How human musculoskeletal system deals with the heel strike initiated impact waves

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ABSTRACT

The objective for this work was to investigate how the human musculoskeletal system attenuates and dissipates the impact wave initiated at the heel strike. An experiment was designed to evaluate amplitude of the shock wave measured at both tibial tuberosities and forehead due to the heel strike. By analyzing data collected from the experiments, this work aimed to find how the human body could attenuate the heel strike initiated impact waves and to protect the brain from excessive accelerations. Ten young healthy volunteers participated in this study. Each subject was walking and running on the treadmill at four different progressive speeds for 30 seconds at each speed. The heel strike induced shock waves were recorded by externally attached accelerometers on both tibial tuberosities and forehead. The data analysis reveals that the heel strike induced impact waves recorded on both tibial tuberosities and forehead increases when the walking or running speed increases. However, the heel strikes induced impact wave on the two tibial tuberosities increases significantly faster compared to that of the forehead. The obtained result supports the conclusion that the human body acts as a natural shock absorber. It tries to attenuate invading impact waves in order to protect the head (brain) from exposure to excessive accelerations from the heel strike initiated shock wave. Its attenuational capacity is not constant, but rather increases with increase of the speed of gait. It is suggested here that such behavior is a manifestation of the built-in mechanism that attempts to protect the brain from overexposure to acceleration resulted from the impact wave initiated by the heel strike.

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1. Introduction

Gait is one of the most frequent activities humans are involved in. Process of gait, which includes walking, running, stair climbing or jumping, is always initiated by a contact between the foot and the surface. This contact has a limited time duration in order of 0.01 to 1 second depending on the way the subject moves, the surface and footwear properties. Relatively short duration of the contact leads to generation of an impact wave that propagates through the human musculoskeletal system (Voloshin et al., 1986, Ziegert and Lewis, 1979, Voloshin et al., 1981, Voloshin and Wosk, 1982). Such impact waves are also generated while dancing (Voloshin et al., 1989). These impact waves are propagating along the appendicular skeleton and can be measured at various locations such as ankle, tibial tuberosity, medial femoral condyle, hip and forehead. The study of the impact wave patterns at various points along the skeleton was performed by externally attached light weight...
accelerometers that did not impede the natural motion pattern of the subjects. It was found that the amplitude of the foot ground impact generated wave was gradually diminished on its way from the point of initiation (Voloshin, 1988) up to the forehead. This process of impact wave attenuation and dissipation is a natural way for human musculoskeletal system to deal with the onslaught of energy invading the system at each heel strike. It was found that the joints of the human musculoskeletal system: ankle, knee (Voloshin and Wosk, 1983), hip, vertebrae (Voloshin and Wosk, 1982) are capable to dissipate these waves at different rates, depending on joint and the subject’s health status. In order to attenuate these shock waves, increased muscular activity is needed (Oakley and Pratt, 1988). It was shown that the reflex is most effective for damping lower frequencies, while bone and soft tissue attenuate peak vibrations at higher frequencies (Voloshin and Wosk, 1982). The foot is the first organ that deals with the heel strike induced impact wave. Gait studies have shown that a heel strike triggers an impact wave having the amplitude of 0.5-25 ‘g’, depending on the activity. This wave is clearly noticeable at the calcaneus, knee, hip and forehead levels (Ledoux and Hillstrom, 2001, Saha and Lakes 1977). Along its path this wave is attenuated by a factor of about 1.5 at the knee joint while walking barefoot (Voloshin, 1988). The external environment of the foot (i.e. the footwear and the floor) can be manipulated to modify the impact waves that invade the musculoskeletal system. In general, footwear modifies dynamic (Light et al., 1980) behavior of the foot and provides a degree of protection by modifying and dissipating the heel strike initiated impact waves. Several investigators noticed that the walking pattern is an important contributor to the amplitude of the impact waves invading the human locomotor system. Encarnación-Martínez et al. (2015) compared the levels of impact between Nordic walking and walking. They find out that the levels of acceleration on the tibia and forehead were significantly higher during Nordic walking than during normal walking. Derrick et al. (1998) showed that increase in the stride length results in the increase of the acceleration magnitude recorded at the tibial tuberosity and the forehead. Nigg et al. (2015) recently proposed an interesting concept of the “preferred movement path”. The idea is that the skeleton of a person performing particular task (running or walking) attempts to stay in the same movement path. Muscle activity is modified to ensure that the skeleton stays in this path that is preferable to the musculoskeletal system of the individual. A number of researchers have investigated the influence of the heel strike initiated impact waves on the well-being of the human body (Voloshin and Wosk, 1982, Oakley T, Pratt DJ., 1988). Voloshin and Wosk (1981) examined the absolute values of the amplitude of the propagated waves. These studies also introduced the contemporary methods for evaluation of the human musculoskeletal system capability to attenuate these waves by using the non-invasive in-vivo technique for quantitative characterization of the locomotor system’s impact absorbing capacity. Nigg et al. (2015) analyzed the rating of a running shoe impact absorbing capacity by testing how well a particular pair of running shoes absorbs the impact and transmits the heel strike generated impact wave through the body. The results showed that there was no specific shoe that could provide the best impact absorbing performance for everyone. Number of gait parameters, like impact peak, loading rate, thrust maximum, decay rate, average vertical GRF, change in vertical velocity, braking impulse, propulsive impulse was determined to be speed dependent (Munro et al., 1987). Increased walking speed caused significant increase in the muscle activities of lumbar erector spinae, biceps femoris, and medial gastrocnemius, lumbar motion, as well as the vertical ground reaction force in the loading response and mid-stance phases (Chiu and Wang, 2007). The magnitude of the heel strike initiated impact waves also depends on the gait speed. It was shown (Clarke et al., 1985, Voloshin 1988, Voloshin 2000, Gruber 2014) that the increase in the speed leads to the increase in the amplitude of the waves. During a gait the body is in a state of dynamic imbalance (Mackinnon and Winter, 1993), therefore adjustments are continuously done that influence the foot placement that in turn affects the internal moments and joint rotations (Sekiya, 1997, Owings and Grabiner, 2003, Winter, 1984). An important question still is not answered – does the increase in the gait speed lead to the modification of the gait pattern or musculoskeletal system ability to dissipate and attenuate these waves in order to limit the exposure of the forehead to the heel strike initiated impact waves? The main hypothesis to be tested in this paper is: does the increase in the gait speed lead to the same increase in the amplitude of the heel strike initiated impact waves at the tibial tuberosity and on the forehead?
2. Materials and Methods

2.1 The instrumentation

Accelerometers were used to measure the amplitude of the heel strike initiated impact wave that propagates throughout the human musculoskeletal system by attaching them non-invasively to the bony prominences along the axial and appendicular skeleton, like tibial tuberosity, medial femoral condyle, sacrum or forehead (Voloshin et al., 1986, Oakley and Pratt, 1988, Ledoux and Hillstrom, 2001). Lightweight accelerometers (303A02-SN-9227, PCB) were used for data acquisition. They were mounted into an aluminum holder (Figure 1) that was tightly strapped to the points of interest (forehead and both tibial tuberosities) using Velcro strips. The strips were attached as tightly as possible, but did not to cause any discomfort to the subject. Such attachment can faithfully measure the amplitude of an impact wave as was shown in many works (Saha and Lakes, 1977, Ziegert and Lewis, 1979, Gruber et al., 2014).

The accelerometer was connected to an analog-to-digital converter (A/D) (model 480D06, PCB) that performs the conversions at a preset sampling rate of 500 samples/seconds/channel.

![Figure 1. Accelerometer mounted to a holder.](image)

2.2 Experimental Procedure

Ten randomly selected undergraduate students that were enrolled in a Biomechanics laboratory course participated in this study. The data was collected during regular scheduled laboratory activities that were a part of the required laboratory course. All subjects were in good health and were classified as habitual rearfoot runners. No histories of muscle weakness, neurological and musculoskeletal disease, or drug therapy have been recorded. The average and standard deviation age of the subjects was 20.9 ± 1.2 years, mass 59.3 ± 11.20 kg and height was 168.8 ± 13.12 cm. To ensure the uniformity of the experiment conditions, all subjects were asked to run on the same treadmill (SMOOTH 9.3P, Smooth Fitness, Mt. Laurel, NJ, USA) and were instrumented with the same piezoelectric accelerometers. The accelerometers were attached to the forehead and both tibial tuberosities (Figure 2). These locations were selected to reduce the effect of soft tissue on the wave transmission from the bone to the accelerometer (Wosk and Voloshin, 1981, Gruber et al., 2014). The subjects’ height, weight and gender were recorded. The subjects were asked to walk and run at prescribed speeds: 0.894 m/s (2 mph), 1.341 m/s (3 mph), 1.788 m/s (4 mph) and 2.24 m/s (5 mph).

After the three-minute warm-up each subject was walking or running on the treadmill for two minutes. These two minutes were divided into four equal periods, each period lasted 30 seconds. During each period the subject was walking or running at the prescribed speed. The subjects walked at the two slower speeds and ran at the two faster speeds. During each 30 seconds interval the acceleration data were acquired by an A/D converter at a sampling rate of 500 Hz per channel. The earlier study showed that the main frequency range of the acquired signal was below 100 Hz (Ziegert and Lewis, 1979, Voloshin and Wosk, 1989, Kim et al., 2011), thus the selected sampling rate was more than adequate to acquire the data without any bias. This setting provided information on about 30 heel strikes, depending on the subject’s speed. The data was acquired by a custom MATLAB (Mathworks, Inc., Natick, MA, USA) routine and stored for the off-line processing and analysis.

For each subject at each speed the following data was acquired: time and acceleration magnitudes on the left tibial tuberosity, on the forehead and on the right tibial tuberosity.
In order to exclude the possible gait modifications, the subjects were not informed when exactly the data acquisition was started. The repetitive nature of the test is evident from the typical plot of the data recorded. Figure 3 shows two consecutive tibial tuberosities strikes. One can see that following the right foot heel strike (R), impact wave propagates to the forehead where it is identified by F(R). Due to the left heel strike (L) the corresponding impact wave is observed on the forehead as indicated by F(L). This process repeats itself until the end of data acquisition. For the data analysis a MATLAB procedure was developed. It utilized the known running speed and this information was used to automatically locate the positions (time) and amplitudes of each heel strike generated impact wave. It proceeded automatically and detected all occurrences of the impact wave at the tibial tuberosities and forehead. Since there was always a possibility of a “bad” data (i.e. subject stumbled or misplaced the foot), the detected strikes were shown on the screen and confirmed by an operator. The sample output of this procedure used for the data recorded by the accelerometer attached to tibial tuberosity is shown in Figure 4. One can see that the peak at the time 2 seconds was not detected, thus it was added manually and used for the overall analysis.
Figure 3. The sample plot of the acceleration recorded on the forehead and both tibial tuberosities. “L” represents the peak acceleration on left tibial tuberosity due to the left heel strike. “R” represents the peak acceleration on right tibial tuberosity due to the right heel strike. “F (L)” represents the acceleration on the forehead corresponding to the acceleration peak recorded at the left tibial tuberosity. “F (R)” represents the acceleration on the forehead corresponding to the acceleration peak recorded at the right tibial tuberosity.

Figure 4. The time and amplitude of each heel strike as detected by MATLAB procedure. The red circle shows the detected peak.
Figure 5. The average acceleration on the forehead and tibial tuberosities as function of the speed. Solid lines represent linear fit to the data; the associated equations are shown below the solid lines. The vertical bars show ± 1 SD. Each value is an average over all trials and all subjects.

\[ y = 0.6375x - 0.3973 \]
\[ y = 1.8202x + 0.0784 \]

Figure 6. The acceleration on forehead (g). Error bars represent standard deviation. ("ns" denotes insignificant difference; all other differences are significant).
3. Results and Discussion

For each subject the averages of the accelerations due to each heel strike measured on the tibial tuberosities and forehead were calculated as a function of the speed. Data from all subjects were averaged and the results are shown in Figure 5.

The observation of Figure 5 reveals that the slope of the acceleration recorded on the tibial tuberosity (1.82) versus speed is much higher than the slope of the acceleration recorded on the forehead (0.64). The analysis of the slopes difference shows that it is significant at the 90% confidence interval.

From the obtained results it is obvious that the musculoskeletal system is capable to modify its rate of attenuation of the heel strike initiated impact wave on its path toward the forehead. One can suggest that the musculoskeletal system modifies its impact absorbing capability in order to prevent the brain from overloading by the high amplitude waves. At this point it is hard to say how this modification takes place. One can speculate that the gait pattern is changed or that the attenuation capacity of the musculoskeletal system is not constant. The bottom line is: the increase in the amplitude of the impact waves reaching the forehead is smaller than the increase of the amplitude of the impact waves on the tibial tuberosity.

Analysis of the average acceleration recorded on the forehead for all subjects as a function of the speed reveals that the difference between the accelerations is significant at 5% level, except between the speeds of 0.894 m/s (2 mph) and 1.341 m/s (3 mph) as shown in Figure 6.

Figure 6 shows that the amplitude of the impact waves recorded on the forehead increases when the speed of locomotion increases. When the speed is at a relatively low level (0.894 m/s and 1.341 m/s), the acceleration amplitude at the forehead is basically at the same level since there is no significant difference between the acceleration recorded at speed of 0.894 and 1.342 m/s. However, at the increased speed the difference becomes significant.

Analysis of the average amplitude of the impact waves recorded on the tibial tuberosity for all subjects also increases when the walking or running speed increases (Figure 7). The differences between the accelerations are significant for all speeds except between the speeds of 1.788 m/s (4 mph) and 2.24 m/s (5 mph). The general trend of the increase of the measured acceleration that represents the amplitude of the impact wave propagating through the human locomotor system with the increase of the speed is obvious.

![Figure 7. The acceleration recorded on tibial tuberosities (g). Error bars represent standard deviation. (“ns” denotes insignificant difference, all other differences are significant).](image)

**Conclusion**

The effect of the speed of locomotion on the amplitude of the heel strike generated impact wave as recorded on the tibial tuberosity and the forehead was studied. As it was already shown (Voloshin 1988, Gruber, 2014) the amplitude of the impact wave increases with the increase of the gait speed. However, it is important to evaluate the rate of increase on the forehead relatively to the tibial tuberosity as a function of a speed. The results obtained in this paper clearly indicate that the rate of acceleration increase on the tibial tuberosity is significantly higher than the rate of increase on the forehead. Therefore, the initial hypothesis that the increase in the gait speed leads to the same increase in the amplitude of the heel strike initiated impact waves at the tibial tuberosity and on the forehead was not correct. The analysis of the obtained data indicates that due to increase in the gait speed the human body modifies the way it attenuates and dissipates the heel strike initiated impact waves on their way toward the head. One may suggest that such behavior is the result of the built-in defense mechanism that attempts to keep the head (probably the brain) from overloading by excessive accelerations resulting from the heel strike initiated impact waves. This may be a result of the change in the gait pattern (Derrick et. al, 1998) or variable impact attenuation capacity of the human body due to the gait speed change. The changes in the gait patterns may be
an interesting topic for further studies.

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References


