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The potential influence of the heel counter on internal stress during static standing: A combined finite element and positional MRI investigation

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Abstract

Confinement of the heel due to the counter of the shoe is believed to influence heel pad biomechanics. Using a two-dimensional finite element model of the heel pad and shoe during a simulation of static standing, the aim of this study was to quantify the potential effect of confinement on internal heel pad stress. Non-weightbearing MRI and weightbearing MRI with plantar pressure and ground reaction force data were recorded for a single subject. The non-weightbearing MRI was used to create two FE models of the heel pad, using either homogeneous or composite material properties. The composite model included a distinction in material properties between fat pad and skin. Vertical and medial–lateral forces, as measured on the subject's heel, were applied to the models and vertical compressive strains for both models were comparable with those observed by weightbearing MRI. However, only for the composite model was the predicted plantar pressure distribution comparable with measured data. The composite model was therefore used in further analyses. In this composite model, the internal stresses were located mainly in the skin and were predominantly tensile in nature, whereas the stress state in the fat pad approached hydrostatic conditions. A representation of a running shoe, including an insole, midsole and heel counter was then added to the composite heel pad to form the shod model. In order to investigate the counter effect, the load was applied to the shod model with and without the heel counter. The effect of the counter on peak stress was to elevate compression (0–50%), reduce tension (22–34%) and reduce shear (22–28%) in the skin. In addition, the counter reduced both compressive (20–40%) and shear (58–80%) stress in the fat pad and tension in the fat pad remained negligible. Taken together the results indicate that a well-fitted counter works in sympathy with the internal structure of the heel pad and could be an effective reducer of heel pad stress. However, further research needs to be undertaken to assess the long-term effects on the soft-tissues, practicalities of achieving good fit and behavior under dynamic events.

Keywords: Soft tissues; Footwear; Shod; Barefoot; Plantar pressure

1. Introduction

The heel pad consists of the fat pad surrounded by a thick subdermal layer of fibrous tissue embedded in a layer of skin (Jahss et al., 1992). Tissue changes resulting from an overload or pathology can render the healthy heel susceptible to biomechanical trauma. This trauma manifests itself in the form of a variety of pathological heel conditions (e.g. plantar fasciitis, subcalcaneal pain and soft tissue disruption) and can lead to a reduction in the pad's force dissipation characteristics (Tong et al., 2003). The latter has been linked to longer-term problems (e.g. degenerative joint disease, prosthetic joint loosening (e.g. Rome, 1997).

The use of appropriate footwear may aid in the prevention of these conditions. Modern athletic shoes

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employ a sole comprising a midsole/insole and counter to cushion and support the heel. When studying the effectiveness of shoe design on internal foot mechanics, researchers have adopted finite element stress analysis. Based on these analyses, it was found that sole cushioning increases heel–sole contact area (Verdejo and Mills, 2004) and speculated that an optimum midsole/insole stiffness ratio exists which minimizes stress within the soft tissue (Nakamura et al., 1981). However, the models to date have not yet investigated the effects of the heel counter.

The heel counter is a semi-rigid cup designed to fit around the heel. While its primary function is to restrict overpronation during gait, the resulting confinement of the heel (known as the counter effect) reduces the severity of external loads acting on the heel (Jorgenson and Ekstrand, 1988). Furthermore, cineradiographic evidence (de Clercq et al., 1994) reveals that the heel pad undergoes reduced compressive strain during shod when compared to barefoot running. This reduced strain was attributed to the counter and subsequently led to speculations that the counter, by ensuring that a critical heel pad thickness is maintained, has the potential to reduce internal stress during shod locomotion (Tong et al., 2003).

The aim of this investigation is to quantify the influence of confinement on heel pad stress.

2. Material and methods

A healthy male volunteer with no history of heel pain volunteered for the study (age at time of study = 36 yr, height 183 cm, weight 95 kg, shoe size 12US). Scans were taken of the right foot using a positional MRI scanner (0.6 T FONAR Positional Open Scanner; Jorgensen et al., 2005) based at the University of Aberdeen. Images of the foot with 256×256 pixels (resolution = 1 mm) and slice thickness of 7 mm were found to be optimal in terms of quality and scanning time. Scans were taken (TR = 3262 and TE = 140) throughout the entire foot with each slice being oriented perpendicularly to the transect joining the 2nd metatarsal and the medial calcaneal tuberosity. Scans in non-weightbearing (Fig. 1a) and weightbearing positions for both unshod (Fig. 1b) and shod (Fig. 1c) conditions were made and in all cases, the subject's feet were positioned 60 cm apart (i.e. approximately shoulder width). Athletic shoes were used (Silver Shadow, Hi Tec, UK). During the weightbearing scans, the vertical compressive strain in the heel pad (change in maximum heel pad thickness/unloaded maximum heel pad thickness) was measured for the barefoot (0.25) and shod (0.21) conditions.

The subject then remained stationary and plantar pressures were recorded using calibrated insoles (Footscan, Rsscan Lab Ltd., Suffolk, UK) over a 10 s standing period for barefoot conditions. The insole sensor dimensions were 8 mm (anterior–posterior) and 6 mm (medial–lateral). The resulting pressures were segmented into anatomical regions and data in the heel region were extracted for further analysis (Fig. 3). The summed force (52 N) and pressure distribution across the medial–lateral row of insole sensors in the location of maximal heel pressure was recorded and used for the application of load to the models and for validating the models (Fig. 3). The subject was then moved to the biomechanics laboratory at the University of Teesside and stood with the right foot on a Kistler force platform (Kistler, Winterthur, Switzerland) and the left foot on the ground thus allowing measurement of the medial–lateral force (35 N) on the right foot during standing.

The pixels of the non-weightbearing scans in the region of the medial and lateral calcaneal tuberosities (Fig. 2a) were used to create the finite element model of the heel pad (Figs. 2b and c). Boundaries (Fig. 2b) between the fat pad, skin and calcaneus were identified using isocontour edge detection (Visual toolkit, Public Kitware.com). The resulting curves were exported to the finite element preprocessor (Msc.Mentat, MARC Analysis, Palo Alto, CA, USA). The areas enclosed by the curves were then meshed using two-dimensional plane–strain elements (e.g. Goske et al., 2006) (Fig. 2c). An out-of-plane thickness (7.5 mm) for all elements was specified which is midway between the MRI slice thickness (7 mm) and the length of the pressure sensors in the anterior–posterior direction (8 mm). Following Spears et al. (2005), the effects of bone and periosteum on stress distribution were simulated using zero-displacement boundary conditions. The heel pad thickness of the subject (i.e. the minimum distance from the medial calcaneal tuberosity to the plantar surface of the skin) in the non-weightbearing condition was 13 ± 1 mm. This is at the lower end of the range (11.2–16.6 mm) reported in the literature (Gefen et al., 2001; Gooding et al., 1985). To account for thicker heels, the model was scaled in the vertical direction to a heel pad thickness of 18 mm. The maximum heel pad width in the unloaded condition in the plane of interest was determined by MRI (60 ± 1 mm, Fig. 1) which is within the range (50–70 mm) reported for healthy heels (Fuller and Hogge, 1998) and was not varied. To overcome numerical errors associated with discretization, the number of elements was gradually increased by a factor of 4 until the difference in force–displacement curves between the current and subsequent refinement was less than 1%. Based on these criteria, a heel pad model consisting of 4435 elements was created (Fig. 2c).

The non-linear, incompressible behavior of the heel pad (Aerts et al., 1995) has previously been represented using an Ogden material model (e.g. Erdemir et al., 2006). The form (MSC.MARC User Guide) of this strain energy function (W) for incompressible behavior and single-order models

Fig. 1. MRI scans of the heel during (a) non-weightbearing, (b) barefoot weightbearing and (c) shod weightbearing. Note that the error in all linear measurements from the MRI (± 1 mm) equates to the pixel dimensions in the scans. In (c) the counter geometry that does not show up in MRI is superimposed. Note that the shod condition had no contact between the heel and counter.