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ORIGINAL

Comparative Analysis of Peak Impact Acceleration and Impact Shock Frequency Components According to the Type of Treadmill for Treadmill-running

Jae Seok Lee[1], Seung Eun Hur[2], Seong Han Park[3], Hwang Woon Moon[2], Bon Ho Koo[1]

[1]Department of Physical Education, Kyung Hee University, Yong-in, South Korea
[2]Department of Sports Outdoor, Eulji University, Sung-nam, South Korea
[3]Department of Physical Education, Kyung Nam University, Chang-won, South Korea

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Corresponding Author
Bon Ho Koo
The Graduate School of Physical Education, Kyung Hee University, 1732 Deogyeong-daero Giheung-gu, Yongin-si, Gyeonggi-Do 17104, South Korea
Tel : +82-10-7105-5550
Fax : +82-31- 204-8118
Email : bhkoo.caleb@gmail.com

Objective: The purpose of this study was to show differences in impact variables between treadmills with (treadmills B, C, and D) and treadmills without a shock-absorbing function (treadmill A) to propose the development of a treadmill with improved or added shock-absorbing function to reduce impact shock.

Method: Thirteen male students in their twenties who had habitual rear foot strike during running ran on four treadmills at 2.67 m/sec while ankle and neck acceleration data were collected. The magnitude of the ankle and neck acceleration peaks and peak positive ankle acceleration were calculated. The power spectral density of each signal was calculated to transform the ankle and neck accelerations in the frequency domain.

Results: The peak positive ankle acceleration on treadmill B was significantly lesser than that on treadmills A and D, and that on treadmill C was significantly less than that on treadmill A ($p < .01$). Peak positive neck acceleration was not statistically different between the treadmills. The frequencies of the peak power of the ankle and neck acceleration signal within the lower and higher frequency ranges were not statistically different between the treadmills. The signal power magnitude of the ankle in higher frequency ranges on treadmill B was significantly less than that on treadmills A, C, and D ($p < .01$). The signal power magnitude of the ankle in higher frequency ranges was not statistically different between the treadmills. The signal power magnitudes of the neck acceleration signal within the lower and higher frequency ranges were not statistically significantly different between the treadmills.

Conclusion: Our results indicate that the shock-absorbing function of a treadmill plays a role in reducing impact shock. Therefore, in future treadmill development, shock-absorbing function should be improved or incorporated to reduce impact shock to the body.

Keywords: Treadmill, Running, Impact, Frequency, Shock, Absorbing

INTRODUCTION

Vertical impact variables that are generated at the moment of ground strike during running, such as the magnitude of impact peak, magnitude of impact shock, and loading rate, have garnered the interest of many researchers for a long time as a major cause of running injuries (Gruber, Boyer, Derrick & Hamill, 2014; Hreljac, Marshall & Hume, 2000; Kwak, Mok & Kwon, 2005; Milner, Ferber, Pollard, Hamill & Davis, 2006; Zifchock, Davis & Hamill, 2006; Ryu, 2005). Many related precedent studies reported mechanical characteristics such as running speed and stride frequency, as well as the magnitude and duration of vertical ground reaction force that is generated when the foot strikes the ground, as the primary cause of impact shock transmitted to the body through the legs during running (Bobbert, Schamhardt & Nigg, 1991; Derrick, Hamill & Caldwell, 1998; Edwards, Derrick & Hamill, 2012; Hennig & Lafortune, 1991; Shorten & Winslow, 1992). In particular, subjects who show rear foot strike characteristically have frequency components of tibial acceleration generated in the stance phase and signal power of impact shock that appears separately in the lower frequency range of 4~8 Hz and the higher frequency range of 10~20 Hz (Bobbert et al., 1991; Derrick et al., 1998; Edwards et al., 2012; Hamill, Derrick & Holt, 1995; Shorten & Winslow, 1992). The lower frequency range shows the acceleration in leg movement and up-and-down change in the center of mass, while the higher frequency range shows the sudden deceleration in the foot and leg during initial foot strike (Bobbert et al., 1991; Shorten & Winslow, 1992).

Meanwhile, when the shock from such vertical impact is not properly absorbed and is transmitted to the head, it can cause damage to the vestibular organs and visual system (Derrick et al., 1998; Edwards et al., 2012; Gruber et al., 2014; Hamill et al., 1995), which can lead to injuries

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such as tibial fatigue fracture, knee chondromalacia, cartilage damage, and lumbar damage (Dufek & Bates, 1990). Accordingly, studies on the development of materials and products with improved shock-absorbing function as a measure for preventing running injuries that can be caused by such vertical impact are ongoing (Ko, Choi, Kim, Roh & Lee, 2004; Boo & Lee, 2005; Lee, 2004). However, such studies have focused mostly on the heel area of shoes or insoles by using functional materials (Oh & Lee, 2009), while previous studies and assessment data on the shock-absorbing function of exercise equipment such as treadmills are rare. A treadmill is an aerobic exercise equipment commonly used today owing to its advantage of not being affected by location or space (Yoon, Yi, Kim, Mun & Yang, 2000). It offers high research value because it is used in various fields with numerous methods, including the use in prescribed exercise for rehabilitation, training for athletes, and aerobic exercise by the general public.

Accordingly, the researchers in the present study established the research hypothesis that incorporating a shock-absorbing function in treadmills will reduce the impact force and shock transmitted to the body, thereby reducing the potential risk of running injuries from repeated treadmill use. The study also aimed to investigate the differences among treadmills with no shock-absorbing function, treadmills with a shock-absorbing function, and treadmills with improved shock-absorbing function. To achieve the research goals of the present study, four treadmills with different shock-absorbing functions were prepared for comparative analysis of the magnitude of the peak impact acceleration transmitted to the legs and head during treadmill running and the magnitude of the impact shock frequency signal power. As a result, the study confirmed the differences in the impact force and impact shock transmitted to the body on the basis of the shock-absorbing function that is incorporated in a treadmill. As such, we suggest that future development of treadmills should recognize the need for adding or improving the shock-absorbing function of treadmills to reduce the impact shock transmitted to the body.

METHODS

1. Participants

The candidates in the study consisted of 15 male students in their twenties who were enrolled in "E" University, located in Seongnam, Gyeonggi Province. The candidates responded to the recruitment advertisement that had been posted and volunteered to participate in the study. Among the candidates, 13 participants who did not have any abnormal orthopedic or cardiovascular findings in the past 3 months and had the characteristic of rear foot strike during running were selected for the study. Whether the participant had the characteristic rear foot strike was determined by analyzing images acquired at 120 fps using Sony a6300 (Sony Corp., Japan), which was installed on the sagittal plane, while the participant was running on the treadmill. All the participants received explanation from the researcher on the procedures and methods involved in data collection prior to their participation and subsequently provided written informed consent. The physical characteristics of the participants are shown in (Table 1).

Table 1. Physical characteristics of the participants (n = 13)

Age (yr)	21.7 ± 1.4
Height (cm)	174.9 ± 6.1
Weight (kg)	68.5 ± 10.9
Body mass index (kg/m^2)	22.4 ± 3.2

2. Data collection

During each measurement, each participant ran on a randomly selected treadmill and all participants wore the same shoes (PW0MW-16F512 Impulse 142; Prospecs, Korea) to control any effects from the shoes. The participants were given sufficient warm-up time before the measurement. While the participants ran for 120 sec on one of four different treadmills with the speed set to 2.67 m/sec, the impact acceleration transmitted to the body was measured. To eliminate any effects of residual fatigue from the previous treadmill running, a 7-day washout period was given in between measurements. Moreover, the belts in all treadmills were adjusted to rotate at the same speed. The experimental design in the present study is shown in (Figure 1).

The accelerometer used for measuring the impact acceleration transmitted to the body was G-Link-LXRS (LORD Microstrain, USA), which is a wireless acceleration measurement system that can measure acceleration by ±10 g with a maximum sampling rate of 512 Hz, and has a gateway that receives acceleration data by the radiofrequency method from nodes equipped with MEMS 3-axis acceleration sensors. The size of each node is 58 × 43 × 21 mm with a mass of 40 g, and the system has an accuracy of 10 mg and 12-bit resolution. In the present study, two nodes were used for measuring impact acceleration trans-

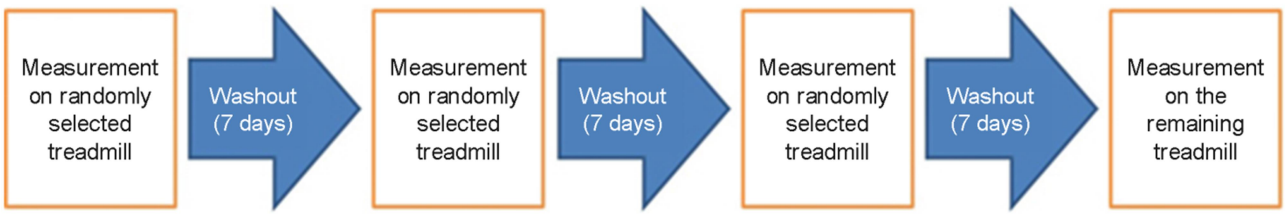


Figure 1. Data collection procedure

mitted to the body during running. One node was attached on the right ankle area (above the lateral malleolus), and the other node was attached on the neck area (below the seventh cervical vertebra). The node on the ankle was used to measure the impact acceleration transmitted from the impact source to the legs, while the node on the neck was used to measure the impact acceleration transmitted from the body to the head. As shown in (Figure 2), the nodes were attached firmly to the body by using elastic tape and joint brace to minimize any vibrational effect due to elasticity of the skin and soft tissues. The nodes were attached so that the longitudinal axis of the node and body segment would align when standing in the upright position. The software used to calculate the acceleration values was Node Commander (LORD Corp., USA), while the acceleration values measured from each node were synchronized by time.

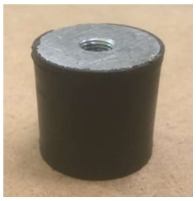
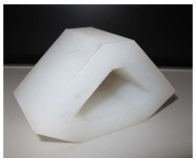
Figure 2. Experimental setup (accelerometers)

3. Treadmills

The four different treadmills with different shock-absorbing functions were as follows: a treadmill without a shock-absorbing function (treadmill A), treadmills with a shock-absorbing function (treadmills C and D), and a treadmill with an improved shock-absorbing function (treadmill B). The shock-absorbing function characteristics of each treadmill are shown in (Table 2).

Among these, treadmills C and D, which had a shock-absorbing function, differed from each other in their shapes and lengths (C: short "O" type, D: long "D" type), material (C: silicon, D: rubber), and the number of springs (C: 6, D: 4), which are responsible for absorbing the impact at the bottom of the deck, where the body comes in contact with the treadmill. Meanwhile, treadmill B, which had an improved shock-absorbing function, was similar to treadmill C in having 6 silicone springs at the bottom of the deck, but the springs had different shapes and improved function from the silicone springs that were installed on the treadmill foot, where the treadmill came in contact with the ground surface.

Table 2. Impact of the shock-absorbing functions of treadmills

Treadmills	Function 1 (deck spring)	Function 2 (foot)	Remarks
A	 No springs	None	No function
B	 6 Silicon springs	2 Silicon rings	Improved function
C	6 Silicon springs	None	
D	4 Rubber springs	None	

Function 1: Number, shapes, and materials of springs below the deck.
Function 2: Number and shape of silicon rings for treadmill foot.

4. Data processing

The collected acceleration data underwent DC offsetting using Matlab 2016a (MathWorks, Inc., USA) and fourth-order Butterworth low-pass filtering. Here, the cutoff frequency was set to 60 Hz (Hennig & Lafortune, 1991), while the linear trend of the signal was adjusted for detrending. To minimize the effect of step deviation, the analysis interval consisted of the stance phase, starting from 10 strides after the start of running and subsequent 10 strides, and the 10 peak impact acceleration values (peak g value) measured from each stride were used derive the mean peak impact acceleration value (Ryu & Lim, 2015; Gruber et al., 2014). Moreover, for comparative analysis of impact shock components, fast Fourier transformation (FFT) was used to transform the acceleration signal to frequency domain; after which, a rectangular window function was applied to determine the power spectrum density

Table 3. Time and frequency domain characteristics and statistical results for each treadmill

Variables	Treadmill A	Treadmill B	Treadmill C	Treadmill D	*F*		*p*
APA (*g*)	4.61 ± 1.76[bc]	3.58 ± 1.48[ad]	3.90 ± 1.22[a]	4.32 ± 1.53[b]	5.555	**	.003
NPA (*g*)	1.94 ± 0.49	1.86 ± 0.45	2.03 ± 0.48	1.98 ± 0.40	0.800		.502
$APPF_{low}$ (Hz)	5.1 ± 1.1	4.8 ± 0.9	4.4 ± 0.5	4.6 ± 0.6	1.651		.211
$APPF_{high}$ (Hz)	12.8 ± 2.3	12.1 ± 2.0	12.4 ± 1.6	13.2 ± 1.7	1.512		.228
$NPPF_{low}$ (Hz)	5.9 ± 0.8	6.1 ± 0.9	6.0 ± 1.0	6.0 ± 0.8	0.649		.535
$NPPF_{high}$ (Hz)	14.1 ± 3.7	12.4 ± 2.5	12.8 ± 3.5	12.9 ± 3.1	1.419		.253
$ASPM_{low}$ (g^2/Hz)	0.431 ± 0.137	0.493 ± 0.203	0.427 ± 0.189	0.500 ± 0.126	0.995		.406
$ASPM_{high}$ (g^2/Hz)	0.453 ± 0.214[b]	0.294 ± 0.142[acd]	0.380 ± 0.213[b]	0.447 ± 0.174[b]	5.617	**	.003
$NSPM_{low}$ (g^2/Hz)	0.068 ± 0.044	0.076 ± 0.050	0.076 ± 0.047	0.076 ± 0.055	0.401		.682

Variables: peak positive acceleration of above ankle (APA) and neck (NPA); frequency of peak power of ankle ($APPF_{low,high}$) and neck ($NPPF_{low,high}$) acceleration signal within the lower and higher frequency ranges; signal power magnitude of the ankle ($ASPM_{low,high}$) and neck ($NSPM_{low,high}$) acceleration signal within the lower and higher frequency ranges (mean ± SD).
a, b, c, d: significantly different between treadmills; $^{**}p < .01$.

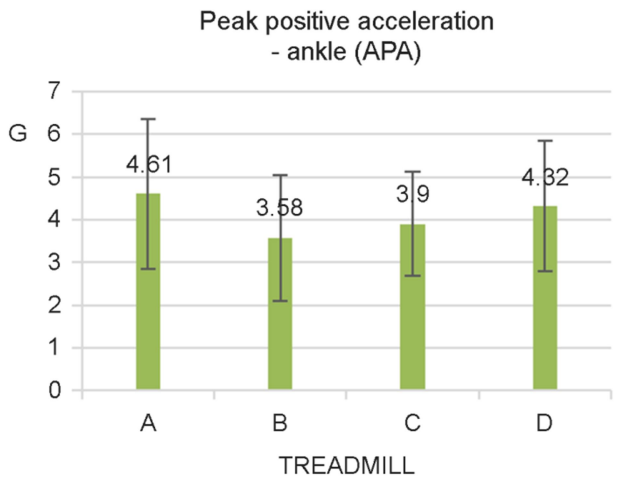


Figure 3. Peak positive acceleration of ankle (APA)

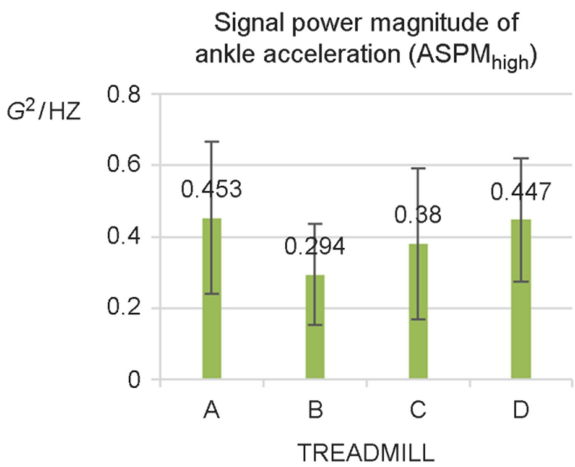


Figure 4. Signal power magnitude of ankle acceleration within the higher frequency ranges

(Gruber et al., 2014). The frequency power derived was divided into two components, the lower frequency range of 4~8 Hz and the higher frequency range of 10~20 Hz. The components were quantified by integrating the signal power magnitude of these two frequency ranges, while observations were made on the frequency components that generated peak power in each of these two frequency ranges (Bobbert et al., 1991; Derrick et al., 1998; Edwards et al., 2012; Hennig & Lafortune, 1991; Shorten & Winslow, 1992).

5. Statistical analysis

In the present study, SPSS 23 (IBM Corp., USA) was used to present the descriptive statistics of all variables. Moreover, one-way repeated-measures analysis of variance was performed to test the differences between the treadmills, while the least significant difference was used for post hoc testing. The statistical significance level for all tests was set to <5%.

RESULTS

The present study evaluated male participants in their twenties who had rear foot strike for the magnitude of peak impact acceleration transmitted to the body, frequency components that generate peak impact shock power for each frequency range, and magnitude of signal power of impact shock for each frequency range when running on four treadmills with differences in shock-absorbing function. The analyses results were as shown in (Table 3) and (Figures 3 and 4).

During the stance phase, the magnitude of the peak impact acceleration exerted on the ankle (APA) was smaller in treadmill B (3.58 ± 1.48 *g*) than in treadmills A (4.61 ± 1.76 *g*) and D (4.32 ± 1.53 *g*), and

smaller in treadmill C (3.90 ± 1.22 *g*) than in treadmill A (4.61 ± 1.76 *g*]; $p < .01$). Meanwhile, the magnitude of the peak impact acceleration exerted on the neck (NPA) did not show statistically significant differences between the treadmills. The peak power frequency in the lower and higher frequency ranges in the ankle ($APPF_{low,high}$) and neck ($NPPF_{low,high}$) did not show statistically significant differences between the treadmills. The signal power magnitude in the lower frequency range in the ankle ($ASPM_{low}$) did not show statistically significant differences, but the signal power magnitude in the higher frequency range in the ankle ($ASPM_{high}$) was smaller in treadmill B (0.294 ± 0.142 g^2/Hz) than in the other treadmills ($p < .01$). Meanwhile, the signal power magnitude in the lower and higher frequency ranges in the neck ($NSPM_{low,high}$) did not show statistically significant differences.

DISCUSSION

In the present study, significant differences in APA were found among the four treadmills used ($p = .003$). This suggests that, as reported in a study by Ryu and Lim (2015), decreased magnitude of the peak impact acceleration transmitted to the ankle causes the impact force transmitted to the neck to decrease as well. As a result, the impact force transmitted to the head is also lower, which can be helpful in preventing injury from impact shock. However, treadmill D, equipped with long rubber springs in the bottom of the deck, did not show a significant difference in APA as compared with treadmill A, which had no shock-absorbing function. This suggests that a silicone spring may be more effective in absorbing shock than a rubber spring. Moreover, treadmill B with improved shock-absorbing function (3.58 ± 1.48 *g*) tended to show a relatively smaller magnitude peak impact acceleration value than treadmill C with a shock-absorbing function before any improvement (3.90 ± 1.22 *g*). Although the treadmill with a shock-absorbing function did not show a statistically significant difference from the treadmills with a shock absorbing function before any improvement, but an improved shock-absorbing function appeared to be more effective in reducing impact force. It is believed that this difference would be clearer if the impact force transmitted to the legs increases with increased running speed, as reported by Boo and Lee (2005).

Meanwhile, no significant differences in NPA were found between the four treadmills used. However, the study by Ryu and Lim (2015) reported that at a running speed of 3.22 m/sec, the magnitude of peak impact acceleration transmitted to the head was affected by the magnitude of peak impact acceleration transmitted to the legs. By contrast, Boo and Lee (2005) reported that at running speeds of 3.0 and 6.0 m/sec, the magnitudes of impact acceleration transmitted to the lower legs and head were 3.4- and 5.5-fold greater, respectively. Therefore, in the present study, we determined that the running speed of 2.67 m/sec, which is used for measuring impact acceleration, may have been too slow for observing the impact force transmitted through the lower legs to the neck.

Meanwhile, numerous previous studies reported that when the frequency components of tibial impact acceleration at rear foot strike during running were divided into a lower frequency range of 4~8 Hz and a higher frequency range of 10~20 Hz, the lower frequency range showed the acceleration in leg movement and up-and-down change in the center of mass, while the higher frequency range shows sudden decelerations in the foot and leg during initial foot strike (Bobbert et al., 1991; Derrick et al., 1998; Edwards et al., 2012; Hamill, Derrick & Holt, 1995; Shorten & Winslow, 1992). In the present study, $APPF_{low}$ did not show differences between the treadmills used. However, a study by Gruber et al. (2014) reported 6.4 ± 0.5 Hz as the lower frequency component that generated the peak power of impact shock that acted on the tibia when track athletes with rear foot strike ran on a treadmill operating at a speed of 3.5 m/sec, which was slightly higher than that in the present study. This may be attributable to the difference in running speed used to measure impact acceleration and to that fact that running speed had the major effect regardless of the shock-absorbing function of the treadmill according to the characteristics of the lower frequency range, which shows the acceleration in leg movement and up-and-down change in the center of mass.

Moreover, $APPF_{high}$ shows the impact from the rapid deceleration generated in the initial stage of rear foot strike. In the present study, relatively lower higher frequency components were found in treadmills B, C, and D than in treadmill A, but the differences between the treadmills were not statistically significant.

However, $NPPF_{low,high}$ showed the characteristic of not showing any difference from the value corresponding to the frequency component in the ankle. Such results contradicted precedent studies that reported that higher frequency components were attenuated as the impact force moved up toward the upper segments of the body (Oh & Lee, 2009; Gruber et al., 2014).

No significant difference in $ASPM_{low}$ was found according to the treadmill used, but the results were similar to the magnitude of signal power in the lower frequency range (0.355 ± 0.092 g^2/Hz) measured in the tibia of track athletes with rear foot strike reported by the study by Gruber et al. (2014). Moreover, treadmill B with improved shock-absorbing function showed significantly lower $ASPM_{high}$ value. Although the differences were not statistically significant, treadmills B and C, which achieved a shock-absorbing function from using silicone springs, tended to show lower values than treadmill A, which had no shock-absorbing function, and treadmill D, which used rubber springs.

No significant difference in $NSPM_{low,high}$ was found on the basis of the treadmill used. We suspect that such results were due to the body responding appropriately to the impact transmitted through the legs to the upper segments to absorb the impact (Derrick et al., 1998; Lafortune, Lake & Hennig, 1996). Moreover, to observe the effects of impact shock transmitted from the treadmill to the neck, a running speed of at least 3 m/sec would be required. The fact that the characteristics of the treadmills, such as the material and number of springs used to achieve shock-absorbing function, were not controlled may be one of the limitations of the present study. Accordingly, we believe that a follow-up study is needed for a clearer comparison of the differences in impact force transmitted to the body based on the shock-absorbing function of treadmills by comparing the shock-absorbing functions at different speeds in treadmills manufactured by the same company that share similar characteristics.

CONCLUSION

The present study used four different treadmills with different shock-absorbing functions to analyze the magnitude of peak impact acceleration, frequency components that generate peak impact shock power for each frequency range, and magnitude of the signal power of the impact shock for each frequency range in 13 male participants in their twenties who showed rear foot strike during running, to investigate the differences in impact transmitted to the head during treadmill running.

The results showed that treadmills with a shock-absorbing function or improved shock-absorbing function reduced the magnitude of the peak impact acceleration transmitted to the legs and signal power in the higher frequency range, which was expected to have a positive effect on reducing the potential risk of injuries due to impact shock. Therefore, we suggest that future development of treadmills should recognize the need for improved or added shock-absorbing function to reduce impact shock.

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