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Journal of Biomechanics 35 (2002) 621–628

JOURNAL OF BIOMECHANICS

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Effects of fatigue and load variation on metatarsal deformation measured in vivo during barefoot walking

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Accepted 28 November 2001

Abstract

This in vivo study presents information to assist in the understanding of metatarsal stress fracture etiology. The aims were (a) to provide a fundamental description of loading patterns of the second metatarsal (MTII) during barefoot walking, and (b) to investigate the hypothesis that MTII dorsal strain increases with fatigue and external carrying load. Dorsal MTII strain was measured in vivo under local anaesthetic with an instrumented staple in eight subjects. Experimental conditions were external loading with a 20 kg backpack and pre- and post-fatigue. *M. flexor digitorum longus* electromyography tentatively indicated fatigue after an extended walking treatment. A reproducible, cyclic temporal pattern of dorsal MTII surface deformation was described. Mean peak compression and tension strains in unloaded barefoot walking were -1534 ± 636 and $363 \pm 359\,\mu\varepsilon$, respectively. Mean peak compression strain rate (SR) was $-4165 \pm 1233\,\mu\varepsilon/s$. Compression strain increased significantly ($\alpha = 0.05$) both with the addition of the backpack and post-fatigue while maximum tension decreased significantly post-fatigue. SR increased significantly with the addition of the backpack. The highest plantar force time integrals were recorded underneath the heads of metatarsals II–V for all conditions (1561 N s pre-fatigue, without backpack; 2123 N s post, with). EMG and plantar pressure data presented a comprehensive description of biomechanical parameters influencing dorsal MTII deformation and alterations in strain following two experimental conditions were suggested as contributing factors in the pathogenesis of metatarsal stress fractures.

Keywords: Second metatarsal; Stress fracture; Local bone deformation; In vivo biomechanics

1. Introduction

Stress fractures are fractures that occur in the absence of a specific acute precipitating event (Buckwalter and Brandser, 1997) and are commonly reported in normal bones of young adults during periods of increased physical activity such as military or sports training. In a prospective study of 295 soldiers Milgrom et al. (1985) reported 184 stress fractures of the lower extremity of which 14 (7.6%) were of the metatarsals. A study of 169 stress fractures by Ha et al. (1991) showed 12 located at the second metatarsal (MTII). Up to 20% (11% MTII) of stress fractures in athletes and 23% in military recruits are located in the metatarsals (McBryde, 1985; Matheson et al., 1987; Hulkko, 1988). One hypothesis describing the occurrence of stress fractures is that insufficient remodeling of stressed bone results in fatigue related trabecular microfractures (Carter et al., 1981; McBryde, 1985; Fazzalari, 1993; Buckwalter and Brandser, 1997). Stress fractures may be initiated by excessive repetitive muscular forces (Stanitski et al., 1978; Orava et al., 1995).

Local bone deformation is regarded as an indicator of stress fracture risk and the measurement of local bone deformation using strain gauges has been performed for many years (Fries, 1972). This method is, however, not free of difficulties. There are disadvantages involved in the in vivo application of strain gauges directly onto osseous surfaces. It is difficult to provide a completely dry, chemically clean bone surface for the strain gauge bond and the effects of the bonding between the bone and the strain gauge are unknown and uncontrolled.

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0021-9290/02/$ - see front matter
PII: S0021-9290(01)00241-X

Until recently cyanoacrylate adhesives were used for strain gauge application to the bone surface but this is not approved for human experimentation due to carcinogenic risks. Good results were reported by Hoshaw et al. (1997) for polymethylmethacrylate (PMMA) bonding, which is now being applied in human in vivo experiments. This technique is, however, quite invasive as the periosteum must be removed for strain gauge application and the bone is 'roughened' with a metal punch for optimal bonding surface preparation.

The insertion of instrumented staples involves less traumatic surgery and is thus more appropriate for human in vivo application. Strain on the human tibia has previously been measured in vivo with rosette strain gauges (Lanyon et al., 1975), transcutaneous extensometers (Fyhrie et al., 1998) and inserted staples (Burr et al., 1996; Ekenman et al., 1998a). Excellent linearity and good long-term results have been reported in measurements of in vivo loading of the canine lumbar spine with instrumented staples (Butterman et al., 1994). No such investigation has, however, been performed on metatarsals or bones of corresponding size and previous values for human metatarsal loading have been limited to theoretical biomechanical models (Stokes et al., 1979; Gross and Bunch, 1989) and in vitro investigations of human cadaver specimens (Lease and Evans, 1959; Sharkey et al., 1995; Courtney et al., 1997; Donahue and Sharkey, 1999).

The MTII is subject to disproportionally high external loads relative to its physical dimensions. Donahue and Sharkey (1999) measured metatarsal deformation on a cadaver lower leg model simulating walking and found the mean peak dorsal strain to be $-1897\,\mu\varepsilon$, which was significantly higher than strain measured on the fifth metatarsal. A dorsal compressive strain of $-6662\,\mu\varepsilon$ was calculated by Gross and Bunch (1989) for the MTII which exceeded that of the first metatarsal by a factor of 6.9. Excessive bending moments resulting from inactivity or fatigue of the toe flexor muscles are commonly mentioned in the etiology of metatarsal overuse injuries (Tanaka, 1981; Sharkey et al., 1995; Voloshin et al., 1998; Donahue and Sharkey, 1999). Increased local bone deformation resulting from lacking soft tissue support can lead to forces responsible for stress fractures, metatarsalgia and flattening of the longitudinal arch of the foot. In military recruits vigorous marches at an early stage of military service provide the prerequisites for both repetitive external loading and muscular fatigue, which may explain the exceptionally high rate of stress fractures. In this study in vivo measurement of metatarsal strains was utilized to address these factors by comparing two experimental treatments: with and without external loading with a 20 kg backpack and the effect of fatigue.

2. Methods

2.1. Staple mounted strain gauge

The application of the staple technique to bones approximating the dimensions of MTII had been previously tested in an in vitro pilot study using chicken tibiae (Arndt et al., 1999). The results showed a linearity of strain measurements relative to a controlled input force of $R^2 = 0.992 \pm 0.006$ and an intraclass correlation coefficient of 0.92.

Two strain gauges (types EA-06-031DE-350 and EA-06-031EC-350, Measurement Group Inc., USA) were mounted perpendicular to each other on the underneath bridge surface of a 16 mm × 15 mm titanium staple with a staple bridge thickness of 1.0 mm (3M, St. Paul, USA). This configuration facilitated the registration of uniaxial strain in the longitudinal direction of the bone. After local anaesthetic (Marcain®, AstraZeneca, Sweden) an incision was made over the dorsal aspect of the MTII and the bone surface exposed with periosteum intact. A guide was used to drill two holes through the cortex and the staple was inserted in the pre-drilled holes with the aid of an insertion tool providing a pretensile staple leg distention of 0.4 mm (Ekenman et al., 1998b). Staples were inserted to a depth of 4 mm ensuring the legs passed completely through the cortex (mean thickness: 2.7 mm; from Gross and Bunch, 1989). Strain gauge signal quality was controlled in the operating theater prior to the beginning of the experimental protocol. A lateral X-ray view of an inserted staple is illustrated in Fig. 1.

After completion of the experiment (maximum of 3 h after staple insertion) the staple was removed and the incision sutured. Subjects experienced initial swelling and tenderness the following day but no further complications were reported. The signal was amplified by a strain gauge conditioner (2120A, Measurements Group Inc., USA), sampled at 1000 Hz and recorded

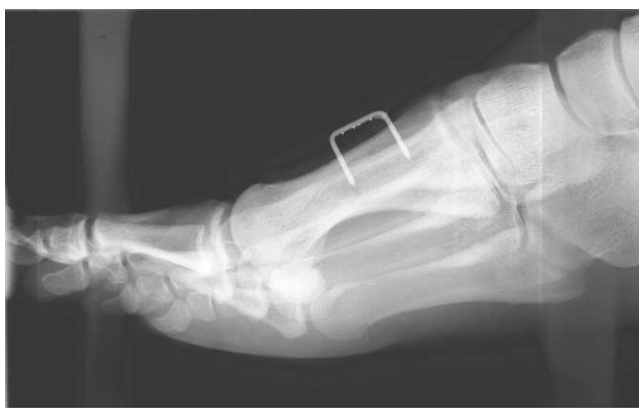

Fig. 1. Medial X-ray view of instrumented staple inserted in the second metatarsal.

with standard analog data collection software (Bioware, Kistler, Switzerland). Transformation of the signal measured from the staple strain gauge to a value representing the strain upon the dorsal cortical metatarsal surface proceeded as follows:

The strain measured by the staple mounted strain gauge was

$$\varepsilon s(t) = (x(t)/1.6) \times 1000\,\mu\varepsilon$$

and the strain at the bone was

$$\varepsilon b(t) = \varepsilon s(t) \quad k = (x(t)/1.6)\,1000 \times 8.96 = 5600\,x(t)\,\mu\varepsilon,$$

where x (V) was the sampled signal at a certain point in time (t), 1.6 was the preset gain (i.e. 1.6 V = 1000 με), k (= 8.96) was the calibration factor derived from the in vitro pilot study by Arndt et al. (1999).

The raw staple strain gauge signal was therefore, multiplied by 5600 to approximate the strain acting upon the MTII cortical surface. The signal was filtered with a digital 4 Hz low pass, dual pass Butterworth filter prior to data analysis. The cutoff frequency was chosen by visual examination of the data signal as the optimal frequency for preserving signal shape while removing obvious high frequency noise. Strain rate (SR) was calculated as the slope of the signal from each maximum tension peak to the subsequent maximum compression peak (see Fig. 2). Throughout the experimental procedure subjects were regularly instructed to unload the instrumented foot with no conscious contraction of intrinsic or extrinsic foot musculature and the strain gauge voltage output was set to 0 mV in this condition. This provided the baseline measurement for the strain gauge.

2.2. Electromyography

Intramuscular wire electrode electromyography (EMG) was conducted to record muscular activity of *M. flexor digitorum longus* (FDL). Two silver electrodes (AG7/40 T, Medwire, USA) were inserted in the muscle under local anaesthetic postero-medial to the tibia ≈10 cm proximal from the medial malleolus. The ground surface electrode was positioned over the fibular head. EMG signals were recorded telemetrically (Telemyo, Noraxon, Finland) and sampled at 1000 Hz synchronous to the strain gauge signal. EMG signal amplitude was analyzed by the subjects performing initial maximum voluntary toe flexion contractions (MVC) and subsequent normalization of 200 ms linear envelope peaks (% MVC). The frequency content of the signal was analyzed by determining the mean power frequency (MPF) of the 9 Hz high pass filtered signal with a fast Fourier transform.

2.3. Experimental protocol

Eight male subjects (age: 45 ± 10 years, weight: 78 ± 13 kg) with no obvious foot abnormalities and no clinical symptoms participated in the study. All walking trials were conducted on a level treadmill operating at 3 km/h. Each subject performed three trials of each experimental condition and the duration of each trial was 15 s to permit the registration of multiple step cycles (range: 10–13). Two different variables were tested: the effect of additional external loading with a 20 kg backpack and walking wearing the backpack to test for fatigue effects. This resulted in four experimental conditions (pre-fatigue without backpack, pre with, post

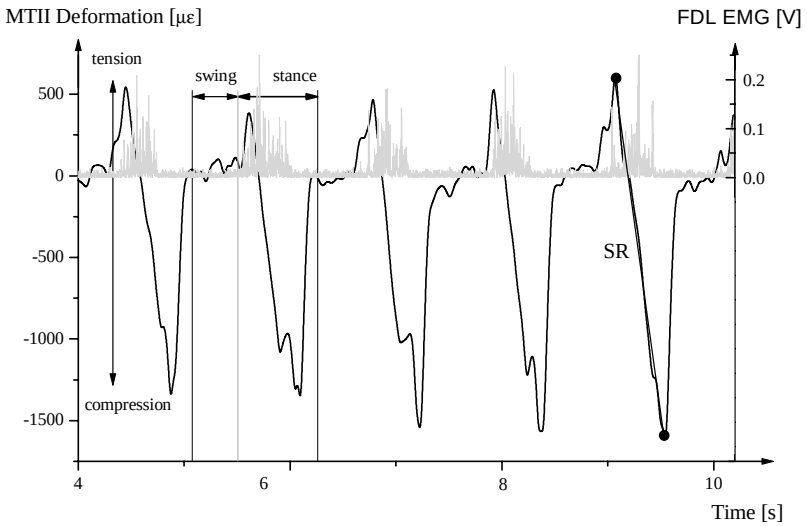

Fig. 2. Illustration of cyclic pattern of local second metatarsal deformation and FDL activity during barefoot walking. A 6 s excerpt of a no backpack, pre-fatigue trial illustrating swing and stance phases. The black line represents deformation and the gray line FDL activity. SR indicates the gradient chosen for defining the SR from tension to compression.

without, post with). Subjects walked until voluntary fatigue, however, a maximum of 1 h due to time restrictions imposed by the invasive methodology.

Allocation of data to the temporal reference points heel touch-down (TD) and toe-off (TO) occurred via the FScan (Tekscan, USA) plantar pressure distribution system for the first five subjects and the Pedar system (Novel Gmbh, Germany) for the remaining three. Due to problems encountered over the extended time period for the first five subjects, quantified plantar pressure results were only presented for the last three subjects. The measuring sole was firmly attached to the plantar surface of the staple instrumented foot with a tight fitting compression sock. Pressure distribution data were sampled at 50 Hz and synchronized with the strain gauge and EMG signals by an analog pulse. TD was defined as the frame after swing phase in which the first heel sensor was active and TO the first frame after stance in which no sensor was active. Maximum force and force time integrals were determined underneath the heads of metatarsals II–V to describe possible direct effects upon dorsal MTII strain. However, further plantar regions (hallux, toes, head of metatarsal I, midfoot and heel) were also analyzed to provide an overall description of pressure variations relative to experimental condition.

The effect of external load was studied in four of the eight subjects. There was no strain gauge signal of one subject after the walking treatment and seven subjects were therefore, included in the analysis of the effect of fatigue. The parameters of interest were averaged over all steps for the relevant subjects for the four experimental conditions, permitting the determination of differences between pre- and post-fatigue both with and without the backpack. A one-way analysis of variance (ANOVA) with post-hoc Tukey–Kramer analysis was conducted for each parameter to identify statistical differences between experimental conditions. The significance level was set at $\alpha = 0.05$ throughout the study. No statistical analysis was conducted on the quantified plantar pressure data as this was only available for three subjects.

Ethical approval for this study was acquired from the Huddinge University Hospital Research Ethical Committee (approval #392/97). The subjects were informed of the procedures and risks involved prior to participation.

3. Results

3.1. Temporal description of MTII deformation

A repetitive cyclic pattern of local MTII deformation was seen (Fig. 2). The following fundamental temporal pattern describes the mean ± standard deviation of all subjects for barefoot, pre-fatigue, no backpack treadmill walking. TD was followed by peak tension, which occurred at 8 ± 7% stance phase followed by a transition to maximum compression during the majority of stance phase with peak compression at 65 ± 15%. Immediately after peak compression local deformation decreased to $\approx 0\,\mu\varepsilon$ around which level it fluctuated during swing phase. Mean peak dorsal strains were $-1534 \pm 636\,\mu\varepsilon$ in compression and $376 \pm 359\,\mu\varepsilon$ in tension. The mean SR of the slope to compression peak was $-4165 \pm 1233\,\mu\varepsilon/s$.

FDL activation began approximately at the same time as tension deformation of the dorsal metatarsal surface following TD with peak activation occurring at 48 ± 11% of the stance phase during the transition to compression deformation. In addition to the temporal description of activity of a flexor inserting in the second metatarsal, the EMG results were also used to ascertain whether fatigue occurred. An increase in EMG signal amplitude and a corresponding decrease in the frequency spectrum were regarded as potential indicators of muscular fatigue. The increase in the amplitude of FDL activity following the walking treatment both without and with the external load (95 ± 37% MVC to 114 ± 98% and 86 ± 50% to 98 ± 55%, respectively) indicated fatiguing although the differences were not statistically significant. Furthermore, the frequency spectrum (MPF) showed a slight decrease post-fatigue with the backpack but a significant increase without. For the purposes of this study the plantar musculature was tentatively regarded as fatigued subsequent to the walking treatment.

3.2. Effects of external load and fatigue

Mean values for deformation, EMG and timing parameters for the four experimental conditions are presented in Table 1 and the corresponding results for the plantar pressure parameters maximum force and force time integral follow in Fig. 3. Peak compression pre-fatigue without backpack was significantly less than for all other conditions, peak tension decreased significantly post-fatigue and the lowest mean peak tension was recorded post-fatigue with backpack ($142 \pm 172\,\mu\varepsilon$). SR increased significantly with addition of the backpack but not post-fatigue. The time of peak compression was significantly later in the stance phase for post-fatigue without backpack (72 ± 8%) than for any other condition, while peak tension occurred significantly earlier post without than pre without. Peak FDL activity occurred later post-fatigue, although this difference was only significant wearing the backpack.

Analysis of the maximal force and force time integral underneath the heads of metatarsals II–V showed increases in both parameters both after fatigue and with the addition of external load.

Table 1
Results of experimental conditions; means ± SD

	1 Pre without	2 Post without	3 Pre with	4 Post with
EMG %MVC	95 ± 37	114 ± 98	86 ± 50[2]	98 ± 55
EMG MPF (Hz)	91 ± 17	100 ± 40[1]	96 ± 12	94 ± 12
comp. (με)	−1534 ± 636	−2166 ± 935[1]	−2187 ± 840[1]	−2327 ± 1173[1]
Tension (με)	376 ± 359	174 ± 64[1]	435 ± 318[2]	142 ± 172[1,3]
SR (με/s)	−4165 ± 1233	−4655 ± 1959	−5458 ± 1554[1,2]	−5030 ± 2424[1]
*T*tension (% stance)	8 ± 7	3 ± 3[1]	7 ± 7[2]	7 ± 7[2]
*t*FDL (% stance)	48 ± 11	51 ± 14	47 ± 10	54 ± 12[1,3]
*t*comp. (% stance)	65 ± 15	67 ± 7	66 ± 10	72 ± 8[1,2,3]

Superscripts indicate significant differences ($\alpha = 0.05$) between conditions. Times of occurrence (% stance phase) of peak compression, tension and FDL activity are denoted as *t*comp, *t*tension, *t*FDL, respectively. Means were calculated from all steps of all available trials.

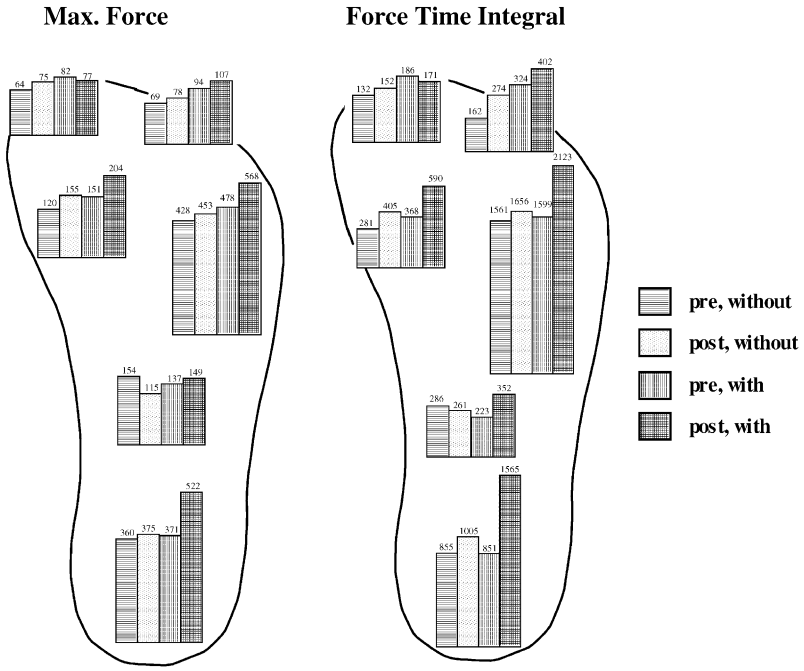

Fig. 3. Pressure distribution results: (A) maximum force, (B) force time integral. Illustrated masks represent the regions: hallux, toes, head of 1st metatarsal, metatarsal heads II–V, midfoot and heel. NB: Intergraph scaling is approximate.

4. Discussion

There are obvious limitations involved when measuring strain at a single location in a single direction on a bone surface and attempting to apply the results to general behaviour of the bone during loading as different strains can of course be acting at other locations and in other directions. The dorsal MTII surface was, however, chosen as it is a relevant site in the discussion of stress fractures and was easily accessible for the in vivo technique used. No limitations in terms of complications of the invasive technique were encountered.

Peak compression strain measured at cortical sites in a variety of mammalian and avian species during strenuous activity ranges from −1700 to −5200 με (Gross et al., 1992; Biewener, 1993; Burr et al., 1996) and the results in this study for normal walking in humans were within this range (peak mean for any condition: −2327 ± 1173 με). The results also corresponded well with some in vitro measurements and biomechanical models describing MTII loading (Stokes et al., 1979; Sharkey et al., 1995; Donahue and Sharkey, 1999). Due to the fact that their cadaver model could not simulate the stance roll-over process in realistic time, Donahue and Sharkey (1999) calculated that their in vitro strains overestimated in vivo values by ≈15%. The resulting −1612 με would correspond well with the pre-fatigue, no backpack state in this study (−1534 με) and the significant dorsal compression increase with severe

fatigue reported by Donahue and Sharkey (1999) also corresponded well with the post-fatigue in vivo results reported here. A biomechanical model of distance running by Gross and Bunch (1989) predicted considerably greater compression strains ($-6662\,\mu\varepsilon$). Compact bone fails in longitudinal compression at strains of -14000 to $-21000\,\mu\varepsilon$ and begins to yield at strains between $-6000\,\mu\varepsilon$ and $-8000\,\mu\varepsilon$ (Biewener, 1993). The MTII compression strain measured in this study was therefore, $\approx$3–4 times less than a predicted yield strain.

The EMG results did not provide significant changes in FDL activity as objective evidence of fatigue. The walking treatment was regarded at face value as a cyclic, walking loading of the foot for a maximum of 1 h wearing a 20 kg backpack and some fatigue of the plantar musculature was assumed to have occurred. Compression strains increased with both the experimental conditions fatigue and external load. The similarity of the compression strain increases (both significant) between pre-fatigue without and both pre-fatigue with external load and post-fatigue without (from -1534 ± 636 to $-2187\pm840\,\mu\varepsilon$ and $-2166\pm840\,\mu\varepsilon$, respectively) indicated that the moderate fatigue induced in this study was sufficient to increase local bone deformation to the same extent as a 20 kg increase in external carrying load.

Mechanical fatigue damage accumulates more rapidly in compressive areas than tensile areas of bone and fatigue failure has been predicted to occur after as few as 100,000 cycles of $\approx3000\,\mu\varepsilon$ (Carter et al., 1981). The greater compressive strain measured indicated that the dorsal aspect of the metatarsal is a likely site of such accumulated damage although its accompanying cyclic tensile loading may cause tension failure if this is sufficiently high. A mean peak tension strain of $435\pm318\,\mu\varepsilon$ was measured in this study. The staple position on the diaphysis of MTII corresponds to the primary location of stress fractures (Brudvig et al., 1983; McBryde, 1985) although McBryde (1985) stated that the origin of MTII stress fractures is most frequently on the medial aspect. Maintenance of soft tissue integrity rendered the staple method unsuitable for in vivo measurement of MTII medial aspect strains. No significant differences were recorded in tension strains with the addition of the backpack although these decreased post-fatigue both with and without the backpack. As constant stress causes damage accumulation in compact bone (Caler and Carter, 1989; Mauch, 1992), lower tension deformation may accentuate the corresponding compression increase by increasing the average strain as suggested by Donahue and Sharkey (1999). A hypothesis contributing to the understanding of stress fracture etiology emerged from the results of this study. In the non-fatigued state the bone experiences a cyclic 'tension unloading' between the compression peaks, however, when the foot is fatigued peak compression strain not only increases but this tensile unloading mechanism decreases. Cellular surface shear forces related to fluid flow resulting from surface deformation may be temporarily relieved by the intermittent tension deformation in the pre-fatigue state. Post-fatigue such shear forces could dominate in one direction causing excessive cellular stresses and subsequently osseous microdamage.

Various studies have demonstrated that inactivity or fatigue of supporting musculature will increase bone deformation and relaxation of plantar structures during late stance will increase bending stress. In vitro second metatarsal compression strain is significantly less if all plantar muscles are under tension compared to only the achilles tendon (Tanaka, 1981; Sharkey et al., 1995; Donahue and Sharkey, 1999). FDL activity was analyzed due to its assumed role in assisting the other auxiliary plantar flexors in maintaining normal metatarsal strains during gait. The precise nature and extent of this role was an important aspect of this investigation. Fig. 2 indicated that FDL activity began prior to dorsal tension immediately after TD. Fatigue and presumably decreased FDL tension corresponded to lower peak tension strains. Peak tension subsequent to TD was regarded as a downward forefoot response of the vertical impact force acting underneath the heel. Dorsiflexion muscle forces at TD must control this inertial mechanism and minimize dorsal tension, while the measured FDL activity would counteract this effect. FDL fatigue would therefore, entail a decrease in such antagonistic activity—a theory which would explain the significantly reduced tension post-fatigue. Extrapolation of this result to more extreme plantar muscle fatigue may imply that MTII is loaded continually under compression throughout the stance phase—including the impact phase. Furthermore, the majority of FDL activity during the transition to peak compression indicated that muscular fatigue could well facilitate the increased compression deformation measured post walking treatment. A suggested mechanism for stress fracture is thus an incessant compression loading resulting in accumulated trabecular microdamage in combination with an increase in the amplitude of compression strain due to fatigue.

The unique combination of the EMG and plantar pressure results with simultaneous in vivo measurement of bone deformation supported the theory that a high force time integral subjects bones to stress fracture risk (Fuller, 1996). Both the maximum force (568 N) and force time integrals (2123 N s) underneath the heads of metatarsals II–V were higher than in any other defined mask and increased with the addition of external load and post-fatigue. Although it appears logical that greater force underneath the metatarsal heads increased dorsal compressive surface deformation, this somewhat contradicts the muscle fatigue hypothesis. If FDL

tension decreased, thus permitting greater dorsal excursion of the distal metatarsal, this would appear to have resulted in decreased plantar force underneath this region, although the force time integral may still have increased due to the force being applied over a longer time period. Conclusive data describing plantar pressure underneath MTII and its possible interaction with dorsal strains requires the investigation of more extreme fatigue treatment than that applied here.

The effect of footwear on metatarsal deformation was not addressed in this study. Footwear affects both the mechanical support of the foot and also the rate of fatigue of foot musculature. It would thus not only directly influence dorsal strain magnitudes, but also the role of fatigue. Footwear is therefore, an important factor when considering the predominance of MTII stress fractures in long-distance athletes and military recruits and will be addressed in future research.

5. Conclusions

Maximum strains measured on the human second metatarsal during walking were in the range described in the literature for other bones during physiological loading. The direct in vivo method here described provided concrete information on the biomechanical loading of the human foot during gait.

The fatigue implications raised by the results of this study are treated with caution as the extent of fatigue experienced by long-distance athletes or military recruits greatly exceeds that produced in this study. Furthermore, military recruits frequently carry greater loads. The results of this study confirmed the hypotheses that an increase in external loading and fatigue of plantar musculature will influence dorsal MTII strain in that both conditions resulted in increased dorsal MTII compression strain while tension strains decreased. These effects of the two chosen experimental conditions are suggested as contributing factors in the etiology of stress fractures of the human second metatarsal.

Acknowledgements

Financial support for this study was provided by the Swedish Defense Material Administration and the Karolinska Institute, Sweden.

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