30 ± 0.55
70%	1.24 ± 0.26	1.26 ± 0.48
80%	1.25 ± 0.30	1.26 ± 0.25

osseous load may still be reduced in view of the provably lower overall physiological cost of hypogravity running: Because of the latter, mean muscular work is presumably lower, thereby – hypothetically – reducing tensile stress on the bone at the tendon insertion points, e.g., those of the knee-extending quadriceps muscles. Thus, hypogravity unloading might reduce tensile osseous load via a reduced muscular performance, even though compression stress, as experienced during impact and active push-off, is not, as shown by our data. Hence, future research is recommended to elucidate the overall osseous stress load in hypogravity running.

Altogether, we conclude that athletes with an injury background at their distal tibiae should be cautious when including hypogravity running sessions in their return-to-sports programme. In particular, to avoid undesired additional stress to the tibia, the individually familiar (absolute) running speed (as chosen on a standard treadmill or in a track/road session) should not be exceeded during first phases of rehabilitation, even though physiological costs are likely to be lower on the AlterG® than on a normal treadmill. Nonetheless, we emphasize that these precautions are only indicated by our results concerning injuries of the precise anatomic site we studied, i.e. the distal tibiae. Generalisations are discouraged. For instance, injury sites near the point where ground reaction forces act on the athlete, i.e. the athlete's centre of mass, are likely to be unloaded by AlterG® running, as other researches showed that ground reaction forces are reduced by hypogravity unloading [9,11,25–27], e.g. the pelvis or the lower back area. In fact, the successful employment of AlterG® running for rehabilitation purposes after a sacral stress fracture has been recently reported in elite marathon running [28]. Hence, in essence, a clear differentiation is generally needed between ground reaction forces (acting on the centre of mass) and inertial forces (acting at the distal body segments, such as the tibiae).

To finally answer the question why tibial accelerations are not reduced under hypogravity, it is necessary to briefly look at the principles of human running gait motor control under both biomechanical and energetic aspects. Polet et al [29,30]. recently found in experiments that in hypogravity running a counter-intuitive change in spatiotemporal stride structure takes place, manifesting itself in a reduced leap height – where intuitively an increased leap height would have been expected because of a lower effective acceleration of free fall $g_{eff} < g_0$.

To understand this surprising behaviour, Polet et al. investigated in detail the different contributions to energy cost of running. They found that the stance-phase cost, i.e. the work associated with the generation of the vertical impact/push-off impulse (and its absorption in the next step), represents only one of two energetic key contributions. The second is given by the muscular cost of swing phase, i.e. the motion of moving back the

leg from behind the body's centre of mass to its front after take-off during flight phase. The researchers experimentally demonstrated and mathematically showed that human locomotion (apparently subconsciously) adapts to an unnaturally low gravity cost by re-adjusting the ratio between stance-phase cost and swing-phase cost, i.e. shifting the ratio of vertical vs. horizontal work per step towards a new optimum. Otherwise, the energetic cost of vertically deaccelerating and accelerating the centre of mass during stance phase would be uneconomically high as compared to the effort for swinging back the legs, a disproportion for which the human organism is obviously very sensitive for [29,30]. That adaptation results in a *smaller* vertical take-off velocity under hypogravity, yielding the observed counterintuitive effect of a reduced leap height under hypogravity [29,30]. By that, the ratio of the horizontal to the vertical component of the take-off velocity is changed towards a higher horizontal contribution. From other studies, it is known that the stance phase period is practically independent of gravity level [23]. Thus, a higher horizontal velocity during the same ground-contact period can only be achieved by higher absolute horizontal accelerations. Except for the very short moment in time when the tibia is exactly aligned perpendicular to the treadmill belt so that the vertical positive pressure lift force may support, this relation implies that higher horizontal acceleration magnitudes (as seen in the global reference frame) must act along the tibia during the stance phase. (Once again, it is emphasised in this context that only the vertical effective bodyweight force is reduced by the positive pressure inside the AlterG® chamber, whereas any horizontal forces are unchanged!) Finally, this picture is also supported by our findings of rather stable ratios of axial and shear components in peak tibial accelerations, suggesting that the direction of osseous load does not change, whereas the mean angle of propulsive force generation and absorption changes towards a more horizontal characteristic, i.e. flatter leaps, as seen in the global frame of reference.

To sum up, an adaptation in human running motor control under hypogravity towards lower vertical and higher horizontal work per step implies that tibial acceleration load is both partially decreased because of a reduced vertical take-off velocity, and partially increased due to a higher horizontal impulse. According to our experimental data, these two effects on 3D acceleration magnitude with opposing directions compensate for each other for a given absolute running velocity within measurement uncertainty, resulting in an unchanged tibial triaxial acceleration magnitude. Given the fact that the runners achieved higher maximal running speeds under hypogravity as compared to standard gravity, this also explains why in terms of relative speeds, observed tibial accelerations were higher under hypogravity.

Conclusion

In summary, the present study revealed that hypogravity unloading does not necessarily reduce peak impact/push-off acceleration magnitudes in lower limb segments. In fact, it was found that the distal portion the tibia is exposed to the same magnitudes of triaxial peak acceleration as compared to normal gravity for a given absolute running speed. However, when running under hypogravity at a speed matching the same physiological effort as under standard conditions, peak tibial acceleration magnitudes have been shown to be higher. Notably, these results do not contradict the well-accepted finding that vertical ground reaction forces are reduced by hypogravity unloading. Instead, a reduction in vertical ground reaction forces only implies that body segments near the athlete's centre of mass (e.g. the pelvis or the lumbar spine) are exposed to a lower acceleration load. There is no biomechanical principle, however, from which an analogous conclusion for distal segments like the calf, thigh or ankle etc. could be derived. In fact, our experimental findings and those of Moran et al [14]. suggest the contrary and are supported by the recent experimental and theoretical of Polet et al. on human motor control under hypogravity [29,30].

In essence, our results underline the need for a clear definition of the settings of a hypogravity running sessions in terms of session aim, unloading level, running speed and concomitant risks, especially within the context of return-to-sports and return-to-completion after tibial injuries. Future research is needed to further elucidate the biomechanics of hypogravity running and its implications for sports orthopaedic practitioners. For instance, a study of changed muscular recruitment patterns under hypogravity as assumed above, possibly manifesting themselves in, e.g., a higher hamstring vs. a lower quadriceps electromyographic activity, would be highly beneficial and would further support the biomechanical argumentation laid out above.

Conclusions for practitioners

Hypogravity running (e.g. in the AlterG® treadmill) does not reduce

peak impact/push-off acceleration magnitudes at the distal tibia, whereas vertical ground reaction forces are reported to be diminished. The underlying reason is most probably a subconscious shift in human running motor control towards flatter leaps, i.e. a higher contribution of horizontal and a lower contribution of vertical impact/push-off accelerations under hypogravity. Therefore, AlterG® running seems to be contraindicated for return-to-sports or return-to-competition measures after a stress fracture or stress syndrome of the distal tibial.

Conflict of interest

There is no conflict of interest.

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