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Acceleration spikes and attenuation response in the trunk in amateur tennis players during real game actions

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Abstract

Although there are numerous locomotion studies analyzing the degree of attenuation of the acceleration spikes in the lower limbs and the trunk, few of these studies relate to tennis, where a high percentage of injuries occur in these body segments. The aim of this study was to describe the acceleration spikes and the attenuation response along the trunk, in real game actions. For this purpose, accelerometers were placed on the lower trunk, the upper trunk, and the head on a sample of 19 players while playing tennis matches. An average of 530 ± 146 acceleration spikes per match were selected in the upper trunk and a clear attenuation response between the upper trunk and the head was found (acceleration spike magnitude was approximately 25 m/s^2 in the upper trunk and approximately 20 m/s^2 in the head; $p < 0.05$; with attenuation percentages above 15%). In all players acceleration spikes of the head were below lower and upper trunk acceleration ($p < 0.05$ in all repeated measures ANOVAs and effect sizes were above 0.8, or large effect sizes). However, between the lower trunk and upper trunk no clear attenuation was found and although in some players the impact peaks were higher in the lower trunk ($p < 0.05$) the effect sizes were negligible or medium (Cohen $d < 0.5$). In other players the upper trunk peaks were higher than the lower trunk peaks ($p < 0.05$) and in a few players there was no significant difference ($p > 0.05$). The attenuation in the upper trunk, probably serves as a head protection/stabilization mechanism and more studies are needed to analyze the biomechanics actions underlying this attenuation response.

Keywords

Damping response, impact shock, IMU, acceleration peaks, biomechanical loading, musculoskeletal injuries, articular damage, disturbance of vestibular system, racket sports, tennis training

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Introduction

Tennis is an intermittent sport, with periods of high intensity followed by periods of rest of longer duration.[1] In a tennis match, sprints, jumps, changes of direction and jumps are performed in rallies that have a variable duration of 5–10 s, depending on the level of play, age or sex.[2] Running, jumping, or changing direction actions generate acceleration spikes in all segments of the body. Acceleration spikes have usually been measured using accelerometers as an indicator of the biomechanical load. The segments and joints play an important role by reducing the acceleration spikes in the head, otherwise the brain and other organs located in the cranium could be damaged.[3] Moreover, an appropriate attenuation of the accelerations during locomotion will minimize the disturbance of the visual and vestibular systems, and preserve head and gaze stability.[4,5] High acceleration spikes and inadequate attenuation mechanisms could also be related to an increased risk of musculoskeletal injuries (such as

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articular cartilage tissues injuries),[3,6] although this relation is not clear, and has been studied using either in vivo animal models of joint trauma, or by following the changes after impaction of cartilage explants.[7,8] The study of these attenuation mechanisms has a solid basic and long history. For example, in the late 1960s, Pradko and Lee[9] discussed the theory of human vibration response and explained the concept of power spectral density and absorbed power, which were widely used in posterior studies. In the late 1970s, Wosk and Voloshin[3] estimated the shock absorption capacity of healthy locomotor systems by placing accelerometers at specific anatomical landmarks on the legs, body, and head and recording the vertical component of acceleration. Later, Smeathers[10] presented the theory and details of a practical method based on skin-mounted accelerometers to assess the transmissibility of the human spine under various conditions.

Based on the assumption that accelerations of the trunk are a representation of whole-body center of mass accelerations, trunk accelerometry derived load measures have been used to quantify and assess whole-body biomechanical loads.[11–13] In addition, accelerometers are a reliable tool to measure physical activity in sports with repeated bouts of activity,[11] and are correlated with other measures of biomechanical load such as the session rating of perceived exertion,[12] biochemical fatigue markers[14] or even directly with an increased risk of injury.[13] Despite this issue, few studies have measured the biomechanical load in tennis, let alone during a match or training, by using accelerometers placed on the trunk. Running, jumping or changing direction actions are associated with axial accelerations of all body segments, which generate stress on the ligaments, muscles, and bone structures and may be associated with an increased risk of injury.[15] Considering this concern, studies of this nature are of great importance particularly in the case of tennis, because a high percentage of injuries occur in the lower trunk.[16,17]

Therefore, the aim of this study was to analyze the degree of reduction of the acceleration spikes from the lower trunk to the head during real game actions. It was hypothesized that there is an attenuation of the acceleration spikes from the lower trunk to the head. This study may be of interest to future racket sports specialists and may provide information on the magnitude and attenuation of acceleration spikes in the spine, both issues related to biomechanical loading, musculoskeletal injuries and/or mechanisms likely involved in reducing the disturbance of the visual and vestibular systems.

Methods

Sample

The sample consisted of 19 amateur male tennis players. The anthropometric characteristics of the sample were: age = 24.2 ± 9 years; weight = 72.7 ± 9.4; height = 176.6 ± 6.4 cm; skeletal muscle mass = 34.7 ± 4.1 kg; body fat percentage = $15.5\% \pm 4.8\%$; body mass index = 23.3 ± 2.5 kg/m^2.

A total of 10 friendly single matches were recorded. Eighteen players participated in one single match while one player participated in two matches (to obtain singular information this player was analyzed in only one of the matches). We recruited a heterogeneous sample with players of different ages and playing levels, representative of a large part of the population. Six of them were categorized as advanced tennis players with an International Tennis Number (ITN) of two; seven of them were considered advanced tennis players with an ITN between three and four, and six of them were categorized as intermediate tennis players with an ITN of six. All of them had a minimum of 8 years of experience playing tennis, were not injured at the time of the study, and had not engaged in vigorous physical activity in the 48 h prior to participation in the study. In addition, all players signed an informed consent form. The study followed the guidelines set out in the Declaration of Helsinki.

Instruments

For the current research, inertial 3D accelerometers Nexgen I2M SXT (Nexgen Ergonomic, Montreal, Canada; size: $48.5 \times 36.5 \times 13.5$ mm; weight: 22 g), at a sampling frequency of 128 Hz, and with an acceleration range of ± 60 m/s^2 were used. The same model of accelerometers have been used in previous studies analyzing the attenuation of acceleration spikes during running.[18] In sports with frequent changes of direction, devices of 100 Hz sampling frequency and 60 m/s^2 accelerometer range were also suitable to measure the player biomechanical load during real matches.[11–13] Furthermore, in a pilot study it was found that most of the acceleration spikes in real tennis matches did not exceed this accelerometer range. The x-axes of the accelerometers were manually aligned with the axial axis of the trunk and the sensors were configured to record in the internal memory in a synchronous way.

Procedures

Analysis of the acceleration signal

The acceleration spikes on the x-axis of the sensors (aligned with the axial axis of the trunk), produced in most cases by the players' jumps, sprints or changes of direction, were selected for the current study,[15,19,20] using a specific application (Peak Analyzer) of the software OriginPro version 2019b (OriginLab, Northampton, USA). The positive peak generated by the foot-floor contact was selected.[21] In a previous study, accelerations in *axial* direction showed greater accuracy than the resultant accelerations when comparing the data with those from a force platform.[19] Also, using the data on the accelerometer axis aligned with

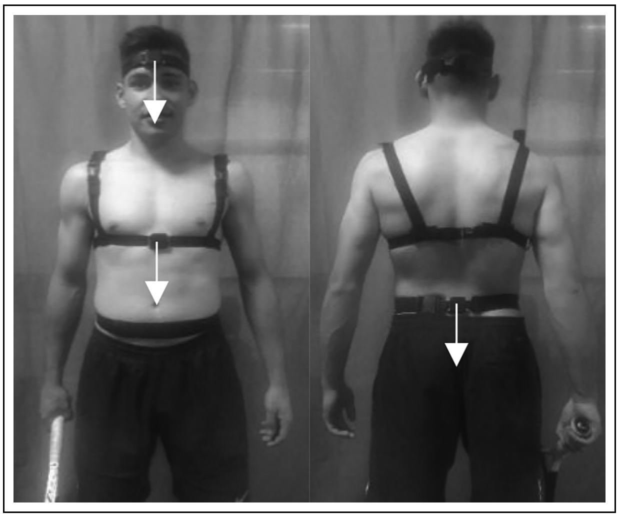

Figure 1. Placement of the accelerometers and direction of the x axis (the one used in the study) of the triaxial accelerometer.

the segment axial direction provides greater assurance that the acceleration spikes relate to axial accelerations,[22] which are expected to induce damage to muscles, bones, ligaments, and tendons and are proposed to be associated with joint pathologies.[23,24] The resulting acceleration signal could contain acceleration spikes due to actions such as sprinting or braking with a high acceleration component in a lateral or frontal direction. The acceleration signal was filtered through a 12 Hz zero-lag, fourth-order Butterworth low pass filter based on a residual analysis[25–27] (see supplemental material for more information). The signal cut-off frequency of 12 Hz was also chosen in a previous study[28] where an accelerometer was placed at a similar height on the trunk segment as in the case of the upper trunk accelerometer in this study (in this case at the back of the trunk) and at a similar sampling frequency (100 Hz). In the aforementioned study,[28] the signal from the accelerometers during running and change of direction tasks was compared with that recorded by a 3D capture system and the 12 Hz *Butterworth filter* showed the strongest relationship.

Assessment of acceleration spike magnitudes and attenuation response in real game actions

After a standardized warm-up consisting of a 3-min rally and 2-min serve, the participants played a match lasting approximately 40 min. During the match, rest times were controlled so that the maximum authorized time between games (90 s) was not exceeded. Three accelerometers were placed on each participant[20] (Figure 1): one at sacral level S1 (using an elastic belt), another at sternum level at the xiphoid process (using a harness made specifically for the study, in order not to impair the mobility of the shoulder) and a third centered at forehead level (using an elastic belt). The elastic belts were firmly placed in bone areas to prevent the sensor's own movements from generating unwanted accelerations. From here on, they will be referred to as the lower trunk accelerometer, the upper trunk accelerometer and the head accelerometer. The acceleration signal of each player was analyzed for 20 min of play (19.6 ± 1 min).

Upper trunk acceleration spikes higher than 18 m/s^2 were examined. This threshold was based on a previous paper in which acceleration spikes below that cut-off would be deemed extremely low intensity actions.[29] The cut-off criteria (18 m/s^2) was very consistent with our pilot tests in the field during game displacements, where acceleration spikes usually exceed this value (of approximately 2Gs) but not while walking. To find the acceleration spikes, time windows of 20 samples – or approximately 160 ms – were used. This time was similar to the duration of the foot contact on the floor during short sprint[30] or during a split-step when returning a tennis service.[31] Subsequently, acceleration spikes of the lower trunk and head were found, but in this case, the peak location windows were of five samples (to select all possible acceleration spikes) and the threshold magnitude value was 10 m/s^2 (approximately the acceleration of gravity, the accelerometer baseline). The acceleration spikes of the lower trunk and of the head closest to the upper trunk acceleration spikes (some authors use the term *neighboring acceleration spikes*) were found using a custom-built Excel template. This template, with a real example and more detailed explanations of the method is included as a supplemental file. Based on the acceleration spikes, the following variables were selected to measure the biomechanical load:[12,32] (I) the magnitude of the acceleration spikes in lower trunk, upper trunk and head (m/s^2); (II) the time (no. of samples) between the lower trunk and upper trunk acceleration spikes and between the head and upper trunk acceleration spikes; (III) the spike rate as number of selected acceleration spikes per minute of match (spikes/min).

The percentages of attenuation of the acceleration spikes between the lower trunk and upper trunk (AT1) and between the upper trunk and head (AT2) were calculated using the following formulas [33–35]:

$$AT1\ (\%) = \frac{AS\ lower\ trunk - AS\ upper\ trunk}{AS\ lower\ trunk} \times 100;$$

$$AT2\ (\%) = \frac{AS\ upper\ trunk - AS\ head}{AS\ upper\ trunk} \times 100;$$

with AS being the acceleration spike magnitude in m/s^2.

Any acceleration spikes that exceeded the sensor's measurement range were discarded for the calculation of attenuation indices and for comparison of means (explained in the statistical analysis section).

Statistical analysis

The statistical analyses were carried out with the tool *Real Statistic Using Excel* and with the *OriginLab* software. The mean and standard deviation of the selected variables were computed. Also, descriptively, and to have visual information, the *Kernell density curve* of the frequency histograms of the peaks was represented, taking into account the peak magnitude.

The differences in magnitude between the acceleration spikes of lower trunk and upper trunk and between upper trunk and head were evaluated by means of an *ANOVA* test of repeated measurements for each individual player and for the entire sample. In the case of the analyses for each individual, no normality test was performed because the number of accelerations registered was high enough to apply *the central limit theorem*. The *post-hoc* analysis was done by the *Tukey's Honest Significant Difference* (HSD) test. At the level of the whole-sample, the *Shapiro-Wilk normality test* and the *sphericity test* (homogeneity of variances) were performed, establishing the limit of the *epsilon* value at 0.5 $(1/(k\text{-}1)$,[36] where k is the number of groups of the independent variable, in this case three.

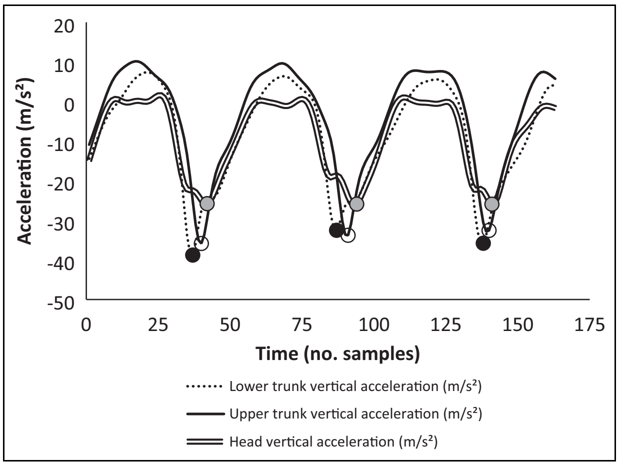

Figure 2. Example of the acceleration signals in the lower trunk, upper trunk and head on three consecutive steps during a tennis match. The reference peak in the upper trunk is marked with a black circle. The closest neighboring acceleration spikes of the lower trunk and head are marked with a white circle.

A two-sided threshold of significance was established at $p < 0.05$. The effect size (*Cohen's d*) was calculated using the free program *Psychometrica* and was considered as follows [37]: (I) 0–0.20, "negligible effect"; (II) 0.20–0.50, "small effect"; (III) 0.50–0.80, "medium effect"; (IV) 0.80–1, "large effect."

Results

Figure 2 shows an example of the acceleration spikes in the accelerometer of the lower trunk, upper trunk and head in three consecutive steps by one of the players who participated in the study. In all three cases the first peak that appears is in the lower trunk, followed by the upper trunk and the head. The plot also displays a clear attenuation of the acceleration spikes between the upper trunk and the head.

Figure 3 shows that AT2 (i.e. the percentages of attenuation of the acceleration spikes between the upper trunk and head) was positive and in 16 of the 19 players it ranged between 18% and 27%, also reveling that between the upper trunk and the head there was an attenuation of the acceleration spikes. The AT1 (i.e. the percentages of attenuation of the acceleration spikes between the lower trunk and upper trunk) was negative in 11 of the 19 players and below 7% in 14 of the 19 players.

The graphs containing the *Kernell density curves* of the peak frequency histogram (Figure 4), illustrate that the acceleration spikes in the high and lower trunk differ between the players, but head accelerations spikes were very similar between players (the maximum value of the *Kernell density curves* was close to $20\,m/s^2$). Table 1 also yielded very similar mean values for the head acceleration spikes among the players (the highest value was $21\,m/s^2$ for Player no. 3).

The mean time (number of samples) between the appearance of the lower trunk acceleration spikes and the appearance of the upper trunk acceleration was

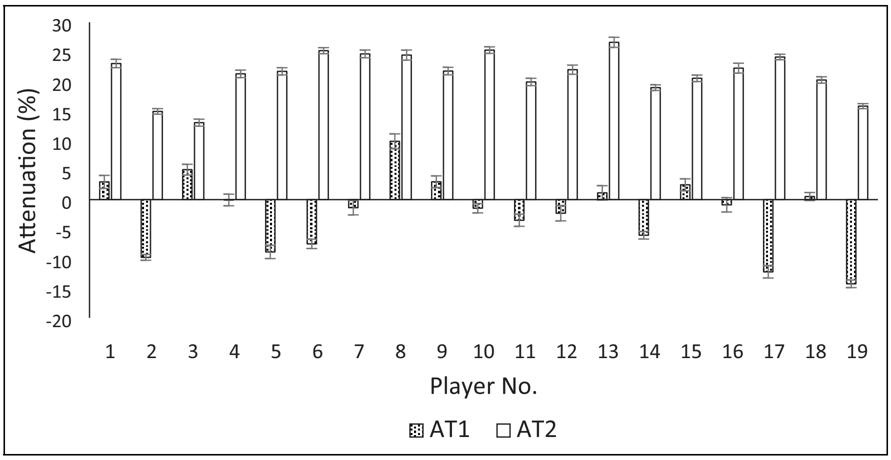

Figure 3. AT1 and AT2 of each player (averages and standard errors).

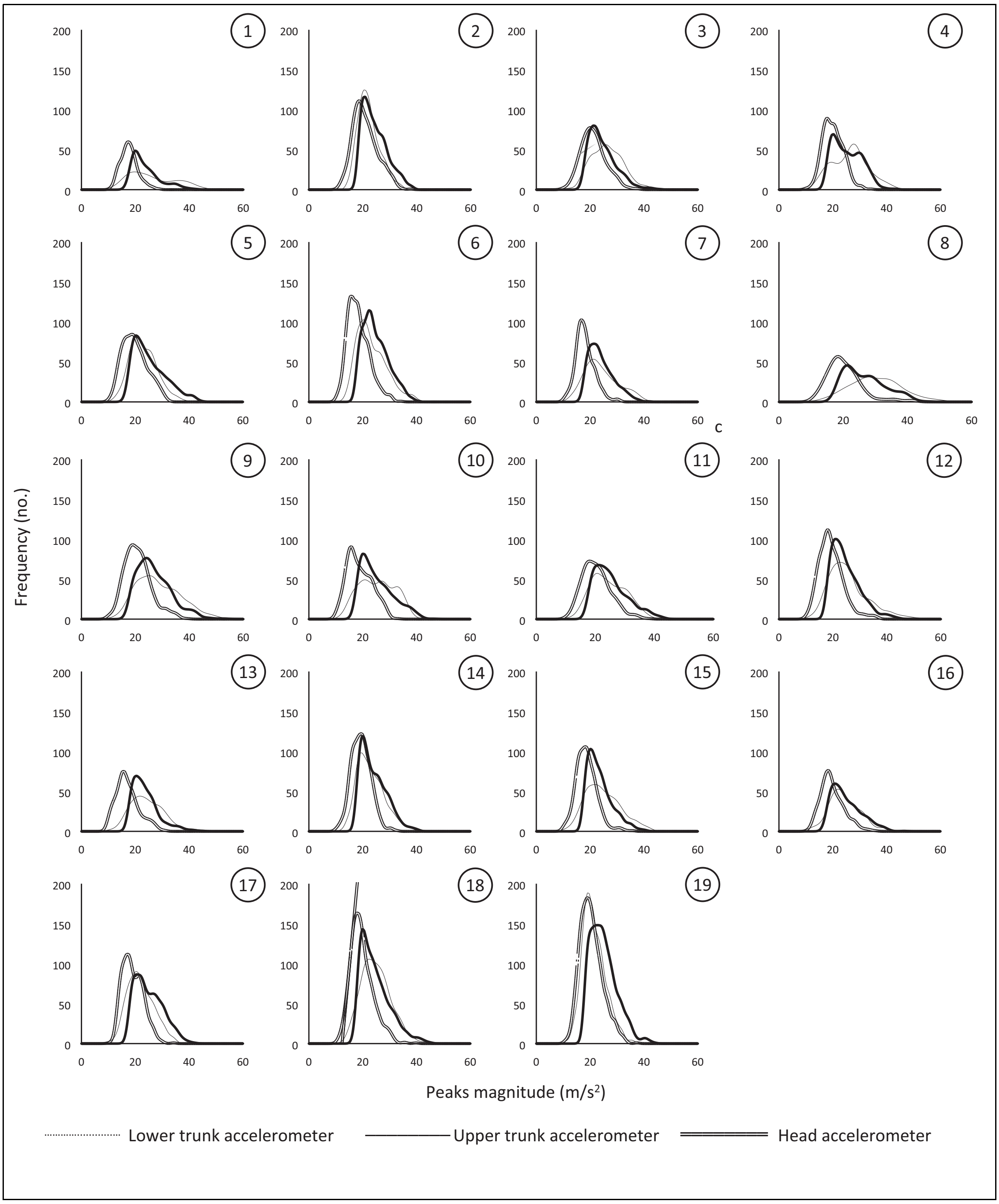

Figure 4. Kernell density graphs showing the frequency of acceleration spikes according to the acceleration magnitudes.

1 ± 0.5 samples and between the upper trunk acceleration spikes and the head acceleration spikes was 0.6 ± 0.5 samples. In all players those values were between zero and two samples. These results indicate that the acceleration spikes on all three accelerometers occurred at practically the same time. Although in most of the acceleration spikes studied, the lower trunk acceleration spikes preceded the upper trunk acceleration spikes and the upper trunk acceleration spikes preceded the head acceleration spikes (in 6592 and 6104 cases respectively), sometimes the upper trunk acceleration spikes occurred before the lower trunk acceleration spikes (in 2554 cases) and the head acceleration spikes occurred before the upper trunk acceleration spikes (in 2985 cases), but with a difference of less than five samples (i.e. less than 40 ms). Only 258 acceleration spikes occurred in the upper trunk before the lower trunk with a time offset greater than five samples and only 315 acceleration spikes occurred in the head before the upper trunk with a time offset greater than five samples.

The average peak magnitude, peak rate per player, and the results of the repeated measurements *ANOVA* at the player level are shown in Table 1. Of

Table 1. Acceleration spikes magnitudes, total numbers of peaks, peak rate and results of the individual repeated measures ANOVA (including effects sizes).

Player no.	Lower trunk peak magnitude (m/s^2)	Upper trunk peak magnitude (m/s^2)	Head peak magnitude (m/s^2)	Total no. of Peaks [a]	Peak rate (peaks/min) [b]	*Cohen's* d_1	*Cohen's* d_2	*Cohen's* d_3
1	24.2 ± 6.9	23.2 ± 5.2	17.4 ± 3.1	367	18	0.28	1.56	1.85
2	21.7 ± 3.9	23.9 ± 4.4	20.2 ± 4.3	601	30	1.08	1.83	0.75
3	26.3 ± 5.8	24.7 ± 5.1	21.2 ± 4.6	484	26	0.48	1	1.48
4	26.4 ± 6.7	25.4 ± 5.3	20.1 ± 3.8	454	24	0.26	1.41	1.67
5	23 ± 5.1	25.4 ± 6	19.8 ± 4.6	568	29	0.64	1.49	0.85
6	23.4 ± 5.6	24.8 ± 4.7	18.4 ± 4	661	35	0.48	2.11	1.63
7	24.2 ± 6.3	24.1 ± 4.6	18.1 ± 3.5	422	21	0.02	1.64	1.66
8	30.1 ± 8	26.7 ± 6.3	20.3 ± 5.4	425	21	0.66	1.23	1.89
9	27.1 ± 6.8	26.2 ± 5.5	20.5 ± 4.7	564	28	0.2	1.37	1.57
10	25.3 ± 6.6	25.3 ± 6	18.8 ± 4.6	554	28	0.01	2.1	2.09
11	24.6 ± 6	25.3 ± 5.6	20.4 ± 4.8	487	24	0.22	1.4	1.18
12	22.7 ± 4.8	23.4 ± 4.1	18.2 ± 3.7	321	18	0.21	1.66	1.45
13	24.5 ± 6.4	23.8 ± 4.9	17.1 ± 4.3	422	26	0.18	1.61	1.79
14	23 ± 5.2	24.1 ± 4.7	19.4 ± 3.6	586	29	0.43	1.86	1.43
15	24.7 ± 6.5	23.6 ± 4.6	18.6 ± 3.9	516	26	0.33	1.39	1.73
16	24.4 ± 5.8	25 ± 5.5	19.8 ± 4.3	380	19	0.18	1.52	1.34
17	22 ± 4.8	25.1 ± 5.1	19 ± 3.8	559	28	1.12	2.18	1.06
18	25 ± 5.6	24.7 ± 5.4	19.6 ± 4.2	783	39	0.1	1.53	1.63
19	21.4 ± 4.3	25 ± 4.9	20.4 ± 4.1	918	45	1.39	1.81	0.42
Average	24.7 ± 0.9	24.4 ± 2.1	19.3 ± 1.1	530 ± 145	27 ± 7	0.4 ± 0.4	1.6 ± 0.3	1.4 ± 0.4

[a]In the variables "Total no. of Peaks" and "Peak rate" all the peaks were taken into account, without eliminating those that exceeded the sensor's measurement range (not in the rest of the variables, where they were eliminated).
[b]Cohen d_1 corresponds to the comparison of lower trunk peaks and upper trunk peaks; Cohen d_2 corresponds to the comparison of upper trunk peaks and head peaks and Cohen d_3 corresponds to the comparison of lower trunk peaks and head peaks.

the total number of analyzed acceleration spikes (10,072), 615 exceeded the measurement range of the lumbar sensor, 112 the measurement range of the trunk sensor and only 13 the measurement range of the head sensor. In 17 of 19 players, there were significant differences with large effect sizes between the lower trunk and head acceleration spikes magnitude and between the upper trunk and head acceleration spikes magnitude. Between the lower trunk and upper trunk acceleration, there were significant differences with higher magnitudes in the lower trunk only in eight players, and the effect sizes were negligible to small in seven of those eight players (and moderate in Player no. 8). In Players no. 2, 5, 17 and 19 there were significant differences ($p < 0.05$) with moderate/large effect sizes but with the higher values in the upper trunk. In Players no. 7, 10 and 18 there were no significant differences ($p > 0.05$) between the lower trunk and the upper trunk. When the means of all players were compared, the results were consistent with those of the individual player analysis. There were significant differences between the magnitude of acceleration spikes of lower trunk, upper trunk and head ($p < 0.001$; $F = 3.25$; df = 2). The *post-hoc* analysis showed differences between the acceleration spikes at the lower and upper trunk versus the acceleration spikes at the head ($p < 0.001$ in both cases) with a large effect size (*Cohen's d* was 4.32 and 4.57, respectively). However, there were no significant differences between the acceleration spikes of lower trunk and upper trunk ($p = 0.72$).

Discussion

Numerous studies have analyzed how acceleration spikes attenuate from legs to head, however, most of them focus on gait and running. To our knowledge, this is the first study that analyses the attenuation of acceleration spikes in tennis. The acceleration spikes in the lower and upper trunk were very similar in most of the players (there are even a few players in which higher acceleration spikes were reached in the upper trunk more than in the lower trunk). However, between the lower/upper trunk and the head there is an attenuation of the acceleration peaks, as the acceleration peaks in the head were considerably lower compared to the lower/upper trunk.

Regarding the number of total impacts found, the data in the present manuscript are difficult to compare with those of previous studies due to methodological issues, such as the characteristics of the measurement devices used, the acceleration threshold established, or the nature of the sport. In rugby matches, 451 ± 493 total acceleration spikes and 13 ± 15 spikes per minute were found, which are in agreement with our work, however, collisions between players were also taken into account.[32] In soccer matches (the data were collected in approximately 1 h of the 90 min match), 1898 ± 730 acceleration spikes were found,[12] and the authors also took collisions between players into account. When the data were normalized with respect to time, the peak rate was 63 ± 8 spikes per minute,[12] which is also somewhat consistent with the data from the present investigation (Table 1). In the women's EuroBasket matches the referees recorded more acceleration peaks (approx. 1000)[38] than most of the tennis players in the present investigation. However, in the above-mentioned study, data from complete basketball matches were analyzed, longer in duration than the recordings of the current study. Only two players showed a similar number of acceleration spikes (Players no. 18 and 19 [Table 1]), compared with the data of the EuroBasket referees.[38] In other studies that also used the acceleration spikes to estimate external training load, the number of measured impacts was highly variable between players and measurements. For example, Clemente et al.[39] found a wide range of values in the case of soccer players during a season. In addition, in the present study a high variability between tennis players was found, both in the total acceleration spikes (ranging between 367 and 918) and in the rate of spikes per minute (ranging between 18 and 45) (Table 1).

In terms of the magnitude of the acceleration spikes, there is scarce literature analyzing actual tennis matches or any similar racket sport with which to compare the results of the tennis players in the present study. In the case of walking, the accelerations found in the pelvic area were evidently lower than in the lower trunk of the present study, being even below $10\,m/s^2$.[40] For runners,[20] accelerations between 30 and $50\,m/s^2$ were found in the lower and upper trunk, when running at speeds between 8 and 15 km/h. In this research,[20] the acceleration spikes were larger in the upper trunk than in the lower trunk (dependent on running speed). It differs from what was found in the present study, where in addition to the fact that accelerations were lower than in the mentioned investigation[20] in both the lower trunk and upper trunk, in some players the peaks were higher in the upper trunk, in others in the lower trunk or in other players there were no differences (Table 1). Macdermid et al.[18] suggest that vertical movement related to shock absorption may be responsible for the larger amplitude in the upper trunk and emphasize the need to investigate the underlying mechanism through video-based analysis. Similarly, while completing running and change of direction tasks in tennis, movements of the trunk can contribute to an increase in the acceleration signal, making peak magnitudes greater at more superior locations. With the methodology used in this study it is not possible to distinguish whether no attenuation occurs between the lower and upper trunk or if the degree of attenuation occurring is negated by movements of the trunk that increase the acceleration signal. To address this issue, it would be interesting to include trunk tilt data in future related studies, which can be obtained with fusion algorithms that integrate the information from accelerometers, magnetometers and gyroscopes or using a 3D photogrammetric system synchronized with accelerometers. This type of analysis could also provide information on the different attenuation mechanisms among players. At the level of the head, the accelerations were about 14 and $19\,m/s^2$ during running,[41] which is very similar to our findings (Table 1). Clemente et al.[39] suggested that the acceleration spikes at the head did not increase in the same proportion as for the legs, concluding that there were a number of attenuation mechanisms set in motion that prevented the head acceleration spikes from exceeding a certain threshold.

Finally, in the present study, the attenuation of the acceleration spikes between the upper trunk and the head was higher than 15% in all players, which also indicates that between the upper trunk and the head there is an impact peak attenuation, similar to what has been found in the case of running. Previous authors state that these attenuations could serve to improve head stability, facilitate vision during locomotion and dynamic stability, or to prevent injuries.[4–6,42] Head stabilization may be especially important in the case of tennis, where the player must always keep his eyes on the opposite side of the court and must achieve a good dynamic balance in the steps prior to and during hitting. Future studies should study the kinematic and neuromuscular control actions responsible for attenuating acceleration spikes in the cervical area in the case of tennis. On the other hand, attenuation of the acceleration spikes between the lower trunk and the upper trunk was very variable between players. Although no significant differences were found, in the majority of the sample (in 11 of 19 players, Figure 3) the attenuation index was negative. In only two players (no. 3 and no. 8), this attenuation was positive and greater than 5% (Figure 3), although the effect size was small. These two players registered the highest acceleration spikes in the lower trunk (Table 1) and may need to compensate it by increasing the attenuation in the thoracic trunk. This counter action may be in response to the activity of the trunk muscles, which is the main factor influencing dynamic lumbar stiffness and consequently the damping response.[6] Another case that differs from most of the sample is that of Players no. 2, 5, 17, and 19 who have an evident negative attenuation index of the acceleration spikes between the lower trunk and the upper trunk (confirmed by repeated measures ANOVAs performed at the individual level, with significant differences, ranging from moderate to large effect sizes) and were also the players that reached the lowest acceleration spike magnitudes in the lower trunk (Table 1). As already stated, another methodological approach is required -which allows estimating trunk movements in the different planes of movement- to know whether or not there is an attenuation along the spine. In addition, the differences in the attenuation pattern and the possible relation with trunk injuries should be addressed in future research.

As mentioned, the main limitation of this study is that trunk kinematics was not considered. This limitation prevents us from knowing in which sector of the trunk most of the attenuation response occurs and which action(s) help to dissipate most of the acceleration spikes. Also, the measurement range of the accelerometers may not allow one to measure impacts of very high intensity, although the data of our work indicates that this issue arose on relatively few occasions. Above all, this could be of more concern for the sensor placed in the lower trunk, which is where the higher accelerations were observed. In addition, in other sports with frequent changes of direction (e.g. rugby), players spent the majority of match-play (73.82–92.06%) in extremely low (0–2 g) to low (4–6 g) acceleration intensities, and less than 1% of the time in very high accelerations.[29] Finally, fatigue effects have not been taken into account, which could affect the effectiveness of technical movements and displacements and the attenuation responses. Despite those limitations, we believe that the data from this work should be taken into account, considering that it is, according to our knowledge, the first study in which *acceleration spikes* are analyzed in real tennis game actions, similar to those encountered during competition. Future work in this area could study the relationship between acceleration spikes and attenuation mechanisms, and risk of injury in real game actions. It would also be interesting to perform a comparative analysis of the acceleration spikes during different types of strokes (serve, forehand, backhand) and different types of supports (single or dual-leg support, with the hips facing the net or perpendicular to it), which could be done by synchronizing the acceleration signals with a high-speed camera. As shear forces in the spine could also damage the osteoarticular system, it would also be interesting to estimate the number of trunk rotations (flexion/extensions, rotations along the axial axis and lateral inclinations), for example, by combining the usage of the accelerometers with that of gyroscopes. Consideration should also be given to combining the time-domain approach with a frequency-domain one, considering that they capture a different kind of information, and frequency-domain could be sensible to collect the attenuation of the acceleration waves (partly because this type of analysis is less influenced by trunk movements in the sagittal, frontal, or transverse plane). Knowing the magnitude of acceleration peaks and their attenuation along the trunk in real game actions is important considering that these factors could be related to biomechanical loading, to musculoskeletal injuries and to mechanisms involved in reducing the disturbance of the visual and vestibular systems. Finally, the results of the present manuscript could also help to develop accelerometer-based biomechanical load monitoring systems for tennis.

Conclusions

This is one of the few studies that analyses acceleration spikes in racquet sports and in a competitive environment. As found in walking and running research, the magnitude of head acceleration spikes was lower than that of the trunk, suggesting an attenuation of the acceleration along the upper spine. Attenuation data between the lower and upper trunk do not reveal clear results, and future manuscripts need to include trunk kinematic measurements. Finally, acceleration spikes were variable at the between players level and some participants showed different patterns of attenuation. More studies are needed that analyze attenuation spike magnitudes in real tennis game actions and try to understand the biomechanical mechanisms involved in the attenuation of these spikes, based for example on high-speed video synchronized with IMUs.

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Supplemental material

Supplemental material for this article is available online.

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