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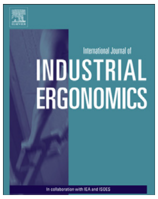

Effects of heel cushioning elements in safety shoes on muscle–physiological parameters

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ABSTRACT

Safety shoe designs are primarily based on safety requirements. But all-day comfort should not be luxury: Heel strike associated impact loads on joints need to be compensated by active muscular effort and safety shoes should support this protective function of muscle activation. In 10 healthy men, 12 trunk and leg muscles were analyzed with surface electromyography. Subjects walked on a walkway while wearing different safety shoes with the test shoes being equipped with exchangeable cushioning heel inserts according to individuals' body weight. While wearing the optimally cushioned shoes the cumulative muscle activity per distance travelled dropped clearly compared to the regular safety shoes, demonstrating reduced muscular effort. Also, the heel strike associated amplitude peak of back muscles occurred earlier within the stride while wearing the test shoes. Thus weight-balanced cushioning heel inserts in safety shoes proved able to reduce muscle strain, logically delaying muscular fatigue and extending muscular joint protection.

Relevance to industry: Adjustable heel inserts in safety shoes are suited to improve the health status of employees by reducing muscular effort so that active joint protection can be prolonged.

1. Introduction

Walking is the natural way of locomotion for humans. However, today we face environmental conditions that differ fundamentally from those of our ancestors: they travelled barefoot or practically so over sand, grass, or through forests. These grounds are elastic and can therefore absorb impacts. Today we mainly tread on firm surfaces wearing shoes and a greater proportion of the impact forces from walking need to be compensated by our bodies.

The human body offers several shock absorbing structures to protect bones, joints, and brain from the forces produced by every heel contact (Hoshino and Wallace, 1987). Besides the construction of our feet and spine, muscles provide the most important element that absorbs impact loads on joints during locomotion (Voloshin and Wosk, 1982; Wee and Voloshin, 2013). To accomplish this requirement a just-in-time activation during stride is of great importance (Lamoth et al., 2004). However, continuous long-lasting activity leads to an increase in muscular strain as the body tries to maintain adequate load compensation. Sustained effort bears the danger of fatiguing the muscles and putting the musculoskeletal system at risk for injuries and disorders (Folman et al., 1986; Garner et al. 2013, Lee et al., 2001).

Attempts to reduce the impact forces of walking, thereby lessening muscle strain, have involved two principal approaches: The first involves the use of anti-fatiguing mats on the floor to change elastic ground characteristics (Cham and Redfern, 2001). However, these mats have fixed elastic properties which may not suit every demand. Also frequent turning leads to torque loads on the joints limiting anti-fatiguing mats to situations where employees stand more or less still for long periods.

A second approach aims at reducing impact forces by improving the elasticity of the shoe's foot bed (Hansen et al., 1998), restricting movements much less. In this case improving shock absorption can be as simple and inexpensive as inserting cushioning elements in

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Non-standard abbreviations: RA, M. rectus abdominis; OE, M. obliquus externus abdominis; OI, M. obliquus internus abdominis; LO, M. erector spinae (longissimus); ICO, M. erector spinae (iliocostalis); MF, M. multifidus (lumbar part); VL, M. vastus lateralis; VM, M. vastus medialis; TA, M. tibialis anterior; BF, M. biceps femoris; ST, M. semitendinosus / semimembranosus; GC, M. gastrocnemius; CMAPD, cumulative muscle activity per distance.

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the heel area of shoes (Wosk and Voloshin, 1985). The shoe construction and safety are not affected, i.e. the firmness of the safety shoe's sole and shaft is not altered, but impact forces within the body are reduced. In many occupations safety shoes have to be worn as a mandatory safety obligation. Combining safety and optimized shock absorbing capacity offers an ideal solution to reduce fatigue. This would provide greater physical comfort on firm ground and reduce the development of work-related musculoskeletal disorders.

In developing safety shoes researchers must consider a wide variability in foot anatomy including adjusting for foot length to width ratio (Hawes and Sovak, 1994), sex (Luo et al., 2009), body weight and age (Scott et al., 2007) of wearer. Many manufacturers already account for this variability by providing multiple-width systems along with models for men or women.

In addition, if cushioning elements were exchangeable, the shock absorption would be adjustable to the individual's weight.

In this study we evaluated the effect of the cushioning elements in these safety shoes on leg and trunk muscle activation characteristics by applying surface electromyography (SEMG). Global parameters were derived that are related to muscle economy.

2. Materials and methods

2.1. Subjects

For this study, 10 healthy men aged between 25 and 48 years (median = 38.5) were investigated. Their health status was verified by physical examination and self-reported history. All subjects were free of back pain and had no history of any back injury. The study followed the ethics requirements for human investigations and was approved by the local ethics committee (3352-01/12). Prior to investigation subjects signed written informed consent. Every investigation was carried out by the same examiners to ensure reliability, as of course was the case for positioning the electrodes. Subjects were employed at a canteen kitchen and therefore obliged to wear safety shoes the entire working day.

2.2. Experimental design

The safety shoes previously worn by the subjects (in the following referred to as control shoes) were equal models of one footwear manufacturer. The test shoe was a safety shoe designed to house an exchangeable cushioning element in the heel area of the inner shoe sole (VX 5 Perbunan, Steitz Secura, S2 category according to German safety shoe specifications). The investigation took place on a walkway that allowed subjects to walk freely over a distance of about 20 m. Heel strike was detected through pressure sensor foils (CZN-CP42, IEE) fixed underneath the shoe heel. Prior to investigation the appropriate cushioning element for each subject was determined through body weight measurements. For the test shoe three different cushioning degrees were investigated: no cushioning (dummy insert), optimal (recommended), and too soft cushioning.

The cushioning elements are grouped into four recommended body weight categories provided by the manufacturer: <57 kg, 58–79 kg, 80–91 kg, and >91 kg. The optimal cushioning element was determined individually according to the provided body weight categories. Then based on this, the too soft cushioning was one increment lighter weight category as recommended for that individual. Required size and width of the shoes were determined using a sizing kit provided by the manufacturer.

For every investigation three walking speeds needed to be completed: preferred walking speed (in the following referred to as preferred), consciously slow (slow), and brisk (fast) walking speed. For the calculation of walking speeds stride lengths were always measured. Subjects were investigated four times (Fig. 1).

The initial measurement was carried out in the control shoes, so far worn by the subjects, to establish baseline SEMG information. Afterwards all subjects were equipped with the test shoes without cushioning (dummy cushioning element). Subsequent to this first investigation subjects had worn the test shoes to accomplish habituation. Except for the second investigation, for which as a precaution the habituation period was extended to five working days to break in the new shoes, two working days in a row were given for habituation. All investigations were executed at the end of the working day. The order of the two remaining cushioning elements (optimal and too soft) was randomized beforehand but evenly distributed between subjects. After each investigation, cushioning elements were exchanged for the next ones scheduled and habituation was started again (Fig. 1). All measurements took place at the subjects' place of work.

2.3. SEMG data acquisition and analysis

For SEMG measurements disposable Ag–AgCl electrodes (H93SG, Covidien, Germany) with a circular uptake area of 1.6 cm diameter were used. SEMG was amplified (gain: 1000, Biovision, Germany) using a bipolar montage (inter-electrode distance: 2.5 cm). The signals were analogue to digital converted at a rate of 2048/s (Tower of Measurement, DeMeTec, Germany; anti-aliasing filter: 1024 Hz, resolution: 24 bit, software: GJB, Germany). SEMG was taken from 12 muscles of legs and trunk, as shown in Fig. 2. Electrode positions were chosen in accordance with the SENIAM recommendations (Hermens et al., 1999), and, if not available for the respective muscle, according to Ng et al. (Ng et al., 1998). For reasons of clarity investigated muscles were grouped into abdominal muscles (RA, OE, OI), back muscles (LO, ICO, MF), and leg muscles (VL, VM, TA, BF, ST, GC).

Electrocardiographical activation (ECG) was also measured by placing electrodes along the heart axis. Since ECG-artifacts appear in almost every SEMG recording, its additional recording serves the possibility of separating it from the muscle signal in further data analysis. Throughout the investigation signals were monitored permanently and detached electrodes replaced if necessary.

Data was stored on computer for offline analysis. All data was processed equally: elimination of DC components, high-pass filter at 65 Hz, low-pass filter at 400 Hz, and 50 Hz notch filter. Elimination of ECG-artifacts was accomplished through a template function (Mörl et al., 2010). Roughly explained, all R-waves were identified, individual templates for every channel calculated and subtracted from the original signals by applying a cosine weighting function to avoid non zero steps in a time window of ±100 ms for all detected events (Anders et al., 1991). Heel strikes were detected by pressure sensors, median intervals calculated, and only full strides varying less than 10% from this value were included in the calculations. All valid strides were time normalized to 100% and were used for the calculation of grand averaged amplitude curves for every single SEMG channel, measurement and subject, respectively. The time normalized curves were calculated with an accuracy of 0.5% (201 single values). These values were included in the analysis and will be referred to as time-dependent parameters. From the grand averaged curves time-independent parameters were calculated: mean values and the minimum to maximum range normalized by mean. High range values indicate a large span between exertion and relaxation that improves blood flow into muscles, thus reducing the likelihood of fatigue.

Furthermore the cumulative muscle activity per distance travelled (CMAPD) was calculated as an indirect measure for the required energy expenditure to travel a given distance (Carrier