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The measurement of tibial acceleration in runners—A review of the factors that can affect tibial acceleration during running and evidence-based guidelines for its use

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ABSTRACT

Background: Impact loading in runners, assessed by the measurement of tibial acceleration, has attracted substantial research attention. Due to potential injury links, particularly tibial fatigue fractures, tibial acceleration is also used as a clinical monitoring metric. There are contributing factors and potential limitations that must be considered before widespread implementation.

Aim: The objective of this review is to update current knowledge of the measurement of tibial acceleration in runners and to provide recommendations for those intending on using this measurement device in research or clinical practice.

Methods: Literature relating to the measurement of tibial acceleration in steady-state running was searched. A narrative approach synthesised the information from papers written in English. A range of literature was identified documenting the selection and placement of accelerometers, the analysis of data, and the effects of intrinsic and extrinsic factors.

Results and discussion: Tibial acceleration is a proxy measurement for the impact forces experienced at the tibia commonly used by clinicians and researchers. There is an assumption that this measure is related to bone stress and strain, however this is yet to be proven. Multi-axis devices should be secured firmly to the tibia to limit movement relative to the underlying bone and enable quantification of all components of acceleration. Additional frequency analyses could be useful to provide a more thorough characterisation of the signal.

Conclusions: Tibial accelerations are clearly affected by running technique, running velocity, lower extremity stiffness, as well as surface and footwear compliance. The interrelationships between muscle pre-activation and fatigue, stiffness, effective mass and tibial acceleration still require further investigation, as well as how changes in these variables impact on injury risk.

1. Introduction

Running is a popular activity, but the high participation rate is accompanied by a high incidence of injuries [1]. The majority of running-related injuries occur in the lower limbs, are chronic in nature, and are related to cumulative loading [2]. The repetitive impacts associated with running is thought to play an important role in the pathophysiology of many common running injuries, especially bony fatigue fractures (commonly termed stress fractures) [3–5]. In runners, between 35% and 49% of all fatigue fractures occur in the tibia [6–9].

While many factors influence bony remodeling and ultimately the manifestation of a fatigue fracture [10], biomechanics dictate the level of mechanical loading on bone during running [11,12]. When the foot strikes the ground, its velocity decelerates to zero and large ground reaction forces (GRF) are generated [13]. This momentum change produces compressive loading of the lower limbs, and results in an impact shock transmitted through the musculoskeletal system, with local segment peak accelerations occurring at successively later times [14,15]. To minimise damage to proximal structures the shock is attenuated, which is accomplished through an interaction of passive and active mechanisms [16–20]. A failure of the lower extremity muscles to adequately absorb the energy of impact may lead to an over-reliance on passive mechanisms for attenuation [20].

Direct in-vivo measurement of bone strain would be ideal for

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monitoring injury risk in runners, however this is invasive and impractical [21,22]. Measuring the tibial acceleration (TA) via segment mounted accelerometers is a commonly used proxy measurement for the impact forces experienced at the tibia by virtue of Newton's second law (F = ma) [23,24]. While the relationship between TA and bone strain is unclear, and likely to be complicated by local muscle forces, peak TA measured via devices attached directly to the tibia bone have revealed reasonable correlations with key GRF parameters (vertical impact peaks r = 0.7–0.85; loading rates r = 0.87–0.99) [25]. While the correlations are weaker when using skin-mounted accelerometers, average loading rate (r = 0.274–0.439) and instantaneous loading rate (r = 0.469) of the vertical GRF have all been significantly correlated with peak TA [26]. The moderate correlation between peak TA and GRF is not surprising as the GRF represents the summed acceleration of all body segments. These points withstanding, the axial component of TA has been shown to discriminate between runners with and without tibial fatigue fractures [27], and between runners injured and uninjured limb [28]. Additionally, the likelihood of the history of tibial fatigue fracture has been shown to increase by a factor of 1.4 for every 1 g increase in axial TA [29].

Previous literature reviews on the use of accelerometers in running have highlighted some of the key elements for consideration, such as the attachment method and placement location of the accelerometer, and the need for a low mass multi-axis device for increased measurement accuracy [23,24]. Despite this, the scope of these reviews did not address many of the issues and potential limitations that must also be considered when measuring TA from runners, including the influence of running velocity, technique, fatigue and surface characteristics. The objective of this review is to update current knowledge of the measurement of TA in runners and to provide recommendations for those intending on using this assessment method in research or clinical practice.

2. Methods

PubMed, Web of Science, SPORTDiscus and Google Scholar were searched to Jan 2018 using the following terms linked with the Boolean operators ('AND' and 'OR'): 'run*', 'tibia* acceler*', 'shock', 'inertia*' and 'biomech*', with no limits. Additional relevant studies were identified using article reference lists. Titles, abstracts and full-texts of retrieved documents were sequentially reviewed to determine their relevance. Only papers published in English, that specifically measured TA during steady-speed running, were included. Papers were excluded if they only assessed sprinting, or where participants used bodyweight support, or any form of implement or aid.

Findings from the literature covering the selection (Sect. 4.1), placement (Sect. 4.2) and attachment (Sect. 4.3) of accelerometers, as well as data analysis (Sect. 5) and key outcome measures (Sect. 6) are consolidated in the first half of the review. The second half of the review assesses the intrinsic (Sect. 7) and extrinsic (Sect. 8) factors that impact TA during running.

3. Definition of terms

A number of terms are used interchangeably to describe different aspects of TA, including peak TA, peak shank deceleration, peak positive acceleration and tibial shock. For the purpose of this review, axial (TA-A), anterior-posterior (TA-AP), and medio-lateral tibial acceleration (TA-ML) are used where time-domain peak acceleration magnitude components from a device aligned to the long axis of the tibia are reported. Resultant tibial acceleration (TA-R) is where the peak acceleration magnitude from all axes are used to calculate the resultant vector.

4. Tibial acceleration measurement

4.1. Device selection

Devices contain one, two or three accelerometers mounted at right angles, each reacting to the orthogonal component acting along their axis [30]. They operate relative to the Earth's gravitational field, constantly registering 9.81 m/s/s (1 *g*) as a reaction to gravitational acceleration [31]. The maximum contribution of the acceleration due to gravity is 1 *g* (when the shank is vertical), and some accelerometers will register 9.81 m/s/s or 1 *g* in this position at rest, while others may read zero [31]. During the stance phase of running, the tibia undergoes angular and linear motions, with tibial angular motion largely confined to the sagittal plane, rotating about the ankle joint [32]. The TA measured by an accelerometer is the summation of the acceleration due to gravity, angular motion and the linear acceleration resulting from ground impact [33], but depending on the angle of the shank at impact, the measured acceleration contribution due to gravity will vary [32].

Recent improvements have enabled sensors that are small, light and transmit wirelessly, allowing for monitoring outside of the laboratory environment [34,35]. Accelerometers can differ across a range of parameters, which can impact on the quality of the signal. One of the main differences can be the range captured; if the signal range exceeds the capture range of the device, the measured signal will be clipped at the extremities. Some devices capture to on-board memory cards, which often have restrictions to the speed of their read-write capacity. Additionally, wireless transmitting devices can exhibit a variable length signal delay, or complete dropout. While on-board processing of data can in some cases alleviate these problems, this can also result in a reduction in the fidelity of the data. Careful assessment of all of these points is necessary when selecting a device. Where accelerometer specifications are not aligned to the task, subsequent data interpretation may be questionable. It should also be mentioned that researchers and clinicians may have access to accelerometers, that also measure other data such as EMG or gyroscopes, however these units are typically greater mass and therefore less accurate for measuring TA [36].

4.1.1. Uniaxial and triaxial accelerometry

The acceleration of the tibia occurs in three dimensions, often referred to with respect to a local tibial coordinate frame: axial, anterio-posterior and medio-lateral [37]. Lafortune and Hennig [38] measured all three TA components using a triaxial accelerometer, and at 4.7 m/s the TA-AP component exhibited the highest peak values (7.6 g) followed by the TA-A (5.0 g) and TA-ML component (4.5 g). The TA-AP and TA-A components were reduced at 3.5 m/s, while TA-ML components remained constant. The authors concluded that in order to accurately quantify the total acceleration passing through the musculoskeletal system, it is important to measure all three components of acceleration. The existence of high TA-AP components supports the hypothesis proposed by MacLellan [39] who, using high-speed films of the shank, identified a horizontally transmitted shock at heel-strike. Despite these recommendations many researchers have solely reported peak TA-A [37,40–44] (Table 1). When measuring TA using a uniaxial accelerometer, or when there is the intention to extract the components relative to anatomically defined axes, there is a need for careful alignment of the device to the long axis of the tibia [37,40–44]. If the correct alignment is not achieved, the acceleration will not accurately reflect the actual TA-A. Using all axes from a triaxial accelerometer to calculate the TA-R is one method to eliminate the need to specifically align the device to the tibial coordinate frame, thus improving repeatability of the measurement [45].

Only a small number of studies have accounted for additional acceleration components applied to the tibia [45–49] (Table 1), with one research group reporting that cadence influenced the acceleration components independently, where an increase in cadence resulted in lower TA-A and TA-R peaks, but greater TA-AP acceleration [48]. These

Table 1
Summary of literature related to tibial acceleration measurement and analysis.

Reference	Participant Details Age (years) | Height (cm) | Weight (kg)	Accelerometer Placement (Sampling | Filtering Frequencies)	Steps / Time Recorded	Surface | Running Speed	Main Tibial Acceleration Results (g)	
Derrick [14]	10 RFS runners 25.3 ± 6.5 | NR | 68.6 ± 8.0	NR (3600 Hz | NR)	Steps = 10	Overground Speed NR	Normal stride length TA-A: 6.4 Preferred stride length TA-A: 6.2 Preferred stride length TA-A: 8.2	
Mizrahi [18]	14 M volunteer runners 24.2 ± 3.7 | 175.5 ± 5.9 | 73.2 ± 8.3	Uniaxial acc. attached to proximal tibia (1667 Hz | NR)	Time = 20 s	Treadmill Varied (individual max effort)	Minute 1 TA-A: 6.9 ± 2.9 Minute 15 TA-A: 10.5 ± 4.7 Minute 30 TA-A: 11.1 ± 4.2	
Derrick [20]	10 M uninjured university students 27.0 ± 5.0 | 179.0 ± 5.0 | 75.5 ± 12.2	Uniaxial acc. attached to distal tibia (1000 Hz | NR)	Steps = 6	Overground 3.83 m/s ± 5%	Preferred stride length TA-A: 6.1 Stride length +20% TA-A: 11.3 Stride length +10% TA-A: 7.9 Stride length -10% TA-A: 5.9 Stride length -20% TA-A: 5.7	
Hennig [25]	6 M 29 | 181.0 | 76.2	Triaxial acc. attached via bone pins to the proximal tibia (1000 Hz | 60 Hz LP)	Steps: 5	Overground 4.5 m/s	TA-A: 5.32 ± 1.5	
Milner [29]	TFF: 20 F RFS runners 26 ± 9 | NR | NR Control: 20 F RFS uninjured runners 25 ± 9 | NR | NR	Uniaxial acc. attached to distal tibia (960 Hz | NR)	Steps = 5	Overground 3.7 m/s ± 0.2	TFF TA-A: 7.70 ± 3.21 Control TA-A: 5.81 ± 1.66	
Lafortune [32]	1 M recreational runner 32 | 179.0 | 76.0	Triaxial acc. attached via bone pins to the proximal tibia (1000 Hz | 60 Hz LP)	Steps: 10	Overground	TA-A: 2.98 ± 0.19 TA-A: 5.19 ± 0.77	
DeBeliso [37]	10 M uninjured RFS runners 20-30 | NR | 78.9 ± 11.9	Acc. attached to distal tibia (NR | NR)	Steps = 10	Treadmill 2.68 & 3.58 m/s	*2.68 m/s* BF TA-A: 3.9 ± 1.4 BF innersole TA-A: 2.8 ± 1.3 Shoe TA-A: 2.0 ± 0.6 Shoe with innersole TA-A: 1.0 ± 0.6	*3.58 m/s* 5.7 ± 2.4 4.2 ± 2.1 3.2 ± 1.5 2.8 ± 1.1
Lafortune [38]	6 M 29 | 181.0 | 76.2	Triaxial acc. attached via bone pins to the proximal tibia (1000 Hz | NR)	Steps: 5	Overground 4.5 m/s	TA-A - measured: 5.32 ± 1.6 TA-A - estimated: 9.39 ± 1.8	
Crowell [40]	4 M / 6 F RFS recreational runners with TA-A > 8 g 26 ± 2.0 | 172.0 ± 0.07 | 81.5 ± 21.0	Uniaxial acc. attached to distal tibia (1080 Hz | 100 Hz LP)	NR	Overground 3.7 m/s	Baseline TA-A: 8.2 ± 2.5 Post FB TA-A: 4.3 ± 1.5 1 month post FB TA-A: 4.6 ± 1.5	
Crowell [41]	5 F recreational runners 26.0 ± 2.0 | 164.0 ± 0.06 | 59.3 ± 5.4	Uni-axial acc. attached to distal tibia (1080 Hz | 100 Hz LP)	Steps = 20	Treadmill Self-selected (2.4–2.6 m/s)	Baseline TA-A: 9.0 ± 1.6 Post FB TA-A: 7.2 ± 4.9 10 min post FB TA-A: 6.3 ± 3.5	
Dufek [42]	11 F pre-menarche 9.2 ± 1.9 | 139.9 ± 12.5 | 32.9 ± 7.7 11 F normally menstruating 25.2 ± 3.9 | 1.64.3 ± 3.2 | 63.6 ± 9.2 12 F post-menopausal 53.2 ± 4.6 | 163.0 ± 8.2 | 67.2 ± 13.0	Uniaxial acc. attached to distal tibia (1000 Hz | NR)	Time = 45 s	Treadmill Varied speeds	Preferred velocity TA-A: 4.87 ± 1.88 Preferred velocity +10% TA-A: 6.07 ± 2.41 Preferred Velocity TA-A: 4.36 ± 1.32 Preferred velocity +10% TA-A: 4.77 ± 1.50 Preferred Velocity TA-A: 3.56 ± 1.74 Preferred velocity +10% TA-A: 4.05 ± 2.39	
Sheerin [45]	14 M recreational runners 33.6 ± 11.6 | 177.2 ± 6.6 | 75.6 ± 9.5	Triaxial acc. attached to the distal tibia (1000 Hz | 60 Hz LP)	Steps = 61 ± 1.5	Treadmill 2.7 m/s 3.0 m/s 3.3 m/s 3.7 m/s	TA-R: 7.8 ± 2.9–8.6 ± 3.4 TA-R: 9.1 ± 2.7–9.7 ± 3.4 TA-R: 10.4 ± 3.4–11.7 ± 3.8 TA-R: 12.9 ± 4.3–11.9 ± 3.6	
Glauberman [46]	20 F uninjured distance runners 27.8 ± 3.7 | 168.1 ± 6.2 | 59.2 ± 7.3	Triaxial acc. attached to distal tibia (NR | NR)	Time = 60 s	Treadmill 3.13 m/s	Natural RFS (control shoes) TA-R: 11.5 ± 2.8 Natural non-RFS (control shoes) TA-R: 8.7 ± 2.8	TA-A: 6.3 ± 1.1 TA-A: 7.4 ± 0.8 NR

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Table 1 (*continued*)

Reference	Participant Details Age (years) | Height (cm) | Weight (kg)	Accelerometer Placement (Sampling | Filtering Frequencies)	Steps / Time Recorded	Surface | Running Speed	Main Tibial Acceleration Results (g)	
					Natural RFS (baseline) TA-R: 9.3 ± 2.6 Natural RFS (altered strike) TA-R: 11.2 ± 1.6 Non-natural RFS (baseline) TA-R: 13.1 ± 1.2 Non-natural RFS (altered strike) TA-R: 9.5 ± 1.6	NR NR NR
Thompson [47]	5 M | 5 F uninjured RFS runners 26 ± 7.3 | 174.0 ± 9.0 | 65.6 ± 10.2	Bi-axial acc. attached to the distal tibia (1000 Hz | NR)	Steps: 10	Overground 2.25 ± 0.19 m/s	TA-R shod: 11.27 ± 1.73 TA-R BF: 11.32 ± 1.48 TA-R BF RFS: 13.55 ± 1.51	
Giandolini [48]	1 M elite trail runner 26 | 171 | 56.5	Triaxial acc. attached to the proximal tibia (1300 Hz | 50 Hz LP)	Steps: 5530	Overground trail race Variable speed	TA-A: 8.41 ± 3.37 TA-R: 10.4 ± 3.42	
Wood [49]	3 M | 6 F uninjured recreational runners 20 ± 1.5 | 170.2 ± 8.7 | 59.1 ± 8.2	Triaxial acc. attached to the distal tibia (612 Hz | NR)	Steps: 20	Treadmill 3.13 ± 2.5 m/s	Baseline PTA(R): 5.9 ± 0.7 FB1 PTA(R): 5.30 ± 0.80 No FB1 PTA(R): 5.60 ± 1.10 FB2 TA-R: 5.20 ± 0.60 No FB2 TA-R: 5.4 ± 0.70	*Proximal* 4.34 ± 1.29 g 5.03 ± 1.45 g 5.71 ± 1.64 g
Garcia-Perez [51]	11 M | 9 F uninjured recreational runners 34 ± 8 | 172 ± 8 | 63.6 ± 8.0	Uniaxial acc. attached to proximal tibia (100 Hz | NR)	Time = 10 s	Treadmill & overground 3.81 ± 0.40 m/s	*Overground* Pre-fatigue TA-A: 24.6 ± 10.8 Post-fatigue TA-A: 22.2 ± 10.3 *Treadmill* 15.3 ± 6.8 17.2 ± 9.5	
Gruber [52]	12 M | 7 F habitual RFS runners 26.7 ± 6.1 | 175.0 ± 9.0 | 70.1 ± 10.0 14 M | 5 F habitual FFS runners 25.4 ± 6.2 | 176.0 ± 10.0 | 68.8 ± 9.5	Uniaxial acc. attached to distal tibia (1200 Hz | 60 Hz LP)	Steps = 11000	Treadmill 3.47 ± 0.90 m/s 3.73 ± 0.24 m/s	RFS TA-A: 5.07 ± 1.49 FFS TA-A: 3.87 ± 1.36	
OLeary [53]	7 M & 9 F uninjured runners 20-30 | 1.73 ± 0.09 | 68.4 ± 12.0	Uniaxial acc. attached to distal tibia (2000 Hz | 100 Hz LP)	Steps = 5	Overground 3.2 ± 0.3 m/s	Without innersoles TA-A: 4.81 ± 1.45 With innersoles TA-A: 4.05 ± 1.69	
Verbitsky [54]	22 M uninjured runners 30.8 ± 5.1 | 173.9 ± 7.3 | 70.4 ± 9.2	Uniaxial acc. attached to proximal tibia (1667 Hz | NR)	Time = 30 s every 5 min.	Treadmill Varied speeds	*Fatigue group* Baseline TA-A: 6.0 ± 0.5 5 min TA-A: 7.5 ± 0.8 10 min TA-A: 9.0 ± 1.7 15 min TA-A: 9.5 ± 0.9 20 min TA-A: 8.7 ± 1.2 25 min TA-A: 8.8 ± 1.2 30 min TA-A: 9.5 ± 1.0	*Non-fatigue group* Baseline TA-A: 7.0 ± 1.7 5 min TA-A: 6.9 ± 0.8 10 min TA-A: 6.9 ± 1.1 15 min TA-A: 6.3 ± 0.7 20 min TA-A: 7.0 ± 1.2 25 min TA-A: 6.3 ± 1.1 30 min TA-A: 7.0 ± 1.3
Lake [56]	2 M recreational runners NR | NR | NR	Uniaxial acc. attached to distal tibia (2000 Hz | 60 Hz LP)	Steps = 5-10	Overground Self-selected speed (~4.5 m/s)	BF TA-A: 16.34 ± 1.60–17.76 ± 1.74 BF corrected TA-A: 17.87 ± 2.07–20.73 ± 2.56 Shod TA-A: 8.51 ± 1.43–8.67 ± 0.97 Shod corrected TA-A: 10.07 ± 1.29–10.52 ± 1.27	
Creaby [62]	11 M uninjured runners - Clinician guided FB 28.1 ± 7.8 | 178.0 ± 0.05 | 76.5 ± 7.7	Triaxial acc. attached to distal tibia (1500 Hz | 100 Hz LP)	Time = 10 s	Treadmill 3.0 m/s	*Clinical FB* Baseline TA-A: 5.74 ± 2.25 During FB TA-A:4.37 ± 1.86	*Visual acc. FB* 5.34 ± 1.93 3.81 ± 1.36

(*continued on next page*)

Table 1 (*continued*)

<table>
<tr><th>Reference</th><th>Participant Details
Age (years) | Height (cm) | Weight (kg)</th><th>Accelerometer Placement (Sampling | Filtering Frequencies)</th><th>Steps / Time Recorded</th><th>Surface | Running Speed</th><th>Main Tibial Acceleration Results (g)</th><th></th></tr>
<tr><td></td><td>11 M uninjured runners - visual FB
22.7 ± 4.6 | 179.0 ± 0.05 |
78.8 ± 9.2</td><td></td><td></td><td></td><td>Post FB TA-A: 4.13 ± 1.82
7-days post FB TA-A: 4.48 ± 1.53</td><td>4.32 ± 2.20
4.20 ± 1.54</td></tr>
<tr><td>Sinclair [69]</td><td>12 M uninjured runners
23.7 ± 2.3 | 176.5 ± 5.8 |
75.6 ± 7.6</td><td>Triaxial acc. attached to the distal tibia
(1000 Hz | 60 Hz LP)</td><td>NR</td><td>Treadmill
4.0 m/s</td><td>Conventional shoes TA-A: 4.28 ± 2.28
Light shoes with added mass TA-A: 5.47 ± 1.83</td><td></td></tr>
<tr><td>Sinclair [70]</td><td>12 M experienced runners
24.3 ± 1.1 | 178.1 ± 5.2 |
76.8 ± 9.0</td><td>Triaxial acc. attached to the distal tibia
(1000 Hz | 60 Hz LP)</td><td>Steps: 6</td><td>Overground
4.0 m/s</td><td>Conventional shoes TA-A: 6.60 ± 3.65
Barefoot TA-A: 9.17 ± 2.96
Barefoot inspired TA-A: 10.2 ± 3.48</td><td></td></tr>
<tr><td>Sinclair [71]</td><td>12 M uninjured runners
23.5 ± 2.0 | 177.1 ± 4.6 |
77.5 ± 5.5</td><td>Triaxial acc. attached to the distal tibia
(1000 Hz | 60 Hz LP)</td><td>Steps: 5</td><td>Overground
4.0 m/s</td><td>Conventional shoes TA-A: 6.60 ± 2.47
Energy return shoes TA-A: 7.03 ± 2.79</td><td></td></tr>
<tr><td>Sinclair [72]</td><td>12 M uninjured runners
23.1 ± 5.0 | 178.0 ± 0.1 |
77.1 ± 7.9</td><td>Triaxial acc. attached to the distal tibia
(1000 Hz | 60 Hz LP)</td><td>Steps: 5</td><td>Overground
4.0 m/s</td><td>Conventional shoes TA-A: 6.73 ± 1.79
Maximalist shoes TA-A: 7.99 ± 2.32
Minimalist shoes TA-A: 9.54 ± 4.29</td><td></td></tr>
<tr><td>Sinclair [73]</td><td>12 F uninjured recreational runners
21.45 ± 2.98 |166.0 ± 6.0 |
60.87 ± 4.37</td><td>Triaxial acc. attached to distal tibia
(1000 Hz | 60 Hz LP)</td><td>Steps = 5</td><td>Overground
4.00 m/s ± 5%</td><td>Normal shoes TA-A: 10.70 ± 2.31
Cooled shoes TA-A: 12.75 ± 4.62</td><td></td></tr>
<tr><td>Sinclair [74]</td><td>15 M uninjured runners
21.02 ± 2.02 |176.6 ± 5.3 |
76.82 ± 6.27</td><td>Uniaxial acc. attached to distal tibia
(1000 Hz | 60 Hz LP)</td><td>Steps = 5</td><td>Overground
4.00 m/s ± 5%</td><td>Conventional shoes TA-A: 5.25 ± 1.43
Energy return shoes TA-A: 5.90 ± 1.58</td><td></td></tr>
<tr><td>Sinclair [77]</td><td>13 M uninjured runners
27.81 ± 7.02 | 177 ± 11 |
76.22 ± 7.04</td><td>Triaxial acc. attached to distal tibia
(1000 Hz | 60 Hz LP)</td><td>Steps = 5</td><td>Overground
4.0 m/s ± 5%</td><td>BF TA-A: 5.72 ± 1.34
Cross-fit TA-A: 5.17 ± 1.82
Conventional TA-A: 4.55 ± 1.29
Minimalist TA-A: 5.31 ± 1.55</td><td></td></tr>
<tr><td>Laughton [78]</td><td>15 RFS runners
22.46 ± 4.0 |169.75 ± 6.07 |
66.41 ± 8.58</td><td>Uniaxial acc. attached to distal tibia
(960 Hz | 100 Hz LP)</td><td>Steps = 5</td><td>Overground
3.7 m/s ± 5%</td><td>No Orthotics TA-A: 7.18 ± 2.98
Orthotics TA-A: 6.78 ± 3.14
RFS TA-A: 7.82 ± 3.16
FFS TA-A: 6.15 ± 2.96</td><td></td></tr>
</table>

M – male; F – female; acc – accelerometer; TA-A - peak axial tibial acceleration; TA-R – peak resultant tibial acceleration; TFF – tibial fatigue fracture; RFS - rearfoot strike; FFS - forefoot strike; BF – barefoot; FB – feedback; NR - not reported; LP - low pass.

data were captured from a single subject, running over a highly variable terrain. Thompson et al. [47] reported TA-R calculated from two movement planes only (TA-A and TA-AP). The lack of a third axis, and therefore a true resultant vector, means that data could still be lost through axis misalignment. The resultant TA takes into account all three axes, therefore the magnitudes will always be larger than the TA-A on its own. Some runners will have a dominance of the axial component, in which case the magnitudes of TA-A and TA-R may be similar, however this is not always the case, and these variables are not interchangeable. Following the initial recommendations of LaFortune et al. [38], Glauberman et al. [46] report no differences in TA-A between rearfoot and non-rearfoot strike runners, however TA-R were reported to be greater in non-rearfoot runners. While they did not report the individual components, the additional acceleration present in the resultant signal could only have come from components other than TA-A.

4.1.2. Sampling frequency

Nyquist theory dictates that the minimum sampling frequency should be twice the highest frequency present in a signal [30]. The measurement of human motion adds signal noise, therefore an even higher sampling frequency (5–10 times the highest frequency) is required to obtain an adequate reconstruction [30]. Power spectral analyses have revealed that 99% of the TA signal power captured during running was below 60 Hz [25,32,50]. Based on the conventions previously outlined, this would dictate a capture sampling frequency between 300–600 Hz. While most researchers report a sampling rate of at least 1000 Hz [14,18,25,38,40], some have sampled as low as 100 Hz [51], calling their results into question (Table 1).

4.2. Accelerometer placement on the tibia

The distal tibia is a common location of fatigue fractures in runners, making it an important site for the measurement of acceleration [29,52,53], but many researchers have also measured TA from the proximal tibia [51,54,55] (Table 1). These differing placements may not give comparable results. Running at 4.5 m/s the tibia angle at impact can vary by up to 20° from vertical [32,56]. The linear acceleration of the tibia is influenced by centripetal acceleration due to the sagittal plane angular motion, which acts in the opposite direction to TA-A [56]. The angular acceleration is dependent on the tibial angular velocity and the distance of the device from the axis (i.e. the ankle) [32]. Both measured and modeled estimates have indicated that the TA recorded on a device attached closer to the knee substantially underestimates the TA-A at the distal attachment [32,57]. Taking into account the contributions of gravity and the angular component of TA, Lake et al. [56] reported that the measured TA (at 4.5 m/s) needed to increase by 1.5–3 g depending on the subject and shod condition. Additionally, the correction for angular motion influenced the TA power spectrum, with a gain in signal power particularly prevalent in the 8–13 Hz frequency band. Despite these findings, most researchers don't examine the frequency components, and often simultaneous kinematics are not captured to allow for a correction for gravity and angular motions of the lower extremity [32,56].

4.3. Accelerometer attachment

To determine the best estimate of the acceleration of a segment of interest, an accelerometer attached directly to the bone is most accurate, however this is impractical for regular use [25,32,38,58] (Table 1). LaFortune et al. [58] compared the TA-A measured from bone and skin mounted accelerometers while runners ran overground. For some subjects the skin-mounted accelerometer overestimated TA-A by as much as twice the bone-mounted devices. While the dominant component of these peaks represented the impact, the signal also included acceleration components due to muscular action, and noise due to resonance in the compliant attachment of the accelerometer [50,58]. The absolute differences between the signals was large, but with a low-pass filter, signals from a skin-mounted device adequately represented bone accelerations [58]. There will always be oscillation of skin-mounted accelerometers, therefore it is important to know the characteristics of this oscillation. If the resonance frequency of the accelerometer and mounting system occurs at the same frequencies of those from ground impact (10-20 Hz) the measured acceleration will be elevated [50]. Ziegert and Lewis [59] studied the effect of soft tissue, by comparing the output of a surface-mounted accelerometer with that of a device connected to the tibia bone via a needle. When the leg was impacted with a device, a 1.5-gram surface-mounted accelerometer showed almost identical outputs to the bone, but a 34-gram accelerometer gave outputs with little resemblance to the bone acceleration, appearing to oscillate at its resonant frequency on the soft tissue. Three studies have reported the natural resonant frequency of the accelerometer as 250 Hz [25], 400 Hz [38]and 1000 Hz [58]. Henning, et al. [25] reported mathematically and experimentally deriving this frequency, however no methods outline or appropriate reference was provided.

Accelerometer oscillation can be minimised by tensioning the device attachments [60], with Clarke et al. [61] reporting that a preload force 'as tight as tolerable' improved reliability, both within and between sessions. Forner-Cordero et al. [36] conducted a series of experiments to determine the frequency characteristics of skin-mounted devices under varied attachment conditions, including using elastic bands, a method commonly used in recent research [42,62–64] (Table 1). They also outlined a test to validate the attachment integrity before recording clinical measurements, which involved subjects standing on their tiptoes, and falling freely onto their heels. While this test is unlikely to produce TA magnitudes representative of running, it did show low variability, and could discriminate between different attachment conditions [36]. Once again, without adequate preload force, the frequency of the accelerometer-mounting system was too low, close to the frequency range of the data, increasing measurement error. While there is still no clarity on what constitutes tensioning 'as much as tolerable', and acknowledgement that this will differ for individuals, a simple test, such as the 'heel drop', could be an effective method to compute the frequency of the accelerometer mounting to allow confirmation of the integrity of attachment before testing begins.

5. Tibial acceleration analysis

5.1. Normalisation

To account for variability in absolute magnitudes between sessions, normalisation of TA data has been proposed [65]. Expressing TA-A relative to the mean observed at the slowest running velocity, provided a 'shock ratio', which can be useful considering the absolute values of the peak accelerations are susceptible to noise and vibration. Focusing on the relative magnitudes of acceleration measures can be informative for many applications (e.g. cushioning properties of running shoes), however to be of use in the comparison of datasets, multiple, and consistent running velocities would be required.

5.2. Frequency content of acceleration

While time domain TA components are most commonly reported, the signal is formed by acceleration components of various frequencies, which are superimposed in the time domain signal [50]. The low frequency component (4–8 Hz) is the acceleration associated with voluntary leg motion, while the high frequency component (10–20 Hz) represents the rapid deceleration of the lower extremity at contact [50,52]. These low and high frequency ranges are also representative of the active and impact peaks of the vertical GRF, respectively [50,66]. The resonant frequency of the mounting system also contributes to the time domain signal.

Table 2
Summary of literature related to intrinsic and extrinsic factors that can modify tibial accelerations.

Reference	Participant Details Age (years) | Height (cm) | Weight (kg)	Accelerometer Placement (Sampling | Filtering Frequencies)	Steps / Time Recorded	Surface | Running Speed	Main Tibial Acceleration Results (g)
Greenhalgh [81]	9 M hockey players 21.0 ± 1.69 | 175.75 ± 6.56 | 78.13 ± 12.11	Triaxial acc. attached to distal tibia (1000 Hz | 60 Hz LP)	Steps = 6	Varied surfaces 3.3 m/s ± 5% 5.0 m/s ± 5%	3.3 m/s Synthetic surface TA-A: 4.2 ± 1.2 – 5.0 ± 1.2 3.3 m/s Concrete TA-A: 4.8 ± 1.8–5.5 ± 1.8 5.0 m/s Synthetic surface TA-A: 7.4 ± 2.6–9.1 ± 2.7 5.0 m/s Concrete TA-A: 10.5 ± 2.4–8.4 ± 2.7
Boey [82]	18 M & 20 F uninjured runners Untrained: 22.3 ± 1.8 | 173.2 ± 8.9 | 65.0 ± 9.1 Rec.: 22.3 ± 2.5 | 176.3 ± 9.2 | 65.8 ± 8.1 Trained: 25.4 ± 5.0 | 178.0 ± 7.9 | 63.3 ± 5.1	Triaxial acc. attached to distal tibia (1024 Hz | 60 Hz LP)	Steps = 15	Concrete - 3.1 m/s Concrete - SS Track - 3.1 m/s Track - SS Trail - 3.1 m/s Trail – SS	*Untrained* *Recreational* *Well-trained* TA-A: 9.39 ± 2.98 9.11 ± 2.33 10.55 ± 2.20 9.83 ± 2.67 10.62 ± 3.21 TA-A: 9.31 ± 2.96 8.99 ± 2.22 10.99 ± 2.98 9.91 ± 2.75 10.27 ± 2.95 TA-A: 8.93 ± 2.62 8.38 ± 1.98 10.38 ± 1.83 9.83 ± 2.42 10.17 ± 2.77 TA-A: 10.88 ± 2.79 TA-A: 9.55 ± 1.53 TA-A: 9.84 ± 2.21
Montgomery [83]	15 recreational runners NR | NR | NR	Triaxial acc. attached to mid-tibia (1500 Hz | 60 Hz LP)	Steps = 8	Varied surfaces 2.88 ± 0.35 m/s 4.25 ± 0.37 m/s	*2.88 m/s* *4.25 m/s* Overground: 5.1 8.0 Motorised treadmill: 5.4 8.1 Non-motorised treadmill: 3.7 6.2
Sinclair [84]	10 M RFS recreational runners 20.42 ± 3.55 | 178.75 ± 5.81 | 76.58 ± 6.52	Triaxial acc. attached to distal tibia (1024 Hz | 60 Hz LP)	Steps = 10	Overground 3.5 m/s ± 5% 5.0 m/s ± 5%	*3.5 m/s* *5.0 m/s* BF TA-A: 6.85 ± 3.51 12.81 ± 5.74 BF inspired shoes TA-A: 5.54 ± 1.31 7.92 ± 4.30 Conventional shoes TA-A: 2.28 ± 0.64 4.54 ± 1.14
Giandolini [103]	12 M & 8 F uninjured RFS recreational runners 19.7 ± 1.3 | 177 ± 79 | 70.7 ± 9.0	Uniaxial acc. attached to distal tibia (1000 Hz | 50 Hz LP)	Time = 10 s	Treadmill Self-selected	*MFS training* *Shoe training* Baseline TA-A: 6.80 ± 1.55 5.60 ± 1.04 1-month TA-A: 6.57 ± 2.12 5.73 ± 1.53 2-month TA-A: 7.47 ± 1.71 6.18 ± 1.90 3-month TA-A: 6.70 ± 1.46 6.67 ± 1.48
Fu [107]	13 M uninjured recreational RFS runners 23.7 ± 1.2 | 173.7 ± 5.7 | 65.7 ± 5.2	Biaxial acc. attached to the proximal tibia (1000 Hz | 100 Hz LP)	Steps = 10	Varied surfaces 3.33 ± 0.17 m/s	Concrete TA-A: 2.4 ± 3.1 Synthetic track TA-A: 10.9 ± 3.5 Grass TA-A: 11.1 ± 3.4 Treadmill TA-A: 11.6 ± 3.0 Treadmill EVA TA-A: 10.3 ± 3.1
McNair [112]	10 M RFS runners 75 ± 6 | NR | NR	Uniaxial acc. attached to distal tibia (1000 Hz | NR)	Steps = 8	Treadmill 3.5 m/s	*Shoe conditions* Double density EVA with a cantilever outsole: 10.0 Double density EVA: 10.3 Air filed chambers within double density EVA: 10.0 Encapsulated double density EVA: 9.8 BF: 14.0
Chambon [115]	15 M uninjured recreational runners 23.9 ± 3.2 | 177.0 ± 3.0 | 73.0 ± 8	Triaxial acc. attached to the middle medial tibia (2000 Hz | 50 Hz LP)	Steps = 5	Overground (flat and 10° incline) 3.3 m/s ± 5%	*Flat: New* *Inclined: New* Viscous TA-A: 5.37 ± 1.86 5.22 ± 1.55 Intermediate TA-A: 4.39 ± 1.83 4.22 ± 1.17 3.73 ± 1.17 Elastic TA-A: 4.09 ± 0.87 *Inclined: Fatigued* *Flat: Fatigued* 6.11 ± 1.77 Viscous TA-A: 5.56 ± 1.76 4.29 ± 1.36 Intermediate TA-A: 5.35 ± 1.77 4.68 ± 1.19 Elastic TA-A: 5.03 ± 1.37
Clark [116]	36 F injury free runners (> 30 min per run; > 3x weekly	Triaxial acc. attached to proximal tibia (2000 Hz | 60 Hz LP)	Steps = 16	Treadmill 2.8 m/s	*Day 1* *Day 14* No pill TA-A: 4.17 ± 1.96 4.24 ± 2.02 No pill TA-AP: 1.89 ± 0.40 1.92 ± 0.37 4.67 ± 2.46 Contraceptive pill: TA-A: 1.78 ± 0.37 4.99 ± 2.02 Contraceptive pill: TA-AP: 1.79 ± 0.35
Clansey [117]	Uninjured recreational runners (RFS) with elevated TA-A (> 9 g) 12 M (intervention) 33.3 ± 9.0 | 180.0 ± 0.1 | 77.2 ± 11 12 M (control) 33.9 ± 11.3 | 180.0 ± 0.1 | 75.1 ± 6.9	Triaxial acc. attached to distal tibia (1500 Hz | 60 Hz LP)	NR	Treadmill 3.7 m/s (1% gradient)	*Intervention* *Control* Pre-intervention TA-A: 10.67 ± 1.85 9.78 ± 1.68 Post-intervention TA-A: 7.39 ± 1.48 9.99 ± 1.97 1 month post-intervention TA-A: 9.68 ± 1.87 8.30 ± 1.82
Butler [118]	12 uninjured high arch runners 20.9 ± 3.0 |	Uniaxial acc. attached to distal tibia (1080 Hz | NR)	NR	Treadmill | Self-selected training pace	*Beginning of run* *End of run* High arch cushion shoes TA-A: 5.9 ± 0.9 5.5 ± 0.7 4.6 ± 0.6

(*continued on next page*)

Table 2 (*continued*)

Reference	Participant Details Age (years) | Height (cm) | Weight (kg)	Accelerometer Placement (Sampling | Filtering Frequencies)	Steps / Time Recorded	Surface | Running Speed	Main Tibial Acceleration Results (g)	
	170.0 ± 0.07 | 68.36 ± 5.75 12 uninjured low arch runners 21.8 ± 3.2 | 173.0 ± 0.11 | 70.04 ± 7.35				High arch motion control shoes TA-A: 4.5 ± 0.4 Low arch cushion shoes TA-A: 4.6 ± 1.4 Low arch motion control shoes TA-A: 5.7 ± 1.7	NR NR

M – male; F – female; acc – accelerometer; TA-A - peak axial tibial acceleration; TA-R – peak resultant tibial acceleration; RFS - rearfoot strike; FFS - forefoot strike; BF – barefoot; FB – feedback; NR - not reported; LP - low pass.

It is possible to separate the frequency components using a frequency analysis [50,67]. A fast fourier transform provides the median power frequency of the acceleration signal, or alternatively a joint time-frequency distribution analysis can provide the instantaneous power spectrum [67]. Variations or changes in peak TA observed in the time domain may be a result of changes in low or high frequency bands, or changes in the resonant frequency of the mounting system [67]. These additional signal analysis approaches have been used to provide a more thorough characterisation of the signal components in a range of running studies [15,68–70].

5.3. Signal filtering

All kinematic data contains a true signal representing human movement, as well as noise, therefore some pre-analysis filtering is required [30]. While both the true signal and noise occupy a wide bandwidth, noise is usually at the higher end of the frequency spectrum. If the cut-off is set too low, the resulting signal will be incorrect, whereas if the cut-offs are too high, too much noise will remain in the signal [30]. Most studies measuring TA magnitude in the time domain during running report using low-pass filters with cut-offs between 40 Hz and 100 Hz [25,32,40,51,58,62,63,71], which were in some cases determined via power spectral analyses of the signal [32,58]. Selecting the appropriate filter cut-off frequencies is essential, as over or under filtering data can lead to inaccurate interpretations. A TA signal also contains low frequency components (4–8 Hz) associated with voluntary leg motion, and the acceleration of the body COM, therefore it is possible to supplement the low-pass filtering with a high-pass (e.g. 10 Hz), or use band-pass filter (e.g. 10–60 Hz) to exclusively reveal the frequency component related to the passive impact of running gait. These filtering methods do not appear to be widely used [50,52] (Table 1).

6. Outcome measures

Where triaxial devices are used, TA signals can be resolved into three acceleration components. The coordinate system axes can be defined differently, but commonly the orthogonal axes are defined with respect to the tibia: TA-A, TA-AP and TA-ML. The TA-A corresponds to a line bisecting the proximal and distal ends of the tibia in both the frontal and sagittal planes. The medio-lateral axis runs perpendicular to the axial axis and parallel to a line joining the two malleoli, and the antero-posterior axis is mutually orthogonal to both the longitudinal and medio-lateral axes [38]. A number of additional variables can be calculated from the measured signals. The most commonly reported are peak TA-A magnitude, followed by peak TA-R [38,45–49]. A smaller number of studies have also reported peak positive TA-ML [38] and TA-AP [38,48], as well as time to peak positive [25,32,38,58,72–75], TA-A slope [73,74,76,77], TA-A loading rate [72], duration of peak positive [38], peak negative [38,46], duration of negative acceleration [38], and peak positive to peak negative acceleration [78,79] (Table 1). It should be noted, TA-A magnitude is currently the only parameter linked to running injury [29].

Despite the widespread use, publications describing the acceptable reliability of accelerometers attached to the tibia of runners is limited [61,80]. Clarke et al. [61] collected TA-A data from three subjects running on a treadmill at 3.8 m/s during five separate sessions. The mean within-session step-to-step variability was 6.8%, and the between-session variability was 5.6%. With the between-session variability falling inside the step-to-step variability, it was deemed that accurate comparisons could be made between sessions. Sheerin et al. [80] report the one-week reliability from 20 runners at a range of velocities (2.7–3.7 m/s) on a treadmill. While the TA-A results were acceptable at all velocities, they were generally larger for TA-A compared to TA-R for both the percentage difference in the means (TA-A 0%–5.7%; TA-R 0.9%–5.1%) and the effect sizes (TA-A 0.01-0.17; TA-R 0.01-0.12), indicating slightly better session-to-session TA-R reliability.

7. Intrinsic factors that can modify tibial accelerations

7.1. Running velocity

The seminal work analysing the effect of running velocity report consistently increased peak TA magnitude with faster running velocities (3.5 and 4.7 m/s) across all components of TA (TA-A, TA-AP and TA-ML) from a single recreational runner, using a bone-mounted accelerometer [58]. This increase in TA-A was also reported at a series of faster running velocities (spanning 3.4 to 5.4 m/s) from 10 well-trained runners [65]. Further to this, linear regression analysis revealed that average TA-A increased by 34% for each 1.0 m/s increase in running velocity. Individual linear relationships varied between 0.15 and 0.68, and while the best-fit linear relationship was described as in 'good agreement' with the experimental data, no supporting statistics were provided [65].

All subsequent studies reporting TA-A while running velocity was manipulated as an independent variable confirm that running at faster velocities was associated with increased TA-A, irrespective of running surface, footwear, running experience, or whether the velocity was fixed, or self-selected [42,45,81–85] (Tables 1 and 2). While the focus of two of these studies were on shock attenuation between the tibia and the head, the results provided insight into the characteristics of TA-A change with increasing velocity [42,85]. Investigating the characteristics of shock attenuation across a range of running velocities up to a runners' maximum, Mercer et al. [85] report that the average TA-A remained constant for both 50% and 60% of maximal velocity, but increased at faster velocities. The TA-A variability (SD) remained relatively constant for the first four velocities, before increasing at 90% and 100% of maximal velocity. In a subsequent study, a mixed model design was used to examine the impact of attenuation characteristics of different groups female runners (pre-pubescent girls, normally menstruating women and postmenopausal women) [42]. Participants ran on a treadmill at their preferred velocity (1.9 to 2.6 m/s) and at a velocity 10% faster, while TA-A values ranging from 3.6 to 6.1 g were recorded. The authors claimed that the results demonstrated the anticipated response for velocity, with all groups exhibiting greater peak TA-A during faster running. However, with deeper analysis, it is evident that the TA-A measured from the prepubescent girls were larger than those measured from the normally menstruating women, despite running slower. While speculative, this could be as a result of the younger girls having a reduced tibia mass, and therefore reduced effective mass [33]. These studies were limited by small samples sizes, and the fact that comparisons were made against the percentage of their individual maximum [85], or comfortable running velocity [42], rather than an absolute velocity.

There is still an absence of normative TA values for runners at a range of running velocities. Sheerin et al. [86] measured TA-R for 82 runners running at four different treadmill velocities (2.7–3.7 m/s) and report mean values ranging from 9.8 + 2.7 g at the slowest velocity to 12.1 + 3.1 at the faster velocity. Values from individual runners were spread with 4.5 g the lowest recorded at 2.7 m/s, and 20.6 g the highest recorded at 3.7 m/s. A moderate positive correlation ($r = 0.42$) was reported between velocity and TA-R, and a regression analysis that revealed that for every 1 m/s increase in velocity TA-R would increase by 3.7 g.

7.2. Stride rate and stride length

In the first three of six studies to assess the influence of stride rate and stride length on TA magnitude, stride rate was manipulated to 5% and 10% slower, and 5% and 10% faster, than subjects' preferred, while controlling velocity at 3.8 m/s [61]. Runners adapted to a stride rate 10% and 20% slower, and a 10% and 20% faster than their preferred, while running at their preferred velocity [87], and finally under the same stride rate conditions at 3.8 m/s [20]. Peak TA-A showed a positive linear trend with increased stride length across all three studies [20]. This increase is likely due to a simultaneous decrease in effective mass, which has been closely linked to knee angle (and therefore stride length) at impact [33,88].

Independently manipulating stride length and rate at different velocities has further expanded the understanding of the relationship of TA-A with these fundamental variables. Running with a longer than preferred stride length, leads to increased TA power spectral density [85,89], which were four times greater when stride length, as opposed to stride rate, was varied [89]. When TA-A was compared between preferred stride length and a stride length constrained to 2.5 m at various velocities [90], magnitudes increased by approximately 24% per 1 m/s increase in running velocity. This is lower than the 42% [91] and 34% [65] increases previously reported, however when stride length was constrained, there was no clear relationship between TA-A and running velocity [90]. These results support the notion that kinematic factors, such as the particular orientation of the hip, knee and ankle joints for a given stride length, might be critical in determining TA magnitude.

7.3. Fatigue

While a complex phenomenon, exercise induced fatigue is an important factor in the development of fatigue fractures [92]. Increases in TA-A towards the end of high intensity treadmill running bouts designed to induce central fatigue (related to a failure in neural drive), have been reported, in some cases by as much as 100% [18,54,93–96]. Derrick et al. [93] suggested that increases in knee flexion angle and foot inversion at contact may be responsible for the increased TA-A, and that these adaptations decrease the effective mass of the system, therefore increasing TA-A. Citing a spring-damper model simulating human running vertical GRFs [97], it is reasoned that increased TA-A should not necessarily be linked to an increased injury potential, suggesting that decreasing the effective mass will increase the TA-A, while at the same time decreasing the impact forces [93]. These conclusions are contradictory to the evidence linking increased TA-A magnitude with tibial fatigue fracture development in runners [27,29]. These views do highlight that the evidence is not clear and that researchers disagree on this topic. Contrasting the evidence that TA increases with central fatigue [18,54,93–96], Abt et al. [64] report no changes in any kinematic or acceleration variables after the exhaustive treadmill run. Unclear findings were also reported in a subsequent study where fatigue effects on TA-A were compared when runners ran both overground and on a treadmill [51]. On average, TA-A increased during the treadmill run, but this was not replicated with overground running. Additional kinematic variables were not captured, and therefore the characteristics of the adaptations could not be analysed further.

To which extent local muscle fatigue effects TA in running has not been demonstrated [98]. Several studies have used a human pendulum approach to control kinematic variables such as joint position and impact velocity, while reproducing impact parameters which closely resemble those of normal running [55,98,99]. In contrast to experiments on central fatigue, across a range of different protocols, localised muscle fatigue was found to cause a decrease in TA-A magnitude and slope at impact [55,98,99]. It is thought that these changes are a result of the reduction in the force generating capacity of the muscle due to fatigue. The implications of these findings are not likely to be fully appreciated until more extensive evaluations of the roles of individual muscles on segment and joint stiffness, and how this translates to the actual running environment [55]. Overall, there have been inconsistencies in the fatigue protocols, the varying levels of runners used, and a lack of understanding of the implications of effective mass during ground impact in running. These factors have meant that the effect of both central and localised muscle fatigue on TA is inconclusive.

7.4. Joint kinematics

Lower extremity joint positions at contact are closely connected to stiffness and effective mass, and therefore their position or alignment at initial contact may effect TA magnitude. Denoth [100] and McMahon et al. [101] demonstrated that greater knee flexion angle resulted in smaller effective mass and a reduction in stiffness, leading to greater shock absorption. This concept has been supported with the 'two-mass' running model, and its association with vertical GRF–time waveform patterns [102].

With the association between a foot strike pattern and the absence or reduction in GRF vertical impact peak, it was hypothesised that landing with a strike pattern further forward on the foot (e.g. forefoot or midfoot) would reduce peak TA [46,78,79,103]. However, when viewed in relation to footstrike mechanics, the findings are conflicting. Where runners transitioned from a rearfoot strike to either a midfoot or forefoot strike pattern, increases in TA-R [46], and in signal power in the 9–20 Hz frequency range [52], were reported. However, either no change [103], an increase [52,78], or a decrease [79] in TA-A variables were also found (Tables 1 and 2). Additionally, when non-rearfoot strike runners transitioned to a rearfoot strike pattern they demonstrated a decrease in TA-R [46]. A number of factors could contribute to these conflicting findings, specifically the varied definitions of running kinematics (e.g. forefoot [52] versus non-rearfoot strike [46]), different baseline characteristics [46,52], and the differing intervention durations for retraining habitual patterns [78,103].

To enhance the understanding of the effects of running kinematics on TA, it makes sense to also consider a greater number of segments. While they can't be determined clinically, lower extremity stiffness and effective mass can also have a meaningful impact on TA. Analysing the discrete kinematic parameters associated with the passive attenuation of both time and frequency domain characteristics, knee flexion velocity at foot-strike was found to be the single regulator of time domain peak TA [104]. While a large proportion of variance and associated mechanisms remain unexplained, this provides some evidence that kinematic parameters can influence TA magnitude during running. When kinematics and stiffness parameters were monitored alongside alterations in decline surface gradient, runners could be classified by their shock attenuation [105]. While all runners demonstrated increased accelerations at the tibia and head with increased decline gradients, Runners with reduced shock attenuation (i.e. relatively higher head accelerations) also demonstrated differences in lower extremity and trunk kinematics at both heel-strike and mid-stance. Specifically, these runners exhibited higher COM displacement, heel-strike velocity, and reduced COM stiffness and damping.

Further evidence supporting an influential relationship between running technique and TA was seen where runners were able to actively modify their kinematics to reduce TA-A, by as much as 50%, after a single session of real-time visual feedback [41,62], or 10% reductions in TA-R in response to real-time audio feedback [49]. Similar chages were noted four weeks post intervention, when runners were screened for high pre-intervention TA-A values and exposed to a more extensive feedback schedule [40,63]. Reductions in TA-A were accompanied by lower instantaneous vertical force loading, as well as increased ankle plantar flexion and decreased heel vertical velocity at initial contact, and changes from a rearfoot strike to a midfoot strike pattern [40,63].

8. Extrinsic factors that can modify tibial accelerations

8.1. Running surface

Owing to their cushioning properties, treadmills typically have a lower compliance compared to tarseal or concrete running surfaces. There is evidence to suggest that TA-A measured overground can be substantially higher than running on some treadmills under comparable conditions [51,83,106], however the relationship between TA-A magnitude and surface compliance is not straightforward. Fu et al. [107] found no differences in TA-A across a wide range of surfaces running at 3.3 m/s, whereas Greenhalgh et al. [81] reported higher magnitudes when participants ran at 5 m/s on concrete compared to a synthetic surface, but again not at a slower velocity (3.3 m/s) (Tables 1 and 2). Conversely, Boey et al. [82] reported lower TA-A when runners ran on a more compliant woodchip trail (compared to concrete or synthetic track), but only when restricted to a slower velocity (3.1 m/s), in comparison to the runners' self-selected pace (average 3.7 m/s).

Experiments using non-running external impacts have suggested that the surface compliance explained less than 10% of the variance of the TA-A, with knee angle and muscle pre-activation explained 25%–29% and 35%–48%, respectively [108]. What is clear is that runners rapidly adjust leg stiffness when on different surfaces. By sensing the changes in surface compliance, runners adapt muscle activations and kinematics within a single stride [109]. For surfaces of higher compliance, leg stiffness increases, which serves to keep the path of a runner's COM the same regardless of the surface characteristics [110]. While it has not been examined, it may be that the pre-activation of muscles, and subsequent changes in leg stiffness, is the mechanism runners use to mitigate the effects of surface compliance on lower extremity acceleration.

Negative correlations have been observed between surface gradient, TA-A, TA-ML and TA-R, as well and median frequency [48,105,111]. Hamill et al. [111] reported 30% increases in TA-A on a 8.7% decline gradient, compared to level. Similar, but slightly smaller increases in TA-A magnitude were found by Chu et al. [105], these were accompanied by 51% increases in impact-related frequencies (i.e. power spectral densities within the 12–20 Hz bandwidth). These findings are in contrast to Mizrahi et al. [96] who observed similar magnitude TA-A, and a lower amplitude within the impact frequency range, from runners running on a 7% decline gradient compared to running on the flat.

8.2. Running footwear

Conventional running footwear has been characterised by an ethylene-vinyl acetate (EVA) midsole of approximately 20 mm thickness. Initial research reported substantial reductions in the high frequency components of TA while running in footwear with a midsole, over barefoot conditions. The power spectral density of frequency components above 20 Hz were directly related to shoe midsole hardness [68]. Subsequent studies have shown, that despite some shoes demonstrating significantly reduced cushioning properties when mechanically drop tested [112], no difference in peak TA across various conventional thickness EVA footwear conditions were found [81,112] (Table 2). Tibial acceleration measured in conventional shoes has also been compared to measurements taken running barefoot [70,77], in barefoot-inspired [72,77], and minimalist shoes [69,70,77,103]. In all cases running barefoot produced higher TA magnitudes than in conventional footwear [69,70,77,84]. Additionally, TA magnitude was lower in conventional shoes, compared to the barefoot, barefoot inspired [77,84] or minimalist [69,70,77] footwear conditions.

Recent developments in running footwear have resulted in oversized lower density midsoles (maximalist shoes), expanded thermoplastic polyurethane midsole, and orthotic inserts claiming to provide additional cushioning and reduced energy loss [72,113]. Findings have indicated that TA-A were actually greater in footwear designed to improve energy return [74]. Additionally, running in a maximalist shoe [72], custom [75,78,79], or over the counter [114] orthotics did not provide further reductions in TA-A than conventional shoes. These findings are less surprising when considered in context of the effects of surface characteristics on stiffness, where runners have been shown to increase their leg stiffness when running on softer surfaces [109].

9. Conclusions and recommendations

Clinicians and researchers commonly use tibial acceleration during running as a proxy measurement for the impact forces experienced at the tibia. There is an assumption that this measure corresponds to the acceleration of the bone, and ultimately bone stress and strain, however this is yet to be proven. For users of tibial mounted accelerometers, there are several recommendations that should be adhered to in order to achieve accurate and reproducible results. Devices should be secured firmly to the tibia to limit movement relative to the underlying bone. Differing placements of accelerometers do not necessarily give comparable results; distally attached devices provide higher values, which likely closer represent the accelerations passing through the bone. While the time domain axial tibial acceleration is the only component shown to have construct validity with respect to injury, it is important to quantify the total acceleration passing through the musculo-skeletal system. Where devices of minimal mass can be sourced, triaxial accelerometers should be used to measure all three components of acceleration. Calculating the resultant acceleration can provide a single metric that takes into account all axes, which is independent of accelerometer alignment. Selecting the appropriate filter frequencies are essential, as incorrect filtering can lead to inaccurate interpretation of data. Additional frequency analyses could be useful to provide a more thorough characterisation of the signal.

Tibial acceleration is clearly influenced by running velocity, whereby faster running velocity leads to increased peak tibial acceleration. The extent of tibial acceleration increases are likely dictated by the associated changes to stride rate and stride length. Where substantial stride length increases occur, changes may also occur in knee flexion angle and velocity, heel-strike velocity or subsequent lower extremity stiffness, which are important determinants of impact characteristics. Surface and footwear compliance also have a substantial influence on lower extremity stiffness and tibial acceleration. Runners rapidly adjust to surface compliance, and conditions that are too hard or too soft appear to result in technique modifications and increases in tibial acceleration. There are still considerable gaps in current knowledge, and the interrelationships between muscle pre-activation and fatigue, stiffness, effective mass and tibial acceleration still require further investigation, as well as how changes in these variables impact on injury risk.

Conflict of interest

Dr. Besier is a consultant for IMeasureU-Vicon and is involved in the development of inertial sensor solutions.

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