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RUNNING HEADING: Tibial acceleration in runners. A Review

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ABSTRACT

Impact loading in runners, assessed by the measurement of tibial acceleration, has attracted substantial research attention. [Due to links with injury, tibial acceleration is also used as a clinical monitoring metric, however there are contributing factors and potential limitations that must be considered before widespread implementation.](#) 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. [Tibial acceleration is a good proxy measure of the impact force experienced by the tibia that can be used by clinicians and researchers.](#) The device selected should be triaxial and securely affixed to the tibia to prevent excessive movement. The frequency domain of signals should be analysed in post-processing to select an appropriate filter to obtain valid data. Tibial acceleration in runners has been successfully used to assess the effect of impact attenuation, varied running technique, running speed, surface compliance, leg stiffness footwear and fatigue. It has also been used in gait retraining feedback studies aimed at reducing injury risk and

improving running performance. The inter-relationships between tibial acceleration, running technique, running speed, surface compliance, leg stiffness and fatigue are still not well understood and require further investigation.

KEY POINTS

Tibial acceleration is a good proxy measure of the impact force experienced by the tibia in runners. Clinicians and researchers should pay particular attention to select a device that has appropriate specifications (sampling rate / number of axes / mass), and ensure that it is securely attached to the tibia with minimal tissue overlaying the bone.

As tibial accelerations are known to be affected by running technique, velocity, surface and a number of other factors, users should be careful to control these variables to ensure that any identified differences (or changes) in tibial accelerations are a result of a true difference (or change) in performance, rather than a consequence of changes in other related variables.

Despite the large number of studies that have measured tibial acceleration caution is needed with the interpretation of findings, as a number of factors including running technique, velocity, surface type and fatigue are still not well understood.

1. Introduction

Running is a popular activity with easy accessibility and relatively low cost, but the high participation in running is accompanied by a high incidence of running injuries {Taunton:2003vg, Taunton:2002ey}. The majority (between 50-75%) of lower limb running-related injuries are chronic in nature and are related to cumulative loading {vanGent:2007dp, VanMechelen:1992ts}. A typical runner will strike the ground more than 300 times per leg for each kilometer run {Edwards:2009kk} and this repetitive impact loading is thought to play an important role in the pathophysiology of many common running injuries, especially bony fatigue fractures (commonly termed stress fractures) {Meardon:2014fw, Bennell:1999ti, Beck:1998vb, Jones:2002hz, Pepper:2006cy}. In runners, between 35% and 49% of all fatigue fractures occur in the tibia {Bennell:2004ft, Matheson:1987vk, Crossley:1999ut, McBryde:1985wj}.

While bone loading is necessary to maintain bone mineral density through a continuous process of remodeling {Vico:2000im}, theoretically a fatigue fracture can develop when there is an inability of a bone to with-stand repetitive bouts of mechanical loading {Warden:2006ua}. During running the skeleton is exposed to cycles of mechanical loading, which is influenced by muscle forces, intersegmental joint forces and biomechanical factors, which results in bone strain {Warden:2006ua}{Levenston:1998vx}{Carter:1987vz} (Figure 1). Repetitive strain is naturally associated with micro-damage, which under normal circumstances stimulates a process of remodeling. However, where training / hormonal / dietary factors are sub-optimal, imbalances in the remodeling process can occur, whereby the level of micro-damage surpasses the body's ability for regeneration. Without intervention, and with continued bone strain, the subsequent accumulation of damage can start a pathology continuum that results in bony fatigue reactions and fractures, and ultimately complete bone fractures {Warden:2006ua}.

[INSERT FIG. 1 APPROXIMATELY HERE]

Due to a large ground reaction force, the instant the foot strikes the ground during gait, its velocity rapidly decelerates to zero, while the body continues to move forward {Whittle:1999eq}. This change in momentum is synonymous with the production of compressive loading of the lower limbs, and an impact shock that is transmitted through the skeletal system from the leg to the head, with local segment peak accelerations occurring at successively later times as the

shock moves up the body {Derrick:2004ft, Lake:2000gy}. To minimise damage to proximal structures the shock that travels up the body must be attenuated {Kobsar:2013uz, Mercer:2003ky}. This process is accomplished through a complex interaction of many passive and active mechanisms {Zadpoor:2012cw, Mizrahi:2001tq, Mizrahi:2000cx, Mizrahi:2000uj, Derrick:1998vt}. The viscoelastic properties of bone, articular cartilage and the calcaneal fat pad give rise to the passive mechanisms, while the active mechanisms involve muscle contractions and joint flexion {Pratt:1988kc}. A failure of the lower extremity muscles to adequately absorb the energy of impact may lead to an over-reliance on passive mechanisms for this attenuation {Derrick:1998vt}.

Accelerometers are small devices that directly measure acceleration of a specific body segment by exploiting Newton's second law ($F=ma$). Although the shock associated with limb deceleration is transmitted throughout the entire body, tibial acceleration (TA) during running in particular has attracted substantial research attention over a long period of time. This research has encompassed a range of individuals, both healthy and injured, running over-ground and on treadmills, across a range of speeds and conditions (i.e. different gradients, fatigue states, stride lengths and frequencies, etc.) {GarciaPerez:2014br, Mizrahi:2000cx, Hamill:1995ku, Derrick:1998vt, Chu:2004wx, Killian:2007td, Mizrahi:2001tq, Rickert:2012vf, Crowell:2011ho, Crowell:2010hn, Milner:2006ej, Flynn:2004fk, Clansey:2014gg, Clansey:2012ib, Clansey:2011wq, Liu:2009en, Sinclair:2013cp, Hennig:1991tz, Hennig:1993tn, Lafortune:1991jn, Lafortune:1991vg, Lafortune:1992ty, Lafortune:1995kz, Purcell:2006fo, Dufek:2009va, Oakley:1988ku, Sinclair:2013bi, Greenhalgh:2012vf, Sinclair:2016fe, Sinclair:2016er, Sinclair:2013bl}.

For the highest level of accuracy strain gauges would be attached onto the bone, however in-vivo measurement of tibial bone strains is invasive and difficult to undertake {Burr:1996vu}{Liu:2009en}. Accelerations of the bone, could be considered a proxy measure for the impact forces experienced by the tibia. Peak TA measured via devices attached directly to the tibia bone have revealed high correlations with key ground reaction force (GRF) parameters (vertical impact peaks $r=0.7-0.85$; loading rates $r=0.87-0.99$) {Hennig:1991tz}. While the correlations are not as strong 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 {Greenhalgh:2012vf}. The peak axial component of TA has been reported as a variable that can

discriminate between runners with and without tibial fatigue fractures {Pohl:2008bl}, and between runners injured and uninjured limb {Zifchock:2008jw}. Additionally, for every 1 g increase in axial TA, the likelihood of having a history of tibial fatigue fracture has been shown to increase by a factor of 1.4 {Milner:2006ej}. The relationship between tibial bone strain and acceleration during running have not been undertaken, and therefore these associations are at this point speculative.

Previous reviews that have included the measurement of TA in runners have highlighted some of the key elements for consideration, such as the attachment method and placement location of the accelerometer, as well as the need for a low mass multi-axis device for increased measurement accuracy {Norris:2014jz, Mathie:2004dc}. Despite this, these reviews have failed to address many of the issues and potential limitations that must be considering when measuring TA from runners, including the influence of running speed, technique, fatigue or surface characteristics. In the following sub-sections of this article, key findings from the literature covering the selection (Sect. 3.1), placement (Sect. 3.2) and attachment (Sect. 3.3) of accelerometers, as well as data analysis methods (Sect. 3.4) and key outcome measures (Sect. 3.5) are consolidated and evaluated. The second half of the review is focused assessing the intrinsic (Sect. 4.1) and extrinsic (Sect. 4.2) factors that affect TA. As this review is primarily concerned with providing recommendations for researchers and practitioners who intend on using TA as a key variable in intervention studies designed to reduce the risk of running-related injuries, discussion of the impact of extrinsic factors is limited.

2. Definition of Terms

A number of terms used interchangeably in the literature 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 authors have reported peak accelerations components from a device aligned to the long axis of the tibia. Resultant tibial acceleration (TA-R) is used where researchers have used the peak accelerations from all three axes of an accelerometer to calculate the resultant vector.

3. Review of the Literature on Tibial Acceleration Measurement

3.1. Device Selection

A detailed outline of the characteristics of all the different types of accelerometers goes beyond the scope of this paper, but this can be found in other sources {Mathie:2004dc}. Devices contain one, two or three accelerometers mounted at right angles to each other, each reacting to the orthogonal component acting along its axis {Winter:2005ud}. Accelerometers operate relative to the reference frame of the Earth's gravitational field, therefore all devices will constantly register 1 g (9.81 m/s) upwards as a reaction to the downward gravitational acceleration {Pottie:2005fq}. Recent improvements in sensors have enabled the manufacture of accelerometers that are small, light, consume low power and include wireless transmission capability. Known as micro-electro-mechanical systems (MEMS), these sensors open the door for monitoring human movement in the real world {Patel:2012jx, Iosa:2016fq}. MEMS accelerometers can differ across a range of parameters, which may impact on the quality of the final signal. One of the main differences can be the range and bit resolution of the acceleration captured. If the signal range exceeds the capture range of the chosen device, the measured signal will be clipped at either the upper or lower extremities. Some devices capture to on-board memory cards, which often have restrictions to the speed of their read-write capacity. Wireless transmitting devices can be associated with a variable length signal delay, or complete dropout. While on-board processing of data can, in some cases alleviate these problems, it can also result in a reduction in the fidelity of the data. Careful consideration of all of these points is necessary when selecting a device.

3.1.1. Uniaxial and Triaxial Accelerometry

The acceleration of the tibia occurs in three dimensions, which is often referred to with respect to a local tibial coordinate frame: axial, antero-posterior and medio-lateral {DeBeliso:2012uf}. Lafortune and Hennig {Lafortune:1991vg} measured TA-A, TA-AP and TA-ML components using a triaxial accelerometer. At running speed of 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 components (4.5 g). The TA-AP and TA-A components were reduced at running speeds of 3.5 m/s while TA-ML components remained constant. These findings led the authors to conclude that in order to accurately quantify the magnitude of the shock, it is important to measure all three components of acceleration. The existence of high TA-AP components supports the hypothesis proposed by MacLellan {MacLellan:1984vk} who, using high-speed films of the distal leg, identified a horizontally transmitted shock wave at heel-strike. Despite these recommendations many

researchers have used uniaxial accelerometers to measure peak TA-A only {Crowell:2011ho, Crowell:2010hn, DeBeliso:2012uf, Dufek:2008ko, Dufek:2009va, Kersting:2011hv}. Some researchers have used triaxial accelerometers to measure TA in runners, but have only reported the TA-A {Chambon:2014hh, Clansey:2014gg, Clark:2010dv, Creaby:2016fm, Greenhalgh:2012vf, LucasCuevas:2016ej, Sinclair:2013bi, Sinclair:2016fe, Sinclair:2016er, Sinclair:2013bl} (Table 1).

When measuring TA using a uniaxial accelerometer there is a need for careful alignment of the device to the long axis of the tibia {Crowell:2011ho, Crowell:2010hn, DeBeliso:2012uf, Dufek:2008ko, Dufek:2009va, Kersting:2011hv}. If the correct tibial alignment is not achieved, the acceleration will not accurately reflect the actual TA-A. However, using all axes from a triaxial accelerometer to calculate the TA-R eliminates the need to carefully align the device to the tibial coordinate frame, thus improving repeatability of the measurement {Sheerin:2017bu}.

Only a small number of studies have made an attempt to account for additional acceleration components applied to the tibia {Glauberman:2014es, Thompson:2016if, Giandolini:2015kw, Wood:2014jx, Sheerin:2017bu} (Table 1), with one research group reporting that step frequency influenced the acceleration components independently, where an increase in step frequency resulted in lower TA-A and TA-R peaks, but greater TA-AP acceleration {Giandolini:2015kw}. These data were captured from a single subject, running over a highly variable terrain. While Thompson and colleagues {Thompson:2016if} reported TA-R, it should be noted that this was calculated from two movement planes only (TA-A and TA-AP) {Thompson:2016if}. While this study appears to support the original recommendations of LaFortune and colleagues {Lafortune:1991vg}, being that the measurement of accelerations from more than one movement plane was important, the lack of a third axis, and therefore a true resultant vector, means that data could still be lost through axis misalignment. It is worth noting that because the TA-R takes into account all three axes, 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. {Lafortune:1991vg}, Glauberman et al. {Glauberman:2014es} report that no differences in TA-A were found between rearfoot (RF) and non-RF strike runners, however TA-R were reported to be greater in non-RF runners. While they did not report the additional individual components,

the additional acceleration present in the resultant signal could only have come from components other than TA-A.

3.1.2. Sampling Frequency

The matching of an appropriate sampling frequency is an important consideration for capturing accurate acceleration data. Nyquist theory dictates that for a perfect signal the minimum sampling frequency should be twice as high as the highest frequency present in the signal itself {Winter:2005ud}. 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 {Winter:2005ud}. Power spectral analyses have revealed that 99% of the TA signal power captured during running was contained below 60 Hz {Lafortune:1991vg, Hennig:1991tz, Shorten:1992vt}. Based on the conventions outlined above this would dictate a capture sampling frequency between 300 to 600 Hz. While almost all studies report a sampling rate of at least 1000 Hz {Hennig:1991tz, Lafortune:1991jn, Derrick:2004ft, Mizrahi:2000cx, Crowell:2011ho}, some have used a sampling frequency as low as 100 Hz {GarciaPerez:2014br}, while others failed to report the capture frequency they used {Glauberman:2014es}. A sampling frequency of 100 Hz fails to meet the baseline Nyquist recommendations and therefore the results must be called into question.

[INSERT TABLE 1 APPROXIMATELY HERE]

3.2. Accelerometer Placement on the Tibia

The distal tibia is a common site of fatigue fractures in runners, making it an important site for the measurement of acceleration {Milner:2006ej, Gruber:2014kz, OLeary:2007uo}, but many researchers have also measured TA more proximally on the bone {Verbitsky:1998uz, GarciaPerez:2014br, Duquette:2010gy}. However, these differing placements may not give comparable results. During the stance phase of running the tibia undergoes both angular and linear motions, with tibial angular motion largely confined to the sagittal plane, rotating about the distal axis of the ankle joint {Lafortune:1991vg}. 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 {Nigg:1995ge}. The maximum contribution of the acceleration due

to gravity can be 1 g (when the shank is vertical), but depending on the angle of the lower leg at impact, the measured acceleration contribution due to gravity could vary {Lafortune:1991vg}.

It has been reported that running at 4.5 m/s the angle of the tibia at impact can vary by up to 20° from vertical {Lafortune:1991vg, Lake:2005vo}. The TA signal is influenced by centripetal acceleration due to the sagittal plane angular motion, which acts in the opposite direction to TA-A {Lake:2005vo}. The angular acceleration is dependent on the tibial angular velocity and the distance of the device from the hinge point (i.e. the ankle) {Lafortune:1991vg}. Both measured and modeled estimates have indicated that the TA acceleration recorded on a device attached proximally (closer to the knee) substantially underestimates the TA-A at the distal attachment {LucasCuevas:2017ej, Lafortune:1991vg} (Table 1). Taking into account the contributions of gravity and the angular component of TA, Lake et al. {Lake:2005vo} reported that the measured TA (measured at 4.5 m/s) needed to increase by 1.5-3g 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, allowing for a correction for gravity and angular motions of the lower extremity {Lafortune:1991vg, Lake:2005vo}. An additional consideration, covered more thoroughly in the following section, the distal tibia also offers an area where there is minimal soft tissue between the skin and the bone, allowing for relatively firm attachment of a device compared to other locations on the lower extremity.

[INSERT TABLE 2 APPROXIMATELY HERE]

3.3. Accelerometer Attachment

Clearly, to determine the best estimate of the acceleration of a bony segment of interest, an accelerometer attached directly to the bone is most accurate {Lafortune:1995kz, Lafortune:1991jn, Hennig:1991tz, Lafortune:1991vg}, however this is impractical for regular use and therefore an alternative must be sought. LaFortune and colleagues {Lafortune:1995kz} compared the TA measured from bone and skin mounted accelerometers while runners ran over-ground. For some subjects the skin-mounted accelerometer overestimated TA by as much

as twice the magnitude of those mounted to the bone {Lafortune:1995kz}. 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 {Lafortune:1995kz, Shorten:1992vt}. As further highlighted by Shorten and Winslow {Shorten:1992vt}, the soft tissue between the bone and the transducer can act as an amplifier or a filter on the acceleration signal. While the absolute differences between the signals was large, with the use of a low-pass filter, signals from a skin-mounted device can adequately represent the accelerations measured on the bone {Lafortune:1995kz}. [There will likely always be some oscillation of skin-mounted accelerometers, and therefore it is essentially to know the characteristics of this oscillation. If the resonance frequency of the accelerometer and its mounting system occurs at the same frequencies of those from ground impact \(10-20 Hz\) the measured acceleration will be elevated {Shorten:1992vt}.](#) Ziegert and Lewis {Ziegert:1979ul} studied the effect of soft tissue between bone- and a surface-mounted accelerometer in vivo, by comparing its output with that of an accelerometer connected to the tibia bone via a needle. When the bone was impacted with a device, a 1.5-gram surface-mounted accelerometer showed nearly 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 {Hennig:1991tz}, 400 Hz {Lafortune:1991jn} and 1000 Hz {Lafortune:1995kz}. While Henning, et al. {Hennig:1991tz} indicated this frequency was both mathematically and experimentally determined, they did not outline or reference how this was carried out to allow others to replicate.

Accelerometer oscillation can potentially be minimised by tensioning it to the lower extremity. Using an embalmed cadaver leg, Saha and Lakes {Saha:1977gq} demonstrated that the preload force applied to elastic straps during the attachment of externally applied accelerometers had a significant effect on the data measured. Using a similar attachment regime, but with runners, Clarke et al. {Clarke:1985il} reported that as much preload force as tolerable provided the most reliable results, both within and between sessions. Forner-Cordero et al. {FornerCordero:2008jw} conducted a series of experiments in order to determine the frequency characteristics of skin-mounted devices under varied attachment conditions, including using elastic bands, which more have been more commonly used in recent research {Creaby:2016fm,

Clansey:2014gg, Dufek:2008ko, Abt:2011hq}. Secondly, they developed a repeatable ‘heel drop’ test to validate the attachment before taking measurements. The test involved subjects standing on their tiptoes and to fall freely on their heels, and while it is unlikely to produce TA magnitudes representative of running, it did show low variability and could discriminate between different attachment conditions {FornerCordero:2008jw}. These researchers report that without adequate preloading force, the frequency of the accelerometer-mounting system was too low, close to the frequency range of the data, increasing the chance of measurement error. While there is still no clarity from the research on what constitutes ‘hard’ or ‘as much as tolerable’ pre-load tension, and acknowledgement that this will differ for individuals, a simple test such as the ‘heel drop’ could provide an effective method for the frequency of the accelerometer mounting to be computed and difference systems compared before testing begins, or during the course of long experimental sessions.

[INSERT TABLE 3 APPROXIMATELY HERE]

3.4. Tibial Acceleration Analysis

3.4.1. Normalisation

In an attempt to account for variability in absolute magnitudes between sessions, Clarke and et al. {Clarke:1985us} proposed a form of normalisation to improve the accuracy of reporting of TA data. The particular methodology proposed expressed TA-A relative to the mean observed at the slowest running speed, giving what was described as a ‘shock ratio’. This normalisation 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 speeds would be required.

3.4.2. Frequency Content of Acceleration

While TA represented in the time domain are most commonly reported, the signal is formed by acceleration components of various frequencies, which are superimposed in the time domain signal {Shorten:1992vt}. The low frequency component (4–8 Hz) is associated with voluntary leg motion (e.g. stride length, segment alignment and velocity) and the acceleration of the body

centre of mass (COM), while the high frequency component (10–20 Hz) represents the rapid deceleration of the lower extremity at contact {Shorten:1992vt, Gruber:2014kz}. These lower and higher frequency ranges are also representative of the active and impact peaks of the vertical ground reaction force, respectively {Shorten:1992vt, Bobbert:1991ww}. The resonant frequency of the accelerometer mounting system also contributes to the time domain signal.

It is possible to separate the frequency components using a frequency analysis {Lake:1999tu, Shorten:1992vt}. A fast fourier transform (FFT) will provide the median power frequency of the acceleration signal, or alternatively a joint time-frequency distribution analysis can provide the instantaneous power spectrum {Lake:1999tu}. 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 {Giandolini:2015kw}. These additional signal analysis approaches have been used to provide a more thorough characterisation of the signal components in studies assessing running kinematics {Mercer:2002hc, Mercer:2003ib, Gruber:2014kz, Sinclair:2014kr} and kinetics {Lafortune:1995hm}, fatigue {Mercer:2003ky, Derrick:2002jt}, surface characteristics {Chu:2004tr} {Mizrahi:2000tg}, and footwear {Shorten:1986ec, Sinclair:2013bi, Sinclair:2013bl, Lake:2000gy}.

3.4.3. Signal Filtering

All kinematic data contains a true signal representing human movement and signal noise, therefore some pre-analysis filtering is always required {Winter:2005ud}. However, selecting the appropriate filter cut-off frequencies is important as over or under filtered data can lead to inaccurate interpretations. While both the true signal and noise occupy a wide band-width, noise is usually at the higher end of the frequency spectrum. Therefore 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 {Winter:2005ud}. Most studies measuring TA from runners report using low-pass filters with cut-offs set between 40 Hz and 100 Hz {Crowell:2011ho, GarciaPerez:2014br, Clansey:2014gg, Hennig:1991tz, Sinclair:2016fe, Creaby:2016fm, Lafortune:1991vg, Lafortune:1995kz} (see Table 1), which were in some cases determined via power spectral analyses of the signal {Lafortune:1991vg, Lafortune:1995kz}. A small number of studies either did not carry out any signal filtering during their post-processing or they failed to report the details {Lafortune:1991jn, LucasCuevas:2017tq, Wood:2014jx, Thompson:2016if,

Derrick:1998vt, Dufek:2008ko, Derrick:2002jt, Mizrahi:2000cx, Abt:2011hq, Derrick:2004ft, Butler:2007dp, Glauberman:2014es, GarciaPerez:2014br}. Since a TA signal also contains low frequency components (4–8 Hz) is associated with voluntary leg motion and the acceleration of the body COM, 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, but this does not appear to be widely used {Gruber:2014kz, Shorten:1992vt}.

3.5. Outcome Measures

For analysis purposes, when a triaxial device has been used, TA signals can be resolved into acceleration components along the three orthogonal axes defined with respect to the tibia: TA-A, TA-AP and TA-ML. The axial axis 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 {Lafortune:1991jn}. A number of additional variables can be calculated from the measured signals (Fig. 3). The most commonly reported are peak TA-A {Sinclair:2013bi, Greenhalgh:2012vf, Sinclair:2016fe, Sinclair:2016er, Sinclair:2013bl, Lafortune:1995kz, Lafortune:1991jn, Hennig:1991tz, Lafortune:1991vg, [LucasCuevas:2017tq](#), Giandolini:2015kw, Gruber:2014kz, Derrick:1998vt, Sheerin:2017bu, Mercer:2002hc, Dufek:2008ko, Crowell:2010hn, Crowell:2011ho, Derrick:2002jt, Mizrahi:2000cx, Abt:2011hq, Clansey:2014gg, Creaby:2016fm, Derrick:2004ft, Butler:2007dp, Mizrahi:2000tg, GarciaPerez:2014br}, followed by peak TA-R {Lafortune:1991jn, Wood:2014jx, Giandolini:2015kw, Thompson:2016if, Sheerin:2017bu, Glauberman:2014es}. A smaller number of studies have also reported peak positive TA-ML {Lafortune:1991jn} and TA-AP {Lafortune:1991jn, Giandolini:2015kw} axes, as well as time to peak positive {Lafortune:1995kz, Lafortune:1991jn, Hennig:1991tz, Lafortune:1991vg, Sinclair:2016er, Sinclair:2015ct, Sinclair:2014ia, Sinclair:2014ht}, TA-A slope {Duquette:2010wq, Sinclair:2017hi, Sinclair:2014ia, Sinclair:2015ct}, TA-A loading rate {Sinclair:2016er}, duration of peak positive {Lafortune:1991jn}, peak negative {Lafortune:1991jn, Glauberman:2014es}, duration of negative acceleration {Lafortune:1991jn}, and peak positive to peak negative acceleration {Laughton:2003uw, Oakley:1988ku}.

Despite the widespread use, publications describing the acceptable reliability of accelerometers attached to the tibia of runners is limited {Clarke:1985il, Sheerin:2017 bu}. Sheerin et al. {Sheerin:2017bu} report the one-week and six-month test-retest reliability for fourteen and eight runners, respectively. both TA-A and TA-R from 20 runners at a range of speeds (2.7-3.7 m/s) on a treadmill. While the TA-A results were deemed acceptable at each of the time-points, both the percentage difference in the means and the effects sizes, they were generally larger for TA-A compared to TA-R, indicating slightly better session-to-session reliability for TA-R. Clarke et al. {Clarke:1985il} 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 collection sessions.

[INSERT TABLE 4 and FIGURE 2 APPROXIMATELY HERE]

A = Peak positive acceleration; B = Time to peak positive acceleration; C = Duration of positive acceleration; D = Peak negative acceleration; E = Duration of negative acceleration

Fig. 2: Time domain variables calculated from the axial tibial acceleration at a velocity of 3.0 m/s

3.6. Intrinsic Factors That Can Modify Tibial Accelerations

3.6.1. Running Velocity

The seminal work assessing the effects 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. This increase (in TA-A) was also reported at a series of increasing running velocities (3.4, 3.8, 4.5, 5.4 m/s) measured from 10 well-trained runners {Clarke:1985us}. Further to this, linear regression analysis revealed that average TA-A increased by 34% for each 1.0 m/s increase in running speed. The authors report individuals demonstrated a linear relationship, but this did vary between 0.15 and 0.68. While the authors claim that the best-fit linear relationship was in 'good

agreement' with the experimental data, no statistics are provided to support this {Clarke:1985us}.

A further seven studies have reported TA-A while manipulating running velocity as an independent variable. The findings of all studies confirm that running at higher velocities was associated with increased TA-A, irrespective of running surface, footwear, running experience, or whether the speed was fixed, or self-selected {Sheerin:2017bu, Greenhalgh:2012wy, Boey:2017eq, Montgomery:2016bl, Sinclair:2013jm, Mercer:2002hc, Dufek:2008ko} (Table 1). While the focus of two of these studies were on the analysis of shock attenuation between the tibia and the head, their results provided insight into the manner in which TA-A changes with increasing velocity {Dufek:2008ko, Mercer:2002hc}. Mercer et al. {Mercer:2002hc} investigated the characteristics of shock attenuation across a range of running speeds up to a runners' maximum. The average TA-A remained constant (6.1 g) for both 50 and 60% of maximal speed, but increased across the subsequent increases in velocity (70%: 7.2 g, 80%: 7.9 g, 90%: 10.0 and 100 %: 10.9 g). The TA-A variability (SD) remained constant for the first four velocities (1.7-1.9 g), before increasing (2.7-2.8 g) at 90 and 100% of maximal velocity. In a subsequent study, they used a mixed model design to examine the impact of attenuation characteristics during running for specific groups of physically active females (prepubescent girls, normally menstruating women and postmenopausal women) {Dufek:2008ko}. Participants ran on a treadmill at their preferred velocity (1.9 to 2.6 m/s) and at a velocity 10% faster (2.1 to 2.9 m/s), while TA-A values ranging from 3.6 g to 6.1 g were recorded. The authors claimed that the study elicited the anticipated response for speed, with all groups exhibiting greater peak TA-A during faster running. However, when TA-A is compared across groups it is evident that the TA-A values measured from the prepubescent girls were larger than those measured from the normally menstruating women, despite running at a slower velocity. These differences could be as a result of a reduced tibia mass, and therefore effective mass, in the younger girls {Nigg:1995ge}. These particular studies were limited by having small samples sizes and comparisons were made against the percentage of their individual maximum {Mercer:2002hc}, or comfortable running velocity {Dufek:2008ko}, rather than an absolute velocity.

Despite the large number of studies measuring TA in runners, there is still an absence of normative values for runners at a range of running velocities. Sheerin et al. {Sheerin:2017bu} measured TA-R

for 82 runners running at four different treadmill speeds (2.7, 3.0, 3.3 and 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 quite 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 ($r=0.42$), and a regression analysis that revealed that for every 1 m/s increase in velocity TA-R would increase by 3.7 g.

3.6.2. Stride Rate and Stride Length

Although many variables are used to describe running kinematics, the fundamental components that dictate running velocity are stride length and rate {Mercer:2003ib, Mercer:2002hc}. Two research groups have published six studies that assessed the influence of stride rate and stride length on TA magnitude {Derrick:1998vt, Clarke:1985il, Mercer:2003ib, Mercer:2005fe, Mercer:2002hc, Hamill:1995ku} (Table 1). In the first three studies, researchers measured TA-A while manipulating stride rate to 5% and 10% slower, and 5% and 10% faster, than the subjects' preferred rate, while controlling speed at 3.8 m/s {Clarke:1985il}. Runners then ran with a 10% and 20% slower, and a 10% and 20% faster than their preferred stride rate, while running at their preferred speed {Hamill:1995ku}, and finally under the same stride rate conditions at 3.8 m/s {Derrick:1998vt}. Mean peak TA-A showed a positive linear trend as stride length increased across all three studies. This increase is likely due to a simultaneous increase in effective mass, which has been closely linked to knee angle (and therefore stride length) at impact {Nigg:1995ge, Frederick:1986vq}.

Independently manipulating stride length and rate at different velocities has further expanded the relationship of TA-A with these fundamental variables. It was reported that running with a longer stride length than preferred, increased the TA power spectral density {Mercer:2002hc, Mercer:2003ib}. These increases were four times greater when stride length was varied compared with when stride rate was varied {Mercer:2003ib}. When TA-A was compared between preferred stride length and stride length constrained to 2.5 m at various speeds {Mercer:2005fe}, magnitudes increased by approximately 24% per 1 m/s increase in running velocity, which is lower than the 42% {Mercer:2003ky} and 34% {Clarke:1985us} increases previously reported. However, when stride length was constrained, there was no clear relationship between TA-A and running velocity {Mercer:2005fe}. 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. However, these studies have not included sufficient measures to determine which aspects of stride length, such as the sagittal plane angle of the knee, or distance of the foot from the center of mass at contact, are important determinants of impact characteristics.

3.6.3. Fatigue

A review examining the risk factors linked with fatigue fractures concluded that while it is a complex phenomenon, exercise induced fatigue was an important factor in their development {Bennell:1999gs}. Exercise fatigue can be classified as central or peripheral, with central fatigue relating to a failure in neural drive, while the latter related to a failure of skeletal muscle contractile functioning {Noakes:2004kz}. A number of studies have reported increases in TA-A towards the end of high intensity treadmill running bouts designed to induce central fatigue {Derrick:2002jt, Voloshin:1998kx, Verbitsky:1998uz, Mizrahi:2000cx, Mizrahi:1997ur, Mizrahi:2000tg} (Table 1), in some cases by as much as 100% {Mizrahi:2000cx}. Derrick et al. {Derrick:2002jt} suggested that the increase in knee flexion angle and foot inversion at contact may be responsible for the increases seen in TA-A. They explain that these adaptations decrease the effective mass of the system, therefore increasing TA-A. Citing a spring-damper model simulating human running vertical ground reaction forces {Derrick:2000dl}, the authors reason that increased TA-A should not necessarily be linked to an increased injury potential, and suggest that it can be shown that decreasing the effective mass will increase the TA-A while at the same time decreases the impact forces {Derrick:2002jt}. These conclusions are contradictory to those who have linked increased TA-A magnitude with the development of tibial fatigue fractures in runners {Pohl:2008bl, Milner:2006ej}. These views do highlight the fact that the evidence is not clear and researchers are not in agreement on this topic.

While past studies have demonstrated increased TA as a result of fatiguing treadmill runs {Derrick:2002jt, Voloshin:1998kx, Verbitsky:1998uz, Mizrahi:2000cx, Mizrahi:1997ur, Mizrahi:2000tg}, more recently two studies have thrown further doubt over the relationship between TA and central fatigue {Abt:2011hq, GarciaPerez:2014br}. Abt et al. {Abt:2011hq} reported no changes in any kinematic or acceleration variables after the exhaustive treadmill run. Unclear findings were reported in a subsequent study where the effects of fatigue on TA-A were compared when runners ran over-ground and on a treadmill {GarciaPerez:2014br}. On

average, TA-A increased during the treadmill run, but this was not replicated in a similarly fatiguing run over-ground. Additional kinematic variables were not captured as part of this study, and therefore the characteristics of the adaptations could not be classified, or studied further.

Since it is difficult to control kinematic variables in a running situation, it has not been possible to examine the extent that local muscle fatigue effects TA in running {Flynn:2004fk}. Most 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 {Flynn:2004fk, Holmes:2006wq, Duquette:2010gy}. 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 {Flynn:2004fk, Holmes:2006wq, Duquette:2010gy}. It is thought that these changes are likely a result of the reduction in the force generating capacity of the muscle after fatigue. However, 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 {Duquette:2010gy}. 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 fatigue, both centrally and in localised muscles, on TA is inconclusive.

3.6.4. Joint Kinematics

Several researchers have suggested that the position or alignment of the lower extremity at initial contact may effect TA magnitude {Mercer:2005fe, Derrick:2004ft}, but this is not a straight forward concept. Joint positions at contact are closely connected to stiffness and effective mass. Denoth {Denoth:1985vp} and McMahon et al. {McMahon:1987tw} showed 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 more recently with the ‘two-mass’ running model, and its association with vertical ground reaction force–time waveform patterns {Clark:2017hw}.

Some authors have hypothesised that, with the connection between a foot strike pattern, and

the absence or reduction in vertical force impact peak, that landing with a strike pattern further forward on the foot (e.g. forefoot or midfoot) would result in lower peak TA {Laughton:2003uw, Oakley:1988ku, Giandolini:2013de, Glaberman:2014es}. However, the findings of these studies, when viewed in relation to footstrike mechanics, are conflicting.

Where runners transitioned from a RF strike (RFS) to either a midfoot strike (MFS) or forefoot strike (FFS) pattern, authors have reported either no change {Giandolini:2013de}, an increase {Laughton:2003uw, Gruber:2014kz}, or a decrease {Oakley:1988ku} in TA-A variables, a significant increase in TA-R {Glaberman:2014es}, or an increase in signal power in the 9-20 Hz frequency range {Gruber:2014kz}. Additionally, when non-RFS runners transitioned to a RFS pattern they demonstrated a significant decrease in TA-R {Glaberman:2014es}. There are a number of factors that could contribute to these conflicting findings, specifically the varied definitions of running kinematics (e.g. FFS {Gruber:2014kz} versus non-RFS {Glaberman:2014es}), different baseline characteristics of runners {Gruber:2014kz, Glaberman:2014es}, the differing intervention lengths for retraining patterns which are relatively habitual {Giandolini:2013de, Laughton:2003uw}.

To gain better understanding of the effects of kinematics on TA, a greater number of segments, lower extremity stiffness, effective mass, and the frequency content of TA are also considered. Analysing the discrete kinematic parameters associated with the passive attenuation of both time and frequency domain characteristics, it was reported that knee flexion velocity at foot-strike served as the single regulator of time domain peak TA {Sinclair:2014kr}. While there still remained a large proportion of variance and associated mechanisms unexplained, this study does suggest that kinematic parameters can influence TA magnitude during running. Using an alteration in decline gradient Chu et al. {Chu:2004tr} measured the changes in TA, kinematics, and stiffness parameters. While all runners demonstrated increased accelerations at the tibia and head with higher decline gradients, runners could be differentiated by the degree of increases in head acceleration. Those runners with reduced shock attenuation (i.e. relatively higher head accelerations) also demonstrated corresponding differences in lower extremity and trunk kinematics at both heel-strike and mid-stance. Additionally, these runners exhibited higher centre of mass displacement, heel-strike velocity, and reduced centre of mass stiffness and damping.

3.6.5. Active Modification of Joint Kinematics.

While studies have induced alterations in TA during running by imposing kinematic constraints, such as changes in stride length {Hamill:1995ku}, studies have also provided feedback and monitored changes as runners choose their own preferred kinematic patterns {Crowell:2010hn, Creaby:2016fm}. Two studies reported reductions in impact loading (some by as much as 50%) when runners completed a single session of real-time visual gait retraining {Crowell:2010hn, Creaby:2016fm}. However, retention of these changes was either not measured {Crowell:2010hn}, or only assessed in the short term (7 days) {Creaby:2016fm}. Similar positive findings were replicated with runners selected based on the high pre-intervention TA-A values of greater than 8 g {Crowell:2011ho} or greater than 9 g {Clansey:2014gg} running and 3.7 m/s. Runners received visual feedback based on their TA-A during six {Clansey:2014gg} to eight {Crowell:2011ho} treadmill runs. Reduced TA-A was demonstrated immediately after, and at 1-month post-intervention for those who received the feedback training {Crowell:2011ho, Clansey:2014gg}. While both groups of researchers reported that participants demonstrated new post-intervention gait patterns, this was only investigated by Clansey et al. {Clansey:2014gg}. 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 RFS to a MFS pattern. Reductions in TA-A were either not measured {Crowell:2011ho}, or not maintained more than one month post-intervention {Clansey:2014gg}.

Wood & Kipp {Wood:2014jx} investigated the feasibility of using audio feedback to reduce TA-R in runners. Progressively higher pitched beeps, matched to increased TA-R values beyond a preset threshold, runners were able to reduce their TA-R by ~10% while listening to the audio feedback for as little as 10 minutes. Once again, no post-intervention follow up was carried out to determine the longevity of these changes.

These studies provide a positive indication that TA can be modified using realtime feedback {Creaby:2016fm, Wood:2014jx, Crowell:2010hn, Crowell:2011ho, Clansey:2014gg}. However, the feedback has been administered via various modalities and protocols, and each has used different TA thresholds, enabling only limited comparisons to be made. Additionally, some researchers provided a single session of feedback only, while others provided up to 12 sessions over a 5-week period. This disparity was also reflected in follow-up sessions to assess the retention of any changes, which varied from no follow-up, to 12 weeks after the cessation of the

intervention. Further work is required to determine what might be a superior approach to providing feedback for the purposes of reducing TA while running, and how reductions in impact loading can be retained long-term.

[INSERT TABLE 5 APPROXIMATELY HERE]

3.7. Extrinsic Factors That Can Modify Tibial Accelerations

3.7.1. Running Surface

When studying TA, some researchers have opted for the apparent simplicity of overground running {Clansey:2012ib, Lafortune:1991vg, Derrick:2004ft, Boey:2017eq, Montgomery:2016bl}, while many have chosen treadmills as a result of their easier instrumentation, control of speed and slope, and improved repeatability of trials {Abt:2011hq, Butler:2007dp, Clarke:1985us, Derrick:2002jt, Glauberman:2014es, Montgomery:2016bl} (Table 1). However, treadmill running is known to result in modifications to stride frequency {Riley:2008iy}, contact time {GarciaPerez:2013he} and lower extremity joint kinematics {Riley:2008iy}. Treadmills also vary considerably in their mechanical and cushioning properties, both in comparison to other treadmills, but also in regard to running overground. There is some evidence to suggest that TA-A measured overground can be substantially higher than running on some treadmills under comparable conditions {GarciaPerez:2014br, Milgrom:2003ce, Montgomery:2016bl}. However, the relationship between TA-A magnitude and surface compliance is not straightforward. Fu et al. {Fu:2015kz} found no differences in TA-A across a wide range of surfaces at running at 3.3 m/s. Whereas Greenhalgh et al. {Greenhalgh:2012wy} reported higher magnitudes when participants ran at 5 m/s on concrete compared to a synthetic surface, but not at a reduced velocity (3.3 m/s). Conversely Boey et al. {Boey:2017eq} reported lower TA-A when runners ran on a more compliant woodchip trail (compared to concrete or synthetic track), but only when runners were restricted to a slower speed (3.1 m/s), in comparison to when runners chose their own running speed (average 3.7 m/s).

While not a running study, Potthast et al. {Potthast:2010tf} attempted to quantify relative contributions of muscle activity and knee angle to TA-A magnitude after external impacts, using a custom-built pneumatically driven impact device. They concluded that the surface hardness explained less than 10% of the variance of the TA-A, while knee angle and muscle pre-activation

explained 25 to 29% and 35 to 48%, respectively. 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 {Ferris:1999df}. For surfaces of higher compliance, leg stiffness increases, which serves to keep the path of a runner's center of mass the same regardless of the surface characteristics {Ferris:1998ij}. While it has not been studied, 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.

Due to the greater impact forces experienced with downhill running {Devita:2008bt}, impact accelerations are also likely to change. Negative correlations have been observed between TA-A, TA-ML and TA-R, as well as median frequency, during both treadmill and over-ground running {Chu:2004tr, Hamill:1984ks, Giandolini:2015kw}. Hamill et al. {Hamill:1984ks} reported 30% increases in TA-A on a 8.7% decline, compared to level running at the same speed. Similar, but slightly smaller increases in TA-A magnitude were found by Chu et al. {Chu:2004tr}, 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. {Mizrahi:2000tg} who observed similar magnitude TA-A, and a lower amplitude within the impact frequency range, from runners running on a 7% decline compared to running.

3.7.2. Running Footwear

Initial research reported substantial reductions in the higher frequency components of TA (above 20 Hz) during running by the use of footwear with midsoles, over barefoot conditions. The power spectral density of frequency components above 20 Hz were directly related to shoe midsole hardness {Shorten:1986ec}. Traditional running footwear has been characterised by an ethylene-vinyl acetate (EVA) midsole of approximately 20 mm thickness. Recent studies have shown that despite some shoe's demonstrating significantly reduced cushioning properties when mechanically drop tested {McNair:1994wd}, no difference in peak TA across various conventional thickness EVA footwear conditions could be demonstrated {Greenhalgh:2012wy, McNair:1994wd}.

More recently manufacturers have manipulated, designing barefoot-inspired (no midsole), minimalist (less than normal midsole) and maximalist (more than normal midsole) running shoes. In line with recent developments in running shoe midsole properties, several research groups have contrasted TA measured in conventional shoes, with that measured running barefoot {Sinclair:2013bl, Sinclair:2017hi}, in barefoot-inspired shoes (no midsole) {Sinclair:2016er, Sinclair:2017hi}, minimalist shoes (less than normal midsole) {Giandolini:2013de, Sinclair:2013bi, Sinclair:2017hi, Sinclair:2013bl}. In all cases running barefoot produced higher peak TA magnitudes than running in conventional footwear {Sinclair:2013bi, Sinclair:2017hi, Sinclair:2013bl, Sinclair:2013jm}. However, TA magnitude was significantly lower in conventional running shoes, compared to the barefoot, barefoot inspired {Sinclair:2017hi, Sinclair:2013jm} or minimalist {Sinclair:2017hi, Sinclair:2013bi, Sinclair:2013bl} footwear conditions. These findings were supported with further analysis of the TA median power frequency {Sinclair:2013bi, Sinclair:2013bl} and TA slope {Sinclair:2017hi}, leading the authors to conclude that running barefoot or in barefoot-inspired or minimalist shoes may place runners at increased risk from chronic pathologies.

Relatively new commercially available footwear has been developed which utilises either an over-sized lower density midsole (maximalist shoes), or expanded thermoplastic polyurethane (TPU) midsole, with claims of providing additional cushioning and shock attenuation and reduced energy loss {Worobets:2013jr, Sinclair:2016er}. Despite these claims, research findings have been less convincing, with results indicating peak TA's were greater in footwear designed to improve energy return {Sinclair:2014ia}. While no differences were reported between maximalist and conventional shoes, a higher trend in peak TA and TA loading rate were observed {Sinclair:2016er}. These findings suggest that a softer midsole did not provide any further benefits in impact loading than conventional shoes, and potentially they were less effective. These findings are less surprising when considered in context with the effects of surface characteristics on stiffness, where runners have been shown to increase their leg stiffness when running on high compliance surfaces {Ferris:1999df}.

Orthotics, or shoe inserts are an additional component of footwear that have been claimed to include shock absorbing properties. While TA-A magnitude or peak-to-peak TA-A were not reduced in either custom {Sinclair:2014ht, Laughton:2003uw, Oakley:1988ku} or over the

counter {Butler:2003de} orthotics, significant differences were reported in time to PT-A, and both average and instantaneous TA slope {Sinclair:2014ht}.

[INSERT TABLE 6 APPROXIMATELY HERE]

4. Conclusions and Recommendations

While the research on TA in runners is extensive, there are still considerable limitations to existing research and gaps in current knowledge. Where acceleration is measured from a skin-mounted device at a specific location along the tibia, there is the assumption that this relates to the acceleration of the bone, and ultimately bone stress and strain. However, transient bone stresses could be more related to factors such as overall tibial bending loads at impact, and not necessarily the acceleration magnitude at one location on the tibia. [There is however evidence to support acceleration, measured by an accelerometer, as a good proxy measure of the impact forces experienced by the tibia, which can be used by clinicians and researchers.](#)

Tibial accelerations are known to be affected by running technique, running speed, surface compliance, leg stiffness, as well as changes with fatigue and different footwear conditions. Users should be careful to control these variables to ensure that any identified differences (or changes) in tibial accelerations are a result of a true difference (or change) in performance, rather than a consequence of changes in other related variables. Additionally, further work does need to take place to improve the understanding of the inter-relationships between TA and each of these variables. There are also conflicting views on the how changes in effective mass impacts on TA, and what this means for risk of injury. While the alignment of the sensitive axis of uniaxial accelerometers with the tibia, as well as the interpretation of TA-A on its own, pose problems, calculating TA-R from all component axes of the accelerometer presents a potential solution. This still needs further investigation and the relationship between TA-R and injury still needs to be established.

For researchers and clinicians who intend to use accelerometers there are several recommendations that should be adhered to in order to achieve results that are accurate and reproducible:

1. Triaxial accelerometers of minimal mass should be used.

2. The accelerometer should be attached distally on the tibia.
3. The device should be affixed to an area with minimal overlying soft tissue, and strapped firmly to the leg.
4. The different axes of acceleration have meaning, and should be measured.
5. Calculating the resultant acceleration provides a single metric that takes into account all measurement axes.
6. Running velocity clearly influences tibial acceleration, so if values are to be compared between sessions or runners, velocity needs to be measured and controlled for.
7. Tibial acceleration signals should also be analysed in the frequency domain in order to identify signal noise, therefore guiding appropriate filter selection.
8. Running surface characteristics, subject fatigue levels and running footwear selection need to be accounted for and should be kept consistent between measurement sessions.

References

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